Clinical Biomechanics of the Spine

Second Edition

Augustus A. White III
Manohar M. Panjabi

Lippincott Williams & Wilkins

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of the Spine

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To

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Vivian	

Martin Luther King, Jr., and Mohandas Karamchand Gandhi

Foreword

It is now little more than a decade since White and Panjabi's *Clinical Biomechanics of the Spine* first appeared. Meanwhile, this monumental work has become a classic, extensively quoted in papers, basic as well as clinical, that deal with traumatic or painful conditions in the cervical, thoracic, or lumbar spine.

Since 1978 there has been an enormous increase in scientific contributions from engineers and physicians, particularly from orthopedic surgeons interested in the biomechanics of spinal disorders. In the last decade, more papers have been published in this area than ever before.

Biology is implicit in the word "biomechanics," and in this comprehensive text the authors also have included recent basic studies in connective tissue and muscle physiology that are of importance for the mechanics of the spine.

The sad truth, however, is that this astounding amount of increased basic knowledge has not been mirrored at all in any scientific evidence of improved care for our patients with spinal disorders.

The burden now rests heavily on the clinicians to put to work all of the knowledge brought forward by the many studies in spine biomechanics—knowledge that is again collected and presented in an admirable and easily understandable manner by the esteemed researchers White and Panjabi. This knowledgeable physician/engineer team has created an updated platform on which we physicians and surgeons should stand when performing our prospective clinical studies in the next decade, an absolute necessity for further advancement in our field.

All too frequently, clinicians err on even the most fundamental parts of this accumulated knowledge and subject patients to experimentation in the operating theater or treatment rooms that would be regarded as unethical and maybe even unlawful if the same principles were applied, for example, in the field of pharmacologic treatment. An example is the recent and rapid increase in the number of spine implants, many of which have not yet been adequately tested in the fashion exemplified in this text before being used in patients with ill-defined pain syndromes.

In the foreword to the first edition I wrote, "In my eyes, this book is the most important contribution to the literature on spinal diseases since Schmorl and Junghanns' book, *Die Gesunde und die Kranke Wirbelsäule in Röntgenbild und Klinik*, which appeared in 1932." However, this new edition of *Clinical Biomechanics of the Spine* by

Dr. White and Dr. Panjabi has superseded that classic text in importance, and I congratulate the authors on bringing us up-to-date.

Let it be a challenge for all the clinicians who consult the pages of this book, to bring the level of clinical treatment of our patients in the 90s up to the same level of knowledge that biomechanicians reached in the 80s.

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Foreword

Historically, the first edition of *Clinical Biomechanics of the* Spine was a unique contribution to our knowledge of disorders of the spine. Some major questions with regard to spinal instability were answered and could for the first time be applied to clinical treatment of patients with spinal disorders.

The last decade of spine surgery has witnessed a rapid evolution of anterior decompression techniques and internal fixation devices that have a myriad of applications in spine surgery. Clinicians need a scientific and biomechanical basis for our work in orthopedics and spine surgery, and laboratory research provides an indispensable part of our core of knowledge. Very often the surgeon's enthusiasm for new devices and operations outweighs and outpaces the slower but necessary process of basic scientific evaluation of the new technology that we wish to use. This was true in the early 1970s, an era of a multiplicity of total joint replacements that were usually designed on the basis of the practicing orthopedist's ideas, rather than on investigation of biomechanics and wear properties.

In this second edition, there are detailed scientific analyses of the various spinal implants and arthrodeses (both anterior and posterior). These analyses are based on a thorough review of the literature and on basic research in the laboratory. This text is unique because of the special collaborative contribution of superb biomechanical studies from the basic scientist, Dr. Panjabi. This collaboration has resulted in a much greater understanding of the indications and techniques used in spine surgery. Ultimately, the efficacy and durability of these new techniques must also be judged by the clinical results in the patient, with a careful analysis of long-term results.

Philosophically, this magnificent text is a testimony to Dr. White's methodical, academic, and inquisitive mind; but, more than that, it is evidence of his energy, honesty, and integrity as an academic orthopedic surgeon.

The second edition of *Clinical Biomechanics of the Spine* will, without a doubt, serve as a major reference and teaching text for those clinicians who treat a wide variety of spinal disorders.

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Foreword

Biomechanics is a relatively new science. It applies mechanical laws to the living subject under normal and abnormal conditions, in order to study the mechanical and biologic aspects of muscles, joints, ligaments, and their surrounding structures. This science was largely developed in Sweden by the late Professor Carl Hirsch and his disciples, who are now working in the United States and other countries. The authors of this book, Professor Augustus A. White and Professor Manohar M. Panjabi, worked with Professor Hirsch: Professor White as a trainee in Professor Hirsch's department, and Professor Panjabi as a collaborator with Dr. Hirsch. They rapidly gained expertise in their chosen field and are in demand as teachers in the United States and abroad.

For those who deal with the spine, this text simplifies biomechanical complexities and adds immensely to the understanding and management of the spine's many inherent problems, particularly instability and spinal implant surgery. This comprehensive book will be of great value to orthopedic surgeons and residents alike, who have an interest in this sometimes complex anatomic jungle. The text is well-written, and the many artistic illustrations complement the text for maximum educational efficiency.

Michelangelo, in his magnificent painting on the Sistine Chapel in Rome, demonstrated great anatomic knowledge. Although he perceived form, he may not have always completely understood function. For the contemporary anatomist, this book will greatly augment the knowledge of spinal function as it relates to structure. For those involved in diagnosis and treatment of spinal disorders, this text will add another important tool to their armamentarium. This second edition of *Clinical Biomechanics of the Spine* takes a refreshingly new approach to spinal biomechanics, and adds not only to our knowledge and concept of spinal function and stability, but also to our understanding of structural anatomy. The work is timely, especially in this era of often complex spinal instrumentation, and is a welcome and important addition to our ability to evaluate these new devices.

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Foreword

Spinal disorders, particularly low back pain, remain at epidemic proportions. Often, a definitive diagnosis is impossible, and a variety of treatment regimens are debated. It is only by the study of basic scientific principles that advances in diagnosis and treatment are possible, and it is precisely in this area that *Clinical Biomechanics of the Spine* excels. The excellent approach of this text has been to combine the talents of a leading biomechanical engineer with those of a leading orthopedic surgeon. Their productive collaboration in research, as well as in this book, has been an example and inspiration to us all. This close research collaboration enables the authors to report firsthand on many of their own important experimental observations. As a result, the text is written with authority and is referenced with the rigor of a scientific article.

A small example is in order. Several years ago, Malcolm Fidler asked members of the International Society for the Study of the Lumbar Spine to define "clinical instability." He received a different answer from each of the respondents! White and Panjabi have grappled with this problem by proposing a rigorous biomechanically based definition and then showing us how the definition is clinically relevant. This approach of giving the scientific background of a problem has been used throughout the book. The chapters on spinal injuries are particularly clear. They explain the mechanisms of injuries, as well as the principles of surgical repair.

The first edition of this book was a particular favorite of mine and an invaluable reference text. Only a book of the quality of this new edition could possibly displace the first edition from my shelf. An interesting feature of this book is that the section at the back of each chapter is available to the clinician desirous of a quick overview of clinically important concepts. On the other hand, the student of biomechanics will prefer the more detailed treatment that is provided in each chapter.

This new edition of *Clinical Biomechanics of the Spine* deserves a place on the bookshelves of all clinicians dealing with spinal problems and of all researchers in the field. The treatment of the subject is comprehensive, the book is lucidly written, and the basic principles are not obscured by mathematical or engineering complexity. The book is a delight to read, and I recommend it highly.

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Preface

We define *clinical biomechanics* as that body of knowledge that employs mechanical facts, concepts, principles, terms, methodologies, and mathematics to interpret and analyze normal and abnormal human anatomy and physiology. Our purposes in writing *Clinical Biomechanics of the Spine* are to advance that knowledge by presenting a comprehensive review and analysis of the clinically relevant scientific data on the mechanics of the human spine, and to combine clinical experience and observations with scientific data in order to improve patient care.

During the course of our study of the biomechanics of the spine, we became aware of the value of the team approach, that of an orthopaedic spine specialist interested in biomechanics working with a mechanical engineer interested in clinical problems. This approach was beneficial in the improvement of research techniques and the application of engineering theory for a better understanding of clinical problems in the human spine. We recognize that a considerable amount of information concerning mechanical behavior of the spine is based on reliable research that may be used to improve patient care. There are engineering theories and principles that are crucial to the basic understanding of the normal and abnormal functioning of the spine. This information is useful in the understanding and treatment of patients with diseases of the spine, provided that it is presented in a clinical context. Our goal is to present this information in a manner that is understandable and useful to the clinician who may not be a student of biomechanics. In doing so, we hope to make some contribution toward a stronger scientific basis for the practice of medicine as it relates to the spine.

Since the clinician increasingly employs biomechanical terms and interprets problems with the assistance of the engineer, and since biomechanical engineers are showing more interest in the field of spine care and surgery, it is important to facilitate communication between the two disciplines. This book is particularly designed to improve the accuracy and effectiveness of communications between the clinician and the engineer.

The text is written and organized so that it can be read in different ways, depending on the reader's goals. The spine care clinician who is interested in a quick overview of the essential clinically relevant biomechanical concepts and information should study the figures, captions, and the "Clinical Biomechanics" section at the end of each chapter. Readers seeking the comprehensive educational impact of the book should read the full text, minus the "Notes" at the end of each chapter. These notes are included for the engineer or the clinician who is interested in more detail and theory, or in specific technical biomechanics information.

In this edition, we have not changed our rationale or our goals. We have removed information that has been shown to be inaccurate, and we have reorganized material. Since the first edition of this text was published in 1978, the amount of information available on the clinical biomechanics of the spine has more than doubled. Therefore, we have attempted to use good judgment in selecting the most important and most reliable new information for presentation. We have provided detailed illustrations and captions for emphasis of the most cogent material.

Chapters 1 and 2 describe physical properties and kinematics of the spine and present the *basic* science of spine biomechanics. These chapters constitute a review of the literature and offer our own selective interpretation and analysis of the most reliable and clinically important information.

Chapter 1 describes the physical properties of structures that are responsible for the behavior of the spine. Entirely new to this chapter is a presentation of spine anatomy from the engineering perspective. We think of it as "quantitative" or "blueprint" anatomy. Because structure is so important to function and to clinical factors, this emphasis seems worthwhile. Care has been taken not to oversimplify these data and material. An attempt has been made to combine these data and to provide a biomechanical analysis for a better understanding of the basic functional mechanics of the human spine.

Chapter 2 is a comprehensive presentation of what is known about the kinematics of the spine. Here we have sought to weigh the data according to our best interpretation of their validity and reliability and to offer an overview of the kinematic function of the entire spine. This section contains a considerable amount of new information and discusses new concepts such as the *neutral* zone, *passive* kinematic ranges, and *representative normal* angles of rotation.

Much of clinical spine work is based on engineering principles. We have endeavored to explain and clarify some of these principles. Chapters 3 through 8 are devoted to specific clinical problems, the understanding and treatment of which are largely based on the clinical biomechanics of the spine.

Chapter 3 is unique in that it collects the scientific, mechanical, and clinical studies on scoliosis and kyphosis, and offers an overview of the problem. Some important new information about etiology and surgical and nonsurgical treatment is included.

In Chapter 4, we evaluate a number of well-recognized fracture patterns, from occiput to sacrum. We offer practical theoretical analysis and interpretation of the mechanisms of injury of these fractures. Several new clinical and experimental observations have significantly changed our understanding of the mechanisms of injury. New imaging capabilities have significantly improved the analysis of fractures in the cervical, thoracic, and lumbar spine. In addition, the clinical literature is reviewed, and we recommend methods of management of these fractures.

Chapter 5 combines anatomic and biomechanical data with clinical information in a comprehensive method that will facilitate patient management and decision-making. In addition, checklists for evaluation and diagnosis of clinical stability are provided, and flow diagrams that help with management have been developed. The checklists have been revised to reflect cogent new clinical and biomechanical data.

Chapter 6 reviews the literature on spine pain. This chapter attempts to collect and integrate the various "facets" of this problem and to discuss them from a biomechanical perspective. Biomechanics is involved in the epidemiology, diagnosis, and treatment of spine pain. The following procedures are presented and analyzed: spinal manipulation, spinal traction, and physical therapy; biomechanics involved in diagnostic procedures; and the biomechanical effects of surgical procedures. In this chapter, the major new topics include a discussion of the clinical biomechanics involved in nocioception, vibrations, cervical disc pressures, epidemiological considerations, and evaluation and treatment of spine pain.

In Chapter 7, the basic principles and mechanics of orthotic devices are discussed. This is followed by a clinical review of virtually all spinal braces, with a detailed update based on new studies.

In Chapter 8, all surgical spinal procedures are studied in a special manner. Traditionally, surgical procedures have been described in terms of the anatomic approach, the surgical technique, and the clinical results. In this chapter, we include the preceding information, and, in addition, discuss the *mechanical analysis* of *surgical constructs* employed in spine surgery. This chapter also analyzes the clinical advantages and disadvantages of polymethylmethacrylate.

In recent years, there has been an explosion in the field of spinal instrumentation. There are perhaps as many as 75 new spinal implant devices. Many of these have undergone a variety of biomechanical tests. Therefore, this chapter on surgery is the most extensively revised chapter in the book. It can be difficult to decide *what* operation to do *when*, and *why*—and moreover, which implant should be used when, and why. We have avoided the "catalog" or "cookbook" approach. Instead, we have attempted to provide the reader with the necessary clinical and biomechanical information (including analyses of experimental testing) to choose appropriately from the many options.

Chapter 9 defines and describes over 100 terms relevant to orthopedic biomechanics, and cites numerous examples that relate to the spine. The important biomechanical concepts and terminology are included. We have added new clinically relevant terminology, such as *vibrations*, *neutral zone*, and *bulk modules*. Each entry is defined in scientific terms and is followed by at least one lay or clinical example. Most are accompanied by an illustration. When relevant, the term is further discussed and mathematical formulas are given in explanatory notes. This format allows easy and rapid assimilation of each definition and enriches the reader's general knowledge of biomechanics.

This chapter has been designed for two levels of interest and reading. It has been written for the reader who is primarily interested in a fundamental understanding of biomechanical terms as well as for individuals who seek a more detailed and mathematical grasp of the terms.

Chapters 1 through 8 include "Clinical Biomechanics" sections that summarize the salient practical and clinical features. Explanatory "Notes" also appear at the end of these chapters; they clarify biomechanical concepts by application of mathematical formulas and, in some instances, by citing clinical examples. Notes are indicated by superscript letters in the text. A partially annotated bibliography is included to help the reader select among the various references that may be offered on a given topic. You will note that we think some publications are "imperative reading."

Our general approach to both the scientific and the clinical aspects of the book is to review the literature, to bring forth the valid trends, and to provide some clinical examples and practical applications. We have endeavored to indicate salient unanswered questions or unresolved conflicts and to separate fact, reasoned hypothesis, theory, and speculation. Although we have tried to be as precise and scientific as possible, for pedagogic reasons we have taken the liberty of interpreting and presenting some information in a teleological context. We trust that this will not offend the pure scientist or distract from our attempts at objectivity.

We hope that this integration of theory, fact, and practice involving the biomechanics of the spine will aid both the clinician and the basic scientist in studying and better managing our fellow humans who are faced with spine problems.

Augustus A. White III, MD, DMed Sci
Manohar M. Panjabi, PhD, DTech

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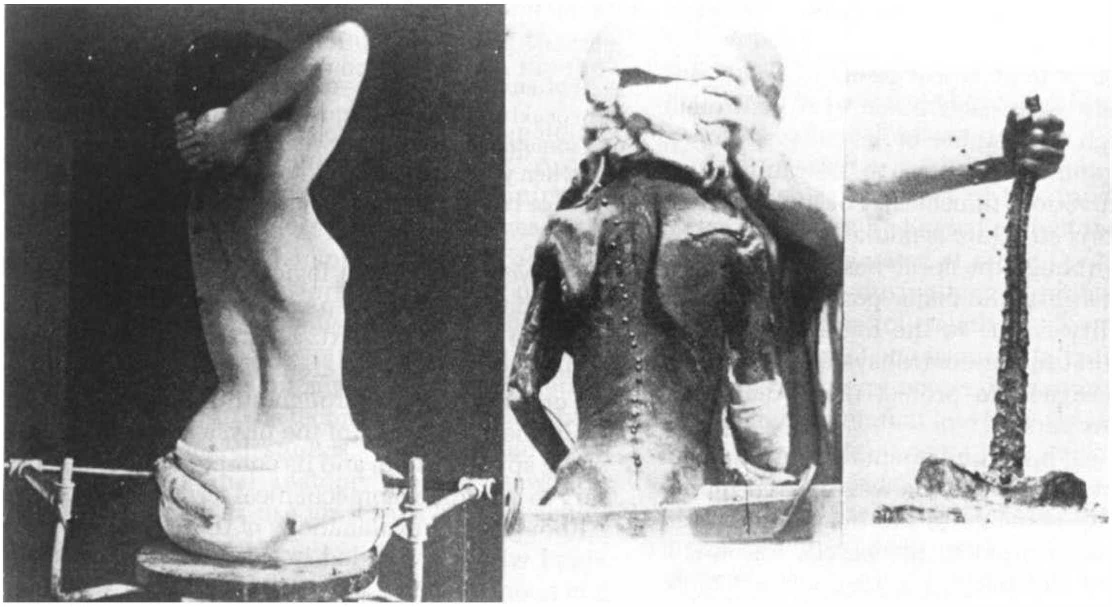
Clinical Biomechanics of the Spine

Physical Properties and Functional Biomechanics of the Spine

Figure 1-1. Physical properties of the human spine may be obtained from studies of: living subjects (Left); whole cadavers (Middle); isolated whole cadaveric spines (Right); and isolated spinal segments (not shown). Each type of study is a compromise between, on one side, reality (represented by a living subject) and, on the other side, simpler bench-type studies (represented by an isolated spinal segment). A living subject provides "realistic" but less accurate measurements. An isolated spinal segment lacks muscles, but can provide highly accurate data and allows the possibility of studying other effects, such as those due to trauma and surgical stabilizations. (Photographs obtained from a classic article by R. W. Lovett. The mechanism of the normal spine and its relation to scoliosis. Medicol and Surgical Journal 153:349, 1905.)



1

> It is a capital mistake to theorize before one has data.
> —THE ADVENTURES OF SHERLOCK HOLMES, 1892

This chapter is a review of the literature of what may be thought of as the *basic* science of spine mechanics. It is scientifically rather than clinically oriented and offers a thorough knowledge and understanding of the biomechanics of the spine. As such, we believe that it will be helpful to the reader in the evaluation and assimilation of subsequent chapters that are more clinically oriented. We have attempted to present the material so that it is palatable and understandable to those who do not have a biomechanical background. The in-depth reader is encouraged to refer frequently to Chapter 9, Biomechanics A to Z, where all of the biomechanical terms used in this book are defined and explained. Although reading this chapter is not required for comprehension of succeeding chapters, we suggest that it be read and referred to as background material for the rest of the book.

The spine is a mechanical structure. The vertebrae articulate with each other in a controlled manner through a complex of levers (vertebrae), pivots (facets and discs), passive restraints (ligaments), and activations (muscles). The long, slender, ligamentous bony structure is markedly stiffened by the rib cage. Although the spine has some inherent ligamentous stability, the major portion of the mechanical stability is due to the highly developed, dynamic neuromuscular control system. The spine structure is designed to protect the spinal cord, which lies at its center.

The spine has three fundamental biomechanical functions. First, it transfers the weights and the resultant bending moments of the head, trunk, and any weights being lifted to the pelvis. Second, it allows sufficient physiologic motions between these three body parts. Finally, and most important, it protects the delicate spinal cord from potentially damaging forces and motions produced by both physiologic movements and trauma. These functions are accomplished through the highly specialized mechanical properties of the normal spinal anatomy.

The material in this chapter is divided into sections, one for each of the spinal components. Each section is further divided into three subsections: anatomy, physical properties, and functional biomechanics. Only biomechanically relevant anatomy is presented. Description of the physical properties constitutes the major part of each section and presents experimental measurements of the physical characteristics of each spinal component or its elements. Finally, we describe the function of the spinal component under physiologic conditions as seen from the biomechanical viewpoint. Often the information needed to understand the functional biomechanics is derived from mathematical models that attempt to simulate the *in vivo* conditions. Seldom is such information obtainable from direct observations *in vivo*.

We have treated the components of the spine with degrees of detail depending upon the clinical importance and the availability of data. While presenting the experimental data, emphasis has been placed on what was found in a given experiment and how it relates to the biomechanical functions of the spine. Although the experimental techniques are discussed, the details are not given. References are provided for those with a more specific interest.

How should the experimental results be presented? Mere description is not enough.

> I often say that when you can measure what you are speaking about and express it in numbers you know something about it: but when you cannot measure it, when you cannot express it in numbers, your knowledge is of a meagre and unsatisfactory kind.[116]
> KELVIN, 1891

However, to keep a logical flow of ideas, an attempt has been made to present as few numbers as possible within the text. The remainder have been collected together and are presented in either tabular or graphic form throughout the chapter.

In the description of the physical characteristics of the spinal system and its components, it is necessary to use some biomechanical terms and concepts. Although brief explanations of these terms are provided wherever needed in this chapter, details are given only in the last chapter. There, each biomechanical term and concept is defined, explained, and exemplified. The terms are arranged alphabetically and therefore are easy to locate. The reader is encouraged and advised to consult the last chapter and to get into the habit of looking up the biomechanical terms encountered in this and any other chapter.

Biomechanically Relevant Anatomy

The spine consists of seven cervical vertebrae, twelve thoracic vertebrae, five lumbar vertebrae, five fused sacral vertebrae, and three to four fused coccygeal segments. As the spine is viewed in the frontal plane, it generally appears straight and symmetrical. In some individuals there may be a slight right thoracic curve, which may be due to either the position of the aorta or the increased use of the right hand. In the lateral or sagittal plane there are four normal curves. These curves are convex anteriorly in the cervical and lumbar regions and convex posteriorly in the thoracic and sacral regions. There is a mechanical basis for these normal anatomic curves; they give the spinal column increased flexibility and augmented, shock-absorbing capacity, while at the same time maintaining adequate stiffness and stability at the intervertebral joint level.

The thoracic curve is structural and is due to the lesser vertical heights of the anterior thoracic vertebral borders, as opposed to the posterior borders.[182] This is also true of the sacral curve. Curvatures of the cervical and lumbar regions are largely due to the wedge-shaped intervertebral discs. Consequently, when distracting forces are applied to the entire spine, there is a greater flattening of the cervical and lumbar lordosis as compared with the thoracic kyphosis.

Bogduk and Twomey have produced an anatomic text devoted entirely to the lumbar spine.[32] A new technique has been developed by Rauschning for presenting very high-quality, detailed images of the sectional anatomy of the spine.[210] Fresh cadaveric spine is put in a given posture, frozen solid, and then shaved in a defined plane using a heavy-duty cryomicrotome. Color photographs are taken, with a scale in place, of the cut surfaces as the sequential shaving progresses. The technique has already been used in some research studies,[211] but its potential for depicting real spinal anatomy, normal as well as pathological, and its use for research and teaching lie in the future.

INTERVERTEBRAL DISC

The intervertebral disc, which has many functions, is subjected to a considerable variety of forces and moments. Along with the facet joints, it is responsible for carrying all the compressive loading to which the trunk is subjected.[100, 207] When a person is standing erect, the forces to which a disc is subjected are much greater than the weight of the portion of the body above it. In fact, Nachemson and his associates have determined that the force on a lumbar disc in a sitting position is more than three times the weight of the trunk.[A161, 165] Such large loads have also been predicted by mathematical models.[225, 227] In addition, with any activity where dynamic loads are involved (e.g., jumping and trauma), the actual loads on the intervertebral disc are much higher, perhaps up to twice as high as those in the static positions. These are mainly compressive loads. The disc is also subjected to other types of loads and stresses. Tensile stresses are produced in certain portions of the disc during physiologic motions of flexion, extension, and lateral bending. Axial rotation of the torso with respect to the pelvis causes torsional loads that result in shear stresses in the lumbar discs. Combined rotation and bending result in stresses in the disc that are a combination of tensile, compressive, and shear stresses.

The loads to which the disc is subjected may be divided into two main categories according to the duration of application: short duration—high amplitude loads (e.g., jerk lifting) and long duration—low magnitude loads due to more normal physical activity. This division is important, because the disc exhibits time-dependent properties such as viscoelasticity characterized by load-rate sensitivity, hysteresis, creep, and relaxation.

Short-duration, high-level loads cause irreparable structural damage of the intervertebral disc when a stress of higher value than the ultimate failure stress is generated at a given point. The mechanism of failure during long-duration, low-level, repetitive loading of relatively low magnitude is entirely different and is due to fatigue failure. A tear develops at a point where the nominal stress is relatively high (but much less than the ultimate or even yield stress), and it eventually enlarges and results in complete disc failure.

Biomechanical behavior of the disc is dependent upon its state of degeneration, which in turn is age-dependent. From a study of 600 lumbar intervertebral discs that had been cut through the mid-disc plane and visually graded for disc degeneration on a scale of 1 to 4, Miller and colleagues found that (1) disc degeneration first appears in males in the second decade and in females a decade later, (2) by age 50, 97% of all lumbar discs are degenerated, and (3) the most degenerated segments are L3–L4, L4–L5, and L5–S1.[155]

Biomechanically Relevant Anatomy

The intervertebral disc has probably received as much attention as any anatomic structure in the entire spine complex, with the exception of the spinal cord. It constitutes 20–33% of the entire height of the vertebral column. The intervertebral disc is comprised of three distinct parts: the nucleus pulposus, the annulus fibrosus, and the cartilaginous end-plates.

Nucleus Pulposus The nucleus pulposus is a centrally located area composed of a very loose and translucent network of fine fibrous strands that lie in a mucoprotein gel containing various mucopolysaccharides. The water content ranges from 70–90%. It is highest at birth and tends to decrease with age.[175] With the help of MRI, it can be measured *in vivo*.[176] The lumbar nucleus fills 30–50% of the total disc area in cross-section. In the low back, the nucleus is usually more posterior than central and lies at about the juncture of the middle and posterior thirds of the sagittal diameter. The size of the nucleus and its capacity to swell are greater in the cervical and lumbar regions.

Annulus Fibrosus The annulus fibrosus is a portion of the intervertebral disc that gradually becomes differentiated from the periphery of the nucleus and forms the outer boundary of the disc. This structure is composed of fibrous tissue in concentric laminated bands (Fig. 1-2A, B). The fibers are arranged in a helicoid manner. They run in about the same direction in a given band but in opposite directions in any two adjacent bands. They are oriented at 30° to the disc plane and therefore at 120° to each other in the adjacent bands (Fig. 1-2B, C). The annulus fibers are attached to the cartilaginous end-plates in the inner zone, while in the more peripheral zone they attach directly into the osseous tissue of the vertebral body and are called Sharpey's fibers. This attachment to the vertebra is a good deal stronger than the other more central attachments, which is a useful characteristic in the clinical evaluation of spine trauma, clinical stability, and surgical constructs. Three-dimensional architecture of the collagen framework has been studied,[108] and morphological changes with aging have also been studied.[206]

Cartilaginous End-Plate This is composed of hyaline cartilage that separates the other two components of the disc from the vertebral body.

An excellent histological study of the lumbar vertebral end-plate and its changes with age (0–37 years) has been provided.[30] Starting with an active growth cartilage, the age changes result in irregularly arranged growth cartilage, which disappears with time and is replaced by bone.

Physical Properties

Elastic Characteristics

The intervertebral disc is a viscoelastic structure. To determine its elastic properties, the mechanical tests must be done at slow loading rate so that the viscoelastic effects are minimized. A proper description of the test should include information concerning the loading rate. Older studies do not provide such data, but one may assume, because of the limited capabilities of the older testing machines, that the tests were done at slow speeds.

The human spine and the intervertebral discs are subjected to physiologic and traumatic loads. These may be relatively simple (e.g., due to flexion of the spine) or complex (e.g., during combined flexion, lateral bending, and torsion to pick up an item from the floor). We describe some experiments that have attempted to quantitatively document the physical properties of the disc, both as a whole structure and as a material. It is important to distinguish between the load applied to the disc as a structure and the stresses produced within the disc material. For example, when we stand in neutral posture, the disc is subjected to a compression load, while its nucleus experiences compressive stresses and its annular fibers experience tensile stresses. We have divided the description of the physical properties of the disc (elastic, viscoelastic, and fatigue) according to the loads applied.

Compression Characteristics The compression test has been the most popular mechanical test for the study of the disc, probably because the disc is the major compression-carrying component of the spine. Many experiments have been done to determine the compressive properties of the disc.[41,67,100,101,102,215,254]

It may be helpful to know how such a test is actually performed. Typically, a test specimen consists of a lumbar disc with anterior and posterior longitudinal ligaments intact and a thin slice of bone on either side. The specimen is placed in a universal testing machine (e.g., Instron or MTS) that is capable of applying large, controlled compressive loads. The load applied to the test specimen and the deformation produced are recorded continuously. The load

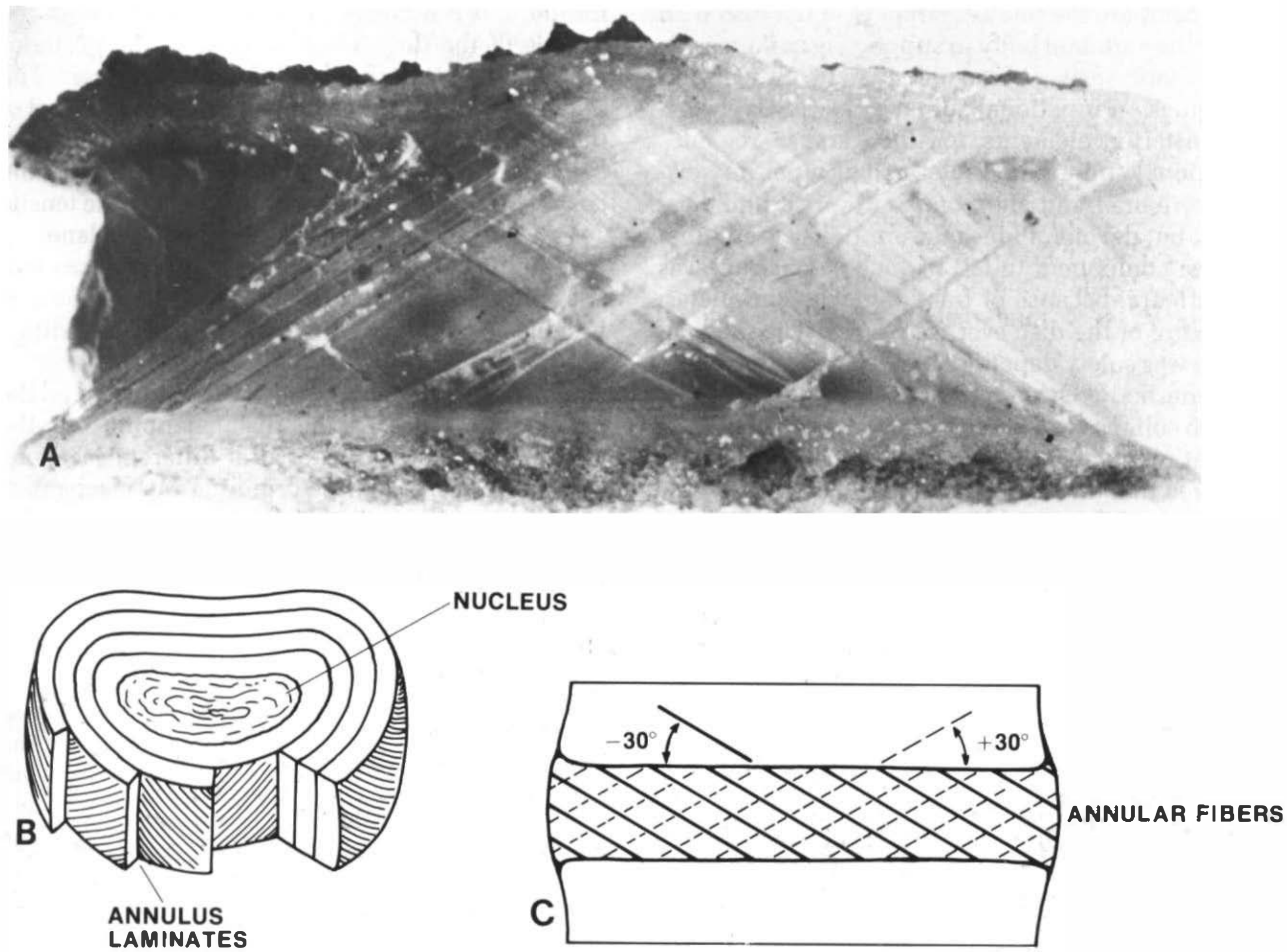


FIGURE 1-2 Intervertebral disc. **(A)** A photograph of a disc clearly shows the annular fibers and their orientation. **(B)** The disc consists of a nucleus pulposus surrounded by the annulus, made of concentric laminated bands of annular fibers. In any two adjacent bands the fibers are oriented in opposite directions. **(C)** The fibers are oriented at about ± 30° with respect to the placement of the disc. *(Photograph courtesy of Dr. Leon Kazarian.)*

(y-axis) and deformation (x-axis) curve has been found to be very useful in documenting the physical behavior of a test specimen. The specimens for the compression test of a disc have greatly varied in different experiments, from a disc with thin slices of vertebrae on each side to a disc with two whole vertebral bodies. Typically, the load–displacement curve is of a sigmoid type, with concavity toward the load axis initially, followed by a straight line, and convexity toward the load axis in the final phase, just prior to failure. Such a curve implies that the disc provides very little resistance at low loads. But as the load is increased, the disc becomes stiffer. Thus, it provides flexibility at low loads and stability at high loads. As we will see, this behavior is even more accentuated in spinal ligaments.

Virgin observed that, although discs were subjected to very high loads and showed permanent deformation on removal of the load, there was no herniation of the nucleus pulposus due to compressive load.[254] Even when a longitudinal incision was made in the posterolateral part of the annulus fibrosus all the way to the center and the specimen was loaded in compression, there was very little change in the elastic properties and definitely no disc herniation. This phenomenon under compressive load has been substantiated in experiments by Hirsch[100] and Markolf and Morris.[146]

To compare the relative strength of the disc with that of the vertebral body in supporting compressive loads, static tests were conducted by Brown and colleagues on functional spinal units (FSUs), without posterior elements, of the lumbar region.[41] (Functional spinal unit is defined as a pair of adjacent vertebrae and the connecting disc and ligaments, but devoid of musculature.) They found that the first component to fail in such a construct was the vertebra, because of fracture of the end-plates. No failure of the disc ever took place. The mode of failure was solely dependent on the condition of the vertebral body. Osteoporotic vertebrae showed extensive collapse of the end-plate and the underlying bone at relatively low loads. Brown and colleagues observed that there were no differences between the vertebrae with "normal" discs and those with degenerated discs. Farfan, to the contrary, proved by his large number of tests that the degenerated disc was actually stronger than the normal disc when subjected to compression,[67] a factor that may be related to the clinical observation of frequent disc rupture and herniation in older age groups (*i.e.*, 50 and above). Experiments were conducted on lumbar spine specimens using discography to demonstrate the movements of the nucleus pulposus under compressive loading. After the first cracks were heard, indicating fracture of the vertebral end-plates, the nucleus was found to migrate into the bodies, resembling Schmorl's nodes. The movement of the nucleus was demonstrated using discography. Thus, we conclude that disc herniation is not caused by excessive compressive loading, although Schmorl's nodes may be the result of such loading.

With central compressive loading, the disc was observed to bulge in the horizontal plane, but not in any particular direction.[38,41,214] This implies that the tendency for the disc to herniate posterolaterally, as seen in the clinical situation, is not inherent in the structure of the disc but must depend upon certain loading situations other than compression (assuming that uniform stress prevails in the disc under compressive loading).

Tensile Properties The disc is seldom subjected to tensile loads under normal physiologic activities. Even under the application of traction to the spine, the discs are under compression load due to muscle activities. However, the disc annulus is subject to *tensile* stresses in various physiologic conditions. In flexion, the instantaneous axes of rotation lie in the frontal and transverse plane and pass through the middle of the disc. Thus, in flexion, the posterior part of the disc is subjected to tensile stresses. The opposite is true in extension (*i.e.*, the anterior part of the disc experiences tensile stresses). In lateral bending, the tensile stresses are produced on the convex side of the bend. In axial rotation, the tensile stresses develop at about 45° to the disc plane.

Finally, compressive loading also produces tensile stresses. Thus, it may be concluded that the disc is subjected to tensile stresses in all different directions under various loading situations.

Two types of studies have been conducted on the tensile properties of the disc: mapping out the strength of the disc material at different locations and orientations, and determining the mechanical properties of the intact disc. Strength of disc material was studied by cutting the vertebra–disc–vertebra construct into multiple, axially oriented, rectangular sections (Fig. 1-3A). The specimens were stretched to failure in a testing machine, and the load–displacement diagrams were recorded. Failure load values were collected from various samples and combined as axial tensile strength maps of the disc. Results for the two discs tested by Brown and colleagues are shown in Figure 1-3B. We note that although there is some variation between the strength maps of the two discs, the anterior and posterior regions of the disc are stronger than the lateral region, and the central region, consisting of the nucleus pulposus, is the weakest. This distribution may be "nature's attempt" to provide strength where most of the failures and herniations tend to occur.

What is the strength along directions other than the axial? With this question in mind, Galante performed extensive biomechanical tests of the disc material.[76] He cut the disc annulus into thin samples (1 × 2 mm) along different orientations and subjected these samples to tensile loads. His results for stiffness[B] of the disc are summarized in Figure 1-4A. The stiffness was found to vary to a great extent with the orientation of the samples; the axial samples were the most flexible, while the samples taken at 15° to the horizontal plane were the stiffest.

Since the loads applied in these experiments were rather low, separate experiments were performed to determine the strength. The results are presented in Figure 1-4B. In comparing the results of the samples taken along the horizontal direction with those taken along the fiber direction, the latter

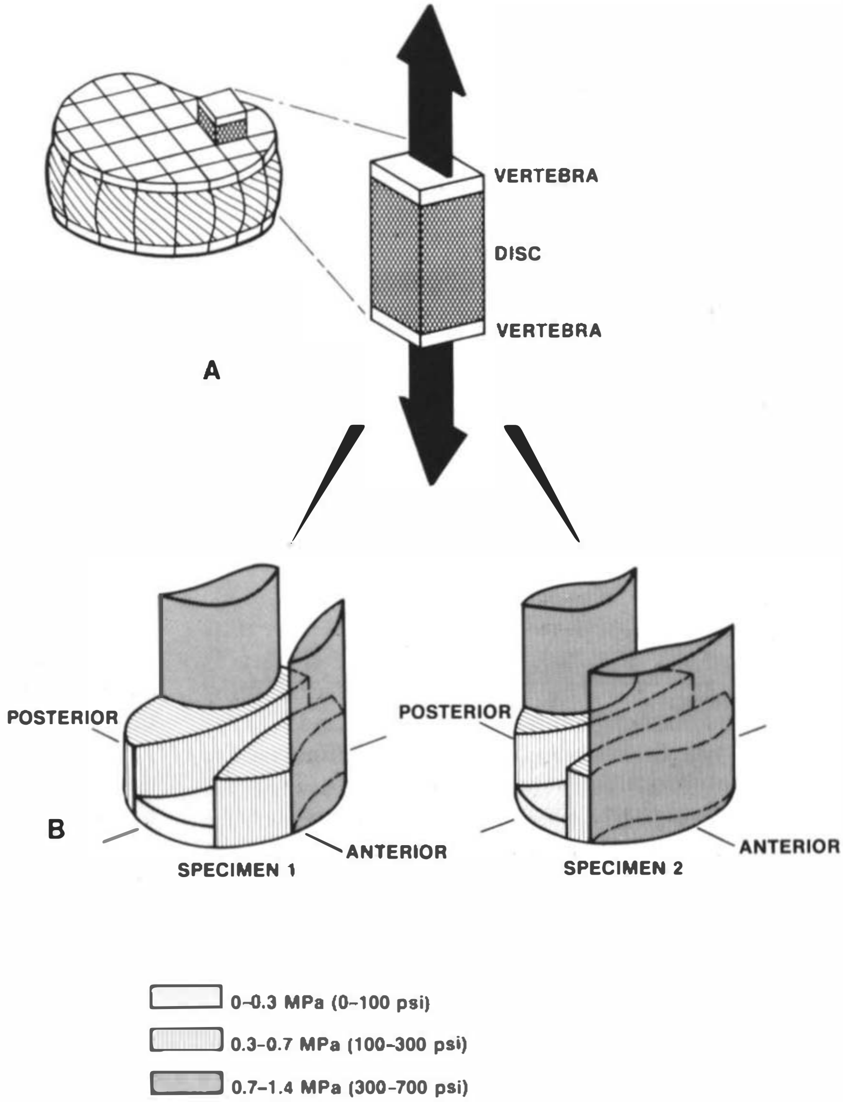


FIGURE 1-3 Tensile strength of disc material. **(A)** For the tensile test, the disc specimens were obtained by dividing the vertebra-disc-vertebra construct into longitudinal sections. **(B)** The results of the tensile tests performed on two specimens are depicted in the form of contour maps, where the height represents the strength in tension at that point. The disc is strongest in the anterior and posterior regions, the center being the weakest. *(Based upon the findings of Brown, T., Hanson, R., and Yorra, A.: Some mechanical tests on the lumbosacral spine with particular reference to the intervertebral discs. J. Bone Joint Surg., 39A: 1135, 1957.)*

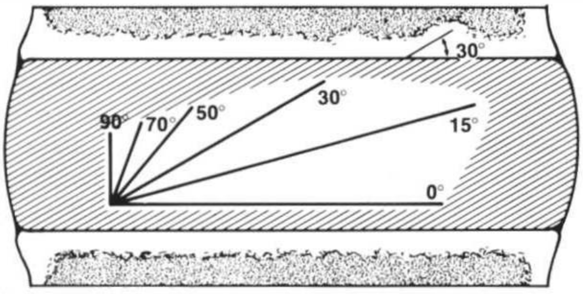
were found to be about three times as strong as the former.

The results of stiffness and strength tests clearly show that the disc is an anisotropic structure. This specialized structure has been optimized to resist certain kinds of loads in the most efficient manner. However, such specialization of the mechanical properties of the disc has a negative consequence; the disc is unable to resist the rest of the loads in an equally optimal manner.

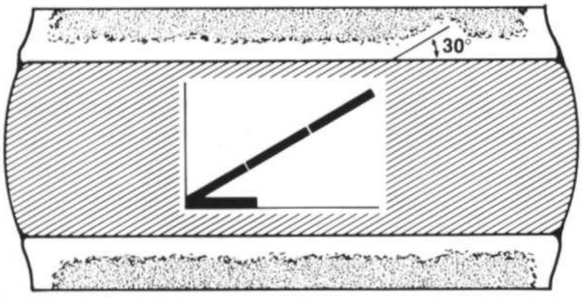
The second type of study on the tensile properties of the whole disc as a structure was first conducted by Markolf.[144] He loaded the vertebra–disc–vertebra specimens from the thoracic and lumbar regions in tension using an Instron testing machine. The disc was found to be less stiff in tension than under compression. This was attributed to the build-up of fluid pressure within the nucleus under compression loading.

Bending Characteristics Bending and torsional loads are of particular interest, because experimental findings suggest that pure compression loads do not damage the disc.[87] A description of the mechanism of disc prolapse using compression load combined with flexion and lateral bend is provided on page 18.

Bending of 6–8° in the sagittal, frontal, and other vertical planes did not result in failure of the lumbar disc. However, after removal of the posterior ele-



A STIFFNESS



B STRENGTH

FIGURE 1-4 Disc anisotropy. **(A)** Tensile stiffness of the disc annulus in different directions is shown. The stiffness is highest along a direction 15° to the disc plane and lowest along the disc axis. **(B)** The strength of the annulus along two directions is compared. Samples taken along the direction of the annulus fiber were found to be about three times as strong as those taken along the horizontal direction. *(Data from Galante, J. O.: Tensile properties of the human lumbar annulus fibrosus. Acta Orthop. Scand., 100 [Suppl.], 1967.)*

ments and with 15° of bending (anterior flexion), failure did occur.[41] A triangular piece of bone was avulsed from the posteroinferior aspect of the superior vertebra in this experiment. Other interesting findings concerned the bulging of the disc during normal physiologic motions. The disc bulged anteriorly during flexion, posteriorly during extension, and toward the concavity of the spinal curve during lateral bending. The disc curved inward on the opposite sides (on the convexity of the curve). Very little motion took place in a direction perpendicular to the plane of motion. These findings were confirmed by Roaf,[215] who noticed that the bulging of the annulus is always on the concave side of the curve and that denucleation seemed to increase bulging. However, no exact measurements were given. With the help of nucleographs of the disc, Roaf found that during flexion/extension the nucleus pulposus does not change in shape or position. This information is useful and offers support to those who emphasize the importance of a "flat back" or the maintenance of a slightly flexed lumbar spine as a treatment and prophylaxis for patients with low back pain or sciatica. Since these early studies,[41,215] several experiments have been done to quantify in greater detail and to provide better understanding of the mechanism of disc bulge and movements of the nucleus under various clinically relevant loading conditions. These are described later in a separate section on disc bulge (p. 16).

Torsional Behavior The hypothesis that torsion may be a major injury-causing load was proposed by Farfan in 1973,[67] based upon the earlier work by his group.[89] In the particular experiment, fresh cadaveric lumbar vertebra–disc–vertebra construct (including the posterior elements) was subjected to torsional loading around a fixed axis passing through the posterior aspect of the disc. Torque was applied, and a continuous record was made of the applied torque and the angle of deformation until the failure occurred. The torque angle curves were found to be of sigmoid shape, with three distinct phases. In the initial phase, 0–3° of deformation could be produced by very little torque. In the intermediate phase, consisting of 3–12° of rotation, there was a linear relationship between the torque and the angular deformation. In the final phase, about 20° of rotation was generally required to produce the failure. The angle of failure was somewhat less for degenerated discs. Sharp, cracking sounds emanating from the specimen were always noted before failure occurred. On close examination, no failure of the end-plates was found. It is believed that the cracking sounds came from the injuries to the annulus. This is reminiscent of the crack or snap that is sometimes felt or heard when patients report acute low back injuries.

Farfan and colleagues[89] tested a total of 21 nondegenerated and 14 degenerated discs from the lumbar region according to the technique described above. They found that the average failure torque for the nondegenerated discs was 25% higher than that for the degenerated discs. The average angle at failure was 16° and 14.5°, respectively. Generally, a large disc exhibited large torsional strength. A round disc was found to be stronger than an oval disc.

Shear Characteristics Experiments on torsion of the disc provide important information regarding the torsional strength of the disc *as an intact structure*. Although the disc is subjected to shear stresses during torsional loading, the stresses are not uniformly distributed. They are high along the periphery and low in the center. Therefore, the torsional experiments do not provide precise information about the horizontal shear characteristics of the disc. Experiments were performed to study the lumbar disc in direct shear. The shear stiffness in the horizontal plane (anteroposterior and lateral directions) was found to be about 260 N/mm.[144] This is a high value and is clinically significant, showing that a large force is required to cause an abnormal horizontal displacement of a normal vertebral disc unit. This means that it is relatively rare for the annulus to fail clinically because of pure shear loading. Most likely, clinical evidence of annular disruption implies that the disc has failed because of some combination of bending, torsion, and tension.

Numerical values for stiffness properties of the disc for various physiologic motions have been collectively provided in Table 1-1.

Viscoelastic Characteristics

The physical properties of a structure that document its time-dependent behavior are called its viscoelastic characteristics. For example, if we were to repeat the elastic tests described above at faster loading speeds, we would find that the load–displacement curves would be different. In fact, they would exhibit stiffer behavior with increasing loading speed. The experiments described below use creep, relaxation, and hysteresis tests to characterize the viscoelastic behavior of the disc.

Creep and Relaxation The intervertebral disc exhibits creep and relaxation.[102] Markolf and Morris studied creep under the application of three different loads and made observations up to 70 minutes.[146] The higher loads produced greater deformation and faster rates of creep. Kazarian performed compression creep tests on FSUs and classified the discs of the specimens into four grades, from 0 to 3, according to their degree of degeneration.[111] (This classification is similar to the one used by Rolander.[217]) He observed that the creep characteristics and the disc grades are related, as shown in Figure 1-5. Note that the shapes of the curves are different. The nondegenerated discs (Grade 0) creep slowly and reach their final deformation value after considerable time, as compared with the degenerates discs (Grades 2 and 3). The Grade 0 curve is characteristic of a more viscoelastic structure, as compared with the curves of Grades 2 and 3. Thus, the process of degeneration makes the discs less viscoelastic. This implies that as the disc degener-

TABLE 1–1 Stiffness Coefficients of the Intervertebral Disc

Authors	Stiffness Coefficients[*]	Maximum Load[*]	Spine Region
Compression (−Fy[†])			
Virgin, 1951	2.5 MN/m	4500 N	Lumbar
Hirsch & Nachemson, 1954	0.7 MN/m	1000 N	Lumbar
Brown, et al., 1957	2.3 MN/m	5300 N	Lumbar
Markolf, 1970	1.8 MN/m	1800 N	Thoracic & lumbar
Moroney, et al., 1988	0.5 MN/m	74 N	Cervical
Tension (+Fy[†])			
Markolf, 1970	1.0 MN/m	1800 N	Thoracic & lumbar
Shear (Fx, Fz[†])			
Markolf, 1970	0.26 MN/m	150 N	Thoracic & lumbar
Moroney, et al., 1988	0.06 MN/m	20 N	Cervical
Axial Rotation (My[†])			
Fairfan, et al., 1970	2.0 Nm/deg	31 Nm	Lumbar
Moroney, et al., 1988	0.42 Nm/deg	1.8 Nm	Cervical

[*] N = newton, kN = 1000 newton, MN = 1,000,000 newton, Nm = newton meter
To convert to the inch-pound system, multiply by the following numbers:
(MN/m) × 5600 = lbf/in (Nm/deg) × 0.738 = in lbf/deg (N) × 0.225 = lbf (Nm) × 0.738 = in lbf

[†] See Figure 1–32 and text for details.

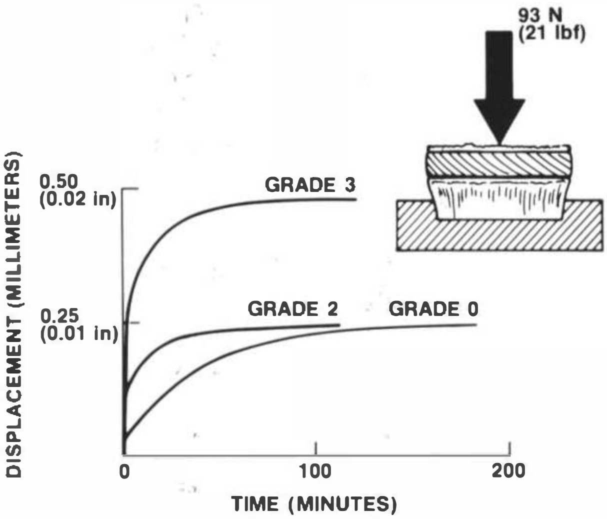


FIGURE 1-5 Creep behavior of the disc. The creep behavior of a structure is documented by applying a sudden load and maintaining it. The deformation of the structure as a function of time is recorded. This behavior seems to correlate with the degree of degeneration of the intervertebral disc. A sample of creep curves for discs with different grades of degeneration are shown. The nondegenerated disc (Grade 0) has smaller overall deformation, and this deformation is reached over a relatively longer period as compared with the degenerated disc (Grade 3). *(Data from Kazarian, L. E.: Creep characteristics of the human spinal column. Orthop. Clin. North Am., 6:3, 1975.)*

ates it loses the capability to attenuate shocks and to distribute the load uniformly over the entire endplate.

Hysteresis All viscoelastic structures, including the disc and FSU, exhibit hysteresis. It is a phenomenon in which there is loss of energy when a structure is subjected to repetitive load and unload cycles. When a person jumps up or down, the shock energy is absorbed on the way from the feet to the brain by the discs and vertebrae because of hysteresis. It may be thought of as a protective mechanism. This phenomenon was first observed in the discs by Virgin.[254] Hysteresis seems to vary with the load applied and the age of the disc, as well as its level. The larger the load, the greater the hysteresis. It is largest in very young people and smallest in the middle-aged. Virgin observed that the lower thoracic and upper lumbar discs showed less hysteresis than the lower lumbar discs. He also observed that hysteresis decreased when the same disc was loaded a second time. This may imply that we are less protected against repetitive loads. Epidemiologic studies show that people who drive motor vehicles have a higher incidence of herniated discs.[115] The repetitive axial vibrations may be a factor.

Fatigue Tolerance

Fatigue tests of the disc are important for establishing the number of load cycles that can be tolerated before radial and circumferential tears develop. Since the biologic capacity for repair and regeneration of the disc is thought to be low, its fatigue properties are important. Unfortunately, very little is known about this subject. Brown and colleagues performed a single fatigue test on the disc by applying a small constant axial load and a repetitive forward bending motion of 5°.[41] The disc showed signs of failure after only 200 cycles of bending, and it completely failed after 1000 cycles. This indicates that the fatigue life is low under such experimental conditions *in vitro*. The fatigue tolerance of the disc *in vivo* is not known.

Functional Biomechanics

Intradiscal Pressure

Measurement of In Vivo Loads There are very few precise studies on the behavior of the spine components *in vivo*. Most of the work is done on cadaver materials. Although these studies have provided large amounts of valuable information, the magnitude of the loads applied to the disc cannot be determined *in vitro*. Nachemson and Morris determined for the first time the actual loads to which a disc is subjected *in vivo*.[165] They used the concept of nucleus pulposus as a load transducer. By means of *in vitro* experiments on vertebra-disc-vertebra preparations, they found that the fluid pressure within the nucleus is directly related to the axial compression applied to the disc (Fig. 1-6A).

The pressure was measured by a transducer in the form of a special needle carrying a miniature electronic pressure gauge at its tip. Thus, by measuring the pressure within the nucleus, they could compute the load on the disc.

Having developed the technique, Nachemson and his group have measured the *in vivo* loads to which the lumbar discs are subjected when a person is resting in different body postures or performing a certain task.[162,163,165] A sample of the results of their work is shown in Figure 1-6B. Observe that the load

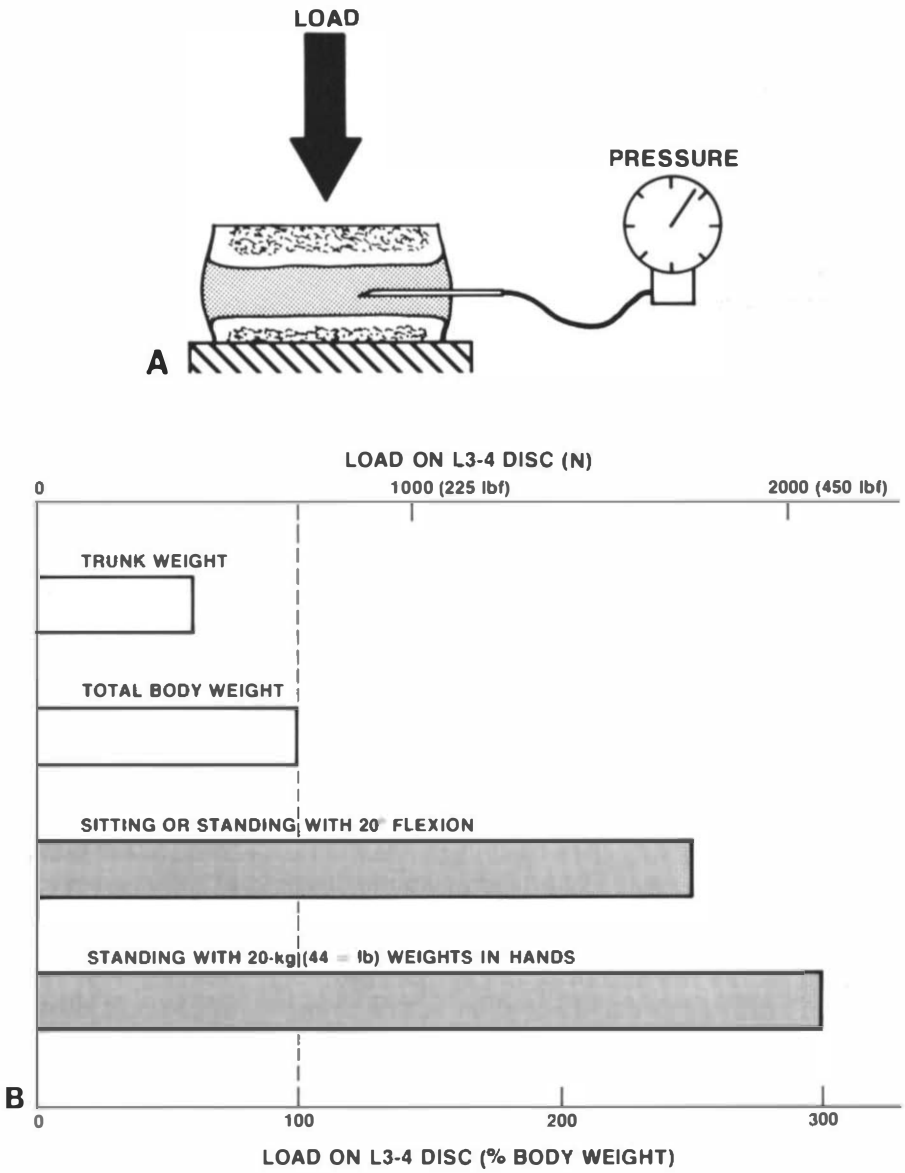


FIGURE 1-6 Intradiscal pressure and loads on the disc. **(A)** The needle pressure transducer is calibrated by introducing it into the nucleus pulposus of a cadeveric functional spinal unit. A correlation is obtained between the compressive load applied and the pressure within the nucleus. **(B)** Using the same needle transducers, *in vivo* measurements were made at the L3–L4 disc in volunteers performing physiologic tasks. The bar graphs record the compressive load on the disc, Note that the disc load while standing with a 20-kg weight in the hands is about three times the weight of the whole body. *(Results based upon those of Nachemson, A.: The load on lumbar discs in different positions of the body. Clin. Orthop., 45:107, 1966.)*

carried by the discs is rather large. Although the portion of the body about the L3 disc constitutes 60% of the total body weight, the load on the L3 disc, in sitting and standing positions with 20° of flexion, is 200%. It becomes 300% with the addition of 20 kg (44 lb) of weight in the hands. A biomechanical analysis of the high load to which the disc is subjected is depicted in Figure 1-30 and is discussed in the Notes.[I]

The same technique was used by Nachemson[161] and Rolander[217] to measure the prestress present in the discs. They measured the nucleus before and after cutting the posterior elements. A pressure of 0.07 MPa (10 psi) was found in the intact FSU, while there was none when the arch was removed. This corresponds to a compressive prestress of about 120 N (26 lbf).

Measurement of Disc Degeneration Discography is used clinically to provide a qualitative description of disc degeneration. Quinnell and Stockdale developed a technique to quantify the discography by recording the intradiscal pressure during this procedure.[209] They reported average intradiscal pressures related to different postures: 154 kPa in prone, 550 kPa in standing, and 700 kPa in sitting. Modify-ing the Quinnell technique to simultaneously measure the pressure as well as the injected volume, Panjabi and co-workers conducted experiments on fresh cadaveric lumbar spines.[180] They found a significant relationship between the disc degeneration grades and the intrinsic intradiscal pressures. The latter is defined as the pressure within the nucleus when the spine is unloaded. If this technique were to be used clinically, degeneration status of the disc could be determined with a defined certainty during discography.

Effects on Mechanical Properties It may seem obvious that the mechanical properties of the intervertebral disc will be affected by fluid injection into the disc, yet there are only a few studies that have looked at this phenomenon in any detail.[15] Using fresh cadaveric lumbar specimens, investigators studied the mechanical properties of the disc before and after injection of the fluid. The discs of all the fluid-retaining specimens became stiffer, while those that could not retain fluid showed, on average, no significant change in stiffness.

Disc Function After Injury

The intervertebral disc may be injured by several mechanisms (e.g., excessive torsion, combined torsion and lateral bending, sudden axial compression while in hyperflexed posture, or surgery for relief of pressure on the nerve root or the spinal cord). A question arises as to the effect of an injury on the mechanics of the disc and FSU at the level of injury as well as the neighboring levels. In an earlier biomechanical study, it was suggested that an injury to the disc does not affect its mechanical properties.[146] This left the impression that a kind of "self-sealing" phenomenon existed in the disc. However, recent studies have shown that the injury to the disc does, in fact, significantly affect its mechanical behavior.[83, 187]

With the use of fresh cadaveric lumbar spine FSUs and well-developed three-dimensional techniques of load applications and motion measurements, the effects of injuries on the elastic and time-dependent physical properties were studied.[187] The injuries were created surgically for the sake of reproducibility. The injuries consisted of (1) cutting of a square window on the right posterolateral aspect of the annulus and (2) removal of the nucleus through this window (Fig. 1-7). The specimens were tested before and after each of the two injuries.

Significant changes in both the main motions (in the directions of the applied loads) and the coupled motions (in directions other than those of the main motions) were observed. Among the main motions, the major effects were caused by the second injury (*i.e.*, removal of the nucleus) (Fig. 1-7). The increases in motion were largest for flexion, left lateral bending, and traction (not shown), all of which produced tensile stresses at the site of injury (on the right side). On the other hand, the smallest changes were due to the compression (not shown) and axial torsional loads. The reason for the small changes under compression, we believe, is the fact that the injury affected only a small area of disc compared with the total area of the disc. In the case of axial rotation, the probable explanation is the fact that it is the facets that resist torsion to a significant degree. Because the facets were not affected by the disc injury, the torsional behavior of the FSU was not affected. The changes seen in the elastic behavior of the disc also reflected the changes observed in the viscoelastic properties.

Our study concluded that even though the compressive behavior of the spine is not significantly altered by injuries to the disc, which agrees with the findings of an earlier study,[146] most of the other three-dimensional properties are. These observations have been confirmed.[83] The changes that do occur are not symmetrical with respect to the sagittal plane. These asymmetrical motions at one joint, one may hypothesize, may lead to adaptive changes in adjacent joints, both inferior and superior to the site of injury.

Stresses in the Disc

Stress at a point in a tissue is defined as the force per unit area. When the stresses at a point exceed the strength of the tissue at that point, the tissue fails. Therefore, in order to understand the failure mechanisms of the disc, it is necessary to know the orientation and magnitude of the stresses generated in response to the various loads applied to the disc. The stresses may be tensile/compressive and/or shear. The first two are called normal stresses and are perpendicular to the plane of observation, while the shear stresses are parallel to the plane. In general, stresses are difficult to measure, especially on the inside of a structure. Therefore, mathematical models are used for this purpose.[124, 125] The studies described below were conducted on computers using special mathematical models called Finite Element Methods, or simply FEM.

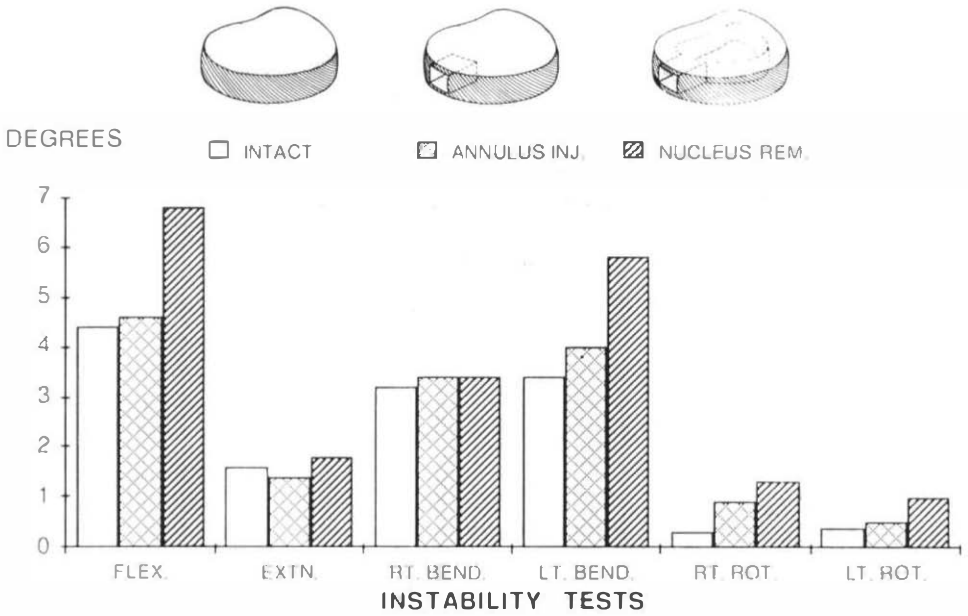


FIGURE 1-7 Effects of disc injury on the mechanical properties of a lumbar functional spinal unit are shown. Three states of the disc were investigated; intact, with annulus injury on left side, and after removal of the nucleus. Instability tests were conducted using pure moments of flexion, extension, right lateral bending, left lateral bending, left rotation, and right rotation. The bar graph shows the main motions for the intact and two injuries due to each of the six physiological loads. Annulus injury with nucleus removal produced greater changes than the annulus injury alone. The maximum absolute changes were seen in flexion and left lateral bending. On the percentage changes, it was the axial rotation that exhibited the greatest effect of the disc injury. *(Data from Panjabi, M. M., et al, Effects of disc injury on the mechanical behavior of the human spine. Spine, 9(7):707, 1984).*

Mathematical Model Since most of the information presented below is based upon computer models in general and a certain model in particular, it may be helpful to introduce the model developed by Shirazi-Adl and co-workers.[238] It is the most comprehensive model available to date. It simulates a lumbar L2–L3 functional spinal unit: the disc, facet joints, and ligaments. The vertebrae are modeled with correct geometry and appropriate stiffness values for cortical and cancellous bones. The facet articulations are modeled as two sliding surfaces mimicking the real facet joints. Ligaments are modeled as nonlinear springs. Finally, the disc is modeled by three components: incompressible gelatinous fluid representing nucleus, fibers representing collagenous fibers of the annulus, and a ground substance surrounding the fibers and simulating the adhesion between the fibers. The model has been validated by direct comparison of some measured parameters (e.g., disc bulge, effects of intradiscal pressure, and load–displacement curves) with the model predictions. The agreements have been good. Therefore, the predicted stresses in the disc should be considered good representations of actual stresses.

Although the loads applied to the disc *in vivo* are certainly very complex, each physiologic load is discussed separately for the sake of understanding.

Compression The compressive load is transferred from one vertebral end-plate to the other by way of the nucleus pulposus and the annulus fibrosus. In the early years of life (up to age 25–30), the nucleus has sufficient water content to act like a gelatinous mass.[24, 166, 208] As the load is applied, a pressure develops within the nucleus, which pushes the surrounding structures in all directions away from the nucleus center (Fig. 1-8A). In other words, the central portions of the two vertebral end-plates are pushed away from each other,[37, 218] and the annular ring is pushed radially outward.[38, 41, 214] The compression load produces complex stresses within the annular ring. Approximate stresses along various directions in the outer and inner laminae of the annulus are indicated in Figure 1-8B. The length of the arrow corresponds to the relative magnitude of the stress. In the outer layers, the stresses are generally small. Axial and circumferential stresses are compressive, while the annular fiber stress is tensile. Arrangement of the fibers at ± 30° accommodates absorption of tensile stresses. The axial and circumferential stresses are still compressive, but their magnitude is larger. The fiber stress is larger and still tensile. The innermost lamina of the annulus is subjected to somewhat different stresses. In addition, there is fluid pressure of the nucleus that supports the inner laminae. The orientation of annular fibers and the nucleus play important roles in transferring the compressive loads from one vertebra to another.

The situation is quite different when the nucleus is dry (see Fig. 1-9A). The load-transferring mechanism is significantly altered because the nucleus is

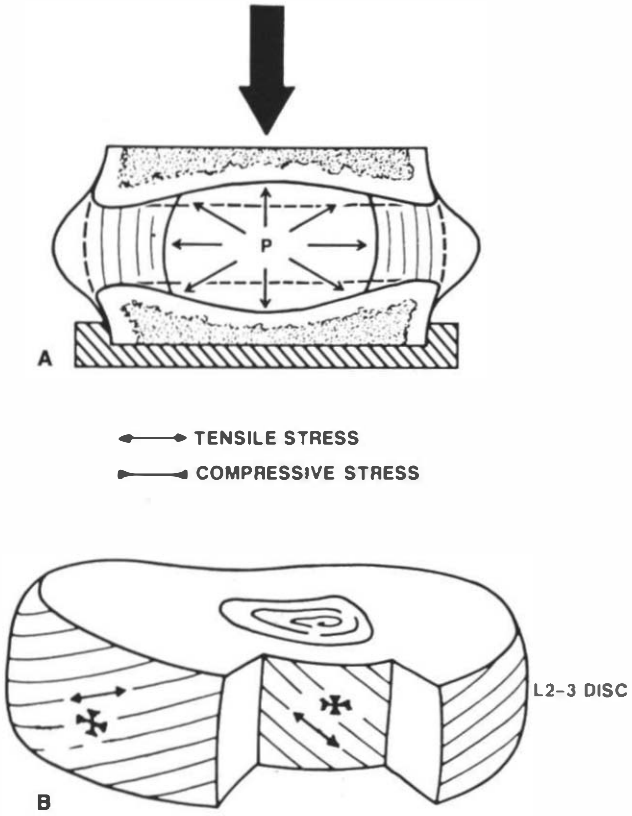


FIGURE 1-8 Non-degenerated disc under compression. **(A)** Pressure within the nucleus is produced because of compression. This pressure pushes the disc annulus and the two end-plates outward. The disc bulges out in the horizontal plane, and the end-plates deflect in the axial direction. **(B)** In the outer layers the stresses are small. Axial, circumferential, and radial stresses are compressive, while the annular fiber stresses are tensile. In the inner layers of the annulus, the axial, circumferential, and radial stresses are still compressive, but their magnitude is larger. The fiber stress is larger and still tensile. *(Based upon mathematical simulation by Shirazi-Adl, S. A., et al: Stress analysis of the lumbar disc-body unit in compression: a three-dimensional nonlinear finite element study. Spine, 9:120, 1984.)*

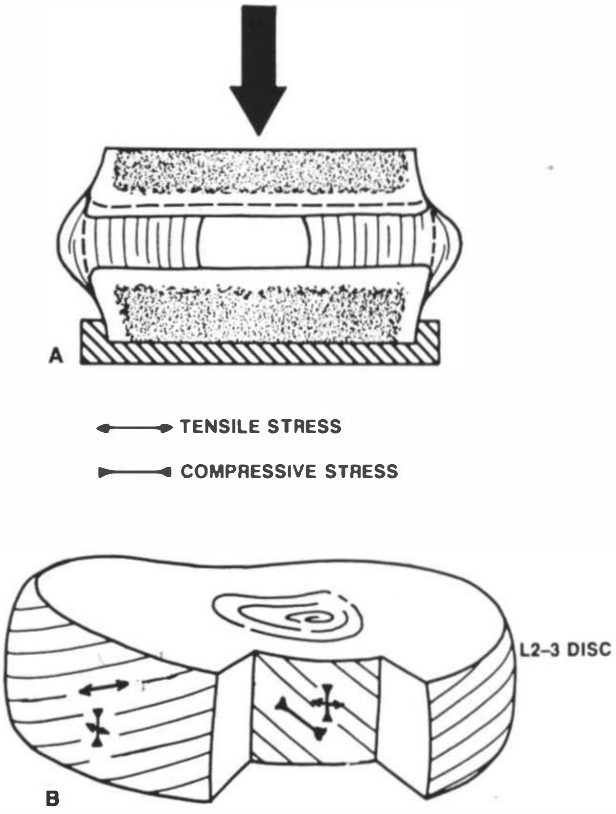


FIGURE 1-9 Degenerated disc under compression. **(A)** The compressive load is carried through a different mechanism. The load is transferred from one end-plate to the other by way of the annulus only, thus loading the end-plates. **(B)** In the outer layers of the degenerated disc, the axial stresses are compressive but smaller. The fiber stresses are tensile. In the inner layers, the stresses are about the same as those in the outer layers. The fiber stress is now compressive. *(Based upon mathematical simulation by Shirazi-Adl, S. A., et al.: Stress analysis of the lumbar disc-body unit in compression: a three-dimensional nonlinear finite element study. Spine, 9:120, 1984.)*

not capable of building sufficient fluid pressure. As a result, the end-plates are subjected to less pressure at the center, and the loads are distributed more around the periphery. The stresses in the annular ring are also changed. Compare Figure 1-9B and 1-3B. In the outer layers of the annulus of a degenerated disc, the axial as well as circumferential stresses are still small. The axial stress is compressive and the circumferential stress is near zero or tensile. In the inner layers of the annulus, the fiber stress is now compressive, while the axial stress is about the same as that on the outer layers. The circumferential stress is very small, annular stress is tensile and small, and the peripheral stress nearly vanishes. These findings are based upon computer simulations af an L2–L3 intervertebral disc subjected to compression by Shirazi-Adl and colleagues.[236]

In addition to the finite element model, which requires large computers and is expensive to run, there are also simpler models of the disc than can be run on personal computers.[39] These models, although incorporating less anatomic detail, are less expensive to run and, therefore, may be more suitable for broader investigations.

In conclusion, the stresses in the disc due to pure compression load are not large enough to cause disc failure. This has been confirmed by experiments in which such a load always caused end-plate failure.

Tension A detailed computer simulation of the disc under tensile loading has not been carried out. We present a highly simplified analysis. To determine the mechanism by which the disc is able to carry tensile loads, imagine the disc being cut by a plane that is perpendicular to the fiber directions of an annular lamina. (This is the technique of free-body analysis.) To support the tensile loads there are two types of stresses that are produced within the annulus—normal and shear, respectively perpendicular and parallel to the cut surface. The shear stresses are relatively larger in magnitude. Although the normal stresses are nicely absorbed by the alternating layers of annular fibers, there is no provision for resisting the shear stresses. Thus, the risk of disc failure is greater with tensile loading as compared with the compressive loading. Another difference between the two types of loadings is the change in the horizontal dimensions of the disc. Because of Poisson's effect, the disc bulges during compression and contracts in tension.

Bending The spine is subjected to tension on its convex side and compression on its concave side when bending loads are applied during flexion, extension, or lateral bending. One part of the disc is subjected to compression, while the other part is loaded in tension, as depicted in Figure 1-10A. Thus, bending loads can be thought of as a combination of the tensile and compressive loads, each applied to one-half of the disc.[C] The resultant effect on the disc is a combination of the effects due to the two load types. The side of the annulus subjected to tension contracts in the horizontal plane, while the side under compression bulges out. The annular fibers resist the compressive and tensile loads by the mechanisms already described. The complete mathematical stress analysis of the intervertebral disc

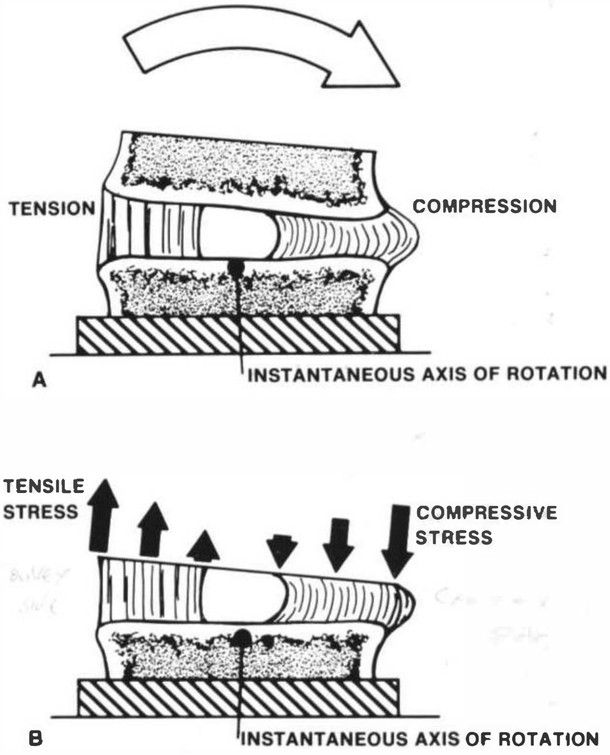


FIGURE 1-10 Disc stresses with bending. **(A)** During bending (flexion, extension, and lateral bending), one side of the annulus is subjected to compression while the other side is put under tensile load. The instantaneous axis of rotation separates the two zones. On the compression side the disc bulges, while it contracts on the tension side. **(B)** The stresses vary in magnitude from maximum in the outer laminae of the annulus to zero at the instantaneous axis of rotation.

subjected to bending has not been performed. However, a simplified version is presented in Figure 1-10B. The instantaneous axis of rotation separates the compressive and the tensile stress zones. Note the increase in stress from the inner toward the outer layers of the annulus.

Torsion When the disc is subjected to an axial torque, the stress distribution in the disc is dependent upon the degree of degeneration of the disc and whether or not the posterior elements are intact. Using a computer model, Shirazi-Adl and co-workers examined the stresses in the innermost annulus layer of an L2–L3 disc. They studied five cases: (1) nondegenerated disc; (2) same as (1) but with 2000 N compression; (3) same as (2) but with loss of intradiscal pressure; (4) same as (1) but with posterior elements removed; (5) same as (4) but with 2000 N compression. In all cases, 60 Nm of torque (upper limit of estimated physiologic torques *in vivo*) was applied. The following results were predicted.

In general, for all five cases, the tensile stresses in the fibers in the direction of the torque were maximum in the anterior half of the disc. Only in cases 1, 2, and 3 were the fibers in the direction opposite the torque stressed, and these were located at the posterior and posterolateral aspects of the disc. Thus, the same fibers in these locations are stressed in both the right and left spinal rotations. Removal of the posterior elements increased the fiber stresses at all locations around the periphery. Loss of intradiscal pressure decreased the stresses, while addition of compressive load did not significantly alter the stress pattern.

Shear To our knowledge, shear load has not been simulated. Therefore, our presentation is simple. This type of force acts in the transverse plane, perpendicular to the long axis of the spine. It probably produces shear stresses that are about equal in magnitude over the entire annulus and are zero on the surface and increase toward the center parallel to the applied shear force. There may also be tensile and compressive stresses at ±45° to the transverse plane.

These mathematical model simulations and simple free-body analyses show how various functional loading modalities have the potential to exert the greatest tensile stresses in the annular fibers. This mechanism of mechanical failure of annular fibers fits best with what is known about the biomechanical, structural, and anatomic characteristics of the intervertebral disc.

Strains in Annular Fibers

The stresses in the disc described above were obtained by constructing mathematical models of the disc and subjecting these to various loads in a computer. The stresses, in contrast to strains, cannot be experimentally measured. The strains (deformation per unit length) in annular fibers have been measured under the application of compression and torsional loads to a functional spinal unit, but without the posterior elements.[242] Seven microphotographic targets (0.8-mm diameter) were glued to the surface of the disc along a single annular fiber from one endplate to the other. Using stereophotogrammetric methods, three-dimensional measurements of the distances between the adjacent targets along the annular fiber were measured. This was done at no load and after the application of incremental loads. The loads studied were compression, up to a maximum of 2500 N, and torsion, up to 17 Nm. Stereophotographs were taken at each load increment. The photographs were digitized, and changes in lengths between the targets were computed. Graphs of fiber strains as functions of each of the two loads were plotted.

The fiber strains were found to be rather small, less than 3%, when the specimen was subjected to 2500 N of compression. Using mathematical models, Klein and co-workers computed fiber strain under compression loading and found it to be about 1%.[120] However, under the application of 12 Nm of torque, the fiber strains were considerably more, about 9%. This probably indicates that the significant compressive loads borne by the spine are carried not by the outer annular fibers but instead by the nucleus and the inner annular fiber layers. Also, it supports the theory that the annular fibers may be torn by the torsional load and not by the compression load.

Disc Bulge

One of the mechanisms of nerve root irritation is thought to be the root impingement by disc bulge. This clinical interest has led to several *in vitro* studies that have carefully measured the disc bulge. In the early studies, only compression load was applied,[102] but in later studies, the effects of various other loads have also been measured along several directions.[36,38,41,120,130,214,235]

In most of these *in vitro* studies, a fresh lumbar vertebra–disc–vertebra preparation with posterior elements removed for observation is used. The specimen is subjected to one of the loads, and the disc bulge is observed by any of several methods: dial gauges, specially made transducers, and photographic methods. Using loads of compression combined with flexion, extension, axial rotation, or lateral bending, the disc bulge was measured.[214] The largest disc bulge occurred in the mid-disc plane when the FSU was subjected to simultaneous compression and lateral bending. The disc bulged most in the lateral and posterolateral directions on the *concave* side of the lateral bend (Fig. 1-11). This value increased more than two times in the case of the degenerated discs. Considering that the space surrounding the nerve in the intervertebral foramen is rather limited, especially in the degenerated spines with decreased disc height,[190] the disc bulges may very well irritate the nerve root. This supports the hypothesis of Alf Breig, who postulated that the low back pain may be caused by constant rubbing of the nerve root surface by the protruding disc during physiologic motions of the spine.[35]

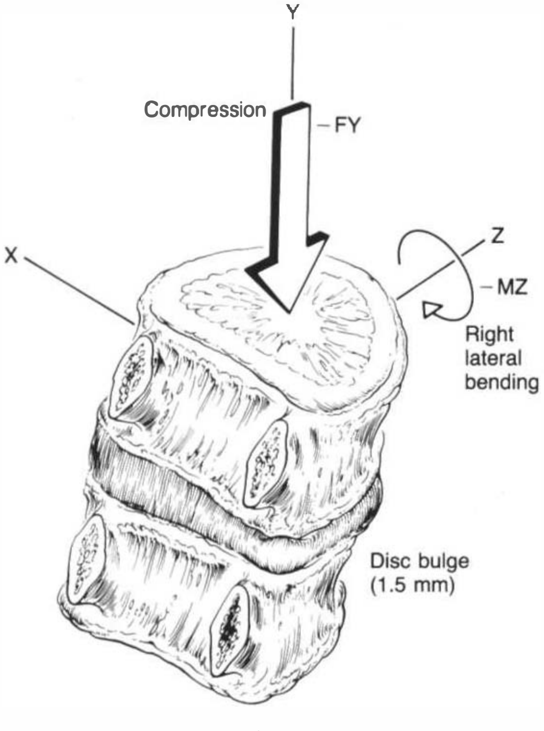


FIGURE 1-11 Maximum bulge in the posterolateral direction is produced because of lateral bending together with compression. The bulge is larger for degenerated discs and for discs with adjacent vertebral end-plate fractures. *(Data from Reuber, M., et al.: Bulging of lumbar intervertebral disks. J. Biomech. Eng., 104: 187, 1982; and Brinckmann, P.: Injury of the annulus fibrosus and disc protrusions.* Spine, *11:149, 1986.)*

In a recent study, the disc bulge was measured along all the directions in the transverse plane, providing a deformation contour in the mid-disc plane.[38] One of the interesting findings in this study was that the disc bulge increased after end-plate fracture when the specimen was subjected to compression loading. In a follow-up study, annular radial tears were simulated by producing a radial surgical division from the center of the disc toward the posterolateral direction. The cut was extended all the way to the periphery, except for the outermost layer of 1.1 mm average.[36] The presence of the radial tear increased the disc protrusion from 0.17 mm at 1000 N compression to 0.48 mm at the same load value. Additional compression fracture of the end-plate further increased the posterolateral bulge by 0.53 mm to a final value of 1.01 mm.

Internal Deformations of the Disc

Although we know precisely how the disc deforms (bulges) under a given load, as seen from the outside, the patterns of deformation within the disc are less well known. Most such information has been obtained from discographic studies. Using a novel method, Krag and co-workers have studied the internal deformations of the disc due to compression, flexion, and extension loads.[122] The method consisted of implanting 0.5-mm steel balls within the discs of L4–L5 functional spinal units, spread out in three rows in the midsagittal plane. Lateral x-rays provided direct visualization, and the measurements were taken by digitizing the images of the steel balls.

In compression, almost all of the internal displacements were in the anterior direction. In flexion, the central part of the disc (nucleus) moved posteriorly, while the peripheral region (annulus) moved anteriorly. In extension, almost the opposite was true (*i.e.*, the central portion moved anteriorly and the periphery of the disc moved posteriorly). After denucleation (surgical removal of the central portion of the disc), the displacements of the internal wall of the annulus were always toward the center of the disc, especially for the anterior portion of the annulus and in compression and flexion. This was so in spite of the increased outward bulging of the disc. Some authors suggest that such stretching of the disc annulus in the anteroposterior direction may lead to lamellar disruption within the disc. This information tends to contradict any suggestion that extension exercises might reduce posterior disc bulge. On the contrary, the extension could be aggravating.

Mechanisms of Disc Prolapse

By using fresh cadaveric lumbar spine segments, many attempts have been made to produce a disc prolapse that is similar to that seen clinically. Compressive loading of the disc, at slow speed or at high speed, is known to result in fractures of the endplates and not in failure of the disc.[93,203,254] This is true irrespective of the status of disc degeneration.[67] Torsional loading of the disc beyond its physiologic limits does result in circumferential tears in the annulus but does not result in disc prolapse.[67] Simple flexion of the specimen generally results in tearing of the posterior ligaments[215] or fractures of the laminae. Therefore, the experiments described below are of significant clinical importance. For the first time, disc prolapse was produced in a laboratory setting with the use of fresh cadaveric specimens. The disc prolapse was attempted by sudden load applications as well as by gradual loading.

Sudden Disc Prolapse The two-vertebrae specimen, laminectomized for observation of disc prolapse, was positioned such that the upper vertebra was laterally bent and hyperflexed so that the posterolateral aspect of the disc annulus was under tension.[3] This was on the side opposite the side of the lateral bend. In this set posture the specimen was suddenly loaded by a compressive force. The result was disc prolapse, similar to that seen clinically, in 26 of the 61 specimens tested.

A close scrutiny of the specimens showed that there was a certain pattern to the specimens that prolapsed and the ones that did not. The prolapse-prone specimens most likely came from the lower lumbar levels (L4–L5 or L5–S1), the 40–49-year age group, and disc degeneration with a grade of 2 (on a scale of 1–4) (Fig. 1-12). These sets of attributes seem to correlate well with the clinical picture of disc prolapse. In addition, these attributes also coincide with the instability stage of the degeneration hypothesis of Kirkaldy-Willis.[118] In this hypothesis, the first stage of degeneration may result in some spinal dysfunction but no instability. In the

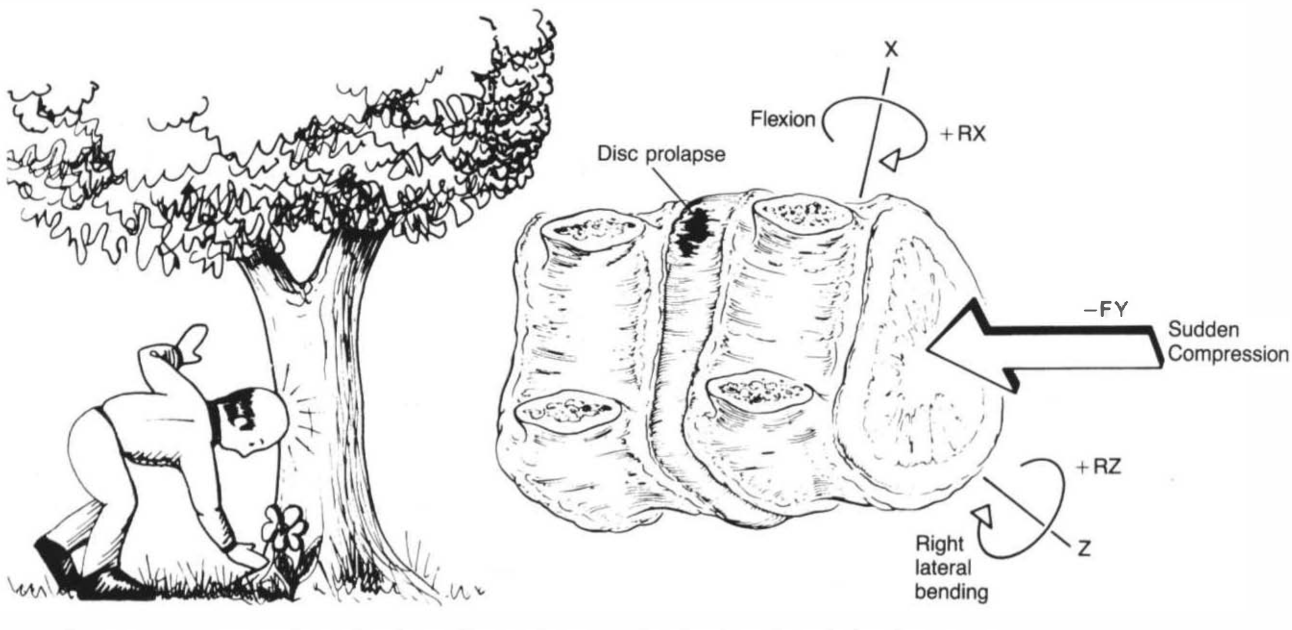


FIGURE 1-12 A mechanism of sudden disc prolapse. Using fresh cadaveric lumbar spine specimens, with posterior elements removed for observation of the disc, experimental disc prolapse was produced in 43% of the experimental trials. The method consisted of placing the specimen in a fully flexed and somewhat laterally bent posture (thus producing tension in the annular fibers) and applying a sudden compression load. The prolapse was produced on the side opposite the side of the lateral bend. Most susceptible discs were those at L5–S1, 40–50 years old, and with a degeneration grade of 2 (on a scale of 1 to 4). *(Data from Adams, M. A., and Hutton, W. C.: Prolapsed intervertebral disc. A hyperflexion injury. Spine, 7:184,1982.)*

third stage, the spine is restabilized, probably because of ligament calcification and osteophytes.

However, in the second stage, which occurs between the ages of 40 and 50 years, the disc degeneration has progressed to the point where the nucleus is still mobile. This is the instability stage. At this stage there is increased risk of disc prolapse at the L4–L5 or L5–S1 levels because of traumatic overload of the spine.

Gradual Disc Prolapse Since the majority of the low back pain patients with disc prolapse seen clinically do not report a precipitating traumatic event, an attempt was made to produce disc prolapse in the laboratory using slowly varying loads.[8]

Fresh cadaveric lumbar FSUs were positioned so that the application of axial compression resulted in simultaneous compression, flexion, and some lateral bending. Cyclically varying compression, between 1500 and 6000 N, was applied at a rate of 40 loadings per minute.

Of the 49 specimens tested, only 6 had gradual prolapses, 35 had end-plate fractures or vertebral collapse, and 8 did not fail at all. One may conclude that the gradual disc prolapse is most likely not caused by the hyperflexion loading used in this experiment. It may be the result of a combination of factors, such as weakened posterior disc annulus, relatively nondegenerated nucleus combined with relatively degenerated annulus with fissures, and another kind of loading (e.g., bending and twisting).

SPINAL LIGAMENTS

Ligaments are uniaxial structures; they are most effective in carrying loads along the direction in which the fibers run. In this respect, they are much like rubber bands. They readily resist tensile forces but buckle when subjected to compression. Nature has designed the spine in such a way that when the functional spinal unit is subjected to different complex force and torque vectors, the individual ligaments provide tensile resistance to external loads by developing tension.

The ligaments have many different functions. First, the ligaments must allow adequate physiologic motion and fixed postural attitudes between vertebrae, with a minimum expenditure of muscle energy. Second, they must protect the spinal cord by restricting the motions within well-defined limits. Third, they share with the muscles the role of providing stability to the spine within its physiologic ranges of motion. Finally, they must protect the spinal cord in traumatic situations in which high loads are applied at fast speeds. In these highly dynamic situations, not only is the displacement to be restricted within safe limits, but large amounts of energy that are suddenly applied to the spine must also be absorbed.

Biomechanically Relevant Anatomy

Anatomy of the ligaments of the upper cervical region (occiput to C2) is distinct and quite different from the rest of the spine,[61] and is described in Chapter 5. Here we describe the biomechanically relevant anatomy of the ligaments from C2 to the sacrum. The ligaments in this region are similar, although they vary in size, orientation, and attachment points. There are seven ligaments of the spine (Fig. 1-13). A short description of each of the ligaments arranged from anterior to posterior follows:

The *anterior longitudinal ligament* is a fibrous tissue structure that arises from the anterior aspect of the basioccipital and is attached to the atlas and the anterior surfaces of all vertebrae, down to and including a part of the sacrum. It attaches firmly to the edges of the vertebral bodies but is not so firmly affixed to the annular fibers of the intervertebral disc. The width of the anterior longitudinal ligament is diminished at the level of the disc. It is narrower and thicker in the thoracic region.

The *posterior longitudinal ligament* arises from the posterior aspect of the basioccipital, covers the dens and the transverse ligament (where it is called the membrana tectoria), and runs over the posterior surfaces of all the vertebral bodies down to the coccyx. It too is thicker in the thoracic region. It has an interwoven connection with the intervertebral disc. In contradistinction to the anterior longitudinal ligament, it is wider at the disc level and narrower at the vertebral body level.

The *intertransverse ligaments* pass between the transverse processes in the thoracic region and are characterized as rounded cords intimately connected with the deep muscles of the back.

The *capsular ligaments* are attached just beyond the margins of the adjacent articular processes. The fibers are generally oriented in a direction perpendicular to the plane of the facet joints. They are shorter and more taut in the thoracic and lumbar regions than in the cervical region.

The *ligamenta flava* extend from the anteroin-

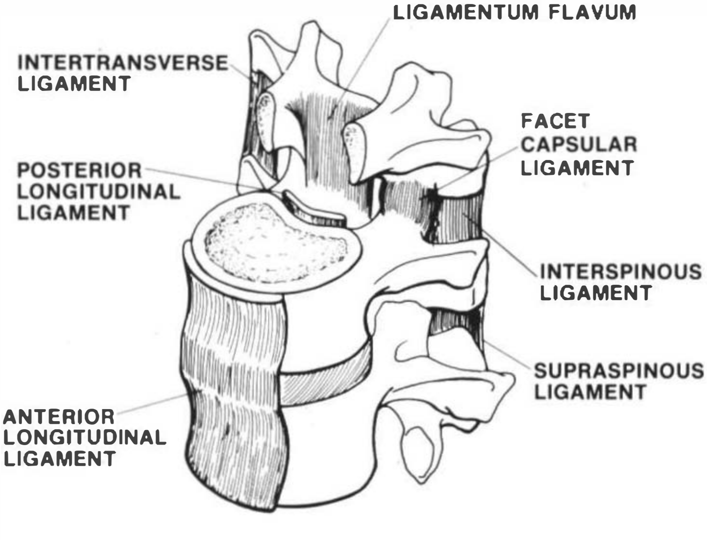


FIGURE 1-13 Ligaments of the spine. Besides the disc, there are seven ligaments that connect one vertebra to the next. Contribution to the spine stability by an individual ligament is dependent upon its cross-section, its distance from the instantaneous axis of rotation, and its orientation in space. The anatomy of the ligaments is such as to collectively provide stability to the spine in its various physiologic motions.

ferior border of the laminae above to the posterosuperior border of the laminae below. They connect the borders of adjacent laminae from the second cervical vertebra to the first sacral vertebra. These ligaments, referred to as yellow ligaments, are thicker in the thoracic region. Although they seem to be paired because of a midline cleavage, each is rather like a single structure that extends from the roots of the articular process on one side to the corresponding process on the other. The ligament is composed of a large amount of elastic fibers and represents the most pure elastic tissue in the human body. It has been noted, however, that with aging there is an increase in the relative amount of fibrous tissue.

The *interspinous ligaments* connect adjacent spines, and their attachments extend from the root to the apex of each process. They are narrow and elongated in the thoracic region, broader and thicker in the lumbar region, and only slightly developed in the neck.

The *supraspinous ligament* originates in the ligamentum nuchae and continues along the tips of the spinous processes as a round, slender strand down to the sacrum. It is thicker and broader in the lumbar region than in the thoracic region.

Quantitative Anatomy

In describing the functional role of ligaments, precise description of their anatomy is necessary. Unfortunately, such information is seldom available from the standard anatomic texts. For each ligament, this should include ligament length, cross-sectional dimensions, three-dimensional coordinates of the attachment points to the vertebrae, direction, and material constituents. Most of this type of data is presently not available. Limited information, concerning the cross-sectional areas and lengths of some of the ligaments,[46,62,84,160] is provided in Table 1-2.

TABLE 1–2 Representative Cross-sectional Areas and Lengths of the Spinal Ligaments

Region	Level	Ligament	Cross-sectional Area (mm^2)	Length
Cervical	C1–C2	Transverse	18	20
		Alar	22	11
Lumbar		ALL	53	13
		PLL	16	11
		LF	67	19
		CL	—	—
		ISL	26	—
		SSL	23	11

Key

ALL = anterior longitudinal ligament;
PLL = posterior longitudinal ligament;
LF = ligamentum flavum;
CL = capsular ligament;
ISL = interspinous ligament;
SSL = supraspinous ligament.

(Based upon data from Chazal, et al.,[46] Dvorak, et al.,[62] Goel, et al.,[84] Myklebust, et al.,[160] and Panjabi, et al.[188])

Physical Properties

The biomechanical functions of the spine described earlier are accomplished in part by the mechanical design of the individual ligaments and their locations and orientations with respect to the vertebrae to which they are attached. This architectural aspect of the ligaments is important and is discussed in the latter part of this section. Here, we describe the physical characteristics of the individual ligaments.

Besides the strength of a ligament, which is important during spinal trauma (and only then), there are other important characteristics that help provide the physiologic functions. One such characteristic is the nonlinearity of the load–displacement curve. A typical load–displacement curve of a ligament is shown in Figure 1-14. To quantify this behavior, we divide the load–displacement curve into three regions: (1) the neutral zone (NZ)—the displacement beyond the neutral position due to application of a small force; (2) the elastic zone (EZ)—the displacement beyond the neutral zone and up to the physiologic limit; and (3) the plastic zone (PZ)—beyond the elastic zone and until failure occurs. The neutral and elastic zones combined together constitute the physiologic range of motion, while the plastic zone is the region of increasing trauma. Viscoelasticity of a ligament is another important characteristic that is responsible for its shock-absorbing capacity. Little data are available concerning such characteristics of spinal ligaments.

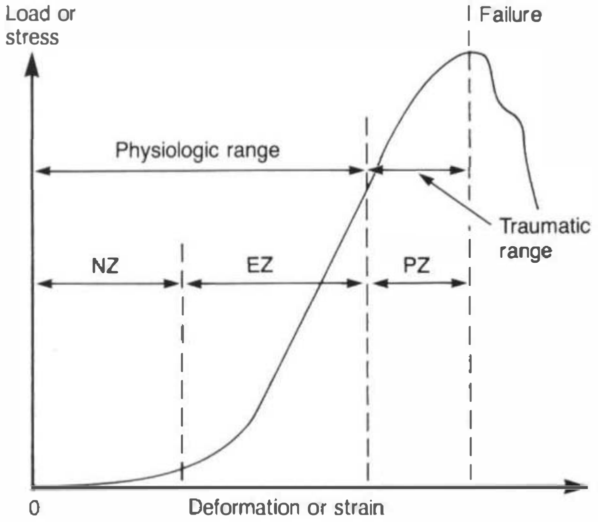


FIGURE 1-14 A typical load–deformation curve of a ligament, obtained in a materials testing machine. One end of the ligament is displaced with respect to the other, while the load and deformation of the ligament are continuously recorded. The deformation, being the independent parameter, is plotted on the horizontal axis. This load–deformation curve may be divided into physiologic and traumatic ranges. The physiologic range may be further divided into two parts. A ligament deformation around the neutral position where very little effort is required to deform the ligament is called the neutral zone (*NZ*). In the second part, a more substantial effort is needed to deform the ligament; we have called this the elastic zone (*EZ*). In the trauma range there is microtrauma with increasing load eventually leading to failure. This has been designated as the plastic zone (*PZ*).

Anterior and Posterior Longitudinal Ligaments

The anterior and posterior ligaments lie on the anterior and posterior surfaces of the disc and are attached to both the disc and the vertebral bodies. Therefore, these ligaments deform not only because of the relative separation between the two adjacent vertebrae but also because of bulging of the disc.

Several functions and characteristics have been attributed to the longitudinal ligament. Traction at the attachment points may produce "anterior lipping" of the vertebrae, as seen clinically.[10] Longitudinal ligaments degenerate with age, as does the disc.[17,94] Roaf claimed that it is not possible to disrupt the anterior longitudinal ligament by flexion or extension of the spine, although it could be accomplished by rotation.[215]

Tkaczuk did an extensive study (484 samples) of the tensile characteristics of both the anterior and posterior longitudinal ligaments of the lumbar spine with the purpose of examining the influence of degeneration and age on the biomechanical properties.[250]

In one set of experiments, specimens of a standard size were used.[D] These samples were loaded up to one-third of the failure load, and the load–deformation curves were plotted. Three parameters were measured: the maximum deformation, the residual or permanent deformation, and the energy loss of hysteresis. All the biomechanical parameters were found to decrease with age. The greatest decrease was found in the energy absorption values. This documents the decrease in the shock-absorbing characteristics of the ligaments with age. Regarding the degenerative changes, Tkaczuk found that the maximum, as well as the residual deformations were lower for the degenerated discs.

In another set of experiments, intact ligament samples were tested to failure.[D] The load–deformation curves were found to be similar in shape to the stress–strain curves of the ligamentum flavum.[164] This implies that the spinal ligaments are similar in functional design. The anterior longitudinal ligament was found to be twice as strong as the posterior ligament. But the *material* properties of the two ligaments were nearly the same, as shown by the strength values of the samples of *equal* cross-section (failure stress) taken from the two ligaments. Just as in the ligamentum flavum, there was some pre-tension present in these ligaments. It was estimated to be about one-tenth of that of the ligamentum flavum.

There have been several follow-up studies, however, with significant variability in the results.[46,57,84,160,186] The variability has been due to age, difficulties in identifying the boundaries of the ligaments from those of the disc, and variations in the experimental techniques. The values given in Table 1-3 are average values from several studies.

Intertransverse Ligaments

The intertransverse ligaments have no mechanical significance in the lumbar region because of their negligible cross-sectional size. They are present in the thoracic region where they replace intertransverse muscles. The thoracic intertransverse ligaments were found to have higher failure stress but

TABLE 1–3 Failure Strength of Spinal Ligaments

	Load (N)		Deformation (mm)		Stress (MPa)		Strain (%)	
	Average	Range	Average	Range	Average	Range	Average	Range
Upper Cervical								
C0–C1								
Ant. atlantooccip. memb.	233		18.9					
Post. atlantooccip. memb.	83		18.1					
C1–C2								
ALL	281		12.3					
Atlanto-axial membrane	113		8.7					
CL	157		11.4					
Transverse ligament	354	170–700						
C0–C2								
Apical	214		11.5					
Alar	286	215–357	14.1					
Vert. cruciate	436		25.2					
Tectorial membrane	76		11.9					
Lower Cervical								
ALL	111.5	47–176	8.95	4.2–13.7				
PLL	74.5	47–102	6.4	3.4–9.4				
LF	138.5	56–221	8.3	3.7–12.9				
CL	204	144–264	8.4	6.8–10				
ISL	35.5	26–45	7.35	5.5–9.2				
SSL								
Thoracic								
ALL	295.5	123–468	10.25	6.3–14.2				
PLL	106	74–138	5.25	3.2–7.3				
LF	200	135–265	8.65	6.3–11				
CL	168	63–273	6.75	3.9–9.6				
ISL	75.5	31–120	5.25	3.8–6.7				
SSL	319.5	101–538	14.1	7.2–21				
Lumbar								
ALL	450	390–510	15.2	7–20	11.6	2.4–21	36.5	16–57
PLL	324	264–384	5.1	4.2–7.0	11.5	2.9–20	26.0	8–44
LF	285	230–340	12.7	12.0–14.5	8.7	2.4–15	26.0	10–46
CL	222	160–284	11.3	9.8–12.8	7.6	7.6	12.0	12.0
ISL	125	120–130	13.0	7.4–17.8	3.2	1.8–4.6	13.0	13.0
SSL	150	100–200	25.9	22.1–28.1	5.4	2.0–8.7	32.5	26–39

ALL = anterior longitudinal ligament
PLL = posterior longitudinal ligament
LF = ligamentum falvum
CL = capsular ligament
ISL = interspinous ligament
SSL = supraspinous ligament

(Data from Chazal, et. al.,[46] Dvorak, et al.,[62] Goel, et al.,[84] Myklebust, et al.,[159] Nachemson and Evans,[164] Panjabi, et al.,[186] and Tkaczuk.[250])

smaller failure strain compared with other ligaments.[46] This seems to be compatible with the restricted intervertebral movements in the thoracic region.

Capsular Ligaments

The contribution of the capsular ligaments in providing flexion stability has been proved in the cervical spine.[192, 263] It has also been shown that under certain conditions of axial loading, such as in pilot ejections from aircraft, the capsular ligaments in the thoracolumbar region are stretched.[207]

The capsular ligaments are generally perpendicular to the joint line. In the lumbar region the joint line is nearly vertical, and, therefore, the ligaments are oriented horizontally. In a recent study, the capsular ligaments were tested using a bone–ligament–bone preparation by applying tensile loads along the direction of the ligament fibers.[186] The results are presented in Table 1-3.

Ligamentum Flavum

The function and importance of the ligamentum flavum in humans has been a matter of discussion from the beginning of this century.[71, 243]

Although some biomechanical aspects of this ligament were explored earlier,[9, 172] only recently have modern tissue handling techniques and mechanical testing machines been used.[46, 84, 160, 164, 186] In the study by Nachemson and Evans, ten specimens of ligamentum flavum and attached laminae, from L3–L4 FSUs, were tested.[164] The specimens were loaded in tension along the spine axis, and the tests were performed at slow speed in an Instron testing machine.

While separating the vertebral laminae from the bodies, Nachemson and Evans found the ligamentum flavum to have pre-tension (the tension present *in situ* when the spine is in neutral position). This "resting" tension in the ligament produces "resting" compression of the disc. The value of these resting tensions was found to decrease with age from about 18 N (4.5 lbf) in the young (<20 years) to about 5 N (1.1 lbf) in the older (>70 years) subjects. These "resting" tensions most probably have some function. Perhaps they prevent protrusion of the ligament into the spinal canal during full extension of the spine when the ligamentum flavum is slack. Also, the resting compression in the disc may add some stability to the spine. Detailed physical properties of this ligament are given in Table 1-3.

Histologically, the ligamentum flavum has the highest percentage of elastic fibers of any tissue in the body.[42, 164] This allows a large amount of extension of the ligament without permanent (residual) deformation. Clinically, this is an important characteristic. In a situation when the spine suddenly goes from full flexion (ligament stretched) to full extension (ligament relaxed), the high elasticity of the yellow ligament, together with its pre-tension, minimizes the chances of any impingement of the spinal cord.

Interspinous and Supraspinous Ligaments

The supraspinous ligament, because of its closeness to skin, was the first, and probably the only, ligament studied *in vivo*.[237] The tension was measured by inserting pins under local anesthesia into the supraspinous and interspinous ligaments between L3 and L4, from the posterior direction, in the sagittal plane. The amount of sideways motion of the pin under the application of a given force was taken as a measure of the tension in the ligaments. It was found that the tension gradually increased as the spine was flexed, reaching its maximum at full flexion. The physical properties of these ligaments have not been studied recently[46, 57, 84, 159, 160] and are summarized in Table 1-3.

Failure Modes: Ligament Versus Bone

The ligaments transfer tensile loads from bone to bone. When they are subjected to large loads *in situ*, the failure may occur either within the ligaments or in the bone at the point of attachment. On what factors does this pattern of failure depend? To our knowledge, there are no studies specifically conducted in the spine that have addressed this question. Noyes and colleagues, using cruciate ligaments of the Rhesus monkey, have reported that the failure pattern (ligament vs. bone) depends upon the rate of application of the loads and the status of the bone.[170, 171]

They conducted tensile tests to failure on bone–cruciate ligament–bone preparations. The tests compared slow and fast rates of loading along with specimens from normal animals and specimens from those immobilized for 6 months. In specimens from normal animals, bone failed more often at slow rates of loading, and ligament failed at high rates of loading. The preparations taken from immobilized animals, tested only at the fast loading rate, always failed by bone avulsion. The general concept here is that in any series structure like the bone–ligament–bone preparation, failure occurs at the weakest point. It has been well documented that the bone and ligament strengths increase with the rate of loading.[151, 170, 193] We may conclude that there is a relatively greater increase in strength for the bone than for the ligament with an increase in the rate of loading, accounting for the ligament failure at higher rates of loading. Furthermore, immobilization decreases the strength of the bone to a greater degree than it does that of the ligament, leading to bone failure in the experiments.

Functional Biomechanics

As seen in Figure 1-15A, the shape of the load–displacement curve of each of the ligaments is quite similar. Each ligament is characterized by its unique combination of stiffness (the slope), maximum deformation, and failure load. These variations reflect the specific functional role of each of the ligaments.

Function of a Ligament

The functional properties of a ligament are a combination of its physical properties, as described above, and its orientation and location with respect to the

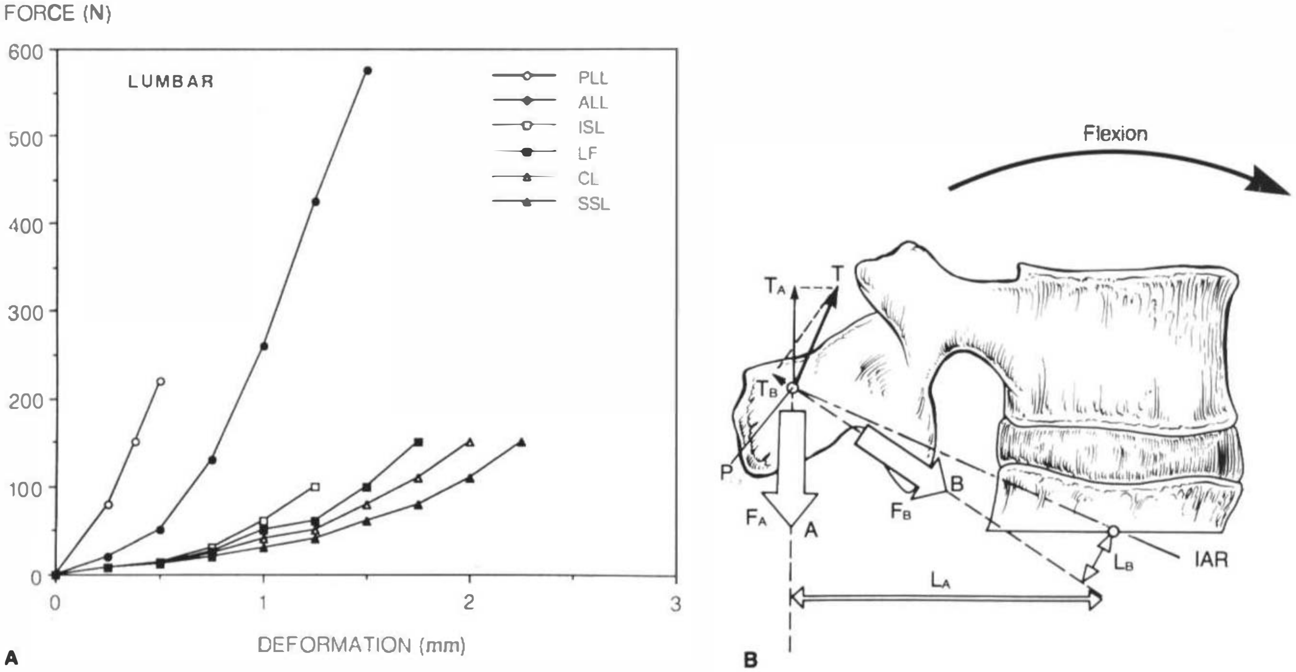


FIGURE 1-15 (A) Force–deformation curves of spinal ligaments of the lumbar region. Notice the nonlinearity of each of the curves (i.e., an initial phase in which a small force produces large deformation and a latter phase in which considerably larger force is required to produce the same deformation). *PLL*=posterior longitudinal ligament; *ALL*=anterior longitudinal ligament; *ISL*= interspinous ligament; *LF*= ligamentum flavum; *CL*=capsular ligament; *SSL*=supraspinous ligament. (B) Stabilizing function of a spinal ligament. Consider two ligaments, A and B, attached at point P to the moving vertebrae and having the same mechanical properties but oriented differently. As the spine flexes, the resistance provided by the two ligaments is proportional to the ligament force and the lever arm. Assuming that the moving vertebra is rotating around the instantaneous axis of rotation (IAR) as shown in the figure, then the resistance provided by ligament A will be $F_A \times L_A$. Similarly, the resistance offered by ligament B will be $F_B \times L_B$. If the ligaments applied equal forces, the resistance due to ligament A would be greater because $L_A > L_B$. In reality, the force F_A will be bigger than the force F_B because of the greater deformation of ligament A, again because $L_A > L_B$. See note M at the end of the chapter for further explanation. *(Data for curves in A from Panjabi, M. M., et al.: Physiological strains in lumbar spinal ligaments, an in vitro biomechanical study. Spine, 7:192, 1982.)*

moving vertebra. For example, a ligament with a larger lever arm provides greater stability to the spine than one with a shorter lever arm. This principle is exemplified in Figure 1-15B. A vertebra with two ligaments (A and B) attached to it is shown. The spine is flexing, causing the vertebra to rotate around an instantaneous axis of rotation (IAR) with respect to the vertebra below, which is relatively fixed. This generates forces F_A and F_B, respectively, in each of the two ligaments. Assuming, for the moment, that the forces generated in each of the ligaments are of equal magnitude, then the resistance to the flexion movement provided by each ligament is directly proportional to the lever arms (*i.e.*, L_A and L_B, respectively). (In reality, the force F_A is bigger than F_B. See details in Note M at the end of this chapter.) Thus, the ligament A will provide greater stability to the spine than the ligament B, even though both are of equal size and have equal physical characteristics. As can be seen in Figure 1-15B, the function of a ligament is dependent upon the location of the instantaneous center of rotation. Shifting this center changes the lever arm. This implies that the function of a ligament may be different in different movements of the spine and even in the same movement as the IAR changes.

Let us look at a simple example of the anterior longitudinal ligament. In flexion, this ligament is subjected to compression and therefore provides no resistance to motion. In extension, on the other hand, the movements of its attachment points to the vertebrae are such that they produce tension in the ligament, thus providing resistance to motion and stability to the spine. Determination of the functional role of each spinal ligament in various spinal movements becomes quite complex. Therefore, it is best done with the help of a mathematical model in a computer. Such an approach has been used and ligament functions determined.[184]

In general, the load–displacement curve is nonlinear and the slope increases with the load. A theoretical analysis, based upon the special shape of this curve, is presented here with the purpose of explaining the three biomechanical functions of a ligament. The analysis is limited to sagittal plane motion and uses the ligamentum flavum as an example. The conclusions of the analysis, with some modifications, can be applied to other ligaments.

A typical stress–strain curve for the ligamentum flavum, reported by Nachemson and Evans,[164] is shown in Figure 1-16. The stress is the load applied per unit area. The strain is the percentage of elongation of the ligament from its unstretched length. Motions of the spine and corresponding deformations of the ligamentum flavum are depicted in Figure 1-16A and B, respectively.

The ligamentum flavum is located posterior to the axes of rotation of flexion/extension; it contracts with extension of the spine and elongates with flexion of the spine.

Panjabi and colleagues conducted a study to determine the *in situ* strains in the lumbar spinal ligament during physiologic motions.[184] Among the findings, the strains in the ligamentum flavum as a function of the physiologic flexion/extension motion were obtained. (Details are given in a later section.) Based upon these findings, the stress–strain curve of the ligamentum flavum was marked to illustrate the functional biomechanics of a ligament (Fig. 1-16B). Starting from the neutral position, there is a decrease in the length of the ligamentum flavum of 13% at full extension. Because the ligamentum flavum has 10% of pre-strain, this results in 3% compression. This is not enough, in normal spines, for the ligament to buckle into the spinal canal and cause clinical problems.

Full flexion of the spine from the neutral position resulted in a 16% increase in the length. Thus, the complete physiologic range of deformation of this ligament was from 3% compression to 26% tension. An additional 21% of stretch, due to further flexion of the spine, resulted in failure.

We have used average values in our example. The results may be different for the degenerated spines. Nachemson and Evans noticed a significant decrease in the prestrain with age. Penning and Wilmink, using myelogram, have reported buckling of the ligamentum flavum into the spinal canal with physiologic extension.[202]

From the curve, the average values of three parameters have been calculated: the force required, the energy stored, and the stiffness of the ligament during the various ranges of motion. A value of 100 was assigned to the average magnitudes of these quantities during the trauma range, and the relative values for the extension and flexion ranges are given in Table 1-4.

The numbers in the table clearly show that during the physiologic ranges of motion, a very small force (1.4 to 5.7) is required to move the spine. There is not much resistance (stiffness 0.4 to 1.3), and not much energy (3.4 to 14.7) is expended to produce

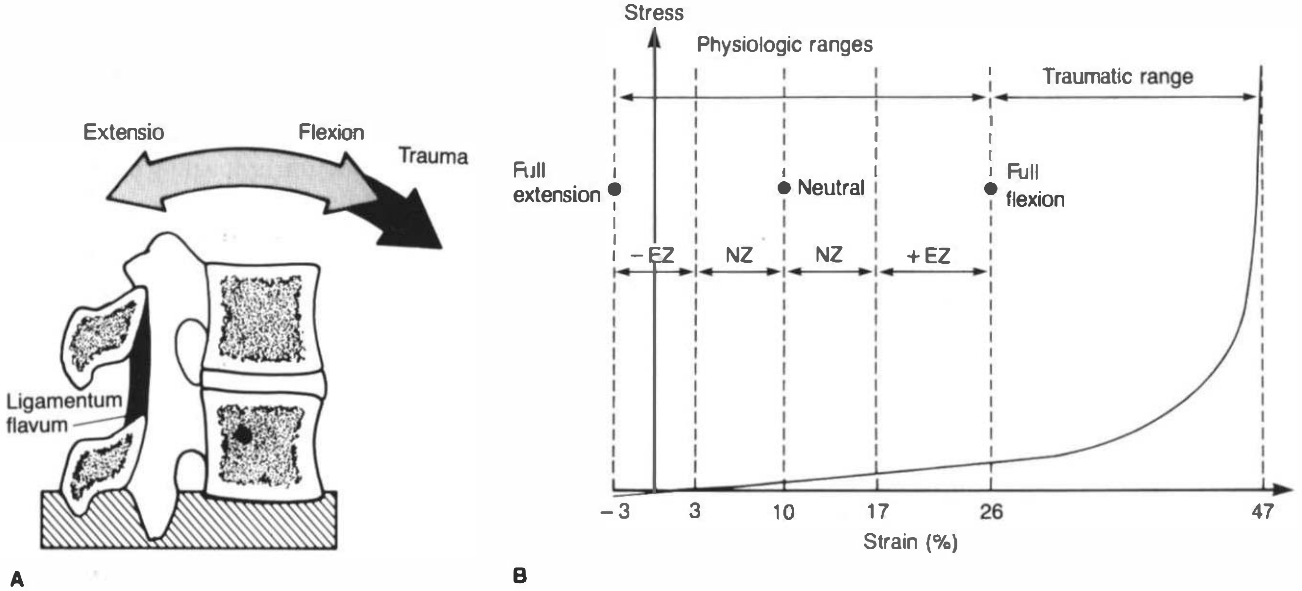


FIGURE 1-16 The functional biomechanics of a ligament are exemplified by the ligamentum flavum undergoing spine motion. (A) Ligamentum flavum in flexion/extension. In flexion of the spine, the ligamentum flavum is stretched, and in extension it contracts. This is due to its location with respect to the instantaneous axes of rotation during these motions. In hyperflexion such as trauma, this ligament may be stretched beyond its elastic limit to failure. (B) A stress–strain curve of deformation of the ligamentum flavum. In the neutral position of the spine, the ligamentum flavum has about 10% of prestrain (i.e., if the ligamentum flavum is transected *in situ*, it will contract by about 10% of its length). During full extension, the ligamentum flavum contracts about 13%, resulting in compression of about 3%. During full flexion of the spine from the neutral position, the ligamentum flavum is stretched by about 16% of its length. Thus, within the physiologic range, the strain in the ligamentum flavum varies from about 3% of compression to 26% of tension. Loaded beyond its physiologic range because of trauma, the ligamentum flavum fails at about 47% of stretch. *(Based upon the experimental findings of Nachemson, A., and Evans, J.: Some mechanical properties of the third lumbar inter-laminar ligament (ligamentum flavum). J. Biomech., 1:211, 1968; and Panjabi M., Goel, V., and Takata, K.: Physiological strains in lumbar spinal ligaments. Spine 7:192, 1982.)*

this useful motion. However, this smooth and efficient motion is effectively limited by the sharp increase in stiffness as the curve leaves the physiologic range. This is shown by the average stiffness in the trauma range, which is 77 times that in the flexion range. Thus, the two functions required of ligaments in the physiologic range are accomplished by the specific shape of the load–deformation curve. When large flexion loads are applied to the spine so that a traumatic situation exists, the shape of the load–deformation curve is such that large amounts of energy are absorbed before failure. Nearly seven times more energy is absorbed in the trauma range as compared with the flexion range.

TABLE 1–4 Mechanical Parameters of the Ligamentum Flavum in the Various Ranges of Motion. (The values are given as percentage of the trauma range.)*

	Physiologic Range		Trauma Range
	Extension	Flexion	
Average force	1.4	5.7	100.0
Energy stored	3.4	14.7	100.0
Average stiffness	0.4	1.3	100.0

* See Figure 1–16B.

(Data from Nachemson, A., and Evans, J.: Some mechanical properties of the third lumbar inter-laminar ligament [ligamentum flavum]. J. Biomech., 1:211, 1968, and Panjabi, M. M., Goel, V. K., and Takata, K.: Physiological strains in lumbar spinal ligaments, an *in vitro* biomechanical study. Spine, 7(3): 192, 1982.)

This analysis clearly shows the means by which the ligamentous mechanism enables the spine to perform two quite different roles: allowing smooth motion within the physiologic range, with a minimum of resistance and expenditure of energy, and at the same time providing a maximum of protection to the spinal cord by restricting motion and absorbing a significant amount of energy in traumatic situations.

Physiologic Strains in Ligaments

Having looked at the theoretical concepts concerning the function of a ligament, we would now like to present experimental data about the real-life deformations of the ligaments during physiologic motions of the spine. Physical properties of isolated ligaments (bone–ligament–bone) provide excellent information concerning that individual ligament, but these studies are less helpful in defining the *in situ* role of each ligament. *In vivo* studies are difficult, except for a ligament that is closer to the skin (e.g., the supraspinous ligament). Therefore, other avenues must be explored. Using an *in vitro* model, quantified ligament anatomy, and a mathematical model, the *in situ* behavior of lumbar spinal ligaments has been studied.[184] First, three-dimensional physiologic intervertebral motions were recorded. Second, the attachment points of each ligament on the moving and stationary vertebrae were quantified in terms of **x**, **y**, and **z** coordinates. Finally, the two data sets—the three-dimensional physiologic motions and the quantified anatomy—were combined using a mathematical model to yield deformations of each ligament as functions of the physiologic vertebral motions.

Each ligament was found to stretch or compress depending upon the particular physical motion. Figure 1-17 shows the average (only tensile) ligament strains due to the physiologic motions. In flexion, significant strains were produced in all ligaments except the anterior longitudinal ligament. In extension, it was the anterior longitudinal ligament that was most stretched. (These findings have been confirmed by direct measurement of strains in longitudinal ligaments using very thin mercury-filled rubber tubes.[98]) In axial rotation, depending upon the direction of rotation, one of the capsular ligaments was maximally stretched—right ligament with right axial rotation, and vice versa. In lateral bending, it was the ligamentum flavum that had the highest strain—the left ligament with the right bending, and vice versa.

This type of information is helpful in some clinically relevant instances. For example, in the case of a patient involved in trauma who has torn supra-

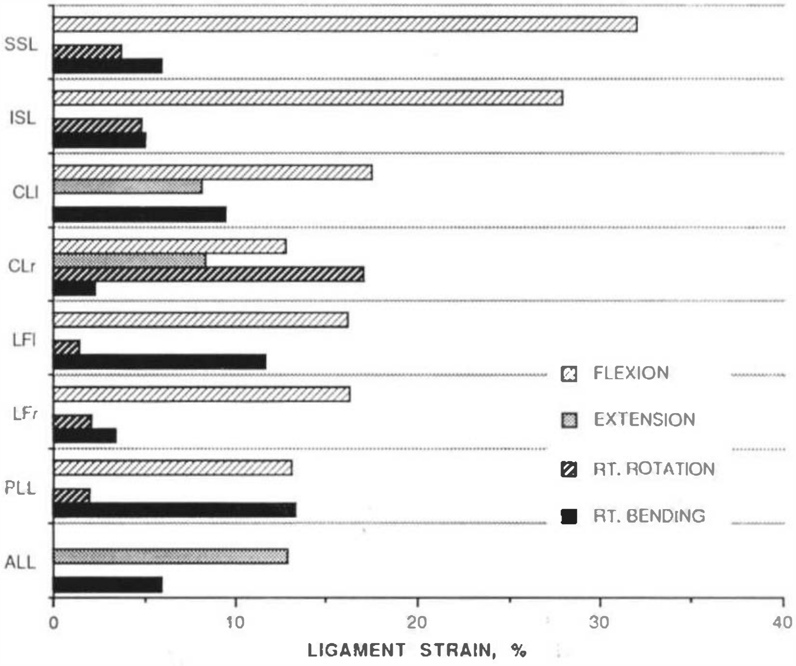


FIGURE 1-17 Physiological strains in lumbar spinal ligaments are shown as functions of the four spinal motions. The ligament strain is defined as the percentage change in its length. *SSL*= supraspinous ligament: *ISL*= interspinous ligament; *CL*= capsular ligament; *LF* = ligamentum flavum; *PLL*= posterior longitudinal ligament; *ALL*= anterior longitudinal ligament. Suffixes *l* and r are respectively left and right. Note that the supraspinous ligament is strained most in flexion, the right capsular ligament in right axial rotation, and the left ligament flavum in flexion right lateral bending. *(Data from Panjabi, M. M., Goel, V., and Takata, K.: Physiological strains in lumbar spinal ligaments. Spine, 7:192, 1982.)*

spinous and/or interspinous ligaments, one is justified in theorizing that it was a hyperflexion injury. In another case, if one finds that it is the right capsular ligament that is disrupted, then one may suspect that the mechanism of injury was a right axial rotation.

Another interesting finding of this study was a significant increase in the amount of neutral zones for the sagittal and horizontal plane rotations due to disc degeneration. This suggests that a degenerated spine may carry a higher risk for the increased ligament strains during these particular physiologic movements.

THE VERTEBRA

Probably the earliest biomechanical study concerning the human spine is that of the strength measurements of the vertebrae, conducted by Messerer over 100 years ago.[153] Since that time a good deal more has been learned about the mechanical properties of the human vertebrae.

Biomechanically Relevant Anatomy

Vertebra A vertebra consists of an anterior block of bone, the vertebral body, and a posterior bony

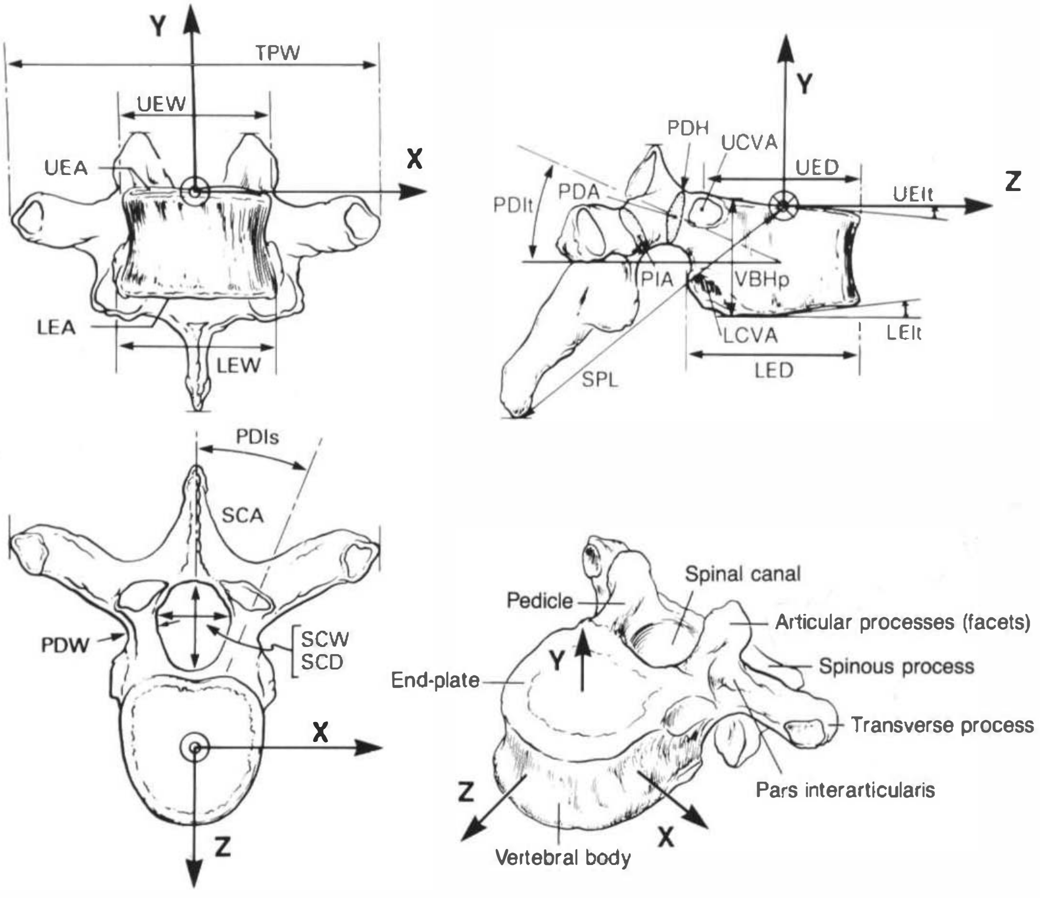


FIGURE 1-18 Quantitative anatomy of a vertebra. Each dimension is defined by a mnemonic consisting of three capital letters followed by a single lowercase letter, when needed. The first two letters represent the name of the anatomic part; the third letter represents the dimension. *TP* = transverse process; *UE* = upper end-plate; *LE* = lower end-plate; *PD* = pedicle; *SP* = spinous process; *SC* = spinal canal; *PI* = pars interarticularis; *VB* = vertebral body; *W* = width; A = area; *D* = depth; *H* = height; *I* = inclination. Suffixes are: t = transverse plane; p = posterior. Values for these dimensions for some representative vertebrae are given in Table 1-5. (*Data from Panjabi, M. M., et al.: Thoracic vertebrae. Quantitative three-dimensional anatomy. Spine, 1990, [in press].*)

ring, known as the neural arch, containing articular, transverse, and spinous processes (Fig. 1-18). The vertebral body is a roughly cylindrical mass of cancellous bone contained in a thin shell of cortical bone. Its superior and inferior surfaces, slightly concave, are the vertebral end-plates. The neural arch consists of two pedicles and two laminae, from which arise seven processes.

Although the basic design of the vertebrae in the various regions of the spine from C3 to L5 is approximately the same, the size and mass of the vertebrae increase from the first cervical to the last lumbar vertebra. This is a mechanical adaptation to the progressively increasing compression loads to which the vertebrae are subjected. There are also other differences. In the cervical region of the spine, there are foramina for the vertebral arteries. The thoracic vertebrae have articular facets for the ribs, and the lumbar spine has mammary processes. Of course the sacral spine, being fused, is unique.

Standard anatomic texts provide visual descriptions of vertebral anatomy[85,255] but seldom any quantitative dimension. The latter type of information is necessary as the biomechanics research becomes more widespread and the clinical practice more precise. Although some quantified anatomic data have been available for some time, detailed quantified three-dimensional vertebral geometry has been obtained only recently.[31,79,132,182] The total data (all dimensions of vertebrae from C2 to L5) are too numerous to be included here. We have selected some important dimensions and some representative vertebral levels. The nomenclature is presented in Figure 1-18, and the data are provided in Table 1-5. It is important to remember that the shape, size, and physical properties of the vertebrae change with age. A description of the changes with aging in the vertebral end-plate has been provided.[30]

It is sometimes necessary to use animal models to study certain phenomena (e.g., healing and fusion).

TABLE 1–5 Quantitative Anatomy of Vertebrae. Average Values of Vertebral Dimensions Shown in Figure 1-18 for Thoracic Vertebrae From T1 to T12.

	T1	T2	T3	T4	T5	T6	T7	T8	T9	T10	T11	T12
UEW (mm)	24.5	24.9	24.6	24.5	24.9	26.2	27.8	29.5	30.6	31.9	34.9	39.0
UED (mm)	18.5	19.6	22.7	23.3	24.3	26.0	27.4	27.9	29.3	30.5	31.9	32.8
LEW (mm)	27.8	27.4	25.9	26.0	27.0	28.2	29.1	30.5	33.0	35.4	39.1	42.1
LED (mm)	19.7	21.6	23.3	24.5	25.8	26.9	28.5	29.4	31.0	31.6	31.8	33.4
VBHp (mm)	14.1	15.6	15.7	16.2	16.2	17.4	18.2	18.7	19.3	20.2	21.3	22.7
UEA (mm^2)	300.	333.	373.	381.	426.	483.	547.	605.	678.	727.	842.	954.
LEA (mm^2)	376.	398.	412.	444.	495.	552.	603.	664.	755.	834.	945.	1024.
UEIt (degrees)	0.8	1.7	2.4	1.5	2.1	2.1	1.6	1.3	0.9	0.5	2.7	2.2
LEIt (degrees)	3.9	1.8	2.1	2.0	1.8	2.0	2.3	1.2	1.2	2.2	1.8	2.0
SCW (mm)	21.8	19.5	18.3	17.0	17.1	17.3	17.3	17.7	17.9	18.2	19.4	22.2
SCD (mm)	16.4	15.3	15.9	16.2	16.3	16.5	16.1	15.9	15.7	15.5	16.0	18.1
SCA (mm)	213.	200.	189.	192.	201.	206.	199.	194.	200.	202.	220.	280.
PDW (mm)	8.2	8.4	7.0	5.5	6.2	6.0	6.5	6.7	7.6	8.3	8.8	8.8
PDH (mm)	9.3	11.1	11.8	11.9	11.2	12.0	11.8	12.5	13.9	14.7	16.9	16.5
PDA (mm^2)	52.2	46.3	38.1	32.5	31.6	3.5	36.8	43.8	52.3	64.8	88.4	90.9
PDIs (degrees)	28.1	28.9	22.5	21.8	20.2	19.4	23.4	22.5	19.3	14.4	12.9	8.0
PDIt (degrees)	4.6	16.5	8.1	6.4	8.6	7.0	10.9	12.1	8.3	6.8	8.9	4.8
SPL (mm)	50.1	52.1	51.7	51.1	52.1	53.8	50.5	52.8	51.3	49.3	45.6	47.4
TPW (mm)	75.3	69.4	60.8	56.9	61.1	61.3	60.4	59.9	59.3	58.4	52.2	46.9

Key: The first two letters indicate anatomic part; the third letter indicates dimension. Figure 1–18 depicts the anatomy of a vertebra in detail.

UE = upper end-plate
LE = lower end-plate
PD = pedicle
SP = spinous process
SC = spinal canal
TP = transverse process
PI = pars interarticularis
VH = vertebral body

W = width
A = area
D = depth
H = height
I = inclination
t = transverse plane
p = posterior

(Based upon data from Berry, et al.,[31] Cotterill, et al.,[47] and Panjabi, et al.[182])

A comparative anatomy of human and bovine spine is available.[47]

Facet Joints The pattern of movements of the spine is dependent, among other factors, upon the shape and position of the articulating processes of the diarthrodial joints. It is the orientation of these joints in space that determines their mechanical importance. Figure 1-19 helps to visualize the changing pattern of the facet orientations, from the inferior

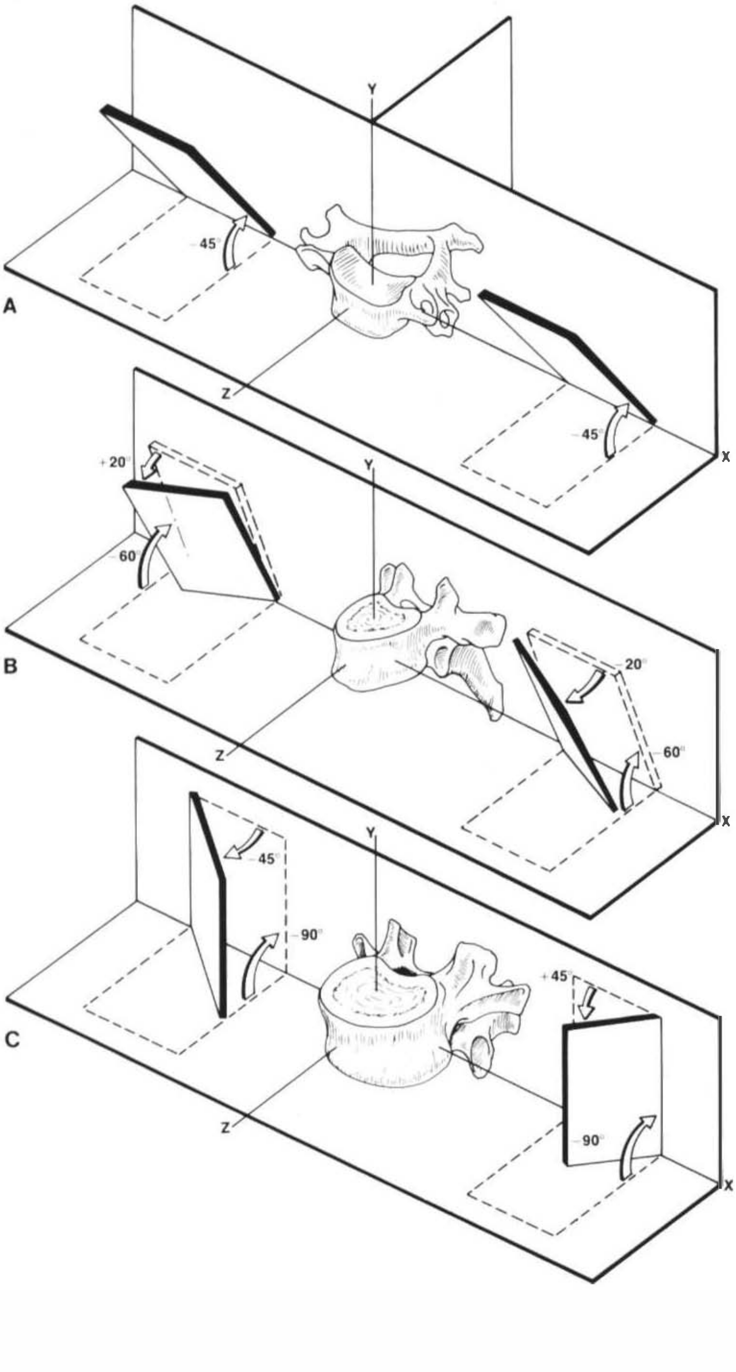


FIGURE 1-19 Orientation of the facet joints. A graphical representation of the facet joint inclinations in various regions of the spine is obtained by rotating two cards lying in the horizontal plane through two consecutive angles, x-axis rotation followed by y-axis rotation. Typical values for the two angles for the three regions of the spine are as follows. **(A)** Cervical spine: −45° followed by 0°. **(B)** Thoracic spine: −60° followed by +20° for right facet, or −20° for left facet. **(C)** Lumbar spine: −90° and −45° for right facet, or +45° for left facet. These are only rough estimates. There are variations within the regions of the spine and between different individuals. Angulations of the lumbar facet joints in the transverse plane are given in Figure 1-20.

facets of C2 to L5. Two cards are initially placed in the horizontal plane. A sequence of rotations of the cards about the various axes of the coordinate system shows the orientation of the facet joints they represent.

In the cervical spine the inclination of the facet joint plane is simulated by first placing the two cards in the horizontal plane and subsequently rotating them through an angle of −45° around the x-axis (Fig. 1-19A). In this position they represent the inclination of the right as well as the left facet joints, C2–C3 to C7–T1.

Orientation of the thoracic facet joints, T1–T2 to T11–T12, is depicted in Figure 1-19B. Again starting with the horizontal plane, a rotation of −60° about the x-axis is followed by a 20° rotation about the y-axis. The latter rotation is positive for the right facet joint and negative for the left facet joint.

The facets of the lumbar region are not plane, but have significantly curved mating surfaces; the inferior facets are convex, while the superior facets are concave. Average planes of inclination of the facet joints, T12–L1 to L5–S1, are depicted in Figure 1-19C. The horizontal cards are first given a negative rotation of about 90° around the x-axis. This is followed by a 45° rotation about the y-axis. This last rotation is positive for the left and negative for the right facet joint.

It should be emphasized that these orientations are only approximate. There is a considerable variation within specific regions of the spine, and transition from one inclination to another does not always coincide with transition from one region of the spine to another. For example, the transition vertebra between the thoracic and lumbar regions could be any vertebra from T9 through L1.

In the lumbar region, the facet joint articulations are oriented approximately perpendicular to the transverse plane. More precisely, this orientation is about 18°, such that the inferior tip of the inferior facet is directed posteriorly with respect to the frontal plane.[238] This makes CT-scan imaging an excellent tool for measuring facet orientations. Based upon such studies, angles and shapes of the lumbar facet joints have been obtained.[8,143,246,253] These are presented in Figure 1-20. The facet angle made with the sagittal plane increases from L1–L2 to L5–S1. There is significant variation in the angle at each level, as shown by the large range of values. There is also significant asymmetry, approximately 30° at L4–L5.[8]

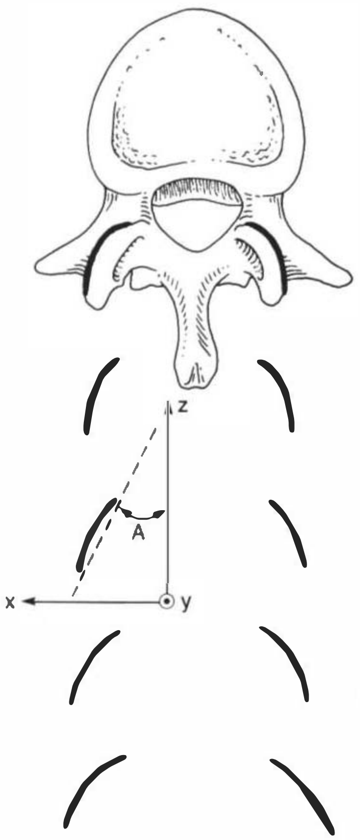
Pedicles Anatomy of the pedicles has gained significant clinical importance because of the accelerated use of pedicle fixation devices in the thoracic, lumbar, and lumbosacral regions. It is not enough to use the illustrations given in the standard anatomic text books. What is needed is a quantitative description of three-dimensional anatomy of the pedicles so that the pedicle screws may be securely and safely anchored into the vertebra. Based upon some recent studies, such data are being made available.[31,123,183,268] Four parameters seem to be necessary. They are pedicle cross-section height (PDH); pedicle cross-section width (PDW); pedicle axis inclination to the sagittal plane (PDIs); and pedicle axis inclination to the transverse plane (PDIt). The four parameters are graphically defined in the Figure 1-18, and their values from T1 to L5 are given in Table 1-6. In the table we have provided both the means and the ranges for each of the parameters. This is so because of the significant variations in these dimensions in the normal population.

Physical Properties

Vertebral Body

Determination of compression strength of the human vertebrae has been the subject of research from the early days of biomechanics. One of the driving forces behind the research has been the problem of pilot ejection. Basically, it involves ejecting the pilot from the high-speed aircraft with the help of a rocket attached to the seat. To minimize the injury to the spine at the time of ejection, it is necessary to use a safe ejection acceleration. This requires a knowledge of the strength thresholds of the vertebrae.

We do not know the design of the experimental setup or the conditions of the cadaveric material used by Messerer in 1880,[153] but his are the only data available, even to this day, that give strength values of the *cervical* vertebrae. Ruff, in his classical paper on the experiments in connection with the pilot ejection problem, reports the results obtained by Geartz.[219] More recently, Perry performed static compression tests on 40 lumbar FSUs in order to study the end-plate fractures.[203] Bell and colleagues also performed similar tests on 32 L4–L5 FSUs.[26] The results of some of these studies, in the form of strength vs. vertebral level, are summarized in Figure 1-21. The trend seems to be clear, although there is some variation between the results of different authors, probably due to differences in the experi-



LEVEL	ANGLE A (degrees)
L1–L2	25 (15–47)
L2–L3	28 (17–51)
L3–L4	37 (15–57)
L4–L5	48 (13–70)
L5–S1	53 (36–70)

FIGURE 1-20 The shape and inclination of the facets of the lumbar spine in the transverse plane (XZ plane) are shown. The facet inclination with the sagittal plane increases toward the lower levels. *(Data from Van Schaik, J. P. J., et al.,[253] Taylor, J. R., et al.,[246] and Ahmed, A. M., et al.[8]).*

TABLE 1–6 Pedicle Dimensions at Selected Thoracic and All Lumbar Levels Are Given as Mean (Range)*

	Width (mm)	Height (mm)	Angle With Sagittal Plane (degrees)	Angle With Transverse Plane (degrees)
C3	6 (4–8)	8 (6–10)	41 (20–55)	−6 (−16–4)
C5	6 (4–8)	7 (5–9)	39 (24–54)	0 (−10–10)
CT	7 (5–9)	8 (6–10)	30 (15–45)	6 (4–16)
T1	8 (5–10)	10 (7–15)	27 (16–34)	13 (4–25)
T5	5 (3–7)	12 (7–14)	9 (2–19)	15 (7–20)
T9	6 (4–9)	14 (11–16)	8 (0–11)	16 (9–14)
T12	7 (3–11)	16 (12–20)	−4 (−17–15)	12 (7–16)
L1	9 (5–13)	15 (11–21)	11 (7–15)	2 (−13–15)
L2	9 (4–13)	15 (10–18)	12 (5–18)	2 (−10–13)
L3	10 (5–16)	15 (8–18)	14 (8–24)	0 (−10–12)
L4	13 (9–17)	15 (9–19)	18 (6–28)	0 (−6–7)
L5	18 (9–29)	14 (10–19)	30 (19–44)	−2 (−8–6)

* The width and height are perpendicular to the axis of the pedicle, while the two angles are made by the pedicle axis with the respective planes.

(Data from Berry, et al.,[31] Krag, et al.,[123] Panjabi, et al.,[182, 189a] and Zindrick, et al.[268])

mental design, testing conditions, and age of the cadaveric specimens. Weaver has shown by the strength measurements of vertebral cancellous bone cubes that the material properties of L3, L4, and L5, at least of the cancellous portion, are about the same.[258] Therefore, the variation in the vertebral strength with the spinal level is most probably due to the size of the vertebrae alone.

In general, the vertebrae decrease in strength with age. Bell and colleagues have shown that there is a definite relationship between the strength (stress of failure) and relative ash content or osseous tissue of the vertebrae (Fig. 1-22).[26] The graph depicts a very important point. Bell and colleagues report that a small loss of osseous tissue produces considerable loss in the vertebral bone strength. From the graph in Figure 1-22A, we see that a 25% decrease in the osseous tissue results in a more than 50% decrease in the strength of a vertebra. This has to do with the columnlike design of the trabecular mesh that forms the central part of the vertebra. An analysis of this

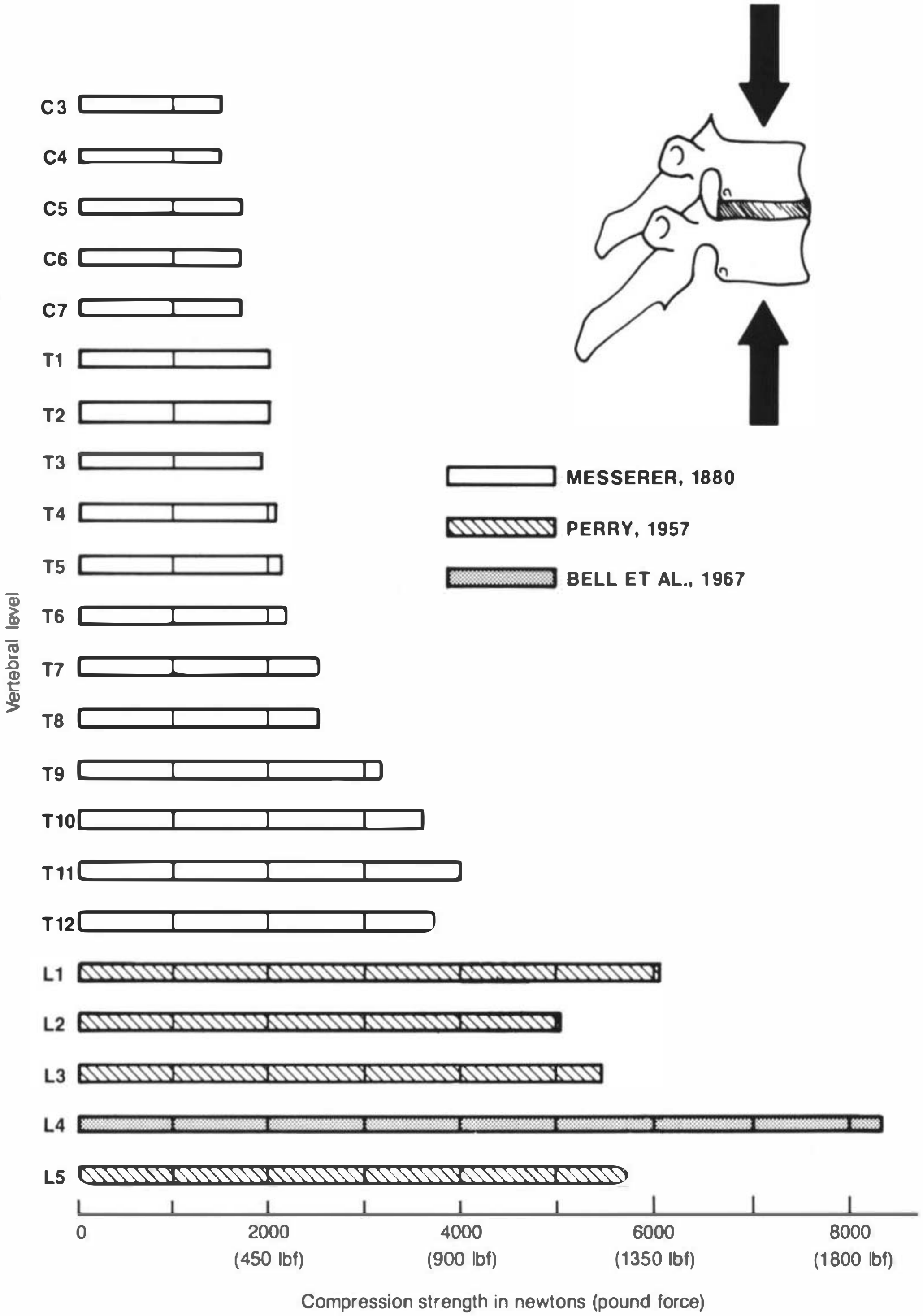


FIGURE 1-21 Vertebral compression strength at slow loading rate. A variation of strength from C3 to L5 is shown. (*Data from Bell, et al.,*[26] Perry,[204] *and Messerer.*[153])

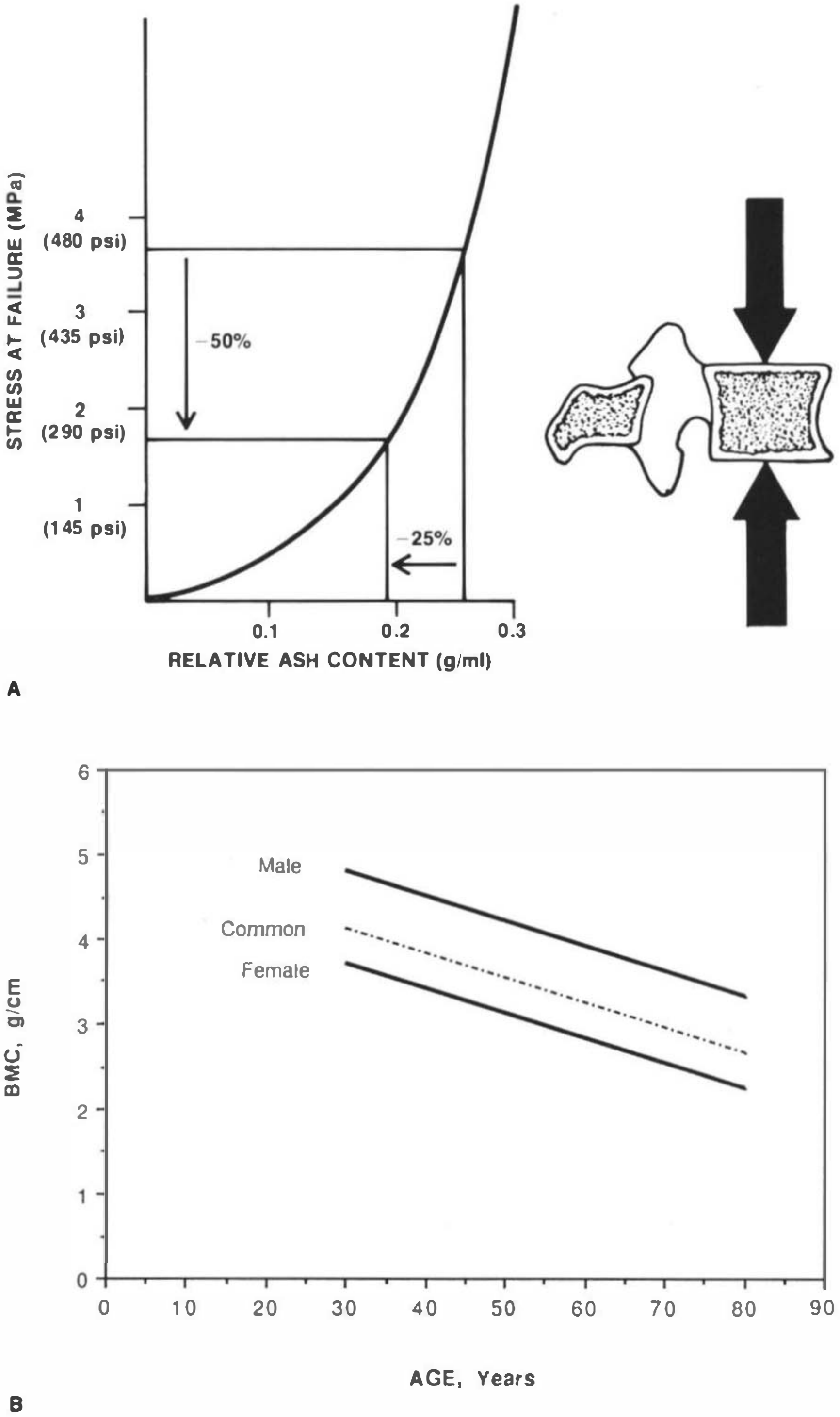


FIGURE 1-22 **(A)** Relationship between osseous tissue and vertebral strength. A 25% decrease in the osseous tissue of the vertebra causes a much larger (50%) decrease in the vertebral strength. This has to do with the load-carrying capacity of vertical and horizontal trabeculae. See Fig. 1-26. *(Data from Bell, G. H., et al.: Variation in strength of vertebrae with age and their relation to osteoporosis. Calcif. Tissue Res., 1:75, 1976.)* **(B)** Bone mineral vs. age. There is a linear decrease in the bone mineral content of lumbar vertebral trabeculae. Although the bone content is less in the female at nearly all ages, the rate of decrease with age is about the same for both sexes. *(Data from Hansson, T., and Roos, B.: The effects of age, height, and weight on the bone mineral content of lumbar vertebrae. Spine, 5:545, 1980.)*

phenomenon is depicted in Figure 1-26 and given in the Notes.[H]

The decrease in the bone mineral content with age has been documented by using a noninvasive technique[91] and is shown in Figure 1-22B. There seems to be a linear relationship between the increase in age and the decrease in bone mineral content. Surprisingly, the rate of decrease is not different between the males and females. However, the bone mineral content, at any age, is less in the female vertebrae. In fact, it is 12% less at, say, 50 years of age. Hansson and co-workers have shown that there is a high correlation between the bone mineral content, as measured by photon absorptiometry, and mechanical strength.[92] It is hoped that such a method may be used clinically to assess the vertebral strength *in vivo*.

Cortical Shell

Although the facets carry some compressive loads, it is the vertebral body that carries the major share in most physiologic situations. This load is transmitted from the superior end-plate of a vertebra to the inferior end-plate by way of two paths, the cortical shell and the cancellous core.

What is the relative share of the load carried by the two paths? The literature concerning this is conflicting. One study concluded that the load-supporting part of a vertebral body is the compact rather than the spongy bone.[65] According to others, the outer wall of a vertebra, unlike that of a long bone, is very thin and can make only a small contribution to its strength.[22, 26] To resolve this conflict, a study was conducted by Rockoff and colleagues.[216] To appreciate their findings, a short description of the experimental procedure is in order.

Vertebrae without posterior elements were obtained from the lumbar spine of cadavers. Nondestructive tests (*i.e.*, applying small loads that may not cause microfractures) for compression strength were carefully performed on each vertebra, and the specimens were then divided into two groups. In one group, the vertebral bodies were hollowed out with the help of a rotating burr introduced into the bodies by way of the basivertebral vein canal. This produced a vertebra with only the cortical shell left intact. In the second group, the outer shell was carefully ground away, leaving only the cancellous core. The vertebrae of the two groups were again subjected to the same nondestructive test. The resulting loss in strength of a specimen, as compared with its intact strength, represented the contribution of the trabecular bone in the first group and the cortical bone in the second group. Other complementary tests, such as bone density, bone volume, and ash content, were also performed. The effects of age were included.

In general, there was a decrease in strength of the intact vertebrae with age.[E] A rapid rate of decrease was observed from 20 to 40 years, while the strength remained more or less constant after age 40. This finding is supported by the bone strength measurements of Bartley and colleagues[22] and Weaver,[258] histologic findings of bone quantity by Bromley and colleagues,[40] and bone surface area measurements by Dunnhill and colleagues.[59]

In addition, Rockoff and colleagues found that under compressive load the trabecular bone contributes 25–55% of the strength of a lumbar vertebral body, depending upon the ash content of the bone.[216] Regarding variation with age, they found that under 40 years of age 55% of load is carried by the trabecular core, while after 40 years this share decreases to about 35%.

This important question of load sharing between the cortical shell and cancellous core has been reexamined.[148] Using 20 fresh cadaveric lumbar spine specimens with and without the cortical shell, the researchers found the cortical shell to provide only a small portion (average 10%) of the total compressive failure load. This is especially surprising since their specimens came from an older population (63–99 years) in which, according to the Rockoff study,[216] the share of the cortical shell is about 65%.

Cancellous Core

Spongy bone of vertebrae has other interesting aspects. In a recent paper by Lindahl, the mechanical properties of this part of the vertebra were studied in detail.[131] He subjected small cubic blocks of the trabecular bone from L2 to L4 to compressive loads until failure occurred, while recording the load–deformation curves. The shape of these curves indicated some remarkable characteristics of the spongy bone. Three types of curves, distinguished by the latter portions, were identified (Fig. 1-23A). Type I shows *decreasing* strength after the maximum load is reached. Type II *maintains* its strength, and Type III shows *increasing* strength after the failure point. Decreasing strength (Type I) was exhibited by only 13% of the specimens tested. About half of the specimens showed constant strength after failure (Type II), and in 38% of the cases the strength kept increas-

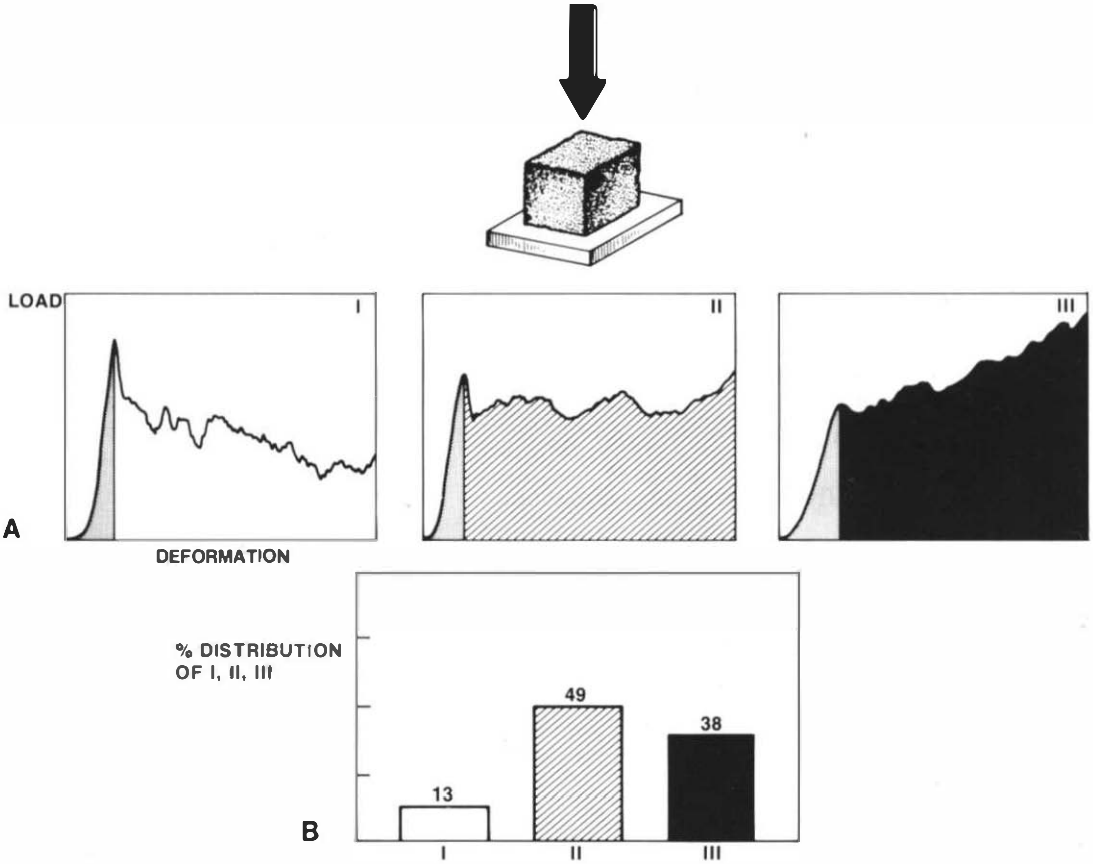


FIGURE 1-23 Cancellous bone failure patterns. **(A)** Cancellous bone samples of vertebrae when subjected to compression fail in three different ways, as shown by the load–deformation curves, Types I, II, and III. It is the latter part of the curve, after the first peak is reached, that differentiates the three types. **(B)** The majority of the curves were found to be of Type II, followed by Types III and I. *(Data from Lindahl, O.: Mechanical properties of dried defatted spongy bone. Acta Orthop. Scand., 47:11, 1976.)*

ing after the "failure" (Type III). These results are depicted graphically in Figure 1-23B. Lindahl further reported that vertebrae with a Type III curve, which is biomechanically the superior of the three,[F] were found most frequently in males under 40 years of age and least frequently in women over 40. These findings may have some relation to the probability of progressive collapse following vertebral compression injuries.

Although there was much variation in the latter part of the curves of different specimens, the mechanical properties represented by the early part of the load–deformation curve were quite consistent. These properties have been quantified and are presented in Table 1-7. Note that the cancellous bone of a vertebra undergoes large compressive deformation, up to 9.5%, before it fails. The corresponding deformation for the cortical bone is less than 2%. Therefore, in vertical compressive loading, injury pain is more likely to be the result of cortical plate fractures than of microfractures in cancellous bone. Obviously, if the magnitude of the force is large enough, then both types of fractures are likely to occur.

The study by Lindahl was done on trabecular samples from which the bone marrow had been removed. In a recent study by Hayes and Carter, they have proved that the shock-absorbing mechanism of trabecular bone is enhanced by the presence of bone marrow, especially in highly dynamic situations such as traumatic injury.[96] Cylindrical specimens of subchondral trabecular bone of bovine femurs were

TABLE 1–7 Compressive Strength Properties of Cancellous Bone of Vertebrae

Physical Property	Magnitude
Proportional-limit stress*	1.37–4.0 MPa
Compression at proportional limit	6.0–6.7%
Modulus of elasticity	22.8–55.6 MPa
Failure stress	1.55–4.6 MPa
Compression at failure	7.4–9.5%

(Data from Lindahl, O: Mechanical properties of dried defatted spongy bone. Acta Orthop. Scand., 47:11, 1976, and Hansson, T. H., Keller, T. S., and Panjabi, M. M.: A study of the compressive properties of lumbar vertebral trabeculae: effects of tissue characteristics. Spine, 12:56, 1987.)

* A point on the load–deformation curve beyond which the elastic portion of the curve is no longer linear.

loaded in a specially designed test fixture with the specimen so confined that the fluid within the specimen could not leak out during the compression loading. They recorded the load–deformation curves at slow as well as very fast rates of loading. The curves were found to be mostly Types II and III, as reported by Lindahl.[131] Type III curves were more often associated with the samples that had high apparent density. Hayes and Carter further reported that the presence of bone marrow significantly increased the compressive strength as well as the energy absorption capacity of the trabecular bone samples. This effect was more significant at higher rates of loading. The suggested mechanism of energy absorption by the cancellous bone was the collapse of an increasing number of intertrabecular spaces as the load was increased. This further constrained the movement of the bone marrow, providing a hydraulic cushion. Therefore, the function of the cancellous core seems to be not only to share the load with the cortical shell, but, at least at high rates of loading, to act as the main resistor of the dynamic peak loads. This is important to keep in mind in the analysis and understanding of vertebral trauma.

The cancellous bone lies at the center of the vertebral strength. Its function can be analyzed by detailed biomechanical[148] and morphological studies.[63,238] These are described in a later part of this section under Functional Biomechanics. Cancellous bone within the vertebra is not uniform as seen biomechanically. In a study that aimed to document this distribution and to possibly relate it to the variation within the adjacent discs, the vertebral body was divided into approximately 1-cm cubes and tested in compression.[113] The deformation of trabeculae under compressive load and recovery after the load is removed have been studied using scanning electron microscopy.[64]

End-Plate

Although failure of the end-plates under compressive loading has been observed by many research workers, it was Perry in 1957 who conducted exhaustive experiments to obtain the basic understanding of the end-plate failure mechanism.[203] His spine specimens were mostly from the lumbar region, with a few lower thoracic specimens also included. The age of the subjects varies from under 40 to over 60 years, covering a wide range of disc degeneration.

The experimental procedure involved an application of increasing compressive force to the intact FSU specimens. The deformation produced and the force applied were continuously recorded. The end point was reached when the load suddenly decreased, indicating failure of the specimen. In these static tests, one-third of the specimens had end-plate fractures with herniation of the nucleus pulposus into the vertebral spongiosa. This fracture pattern was present more often in the younger age group and in the upper lumbar vertebrae. There were no disc herniations. Generally, the strength of the FSU was greater in the lower region than in the upper region of the lumbar spine. However, there was much greater variation as a result of age. Below 40 years, the FSU could bear about 8000 N (1800 lbf) of compressive load. Between 40 and 60 years, the strength decreased to about 55% of this value, and above 60 years it decreased to 45%.

Perry also performed high-speed dynamic tests on 76 specimens. Again, there were no disc herniations. The failures of the specimens were due to either failure of the end-plate or the compression of the vertebra, depending upon the intensity of the load. In these dynamic tests lasting 0.006 seconds, the loads applied were much higher, up to 13500 N (3030 lbf).

Basically, there were three failure patterns of the end-plate observed by Perry: central, peripheral, and one involving the entire end-plate. The central fractures were more often present in the specimens with nondegenerated discs. The opposite was true for the peripheral fractures. The fractures encompassing the whole of the end-plate were the result of higher loads. We can look at these failure mechanisms in some detail.

Figure 1-24A shows an FSU with a nondegene-

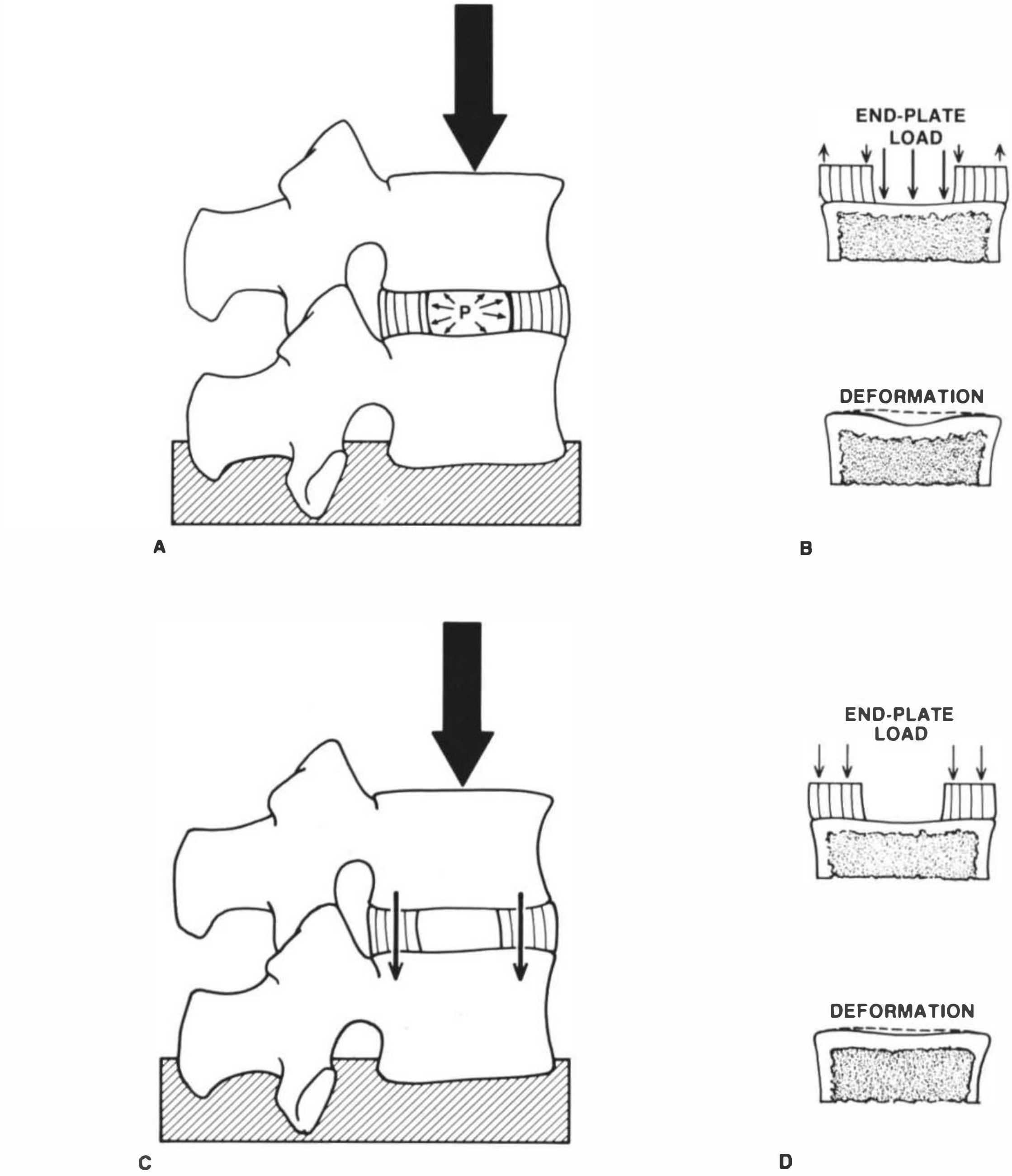


FIGURE 1-24 End-plate failure mechanism. **(A, B)** Compression of a nondegenerated disc produces pressure within the nucleus, which results in compression load at the middle of the end-plate and some tension on its periphery. Loading of the end-plate in this manner produces deflection of the end-plate, so that high bending stresses occur in the center. The latter may cause central fractures of the end-plate and Schmorl's nodes. **(C, D)** In a degenerated disc the compressive load is mostly transferred from one end-plate to the other by way of the annulus. The nucleus may not carry any significant loads. The end-plate is loaded more at its periphery. The stresses are more evenly distributed within the entire end-plate. The failure is by fracture of the vertebral body.

rated jellylike nucleus. When the compression load is applied there is a build-up of pressure within the nucleus. This produces tension in the outer fibers of the annulus and a central compressive loading on the end-plate (Fig. 1-24B). Also shown is the deflection curve of the end-plate. As the stresses are in direct proportion to the bending moment, the fracture will most probably start in the center of the end-plate, where the bending moment is maximum under these circumstances.

A completely different situation arises when the nucleus is not jellylike and is not able to build up any significant fluid pressure. This is depicted in Figure 1-24C. The compressive load is mostly transferred directly from one vertebrae to the other by way of the annulus. The annulus is mostly under compression, and so is the periphery of the end-plate, with much less deflection at its center (Fig. 1-24D). The failure of the vertebra is due to the fracture of the periphery of the end-plate.

Rolander and Blair[218] did experiments similar to those done by Perry, but in addition they measured the deflection of the center of the end-plate.[G] Results showed that the end-plates did indeed buckle away from the disc, increasing the disc height at the center by about 0.6 mm (0.024 in) just prior to the failure. In all of the specimens tested there were fractures of one or both end-plates. None of the failures was due to compression of the vertebral body.

Facet Joints

Facet joints are clinically important for at least two reasons. First, facets have been found to be a direct source of pain. Second, it has been observed clinically that facets are important stabilizing structures, and their surgical excision, unilaterally or bilaterally, often leads to spinal instability.[1,97] The clinically important mechanical roles of the facets have been explored by several biomechanical studies.

Nachemson, using his needle pressure transducer, measured the nucleus pressures (and therefore the disc loads) of an intact FSU and of an FSU in which the posterior elements had been removed. He concluded that the facets carry about 18% of the total compressive load borne by a lumbar spine segment.[161]

Dynamic studies of whole cadavers by King and colleagues have shown that the mechanism of load sharing between the facets and the disc is rather complex. Using cadavers fitted with a special load-measuring device in place of a disc, the facet and disc loads could be separated. The cadavers were subjected to caudocephalad accelerations of varying degrees. After extensive measurements, they concluded that, depending upon the spine posture, the share of the load carried by the facets could be anywhere from 33% to zero. In certain spinal postures, the facets were unloaded and the capsular ligaments were put under tension.[117]

In a comparative study of the various components of an FSU with respect to their contribution toward the torsional strength, Farfan found that the disc and the longitudinal ligaments shared equally with the two facets including the capsular ligaments, about 45% each. The remaining 10% of the torsional strength was contributed by the interspinous ligaments.[67]

White and Hirsch studied the role played by the facets and the posterior ligamentous complex in restricting the physiologic motions of the spine.[262] Using thoracic FSUs, they measured the various ranges of motion of intact FSUs and of FSUs with posterior elements removed. In the upper thoracic region there was a 50–80% increase (greatest during flexion/extension and least during lateral bending). In the lower thoracic FSUs the increase was only 15% during flexion/extension and lateral bending and 40% in axial rotation.

In a study of the flexion/extension stability of the cervical spine *in vitro*, the various components of the FSU were transected in two different sequences under simulated flexion and extension.[192,263] The transection sequences were either anterior to posterior or posterior to anterior. Under the application of a flexion-producing load equivalent to one-third body weight, the transection of the disc and the longitudinal ligaments produced a 33% increase in the horizontal translation as compared with the intact FSU. When the facets were transected next, the corresponding increase was 140%. Thus, the facets provide significant stability to the spine in flexion, especially when the disc is already ruptured.

The importance of facet orientation for the pathology of the intervertebral disc has been well documented by Farfan and Sullivan.[70] Using radiographic measurements, they studied 45 patients admitted for low back pain with sciatica who were treated conservatively. In addition, there were 52 patients who were ultimately treated surgically. From radiographic measurements and operating room findings of these patients, they established a highly significant correlation between the asymmetry of the facet joints and the level of disc pathology and between the side of the more oblique facet orientation and the side of sciatica. However, this idea of tropism has not been supported by recent biomechanical studies.[8,58a]

Neural Arch

There are two types of biomechanical studies that have dealt with the components of the neural arch: experimental and mathematical models. In the experiments, the methods of loading have varied greatly among the authors.[128,217,259] This is a reflection of our imprecise knowledge at the present time concerning the loads applied to the neural arch *in vivo*. The methods of loading and the average failure loads are depicted in Figure 1-25.

Most failures occurred through the pedicles. In the experiments by Lamy and colleagues, about one-third of the failures were through the pars interarticularis.[128] This number increased when the tests were conducted at higher rates of loading. The strength was found to be the same for both male and female subjects as well as for those with either normal or degenerated discs. However, it decreased with age.

Functional Biomechanics

Cancellous Bone

In osteoporosis there is reduction in osseous tissue (ash content).[21,25] There is also a decrease in the vertebral strength with age as observed by Perry[203] and Bell and colleagues.[26] Proportionally, there is much greater decrease in the strength as compared with the loss in the osseous tissue (Fig. 1-22).[26] Using a model described subsequently, one may explain these changes in the mechanical strength on the basis of engineering principles.

The cancellous part of the vertebral body (Fig. 1-26A) may be thought of as an engineering structure composed of vertical trabeculae (vertical columns) joining the two end-plates and horizontal trabeculae (horizontal ties) supporting the vertebral trabeculae from the sides (Fig. 1-26B).[78] This pattern has been observed by Casuccio,[45] Atkinson,[19] and Amstutz and Sissons.[11]

According to a well-established engineering principle, Euler's theory of buckling, the compressive strength of a column is in direct proportion to the square of the area of its cross-section and in inverse proportion to the square of its unsupported length.[11]

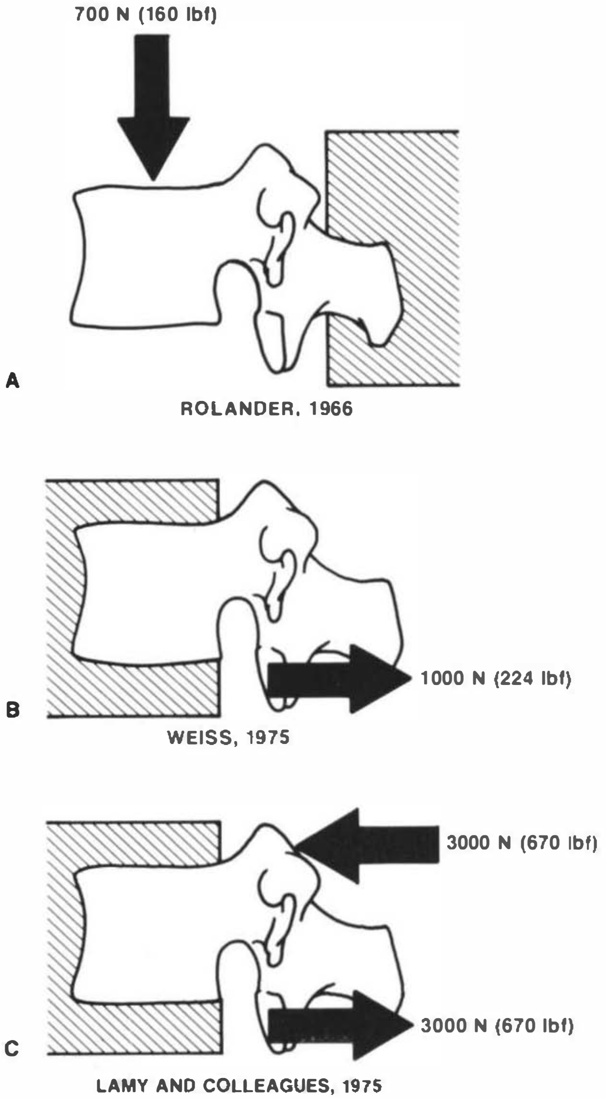


FIGURE 1-25 Failure loads of the neural arch. **(A, B, C)** The variation in the failure loads is representative of the differing methods of load application in the three experiments.

In osteoporosis, the decrease in the amount of osseous tissue of a vertebra may result in a decrease in the cross-sectional area of the vertical trabeculae (columns) and/or a breakdown of some horizontal trabeculae (ties). Thus, a 50% decrease in the mass results in one-quarter of the original strength (Fig. 1-26C). On the other hand, breakdown of horizontal ties effectively increases the unsupported length. If 50% of the horizontal ties (i.e., every alternate tie) were removed, the strength of the structure would be reduced to one-quarter of its original value (Fig. 1-26D). This reduction in strength would be even

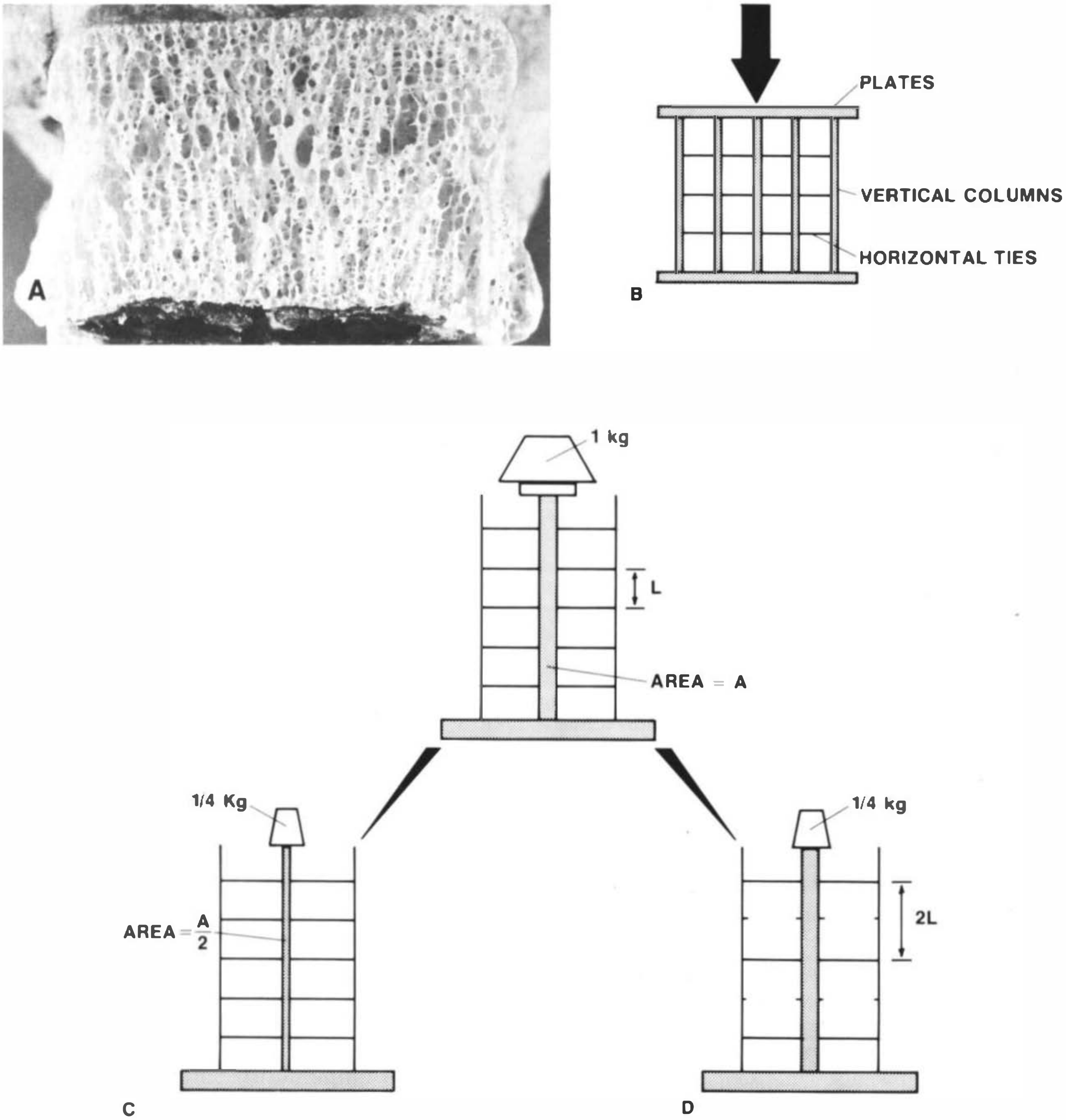


FIGURE 1-26 Biomechanical analysis of vertebral failure in osteoporosis. **(A)** A photograph of a longitudinal section through a vertebral body shows the dominant vertical and horizontal arrangement of the trabeculae of the cancellous bone. **(B)** A model representing the trabecular pattern of the vertebra. Horizontal ties effectively reduce the free length of the vertical columns, thus providing support. **(C, D)** Based on Euler's formula, the model predicts that the compression strength of the cancellous part of the vertebra will decrease to 25% of its original value when there is either a 50% decrease in the cross-sectional area of the vertical trabeculae or loss of horizontal trabeculae, so that the free length of the vertical trabeculae is effectively increased by 100%. *(Photograph courtesy of Yale University, Biomechanics Laboratory. Data from Bell, et al.: Variation in strength of vertebrae with age and their relation to osteoporosis. Calcif. Tissue. Res., 1:75, 1967.)*

greater if adjacent ties instead of the alternate ties were removed.

Atkinson did a histologic study of the vertebral trabeculae and their changing patterns with age. The earliest change seen was the loss of the horizontal trabeculae. However, this was accompanied by simultaneous thickening of some of the vertical trabeculae. Although there was no appreciable loss of osseous tissue on the whole until the age of 50, there was nonetheless a substantial decrease in the mechanical strength. The biomechanical analysis nicely shows the high sensitivity of the strength to the loss of the horizontal trabeculae.[19]

Another observation by Atkinson was that with age there was a loss of horizontal trabeculae in the central region of the vertebral body, while those in the peripheral regions remained unaltered. This implies that the loss of strength with age occurs preferentially in the middle of the vertebrae. This seems to fit nicely with the clinical observations of central collapse of the vertebral body in patients with osteoporosis.

The important question of changes in the internal architecture of the horizontal and vertebral trabeculae as we age was investigated using modern techniques of stereology.[239] Snyder and colleagues quantified the dimensions and spacing of horizontal and vertical trabeculae as well as the relative overall density of midsagittal plane sections of L1 vertebral specimens. Contrary to the expected greater decrease of horizontal trabeculae with decreasing density, they found that both trabeculae get thinner and decrease at the same rate, but the horizontal trabeculae are lesser in number than the vertical trabeculae at all density levels. Thus, the spacing between horizontal trabeculae increases more rapidly than the spacing between vertical trabeculae. It has been shown in a separate study that the vertebral strength was related to density raised to the power of 2.26.[148] This represents triple jeopardy as to vertebral strength. Not only are there fewer trabeculae, but the remaining trabeculae are becoming both thinner and longer. To quantify the trabecular bone morphological changes, quantified computed tomography (QCT) has been found to be a good indicator of bone morphology[63] and vertebral strength.[148]

Biomechanical Adaptation Degeneration of the spine manifests itself in many different ways. Osteophytes, facet degeneration, and disc degeneration are some examples. The degeneration of the disc is well recognized (macroscopically, radiographically, and biomechanically), but its effect on the vertebra itself has not been previously explored. Knowing that Wolff's law (simply stated: form follows function) is valid in most structures, one may expect changes in the trabecular bone as a reflection of the changes in the underlying disc.[264] This hypothesis has been tested and proved correct in recent studies.[89, 113]

Using fresh lumbar spines, the investigators carefully cut each vertebral body into 12 cranial and 12 caudal 1-cm cubes. The cuts were made in such a way that the cubes consisted only of the trabecular bone, no end-plate or cortical shell. The cubes were properly identified by their location, and disc degeneration of the adjacent disc was also recorded. The cubes were compressed in the cranial-caudal direction, and the load–deformation curves were recorded. Parameters of bone stiffness, failure deformation, failure load, and so forth, were determined.

Distribution of compression strength (failure load divided by cross-sectional area of the cube) in the transverse plane is shown in Figure 1-27. The bars represent the strength in megapascals. We find that there is a tendency for the trabecular bone to be strongest in the center, just above the disc nucleus, and weakest in the peripheral region, especially in the posterolateral region. This strength variation was especially noted among the specimens with less degenerated discs. We know from mathematical modeling studies of the spine that it is the central region of the disc that carries the highest compressive stresses. In accordance with Wolff's law, therefore, stronger trabecular bone is needed in the central vertebral region to support this load, as compared with the peripheral region. In the case of the degenerated discs, the compressive stresses are more uniformly distributed over the entire disc plane. The trabecular bone strength was found to have similar uniform distribution.

The End-Plate

The major physiologic load on the spine is axial, which produces compression of the functional spinal units. For example, it has been observed that astronauts (and cosmonauts), on their return to Earth, are several centimeters taller than when they left for the space journey. The increased height was due to loss of the weight of the head and trunk during the weightlessness in space. Conversely, here on Earth, the physiologic compression load on the

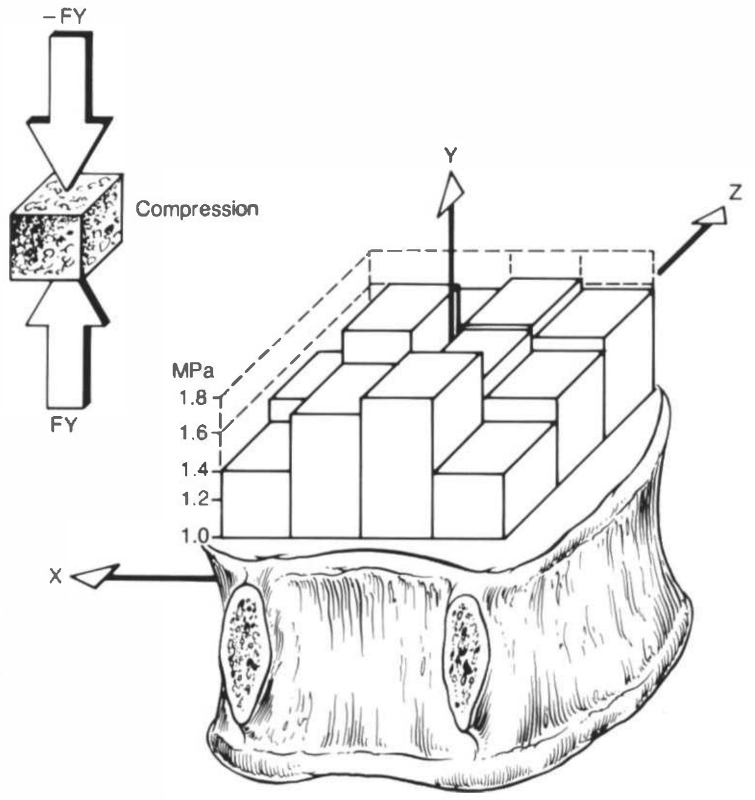


FIGURE 1-27 Trabecular compressive strength distribution in the transverse plane. The vertebra exhibits greatest strength in the center, followed by middle of the posterior region. It is weakest on the outside of the posterior region. *(Data from Keller et al: Regional variations in the compressive properties of lumbar vertebral trabeculae: effects of disc degeneration. Spine, 14:1012, 1989.)*

spine decreases its natural length. This is for two reasons. First, the lordosis of the cervical and lumbar regions and the kyphosis of the thoracic region increase with compression load. Second, all the functional spinal units between the head and pelvis (25 in number) are compressed. It is this latter mechanism that we wish to look at a little more closely.

The decreased height of a functional spinal unit under compression is due to both the outward radial bulge of the annulus and the outward axial bulge of the end-plates. A simple "thought experiment" will help provide an understanding of these phenomena.

Let us take an empty can of relatively large diameter and small height. Fill it with water under controlled pressure, using a syringe. The pressure acts equally on the insides of the cylindrical wall and on the lids. Engineering analysis shows that the stresses generated in the walls are lower than those in the lids. But if the walls were made of a weaker material (e.g., rubber membranes), the walls would bulge radially outward and the lids would bulge axially outward. Depending upon the materials we choose for the walls and the lids, we can make the two bulges different or equal in size.

Let us continue our thought experiment. Attach a pressure gauge, and monitor the internal fluid pressure. Now apply an external axial compression. We will find that the inside pressure increases, and consequently the cylindrical wall bulges further outward radially and the lids bulge further outward axially. Thus, the external compression load has the same effect on our can of water as does the increase in internal pressure.

Experiments, using fresh cadaveric specimens,[15] and *in vivo* measurements[161] have confirmed the above findings of increased internal pressure with the application of external compression. Findings of outward bulging of both the disc annulus and the end-plates have also been observed by direct measurements[37,214,218] and by mathematical models.[37] It has been estimated that in a real spine the end-plate bulge is about half the size of the disc bulge.[38]

The Facet Joints

Clinical observations in the lumbar spine indicating that there is a greater risk for disc herniation in association with a more oblique facet,[70] seen from the biomechanical viewpoint, imply that a more oblique facet allows greater axial rotation and, therefore, increased stresses in the disc annulus, leading to disc injury. In a recent *in vitro* study using fresh cadaveric lumbar spine specimens, Ahmed and coworkers found no correlation between the facet asymmetry and increased axial rotational range of motion.[8] The study supported the view expressed by Adams and Hutton that the facets act as positive stops to axial rotation and that certain tropism does not affect this facet function.[2,52] It may be further argued that the facets protect the disc from any damage that may otherwise occur as a result of the rotation.[4] The disc alone is capable of being torqued up to 22° without failure.[69] The maximum rotation allowed, when the facets are present, is less than 5° measured *in vitro*[217] and *in vivo*.[86] This is clear evidence of the protective role of the facets against torsional damage to the disc.

Another important role of the facets is that of resisting anterior shear loads. It has been estimated that the facet joints carry one-third of the anterior shear while the disc carries two-thirds.[52] This is when the load is suddenly applied. However, because of the viscoelastic properties of the disc, if the load is applied slowly, then the above vertebra creeps forward, further loading the facets. Thus, after sufficient time, most of the shear load may be carried by the facets alone.

We know from indirect measurements that the facets share the spinal load with the disc and that this share varies depending upon the spine posture. Lorentz and co-workers directly measured the facet loads, and the distribution of the load over the facet surface in the form of peak pressure has been quantified.[136] They found, using fresh cadaveric spinal units, that the load carried by the facets varied from about 9% in neutral to about 15% in extension. This corresponded to peak pressures of 1.3 to 2.9 MPa. In comparison between the L2–L3 and L4–L5 levels, the facet loads were found to be higher at the upper level.

The above findings are at some variance with the findings of a recent study by Yang and King.[266] They found the facet loads to be 3–25% of total load, and this could increase up to 47% if the facet joint is arthritic. The authors also proposed, based upon their experimental findings, a new mode of load transmission from one vertebra to the next. As the spine is fully extended, the tip of the inferior facet contacts the pars interarticularis of the vertebra below. Further extension load pivots the vertebra about the facet tip and stretches the capsular ligaments.

Recently a mathematical model of a L2-L3 FSU, including realistic facet articulations, has been constructed. Shirazi-Adl and Drouin found that the compressive force is mostly borne by the disc and that facets carry only about 8%. This load share may increase to 30% if, in addition, the spine is in extension. The opposite, up to 7° of flexion happens in flexion posture. Beyond this limit, the facet load again increases. Clinically, partial medial facetectomies, uni- or bilateral, are often used to find a compromise between the need to decompress and the need to maintain stability. In a recent biomechanical study, this problem was addressed to find the answer to the question, How much spine may be decompressed before there is danger of instability? Using fresh human cadaveric lumbar spine specimens, investigators conducted three-dimensional instability tests on intact and injured specimens. The injuries varied from inter- and supraspinous ligament transections to bilateral facetectomy. The spine became unstable, especially in the axial rotation test, when one facet was completely removed (Fig. 1-28). In conclusion, semi- or bilateral partial medial facetectomies do not make the spine acutely unstable, while a total facetectomy, uni- or bilateral, destabilizes the spine and is not to be recommended, based upon the *in vitro* study results.

The Neural Arch

Mathematical models, using modern techniques of finite element methods (FEM), are useful in simulating vertebral geometry, physical characteristics, and distribution of the cancellous and cortical bones via computer "experiments." With simulated *in vivo* loadings, the distributions of stresses in the posterior elements (and the vertebral body) may be explored. Hakim and King, and later on Balasubramanian and co-workers, developed such models and studied the effects of laminectomy.[20,88] They found this surgical procedure to result in high tensile stresses in the pedicle and pars interarticularis. Loads in the posterior elements have also been studied using mathematical models.[154]

The Intervertebral Foramen

The spinal nerve root, in comparison with a peripheral nerve, has less abundant epineurium, no branching fasciculi, and poor lymphatic drainage.[244] Therefore, the nerve root is more susceptible to injury by mechanical forces (e.g., compression during

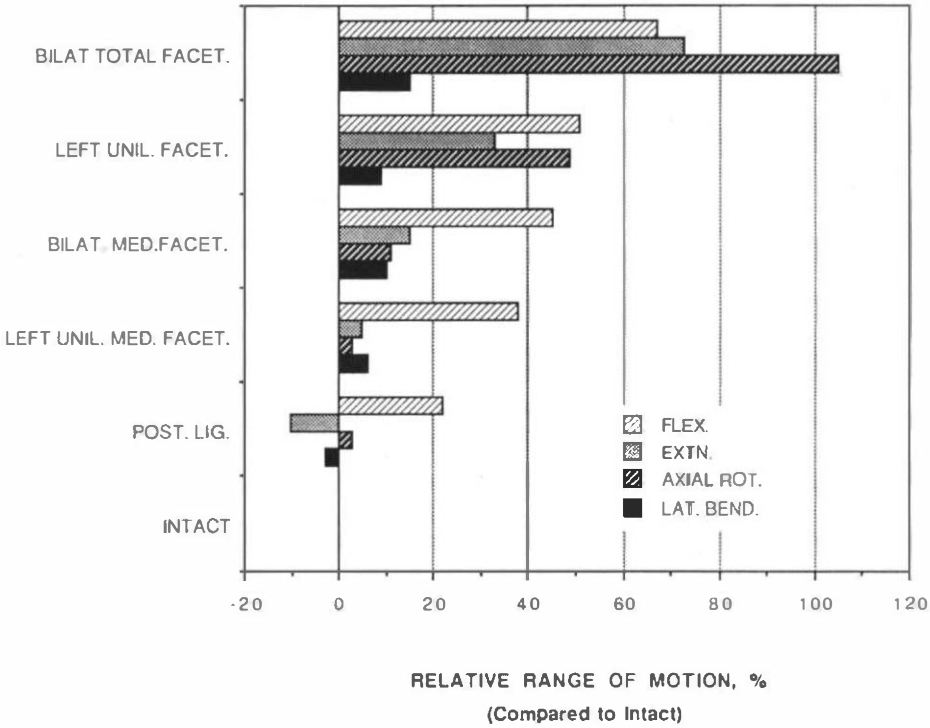


FIGURE 1-28 Results of partial facetectomies, in the form of changes in the ranges of motion as compared to the intact behavior, are shown. The changes are shown as percentages of the intact behavior under physiological loads of flexion, extension, axial rotation, and lateral bending. The flexion motion increased with increasing injury. There were smaller but similar increases in extension. The axial rotation showed significant increase with unilateral and bilateral facetectomies. Lateral bending was affected least. *(Data from Abumi, K., Panjabi, M. M., et al.: Instabilities due to partial and total facetectomies of the lumbar spine. 34th annual meeting of the Orthopaedic Research Society, Atlanta, Georgia, 1988.)*

its passage through the intervertebral foramen). Is it possible to compress a nerve root during normal physiologic movements of the spine, and is this effect aggravated by disc degeneration? Which type of motion produces the most severe compression?

Some answers to these types of questions have been obtained in an *in vitro* study using fresh cadaveric lumbar spine specimens.[190] First, physiologic movements of the moving vertebra with respect to the stationary vertebra were registered three-dimensionally. Next, the shape of the intervertebral foramen was carefully recorded. The foramen shape consisted of two parts: stationary and moving, belonging respectively to the stationary and moving vertebra of the spine specimen. Finally, a mathematical model was used to combine the spinal movements data with the intervertebral foramen shape to compute the changes in the shape and size of the intervertebral foramen due to the motions of flexion, extension, lateral bending, and axial rotation. The results were obtained separately from the nondegenerated and degenerated spine specimens, making it possible to study the effect of disc degeneration.

In flexion, the main finding was that the size of the foramen increased, while the size decreased in extension. The nondegenerated specimen had a relatively large intervertebral foramen, 185 mm^2 versus 108 mm^2 with the spine in neutral position. The foramen opened during flexion by 24% and closed during extension by 20%. Changes due to lateral bending and axial rotation were less significant.

FUNCTIONAL AND MULTISEGMENTAL SPINAL UNIT

The functional spinal unit, or the motion segment, is the smallest segment of the spine that exhibits biomechanical characteristics similar to those of the entire spine. It consists of two adjacent vertebrae and the connecting ligamentous tissues. In the thoracic region, costovertebral articulations are also included. For its biomechanical characterization, the lower vertebra is fixed while the loads are applied to the upper vertebra, and its displacements are measured. The behavior of an FSU is dependent upon, among other things, the physical properties of its components, such as the intervertebral disc, ligaments, and articulating surfaces. Because the spine may be considered as a structure composed of multiple FSUs connected in series, its total behavior may be approximated as a composite of the behaviors of the individual FSUs constituting the spine.

Physical Properties

We have combined the presentation of FSUs and multisegmental spinal units into one section. Even though most of the experimental data available were obtained from the FSU experiments, there is a trend to use multisegmental spine specimens. There are several reasons for this. The multisegmental spinal unit is closer to the *in vivo* situation: the ligaments (e.g., longitudinal) are continuous structures. Certain experiments necessitate multisegmental spinal units (e.g., stability evaluation studies of spinal instrumentations). Finally, the technical difficulties in the testing of long, and, therefore, very flexible, spinal specimens are being successfully tackled. Although the experiments are conducted using multisegmental spinal units, the results are still presented for each individual FSU constituting the multisegmental specimen.

Flexibility and Neutral Zone

Flexibility is the ability of the structure to deform under the application of a load. *Stiffness* is just the opposite. It is that property of a structure by which resistance is offered to an imposed displacement. To quantitate these structural qualities, the concepts of coefficient of flexibility and coefficient of stiffness have been evolved. The *coefficient of flexibility* is defined as the ratio of the displacement produced to the load applied. The *coefficient of stiffness* is defined as the ratio of the resistance offered to the displacement imposed. The flexibility coefficient is approximately equal to the inverse of the stiffness coefficient, and vice versa. (The nature of the approximation has been discussed elsewhere.[179]) We prefer to use the flexibility coefficient because it is closer to the clinical concepts of the range of motion and spinal instability. Also, most experiments are flexibility experiments (i.e., a load is applied and motions are measured)[178]; therefore, it is natural to present the results as flexibility coefficients.

Although it is quite convenient to represent the mechanical behavior of the spine in a certain direction (e.g., flexion) by a single number, the flexibility coefficient, it does not do justice to the complex behavior a real spine exhibits. A single number implies that the intervertebral motion, in response to the applied load, is linear. In other words, if half or double the load were applied, it would produce, respectively, half or double the amount of motion. This is far from the truth. The spine has a nonlinear, elastic behavior. (It also exhibits viscoelastic behavior, as discussed later.) The flexibility coefficient of the spine is different at different load magnitudes. In fact, the behavior is biphasic, more like that of a spinal ligament (Fig. 1-15A). At smaller loads, the spine easily deforms, providing little resistance. As the load increases, so does the resistance, but at an increasing rate. The two phases are quite distinct. Therefore, the spine behavior may be represented more reasonably by two numbers, each representing one phase. The first number represents the low-load response near the neutral position. We have termed this the neutral zone (NZ) (Fig. 1-29). The second number represents the spinal behavior beyond the neutral zone up to the end of the physiologic limit.

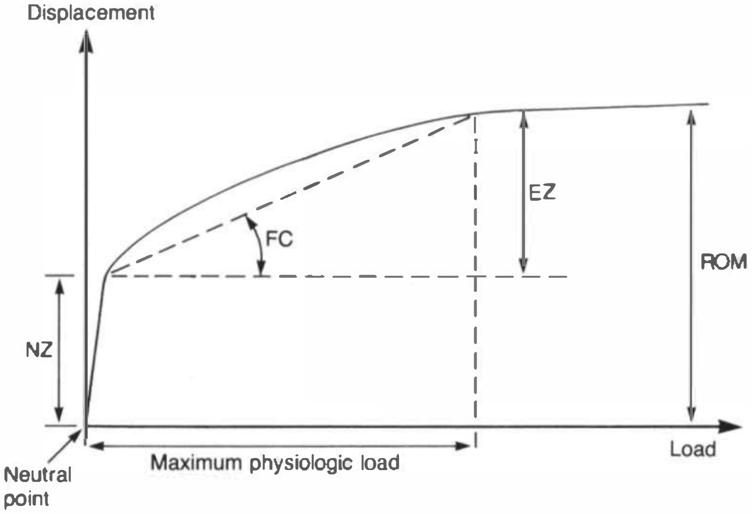


FIGURE 1-29 The load–displacement curve of the spine is generally nonlinear. (Note that the load is plotted on the horizontal axis, while the displacement is on the vertical axis.) At small loads there is relatively large displacement, while at larger loads there is relatively less displacement. This biphasic behavior is documented by four parameters. Neutral zone (*NZ*) is the displacement at low loads from the neutral point; elastic zone (*EZ*) is a displacement from the end of the neutral zone to the maximum physiological load; range of motion (*ROM*) is the sum of the neutral and elastic zones; and average flexibility coefficient (*FC*) is the elastic zone divided by the maximum physiological load.

We call this the elastic zone (EZ). The range of motion (ROM) is, then, the sum of the NZ and EZ. We define the average flexibility coefficient as the elastic zone divided by the maximum physiologic load. Unfortunately, most researchers have not distinguished between the two phases of the spine behavior. Generally, but not always, the stiffness or flexibility coefficients given in the literature represent the second phase. Thus, a significant amount of spinal motion that takes place around the neutral position is not accounted for.

Another complicating factor in the determination of the *elastic properties* of the spine is the fact that the real spine is viscoelastic. This latter concept implies that the mechanical behavior of the spine varies with the speed of loading during the experiment. Furthermore, if the spine is loaded several times in the same direction (e.g., several flexion motions), then the behavior of the spine will be different every time. It is a common practice in spinal biomechanics research to load the spine specimen several times during the experiment. This is the so-called preconditioning. However, it is not always reported in the research papers. These experimental design variations, together with the biological variations of the human spine, result in significant overall variability in the data.

Flexibility Measurements

The flexibility and stiffness coefficients, presented in Table 1-8, are average representative values based upon available studies. The significant variation in the data is due to the different methodologies used and the inherent biological variation, as mentioned. It should be emphasized, once more, that the spine behavior is quite complex (nonlinear, biphasic, and viscoelastic) and should not be represented by single numbers. The tables, however, do serve the purpose of providing a feeling for the relative flexibilities of the spine in different directions.

Compression, Tension, and Shear Compression loading, again because of its assumed clinical importance and simplicity of testing, has dominated the studies of physical properties of the spine.* Tension and shear, on the other hand, are probably the least studied.[29,189]

Experiments have shown that the FSUs are more flexible in tension than in compression in all regions of the spine. In the cervical and thoracic regions, the tensile flexibilities were about 380% and 160%, respectively, of the compressive flexibilities. In the

TABLE 1–8 Average Flexibility and Stiffness Coefficients of a Representative Functional Spinal Unit in Different Regions of the Spine. (Note that these numbers are simple representations of complex spinal behavior.)

	Forces					Moments			
	Tension (+FY)	Compression (−FY)	Ant. Shear (+FZ)	Post. Shear (−FZ)	Lat. Shear (FX)	Flexion (+MX)	Extension (−MX)	Lat. Bending (MZ)	Axial Rot. (MY)
Flexibility Coefficients (mm/kN for forces and deg/Nm for moments)									
Cervical	19.0	5.0	20.0	19.0	19.0	2.33	1.37	1.47	0.86
Thoracic	1.3	0.8	9.1	9.1	9.1	0.45	0.36	0.36	0.40
Lumbar	1.3	0.5	8.3	5.9	6.9	0.74	0.48	0.57	0.20
Lumbosacral						1.00	0.78	0.13	0.55
*Sacroiliac**	6.4	3.4	9.3	5.3	2.6	0.06	0.08	0.03	0.15
Stiffness Coefficients (N/mm for forces and Nm/deg for moments)									
Cervical	53	200	50	53	53	0.43	0.73	0.68	1.16
Thoracic	770	1250	110	110	110	2.22	2.80	2.80	2.53
Lumbar	770	2000	121	170	145	1.36	2.08	1.75	5.00
Lumbosacral						1.00	1.28	7.69	1.82
*Sacroiliac**	156	294	108	189	385	15.67	11.93	30.00	6.76

* One ilium fixed.

(Data from Berkson, et al.,[29] McGlashen, et al.,[150] Miller, et al.,[158] Moroney, et al.[157] Panjabi, et al.,[179,189] Tencer and Ahmed,[247] and Tencer, et al.[248])

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*See references 66, 102, 135, 153, 156, 157, 179, 189, 203, 217, 248, 267.

lumbar and sacroiliac joints, the corresponding values were 260% and 188%, respectively. The lower flexibility values for compression are probably due to the hydrostatic pressure within the disc and the loading of the facets.

The shear flexibility values, in general, are about equal in different directions (e.g., anterior, posterior, or lateral).[179,189,248] In the lumbar region, the shear flexibilities in the horizontal plane are ordered in the following manner, from lowest to highest: posterior shear, lateral shear, and anterior shear.[135,248] There is no consistent pattern of change with the levels of the spine. In general, the spine was found to be much more flexible in shear. When compared with the axial compression stiffness, the shear stiffness was only 25% in the cervical,[189] 9% in the thoracic,[179] 15% in the lumbar,[27] and 36% in the sacroiliac joints.[156]

Flexion, Extension, and Lateral Bending Although the compression behavior of the FSU has been more thoroughly studied, compared with the other load types, it has yet to be related to the clinically observed failures of the disc.[41,203,254] Rotatory loads (bendings and twists), on the other hand, are more likely to produce disc failures.[41,67]

The spine is more flexible (or less stiff) in flexion than in extension in all regions except the sacroiliac joint. The increased flexibility is about 25–60%, the lower value for the thoracic and lumbosacral joints and the upper value for the cervical and lumbar regions. On removal of the posterior elements, the extension flexibility increased so that there was no significant difference between the flexion and extension flexibility and stiffness values. But there was no increased flexion when the posterior elements were removed.[145] This implies that the posterior elements play a significant part in resisting extension but not flexion. There was no consistent variation of these properties with the spinal level in the cervical,[96,157,189,287] thoracic,[179,261] and lumbar[166] regions.

Flexibility values for lateral bending were in between the values for flexion and extension, except for the lumbosacral and sacroiliac joints where they were less than both the flexion and extension values. Similarly, removing the posterior elements had no effect on lateral bending.[145,233]

Axial Rotation This motion is probably more dangerous to the disc than any other, except for a combination of axial rotation and lateral bending.[67] The flexibility characteristics of axial rotation are markedly different from those of the other rotatory motions. In the cervical region, the spine is only 37% as flexible in torsion as compared with flexion. The torsional stiffness within the upper thoracic region is more or less constant, its value being about the same as that for flexion. The lumbar region has the lowest torsional flexibility. Compared with flexion, the average torsional flexibility is about 27%. There is an increase in torsional flexibility at the lumbosacral joint (55% of flexion) and especially at the sacroiliac joints (250% of flexion).

The effect of removal of the posterior elements on the torsional stiffness properties has also been studied. Although only a small change occurred in the upper thoracic region,[259] the effect is significant in the lumbar spine. For example, the stiffness of the L3–L4 FSU was reduced to almost one-fourth of its intact value. In a study of a large number of FSUs from the L1 to L4 region, removal of the posterior elements increased the rotation by 150% for the same torque.[233]

Combined Loads In a study by Lin and coworkers, mechanical tests on the lumbar FSUs were conducted in which combined loads were applied.[130] For example, in one test, compression, shear, and bending were simultaneously applied.[3] Such tests, in general, are better representatives of the situations *in vivo*. The difficulty in such tests, however, lies in choosing the right combination of loads that is representative of reality, because the loads *in vivo* are not known. A certain combination, namely, a maximum flexion combined with some lateral bending, resulted in disc herniation when a sudden compression was applied. For details, see the description in the section on the intervertebral disc. Also, epidemiologically combined loads have been found to be a significant risk factor in disc herniation.[114a]

Neutral Zone Measurements

The neutral zone is a relatively new kinematics parameter. We defined it in precise biomechanical terms earlier. Simply stated, it is a quantitative measure of joint laxity around the neutral position of the joint. It has been shown to increase with degeneration,[183] surgical injury,[1,129] repetitive cyclic loads,[133] and high-speed trauma.[181] Representative values of neutral zones of FSUs for rotatory motions from C0–C1 to L5–S1 are given in Table 1-9. Please note that these values are for motions to one side of the neutral position. No values are given for the

TABLE 1–9 Average Neutral Zones for Rotatary Main Motions in Degrees for a Representative Functional Spinal Unit in Different Regions of the Spine

Region	Flexion/ extension	Lateral bending	Axial rotation
C0–C1	1.1	1.6	1.5
C1–C2	3.2	1.2	29.6
C3–C6	4.9	4.0	3.8
C7–T1/T11–T1	1.5	2.2	1.2
L1–L2/L3–L4	1.5	1.6	0.7
L5–S1	3.0	1.8	0.4

(Data from Panjabi, et al.,[182a,189] and Yamamoto, et al.[265])

thoracic region because the data are not available from this region.

In flexion/extension, the neutral zone is largest in the lower cervical spine and smallest at the C0–C1 joint. In lateral bending, the largest neutral zone is again in the lower cervical spine region, while the smallest is at the C1–C2 joint. In axial rotation there is a large neutral zone of 29.6° at the C1–C2 level. Compared with the corresponding range of motion of about 40°, the neutral zone is a major portion of the range of motion at this level.

Preload Effects

In situ, the motion segment is subjected to the physiologic loads of motion during normal activities. In addition, there are much larger loads present as a result of body posture. Nachemson and Morris found these so-called compressive preloads at the L3–L4 FSU to be very high, about twice the body weight when standing in 20° flexed posture.[165] In our experimental study of the lumbar spine, we found unexpectedly that the addition of preload greatly affected certain stiffness values while hardly changing others. Furthermore, of those values affected, some were increased while others were decreased with the preload.[188] For example, the spine became less stiff in flexion and more stiff in axial rotation as a result of the addition of the preload. We may conclude by saying that the true stiffness and flexibility properties of the spine should be measured in the presence of suitable preload so as to simulate, as closely as possible, the conditions *in vivo*.

The Analysis of Preload

The preload *in situ*, such as the axial load on the disc or FSU due to body posture, has two origins. First, there is the direct compressive load due to the weight of the body part above the FSU (Fig. 1-30A); for example, the lumbar segments are subjected to the weight of the entire torso. Second, because the position of the center of gravity of the supported weight is anterior to the spine, the FSU is also subjected to large flexion (bending) moments that are counterbalanced by the ligament and back muscle forces. These ligament and muscle forces, in turn, apply compression to the FSU. All of these forces and moments that act on the disc–vertebra are shown on the free-body diagram in Figure 1-30B and are analyzed.[1]

Facet Joints

The torsional stiffness of the spine is largely determined by the design of the facet joints. The observations by Markolf of increasing torsional stiffness from T7–T8 to L3–L4, with the peak at T12–L1, and subsequent decrease in these values with the removal of the facet joints[145] can be explained on the basis of the changing patterns of the facet joints.

Two examples of facet joints are shown in Figure 1-31. When the plane of the facet joint is such as to allow nearly unhindered rotation of one vertebra with respect to the other, the motion segment has low rotatory stiffness, which is dependent on the contributions of the ligamentous structures only. The facets do not play any significant role. Such a case is that of the T5–T6 FSU (Fig. 1-31A). On removal of the facet joints, there is minimal change in the segment mechanics (Fig. 1-31B). On the other hand, the T12–L1 intervertebral joint has facet joints that effectively hinder the relative axial rotation (Fig. 1-31C). Here the facet joints play an important role. When they are intact they provide high resistance to axial torsion, and their removal significantly decreases the stiffness (Fig. 1-31D). This important role of the facet joints of the lumbar spine in resisting axial rotation has been well documented experimentally.[1,2,5,69,81,248] The same mechanism is present to varying degrees in the other regions of the spine.

A sudden change in the stiffness properties of a structure at a given point implies a stress concentration at that point that will eventually lead to mechanical failure. Such a point in the spine is represented by the T12–L1 FSU. This hypothesis is well supported by experimental studies[112] as well as clinical observation.[87] The highest frequency of spine injury is in the region of the thoracolumbar junction. As the above analysis showed, the abnormally high stiffness is the result of special orientation of the articulating facets. Anatomically, this articulation

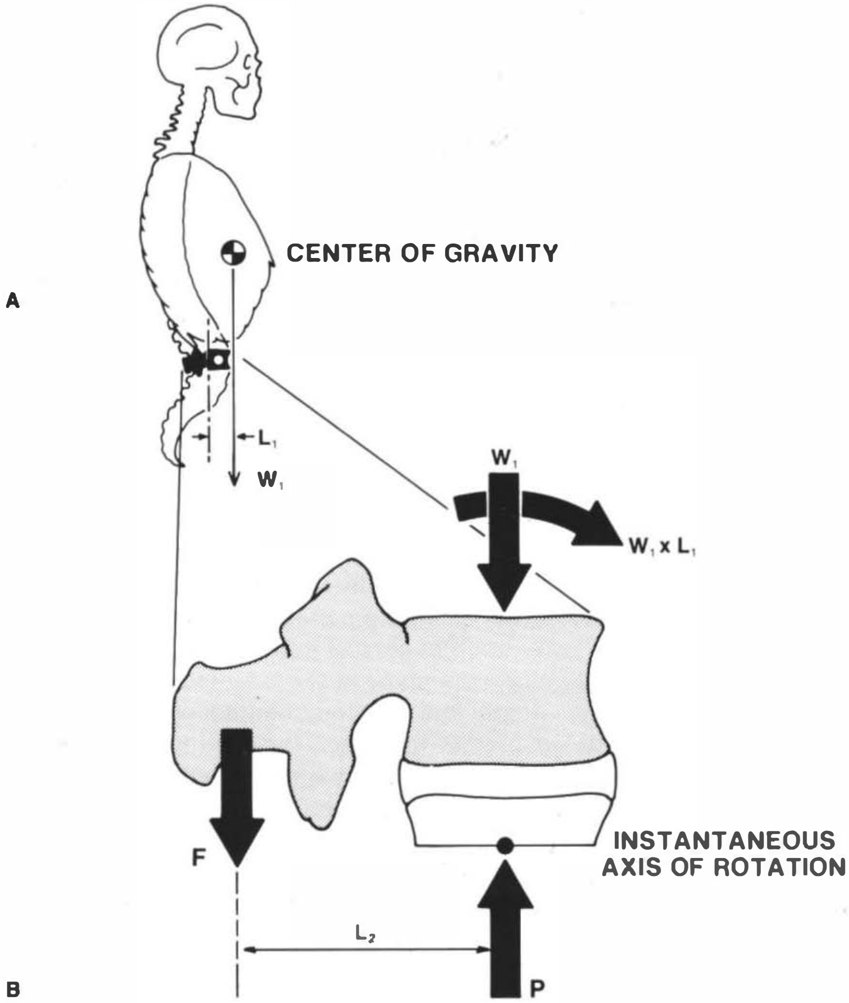

FIGURE 1-30 Analysis of preloads. **(A)** The functional spinal units of the lumbar spine have been observed to bear very large compressive loads (preloads). A simple mechanism of this phenomenon is depicted here. The center of gravity of the weight of the body parts above the functional spinal unit is anterior to the instantaneous axis of rotation of the segment. **(B)** A free-body diagram of the functional spinal unit is shown. The external loads are the weight of the body parts W_1 and the flexion bending moment $W_1 \times L_1$. The internal loads are the ligamentous and muscle forces F and the compressive reaction force acting at the instantaneous axis of rotation in the vertebra (i.e., the preload *P*). Length L_2 is the lever arm for the force F to the instantaneous axis of rotation. Equilibrium of the vertebra provides a measure of the preload. The force $P = W_1 + W_1 \times L_1/L_2$. See text and Note I for details.

(of highest stiffness) may vary among individuals from T9 to L1.

Age, Sex, and Degeneration

It is sometimes assumed that with age the disc space narrows and the discs become stiffer; it is also assumed that a herniated disc is biomechanically unsound.[94] In a carefully conducted study of lumbar cadaveric FSUs, Nachemson and colleagues made the following observations.[166] The disc height, even in a group of grossly degenerated specimens, was found to be average. In general, age was not related to the mechanical behavior of the FSU in any pronounced manner. The same was true for the disc level within the lumbar region. However, females were found to have more flexible spines as compared with the males. The most interesting finding, however, concerned the disc degeneration: no consistent correlation was observed between disc degeneration and the mechanical behavior. In a specimen with a grossly herniated disc, the mechanical tests showed it to have near normal behavior. We would like to make one comment about these quite interesting findings. This study measured only the elastic behavior. The viscoelastic behavior may turn out to be a more significant factor. As reported elsewhere, Kazarian found a relationship between disc degeneration grade and the viscoelastic creep behavior obtained under compression loading.[111] Also, neutral zones have been found to significantly increase with disc degeneration, especially in axial rotation and anterior/posterior shear motions.[183]

Viscoelastic Characteristics

Viscoelastic characteristics of the functional spinal unit have not been studied to the same extent as the elastic behavior, often because of the technical difficulties involved in doing the experiment and, more

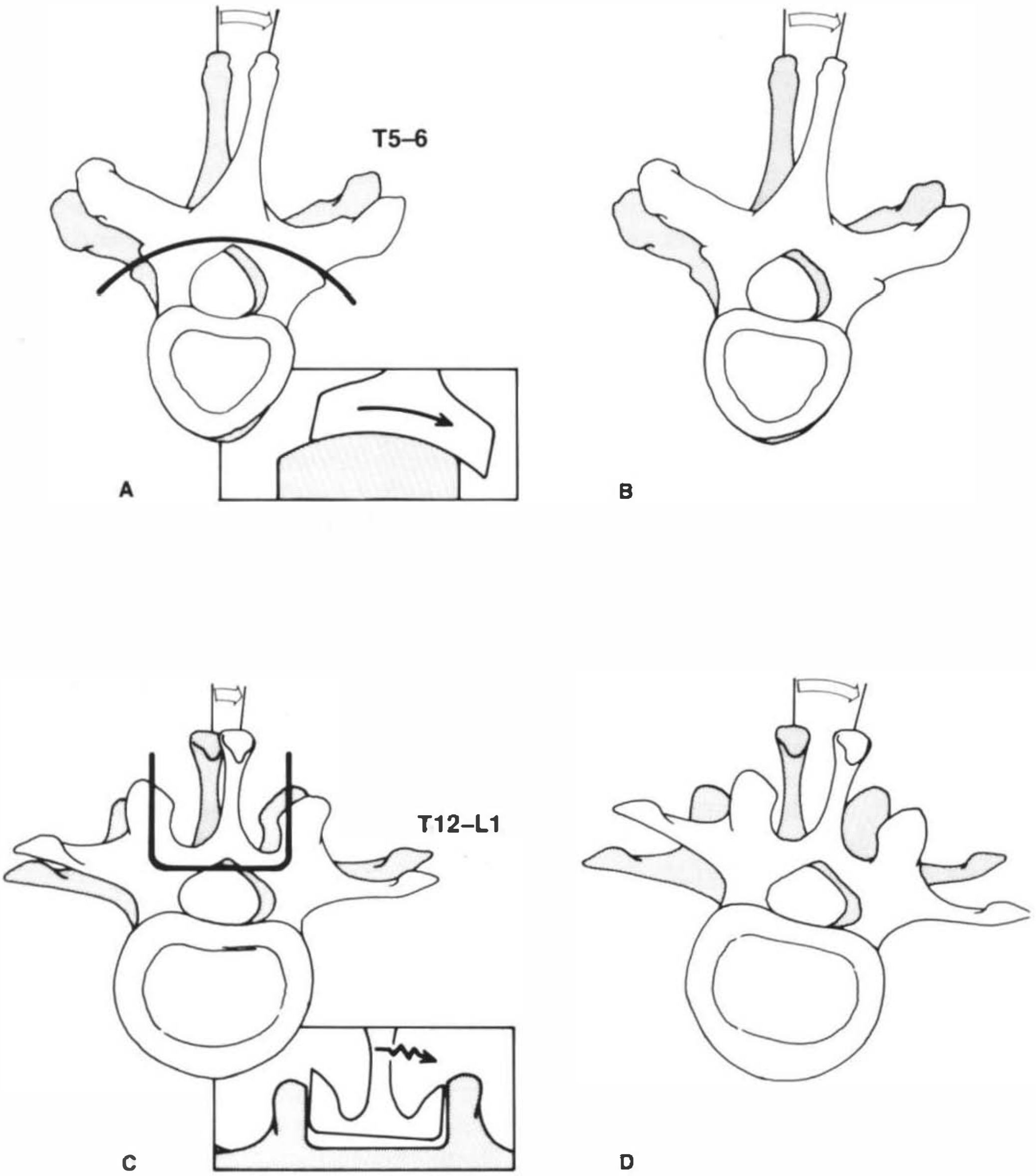

FIGURE 1-31 The role of facet joints. **(A)** The axial rotation of a vertebra with respect to the one below of a T5–T6 functional spinal unit is unhindered by the facet articulation. This is due to the orientation of the plane of the facet joints with respect to the instantaneous axis of rotation. For a given torque, a certain axial rotation is produced. **(B)** On removal of the facet joints there is no significant change in the axial rotation of the T5–T6 functional spinal unit. For the same torque as in A, about the same rotation is produced. **(C)** It is quite different for a functional spinal unit in which the facet joint articulation is oriented so that the two facets impinge on each other when the segment is subject to axial torsion. An example of this type of functional spinal unit is that of T12–L1, which has been found to have the highest rotatory stiffness of any functional spinal unit. Thus, for the same torque as in A, a smaller rotation is produced. **(D)** On removal of the facets of the T12–L1 functional spinal unit, the motion is unhindered and there is a significant change in the axial rotational stiffness. For the same torque as in A, much greater rotation is produced.

important, analyzing the data obtained.[43] In the lumbar spine, phenomena of creep under flexion moment[252] and under other physiologic loads have been studied.[187] In a recent study, the viscoelastic characteristics of the lumbar FSU were investigated under compression loading.[114] Creep test was performed. (The creep test consists of application of constant load and measurement of the resulting deformation as a function of time.) They found that the moderately degenerated specimen exhibited more initial deformation and approached the equilibrium at a more rapid rate than did the nondegenerated specimen. Using a three-parameter mathematical model (see Chap. 9 for description), they analyzed the viscoelastic curves. They found that: (1) the creep rate for the moderately degenerated (and older) specimens was twice as large as that for the nondegenerated (and younger) ones, (2) the mechanical parameters were not related to disc height, disc area, or disc level, (3) there was a trend for higher creep rate for the female specimens as compared with the male specimens, although this was not statistically significant, and (4) there was a direct relation between mechanical properties (viscosity and viscous modulus) and the bone mineral content, indicating interdependence of the mechanical properties of the disc and vertebra. Such interdependence has also been mathematically modeled using the FEM.[126]

Fatigue Tolerance

A functional spinal unit *in situ* is subjected to dynamic loads. Fatigue failure, by definition, may take place as a result of cyclic loads of moderate magnitudes well within the physiologic limits. Only a few studies have examined this behavior. We have mentioned, in the section on the intervertebral disc, one of the first studies by Brown and colleagues. Whole lumbar spines were studied for fatigue tolerance by subjecting these to cyclically varying loads.[41] The loads ranged from about 1800 N to 400 N at two cycles per second. Compressive fractures were found in all specimens at one or more levels after 200 to over one million cycles.

Recently, there have been several fatigue studies that have used modern biomechanical techniques.[4,134] Liu and colleagues studied the fatigue behavior of the lumbar FSU first in pure compression[134] and then in a separate study with torsional loads.[133] In the compression study, cyclic loads at 0.5 cycles per second were applied. The minimum was always 22 N, whereas the maximum ranged between 37 N to 80% of the expected failure loads for each specimen. The tests were stopped at 10,000 cycles. Two types of behavior were seen biomechanically. A group of specimens showed a large, abrupt increase in displacement between 1000 and 2000 cycles, while the rest of the specimens exhibited a moderate and monotonous increase up to 10,000 cycles. On radiographic and morphologic examination, the first group of specimens showed generalized bone failure of the end-plate and subchondral bone, while the second group had intact bone.

In the torsion fatigue study, the specimens were tested up to 10,000 cycles in two ways: (1) cyclic reversals to constant torque values (11–45 Nm) while displacements were monitored and (2) cyclic angular reversals to 1.5°. In the first group, failures were observed only in specimens that showed initial angular deformation of greater than 1.5°. The failures were of various components of the FSU. In the second group, an initial torque of about ± 10 Nm was measured, which decreased exponentially (at an ever-decreasing rate) to a stable value of about ± 2 Nm, reached after 3000 cycles. All specimens, after the 10,000 cycles, had increased "looseness." This last observation implies increased neutral zone due to fatigue. Unfortunately, the researchers did not measure the "looseness" as it was observed by manual twisting of the specimen before and after the fatigue test.

In another compression fatigue study, in which bone mineral content measurements were used, the loads applied were a certain percentage of the estimated failure loads.[90] The cyclic loads (average of 70% of estimated failure load) were superimposed over a physiologic preload at 0.5 cycles per second. In this study, detailed and extensive analysis of the data was carried out. We will mention here some interesting results. First, the axial stiffness (axial load divided by axial deformation) initially increased and then stabilized before failure. The stabilized stiffness value was found to be the highest for the nondegenerated and younger group. The stiffness decreased with increasing degeneration grade as well as age. Second, the investigators also obtained what is called the fatigue life curve: the stress versus the number of cycles to failure. The former is plotted on a linear scale, while the latter is plotted on a logarithmic scale. For engineering materials, this relationship is generally a straight line. The same was found to be true for the FSUs. The fatigue life (number of cycles to failure) was found to be inversely related to the applied cycle stress raised to a power of 14. Thus, even a small increase in stress will significantly decrease the fatigue life.

There have been other fatigue studies designed to answer a specific question, for example, the fatigue of the posterior elements of the lumbar spine[127] and the susceptibility of the pars interarticularis to fatigue failure.[51, 106]

Functional Biomechanics

Coupling of Intervertebral Motions

Each of the physiologic motions of the spine, such as bending and rotation, has been described separately for the sake of simplicity. However, they are inherently connected. This phenomenon, which is called *coupling*, is due to the geometry of the individual vertebrae and the connecting ligaments and disc, as well as the curvature of the spine. Mathematical models may be used to account for the curvature of the spine, but the coupled behavior must also be studied while determining the physical properties of the FSUs as well as the whole spine.

Two or more individual motions are said to be coupled (e.g., lateral bending and axial rotation or anterior translation with flexion) when one motion is always accompanied by another motion. The motion being produced by an external load is termed the main motion, and all the accompanying motions are called coupled motions. For example, when we flex the spine in the sagittal plane, the flexion rotation is the main motion and the accompanying anterior and inferior/superior translatory motions are called the coupled motions. As seen in Figure 1-32, a vertebra can move in six different directions (it is said to have six degrees of freedom). In other words, the three-dimensional motion has six motion components: three translations and three rotations. Theoretically speaking, any one of the motion components may be accompanied by five coupled motions. Therefore, in addition to the sagittal plane coupled translations, there may be, out of the sagittal plane, coupled motions associated with flexion due to congenital, degenerative, or traumatic asymmetry of the facet joints.

Experimental Measurements The phenomenon of coupling has been well documented experimentally. It occurs in the thoracic spine,[179, 261] but it is more common in the cervical spine[139, 157, 189] and the lumbar segments.[121, 188] It has also been observed clinically. To measure the coupled motions, the intervertebral motions must be measured three-dimensionally to provide valid representation of the actual behavior of the spine. Unfortunately, this is not always done.

Some studies have taken into account the coupling phenomenon while measuring the flexibility properties of the cervical, thoracic, and lumbar functional spinal units.[60, 80, 121, 179, 189, 248] Generally, twelve forces and moments are applied, and six translations and rotations of the upper vertebra in three-dimensional space are measured (Fig. 1-32).

In the thoracic region there is strong coupling between all motions in the sagittal plane (e.g., translation and rotation). The coupling of axial rotation with lateral bending is much stronger in the cervical and lumbar region than in the thoracic region. In addition, the lumbar region shows certain cross-coupling of all three rotations. In other words, when the lumbar motion segment is axially rotated, it bends in the frontal and sagittal planes. Also, when bent laterally, it simultaneously bends in the sagittal plane and rotates axially. However, bending in the sagittal plane does not produce two other rotations. This phenomenon of coupling can be visualized with the help of a curved bar. Application of either an axial torque or a lateral bending moment produces twisting and bending in a three-dimensional space. However, bending in the plane of curvature produces motion only in that plane, and, therefore, no lateral bending or axial rotation.

In a recent study, fresh cadaveric whole lumbar spine specimens (L1 to sacrum) were studied, specifically for their coupled motion patterns.[195] A distinguishing feature of this study was, besides measuring the intervertebral motions three-dimensionally, that the posture of the spine was varied from fully extended to fully flexed. In addition to the two extreme postures, the specimen was placed in half-extended, neutral, and half-flexed postures. In each of these postures, the L1 vertebra was subjected to right and left axial rotations as well as right and left lateral bendings. In response to each of the main motions of the top vertebra, the whole spine bent, twisted, and rotated in the sagittal plane. Two examples of this complex spinal behavior, with the spine in neutral posture, are shown in Figure 1-33.

Response to left axial rotatory torque (of 10 Nm) is depicted in Figure 1-33A. The nomenclature and sign convention for RX, RY, and RZ are those shown

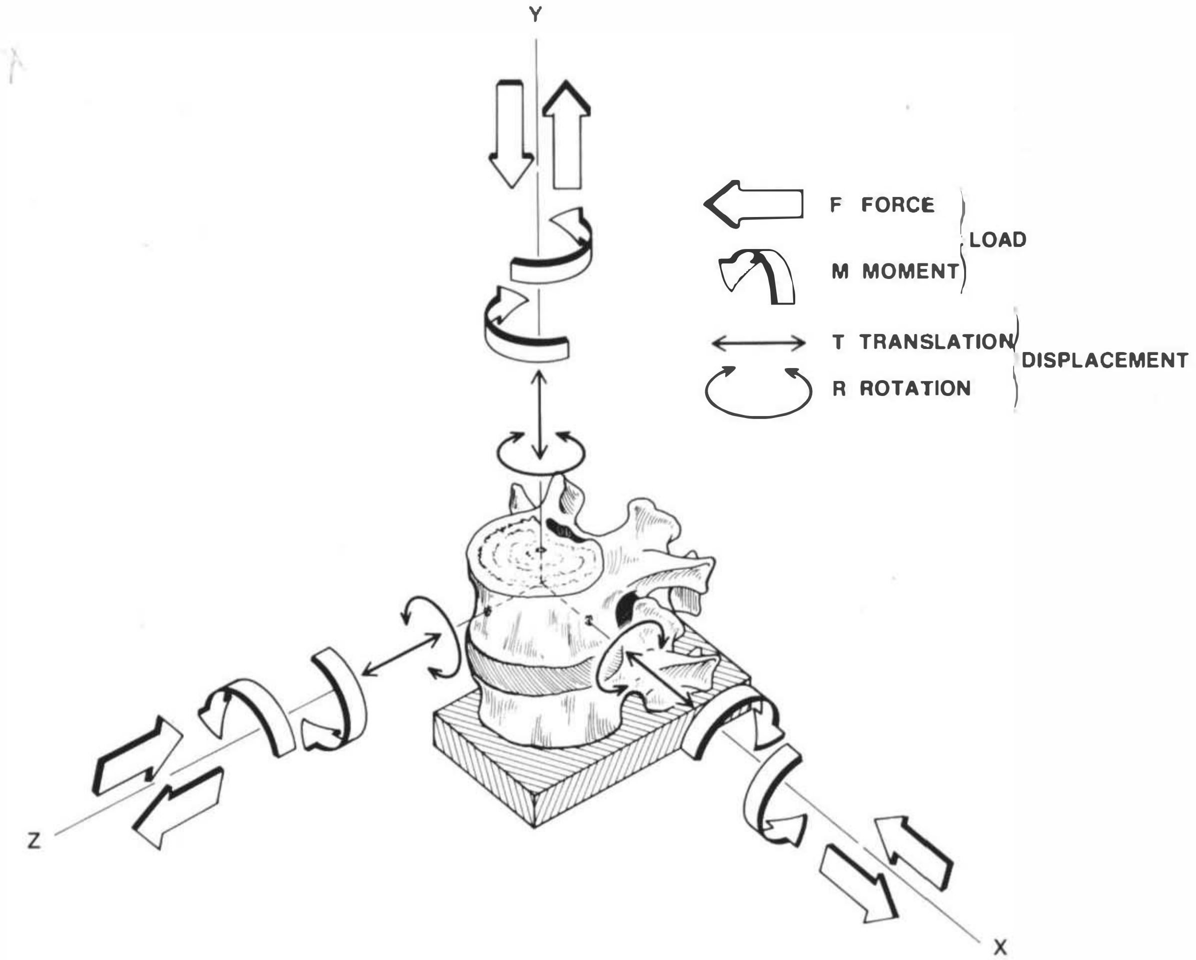

FIGURE 1-32 A three-dimensional coordinate system has been placed at the center of the upper vertebral body of a motion segment. The coordinate system is fixed in space. To document the complete mechanical behavior of the functional spine unit in the form of load-displacement curves, six forces along and six moments or torques about the three axes of the coordinate system are applied. These twelve-load components are depicted. The application of any one of the load components produces displacement of the upper vertebra with respect to the lower vertebra. The displacement consists of translation and rotation. These two motions can be further divided with respect to the coordinate axes. Thus, the three-dimensional displacement has six components, three translations along and three rotations about the three axes of the coordinate system. These are also shown. The point of application of forces and the point at which translations are measured significantly influence the load-displacement curves obtained. Thus, these points should be precisely defined for three-dimensional description of spinal physical properties. This is not true of the applied moments and measured rotations.

in Figure 1-32. Associated with the main motion of left axial rotation (+RY) is lateral bending (RZ), one of the two coupled motions. Its direction is to the right (+RZ) in the upper two joints and to the left (−RZ) in the lower two joints. The other coupled rotation is the sagittal plane rotation, which is flexion (+RZ) for all the lumbar joints.

Response to the right lateral bending moment of 10 Nm is shown in Figure 1-33B. The main motion (RZ) is about the same at all lumbar levels. In general, the coupled motions are less than half of the main motion. The sagittal plane rotation is always flexion (+RX) and is maximum in the middle of the lumbar spine. On the other hand, coupled axial rota-

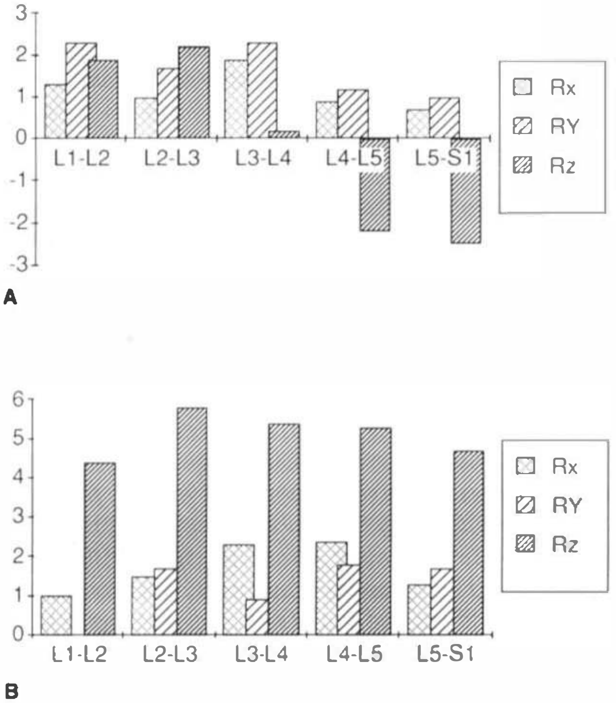

FIGURE 1-33 Coupling of intervertebral motions in lumbar spine. Using fresh cadaveric whole lumbar spine specimens, the coupling patterns between the rotatory motions of the lumbar spine were investigated. **(A)** The left axial rotation of L1 due to left axial torque (+MY) of 10 Nm resulted in the motions shown. The axial rotations (RY) decrease from L1–L2 to L5–S1. The coupled lateral bending (RZ) changes direction from right bending (opposite the applied torque) at L1–L2 and L2–L3 to left bending (same direction as the applied torque) at L4–L5 and L5–S1. The coupled sagittal plane rotation is flexion at all levels. **(B)** The right lateral bending due to 10 Nm of right bending moment (+MZ) resulted in the motions shown. The lateral bending was about the same at all levels. The coupled axial rotation was zero at L1–L2 and about 2° to the left (opposite in direction to the axial rotation) from L2–L3 to L5–S1. The sagittal plane motion was always flexion (+RX) at all levels. These results were obtained with the spine in neutral posture. The coupling patterns changed significantly as the spinal posture changed from flexion to extension. *(Data from Panjabi, M. M., et al.: How does posture affect coupling in the lumbar spine?* Spine, *14:1002, 1989.)*

tion (RY) is absent at L1–L2 and is about 2° at the other joints.

The study found that the coupled motions in the lumbar spine not only vary with the level but also are significantly affected, in magnitude and direction, by the posture of the spine.

In Vivo Studies There have been several observations, not quantified, of the existence of coupling (lateral bending with axial rotation or vice versa).[280] These were the result of studying anteroposterior radiographs in which the subject bent to the side. Quantified studies have been few. Pearcy and Tibrewal studied the three-dimensional movements of the lumbar spine from the neutral position in normal subjects in flexion and extension.[199] Their results are in excellent agreement with our *in vitro* studies described above, except for one coupled motion. They found the lateral bending motion to be accompanied by extension in the upper three lumbar segments, while we found all the lumbar segments to have flexion coupled motions.

MATHEMATICAL MODELS

The reason for including a short description of mathematical models of the spine is the fact that there is a strong link between the models and the physical properties of the spine. One of the goals of a mathematical model is to link the basic physical properties of individual components of a structure to the total behavior of the structure itself. By using the physical properties, the data of the functional spinal units and the mathematical modeling techniques, it is then possible to re-create the behavior of the entire spine. Even the effects of the rib cage and spine musculature may be incorporated by using the mathematical modeling process.

The mathematical model can simulate behavior of the spine in situations in which other means of investigation are not feasible. Clinical investigations are restricted to studies in which the subjects are not put in danger. Animal models are limited by the anatomic differences between the animal and the human spines. *In vitro* experiments using human cadavers are free from the restrictions of the clinical and animal studies, but they are expensive to conduct, provide limited information, and cannot simulate the muscle structures and the neuromusculature controls present *in vivo*. The mathematical models, which have been thoroughly validated for their accuracy, need not have any of these restrictions. In theory, at least, they have the potential for truly simulating the biomechanical behavior of the human spine *in vivo*. However, one must be careful to interpret the results of mathematical models in an appropriate biological and clinical perspective before the results are applied to patients.

With the use of high-speed modern computers, such models, once validated by experiments and clinical applications, can become powerful tools in the understanding, prevention, and treatment of disorders of the spine. Scoliosis serves as an important example. A suitable mathematical model may provide insight into the probable etiology by showing which mechanical anomalies (e.g., disc wedging, rib resection, and asymmetrical bone or muscle development) must be present to produce scoliosis. The methods of present treatment, such as the Milwaukee brace, Harrington rods, and the Dwyer procedure, can be evaluated and optimized. New treatment methods (e.g., correction of scoliotic deformity by stimulation of selective back muscles) can be analyzed.[224] In general, computer simulations are more economical, safer, and should precede the clinical trials.

Some of the earlier mathematical models that have the capability of simulating the three-dimensional behavior of the whole spine are by Belytschko and co-workers[27] and Panjabi.[177] The former model has been successfully tested in the simulation of scoliosis as seen clinically. The latter model, although it has the potential for greater validity, has not yet been tested. There are several new spinal models that have been developed recently. Most of these models address one specific problem, such as facet lumbar articulations,[235a] load determinations of spinal column and muscles in the lumbar region,[126a] or dynamic movements of cervical spine.[53a]

THE RIB CAGE

The rib cage has several important biomechanical functions related to the spine. It is a protective barrier for any traumatic impact directed from the anterior or the sides. It stiffens and strengthens the spine, thus providing greater resistance to displacement, which is advantageous when the spine is injured or has been disrupted by a disease. The stiffening effect of the rib cage is twofold. The costovertebral joint provides additional ligamentous structures that contribute to spinal stiffness. But the more important biomechanical aspect of the rib cage is its moment of inertia effect. The transverse dimensions of the thoracic spine are increased severalfold by the inclusion of the ribs and the sternum. The increased moment of inertia stiffens the spine when it is subjected to any kind of rotatory forces, such as bending moments and torques. Because the rib cage is part of the spine structure, it provides additional strength and energy-absorbing capacity during trauma.

Biomechanically Relevant Anatomy

The ribs are curved bones of elliptical cross-section joining the vertebral column to the sternum and thus forming a closed cylindrical cavity, the thorax. The first seven ribs join the sternum by means of individual costal cartilages, the next three by means of a fused costal cartilage, and the last two are free-floating, ending in the muscles of the abdominal wall.

The ribs articulate with the vertebrae at both the heads and tubercles. The head of the rib articulates with the sides of the corresponding vertebra and the one above, forming the costovertebral joint. This synovial joint has an articular capsule that is strengthened by the radiate ligament, which spreads from the head to the vertebrae and corresponding disc. The tubercle of the rib articulates with the transverse processes of the corresponding vertebra, forming the costotransverse joint. Although the articular capsule of the joint is weak, it is greatly strengthened by three costotransverse ligaments. The medial and lateral costotransverse ligaments extend from the tip of the transverse process to the neck and lateral aspect of the rib, respectively. The superior costotransverse ligament extends from the neck to the transverse process of the vertebra above.

The end of the rib joins its costal cartilage by means of the costochondral joint. The costal cartilage articulates with the sternum in several ways. The first rib is joined firmly to the manubrium by a cartilaginous joint. The cartilage of the second rib articulates with demifacets on both the manubrium and the body of the sternum by way of the synovial joints. The cartilages of the third to seventh ribs have small synovial joints with the body of the sternum.

Physical Properties

The Ribs

The only physical property of the ribs that has been studied is stiffness.[229] Schultz and colleagues fixed the rib at the heads, and applied loads to the free ends of the costal cartilage in six different directions: anterior, posterior, lateral, medial, superior, and inferior. The deformations of the loading point were measured and are summarized in Table 1-10. The highest stiffness was exhibited by the shortest rib (R-2) when pulled in the anterior direction, while the lowest stiffness (the highest flexibility) was shown by the longest rib (R-10) when loaded in the superior and inferior directions.

All ribs exhibited higher stiffness in the anterior direction as compared with the posterior direction. Generally, the ribs were highly flexible: a 10-N load (2.2 lbf) produced a deformation of 25 mm (1 in). The ribs also exhibited coupling effects, which are seen in functional spinal units. For example, superior loading not only produced superior displacement but also posterior and medial displacements. This was probably due to the curved geometry of the ribs. Although other coupled motions such as rotation were probably present, they were not measured.

Costovertebral and Sternocostal Joints

In another unique study, Schultz and colleagues measured the physical properties of the costovertebral and sternocostal articulations.[228] The results of stiffness of the two joints measured in various directions are shown in Table 1-11. The costovertebral joint, especially for the middle ribs, exhibited the highest stiffness in the lateral direction, while the lowest stiffness resulted from loads applied in the superior and inferior directions. Highly nonlinear behavior was observed. Initial motions about the neutral position could be accomplished by very small forces, while beyond this range there was a sudden increase in stiffness. The sternocostal joints, on the other hand, provided maximum resistance when loaded in the superior and inferior directions, especially for the joint of the second rib. The least resistance was offered by the inferior joints in the anterior and posterior directions.

TABLE 1–10 Stiffness Properties of the Human Ribs

Load Directions*	R-2† (N/mm)‡	R-4 to R-8† (N/mm)‡	R-10† (N/mm)‡
Anterior/posterior, ±z	1.50	0.75	0.30
Lateral/medial, ±x	0.75	0.40	0.25
Superior/inferior, ±y	0.40	0.25	0.20

* These directions refer to Figure 1-32.

† These are average stiffness values measured with a load of 7.5 N (1.7 lbf). The load and deformation point was at the head of the rib, while the other end of the rib was fixed. The ribs were much stiffer when the load direction was anterior compared with posterior.

‡ N/mm = 5.6 lbf/in

(Data from Schultz, A. B., Benson, D., and Hirsch, C.: Force deformation properties of human ribs. J. Biomech., 7:303, 1974.)

Panjabi and co-workers found the costovertebral joint to play a pivotal role in providing stability to the functional spinal units of the thoracic spine.[185] When flexion was simulated and all posterior elements, the posterior longitudinal ligament, and the posterior half of the disc were cut, the spine was on the verge of instability. Subsequent transection of the costovertebral joint consistently produced failure. When extension was simulated, the spine was found to be on the border of instability when the anterior longitudinal ligament, the disc, and the costovertebral joint were transected. Therefore, in the clinical situation, if there is evidence of the destruction of the costovertebral joint, one should question the ability of the spine to carry normal physiologic loads.

Functional Biomechanics

Agostoni and colleagues[7] subjected the relaxed rib cage of live subjects to a lateral squeezing force and measured the resulting changes in the lateral and frontal diameters. Patrick and colleagues[197] and Nahum and colleagues[167] studied the load–

TABLE 1–11 Stiffness Properties of Costovertebral and Sternocostal Joints

	Costovertebral Joints†			Sternocostal Joints‡		
Load Direction*	R-2 (N/mm)	R-4 to R-8 (N/mm)	R-10 to R-12 (N/mm)	R-2 (N/mm)	R-4 (N/mm)	R-6 to R-10 (N/mm)
Lateral, ±x	2.50	5.00	2.50	—	—	—
Anterior/posterior, ±z	1.50	1.75	1.50	2.50	2.5	0.50
Superior/inferior, ±y	0.75	1.50	1.00	3.0	1.50	0.75

* These directions refer to Figure 1-32.

† The load and deformation point was on the rib just beyond the costovertebral joint. The vertebra was fixed.

‡ The load and deformation point was on the rib just medial to the sternocostal joint. The sternum was fixed.

(Data from Schultz, A. B., Benson, D., and Hirsch, C.: Force deformation properties of human ribs. J. Biomech., 7:303, 1974. The values are average stiffness, measured with a load of 7.5 N (1.7 lbf).)

displacement behavior of the thorax for an anterior to posterior load applied at the sternum. On the average, the stiffness was found to be 10 to 20 times that of a single rib. From these studies, however, it is not possible to determine the contribution of the rib cage or its components to the stiffness and stability of the spine.

Although the individual components of the rib cage (ribs and their joints) are quite flexible, the rib cage as a whole greatly enhances the stiffness of the spine. Using a mathematical model of the thoracic and lumbar spine and the rib cage, Andriacchi and colleagues[18] performed computer simulations to determine the effect of the rib cage on: (1) the stiffness properties of the normal spine during flexion, extension, lateral bending, and axial rotation; (2) the stability of the normal spine under axial compression; and (3) the scoliotic spine subjected to traction. Also studied were the effects of removing one or two ribs or the entire sternum from an intact thorax. The mathematical model was validated by its ability to simulate reasonably well the experimental results of Agostoni, Patrick, and Nahum.[7, 187, 197] The findings of the model regarding stiffness, stability, and scoliosis were as follows:

1. The stiffness properties of the spine were found to be greatly enhanced by the presence of the rib cage for all four physiologic motions, especially extension (Fig. 1-34). Here, the stiffness with the thorax was nearly 2.5 times that of the ligamentous spine alone during extension.

   Removal of the sternum from the rib cage, on the other hand, had a profound effect, almost completely destroying the stiffening effect of the thorax (Fig. 1-34). The stiffness in all four types of physiologic motion decreased to values that were representative of the ligamentous spine without the thorax. This effect can be illustrated by the behavior of a thin-walled cylinder subjected to bending and torsion loads before and after a narrow longitudinal strip is removed.[I] Removal of one or two ribs, as is sometimes carried out in a scoliosis operation to obtain optimum correction, did not affect the stiffness properties significantly.
2. The rib cage was found to increase the axial mechanical stability of the spinal column in compression by four times. This relative increase can be put into perspective by the fact that, without the muscles and the rib cage, a ligamentous spine in an upright position could support an axial compressive load of only 20 N (4 lbf).[137]
3. Finally, the application of traction to the spine was simulated by the computer model for both the normal and scoliotic spines. Although the axial stiffness of the normal spine increased by 40% because of the presence of a rib cage, there was no corresponding increase for the scoliotic spine. In addition, the scoliotic spine was found to be about 2.5 times as flexible as the normal spine when both were subjected to traction (Fig. 1-35). Although the authors of the paper do not comment upon this finding, we believe it has the following biomechanical explanation. The extra mobility of the scoliotic spine in the axial direction was probably due to the additional curvature in the frontal plane that is present in a scoliotic spine. These curves straighten when the axial load is applied, giving the impression of a more flexible spine. (This is similar to the axial stiffness of a straight steel wire as compared with that of a spiral spring made of the same wire.) In the computer simulation of scoliosis, the physical properties of the ligaments were not altered.[K] Therefore, the observed flexibility was due to the abnormal geometric curvature of the spine only.

SPINAL MUSCLES

The spine, with its ligaments intact but devoid of muscles, is an extremely unstable structure. We have stated previously that a fresh cadaveric spine (without the rib cage), oriented vertically and fixed at the sacrum, could carry a maximum load of 20 N (4 lbf) placed centrally at T1.[137] Any additional load would permanently displace the spine from its central position. The muscles and the complex neuromuscular controls are required: (1) to provide stability of the trunk in a given posture and (2) to produce movements during physiologic activity. The muscles may also play a role in protecting the spine during trauma in which there is time for voluntary control, and possibly in the postinjury phase.

Biomechanically Relevant Anatomy

The muscles that directly control the movements of the vertebral column may be divided into categories according to their position: postvertebral and prevertebral.[77]

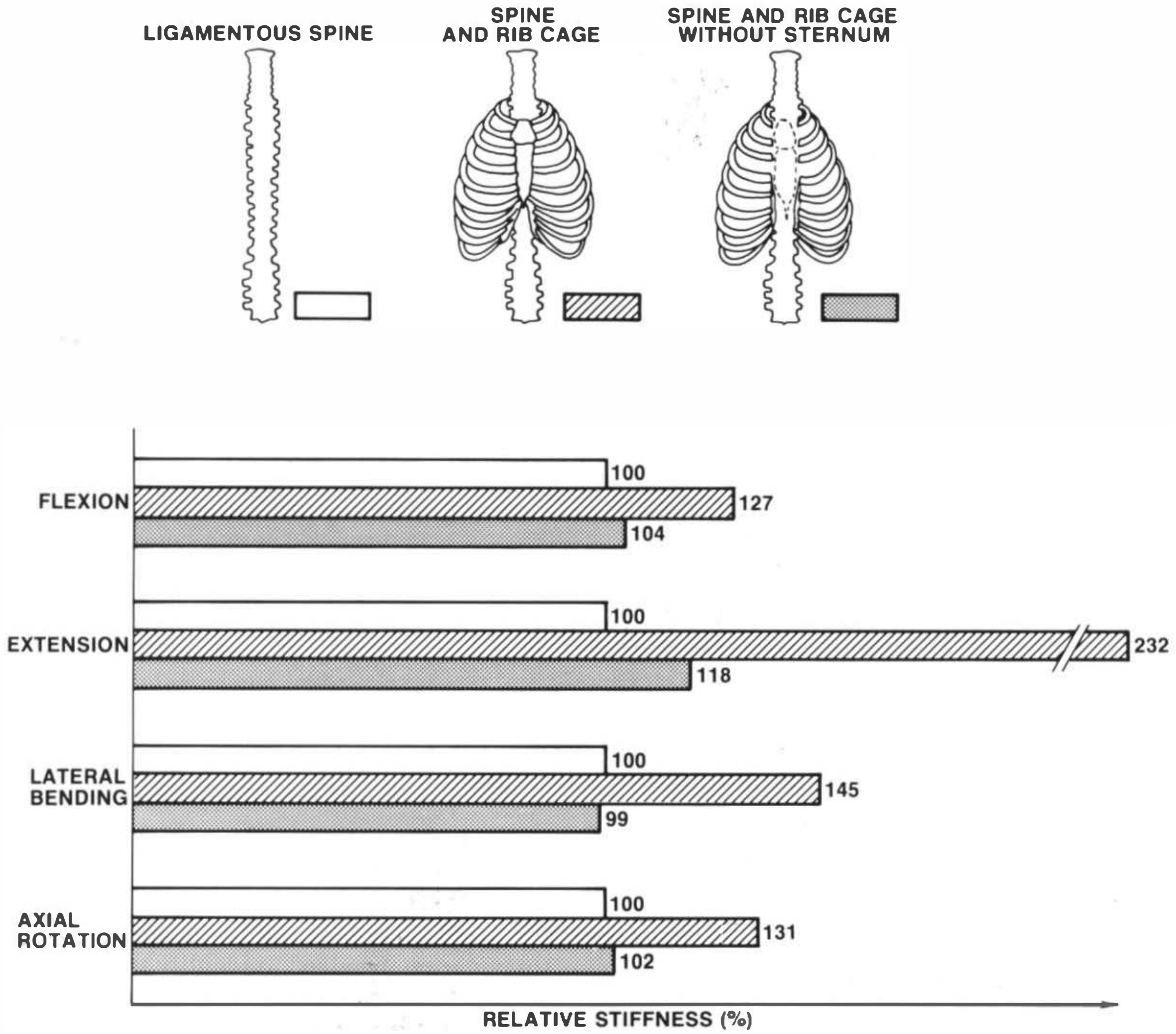

FIGURE 1-34 The role of the rib cage in enhancing the overall stability of the spine is depicted. The stiffness values were computed for each of the three structures during four different physiologic motions: flexion, extension, lateral bending, and axial rotation. The results of the relative stiffness values for the three structures are presented here in the form of horizontal bar graphs. For each motion, the "ligamentous spine" has been assigned a value of 100%. Note that the significant increase in stiffness achieved during all four physiologic motions due to the addition of the rib cage is entirely lost when the sternum is removed. *(Results based upon mathematical model by Andriacchi, T. P., Schultz, A. B., Belytschko, T. B., and Galante, J. O.: A model for studies of mechanical interactions between the human spine and rib cage. J. Biomech., 7:487, 1974.)*

The postvertebral muscles may be further divided into three groups: deep, intermediate, and superficial. The deep muscles consist of short muscles that connect adjacent spinous processes, *musculi interspinales;* adjacent transverse processes, *musculi intertransversarii;* transverse processes below to laminae above, *musculi rotatores;* and in the thoracic region, transverse processes to the ribs, *musculi levatores costarum.* The intermediate muscles are more diffused, but certain components can be identified. These muscles arise from the transverse processes of each vertebra and attach to the spinous process of the vertebra above. According to the regions, they are the *multifidus* (lumbosacral),

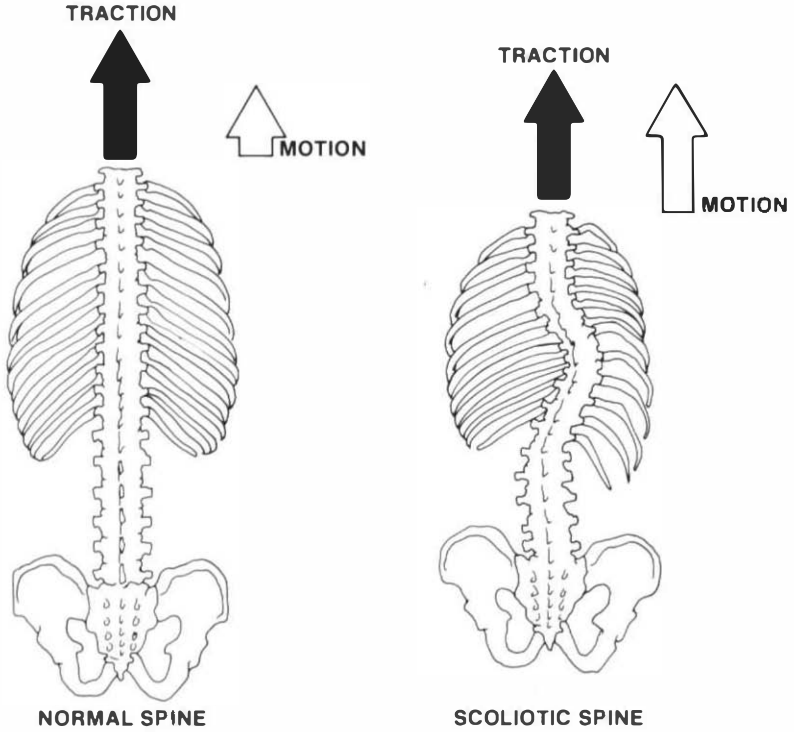

FIGURE 1-35 Relative flexibility of a normal and scoliotic spine in the axial direction is depicted in this diagram. Under the application of the same amount of traction force to the two spines, the motion produced at T1 with respect to the pelvis was found in the scoliotic spine to be about 2.5 times that in the normal spine. *(Results based upon computer simulation by Andriacchi, T. P., Schultz, A. B., Belytschko, T. B., and Galante, J. O.: A model for studies of mechanical interactions between the human spine and rib cage. J. Biomech., 7:487, 1974.)*

semispinalis thoracis, semispinalis cervicis, and *semispinalis capitis.* Finally, the superficial postvertebral muscles, collectively called the *erector spinae,* are the *iliocostalis* (most laterally placed), the *longissimus,* and the *spinalis* (most medially placed).

The prevertebral muscles are the four abdominal muscles. Three of the muscles encircle the abdominal region. They are the *external oblique, internal oblique,* and *transversus abdominis.* The fourth muscle is the *rectus abdominis,* located anteriorly at the midline. The four muscles are arranged in distinctly different directions. Anatomy of some of the lumbar spine back muscles has recently been described.[140,141,142]

A schematic representation of the muscles surrounding the spine in the lumbar region[110] is shown in Figure 1-36. Some recent studies have provided us with quantitative information concerning the location, orientation, and size of the muscles. These types of data are extremely important for the understanding of the biomechanical function of each muscle. Just as for the ligament function (see p. 24), these geometric parameters determine the muscle function.

Goel and co-workers obtained three-dimensional coordinates of the origins and insertions of 22 muscles of the cervical region.[82] The technique they used consisted of implanting radiopaque steel markers (hypodermic needles) to identify the origin and insertion points of various muscles in cadavers. Two orthogonal radiographs (anteroposterior and lateral) were obtained. The marker images were digitized, and, using computer programs, three-dimensional coordinates of the muscles were computed. Dumas and associates used a somewhat different approach.[58] They directly digitized the origin and insertions of the muscles using an electromechanical three-dimensional digitizer. They obtained not only the coordinates of the origin and insertion of the lumbar muscles but also the line of action (straight or curvilinear) of each muscle.

The above methods are limited to the cadaveric studies and do not provide the size of the muscles. With the use of computerized tomography (CT) scans, moment arm lengths (location of the center of the muscle cross-section) and cross-section sizes of some of the muscles of the lumbar spine have been obtained.[149,169] For example, at L4–L5, the psoas muscle had an area of 17.6 cm^2 (one side), while the

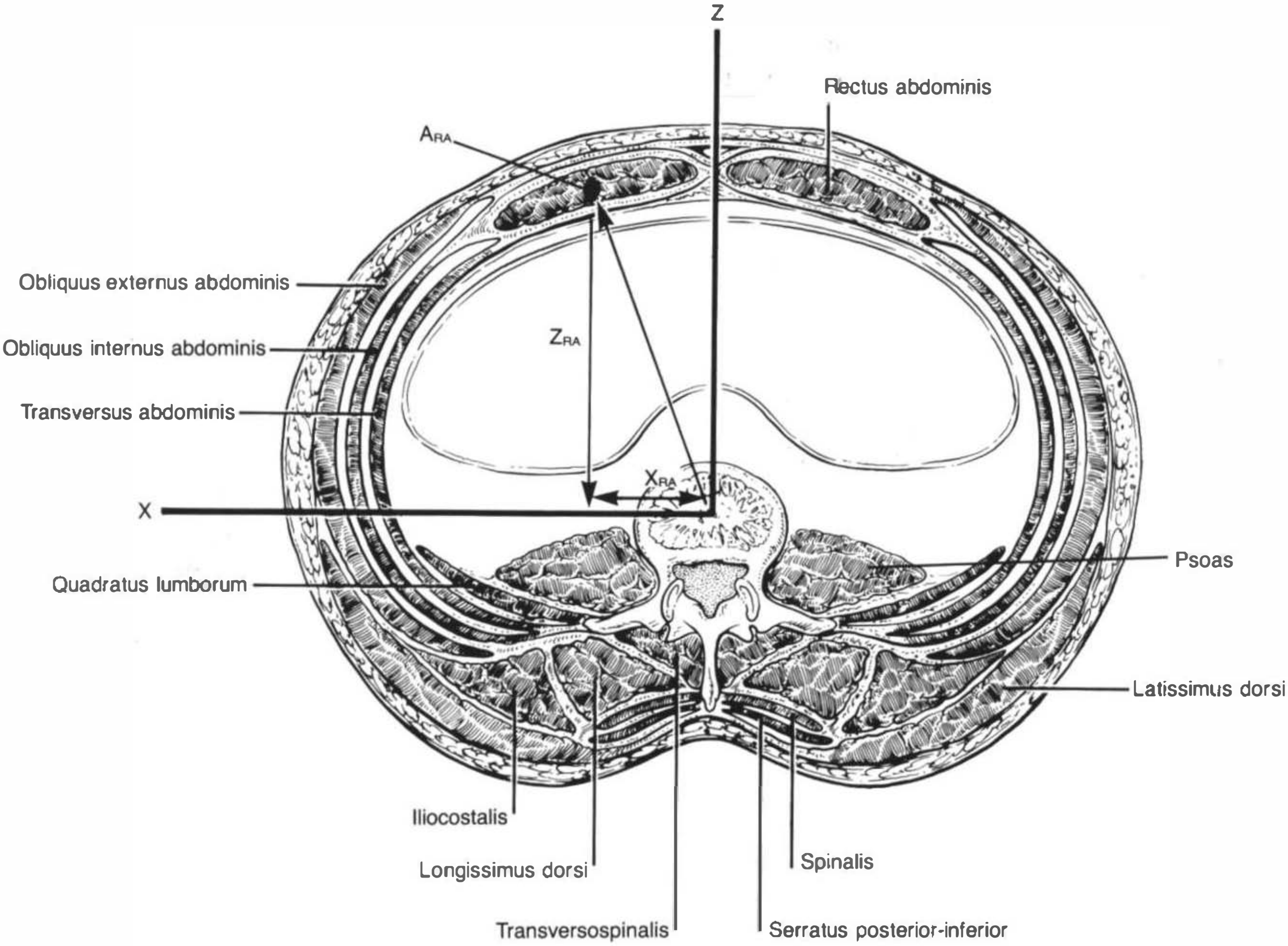

FIGURE 1-36 A horizontal cross-section (plane ZX) through the lumbar spine showing the muscles that help stabilize the spine and produce movements. Quantitative data are exemplified by the left rectus abdominis (*RA*). The coordinates X_{RA} and Z_{RA} of the centroid of the muscle cross-section and its cross-sectional area A_{RA} provide the necessary information to compute its contribution to spinal stability and motion. (This assumes that the muscle force is proportional to its cross-sectional area, which has not yet been fully established.)

external oblique area was 10.6 cm^2. By taking into account their corresponding locations with respect to the center of the disc, the moment-producing effects of the two muscles were in reverse order.

Because of its noninvasive nature and better definition of the soft tissue, magnetic resonance imaging (MRI) is ideally suited for obtaining the quantitative anatomic data of the back muscles. This has been done recently by Tracy and co-workers, who obtained size and location data of six lumbar spinal muscles.[251]

Physical Properties

The spinal muscles, like most other muscles, have several biomechanical functions. Through their activity they produce body movements by generating bending moments and torques. Through the same mechanism they also perform tasks and resist external loads. Most important, they provide dynamic stability to the spine where very little exists. Two mechanical characteristics are necessary to provide these physiologic functions. First, the muscles must generate force isometrically as well as with changing length. Second, they must increase the stiffness of the spinal system (ligamentous column and the surrounding musculature), thus increasing stability.[28,49]

The amount of force a muscle can maximally develop depends upon several factors (e.g., the length at the start of the contraction). Figure 1-37A shows the well-known force–length curve of a muscle.[23,174] Note that the maximal force is developed at about 120–130% of the resting length of a muscle, while at about 50% of its resting length the muscle develops very little force. The passive force–length curve is also shown in Figure 1-37A. The maximum force per unit area of the muscle has been determined by several researchers with widely varying results, ranging from 29 to 93 N/cm^2 for different muscles of the extremities.[107,147] With the use of strength-testing machines and magnetic resonance images, the maximum stress that the erector spinae can develop has been determined to be 48 N/cm^2.[213]

As mentioned earlier, the muscle must provide stiffness to stabilize the spine. Figure 1-37B shows the stiffness characteristics of the cat soleus muscle when the muscle is stimulated (above curve) and when it is not (lower curve).[103,152] Note that the active stiffness increases rapidly at small force levels and then saturates at higher levels of the force. Thus, one may hypothesize that, while resting, the spinal column is quickly stabilized when subjected to reasonable physiologic loads. But as the load increases to a certain level, there is no further increase in the stiffness of the muscles. In effect, this mechanism sets an upper load limit for the spinal column stabilized by the muscles.

Biomechanical Function of a Muscle

The inactivated muscle has physical properties that are similar to those of other noncontractile soft tissues. The mechanical output of an active muscle is dependent upon the external load and the muscle length. The passive muscle resists, and the active muscle produces force that seems to be related to the cross-sectional area of the muscle.[95] A representation of the active muscle function by a mathematical

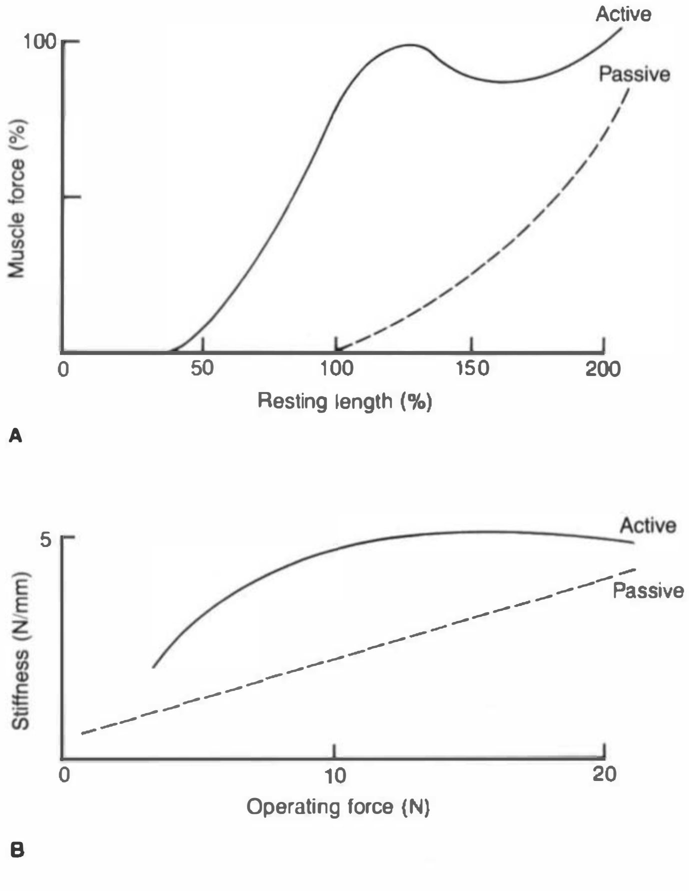

FIGURE 1-37 Physical properties of a muscle consist of its passive and active characteristics. **(A)** The force developed by a muscle (in percent of its maximum capacity) is plotted as a function of the muscle length (in percent of its resting length). Very little force is produced at about half its resting length, while the maximum force develops at 120–130% above the resting length. Also shown is the length force curve of a passive muscle. *(Modified from Basmajian, J. V., et al.: Muscles Alive: Their Functions Revealed by Electromyography. 5th ed. Baltimore, Williams & Wilkins, 1985; and Ottoson, D.: Physiology of the Nervous System. New York, Oxford University Press, 1983.)* **(B)** Stiffness of a muscle (cat soleus) as a function of the operating force. The stiffness of the muscle without stretch reflex increases linearly with the force applied to the muscle. On the other hand, the active muscle stiffness increases rapidly at small force levels and then saturates at the higher levels of the force. *(Adapted from Hoffer, J. A., et al: Factors affecting the gain of the stretch reflex and soleus stiffness in premammillary cats. Soc. Neurosci., [Abstr] 4:935, 1978.)*

model was proposed in 1939 by Hill.[99] A modified Hill's model that also includes the passive behavior of the muscle, according to Fung,[75] is diagrammed in Figure 1-38.

The model consists of three elements—two springlike elastic elements (parallel and series) and one contractile element under the control of a neuromuscular signal. The passive behavior of the muscle is completely represented by the parallel element, because the contractile element is inactive, and, therefore, no force is transmitted by way of the series element. When a muscle is voluntarily contracted, it may remain in a fixed position with no change of muscle length (isometric contraction), or it may contract and shorten (isotonic contraction) to provide work against an external load. In both situations, the series element shares the load together with the parallel element. This effectively increases the muscle stiffness. It should be emphasized that the mathematical model presented in Figure 1-39 is not a physical representation of a muscle, but it is a simple and precise way to describe the actual mechanical behavior of the muscle. Such models have been used to study the protective role of the back muscles of the spine in front-end auto collisions.[240]

In general, the purpose of a muscle force is to produce torque or moment across one or more joints.

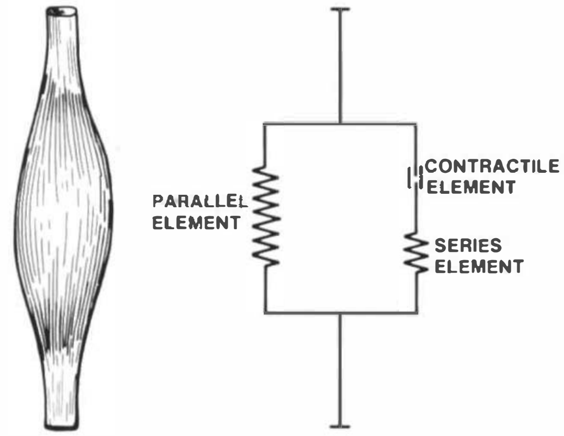

FIGURE 1-38 Functional model of muscle. Physical properties of a muscle are quite different when it is in passive or in active state. Both these aspects of muscle behavior may be represented in a quantifiable manner by the three-element mathematical model shown on the right. The model consists of a parallel element, representing the passive elastic behavior of the muscle, and a series element that, together with a contractile element, represents the active elastic behavior of the muscle. The parallel and the series elements have constant stiffness characteristics for a given muscle, while the contractile element is variable depending upon the activity of the muscle. Such models for individual muscles may be incorporated into the mathematical models of the spine to represent the total active behavior of the entire spine.

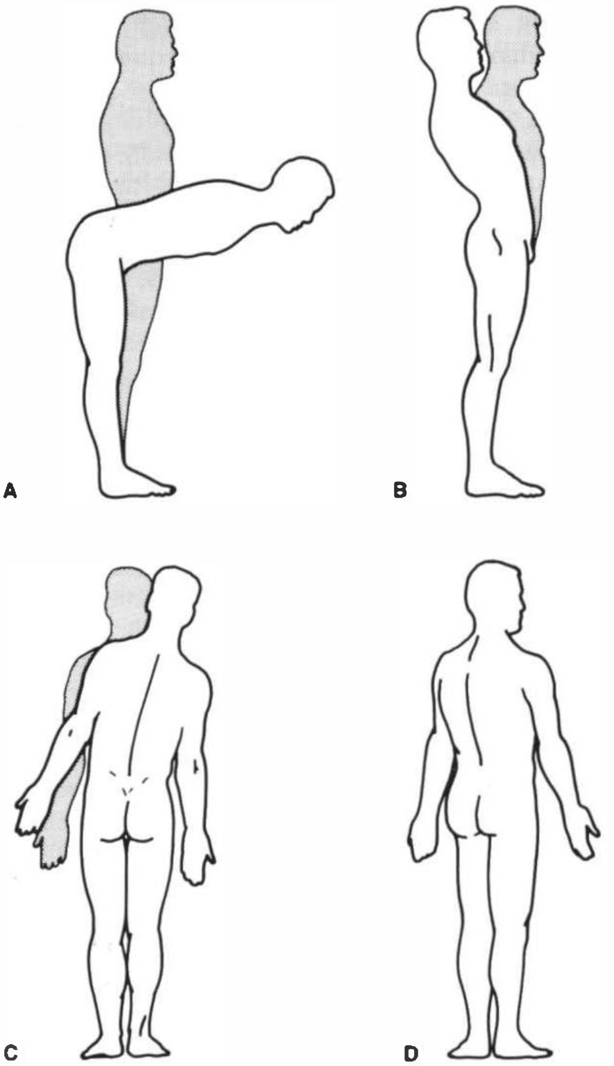

FIGURE 1-39 Muscle activity during the four physiologic motions is presented in this composite diagram. **(A, B)** In flexion and extension, gluteus and erector spinae muscles are active. **(C)** Lateral bending is achieved by an imbalance of muscle forces on both sides of the back. There is greater muscle activity on the ipsilateral side. **(D)** During axial rotation, erector spinae muscles on the ipsilateral side, the rotators and multifidi on the contralateral side, and the gluteals on both the sides were found to be active.

This force results in a torque that resists or does work against an external load. In addition, there are large compressive forces created at the joint between the two bones. This compressive, joint-reaction force is equal in magnitude to the vectorial sum of all the tensile muscle forces across the joint. An example of this is the large preload to which the functional spinal units are subjected in normal erect posture (see Fig. 1-30).

Evidence is accumulating that the muscles have other important functions besides the production of motion or resistance of an external load.[140,205] These functions are of at least two kinds. First, a spinal column without the muscles is not stable and cannot carry even the weight of the trunk. The requirement for maintaining stability becomes even more critical when the spinal column is required to support an external load or carry out a certain task. Muscles provide the necessary spinal stability when the external load is imposed upon it, as well as during every instant of the physiologic motion.

Second, activation of a muscle in response to an external load or a physiologic motion may produce unwanted intervertebral motions. This tendency must be counterbalanced by other muscle forces that help block these unwanted motions. It is for these reasons that some muscles that would ordinarily not contribute to the development of the particular torque or motion have been found to be active.[205]

Measurement of muscle action and forces may be documented by electromyographic studies. Although no definite relationship has been established between the electromyographic signal and the muscle tensile force, it is a monotonic relationship, implying an increasing signal with increasing muscle force.[173] Thus, with the present knowledge it is difficult to quantify the precise force a given muscle exerts. However, its electrical activity can be documented, and this signal gives some indication of the muscle forces.

Functional Biomechanics

One of the two practical motivations for biomechanical studies of the back muscles has been to predict the severity of an industrial task. An increased external load has been observed to worsen low back pain in a patient. Therefore, the loads on the spine, and in the muscles surrounding it, have been taken as indicators of spinal load severity. Because neither the loads on the spine nor tensions in the muscles can be directly measured, a variety of indirect methods of investigation have been developed and used in various studies. These include electromyographic (EMG) studies using bipolar surface and needle electrodes,[13,72] intradiscal pressure measurements using a tiny pressure transducer on the tip of a needle,[14,181] intra-abdominal pressure measurements using a pressure-sensitive radio transducer,[257] and mathematical models of the spine.[228] The studies presented below have used a combination of these methods. It should be re-emphasized that none of these methods is a direct measurement of the loads on the spine or tensions in the muscles. Furthermore, the studies are limited in their scope because of the general limitations associated with using living human volunteers. Therefore, the results obtained need to be interpreted carefully. There are several additional limitations. For example, the electromyographic signals of the muscles have not yet been directly related to the muscle tension; the results presented in a study are valid only for those muscles which were instrumented for electromyographic signals or modeled in a mathematical model; and the mathematical model predictions are limited to those postures which require no antagonistic muscle activities.

The second clinically important reason for the study of the back muscles has been to document the precise role of a particular muscle in producing a body movement or resisting an external load.

Described below are several studies that have increased our understanding of the role of the back muscles in maintaining posture, carrying loads, and producing physiologic motions of the back.

Posture

In the relaxed standing posture, the activity of the back muscles is generally low, especially in the cervical and lumbar regions. Slight activity of the abdominal muscles has been reported, but not simultaneously with activity of the back muscles.[18] Some activity in the vertebral portion of the psoas major muscle has also been measured.[182] These findings can be explained biomechanically. The ligamentous spine supporting the weight of the trunk is inherently unstable in its central position. A shift of the center of gravity of the trunk in the horizontal plane requires an active, counterbalancing muscle force on the opposite side. Therefore, an anterior, posterior, and lateral shift of the center of gravity activates the back, abdominal, and psoas major muscles, respectively. Morris and colleagues found the longissimus dorsi and rotatores spinae to be continuously active during standing.[158]

In the unsupported sitting posture, the muscle activity in the lumbar region was found to be about the same as that in the standing posture. Andersson and Örtengren observed that there was somewhat higher activity of the back muscles in the thoracic region while sitting as compared with the activity found in the standing posture.[12]

External Loads

The loads on the lumbar spine are significantly affected by the loads carried in the hands or supported by the trunk, as well as by the posture of the spine. A weight carried in the hands with the spine in neutral posture did not significantly increase the erector spinae muscle signal, according to Floyd and Silver, while Schultz and co-workers found a linear relationship between the load held in the hands and the EMG activity of the back muscles.[73,226] In the latter study, the muscle activity also increased as the hands holding the loads were moved anteriorly. Flexing the spine produced further increases in the back EMG signals. Mathematical models of the spine predicted muscle tension, supporting the EMG measurements. The models also computed the associated increases in compression loads on the spine. Placement of the load on one side of the spine asymmetrically loaded the spine and activated the back muscles on the contralateral side.[73,227] Horizontal loads, such as pulling and pushing, have also been studied. In general, the spinal loads increased. The rectus abdominis became active during simulated pushing (extension resistance), but this did not happen with pulling (flexion resistance).

The mathematical model used in the above study was validated against the measurements of intradiscal pressures in the third lumbar intervertebral disc.[227] The disc pressure measurements as indicators of spinal compression load had been previously validated with fresh cadaveric specimens.[161] The model predictions showed highly significant correlations with the loads on the spine. Therefore, authors suggest that to evaluate a situation such as an industrial task, where it is necessary to estimate the loads imposed on the spine as a result of a certain posture and/or external load, it is possible to compute the resulting loads on the spine without using any measurements of myoelectric activity, intradiscal pressure, or intra-abdominal pressure. Instead, measurements of posture of the spine, location of the external load, and magnitude of the load are sufficient for a validated mathematical model to accurately predict the spinal loads and back muscle tensions. Such predictions are more accurate when loads are in the sagittal plane and are of reasonable magnitude. They are less accurate if the loads are out of the sagittal plane (e.g., bends and twists)[225] and are of large magnitude.[230]

Muscles produce physiologic motions of the spine (Fig. 1-39) and, as we will see, also help stabilize the spine. A brief description of the role of the muscles during each of the four physiologic motions is provided below.

Flexion

Bending forward is a two-part movement involving both the spine and the pelvis. The first 60° of movement, on the average, are due to flexion of the lumbar motion segments. This is followed by an additional movement at the hip joints of about 25° (Fig. 1-40). In extension from the fully flexed position, the movement is reversed, so that at first the pelvis rotates backward, followed by extension of the lumbar motion segments to the neutral position.[53,68]

The muscle activity closely follows the pattern of motion. Initially, the pelvis is locked, as demonstrated by strong myoelectric activity of the gluteus maximus and medius and the hamstring muscles.[44] As flexion progresses, the increasing bending moment due to the weight of the trunk is balanced by the corresponding increase in the activity of the erector spinae muscles and the superficial muscles of the back.[13,74] This back muscle activity was found to be directly proportional to the sine of the flexion angle. However, on reaching full flexion, there is complete relaxation of these muscles (Fig. 1-40).[73,232] At this point the ligaments provide the major share of the required bending movement, while the passive extension of the muscles supplies the remainder. Morris and colleagues found most of the back muscles to be active during flexion. At full flexion, however, all muscles became inactive except the iliocostalis dorsi.

Extension

Myoelectric activity in the back muscles has been shown to occur at the beginning and at the completion of full extension from the neutral position, with only slight activity between these two extremes.[158] The abdominal muscles, on the other hand, show

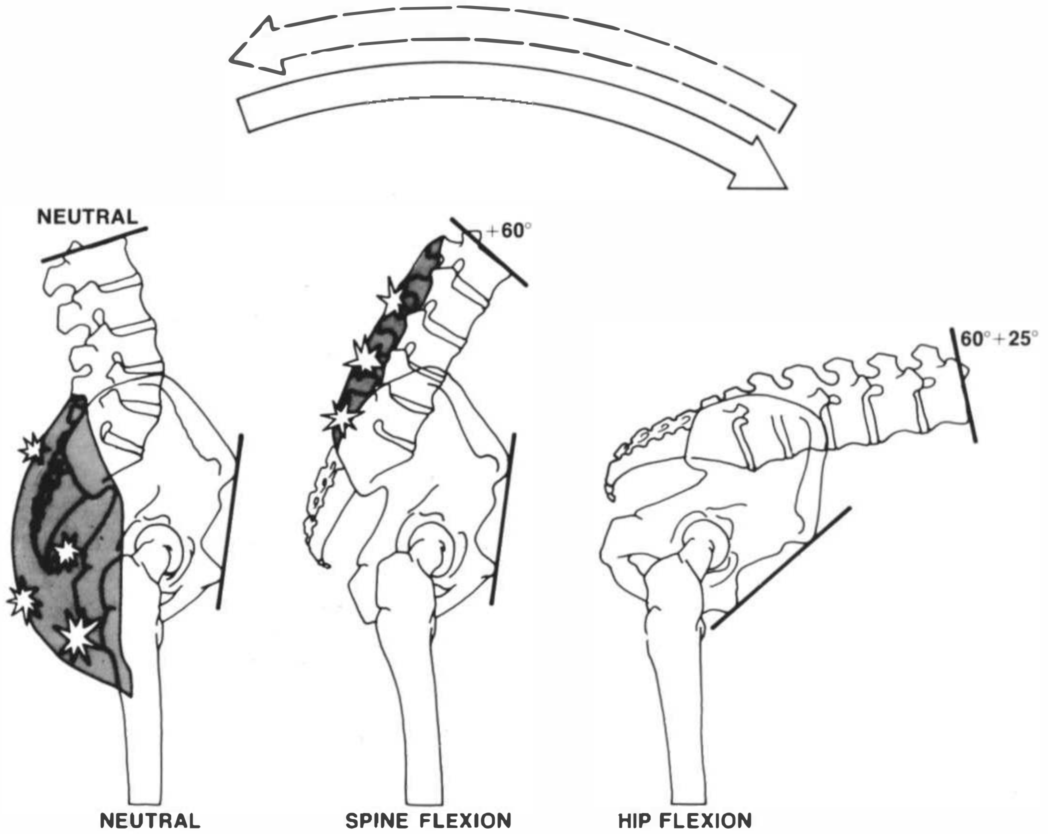

FIGURE 1-40 Muscle activity in forward bending. Bending forward is a two-part movement involving both the spine and the pelvis. In the first 60° of movement, the pelvis is locked by the gluteal muscles, while the lumbar spine is gradually flexed, with the accompanying increasing activity of the erector spinae and superficial muscles of the back. In the second phase there is additional motion of about 25°, which is obtained by relaxation of the pelvis with respect to the femurs. In the fully flexed position, all the muscles are relaxed, and the weight of the trunk is borne by the ligaments and passive extension of the muscles. Extension from the fully flexed position to the neutral position is achieved in the reverse order; the pelvis extension is followed by extension of the lumbar spine.

increasing activity while bending.[72] Extension of the trunk against load increased the activity of the back muscles of the lumbar region.[198]

Lateral Bending

With lateral bending, the activity of the back muscles increased on both sides of the spine, but mostly on the ipsilateral side.[158] Thus, the trunk is bent over to one side by the imbalance of forces. However, if the spine carrying a load is bent laterally, relatively higher activity is registered in both the contralateral side of the lumbar region and the ipsilateral side of the thoracic region.[13]

Axial Rotation

During axial rotation of the spine, erector spinae muscles on the ipsilateral side and musculi rotatores and multifidi on the contralateral side were found to be active.[158] However, Donish and Basmajian found the activity in the thoracic muscles of the back to be symmetrical, while that in the lumbar muscles was present only on the contralateral side.[55]

The abdominal muscles showed only a slight activity. But strong activity was noticed in the gluteus medius and the tensor fasciae latae muscles.[44]

These findings are somewhat in conflict with those of a recent study.[205] With the use of surface electrodes, the bilateral muscle activities of erector spinae, rectus abdominis, obliquus externus, and obliquus internus were investigated while the subjects exerted axial torques. There was significant antagonistic activity together with the activities that would be expected to produce the axial torque. For example, when left axial torque was produced (left shoulder translating posteriorly), the right internal oblique and left external oblique were more active than their corresponding partners, but there was considerable antagonistic activity as well as activity in muscles that do not produce axial torque (e.g., erector spinae). In fact, the erector spinae and obliquus externus showed the largest differences between the left and right sides. These two muscles also carried the largest tension. Authors suggest that the muscles, besides producing the torque, also act as postural stabilizers. This interesting idea is supported by anatomic studies of the multifidus muscle.[140]

SPINAL CORD

Although protection of the spinal cord is crucial to survival, little is known about the physical properties and the functional biomechanics of this vital structure. The delicate spinal cord is enclosed within the relatively hard spinal canal, made of rigid vertebrae connected end-to-end in space. The spinal canal changes in length as a result of physiologic flexion, extension, and lateral bending. Its effective cross-sectional area also undergoes changes with physiologic flexion and extension as a result of soft-tissue bulge, axial rotation, and horizontal displacement. The spinal cord itself is supported and protected by surrounding soft-tissue structures: pia matter, dentate ligaments, the subarachnoid and subdural space filled with spinal fluid, and dura mater.

Biomechanically Relevant Anatomy

Three membranes cover the spinal cord. They are the dura mater, the pia mater, and the arachnoid (Fig. 1-41).

The dura mater is a long, cylindrically shaped sac of dense connective tissue that encloses the cord. It is separated from the periosteum lining the vertebral canal by epidural space containing fat and venous network. The dura also envelops the spinal roots, ganglia, and nerve as they pass through the intervertebral foramina.

The arachnoid is a very delicate cobweblike membrane consisting of fine, elastic, fibrous tissue. It follows the contours of the dura mater. It is separated from the dura by subdural space (moistened by fluid) and from the pia mater by subarachnoid space (filled with cerebrospinal fluid). The arachnoid is attached to the dura by threadlike subdural trabeculae. Strands of arachnoid traverse the subarachnoid space to become attached to the pia mater.[48]

The pia mater is a vascular membrane covering the cord. Its inner layer is composed of a closely fitted network of fine elastic fibers. Its outer layer is formed by a loose meshwork of collagenous fiber bundles continuous with the arachnoid trabeculae.

In the cervical and thoracic region, the pia mater thickens between the anterior and posterior roots and on each side, forming the dentate ligaments. These toothlike processes traverse the subarachnoid and subdural spaces to become fixed to the inner side of the dura. There are 20 dentate ligaments, the last being at the level of T12–L1.

A new technique has been introduced to study the anatomy of the spinal cord and other nervous structures. Fresh cadavers were deep-frozen in defined spinal postures. The spine was cut in the transverse plane and photographed at 0.1–1.0-mm intervals using a special cryomicrotome. The sequential photographs provided an excellent record of the spatial relationships between the spinal cord and the surrounding structures.[210] The technique has the potential, in combination with computer graphics, to demonstrate three-dimensionally the dynamic anatomic relationships between various spinal components (e.g., bone, ligaments, blood vessels, and the nervous tissues). This may be done for the normal, degenerated, and traumatized spines.

Physical Properties

Elastic Properties

The spinal cord (cord and pia mater) is a structure with special biomechanical characteristics. When removed of circumferential attachments, nerves and dentate ligaments, and suspended from its upper end in the vertical position, it lengthens as a result of its own weight by more than 10%.[34] This very flexible behavior changes suddenly into stiff resistance when an attempt is made to produce any further

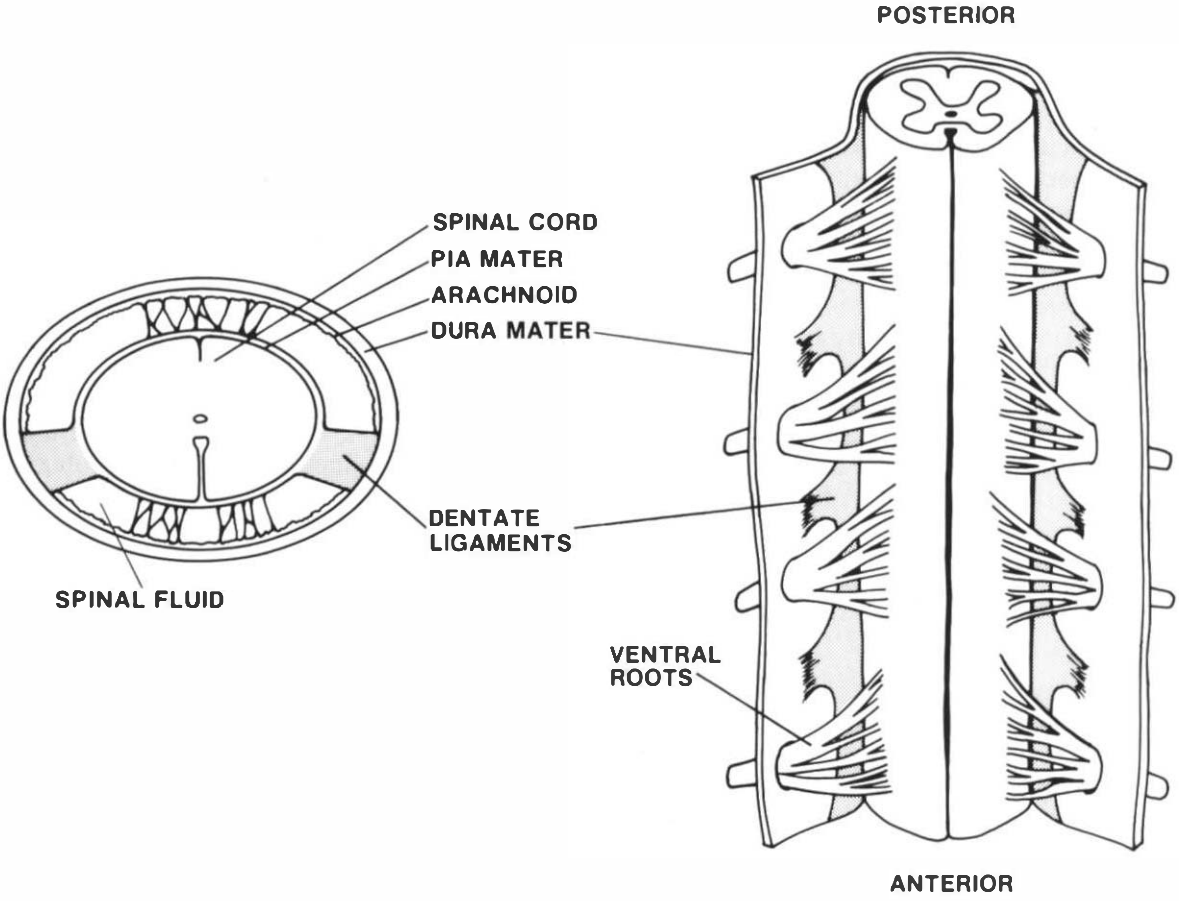

FIGURE 1-41 Anatomy of the spinal cord and the surrounding structures.

deformation. In other words, the load–displacement curve of the spinal cord has two distinct phases: an initial phase in which large displacement is obtained with very small forces, and a second phase in which relatively large forces are required to produce relatively small deformations. There is an abrupt change from one phase to the other. The forces in the initial phase measured less than 0.01 N (0.04 oz), while in the second phase the spinal cord supported 20–30 N (4.5–6.7 lbf) before rupture. This behavior is qualitatively analogous to the behavior of a ligament (Fig. 1-16).

Axial compression of the cord, however, did not show such abrupt change. When compression was applied to a spinal cord specimen, there was large initial deformation (with very small forces), followed by increasing elastic resistance until the specimen buckled. The spinal cord without the pia mater was found to behave like a semifluid cohesive mass.

The large deformations with very small forces are probably due to the design of the spinal cord structure. The extreme mobility of the cord in the initial phase is achieved by folding and unfolding of the cord, much like an accordion. Beyond these limits of unfolding, the tissue is subjected to direct tensile forces. Thus, the second phase of the load–displacement curve truly represents the tissue properties of the spinal cord *material*, while the initial phase of extreme flexibility represents its accordionlike *structural design*.

The variation in length of the cord was accompanied by a change in its cross-sectional area, which increased under compression and decreased on extension. This is due to the incompressibility of the cord tissue. The cross-sectional area *in situ* was observed to change from a rounder to a more oval shape when the spine was bent from full flexion to full extension.

Viscoelastic Behavior

All biological tissues (from ligaments to bone) are viscoelastic (*i.e.*, their mechanical properties are time-dependent). The spinal cord, containing significant amounts of fluid, shows very prominent viscoelastic behavior. Hung and co-workers have studied the stress–strain–time relationships *in vivo* using feline[105] and canine[104] models. Applying axial tension to the cord, they found the cord to exhibit a nonlinear stress–strain relationship beyond 4–5% strain and large hysteresis (loss of energy during a load and unload cycle), which varied with the amount of maximum deformation. They found an initial tensile stress in the spinal cord of 2–3 kPa and an average Young's modulus of 0.27 MPa.

Functional Biomechanics

Motion and Deformation

The cervical, thoracic, and lumbar spinal canals are lengthened during flexion and shortened during extension.[34] The mechanism for these observations is shown in Figure 1-42A, in which the instantaneous axes of rotation (IAR) refer to the thoracic spine motion segments.[1,194] In flexion, the length of the canal as measured by its center line is increased in comparison to that of the neutral position. The anterior border of the canal also increases, but to a lesser extent. The maximum increase, however, is that for the posterior border of the canal. In extension, the canal is shortened as measured by the decrease in its anterior border, center line, and posterior border lengths. The maximum decrease is on the concave side of the curve, on the posterior border.

The changes in length of the bony canal are always followed by similar changes in the spinal cord. The mechanism of folding and unfolding is responsible for an estimated 70–75% of the entire length change from full extension to full flexion (Fig. 1-42B). The rest of the changes at the extremes of physiologic motions are due to the elastic deformation of the spinal cord tissue.[34]

The spinal cord folds like an accordion during extension. The folds are more distinct on its posterior surface, the place of maximum decrease in length, than on its anterior surface. Clinically, these folds are visualized on contrast radiographs as a series of protuberances. Yellow ligament encroachment may also contribute to these folds in older people, because this ligament becomes less elastic with age.

The spinal cord is suspended within the dura by the dentate ligaments, and the nerve roots may also provide some support. During full flexion, the spinal cord, its nerve roots, and the dentate ligaments are under physiologic tension. Because the dentate ligaments are inclined inferiorly, the tensile force in the ligaments has two components with respect to the axis of the spinal cord (Fig. 1-43). The axial component balances the tension in the cord, probably reducing its magnitude. On the other hand, the transverse components balance each other in pairs to position the cord near the center of the canal and anchor it there. The central position of the spinal cord is advantageous because it provides maximum protection from bony impingement or shock during trauma. (In the design of football and military helmets, a similar principle is used. The head is protected against trauma by suspending the helmet from the headband by several radially directed straps.)

There are two other substances that may offer mechanical protection to the spinal cord, namely, the epidural fat and the spinal fluid. Very likely, these aid in reducing friction and in absorbing the energy from physiologic and other forces. The biomechanical and pathophysiologic factors related to spinal cord trauma are reviewed in Chapter 4.

Internal Stresses

Abnormal internal stresses may be generated in the spinal cord in several pathologic conditions (e.g., hypertrophy of the ligamentum flavum, disc bulge, and an osteophyte). A qualitative theoretical analysis of an impingement of the spinal cord provides us with some understanding of the clinical consequences.[191] Figure 1-44A shows the resulting loads due to an anterior impingement: two forces (compression and tension) and a moment (bending) acting on the spinal cord. Let's confine our analysis to the transverse cross-section of the cord at the point of impingement. We will analyze each of the three loads separately and then add up the effects. The compression force results in compression stresses that decrease in magnitude away from the point of impingement. Additionally, there are shear stresses that have maximum values in the middle of the cord. These stress variations are shown in Figure 1-44B and C. Note that in these diagrams the stresses are represented by isostress lines. For the second load

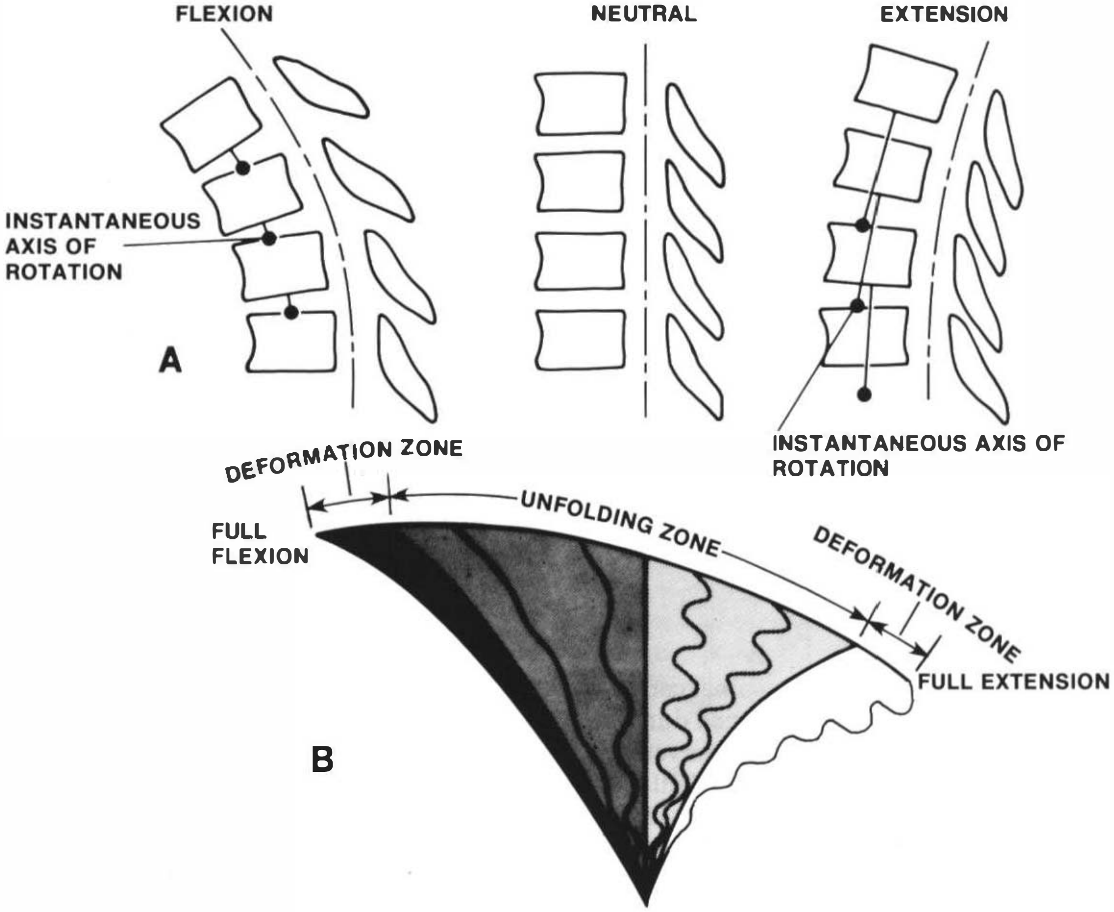

FIGURE 1-42 Spinal canal and cord in flexion/extension. **(A)** The spinal canal is formed by a series of spaces of the neural arch. In flexion, from the neutral position, the length of the spinal canal increases. This is due to the location of the instantaneous axis of rotation, which is anterior to the canal. The greatest increase is at the posterior border of the canal. In extension, the canal length decreases, again for the same reason. The greatest decrease is at the posterior border of the canal. **(B)** The spinal cord is required to follow the changes in length of the spinal canal during physiologic motions. This it does through two mechanisms, unfolding/folding and elastic deformation. In the neutral position, the cord is folded like an accordion and has slight tension. During flexion, the spinal cord first unfolds, with a minimum of increase in its tension, followed by some elastic deformation near full flexion of the spine. During extension, the spinal cord first folds, with a minimum of decrease in the tension, followed by some elastic compression. (*B is based upon experimental findings of Breig, A.: Biomechanics of the Central Nervous System: Some Basic Normal and Pathological Phenomena. Stockholm, Almqvist and Wiksell, 1960.*)

(*i.e.*, the tensile force), the resulting stresses are all tensile and are uniformly distributed across the cross-section (Fig. 1-44D). Finally, the bending moment results in tensile stresses on the convex and compressive stresses on the concave side of the bent spinal cord (Fig. 1-44E). The magnitude of these stresses is highest on the surface and decreases to zero value in the middle (neutral axis) of the spinal cord.

Combining the partial results described above, we obtain the final internal stresses present on the transverse section of the spinal cord at the level of

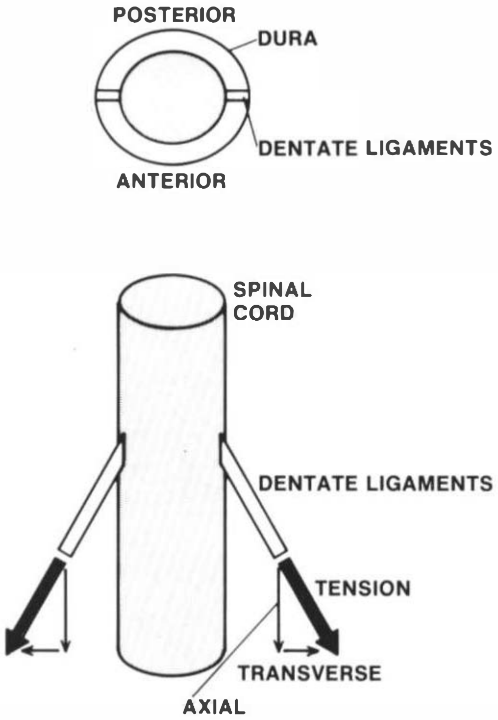

FIGURE 1-43 Role of the dentate ligaments. Besides the support provided by the three meninges (pia mater, arachnoid, and dura mater) and the two fluid-filled spaces (subarachnoid and subdural), the spinal cord is stabilized in its central position within the dura by several pairs of dentate ligaments under tension. Because of inclination, the tension in these ligaments may be divided into axial and transverse components. The dentate ligaments come in pairs; therefore, the axial components are added, and together they balance the axial tension of the cord. The transverse forces of the two dentate ligaments balance each other and provide stability and protection.

impingement. Figure 1-44F depicts the normal (along the length of the spinal cord) stresses: high compressive stresses anteriorly and less severe tensile stresses in the posterior part of the cord. The shear (in the transverse plane of the cord) stresses (Fig. 1-44G) are the same as described earlier. The total stress at any point in the transverse section is the sum of the normal and shear stresses. The failure of the tissue will result when the total stress at a point exceeds the tissue strength at that point. Taking into consideration the qualitative nature of this analysis of a highly complex clinical situation, the results help explain some of the clinical consequences of trauma[212] and anterior compression.[109]

NERVE ROOTS

Biomechanically Relevant Anatomy

Inside the dural sheath, the dorsal and ventral nerve roots approach the intervertebral foramen. The dorsal root continues into the dorsal root ganglion, which is usually located in the central part of the foramen. Then the roots join to form the spinal nerve, which continues into the peripheral nerve.[220]

The nerve roots, spinal nerves, and peripheral nerves constitute the various parts of long cellular extensions from nerve cell bodies located in the anterior horn of the spinal cord and in the dorsal root ganglion[220] (Fig. 1-45). The length of the nerve roots, from the spinal cord to the intervertebral foramen, varies from 60 mm (L1) to 170 mm (S1). The average cross-sectional areas of the nerve roots, dorsal as well as ventral, are about 1.2 mm square.[245] The nerve cell axons are very thin and long—10 μm in diameter and up to 1 m in length, a diameter-to-length ratio of 1 : 100,000.

Intraneural microcirculation is essential for nerve function. As we will see, compression or stretch of the nerve may interfere with this circulation and, hence, the nerve function. For details of this important anatomy, one may consult specific research articles and textbooks.[50, 54, 196]

The peripheral nerves, which have epineural connective tissue surrounding the nerve as a protective layer, are different from the spinal nerves, which lack this tissue but instead are surrounded by cerebrospinal fluid contained within the dura and arachnoid membrane.

The passage of the spinal nerves from the spinal canal to the outside of the intervertebral foramen is complex, as shown by the sectional photographs of Rauschning.[211] Compression of the nerve roots depends upon the effective space available within the intervertebral foramen, accounting for the soft tissue within the canal and the decreased size of the canal due to degeneration as well as physiologic movements by vertebrae.

An excellent study of the anatomy and pathology of the nerve roots has been provided by Kirkaldy-Willis.[119] The spinal nerves may be entrapped at several sites (e.g., posterior to the disc, laterally in the canal, and posteriorly in the facet joints). The spectrum of degeneration in these joints may lead to varying degrees and types of clinical problems. De-

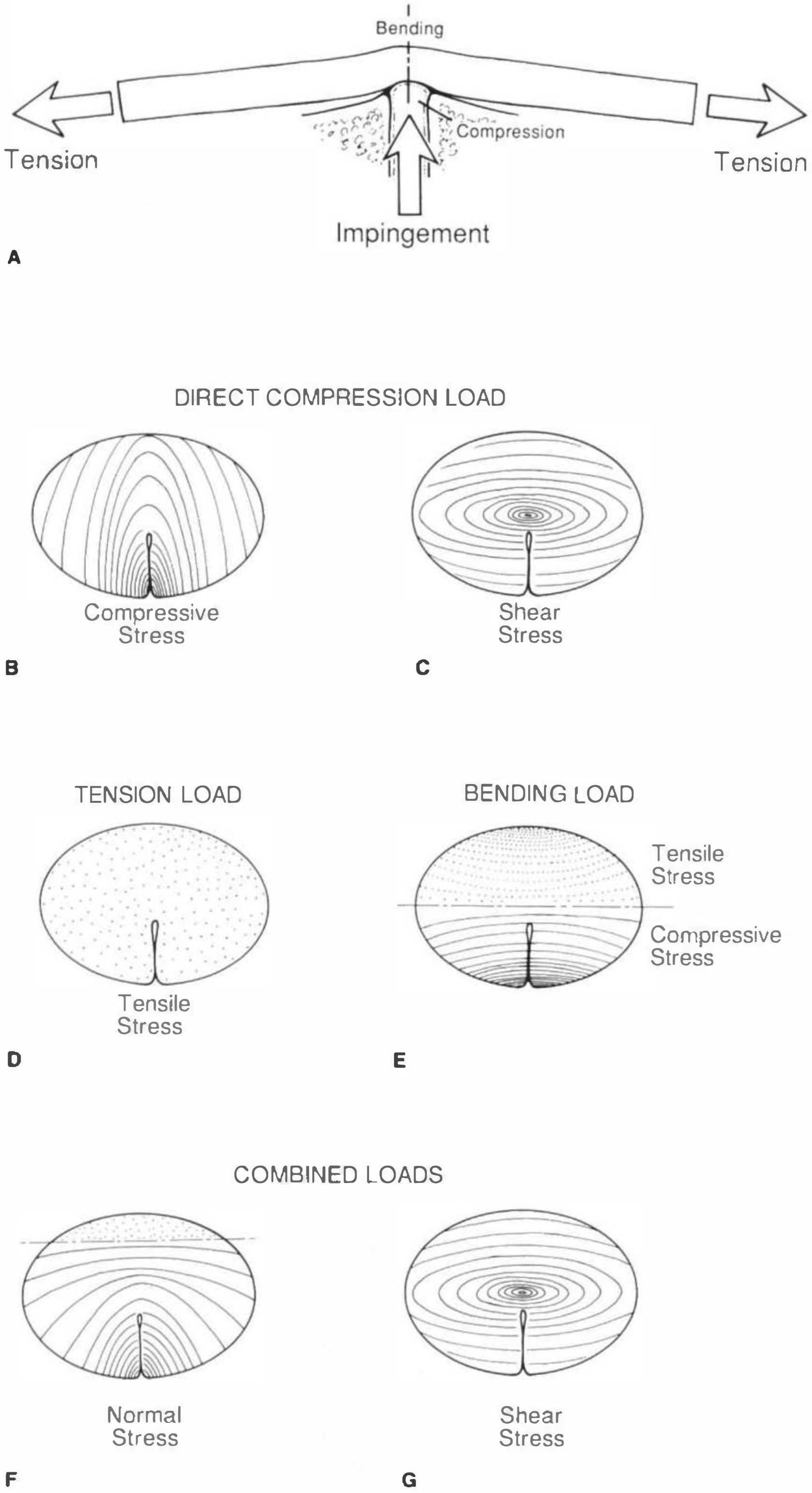

FIGURE 1-44 Qualitative stresses in the spinal cord due to anterior impingement. **(A)** Loads on the spinal cord: Compression, tension, and bending. **(B)** Compression stresses decrease away from the point of impingement. **(C)** Shear stresses are maximum at the center. **(D)** Tensile stresses are uniformly distributed over the cross-section. **(E)** Compression and tensile stresses due to bending—zero at the center and increasing toward the anterior (compression) and posterior (tensile). **(F)** Total normal stresses, equal to sum of stresses shown in **B**, **D**, and **E**. **(G)** Total shear stresses, same as those in C. *(From Panjabi, M. M., and White, III, A. A.: Biomechanics of non-acute cervical spinal cord trauma. Spine,* 13:838, *1988.)*

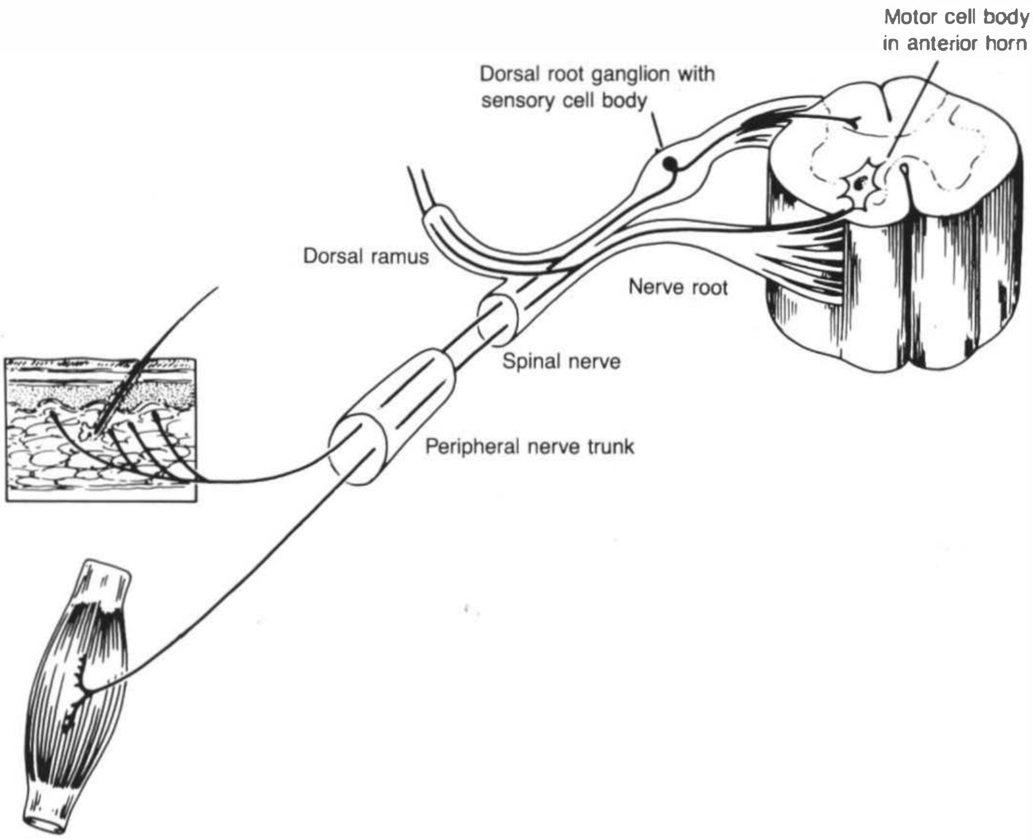

FIGURE 1-45 Anatomy of nerve roots, spinal nerves, and root ganglion. *(Adopted from Rydevik, B., et al.: Pathoanatomy and pathophysiology of nerve root compression. Spine, 9:7, 1984.)*

tailed anatomy of the nerve root canal of the lumbar spine has also been described.[33]

Physical Properties

Although there have been several studies on the physical characteristics of peripheral nerves, the mechanical properties of human spinal nerves have not been studied to the same extent.[245] In a recent study using fresh cadaveric specimens loaded in tension at a slow loading rate, the stress–strain curves were found to be nonlinear with an average elastic modulus of 6 MPa. At the time of failure, the maximum stress and strain reached were 1.6 MPa and 18%, respectively. The failure loads, but not the failure stress, were significantly higher for the foramenal part of the nerve root compared with the intrathecal part.

Functional Biomechanics

The spinal nerves may be subjected to compression in spinal stenosis and combined compression and tension in the case of disc herniation. The latter is based upon a hypothesis by Breig, which states that physiologic sliding movements of the nerve within the intervertebral foramen may be interfered with by disc herniation, resulting in chronic tension in the spinal nerve segment between the foramen and the site of disc herniation.[35]

The effects of compression on the peripheral nerve have been studied by Rydevik and coworkers.[221-223] When external pressure applied to the peripheral nerve exceeds 30–50 mm Hg (1 mm Hg = 133 Pa), there are alterations in blood flow, vascular permeability, and axonal transport. If these pressures are maintained for a longer period of time, there will be deterioration of nerve function.[223]

In an elegant *in vivo* model using pig cauda equina, Pedowitz and colleagues studied motor and sensory nerve root conductions as functions of compression pressure (up to 200 mm Hg) and compression duration (up to 4 hours), followed by recovery.[200,201] Compression of 50 mm Hg produced no significant changes during the entire observation period. In the 100 mm Hg group, after two hours of compression, the efferent motor conduction across the compressed segment of the nerve was reduced to 43% of the original signal, but the conduction was fully restored 40 minutes after the unloading. On the other hand, afferent sensory conduction was impaired to a much greater degree (down to 26% of the original) and did not fully recover. The same compression applied for 3 hours (in another group of animals) resulted in an 80% decrease in the conduction amplitudes. Finally, the compression pressure of 200 mm Hg resulted in complete block. On removing the pressure after 4 hours, there was minimal recovery, but only of the efferent motor conduction. These findings nicely corroborate the clinical observations of sensory deficit or pain in the absence of motor functional loss.

The physiologic effects of stretching of the nerves are also dependent upon the magnitude as well as the duration of the deformation. With stretching of the nerve, the cross-sectional area decreases, causing an increase in intrafascicular pressure, which interferes with the intraneural blood flow.[138] The relationship between the stretch and the decrease in blood flow was found not to be uniform. In the rabbit tibial nerve, 8% of the tensile deformation was sufficient to start the decrease in the blood flow, while 15% of stretch completely shut off the blood supply to the peripheral nerve.

Edema formation in the nerve roots has been studied in greater detail by Olmarker and colleagues, using pig cauda equina.[172a] The time-pressure thresholds for the occurrence of edema were two minutes at both 50 mm Hg and 200 mm Hg for a rapid rate of pressure increase (0.05–0.1 sec) and two hours at 50 mm Hg and two minutes at 200 mg Hg for a slow rate of pressure increase (up to 20 sec). In another study using a similar model, the fluid pressure inside the dorsal root ganglion was measured as function of mechanical compression, using a four-thousandth of a millimeter diameter micropipettes.[222a] The normal value of 2.7 ± 0.2 mm Hg increased to 7.1 ± 1.2 mm Hg because of mechanical compression. Unfortunately, the researchers did not document the magnitude of the compression force or pressure applied, its precise location and its rate of application.

■ CLINICAL BIOMECHANICS

The Intervertebral Disc

■ When a vertebra–disc–vertebra unit is subjected to a compressive load, it generally but not always fails by fracture of the end-plate or of the vertebral body, with no apparent damage to the disc.

■ Compression loading of the disc cannot produce disc herniation. However, the disc bulges in the horizontal plane under such loading, with no propensity for posterolateral bulging.

■ During bending, the disc bulges on the concave side of the curve and collapses on the convex side. Thus, in flexion, the disc protrudes anteriorly and is depressed posteriorly. The converse holds for extension.

■ Maximum posterolateral disc bulge occurs as a result of lateral bending to the same side and axial compression.

■ In flexion, the nucleus moves posteriorly, while the annulus, anterior and posterior parts, move anteriorly.

■ The bending and torsional loads are probably more dangerous to the disc than is axial compression.

■ The mechanism of disc prolapse, based on experiments, consists of a sudden application of compressive force with the spine in flexion and lateral bent posture.

■ The intervertebral disc is viscoelastic and exhibits creep and relaxation behavior. These phenomena may be advantageously used in traction and with spinal instrumentation.

■ There seems to be a correlation between the degree of disc degeneration and its creep characteristic. A degenerated disc exhibits less creep and thus has less capability to attenuate shocks and vibrations from the ground.

■ *In vivo*, the loads on the disc are relatively very high. In a standing posture the compressive load is about two times the whole body weight.

■ An injury to the disc annulus or removal of the nucleus substantially alters the biomechanical behavior of the disc *in vitro* under bending moments and torques.

- Under compression loading, the disc is subjected to relatively much higher stresses when it is degenerated.

Spinal Ligaments

- In a physiologic range of motion of the spine, the vertebrae easily move with respect to each other (the spine is relatively flexible), and there is the least amount of energy expenditure. Furthermore, beyond the physiologic range the ligaments provide substantial protection by resisting forces and absorbing large amounts of energy before failure.
- In flexion, all ligaments, except the anterior longitudinal ligament, are stretched. In extension, the opposite is true. In lateral bending, the ligamentum flavum and the transverse ligaments are stretched. In axial rotation, one of the capsular ligaments and the supraspinous ligament are stretched.
- A bone–ligament–bone system subjected to tensile loads fails through either the bone or the ligament. At slow rates of loading, the failure is more often through the bone, and at high rates of loading it is the ligament that fails.

The Vertebra

- The compressive strength of vertebrae increases from C1 to L5.
- There is a sharp decrease in the vertebral strength with age beyond 40 years. However, the decrease is more gradual after 60 years.
- In osteoporosis, the loss of horizontal trabeculae effectively lengthens and therefore weakens the vertical trabecular beams and compromises the strength of the verebral body.
- New studies have found that the cortical shell of the vertebral body contributes an average of 10% of the compressive strength of the vertebral body.
- Half of the cancellous bone samples tested in an experiment were found to be stronger after the first fracture. Thus, a vertebra with a compression fracture may actually be able to carry equal or higher loads after a fracture.
- Central fractures of the end-plates are more often associated with the nondegenerated discs. The opposite is true for peripheral fractures, which were found to be related to degenerated discs.
- The facet joints may carry large compressive loads (up to 33%), depending upon the body posture. They also provide 45% of the torsional strength of a functional spinal unit.

Functional and Multisegmental Spinal Units

- The highest torsional stiffness is typically exhibited at the thoracolumbar junction. This makes the T12–L1 FSU the site of high stress concentration. The clinical observation of a higher incidence of spine fractures at this level may be related to this factor.
- There is no consistent correlation between the elastic properties of an FSU and its disc level, disc grade, or disc height, or the sex or age of the patient.
- The lumbar spine exhibits coupling. Axial rotation produces lateral bending to the opposite side, in the upper region of the spine, and to the same side for the L4–L5 and L5–S1 levels. Lateral bending, on the other hand, produces axial rotation to the same side at all lumbar levels. Both these motions also produce flexion rotation. The coupling pattern varies with both the spinal level and spinal posture. There is also some variation among different experimental studies of the same region of the spine.

Mathematical Models

- Mathematical models of the spine are helpful in providing information that cannot be easily obtained otherwise (e.g., stresses in the disc or estimation of forces in the muscles).

The Rib Cage

- The rib cage substantially increases the stiffness of the spine in all physiologic motions. The removal of the sternum completely negates the stiffening effect of the rib cage.
- A scoliotic spine is much more flexible in axial traction than is a normal spine. The additional flexibility is due to the more curved shape of the scoliotic spine.

Spinal Muscles

- Muscles are extremely important in maintaining the erect spine.
- The first 60° of flexion are achieved by locking the pelvis and flexing the lumbar spine. Release of the hip joint provides an additional 25° of flexion.
- Lateral bending of the spine is achieved by the imbalance of the forces exerted by the muscles on the two sides of the spine.
- During axial rotation, the erector spinae on the ipsilateral side and musculi rotatores and multifidus on the contralateral side were found to be the most active.

The Spinal Cord

- The spinal cord is very flexible when subjected to small loads. However, it provides considerable resistance before failure. In its unstretched position it is folded like an accordion, thus providing additional flexibility.
- The spinal canal decreases in length when the spine is extended and increases in length when the spine is flexed.
- The spinal cord follows this pattern easily because of its high flexibility. In flexion, the accordionlike spinal cord unfolds, and in extension it folds. There is a small but perhaps significant component of pistoning of the cord within the canal.
- The spinal cord is protected from traumatic forces because of its three membranes and two fluid-filled spaces. Dentate ligaments provide additional protection and stability to the spinal cord.

Nerve Roots

- Compression of 50 mm Hg up to 4 hours in a porcine model had no effect. Compression of 100 mm Hg reduced the efferent conduction to 43% after 2 hours and to 20% after 3 hours. The afferent conduction was much more severely affected by compression than the efferent motor conduction.

NOTES

[A] The main reason for the large magnitude of forces on a lumbar disc is that the center of gravity of the trunk is in front of the disc, causinga bending moment in the sagittal plane. To balance this, large muscular forces are required on the posterior elements. The reaction to these forces, in turn, is an equally large compressive disc load.

[B] Stiffness as used by Galante is the maximum load applied (0.5 N or 0.1 lbf) divided by the displacement produced in millimeters or inches. The load–displacement curve was found to be highly nonlinear; therefore, this stiffness represents an average value.

[C] The precise line of demarcation between the compressive and tensile zones will depend upon the location of the instantaneous axis of rotation. Shown in Figure 1-10B are an instantaneous axis of rotation and the corresponding distribution of the tensile and compressive stresses in a disc. The length of the vertical lines in the stress diagram represents the magnitude of the stress at a given location. As can be seen, the stresses are maximum at the periphery and decrease toward the line passing through the instantaneous axis of rotation.

[D] So that the important differences between the experiments on the longitudinal ligaments are appreciated, a short discussion of the concepts of material and structural properties is necessary. An intact ligament is a structure. It has shape, size, and a certain distribution of material. Because of these fundamental properties, it performs its structural function of providing mechanical stability to the spine. When tests are performed on *intact* ligaments, the failure load in newtons (poundforce) and the load–deformation curve represent the physical properties of the ligament as a *structure*. On the other hand, when samples of *standardized* size, obtained from a ligament, are tested, the failure load is presented as the breaking *stress* in newtons per square meter (poundforce per square inch) and the load–deformation curves are shown as the *stress–strain* curves. These parameters represent the physical properties of the *material* of the ligament. For more information, see the terms in Chapter 9, Biomechanics A to Z.

[E] In the study by Rockoff and colleagues, strength was defined as a point on the load–deformation curve where the curve departed from the linear behavior. This is not the true, ultimate strength. But it may be expected to correlate reasonably with it.

[F] For a given deformation, the area under the load–deformation curve represents the energy that has been absorbed to produce the deformation. Comparing the Type I curve with the Type III curve for a given deformation beyond the failure load (see Fig. 1-23), it is seen that the Type III curve has the greatest reserve of energy. This energy may be advantageously used either during trauma, where the damage to the adjoining soft tissue is diminished, or during the recovery period, where it may serve as a safeguard against further increase in the deformity.

[G] Rolander and Blair took a motion segment and drilled an axial hole from the top in the center of the upper vertebral body and used a displacement gauge to measure the vertical motion of the lower end-plate of the upper vertebra. Another set of gauges was arranged to measure the motion of the periphery of the same end-plate. The difference between the readings of the two sets of gauges represented the true deflection of the center of the end-plate.

[H] Euler's formula. The strength of a slender column of circular cross-section under compressive load F is given by

$$F = \frac{\pi^2 ED^2}{16\ L^2} \times A$$

where $\pi = 3.14$

E = modulus of elasticity of the column material,
D = diameter of the column
A = cross-sectional area of the column, and
L = free length of the column.

This reduces to

$$F = \frac{\pi E}{4} \times \frac{A^2}{L^2}$$
$$= C \times \frac{A^2}{L^2}$$

where C is a constant for a column of a given material. Thus the strength of a column is directly proportional to the square of its cross-sectional area and inversely proportional to the square of its length.

[I] With reference to Figure 1-30, the computations of preload with the body in anatomic posture are as follows:

Sum of forces $= P - F - W_1 = 0$
Sum of moments $= W_1 \times L_1 - F \times L_2 = 0$

Solving for P, we get

$$P = (L_1 + L_2)W_1/L_2$$

Assuming that $L_1 = 80$ mm, $L_2 = 40$ mm and $W_1 = 0.6$ times body weight, we get

$$P = 1.8 \times \text{body weight}$$

(W_1 is the weight of the body portion above the functional spinal unit; L_1 is the lever arm of this weight with respect to the instantaneous axis of rotation; F is the representative ligament and back muscle tension; L_2 is the lever arm of F; and P is the compressive load on the disc.)

[I]As a simple experiment, take the cylindrical core of a roll of paper towels. Subject it to bending and torsion with your hands and feel the resistance. Remove a longitudinal strip. Repeat the bending and torsion tests, and feel the enormous decrease in the resistance.

[K]Waters and Morris showed from their experiments that the tensile properties of the interspinous ligaments of idiopathic and other scoliotic spines are about the same.[256] Although they did not directly compare these results with those of the ligaments from normal spines, it was implied that the mechanical properties of the ligaments are not affected by the various diseases of scoliosis.

[L]The mechanism of lengthening of the spinal cord for the thoracic region is probably similar to that for other regions of the spine; although no precise information regarding the IAR in flexion–extension for other regions has been published, they are most likely anterior to the spinal cord.

[M]There is another factor that further enhances the stabilizing effect of ligament A. As seen in Figure 1-15B, the motion of the attachment point P may be represented by a translation vector *T*. Its projection T_A and T_B onto the lines of actions of ligaments give us the deformations of ligaments A and B, respectively. As seen in the figure, ligament A has greater deformation than ligament B. Because the resistance offered by a ligament is generally proportional to its deformation, the force F_A will be greater than force F_B, thus further enhancing the effectiveness of ligament A in stabilizing the spine.

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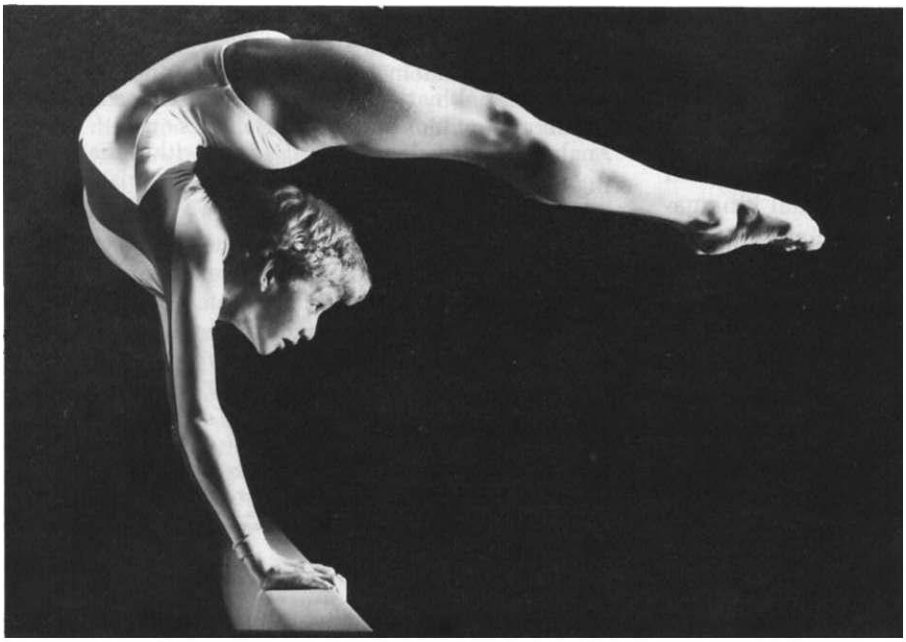
Kinematics of the Spine

Figure 2-1. This Olympic Gold Medal Winner represents the epitome of health. Her spinal column has no pathologic subluxations but represents the maximal range of motion as it is involved in this artistic and athletic form of kinematics. A keen appreciation of normal kinematics is basic to all aspects of the clinical care of the spine. (© 1985, Rocky Theis.)



2

A comprehensive knowledge of spinal kinematics is of paramount importance for the understanding of all aspects of the clinical analysis and management of spine problems. This is true in the evaluation of radiographs, the understanding of clinical stability, spine trauma, scoliosis, the clinical effects of fusions, orthotic prescriptions, and the evaluation of surgical constructs. Much information is available on this complex topic. A selected presentation of the most cogent old and new material on the kinematics of the human spine follows.

TERMS AND DEFINITIONS

Kinematics

Kinematics is that phase of mechanics concerned with the study of motion of rigid bodies, with no consideration of the forces involved.

Coordinate System

The coordinate system employed here is easy and efficient for accurate description of spinal kinematics. Understanding the text is not dependent upon following the conventions used here. However, they are helpful for more precise communications and understanding of the biomechanics literature.

The right-handed orthogonal (90° angle) coordinate system has been recommended for precise orientation about the human body.[87,128] Its orientation in space and its conventions are shown in Figure 2-2.

Occiput—Ocp Versus C0

The designation of occiput has been changed from Ocp to C0, that is, cervical 0. This has been done not only because of its somatic origin but, more important, because of its integral part in the anatomy, biomechanics, and clinical functions of the upper cervical spine.

Regions of the Spine

Based on new information, new hypotheses, and clinical findings, some changes in the subdivisions of the spine are offered. The C0–C1–C2 (occipital-atlanto-axial) complex is grouped and labeled the upper cervical spine. Because of kinematic,[60,67] kinetic,[13] and clinical uniqueness,[116] the cervical region is divided into the middle cervical (C2–C5) and the lower cervical (C5–T1).

A COMPREHENSIVE DESCRIPTION OF KINEMATICS OF THE SPINE*

The important basic and clinical aspects of kinematics of the spine are as follows:

Range of motion for all 6 degrees of freedom
- Rotations
- Translations

Traditional physiologic patterns of motion
- Flexion/extension
- Lateral bending
- Axial rotation

Coupling characteristics and ratios

Instantaneous axes of rotation of motion segments in each of the traditional planes[B]
- Sagittal plane (y,z)
- Frontal plane (y,x)
- Horizontal plane (x,y)

Helical axes of motion located throughout the range of motion[B]

Functions of anatomic elements. A description of the roles played by the various anatomic elements in determining kinematic characteristics

Analysis of cephalocaudal variations within the regions

Analysis and comparison of the regional variations

Upper cervical	C0–C1–C2
Middle cervical	C3–C5
Lower cervical	C5–T1
Upper thoracic	T1–T4
Middle thoracic	T4–T8
Lower thoracic	T8–L1
Lumbar	L1–L5
Lumbosacral	L5–S1
Sacroiliac	S–I

* For a comprehensive review of the subject, the reader is referred to the works of Lysell,[67] Rolander,[102] Werne,[123] and White.[128]

In the thoracic spine, the following division is suggested: upper thoracic (T1–T4), middle thoracic (T4 [illegible] rationale is as follows. The vertebrae are smaller in the upper thoracic, the coupling patterns are similar to those described in the cervical spine,[125] and the kinematics and kinetics[80] are slightly different from those in the middle thoracic and quite different from those of the lower thoracic. The coupling in the middle portion of the thoracic spine is variable but can be distinctly different from the upper. This region is different anatomically in regard to space

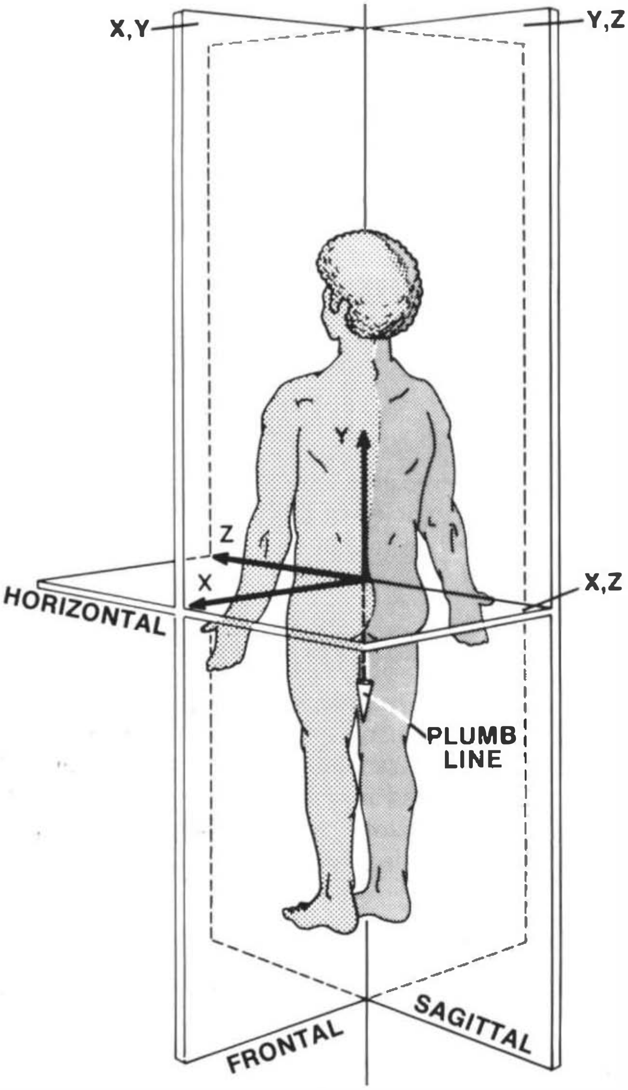


FIGURE 2-2 The suggested central coordinate system with its origin between the cornua of the sacrum is shown. Its orientation is as follows. The −y-axis is described by the plumb line dropped from the origin, and the +x-axis points to the left at a 90-degree angle to the y-axis. The +z-axis points forward at a 90-degree angle to both the y-axis and x-axis. The human body is shown in the anatomic position. There are some basic conventions that are observed which make this a useful system. *The planes* are as shown: The sagittal plane is the y, z plane; the frontal plane is the x, y plane; the horizontal plane is the x, z plane. *Movements* are described in relation to the origin of the coordinate system. The arrows indicate the positive direction of each axis. The origin is the zero point, and the direction opposite to the arrows is negative. Thus, direct forward translation is +z; up is +y; to the left is +x, and to the right is −x; down is −y; and backward is −z. The convention for rotations is determined by imagining oneself at the origin of the coordinate system looking in the positive direction of the axis. Clockwise rotations are $+\theta$ and counterclockwise rotations are $-\theta$. Thus, $+\theta x$ is roughly analogous to flexion; $+\theta z$ is analogous to right lateral bending; $+\theta y$ is axial rotation toward the left. A coordinate system may be set up at any defined point parallel to the master system described above. The location of the coordinate system should be clearly indicated for precise, accurate communications. In spinal kinematics, the motion is usually described in relation to the subjacent vertebra. The secondary coordinate system may be established in the body of the subjacent vertebra. For *in vivo* measurements, the tip of its spinous process may be used. (*Panjabi, M. M., White, A. A., and Brand, R. A.: A note on defining body parts configurations. J. Biomech., 7:385, 1974.*)

available for the thoracic cord and blood supply to the thoracic cord, both of which relate to its vulnerability to trauma. The lower thoracic spine is anatomically different in that the anatomic transition to the lumbar design may occur and the kinematics are distinctly different,[124] and it is the location of a considerable amount of clinical trauma.

The lumbar spine is divided into the lumbar (L1–L5) and the lumbosacral (L5–S1) regions. The anatomy, kinematics, and kinetics of L5–S1 are significantly different from the rest of the lumbar spine.[97]

Finally, the sacroiliac (SI) region, about which a great deal less is known, obviously has a distinctly different articulation and biomechanical structure. The kinematics of this modestly studied joint have been reviewed by White and colleagues.[126]

Functional Spinal Unit (Motion Segment)

The motion segment, or functional spinal unit, is the traditional unit of study in spinal kinematics. It is constituted of two adjacent vertebrae and their intervening soft tissues. Motion is described in terms relative to the subjacent vertebra (see Chapter 9, Biomechanics A to Z). Since a great deal of clinical biomechanics involves kinetics (*i.e.*, the study of forces as well as motions), the term functional spinal unit (FSU) is preferable and is used throughout this book.

Rotation

A body (any piece of matter) is said to be in rotation when movement is such that all particles along some straight line in the body or a hypothetical extension of it have zero velocity relative to a fixed point. Rotation is a spinning or angular displacement of a body about some axis. The axis may be located outside the rotating body or inside it.

Translation

A body is said to be in translation when movement is such that all particles in the body at a given time have the same direction of motion relative to a fixed point.

Neutral Zone

The displacement between the neutral position and the initiation point of spinal resistance to physiological motion. Translatory and rotatory neutral zones are expressed in meters and degrees, respectively. The neutral zone can be expressed for each of the six degrees of freedom (see Fig. 1-14).

Elastic Zone

The displacement between the end of the neutral zone and end of the range of motion. Translatory and rotatory elastic zones are expressed in meters and degrees, respectively. The elastic zone can be expressed for each of the six degrees of freedom (see Fig. 1-14).

Degrees of Freedom

One degree of freedom is motion in which a rigid body may translate back and forth along a straight line or may rotate back and forth about a particular axis. Vertebrae have six degrees of freedom, translation along and rotation about each of three orthogonal axes.

Range of Motion (ROM)

The difference between the two points of physiologic extent of movement is the range of motion. Translation is expressed in meters or inches, and rotation is expressed in degrees. The range of motion can be expressed for each of the six degrees of freedom and is the sum of the neutral and elastic zones.

Pattern of Motion

This is defined by the configuration of a path that the geometric center of the body describes as it moves through its range of motion.

Changes in the normal coupling or the instantaneous axes of rotation are considered abnormal patterns of motion.

Coupling

Coupling refers to motion in which rotation or translation of a body about or along one axis is consistently associated with simultaneous rotation or translation about another axis.

There are certain abnormal patterns of motion described in association with various morphological and clinical situations. Exaggerated coupling patterns have been described in the lumbar spine as a possible sign of instability.[90]

Paradoxical Motion

Paradoxical motion, which may be associated with instability, occurs when there are typical flexion patterns at a functional spinal unit when the overall motion is in extension. Another example is when the converse occurs with typical extension (Fig. 2-3). Paradoxical translation may also occur. We observe in Figure 2-3 that when a posterior ($-z$-axis) translation is expected with extension, there is an anterior ($+z$-axis) translation. Note that with flexion there is an unexpected $-x$-axis rotation of C5. These paradoxical motions are described in the sagittal plane; theoretically, they could occur in the frontal and horizontal planes as well.

Instantaneous Axes of Rotation (IAR)

At every instant for a rigid body in plane motion there is a line in the body or a hypothetical extension of this line that does not move. The instantaneous axis of rotation is this line. Plane motion is fully defined by the position of the instantaneous axis of rotation and the magnitude of the rotation about it (Fig. 2-4).[4]

Rolander,[102] Dimnet,[20] and Gertzbein[39] have

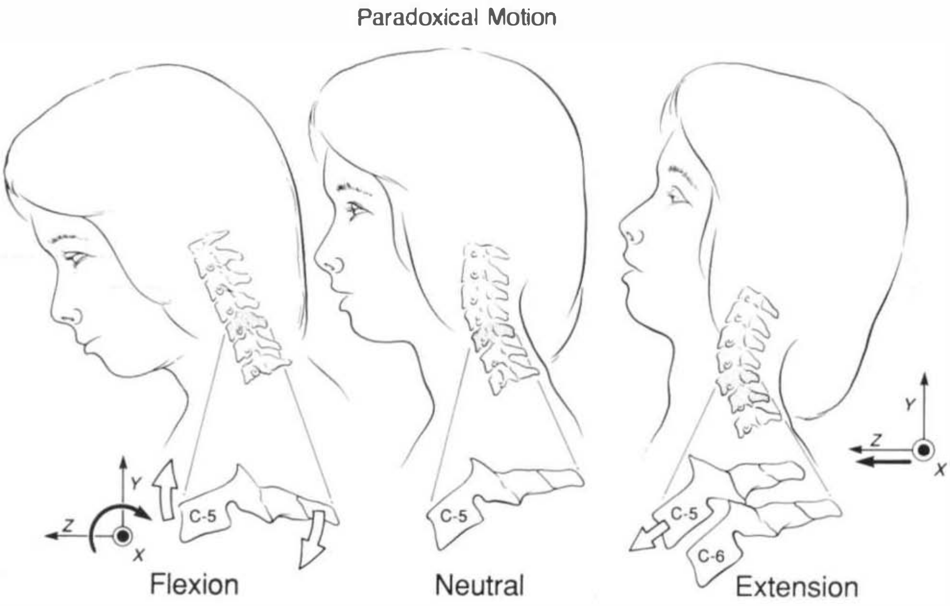


FIGURE 2-3 Paradoxical motion is a theoretical motion, in which a FSU rotates in extension when the overall motion of the segment is in flexion. The figure labelled *Flexion* shows C5 moving as would be expected with extension. The converse may occur with extension. The figure labelled *Extension* demonstrates paradoxical translation. With extension, a small posterior translation of C5 is expected, but instead there is a small anterior translation.

demonstrated abnormal IARs in the lumbar spine associated with disc degeneration and presumed instability.

Helical Axis of Motion (HAM)

The instantaneous motion of a rigid body in three-dimensional space can be analyzed by regarding it as a simple screw motion. The screw motion is a superimposition of rotation and translation about and along the same axis. This axis has the same direction as the resultant of the three rotations about the x, y, and z axes. For a given moving rigid body in space, the location of this axis and the designation of numerical values for rotation and translation constitute a complete, precise, three-dimensional description of the motion (Fig. 2-5). For a more detailed discussion of this concept, consult Chapter 9, Biomechanics A to Z.

Kinematics and Instability

Various writers evaluate instability partially or solely in terms of abnormal kinematics. Abnormal kinematics have generally been discussed in regard to excessive motion. Subsequently, abnormal kinematics have included restricted (too little) motion,[132] abnormal patterns of motion (including abnormal coupling), abnormal distribution of IARs, and paradoxical motion. Problems of clinical instability are discussed in more detail in Chapter 5.

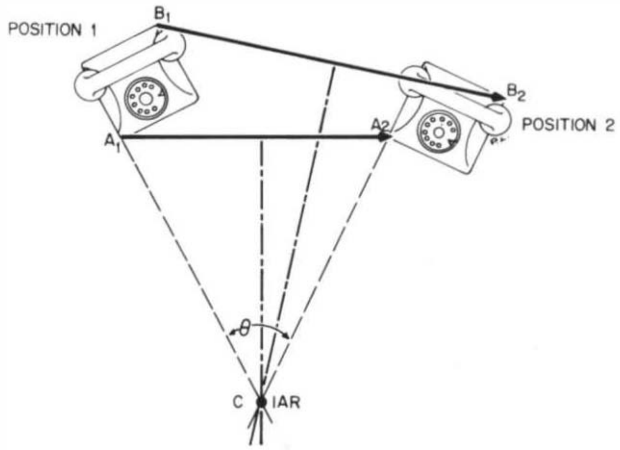


FIGURE 2-4 This figure shows the concept and the actual method of determining the instantaneous axis of rotation in *uniplanar* motion. The IAR is determined by the intersection of the perpendicular bisectors of the two lines A_1A_2 and B_1B_2 (the translation vectors of the two points A and B on the telephone). The angle θ formed at the IAR by points A_1, A_2 or B_1, B_2 is the angle of rotation. *(White III, A. A., and Panjabi, M. M.: Spinal kinematics. The Research Status of Spinal Manipulative Therapy. NINCDS Monograph (No. 15). p. 93. Washington, D.C., U.S. Department of Health, Education and Welfare, 1975).*

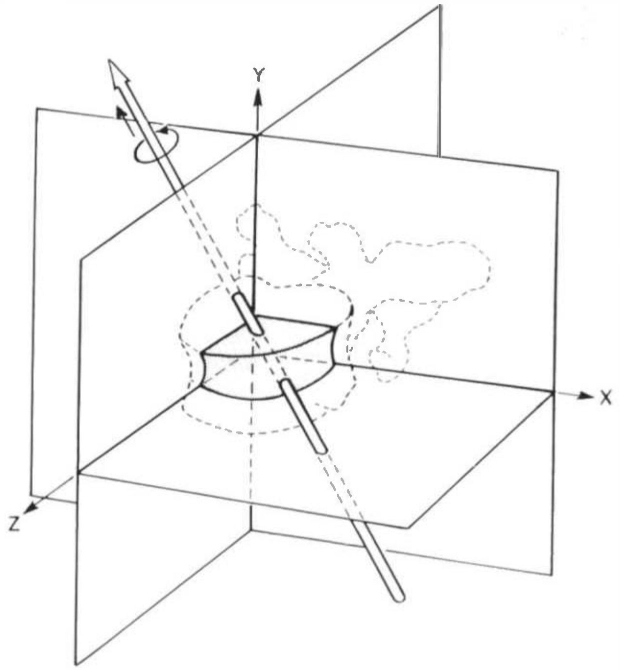


FIGURE 2-5 Helical Axis of Motion. The screw motion that fully describes three-dimensional motion is a superimposition of rotation and translation about and along the same axis. Here a vertebra is shown with a hypothetical helical axis. *(See Chapter 9, Biomechanics A to Z, for a more detailed explanation.)*

Active and Passive *In Vivo* Kinematics

Chapter 1 includes a discussion of neutral and elastic zones within the context of spine kinematics (see p. 21). A relatively new concept has been exemplified in the work of Dvorak[23] and associates. This can be thought of as passive *in vivo* kinetics.

The investigations showed that it is possible to gain additional motion by exerting external forces on the fully flexed or extended neck (Fig. 2-6B). This is important to keep in mind when we evaluate voluntary flexion/extension tests. This also fits with our assertion that the stretch test (passive +y-axis traction) is more likely to show existing pathology than are *active* flexion and extension (Chap. 5, p. 318). The differences in active and passive *in vivo* ranges of motion must be considered in the interpretation of laboratory and clinical studies of normal ranges of motion. The various measures of spinal motion (e.g., *in vitro, in vivo,* active, passive) are further discussed at the end of this chapter.[G]

KINETICS AND MUSCLE ACTIVITY

Kinematics has been defined as that phase of mechanics concerned with the study of movement of rigid bodies, with no consideration of what has caused the motion. *Kinetics* includes the study of the forces responsible for the motion. The muscles are the primary source of force resulting in motion of the vertebrae. Kinetics of the spine are discussed in Chapter 1 and will not be emphasized here.

The muscles that may produce motion of the spine include the anterior muscles (which are in front of the vertebrae), the posterior muscles, and the lateral muscles. The anterior muscles include the abdominal muscles and the iliopsoas. They flex the

VERTEBRAL MUSCLES AND THEIR MOTOR FUNCTIONS

Anterior

Muscles in front flex the spine. If the muscle runs a little obliquely and contracts independently of the corresponding muscle on the opposite side, it rotates and bends the spine laterally, as well as flexes it.

Longus colli*
Longus capitis
Rectus capitis anterior
Rectus capitis lateralis†
Obliquus externus abdominis*
Obliquus internus abdominis*
Psoas major†
Psoas minor†
Iliacus
Quadratus lumborum

Posterior

Muscles in back extend the spine. If the muscle runs a little obliquely and contracts independently of the corresponding muscle on the opposite side, it rotates and bends the spine laterally, as well as extends it.

Superficial stratum

Splenius capitis*†
Splenius cervicis†
Erector spinae (sacrospinalis)
  Iliocostalis*†
  Longissimis*†
  Spinalis*†

Deep stratum

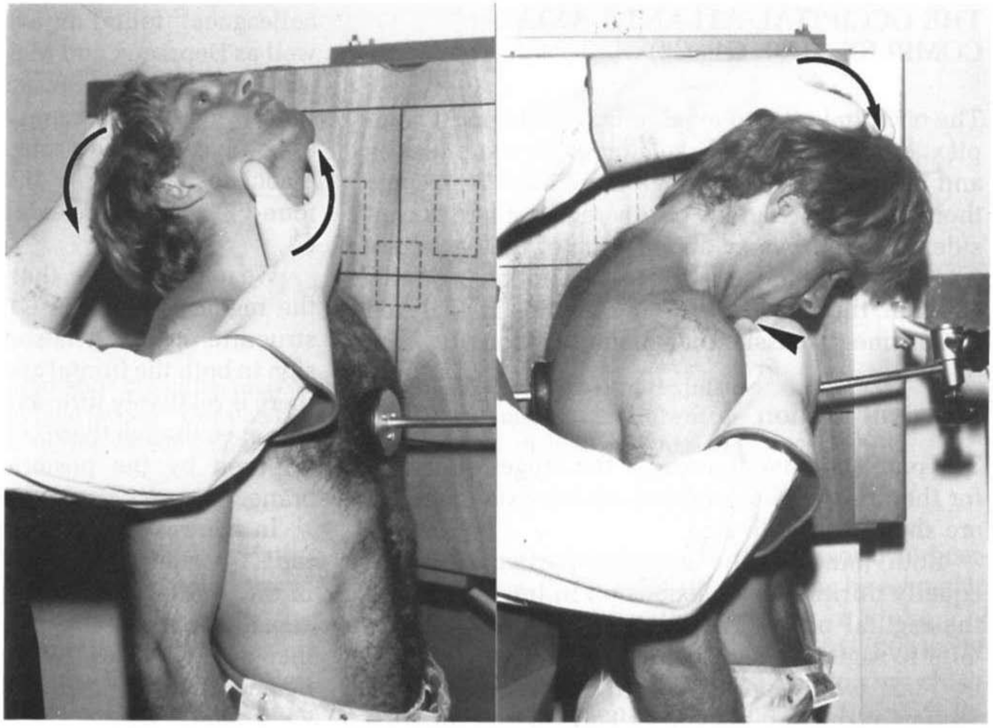
Semispinalis
  Thoracis*
  Cervicis*
  Capitis*
Multifidi*
Rotatores*
Interspinales
Intertransversarii*

Lateral

Muscles on the side bend the spine laterally.

Trapezius
Sternocleidomastoid*
Quadratus lumborum
Scalenus*
  Anterior
  Medial
  Posterior

*Muscles with axial rotation function
†Muscles with lateral bending function



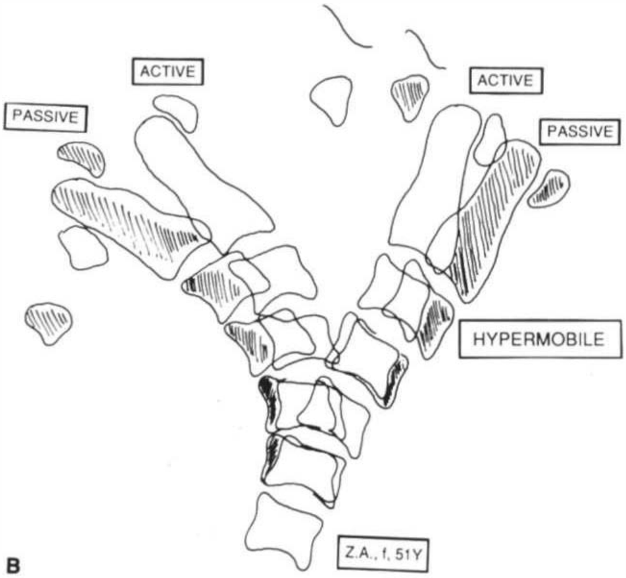


FIGURE 2-6 **(A)** Roentgenograms are taken after the subject fully flexes or extends. This is *active* motion. The neck is further flexed or extended by the examiner, and additional x-rays are taken. This represents the *passive* motion. **(B)** This diagram depicts the active and passive motions of the cervical spine. *(From Dvorak, J., Froehlich, D., Penning, L., Baumgartner, H., and Panjabi, M.: Functional radiographic diagnosis of the cervical spine: flexion/extension. Spine, 13(7):748, 1988.*

spine. If an anterior muscle runs obliquely and contracts independently of the muscle on the opposite side, then it will axially rotate the spine as well as flex it. Similarly, a posterior muscle in the back extends the spine when it contracts. If the muscle runs obliquely and contracts independently of its counterpart on the opposite side, it will axially rotate and bend the spine laterally. If the lateral muscles are on the side and contract, the spine will bend laterally.

THE OCCIPITAL-ATLANTO-AXIAL COMPLEX (C0–C1–C2)

The occipital-atlanto-axial joints are the most complex joints of the axial skeleton, both anatomically and kinematically. Although there have been some thorough investigations of this region, there is considerable controversy about some of the basic biomechanical characteristics. In the following presentation, the best available information is analyzed with some discussion of salient questions.

Range of Motion

The representative figures for the ranges of motion for the units of the occipital-atlanto-axial complex are shown in Table 2-1.

Both joints of the complex participate about equally during flexion/extension in total motion in the sagittal plane. The contribution of the C1–C2 joint to sagittal plane rotation has been questioned by Fick, who reported that there is insignificant motion at this joint.[31] Poirier and Charpy reported 11° of movement.[96] Werne showed that, upon radiographic study, sagittal plane movement is definitely present.[123] An example from his work is shown in Figure 2-7, with an angle of rotation indicated. It was found that the curvature of the dens in the sagittal plane may also allow some additional rotary displacement in that plane.

Some of the curvatures of the dens in the sagittal plane are shown in Figure 4-23 (p. 205), where they are related to a threshold of posterior dislocation of C1 on C2. Subluxations and dislocations are often an exaggeration or overextension of the normal range of motion.

Previously there was thought to be very little or no axial rotation between C0 and C1. However, several investigators have independently observed one side axial rotation in the range of 3–8°. Clark and colleagues[12] found an average of 4.8°; Worth,[132] as well as Depreaux and Mestdagh,[19] reported an average of 3.2°; and Dvorak and colleagues,[27] using computerized axial tomography *in vivo*, noted an average one side axial rotation of 4.3°. Panjabi and associates,[61] using a three-dimensional analysis, found 8° of one side axial rotation between C0 and C1.

We note, however, that the major axial rotation in the region is between C1 and C2.[93] The anatomic structure of C0–C1 is somewhat cuplike in its design in both the frontal and the sagittal planes. Thus, there is relatively little axial (y-axis) rotation. This is true even though there is little ligamentous restraint imposed by the posterior atlantooccipital membrane.

In contrast, however, both articular surfaces of the C1–C2 lateral masses have a convex orientation in the sagittal plane. This geometric design allows considerable mobility. The motion capacity is further enhanced by the absence of any taut yellow ligament connecting the posterior elements. Instead, and contrary to some anatomic diagrams, there is the loose, readily mobile atlanto-axial membrane connecting the posterior elements. The motion here was reported by Werne[123] as 47° to one side. Recent investigators have made similar observations. Dvorak and associates found unilateral C1–C2 axial rotation of 34° in an *in vitro* study[22, 27] and 41.5° in *in vivo* analysis.[24] Representative ranges of Dvorak's *in vitro* and *in vivo* work are presented in Table 2-5. Panjabi and co-workers,[61] employing a three-dimensional *in vitro* methodology, measured 38.9°. Approximately 60% of the axial rotation of the entire cervical spine and occiput is found in the upper region (C0–C1–C2), and 40% is found in the lower region (*i.e.*, below the C0–C1–C2 region).

Results of some of the recent studies described above as well as those provided in the first edition of this book have been summarized in Table 2-5. The

TABLE 2–1 Limits and Representative Values of Ranges of Rotation of the Occipital-Atlanto-Axial Complex

Unit of Complex	Type of Motion	Representative Angle (degrees)
Occipital-atlantal joint (C0–C1)	Combined flexion/extension (±θx)	25
	One side lateral bending (θz)	5
	One side axial rotation (θy)	5
Atlanto-axial joint (C1–C2)	Combined flexion/extension (±θx)	20
	One side lateral bending (θz)	5
	One side axial rotation (θy)	40

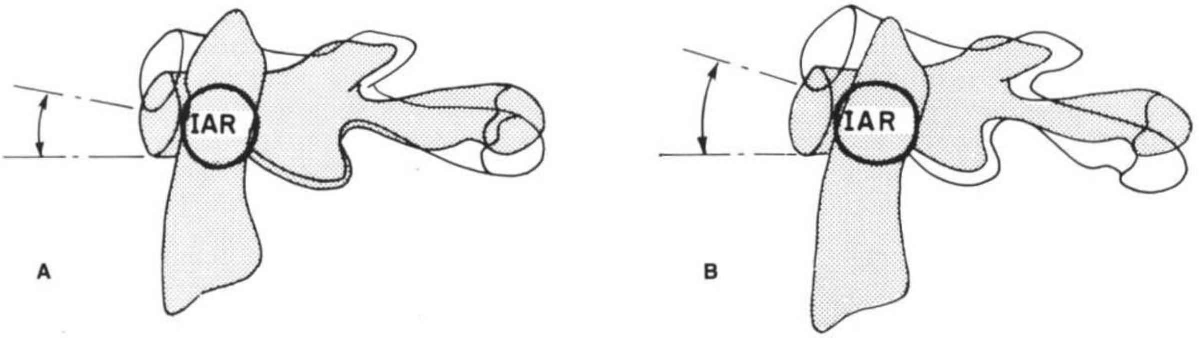


FIGURE 2-7 **(A)** Representation of sagittal plane motion of C1 on C2, with the approximate IAR also indicated. Combined flexion/extension is about 20°. **(B)** The anterior curvature of the dens may permit some degree of additional sagittal plane motion in both rotation and translation.

large rotation that occurs at the C1–C2 articulation can cause clinical problems. Selecki studied the effect of this rotation on the vertebral arteries that ascend vertically in the foramina transversarium and then pass through both the C1–C2 and atlantooccipital region before entering the skull.[105] He found that after 30° of rotation there is kinking of the contralateral vertebral artery. This kinking, which is also accompanied by stretching, first occurs as the vertebral artery exits from the transverse foramina. It becomes more marked as the angle of rotation is increased. At 45° of rotation, the ipsilateral artery also begins to kink (Fig. 2-8).[32] If the flow in both arteries is compromised, symptoms related to decreased flow in the posterior fossa may be elicited.[6] Situations in which this phenomenon may occur include yoga, calisthenics, overhead work, and cervical traction.[103] This last situation may be related to y-axis displacement with stretching or kinking of already compromised vertebral arteries. Similarly, cases of stroke have been reported following chiropractic manipulation of the neck and head.[6, 74] Recently, Schellas and co-workers reported angiographically confirmed vertebrobasilar injuries following chiropractic manipulation.[103] Other authors have reported similar complications in patients without medical problems.[6, 74] Evidently, these accidents may occur in the absence of clinically apparent vascular or cervical spine disease.

According to Miller and Burton, there are usually premonitory symptoms, including nausea, visual disturbance, vomiting, and vertigo, during the pre-

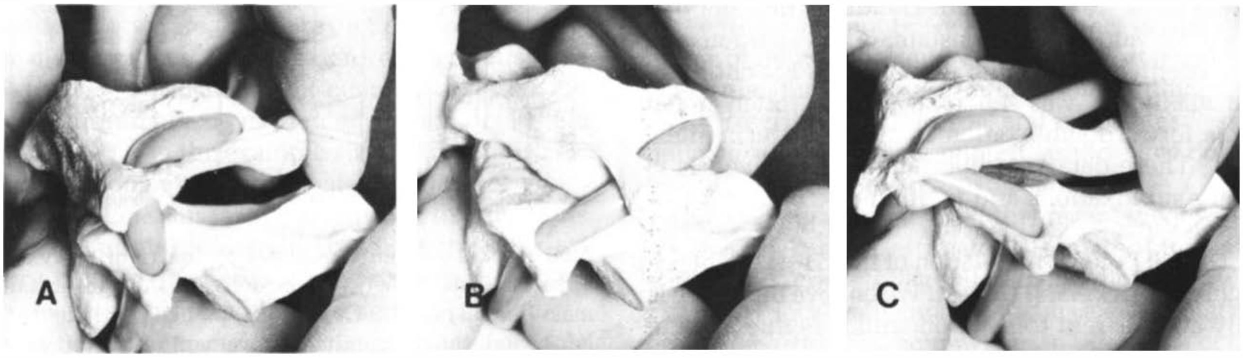


FIGURE 2-8 The vertebral artery is represented by a piece of rubber tubing. **(A)** The atlas and axis are in neutral position. **(B)** The atlas is rotated to the left $(+\theta y)$. **(C)** The atlas is rotated to the right $(-\theta y)$. There is considerable excursion of the vertebral artery when rotary displacement takes place between the vertebrae. *(Fielding, J. W.: Cineroentgenography of the normal cervical spine. J. Bone Joint Surg., 39A:1280, 1957.)*

liminary treatments.[74] If chiropractic treatments are stopped at this stage, further irreversible damage can usually be avoided. In all instances, patients with cervical spondylosis or symptoms of vertebral vascular insufficiency should be warned of the risks of manipulation of the cervical spine. Cervical spine fusion may alleviate this symptom complex; however, further clinical studies are needed to verify this assertion.

An excellent early work on the kinematics of this region was done by Werne.[123] Recently, Dvorak, Panjabi, Clark, and others have significantly improved knowledge of the kinematics of C0–C1–C2. In summary, rotation of the head about the three axes occurs through the occipital-atlanto-axial complex with participation of all three units: the occiput, the atlas, and the axis. These findings are presented in Table 2-1.

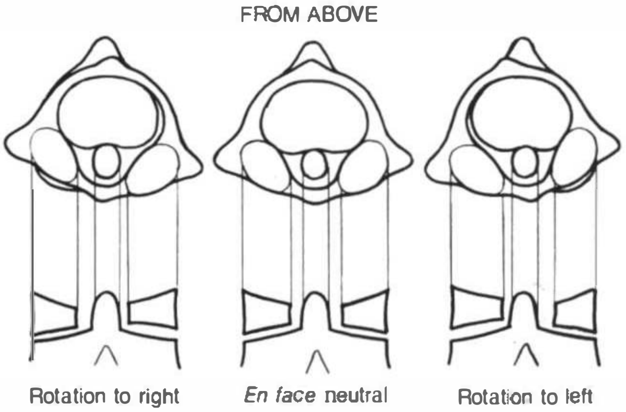
Translatory movements at the occipital-atlanto-axial complex are small. Between the occiput and C1 there is insignificant translation. At the C1–C2 articulation, sagittal plane translations (± z-axis) are minimal because of the snug fit of the ring of C1 about the dens.

During translation in the midsagittal plane, the distance between the anterior portion of the dens and the posterior portion of the ring of C1 is clinically significant. Normal translation is 2–3 mm and is used as a guideline to radiologically evaluate the possibility of transverse ligament inadequacy from either laxity or failure.[54] Jackson carried out radiologic studies of 50 adults and 20 children in which the distance between the posteroinferior margin of the anterior arch of the atlas and the anterior surface of the dens was measured. He found that the distance for adults was constant in full flexion and extension: the maximum was 2.5 mm. For children, the maximum was 4.5 mm. Jackson often noted some forward subluxation in children during flexion.[57] These data are important in the diagnosis of rotary subluxations and fixation of C1 and C2 (see Chapter 5).

Lateral (x-axis) translation of the C1–C2 joint is a highly controversial subject. We believe that there is only an apparent translation, and it is due to axial rotation between C1 and C2. The rotary changes produce a shift in the projection of the lateral masses of C1 in relation to the dens. This has been described by Werne[123] and demonstrated well by Shapiro and colleagues.[106] The rotary displacement pattern and the radiographic projection are shown diagrammatically in Figure 2-9. Although Hohl had a different interpretation of this aspect of C1–C2 kinematics, he too made the point that lateral displacement (we believe *apparent* lateral displacement) of up to 4 mm between the dens and the lateral masses as an isolated radiographic finding is not indicative of subluxation or dislocation.[54] This is nicely confirmed by the lateral (x-axis) translation of point A as depicted in Figure 2-10, which is based on experimental data.

Coupling Characteristics

It is generally accepted that there is a strong coupling pattern at the atlantoaxial joint (Fig. 2-10). The axial (± y-axis) rotation of C1 is associated with vertical (± y-axis) translation. However, there is some disagreement. The problem goes back at least as far as Henke, who in 1863 described a "double threaded screw" joint, due to the biconvexity of the articulations between C1 and C2.[52] This analysis was criticized by Hultkrantz, who studied sagittal sections of the C1–C2 articulations. He found that some of the surfaces were slightly biconvex and others were slightly biconcave.[56] It has been observed that although the actual bony configuration may be concave, the configuration of the cartilage is such that



Rotation to right · *En face* neutral · Rotation to left

FIGURE 2-9 When C1 rotates to the right (− θy), the apparent distance between the dens and the right articular mass (lateral mass) of C1 increases. When C1 rotates to the left (+ θy), the concomitant movement of the left lateral mass results in an apparent increase in the distance between the left lateral mass and the dens as seen on the anteroposterior radiograph. (*Shapiro, R., Youngberg, A. S., and Rothman, S. L. G.: The differential diagnosis of traumatic lesions of the occipito-atlanto-axial segment. Radiol. Clin. North Am., 11:505, 1973.*)

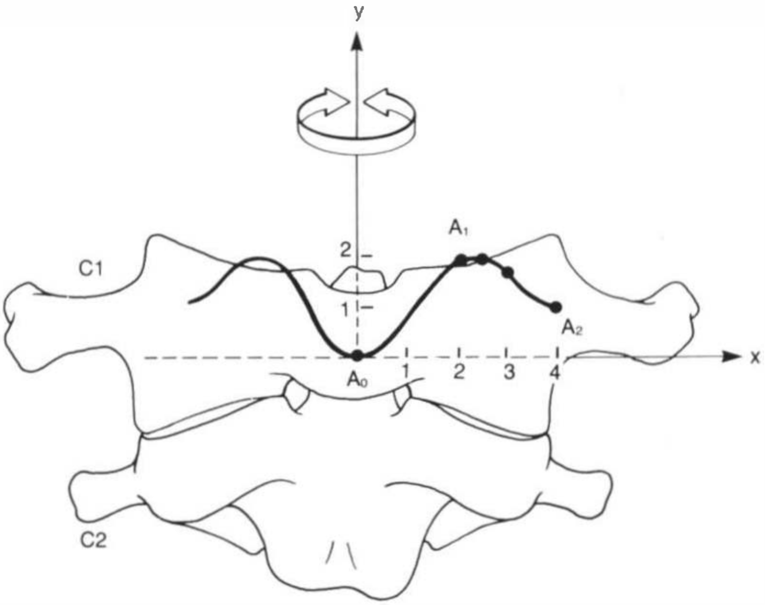


FIGURE 2-10 Translatory movements of the anterior aspect of C1 with respect to C2, when the head rotates (around y-axis) to left and right. As the translatory motion of each point on the vertebra varies,[11] we have chosen point *A* to describe this motion. With the head in neutral position, the point *A* has position A_0. As the head turns left, C1 rotates on C2 and the point *A* follows a steep rise to position A_1, remains on this plateau, and then slowly comes down to A_2, at the end of motion. The x- and y-scales are in millimeters. Note that the scales have been magnified (about five times) to clearly show the path followed by the point *A*. (*Based on unpublished data, Yale University School of Medicine, Orthopaedic Biomechanics Laboratory.*)

the complete articulation has a biconvex design. This design is thought to account for the screw motion. Hultkrantz deduced that the screw movement (y-axis translation) was not characteristic of turning the head but probably occurred only in the extremes of the range of movement. There is more evidence on both sides of this discussion. Hohl has described the coupling of vertical translation of C1 with axial rotation of C1 on C2.[54] His conclusions were based on observations of cineradiographs. Werne's investigations led him to the conclusion that the screw motion depended somewhat on the extent to which the longitudinal axis of the dens correlates with the imaginary longitudinal axis of the body. The more parallel the two are, the more distinctive is the vertical displacement.[123] The extent of parallelism between the vertical axis of the coordinate system and the longitudinal axis of the dens can vary (see Fig. 2-30, p. 116). In the example shown in Figure 2-30, there is an angle of approximately 45° between these two lines. One can readily appreciate that the translation along the longitudinal axis of the dens can carry the atlas posteriorly or vertically, depending on the direction in which the dens is pointed.

Instantaneous Axes of Rotation (IAR)

Henke identified IAR for atlantooccipital motion by determining the centers of the arches formed by the outline of the joints in the sagittal and frontal planes. The x-axis passed through the centers of the mastoid processes (Fig. 2-11), and the z-axis was located at a point 2–3 cm above the apex of the dens. Although these points were identified more than a century ago, they remain the only approximation of the instantaneous axes of rotation for atlantooccipital articulation. The method used does not necessarily give accurate results. In order to locate the axes accurately, experimental investigations must be carried out, involving analysis of plane motion in carefully measured, controlled situations. The authors are not aware of such investigations of this region. The anterior atlantooccipital membrane connects the occiput to the anterior ring of C1. This probably becomes taut with extension. However, because it is a continuation of the cervical anterior longitudinal ligament, we assume that it, like the anterior longitudinal ligament described by Johnson and associates,[59] is a delicate structure. Therefore, it would have only a modest ability to restrict extension.

For the atlantoaxial joint, the instantaneous axes of rotation can be estimated from the kinematic studies of Werne.[123] Sagittal plane motion shown in Figure 2-7 locates the instantaneous axes of rotation for flexion/extension somewhere in the region of the middle third of the dens. For axial ($\pm\theta y$) rotation, the instantaneous axes of rotation may be assumed to lie in the central portion of the *axis*,[31] a fact that attests to the astuteness of the scholars who named the structure (Fig. 2-12).

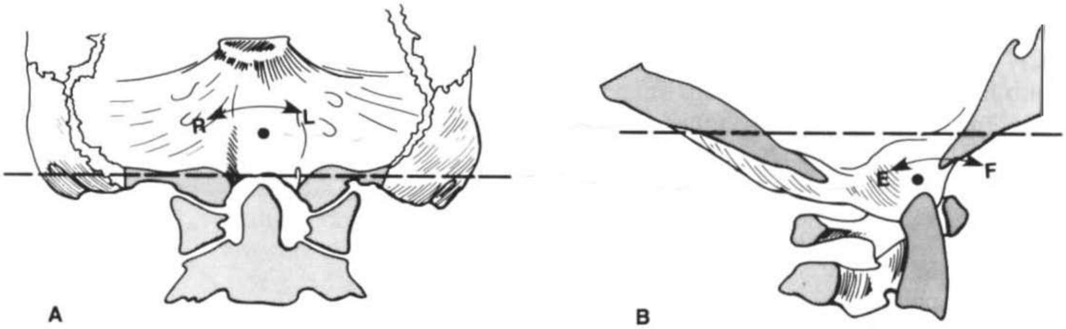


FIGURE 2-11 (A) The approximate location of the IAR for the atlantooccipital joint in the frontal plane is shown here. Lateral bending (R, L) of the occiput on C1 is thought to take place around the indicated dot. The broken line indicates the approximate location of the IAR for the flexion/extension (F, E) motion (± x-axis rotation) in the sagittal plane (z, y plane). (B) The converse is shown in the sagittal plane. The broken line localizes the IAR for lateral bending, and the dot shows the axes for flexion/extension.

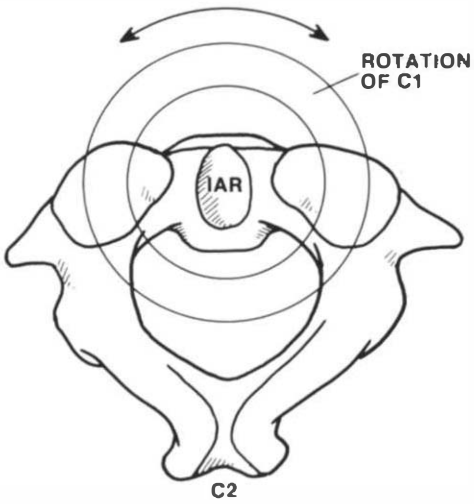


FIGURE 2-12 This is a diagrammatic representation of the approximate location of the IAR for axial rotation (± y-axis) of C1 on C2. The points are located in the region of the appropriately named *axis*.

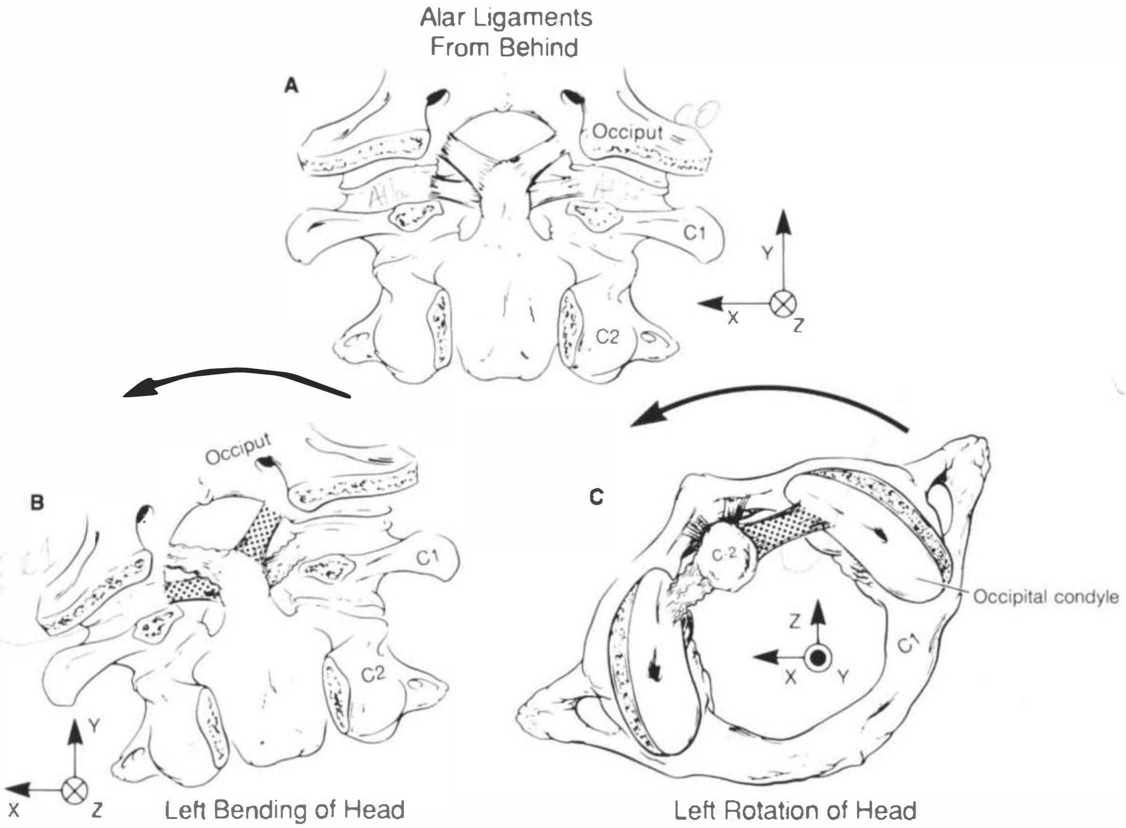
Lateral bending (x, y) plane rotation of the atlantoaxial joint is about 5° to each side. The IAR for this motion is being studied, but the precise determinations have not yet been made.

Functions of Anatomic Elements

At the C0–C1 articulation, flexion movement is checked by skeletal contact between the anterior margin of the foramen magnum and the tip of the dens. Werne discovered a well-developed, previously undescribed bursa that communicated with the joints of the dens. He called it the bursa apicis dentis. Extension is limited by the tectorial membrane (the cephalad continuation of the posterior longitudinal ligament). With flexion of the C0–C1 joint beyond neutral, the tectorial membrane becomes taut and limits forward flexion at the C1–C2 joint. It is reasonable to assume that the anterior atlanto-dental ligaments recently discovered by Dvorak and associates[28] serve to some extent as a checkrein for C1–C2, but this is not fully established at the present time. Similarly, with extension of the C0–C1 joint, the tectorial membrane again becomes taut and limits extension between C1 and C2.[123]

Axial rotation between C0 and C1 is limited by the ligaments and osseous anatomy of the C0–C1–C2 articulations. The joint surfaces are cup-shaped, with the arcuate occipital articulation fitting into the cup of C1. The alar ligaments also provide a checkrein to this motion. Dvorak and colleagues,[27] in an *in vitro* study, demonstrated increased rotation at both the atlantooccipital and atlantoaxial articulation following severing of the contralateral alar ligament. The alar ligaments are symmetrically placed on both sides of the dens, with one portion connecting the dens to the occiput and the remaining ligament connecting the dens to the atlas (Fig. 2-13A).

The mechanism for lateral bending is more complex. This motion involves 5° to one side at C0–C1 and also at C1–C2. The C0-C1 motion is controlled by both components of the alar ligaments (Fig. 2-13B). During *left lateral* bending (− z-axis rotation), the *right* upper portions of the alar ligament, connected to the occiput, and the left lower compo-



Left Bending of Head

Left Rotation of Head

FIGURE 2-13 The role of the alar ligaments in lateral bending and axial rotation. **(A)** View of the alar ligaments from behind with C0–C1–C2 complex in neutral position. **(B)** Left lateral bending (− z-axis rotation) of the head and neck. This motion is checked or restrained by the taut right upper portion of the alar ligament and the left lower portion. **(C)** Left axial rotation (+ y-axis rotation) of the head and neck. In this case, the motion is checked by a taut right alar ligament. *(Jofe, M. H., White, A. A., III, and Panjabi, M. M.: Clinically relevant kinematics of the cervical spine. In The Cervical Spine Research Society: The Cervical Spine, 2nd. ed. Philadelphia, J. B. Lippincott, 1989).*

nent, connected to the ring of C1, check the motion. The opposite is true for *right lateral* bending.

Left axial rotation (+ y-axis rotation), as in turning the head to look to one's left, is checked by the *right* alar ligament (Fig. 2-13C). The opposite is true for *right axial rotation*.

THE MIDDLE AND LOWER CERVICAL SPINE (C2–T1)

Most of the information presented in this section is based on the work of Lysell.[67] He carried out detailed examinations of the kinematics of fresh autopsy specimens using a precise radiographic technique that allowed for measurement of three-dimensional motion. The axis, C2, is a transitional vertebra between the occipital-atlanto-axial complex and the lower cervical spine, and therefore it is also discussed here.

Range of Motion

Rotation ranges for the middle and lower cervical spine are shown in Table 2-2 and Figure 2-23. Most of the motion in flexion/extension is in the central region. The C5–C6 interspace is generally considered to have the largest range. There may be some causal relationship between this observation and the incidence of cervical spondylosis at that interspace.[129] For lateral bending and axial rotation there is a tendency for a smaller range of motion in the

TABLE 2–2 Limits and Representative Values of Ranges of Rotation of the Middle and Lower Cervical Spine

	Combined Flexion/Extension (±x-axis rotation)		One Side Lateral Bending (z-axis rotation)		One Side Axial Rotation (y-axis rotation)	
Interspace	Limits of Ranges (degrees)	Representative Angle (degrees)	Limits of Ranges (degrees)	Representative Angle (degrees)	Limits of Ranges (degrees)	Representative Angle (degrees)
Middle						
C2–3	5–16	10	11–20	10	0–10	3
C3–4	7–26	15	9–15	11	3–10	7
C4–5	13–29	20	0–16	11	1–12	7
Lower						
C5–6	13–29	20	0–16	8	2–12	7
C6–7	6–26	17	0–17	7	2–10	6
C7–T1	4–7	9	0–17	4	0–7	2

more caudal segments. The relationship between disc degeneration and motion was examined by Lysell. The intervertebral disc for each motion segment was cut and graded for degeneration. There was no change in range of motion as a function of disc degeneration.[67] Other investigators have observed that a compensatory increase in motion occurs in cervical spine segments adjacent to interspaces with reduced motion due to either degeneration or post-traumatic changes.[33,60]

Two recent studies have provided additional data on the kinematics of the cervical spine.[76,86] Both studies used fresh cadaveric functional spinal units and measured three-dimensional motions in response to forces alone (Panjabi and co-workers)[86] and forces and moments (Moroney and associates).[76] Neither of these two studies found any significant variation of the physical properties with respect to the vertebral level. Average rotatory ranges of motion reported by Moroney and associates are of the same magnitude as those of Lysell,[67] except for axial rotation, which is only half as large.

The maximum sagittal plane translation (z-axis) occurring in the lower cervical spine under "physiologic loads" simulating flexion/extension has been measured directly.[127] The representative value was 2 mm and the maximum was 2.7 mm. This was the anterior translation of the anterior-inferior corner of the moving vertebra. The same measurement on a radiograph would vary with the technique employed in taking the film. The authors suggest 3.5 mm as a guide for the upper limits of normal, taking into account approximately 25% radiographic magnification (see p. 279).

The two recent studies mentioned above have also provided fresh data concerning the translations in the middle and lower cervical spine.[76,86] However, when comparing the translation findings of different studies, care must be taken to compare translations of the same anatomic point, e.g., the anterior-inferior corner of the moving vertebra.[H] In both studies, the translation was measured at the geometric center of the vertebral body. In the first study, Panjabi and associates found an average anterior translation of 1.9 mm and posterior translation of 1.6 mm.[88] Thus, the total sagittal plane translation was 3.5 mm. This result is somewhat higher than that of an earlier study,[127] and may be explained on the basis that the instantaneous axis of rotation of the vertebra in flexion/extension is generally below the moving vertebra. Therefore, the anterior translation of the geometric center of the vertebral body will be somewhat larger, being further away from the IAR, than that of the anterior-inferior corner. The study also recorded one-side lateral translation of 1.5 mm and inferior and superior translations, respectively, of 0.7 mm and 1.1 mm. In the second study, Moroney and associates found anterior and posterior translations of only 0.15 mm and 0.37 mm respectively.[76] Similarly, their measurements of lateral (0.14 mm) and inferior (0.08 mm) translations were also considerably smaller. They did not determine the superior translation. Both studies used nearly identical experimental techniques;[79] however, the magnitude of the physiological forces used to produce motion were different (50 N by Panjabi and associates and 19.6 N by Moroney and associates). This may explain the small translatory ranges of motions observed in the latter study. Results of some of the recent studies described above as well as those provided in the first edition of this book have been summarized in Table 2-5.

Patterns of Motion

As a vertebra goes through its ranges of motion, the pattern of motion is determined by a combination of the geometric anatomy of the structures and their physical properties. The positions of a vertebra from full extension to full flexion, for example, have certain similarities throughout the spine, and yet there are some characteristic regional differences and even gradations of differences within regions. Lysell showed clearly that the routes (patterns) were the same for any given vertebra whether it was going from flexion to extension or vice versa.[67] The movement is a combination of translation and rotation. He used what he called the "top angle" to indicate the steepness of the arch that was described by the vertebra while moving from full extension to full flexion.[C] The arches were flat at C2. The steepest was at C6, followed by C7. Those in between were all about the same.

The pattern of motion in the sagittal plane is shown diagrammatically in Figure 2-14. The acuity of the arc was found to decrease in association with disc degeneration; this overall pattern was shown to be a statistically significant variation.[67] The pattern of motion in the sagittal plane involves a strong coupling element.

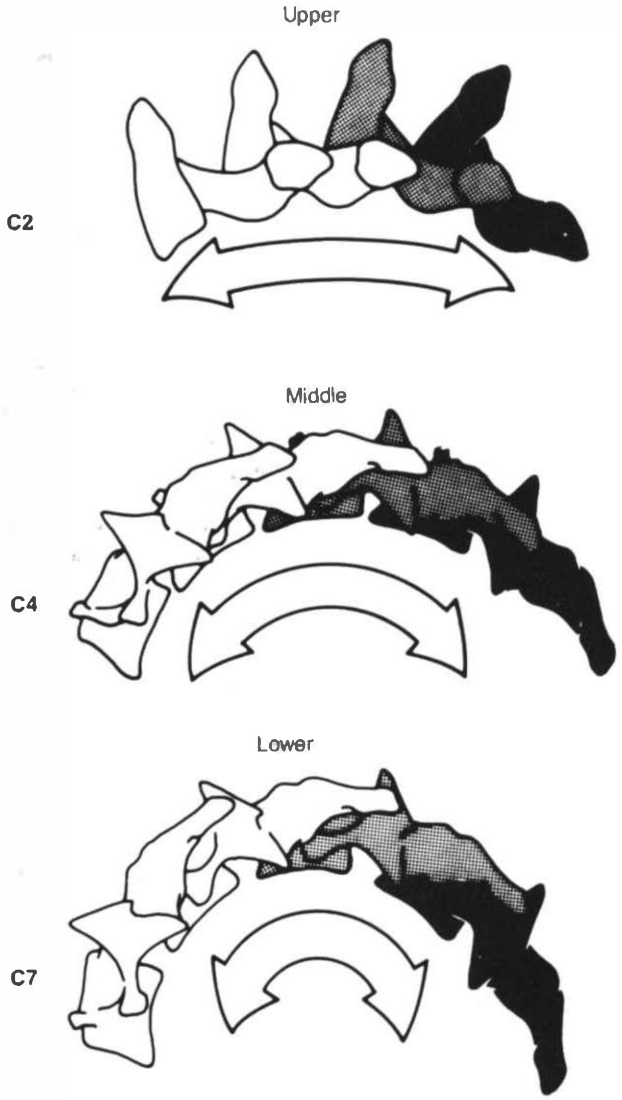


FIGURE 2-14 A diagrammatic approximation of the relative regional cephalocaudal variations in radii of curvature of the arches defined by the cervical vertebrae as they rotate and translate in the sagittal plane. The diagram depicts the patterns of motion of C2, C4, and C7, moving back and forth between full flexion and full extension.

Coupling Characteristics

The coupling patterns in the lower cervical spine are dramatic and clinically important. The coupling is such that with lateral bending the spinous processes go to the convexity of the curve.[67] In lateral bending to the left, the spinous processes go to the right, and in lateral bending to the right, they go to the left (Fig. 2-15). (In the coordinate system, +z-axis bending is coupled with −y-axis axial rotation, and −z-axis bending is coupled with +y-axis rotation.) This coupling is significant in understanding scoliosis as well as some aspects of spine trauma and its treatment. For example, a dislocation may result when a traumatic force carries a joint beyond its normal range of motion. The coupling phenomenon plays a role in that some ratios of axial rotation and lateral bending may result in a unilateral facet dislocation (see Chap. 4, p. 220).

The amount of axial rotation that is coupled with lateral bending at various levels of the spine has been studied and described.[67] At the second cervical vertebra there are 2° of coupled axial rotation for every 3° of lateral bending, a ratio of 2 to 3 or 0.67. At the seventh cervical vertebra there is 1° of coupled axial rotation for every 7.5° of lateral bending, a ratio of 1 to 7.5 or 0.13. Between C2 and C7 there is a gradual cephalocaudal decrease in the amount of axial rotation that is associated with lateral bending. This phenomenon of gradual change in the coupling ratio may be related to a change in the incline of the

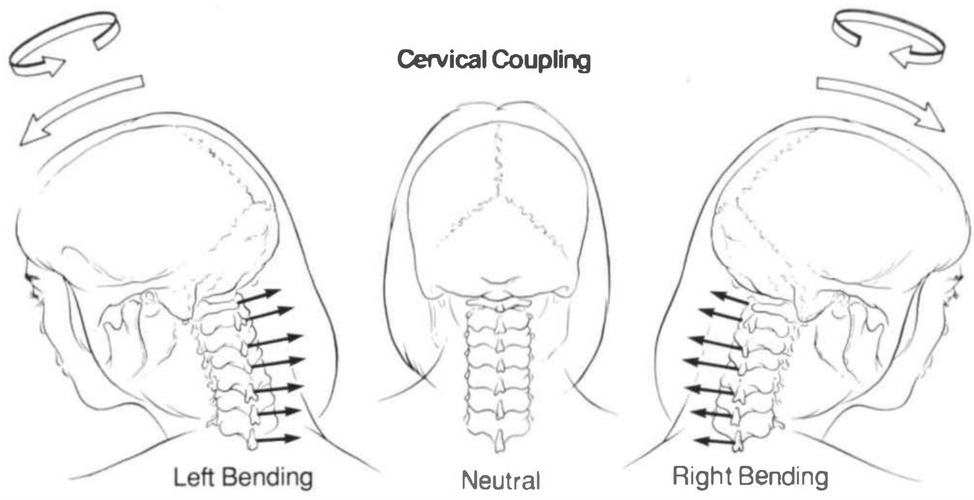


FIGURE 2-15 An important cervical spine coupling pattern. When the head and neck are bent to the right, the spinous processes go to the left. The converse is also shown. (Expressed in the coordinate system, +z-axis rotation is coupled with −y-axis rotation, and −z-axis rotation is coupled with +y-axis rotation.) This is a good frame of reference for describing and remembering the axial and lateral-bending coupling relationships in other regions of the spine. See also Figure 2-20.

facet joints. Although this has not been measured and proved, we believe that the angle of incline of the facet joints in the sagittal plane increases cephalocaudally.

In a three-dimensional *in vitro* study by Panjabi and colleagues,[86] it was noted that there is coupled lateral bending when axial rotation is imparted to the spine. Left lateral bending of 0.75° was found to be associated with 1° of left axial rotation. Expressed in the coordinate system, −z-axis rotation is coupled with a +y-axis rotation.

In the cervical spine study by Moroney and coworkers, the average ratio of coupled axial rotation to lateral bending was 0.32,[76] which nicely correlates with the average results of Lysell presented above. They also found the average ratio of coupled lateral bending to axial rotation of 0.51, which is close to the 0.75 value of Panjabi and coworkers. The direction of coupled rotation was similar to that found by Lysell.

Instantaneous Axis of Rotation

The instantaneous axis of rotation for the cervical functional spinal unit has been placed in a variety of different locations by different research workers. This is partially the result of a lack of consistency among investigators (Fig. 2-16).[D] Suggested locations for ±x-axis rotation (flexion/extension) include the body of the subjacent vertebra, the center of the vertebral body, the disc, and the nucleus pulposus. An additional theory contends that the instantaneous axis of rotation for C2 (±x-axis rotation) lies in the posterior, caudal portion of the subjacent vertebra but that there is a progression in which the instantaneous axis of rotation moves anterior and cephalad, such that for C6 it is located at the anterior, cephalad portion of the subjacent vertebra. Others suggest that there are large numbers of motion centers for each vertebra. Work is in progress that is designed specifically to locate the instantaneous axis of rotation for cervical spine motion.

Based on personal judgment of observations of patterns of motion, Lysell postulated the locations of the instantaneous axes of rotation in the cervical region.[87] The locations were not determined quantitatively.

For sagittal and horizontal plane motions, the instantaneous centers are thought to lie in the anterior portion of the subjacent vertebra. The more anterior location is suggested by Lysell, who observed very little movement of his anterior measuring point. For lateral bending, they are probably in the region of the question mark shown in Figure 2-17, but this is even more speculative.

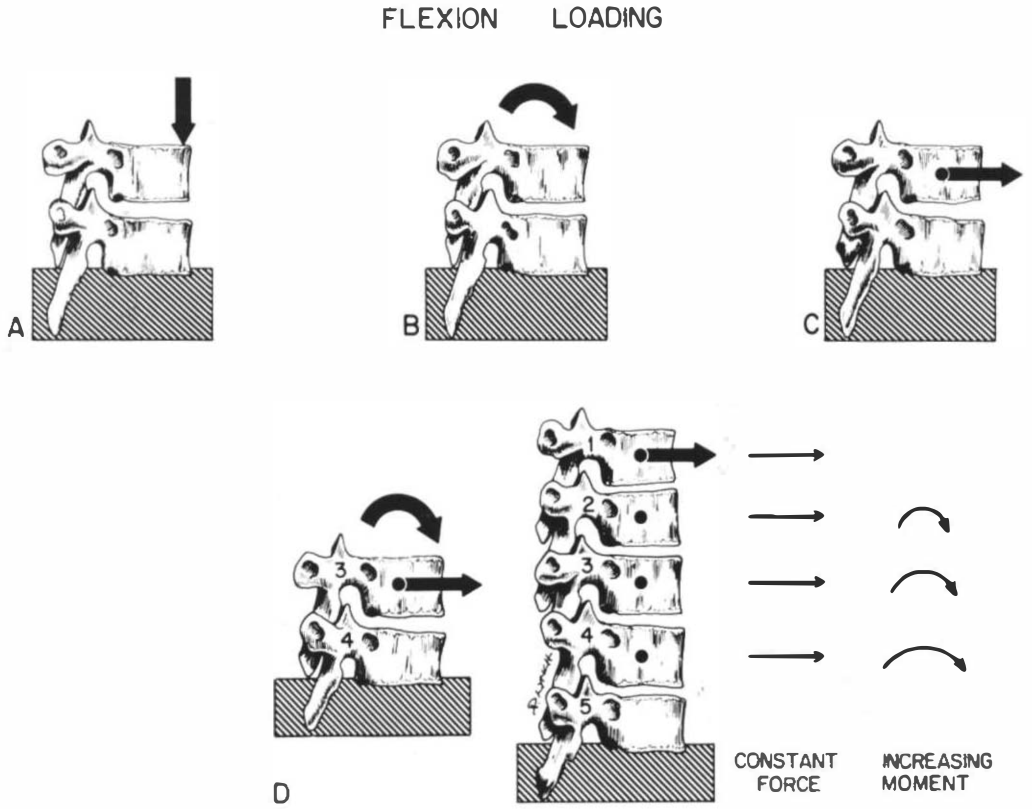


FIGURE 2-16 Various methods employed in the *in vitro* experiments to produce flexion of the spine. (**A**) Compressive load applied at an anterior point results in compression, in addition to the flexion bending movement. (**B**) Pure flexion movement. (**C**) Anterior shear force results in anterior shear plus flexion moment. (**D**) Shear applied to top vertebra of a multisegmental specimen. Result is same constant shear force at all the functional spine units, but the moment increases caudally. *(White III, A. A., and Panjabi, M. M.: Spinal kinematics. The Research Status of Spinal Manipulative Therapy. NINCDS Monograph (No. 15), p. 93. Washington, D.C., U.S. Department of Health, Education and Welfare, 1975.)*

Functions of Anatomic Elements

The function of anatomic elements in the cervical spine is discussed in detail in Chapter 5. Investigations comparing the kinematics of a motion segment under "physiologic loads," with and without various elements, show that as long as either all anterior elements or all posterior elements are intact, there is no grossly abnormal motion.[127]

The strength and orientation of the annular fibers, along with the tenacious attachment to the periphery in all regions of the vertebral body and end-plate, contribute to the great resistance of the annulus to horizontal translation. This has an important role in the clinical stability of the spine.

The range of motion of flexion/extension is to some extent dictated by the geometry and stiffness of the disc.[31,61,65,131] For example, in flexion/extension, the greater the height of the disc and the smaller the anteroposterior diameter, the greater is the motion. Similarly, if lateral bending is analyzed, the motion would be greater when the disc is higher and its lateral diameter is smaller. In addition, the greater the stiffness of the disc, the smaller is the motion.[E]

When there is a smaller diameter in the plane of motion, other things being equal, bony impingement is less likely and more motion is possible. In the cervical spine, where there is the greatest motion, the disc diameters of the sagittal and coronal plane

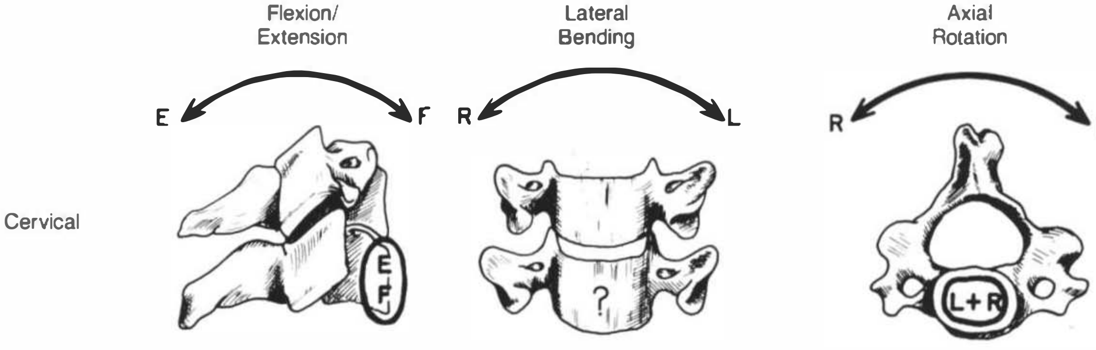


FIGURE 2-17 The approximate locations of instantaneous axes of rotation in the lower cervical spine. E is the location of IAR in going from a neutral to extended position. F is the location in going from a neutral to a flexed position. L shows the axes in left axial rotation, and R shows them in right axial rotation. The question mark indicates that there are at present no convincing estimates of the IAR for lateral bending in the cervical spine. *(White, III, A. A., and Panjabi, M. M.: Spinal kinematics. The Research Status of Spinal Manipulative Therapy. NINCDS Monograph (No. 15). p. 93. Washington, D.C., U.S. Department of Health, Education and Welfare, 1975.)*

are less than in the thoracic and lumbar regions. In addition to the disc, the stiffness of the other ligaments, especially the yellow ligament, also plays a significant role in the kinematics of the cervical spine.[5]

The uncinate processes, which begin to develop at 6 to 9 years of age and are fully developed at 18 years, may be of importance in the patterns of motion in the cervical spine. They are thought to prevent posterior translation and also to limit lateral bending. In addition, the uncinate processes serve as a guiding mechanism to the patterns of flexion/extension.[14,29,36,92]

THE THORACIC SPINE

Range of Motion

The range of sagittal plane rotation (flexion/extension) for the thoracic spine is given in Table 2-3 and Figure 2-23. The median figure is 4° of motion in the upper portion of the thoracic spine and 6° of motion in the middle segments. In the lower portion (T11-12 and T12-L1), there are 12° of motion at each segment. In the frontal plane (lateral bending) there are 6° of motion in the upper thoracic spine, with 8° or 9° in the two lower segments. In the horizontal plane (axial rotation) there are 8–9° of motion in the upper half of the thoracic spine and 2° for each interspace of the three lower segments. Here, the values for axial rotation coincide somewhat with the in vivo findings of Gregersen and Lucas, who studied axial rotation in the thoracic spines of seven medical students by inserting Steinmann pins into the spinous processes.[47] They noted an average of 6° of rotation at each level, and when their subjects were walking, the maximum amount of rotation was observed at the middle portion of the thoracic spine. Figures for each interspace are given in Table 2-3.

Patterns of Motion

The pattern of motion in the sagittal plane for the thoracic spine is somewhat similar to that in the cervical spine. In describing the patterns of cervical spine motion, the T angle, or "top angle," was employed to indicate the acuity of the arch formed by a given point as a vertebra moved in a plane.[124] To evaluate thoracic spine motion in the sagittal and frontal planes, the average curvature (the reciprocal of the radius of the arch) is used. In sagittal plane motion (flexion/extension), the average curvature is quite small, indicating a rather flat arch (Fig. 2-18A). There is no pattern of cephalocaudal variation. The average curvature in the frontal plane is also flat, but nevertheless greater, or steeper, than the arches of the sagittal plane (Fig. 2-18B). Also, in the frontal plane there is a cephalocaudal variation. The acuity of the arch tends to increase between T1 and T12.

TABLE 2–3 Limits and Representative Values of Ranges of Rotation of the Thoracic Spine

	Combined Flexion/Extension (±x-axis rotation)		One Side Lateral Bending (z-axis rotation)		One Side Axial Rotation (y-axis rotation)	
Interspace	Limits of Ranges (degrees)	Representative Angle (degrees)	Limits of Ranges (degrees)	Representative Angle (degrees)	Limits of Ranges (degrees)	Representative Angle (degrees)
T1–T2	3–5	4	5	5	14	9
T2–T3	3–5	4	5–7	6	4–12	8
T3–T4	2–5	4	3–7	5	5–11	8
T4–T5	2–5	4	5–6	6	5–11	8
T5–T6	3–5	4	5–6	6	5–11	8
T6–T7	2–7	5	6	6	4–11	7
T7–T8	3–8	6	3–8	6	4–11	7
T8–T9	3–8	6	4–7	6	6–7	6
T9–T10	3–8	6	4–7	6	3–5	4
T10–T11	4–14	9	3–10	7	2–3	2
T11–T12	6–20	12	4–13	9	2–3	2
T12–L1	6–20	12	5–10	8	2–3	2

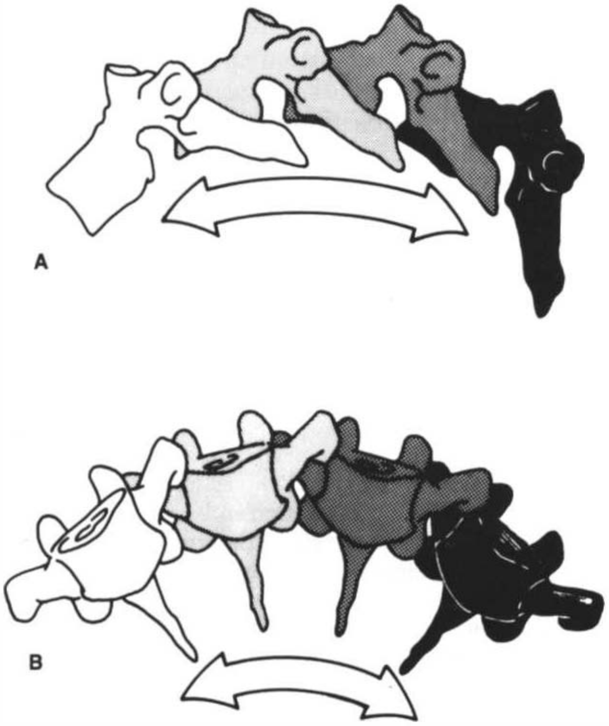


FIGURE 2-18 A diagrammatic approximation of the relative variations in the radii of curvature of the arches defined by the thoracic vertebrae as they move in the sagittal and in the frontal planes. The arches described in the pattern of motion for lateral bending (B) are more accentuated than those described in flexion/extension (A).

Coupling Characteristics

There are a number of different coupling patterns, many of which may prove to be of clinical significance in the future. Of most interest at present in both the cervical and thoracic spines is coupling between lateral bending and axial rotation. Considerable interest in the thoracic spine is due to the significance of normal coupling and abnormal coupling in scoliotic deformities (see Chap. 3, p. 135).

Coupling has caused considerable literary controversy; not only are there debatable characteristics, but the occurrence of such a phenomenon is sometimes questioned. The historical aspects of this controversy have been reviewed.[124] The disagreement is due to a wide variety of different techniques as well as to the complexity of the motion under analysis.

The pattern of coupling in the thoracic spine is of the type that has been described in the cervical spine. Lateral bending is coupled with axial rotation such that the spinous processes move toward the convexity of the lateral curvature. The cephalocaudal variation of this coupling pattern within the thoracic spine is of considerable interest. In the upper portion of the thoracic spine the two motions are strongly coupled, although not as strongly as in the cervical spine. In the middle portion of the thoracic spine the coupling pattern is by no means as distinct; moreover, it is inconsistent, and in some instances in the middle portion the spinous processes rotate toward the concavity of the lateral curvature. Also, in the lower portion of the thoracic spine the coupling pattern is not as strong as in the upper portion. These patterns have been documented in detail elsewhere.[F,124]

Actually, a study by Panjabi and colleagues has shown that all of the six degrees of freedom demonstrate coupling patterns of varying degree.[79]

There has been relatively more interest in the coupling of axial rotation and lateral bending. This is largely due to its relevance in the etiology, evaluation, and treatment of scoliosis. This coupling is also important in the mechanisms of injury in the cervical spine. Abnormal coupling patterns have been viewed and analyzed in the context of possible evidence of instability. Changes in coupling patterns have also been noted adjacent to spinal fusions. Finally, this particular coupling characteristic may have relevance in the basic biomechanics of different regions of the spine. Because of these important factors, which are discussed in various parts of this text, we've elected to prepare composites, which are shown in Figure 2-19.

Instantaneous Axis of Rotation

The shortcomings of current descriptions of the location of the instantaneous axis of rotation have been discussed.[D] The approximate locations of these centers for the thoracic spine are represented diagrammatically in Figure 2-20.

In a recent study using fresh cadaveric functional spinal units covering all levels of the thoracic spine,

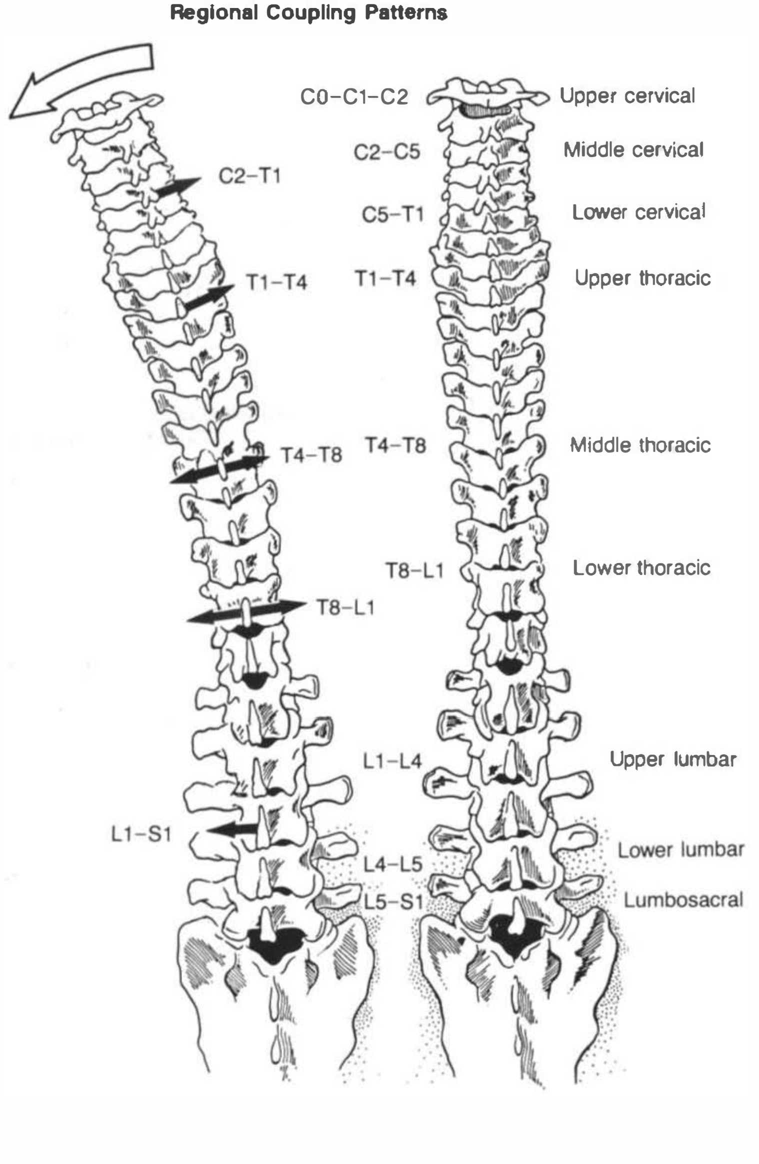


FIGURE 2-19 This diagram summarizes the coupling of lateral bending and axial rotation and depicts the new biomechanical subdivisions of the spine. The actual coupling is between ± z-axis rotation and ± y-axis rotation. It can also be thought of in terms of the direction of movement of spinous processes with left lateral bending. Note that in the middle and lower cervical spine as well as in the upper thoracic spine there are the same coupling patterns. In the middle and the lower thoracic spine, the axial rotation, which is coupled with lateral bending, can be in either direction, that is, it can be ± y-axis rotation. The direction of this axial rotation apparently varies between different specimens. In the lumbar spine there is − y-axis rotation associated with − z-axis rotation. That is, the spinous processes go to the left with left lateral bending. The same pattern is also present at the lumbosacral FSU.

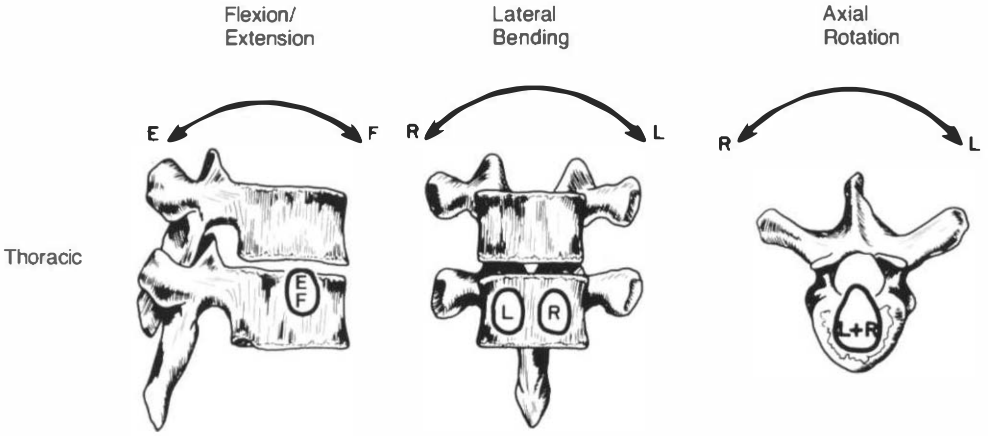


FIGURE 2-20 The approximate locations of the instantaneous axes of rotation in the thoracic spine. E is the location of the axes going from a neutral to an extended position. F is the location of the axes going from a neutral to a flexed position. L shows the IAR in left lateral bending or left axial rotation (+ y-axis), and R shows the axes in right lateral bending or right axial rotation. *(White III, A. A., and Panjobi, M. M.: Spinal kinematics. The Research Status of Spinal Manipulative Therapy, NINCDS Monograph (No. 15). p. 93. Washington, D.C., U.S. Department of Health, Education and Welfare, 1975.)*

centers of rotation for the sagittal plane motions have been obtained.[84] Locations of the centers of rotations were obtained by the application of four different loads, all producing flexion/extension motions. When anterior shear force (producing flexion) was applied at the geometric center of the vertebra, the IAR was located at the inferior end-plate of the lower vertebra. It moved even further down when posterior shear force (producing extension) was applied. Producing the same motions by applying flexion and extension pure moments, respectively, the IARs moved superiorly. For both flexion and extension moments, the IARs were located at the superior end-plate of the inferior vertebra of the functional spinal unit.

Helical Axis of Motion

Although the concept of a helical axis of motion in the thoracic spine has been introduced,[124] no studies have been designed to precisely determine the site and orientation for a representative sample of vertebral motion.

Functions of Anatomic Elements

The functions of various anatomic elements have been studied in thoracic spine kinematics. With regard to the effect of removal of all posterior elements on the mechanics of the thoracic spine, several parameters were studied in individual motion segments.[124] In the movement where extension was simulated there was a statistically significant increase in extension following removal of the posterior elements. This is due to the fact that the intervertebral joints and the spinous processes limit the amount of extension that occurs in this region. This also supports the description of these structures as load-bearing elements. The differences throughout the full range of flexion/extension are shown in Figure 2-21. The increase in rotation also occurred in the horizontal plane (y-axis rotation) upon removal of the posterior elements. These biomechanical changes were also shown to be statistically significant. Because of the spatial alignment of the facet articulations, bony impingement resisting axial rotation is not believed to occur (Fig. 2-22). The posterior ligaments, primarily the yellow ligaments and also the facet joint capsules, are probably the major structures that resist axial motion. The resistance results from the development of tension in spinal structures. After the posterior elements are removed, the motion is restricted solely by the annulus fibrosus and the muscles.

The effect of the removal of the posterior elements on the instantaneous axes of rotation has been studied in the three traditional planes of motion. Only a slight shift of the points was noted. This

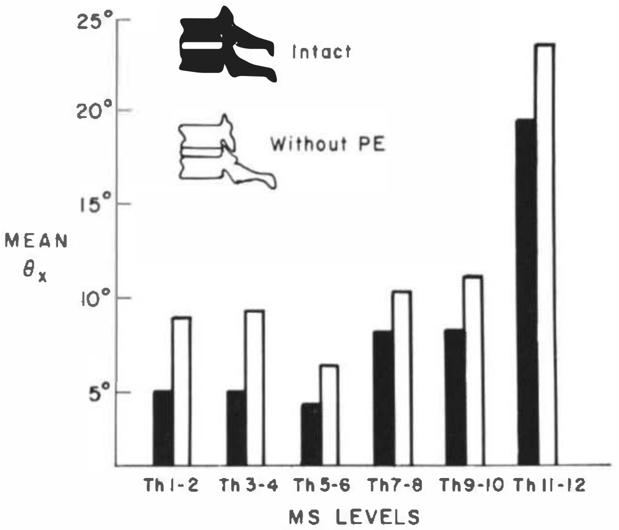


FIGURE 2-21 This is a graphic representation of average total sagittal plane rotation (flexion/extension) at the different thoracic levels, with and without the posterior elements.

observation is of theoretical interest because of the significance of the spatial alignment of the facet joints with regard to the patterns of motion. If the removal of these joints does not affect the rotation axes, the role of the facet joints in the mechanics of the spine comes into question. At present, resolution of this question awaits further study.

THE LUMBAR SPINE

Range of Motion

The representative rotations in flexion/extension, lateral bending, and axial rotation are shown in Table 2-4 and Figure 2-23. In flexion/extension there is usually a cephalocaudal increase in the range of motion in the lumbar spine. The lumbosacral joint offers more sagittal plane motion than do the other lumbar joints. For lateral bending, each level is about the same, except for the lumbosacral joint, which shows a relatively small amount of motion. The situation is about the same for axial rotation.[86] It is not unreasonable to speculate that the high incidence of clinically evident disc disease at L4–L5 and L5–S1 may be related to mechanics. These two areas bear the highest loads and tend to undergo the most motion in the sagittal plane.

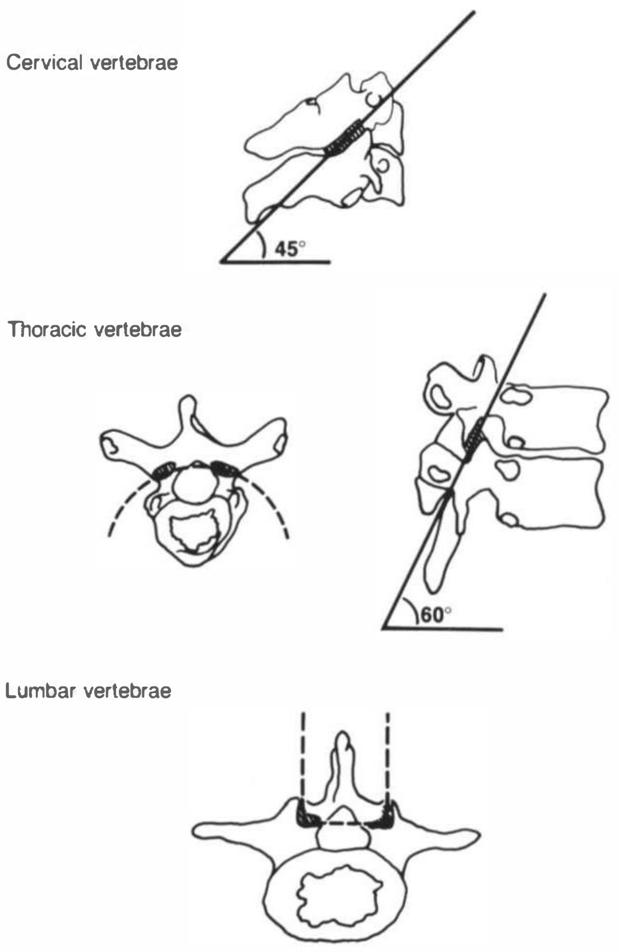


FIGURE 2-22 Characteristic facet orientation in the cervical, thoracic, and lumbar regions. The spatial alignment of the facet joints determines, to a large extent though not completely, the characteristic kinematics of different regions of the spine.

An important component of lumbar spine kinematics is that of sagittal plane translation. This is because measurement of this parameter is frequently used to determine whether or not there is instability. There is considerable variation in measuring techniques. The work of Pearcy[95] is based on sound methodology and suggests that 2 mm of anterior sagittal plane translation is normal for the lumbar spine. The *in vitro* work of Posner and colleagues,[97] who used preloads to simulate physiologic conditions, suggested 2.8 mm of anterior displacement as the upper limits of normal. Thus, after careful consideration of a number of factors, we suggest 4.5 mm for evaluation of clinical instability (see p. 354).

TABLE 2–4 Limits and Representative Values of Ranges of Rotation of the Lumbar Spine

	Combined Flexion/Extension (±x-axis rotation)		One Side Lateral Bending (z-axis rotation)		One Side Axial Rotation (y-axis rotation)	
Interspace	Limits of Ranges (degrees)	Representative Angle (degrees)	Limits of Ranges (degrees)	Representative Angle (degrees)	Limits of Ranges (degrees)	Representative Angle (degrees)
L1–L2	5–16	12	3–8	6	1–3	2
L2–L3	8–18	14	3–10	6	1–3	2
L3–L4	6–17	15	4–12	8	1–3	2
L4–L5	9–21	16	3–9	6	1–3	2
L5–S1	10–24	17	2–6	3	0–2	1

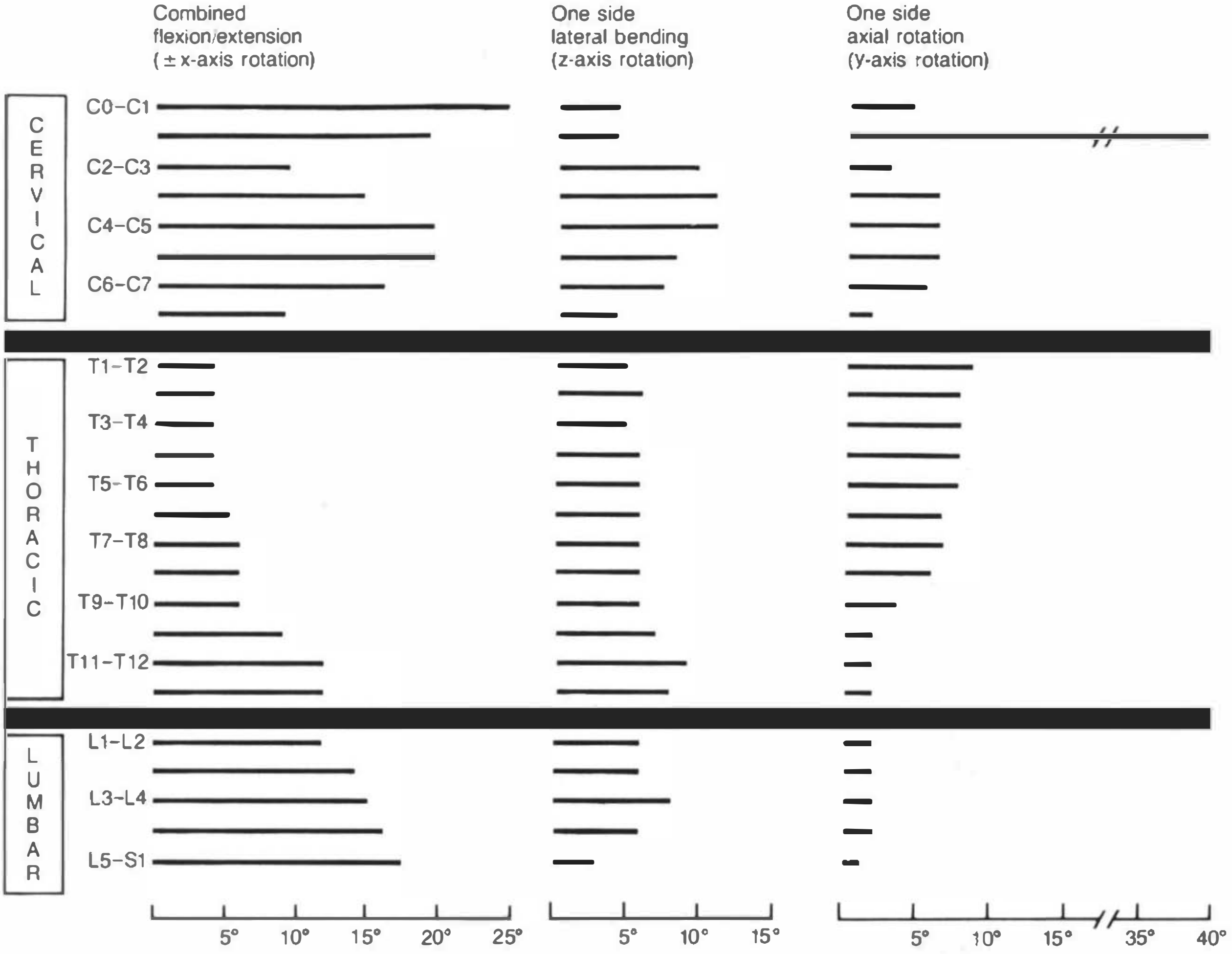


FIGURE 2-23 This is an updated composite of what the authors consider, based on careful review of the literature, to be the most representative values for rotatory ranges of motion at different levels of the spine (in the traditional planes of motion).

In a new study of 41 normal subjects, lumbar spine motions of flexion/extension and lateral bending were studied using x-rays and specially developed computer programs for the analysis.[26] One important aspect of this study was the measurement of translatory motions, which were determined for the inferior as well as the superior-posterior corners of the moving vertebra. There were significant differences between these two measurements. In general, the superior end-plate motion was larger (up to 2.5 times greater) than the lower end-plate motion. Total anterior translation from extended posture to flexed posture was on average 5 mm for the lower end-plate and 12 mm for the superior end-plate. Equivalent results for side-to-side bending were respectively 1.4 mm and 7.4 mm in the frontal plane.

Results of some of the recent studies described above as well as those provided in the first edition of this book have been summarized in Table 2-5, p. 110.

Coupling Characteristics

Traditionally, we have looked at coupling of axial rotation and lateral bending as reciprocal. In other words, we have assumed that if we bend the spine laterally and measure the associated axial rotation, then the ratio between axial rotation and lateral bending would be the same as if we had actually rotated the spine and measured the lateral bending. Recent experiments have shown that this assumption is not true, at least in the lumbar spine.[88] The specifics of these recent observations are discussed later.

There are several coupling patterns that have been observed in the lumbar spine. Rolander observed an interesting coupling of y-axis rotation (axial) with +y-axis translation.[102] However, in more detailed experiments, this particular coupling was found to be rather weak.[85] In recent stereoradiographic studies of the lumbar spine, observed coupling of 2° of axial rotation and 3° of lateral bending with flexion/extension (i.e., y-axis and z-axis rotations coupled with x-axis rotation).[89,95] Because the spinal column in general may be assumed to be symmetrical about the sagittal plane, one would not expect any coupled rotations in association with sagittal plane motions. We suggest that the observed coupled motions may be due to the suboptimal muscle control.

One of the strongest coupling patterns is that of lateral bending (z-axis rotation) with axial rotation (y-axis rotation) (Fig. 2-24). The direction of lateral bending coupled with axial rotation is such that the spinous processes point in the same direction as the lateral bending.[72] This pattern is the opposite of that in the cervical spine and the upper thoracic spine.

Work by Pearcy and Tibrewal[90] has confirmed previous lumbar spine kinematic investigations and has added some new coupling information. These researchers noted a coupling pattern of lateral bending with axial rotation at the lumbosacral joint that is the opposite of that found in the lumbar spine and the same as that observed in the cervical spine (below C2). The lumbar and lumbosacral spine coupling patterns are illustrated in Figures 2-19 and 2-24, respectively.

Panjabi and associates[88] have confirmed most of the coupling patterns observed by Pearcy and Tibrewal. They used fresh cadaveric whole lumbar spine specimens and studied the three-dimensional motions of each lumbar vertebra in five different sagittal plane postures, from full extension to full flexion. There were two findings that were somewhat different from those of Pearcy and Tibrewal. First, the left axial rotation, performed in neutral posture of the specimen, produced right lateral bending only in the upper three lumbar levels, i.e., L1–L2, L2–L3, and L3–L4. The levels L4–L5 and L5–S1 showed left lateral bending. A similar pattern was seen for the right axial rotation, i.e., the upper three lumbar levels showed lateral bending in the opposite direction, whereas the L4–L5 and L5–S1 showed bending on the same side as the axial rotation. (Note that in both studies the coupled axial rotation did not switch direction in response to lateral bending, at any levels.) Second, the two studies found different directions of observed coupled sagittal plane rotation. In the study by Panjabi and associates, the accompanying sagittal plane rotation, with the spine in neutral posture, was a flexion at all levels. In contrast, Pearcy and Tibrewal found mostly extension, except at the lumbosacral junction where they measured flexion for right bending but not for left bending. In addition, Panjabi and associates discovered an interesting effect of posture on the sagittal plane rotations. They found that in the extension posture the coupled motion was a flexion, while in the flexion postures the coupled motion was an extension. In other words, as the spine is laterally bent or axially rotated, it has a tendency to straighten (go into neutral posture) from the flexed as well as the extended postures.

Although these coupling patterns constitute fundamental and essential elements in the understand-

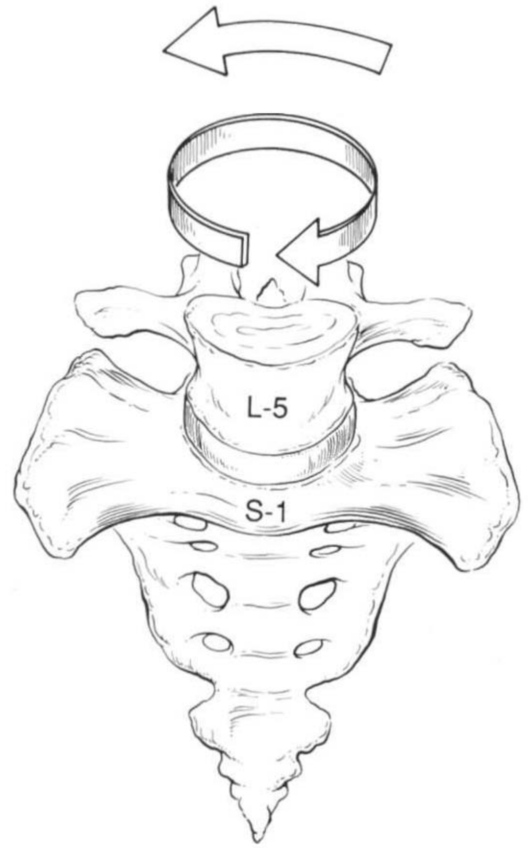


FIGURE 2-24 Axial rotation coupling patterns. In *lumbosacral coupling* (shown above), right axial rotation (−y-axis) results in right lateral bending (+z-axis). This pattern is similar to that of the middle and lower cervical spine, but opposite to that in the lumbar region. In *lumbar coupling* (not shown), right axial rotation (−y-axis) results in left lateral bending (−z-axis), and vice versa. Note that this coupling pattern is present in the upper three lumbar FSUs and is opposite to that in the cervical spine. (*Data from Panjabi, M. M., Yamamoto, I., Oxland, T., and Crisco, J. J.: How does posture affect the coupling? Spine, 14(9):1002, 1989 and Pearcy, M. J., and Tibrewal, S. B.: Axial rotation and lateral bending in the normal lumbar spine measured by three-dimensional radiography. Spine, 9(6):582, 1984.*)

ing of lumbar spine kinematics, investigators have not been able to attach any definite clinical significance to them. However, one can justifiably hypothesize that the transitional nature and the increased motion[95] of the L4–L5 FSU in comparison to the lumbar and lumbosacral levels may be the mechanical factors contributing to the higher incidence of clinical instability at that level.

Instantaneous Axes of Rotation (IAR)

The rotation axes for the sagittal plane of the lumbar spine have been described in several reports. Calve and Galland in 1930 suggested that the center of the intervertebral disc is the site of the axes for flexion/extension.[10] Rolander showed that when flexion is simulated starting from a neutral position, the axes are located in the region of the anterior portion of the disc,[102] as shown in Figure 2-25. Reichmann and colleagues[98] reported that the instantaneous axis of rotation is occasionally in the region of the disc, but in the majority of situations it is outside the disc and a considerable distance from it. In lateral bending, frontal plane rotation, the axes fall in the region of the right side of the disc with left lateral bending and in the region of the left side of the disc with right lateral bending (Fig. 2-25). In an *in vivo* study of lateral bending, there was much spread in the results, but, in general, the IARs were much lower than those shown in Figure 2-25.[20]

For axial (y-axis) rotation, the instantaneous axes of rotation are located in the region of the posterior nucleus and annulus.[15] A pattern of displacement of the rotation axes was not apparent according to evidence of disc degeneration. It should be noted that because of the strong coupling between the axial and lateral bending rotations, the intervertebral motions are truly three-dimensional. Therefore, as discussed earlier, in such situations there is no instantaneous axis of rotation. Instead, the motion must be represented by the helical axis of motion.

In the lumbar spine, the location of the IAR (or some analogous concept) has received considerable attention.[38,39] The major thrust of the investigations has been to show differences in the IAR points of a diseased lumbar spine as opposed to those of the healthy state, with the rational expectation that this may provide some basic knowledge related to or leading to some useful diagnostic tool. The hope is that this type of mechanical observation may provide some insight or understanding concerning the cause of pain or morphologic changes. The other potential value of IAR studies is the possible development of a diagnostic tool that could help identify the source and location of pain.

In both sagittal and frontal plane rotation with a normal disc, the instantaneous axes of rotation were found in a relatively concentrated area.[102] However, in the presence of disc degeneration there was a distinct tendency for the axes to be spread out, as shown in Figure 2-26. Similar observations were

TABLE 2–5 Summary of Cervical and Lumbar Spine Rotations (Comprehensive References)

CERVICAL SPINE RANGES OF MOTION

FLEXION PLUS EXTENSION

	1. DVORAK, 88			2. DVORAK, 88			3. PANJABI,88			4. PENNING, 78			W& P, 78*		
	SPINE 13/7:748			SPINE 13/7:748			SPINE 13/7:728			AJRoentlg 130:317,1978					
	In vivo/active			*In vivo*/passive			*In vitro*/whole spine			*In vivo*/active					
	MEAN	LOWER	UPPER	MEAN	LOWER	UPPER	MEAN	LOWER	UPPER	MEAN	LOWER	UPPER	*MEAN*	*LOWER*	*UPPER*
C0-C1							24.5	9.9	37.4	30.0	25.0	45.0	*13.0*		
C1-C2	12.0	5.0	20.0	15.0	8.0	22.0	22.4	3.0	41.3	30.0	25.0	45.0	*10.0*		
C2-C3	10.0	5.0	15.0	12.0	6.0	17.0				12.0	5.0	16.0	*8.0*	*5.0*	*23.0*
C3-C4	15.0	7.0	23.0	17.0	10.0	24.0				18.0	13.0	26.0	*13.0*	*7.0*	*38.0*
C4-C5	19.0	13.0	26.0	21.0	14.0	28.0				20.0	15.0	29.0	*12.0*	*8.0*	*39.0*
C5-C6	20.0	13.0	28.0	23.0	16.0	31.0				20.0	16.0	29.0	*17.0*	*4.0*	*34.0*
C6-C7	19.0	11.0	26.0	21.0	13.0	29.0				15.0	6.0	25.0	*16.0*	*1.0*	*29.0*
C7-T1													*6.0*	*4.0*	*17.0*

LATERAL BENDING (One Side)

	1. PANJABI, 88			2. MORONEY,88			3. PENNING, 78						W& P, 78*		
	SPINE 13/7:728			J BIOM21/9:769,88			AJRoentlg 130:317,1978								
	In vitro/whole spine			*In vitro*/FSU			*In vivo*/active								
	MEAN	LOWER	UPPER	MEAN	LOWER	UPPER	MEAN	LOWER	UPPER				*MEAN*	*LOWER*	*UPPER*
C0-C1	5.5	1.7	13.3				5.0						*8.0*		
C1-C2	6.7	0.8	16.5										*0.0*		
C2-C3				4.7			6.0						*10.0*	*11.0*	*20.0*
C3-C4				4.7			6.0						*11.0*	*9.0*	*15.0*
C4-C5				4.7			6.0						*11.0*	*0.0*	*16.0*
C5-C6				4.7			6.0						*8.0*	*0.0*	*16.0*
C6-C7				4.7			6.0						*7.0*	*0.0*	*17.0*
C7-T1				4.7			6.0						*4.0*	*0.0*	*17.0*

AXIAL ROTATION (One Side)

	1. PANJABI			2. Dvorak			3. Penning						W& P, 78*		
	SPINE 13/7:728			SPINE 12/8:726,1987			SPINE12/8:732,1987								
	In vitro/whole spine			*In vivo*/passive			*In vivo*/active								
	MEAN	LOWER	UPPER	MEAN	LOWER	UPPER	MEAN	LOWER	UPPER				*MEAN*	*LOWER*	*UPPER*
C0-C1	7.3	3.8	11.2	4.0			1.0	-2.0	5.0				*0.0*		
C1-C2	38.9	27.1	49.0	41.5	38.0	44.0	40.5	29.0	46.0				*47.0*		
C2-C3				3.0	-2.7	8.7	3.0	0.0	10.0				*9.0*	*6.0*	*28.0*
C3-C4				6.5	1.2	11.8	6.5	3.0	10.0				*11.0*	*10.0*	*28.0*
C4-C5				6.7	1.3	12.1	6.8	1.0	12.0				*12.0*	*10.0*	*26.0*
C5-C6				7.0	1.3	12.7	6.9	2.0	12.0				*10.0*	*8.0*	*34.0*
C6-C7				5.4	0.0	10.8	5.4	2.0	10.0				*9.0*	*6.0*	*15.0*
C7-T1				2.1	-2.8	7.0	2.1	-2.0	7.0				*8.0*	*5.0*	*13.0*

(continued)

TABLE 2–5 Summary of Cervical and Lumbar Spine Rotations (Comprehensive References) *(continued)*

LUMBAR SPINE RANGES OF MOTION

FLEXION PLUS EXTENSION

	YAMAMOTO,89			HAYES, 89			PEARCEY,84			DVORAK,89			*W& P, 78**		
	ISSLS, Kyoto			SPINE14/3:327-331			SPINE9/3:294-297			ISSLS, Kyoto					
	In vitro			*In vivo*/active			*In vivo*/active			*In vivo*/passive					
	MEAN	LOWER	UPPER	MEAN	LOWER	UPPER	MEAN	LOWER	UPPER	MEAN	LOWER	UPPER	*MEAN*	*LOWER*	*UPPER*
L1/2	10.7	5.0	13.0	7.0	1.0	14.0	13.0	3.0	23.0	11.9	8.6	17.9	*12.0*	*9.0*	*16.0*
L2/3	10.8	8.0	13.0	9.0	2.0	16.0	14.0	10.0	18.0	14.5	9.5	19.1	*14.0*	*11.0*	*18.0*
L3/4	11.2	6.0	15.0	10.0	2.0	18.0	13.0	9.0	17.0	15.3	11.9	21.0	*15.0*	*12.0*	*18.0*
L4/5	14.5	9.0	20.0	13.0	2.0	20.0	16.0	8.0	24.0	18.2	11.6	25.6	*17.0*	*14.0*	*21.0*
L5/S1	17.8	10.0	24.0	14.0	2.0	27.0	14.0	4.0	24.0	17.0	6.3	23.7	*20.0*	*18.0*	*22.0*

LATERAL BENDING (One Side)

	YAMAMOTO,89			PEARCEY,84			DVORAK, 89						*W & P, 78**		
	ISSLS, Kyoto			SPINE9/6:582-587			ISSLS, Kyoto								
	In vitro			*In vivo*/active			*In vivo*/passive								
	MEAN	LOWER	UPPER	MEAN	LOWER	UPPER	MEAN	LOWER	UPPER				*MEAN*	*LOWER*	*UPPER*
L1/2	4.9	3.8	6.5	5.5	4.0	10.0	7.9		14.2				*6.0*	*3.0*	*8.0*
L2/3	7.0	4.6	9.5	5.5	2.0	10.0	10.4		16.9				*6.0*	*3.0*	*9.0*
L3/4	5.7	4.5	8.1	5.0	3.0	8.0	12.4		21.2				*8.0*	*5.0*	*10.0*
L4/5	5.7	3.2	8.2	2.5	3.0	6.0	12.4		19.8				*6.0*	*5.0*	*7.0*
L5/S1	5.5	3.9	7.8	1.0	1.0	6.0	9.5		17.6				*3.0*	*2.0*	*3.0*

AXIAL ROTATION (One Side)

	YAMAMOTO,89			PEARCEY,84									*W & P, 78**		
	ISSLS, Kyoto			SPINE9/6:582-587											
	In vitro			*In vivo*/active											
	MEAN	LOWER	UPPER	MEAN	LOWER	UPPER							*MEAN*	*LOWER*	*UPPER*
L1/2	2.1	0.9	4.5	1.0	-1.0	2.0							*2.0*	*1.0*	*3.0*
L2/3	2.6	1.2	4.6	1.0	-1.0	2.0							*2.0*	*1.0*	*3.0*
L3/4	2.6	0.9	4.0	1.5	0.0	4.0							*2.0*	*1.0*	*3.0*
L4/5	2.2	0.8	4.7	1.5	0.0	3.0							*2.0*	*1.0*	*3.0*
L5/S1	1.3	0.6	2.1	0.5	-2.0	2.0							*5.0*	*3.0*	*6.0*

*White, A. A., and Panjabi, M. M.: Clinical Biomechanics of the Spine. ed. 1. Philadelphia, J. B. Lippincott, 1978.

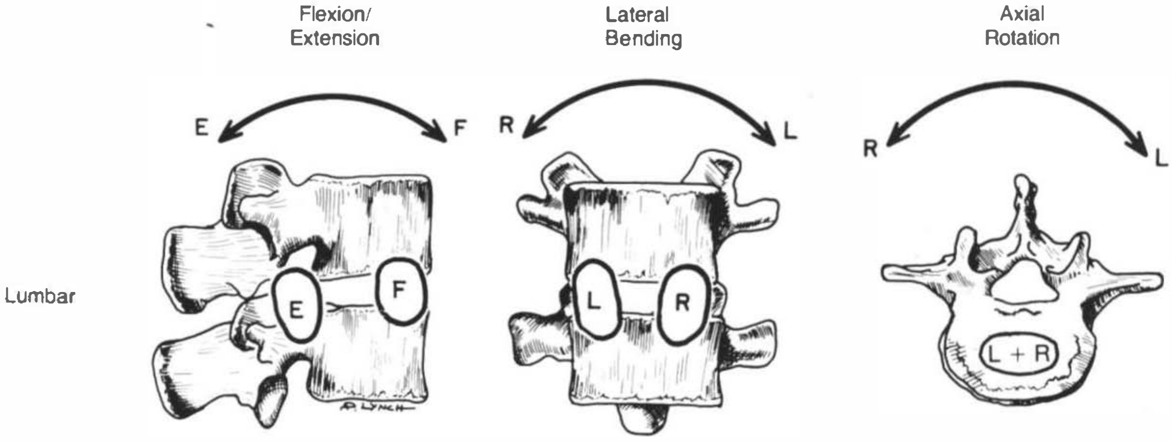


FIGURE 2-25 The approximate locations of the instantaneous axes of rotation in the lumbar spine. *E* is location of the axes going from a neutral to an extended position. *F* is the location of the axes going from a neutral to a flexed position. *L* shows the IAR in left lateral bending or left axial rotation, and *R* shows the axes in right lateral bending or right axial rotation. *(White III, A. A., and Panjabi, M. M.: Spinal kinematics. The Research Status of Spinal Manipulative Therapy. NINCDS Monograph (No. 15). p. 93. Washington, D.C., U.S. Department of Health, Education and Welfare, 1975.)*

made by Pennal and colleagues[91] using radiographic technique *in vivo*. This general approach of determining "abnormalities" of IAR to diagnose disc degeneration or other diseases is appealing.

Gertzbein and associates[38,39] have used *in vitro* and *in vivo* methodologies to study IARs in the lumbar spine. The researchers used the term "centrode" for IAR. Like Rolander, they found in flexion/extension a greater scatter of centrodes in those FSUs in which there was morphologic evidence of disc degeneration.

Although at present there is no distinct, clinically workable use for IAR, there are some speculations that merit discussion. Theoretically, once measuring techniques are developed, it should be possible to recognize, in addition to disc degeneration, clinical instability and changes in the physical properties of the ligamentous structures. In addition, the reliable identification of IAR could be of value in predicting the behavior of the spine motor units in response to different injurious vectors (see Chap. 4). Finally, the instantaneous axis of rotation is significant in determining the efficacy of different constructs of spine fusions (see Chap. 8).

Helical Axis of Motion

We know of only one preliminary study in which the helical axes of motion in the lumbar spine were determined.[43] Using fresh cadaveric lumbar functional spinal units, intervertebral motions due to lateral shear force and pure lateral bending moments were studied. For the application of right lateral bending moment, the HAM axis intersected the frontal plane (xy-plane) at about 10 mm to the right ($x = -10$ mm) and 21 mm inferiorly ($y = -21$ mm) with respect to the geometric center of the moving vertebra. The axis was inclined such that it made a longitude of 15° with respect to the sagittal plane and latitude of 85° with respect to the horizontal plane.

Functions of Anatomic Elements

The lumbar intervertebral joints are thought to be anatomically designed to limit anterior translation and permit considerable sagittal and frontal plane rotations.[63] The intervertebral joints are aligned to resist axial rotation. In general, these joints are thought to serve as a guide for the patterns of displacement of the motion segments.[4,77]

THE SACROILIAC REGION

Relatively little is known about the kinematics of this important set of articulations, which constitute a fertile region for future research. This is the link through which the weight of the trunk is transmitted

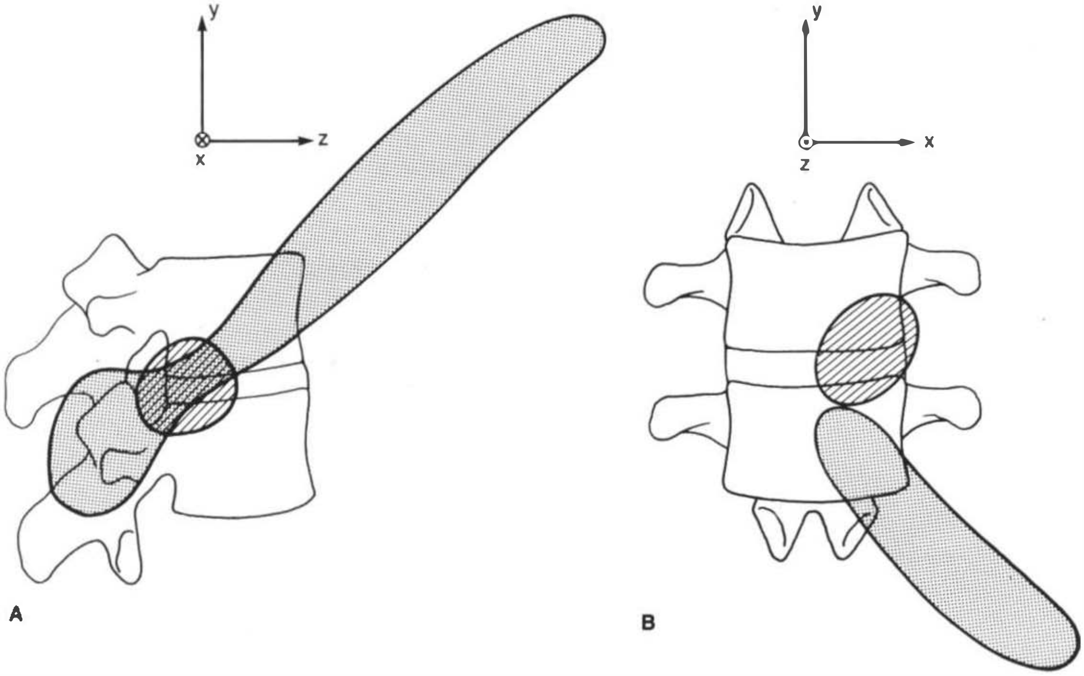


FIGURE 2-26 The changes in the location of the instantaneous axes of rotation in lumbar spine motion segment, with and without degenerative disc disease. The axes for the normal discs are shown in the dark areas with longitudinal lines, and those for the degenerated discs are shown in the lighter gray areas. Flexion is represented by **A** and right lateral bending by **B**. Left lateral bending would be represented by a mirror image of **B**. *(Data from Rolander, S. D.: Motion of the lumbar spine with special reference to the stabilizing effect of posterior fusion [thesis]. Acta Orthop. Scand., 90 [Suppl.], 1966.)*

to the legs and a region to which the patient will frequently point to localize back pain.

The sacroiliac joint is partly synovial and partly syndesmotic. It may be completely ankylosed in as much as 76% of subjects over 50 years of age.[9] This fact makes the kinematic study of the joint a moot issue for a significant portion of the population. However, for many others these are rather stiff joints whose overall motion and stability depend largely upon the coarseness of the interdigitating articular surfaces.[107]

Range of Motion

Studies of the motion of these joints have yielded a broad range of results. The motion was described in 1911 by Fick, the distinguished German anatomist, as being slight and merely of a rocking type.[31] Weigel in 1955 reported cineradiographic studies.[122] He described a 5-mm ventral shift of the sacrum in relation to the ilium around an axis of motion located about 10 cm below the sacrum. Frigerio and colleagues in 1974 reported a movement of 2.6 cm of the iliac crests in relation to the sacrum.[35] Schunke (1938) observed a pelvic shift when an individual supports his weight on one leg and suggested that there may be sacroiliac joint motion in the stance phase of normal gait.[104]

Walheim and Olerud[119, 120] used an accurate (0.1 mm for translation and 0.1 degree for rotation) electromagnetic measuring technique in which they affixed two pins to the pubic bone on either side of the pubic symphysis *in vivo*. This unique experiment correlated with the radiographic technique of Chamberlain[11] and was successful in recording motion between the two pins during active straight leg raising, hip abduction, and one-leg standing. *In vivo* measurements by Walheim and associates[121] show vertical translations of 2–3 mm and rotations of up to 3° at the pubic symphysis with alternate right and left leg standing. It is of interest that the symphysis motion was the same for males and nulliparous females but was slightly greater in multiparous females.

Following the lead of Olerud, a study was conducted in Sweden to investigate the sacroiliac joint motions in patients in a truly three-dimensional manner using a stereoradiographic technique.[112] At the time of examination, at least four tiny (0.5 mm) tantalum balls were inserted percutaneously, under local anesthesia, into both the pelvis and sacrum. Stereoroentgenograms were taken with the patient in five different positions: (1) supine, (2) prone with hyperextension of the left leg, (3) prone with hyperextension of the right leg, (4) standing, (5) sitting with straight knees. In general, movements were very small (1 to 2° and 0.5 mm to 1.0 mm). The authors concluded that the hypotheses of hypo- and hypermobility causing pain were not supported by the findings of this investigation.

Miller and co-workers[73] studied the kinematics of the sacroiliac joints in eight fresh cadaver specimens aged 59–74. The joints were loaded, and displacements of the sacrum were measured in relation to one or both ilia. The key kinematics findings are expressed in Figure 2-27. Lateral (x-axis) translation was measured at 0.76 mm (standard deviation [SD] 1.41), and anterior (+z-axis) translation was observed to average 2.74 mm (SD 1.07). Lateral rotation

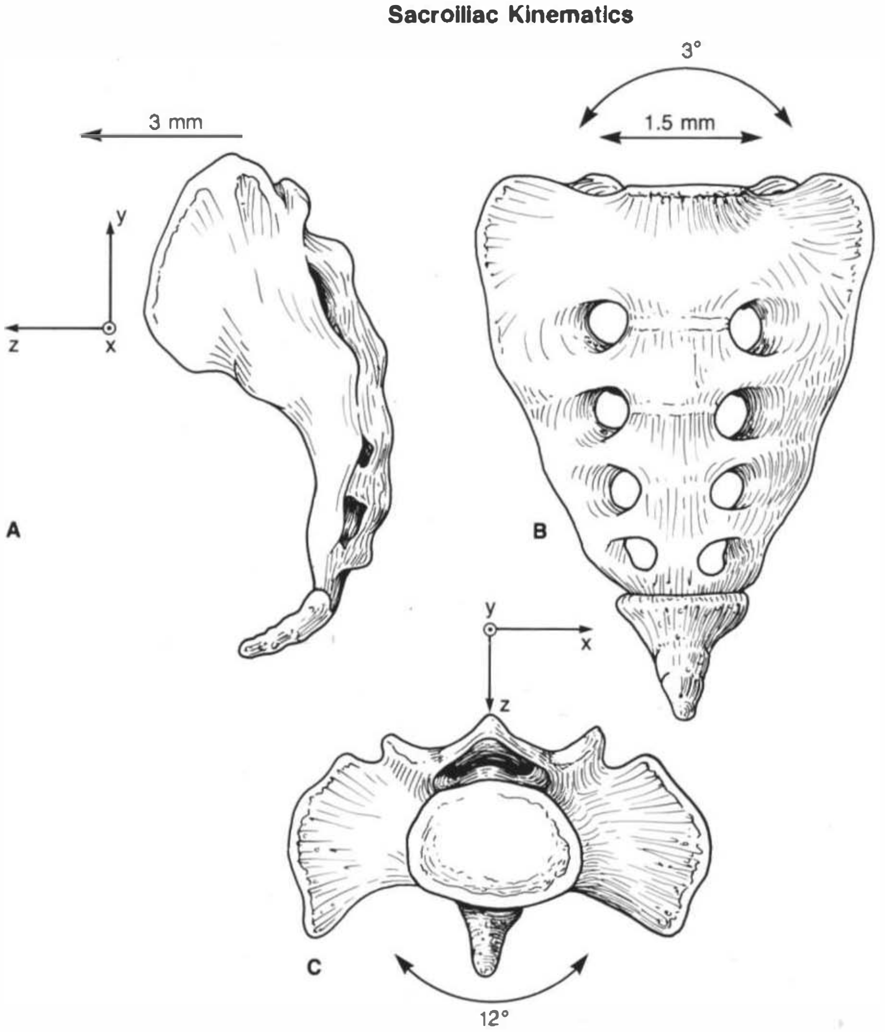


FIGURE 2-27 Representative kinematics of the sacrum in relation to one or both ilia. **(A)** This shows 3 mm of anterior translation (+z-axis). **(B)** Representative z-axis rotation or lateral bending is 1.5° to either side of a complete arch of 3°. **(C)** Representative y-axis rotation is 6° to each side or a total arch of 12°. *(Based on data from Miller, J. A., Schultz, A. B., and Anderson, G. B.: Load-displacement behaviors of sacroiliac joints. J. Orthop. Res. 5:92, 1987.)*

to one side (z-axis) averaged 1.40° (SD 0.71), and axial rotation (y-axis) in one direction was 6.21° (SD 3.29). These specimens were in the older age group and may be skewed so as to show relatively less motion.

Instantaneous Axes of Rotation

Wilder (1980) and associates[130] evaluated topography and the sagittal and frontal plane motion of the sacroiliac joint. There were several interesting observations about this articulation. There was broad scatter of the IARs in both the sagittal and frontal plane motions (Figs. 2-28 and 2-29). Because of the irregular contour of a portion of the joint surface, there must be a separation with enough force to overcome ligamentous resistance. The authors suggested that this mechanical factor may constitute a shock-absorbing mechanism.

Comments

Much more basic research is needed to describe and understand the kinematics and function of the sacroiliac articulation. We believe that currently the best available data on this topic is presented in Figures 2-27 through 2-29. The important question of sacroiliac instability can be answered only by careful investigations that include important quantitative observations based on sound biomechanics.

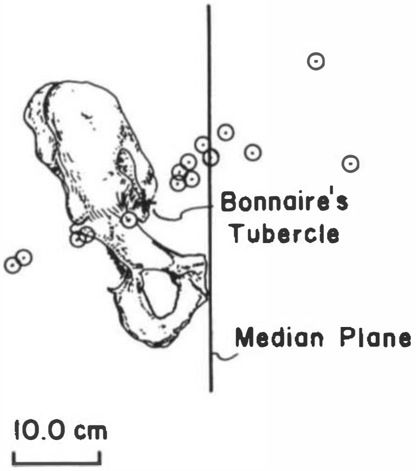


FIGURE 2-28 Front view of right innominate bone with best-fit axes of rotation to show scatter. *(Adapted from Wilder, D. G., Pope, M. H., and Frymoyer, J. W.: The functional topography of the sacroiliac joint. Spine, 5:575, 1980.)*

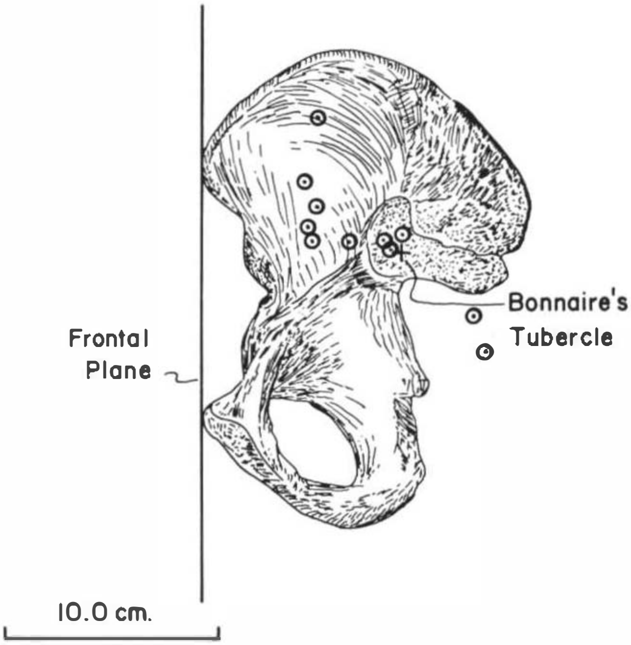


FIGURE 2-29 Sagittal view of right innominate bone with best-fit axes of rotation to show scatter. *(Adapted from Wilder, D. G., Pope, M. H., and Frymoyer, J. W.: The functional topography of the sacroiliac joint. Spine, 5:575, 1980.)*

COMPARISON OF REGIONAL CHARACTERISTICS AND VARIATIONS

This section includes some generalizations about the characteristic kinematics of the four spinal regions. At the risk of seeming to "second-guess" nature and evolution, some etiologic considerations about the engineering design of the human spine are discussed.

Anatomic considerations relating to the spine as a whole are important. It is worthwhile to observe, study, and reflect upon regional characteristics and the spatial orientation of the vertebrae in different regions of the spine, especially in relation to clinical biomechanics of the spine. Much thought and analysis goes into a discussion of the "motion segment." Generally, it is shown with the intervertebral disc parallel to the bottom of the page, and one tends to assume that each motion segment is oriented such that the disc is horizontal. In Figure 2-30, a radiograph in which the subject is standing erect, only a few of the motion segments are parallel or nearly parallel to the horizontal plane. In the normal individual, this occurs in the region of C3–C4, the middle thoracic region (T6–T7), and the lumbar region (L3–L4). In other regions, the plane of the intervertebral disc is not parallel to the horizontal plane. Frequent referral to Figure 2-30 may provide valuable spatial orientation for the study of *in vivo* spine biomechanics.

As the radiographs in Figure 2-30 are studied, it may be useful to review some anatomic factors. In the frontal plane, the spine is straight and symmetrical, with the exception of a slight right convex thoracic curve. This may be due to the position of the aorta.[109] Other investigators suggest that it is due to increased use of the right hand.[17,34] The relation of handedness has been supported by the observation of left convex curvature in left-handed individuals.[45]

There are four normal curves in the sagittal plane. They provide increased axial flexibility with stability and augmented shock-absorbing capacity. The curves are convex forward in the cervical and lumbar region and concave forward in the thoracic and sacral regions. The lumbar curve is slightly more accentuated in women. The dorsal or the thoracic curve is structural and can be looked upon as the persisting curve of the embryonic axis.[34] This curve is the result of an intravertebral variation in height, the anterior height being less than the posterior height.[46] Its convexity is 20–40°. When it exceeds 40° it has been considered abnormal.[100] The thoracic curve has been observed to increase with age. This increase with age occurs at a higher rate for women than for men.[33a] This may contribute both to the observation of the dowager's hump and to the progression of kyphosis with osteoporosis. The lordosis of the cervical curve is due to the wedge-shaped disc and the greater anterior height of the vertebral body. Note that from C1 to L5, the vertebral bodies increase consistently in volume.[46,109] In the frontal plane, the width of the vertebral bodies increases from C2 to L3.[46] In the thoracic spine, the sagittal and frontal diameters are about equal[46] or slightly greater in the sagittal plane.[111]

The spatial orientation of the facet joints in the human spine is shown in Figures 2-22 and 1-19. These structures play a significant role in the characteristic regional variations in the kinematics of the human spine.

The representative ranges of motion for all segments in the traditional planes are given in Figure 2-23, which summarizes the basic kinematics and regional variations.

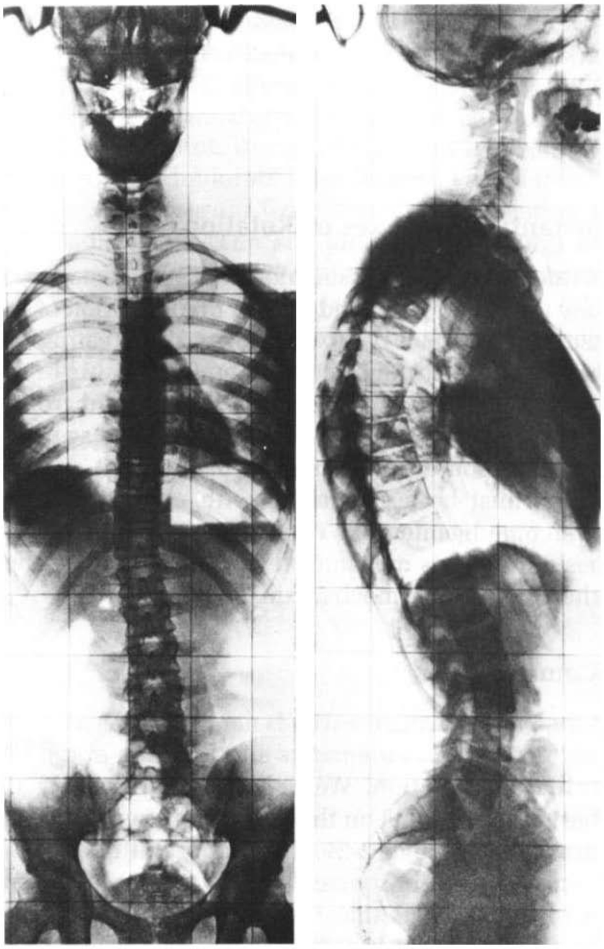


FIGURE 2-30 Standing frontal and sagittal view of the entire human spine. This is a valuable picture that is important to study for several reasons: It gives an accurate account of the relative sizes of the vertebrae in different regions of the spine, and it shows the spatial relationships of all vertebrae and provides a view of the relative curvatures of different regions in various planes. This picture also allows the viewer to better appreciate the spatial orientation of the facet joints in the standing position, and it reminds the viewer that, although functional spinal units are usually depicted with the disc horizontal, only a few of the functional spinal units actually have their discs horizontal in a standing position. *(Schmorl, G., and Junghanns, H.: The Human Spine in Health and Disease. 2nd English edition. Stuttgart, Georg Thieme Verlag, 1968.)*

Occipital-Atlanto-Axial Complex

This region contains the most complex, unique, and highly specialized structures. It is the transition zone between the more standard vertebral design and the radically different skull. The three units maintain structural stability and at the same ti combine to allow sizable quantities of motion in flexion/extension, lateral bending, and especially axial rotation. In order to protect the vital medullary structures in the area, more free space for the spinal cord is present here than anywhere else in the spine (see Fig. 4-21). In addition, an anatomic mechanism for axial rotation has evolved in which the instantaneous axis of rotation is placed as close as possible to the spinal cord, permitting a large magnitude (40°) of rotation without bony impingement upon the spinal cord. In addition to the crucial medullary structures in this area, there is also the unique problem of allowing motion and at the same time achieving and protecting the transport of the vertebral arteries into the calvarium. A large amount of axial rotation is allowed at C1–C2, but there is virtually none between the occiput and C1, where the arteries enter the calvarium. All loads are borne through the occiput and the lateral masses of C1 on to those of C2. These articulations are crucial to structural stability, and without an intervertebral disc, good load-bearing and motion are not possible. Consequently, there is very little or no lateral bending at this articulation.

The axis (C2) is also anatomically unique and is certainly a transitional vertebra with its large spinal canal. The axial rotation is much less at C2–C3, largely because the yellow ligament that starts at this level is much stiffer and more restricting than the lax atlantoaxial membrane that lies at the level above. Moreover, the anatomic complex of the dens and the transverse, apical, and alar ligaments are designed to allow significantly more axial rotation than would be expected of the intervertebral disc and the facet joint.

Middle and Lower Cervical Spine (C2–T1)

The second cervical vertebra is also part of the middle cervical spine. In this region and throughout the spine, stability and mobility must be provided, and at the same time the vital spinal cord and, in the middle and lower cervical spine, the vertebral arteries must be protected. There is a good deal of flexion/extension and lateral bending in this area. These regions have at least one distinct characteristic coupling pattern in which lateral bending and axial rotation are coupled such that the spinous processes point in the direction opposite of that in which lateral bending takes place. Arkin put forth an interesting hypothesis that coupling is due to the mechanics related to soft-tissue tensions.[3] However, the spatial orientation of the facet joints is a more plausible explanation, illustrated in Figure 2-22. Because these joints are oriented at about a 45° angle to the frontal plane, the lateral bending results in axial rotation. For example, during lateral bending to the left, as the left facet of the upper vertebra moves down the 45-degree incline to the left, it is displaced somewhat posteriorly. At the same time, the facet on the right moves up the 45-degree incline, displacing itself somewhat anteriorly. The total effect is an axial rotation such that the left lateral bending ($-z$-axis rotation) results in axial rotation in which the spinous process points to the right ($+y$-axis rotation).

Clinical Aspects

If the shoulders are held in place and the patient is asked to look to the right and to the left ($\pm y$-axis rotation), we have a useful clinical test. If the patient is perfectly normal, there will be little noticeable lateral bending ($\pm z$-axis rotation). There will be a smooth, normal side-to-side gaze. However, if there is an arthrodesis or a fixed subluxation of C1–C2, distinct lateral bending will be noted in association with the axial rotation. This is because with the locked C1–C2 articulation, all of the axial rotation must take place in the middle and lower cervical spine. In this region, there is strong coupling of axial rotation to lateral bending (ratio 0.32) (see Fig. 2-15).[76] Therefore, when the patient axially rotates, there is lateral bending at each FSU from C2 to T1. This useful and interesting clinical sign has been described by Professor B. Jeanneret, St. Gallen, Switzerland.[57a, 57b]

Thoracic Spine

The thoracic spine is a transitional region between the relatively more mobile cervical and lumbar regions. It is thought to be designed for rigidity, to be vital for general, erect bipedal support and protection of the cord as well as the other organs in the thoracic cavity, and to facilitate the mechanical activities of the lungs and rib cage.

First, consider the anatomic and kinematic factors. The upper thoracic spine is similar to the cervical region, and the lower thoracic spine is similar to the lumbar region. The upper thoracic vertebrae are relatively small, similar to those in the cervical region. The spatial orientation of the facet joints is somewhat similar to that of the cervical spine, but the angulation in the sagittal plane is less acute (Fig. 2-22). The spatial orientation of the facets in the thoracic spine changes from the upper to the lower region.[18,111] In a given individual, the spatial orientation of the facet joints may change abruptly to that of the lumbar region anywhere between T9 and T12.[46,101] In the lower thoracic spine the vertebral bodies and discs are larger than in the upper.

In regard to kinematic changes, the upper thoracic spine exhibits more axial rotation than the lower thoracic spine. Axial rotation in the cervical spine is ample, but in the lumbar spine there is relatively little (Fig. 2-23). The lower portion of the thoracic spine, unlike the upper portion, is capable of ample flexion/extension, a motion that is sufficient and gradually increasing in the lumbar spine. The coupling characteristic of the cervical spine is clearly present in the upper portion of the thoracic spine. However, it becomes weaker or changes in the middle and lower portion of the thoracic spine.

Lumbar Spine

The most unique biomechanical characteristic of this area is that it must bear tremendous loads. This is due to the large, superimposed body weight, which interacts with additional forces generated by lifting and other activities that involve powerful muscle forces. In addition, the lumbar spine in conjunction with the hips is responsible for the mobility of the trunk. Together, these impose formidable mechanical demands on the region, and it is not surprising that the lumbar spine may not always be capable of meeting them.

The anatomic location of the fifth lumbar vertebra and the unique spatial orientation of its facet articulation with the sacrum qualifies it as a transitional vertebra. The ample flexion/extension may be attributed to the sizable intervertebral disc.

There is an interesting coupling pattern in the lumbar spine, which is the opposite of a salient pattern in the cervical and upper thoracic spine. In the cervical and upper thoracic regions, with lateral bending, the spinous processes move toward the convexity of the curve. However, in the lumbar and lumbosacral region, the spinous processes may move toward the concavity of the curve.

Clinical Aspects

There are several clinically relevant aspects of lumbar spine kinematics that should be mentioned. In earlier *in vitro* studies, in which effects of injury to the disc were investigated under compression loading conditions, it was found that the injury had only marginal effect on the the biomechanical properties of the functional spinal unit.[70] In a recent study, the opposite has been found.[83] Using three-dimensional loading conditions, increased motions were observed under the application of all physiological loads. The changes were most in axial rotation and least in axial compression. The latter finding may explain the conclusions of earlier studies in which, probably due to limited accuracy of the measuring system, the authors could not measure the small changes produced under axial compression loading. In another *in vitro* study, significantly less motion was observed at the injury site following experimental *subtotal* discectomy in comparison with *total* discectomy.[42] With both subtotal and total discectomy there was increased motion above the discectomy site. Stokes and colleagues,[110] in an *in vivo* biplanar radiographic study, found that disc space narrowing was associated with increased motions, especially the coupled motions. They also observed increased motion, particularly lateral bending, above spinal fusions. Pearcy and colleagues,[89] using biplanar radiographs of live subjects, concluded that when the axial rotation coupling that is associated with flexion or extension is greater than 4°, this is abnormal. Although these observations need more clinical correlations, they serve as a basis for cogent hypotheses.

AGE, SEX, AND SPINE KINEMATICS

Many observations have been made with regard to the decreased mobility of the spine associated with aging. It is not clear whether this is an independent variable, having to do with the spontaneous changes in the mechanical properties of the tissues, or whether it is related to changes that occur as a result of inactivity (using spinal mobility less).

Penning states that cervical spine motion decreases with age.[92] Moll and Wright carried out measurements of thoracolumbar spine motion in 237 normals—119 males and 118 females. Their results showed an increase in spinal mobility from age 15 to 24 and from age 25 to 34, followed by a progressive decrease with advancing age. They also found an interesting sex difference related to spine kinematics. Male mobility exceeded female mobility in the sagittal plane during flexion/extension, whereas during lateral flexion, the converse was true.[75] These findings may challenge the imagination of even the most prudish sexologist.

A recent *in vivo* surface marker television study of the motion of the entire lumbar spine and pelvis in the frontal and sagittal planes showed a decrease in motion with aging.[115]

Galante noted that after age 35 there was no significant change in the tensile properties of the lumbar annulus fibrosus as a function of aging.[37] Tanz, in studies of the lumbar spine, observed a significant motion loss between childhood and adolescence or young adulthood, but he found that the motion loss after age 35 was not significant.[114] This observation is better explained by *intrinsic* tissue changes than by disuse of the full range of motion.

Although the number of women included in his study group was small, Allbrook did not find any sex difference in the range of lumbar spine motion in his study of live subjects. He did, however, note a decreasing range of flexibility with age. This study also showed that there was less movement in the presence of osteophytes at the interspace.[1] Evans and Lissner studied a group of 11 specimens of lumbar spines and pelves and found no apparent relationship between the age of the individuals and the biomechanical properties that affect kinematics.[30]

DISEASE AND SPINE KINEMATICS

One of the primary reasons for studying spine kinematics is to identify and study any changes that may occur in relation to disease.

Reports have suggested that variations in mobility associated with a given interspace are indicative of pathologic changes. A bibliography of these studies is available in the work of Pennal and colleagues.[91]

Increased sagittal plane translation may be an early sign of disc degeneration[48] and low back problems.[62,40,133] Some investigators have not agreed with this observation. Hirsch and Lewin studied the lumbosacral joint and found that disc degeneration did not affect the range of motion of the L5 facet in relation to the sacrum.[53] Rolander showed that there is generally only 1–2 mm of translation in the frontal or sagittal planes. Rolander found no increased translation with degenerative discs.[102] Nachemson and colleagues also found no correlation between the ranges of motion in any of the degrees-of-freedom and the disc degeneration.[78] On the other hand, Panjabi and co-workers found some evidence of increased posterior to anterior coupled translatory motion, during flexion, with increasing disc degeneration.[82] In this *in vitro* study, they also found a similar tendency for the axial rotation to increase, especially the neutral zone. These *in vitro* studies involved direct measurements and observations on the discs.

Mensor and Duvall studied the motion of the lumbosacral joint in 527 consecutive patients in their office. This was done with lateral radiographs in full flexion/extension with the subject sitting. The motion was measured by superimposing the two films and measuring changes in the angle at the lumbosacral junction. The findings were compared with a control group of 94 healthy individuals. Although there is no statistical analysis, the authors observed that while only 15% of the normal control group showed absence of mobility, 43% of the patients with low back pain from a variety of diseases had no motion of L4 and L5.[71]

Jirout and Tanz have noted that 11% and 20% of normals (individuals without back pain) have restriction of spinal movement at the L4–L5 and L5–S1 interspaces, respectively.[58,114] Tanz observed no differences in lumbar motion between normals and those with a history of low back pain. Howes and Isdale reported greater joint mobility in women patients with no specific back pain.[55] However, Sweetman and colleagues found no relationship between sagittal mobility of the lumbosacral spine and past history of low back pain in a study of 500 postmen.[113]

Some hypotheses suggest an increase in motion associated with disease; others propose a decrease. Generally, this type of approach to analysis of disease has not been used, for several reasons. First of all, there is a considerable normal variation. In addition, the measuring techniques in most studies have been cumbersome and of questionable reliability.

A large number of studies have employed a variety of devices, techniques, and methods in an attempt to develop a useful measure of various segments of the spine in order to distinguish between health and disease. None of these appear to be practicable techniques. A description of various methods follows.

In 1924, Cyriax described a torsionometer designed to measure lateral curvature in the lumbodorsal region.[16] Since that time, a number of techniques have been proposed. These include a photographic technique,[49] spondylometers,[21,44,50] inclinometers,[117] skin measurements,[68,118] and recently a combined skin and pendulum method.[75] Finally, there is the combined flexirule/hydrogoniometer.[2] This device combines two methods and gives a record of the curvature of the lumbar spine as well as a numerical figure that represents the mobility of the segments measured. None of these seem to have achieved widespread clinical usage. This type of gross measurement of motion has been used to follow patients with ankylosing spondylitis[50,68] but has not been helpful in evaluating or predicting low back pain.[113]

CLINICAL BIOMECHANICS

- A knowledge of basic normal kinematics of the spine is crucial to the understanding of the pathology, clinical interpretation, and treatment of a large majority of spine disorders. Moreover, an astute analysis and follow-up of the development of new kinematic information provides an important base for creating new insights into the challenging problems of clinical instability and idiopathic spine pain.
- Accurate description and communication of this knowledge requires certain terms, definitions, and conventions.
- Some new observations have been recently made concerning the multidirectional kinematics of the occipital-atlanto-axial joint complex. This knowledge may be helpful in providing better diagnostic and treatment methods for soft tissue problems of this region, such as whiplash.
- There is a large amount (40°) of axial rotation possible at the C1–C2 articulation. If there is compromise of the flow of the vertebral artery, symptoms of basilar insufficiency may result. Stretching and narrowing of the artery due to axial rotation between C1 and C2 may cause this.
- The kinematics of this extremely complex articulation must be carefully studied in order to understand rotary subluxation between C1 and C2. The two key points to appreciate here are that normal axial rotation on an anteroposterior radiograph gives the appearance of abnormal frontal plane motion of C1 in relation to C2, and that true rotary subluxation and dislocation are manifested by an abnormal separation between the dens and the anterior ring of C1 on a true lateral radiograph of C1. This finding can now be definitively evaluated by a CT scan of the upper cervical spine.
- The large amount of motion in the C5–C6 area of the cervical spine, especially in the sagittal plane, may be related to the higher incidence of cervical spondylosis there. Studies of the maximum range of sagittal translation ($\pm z$-axis) in the cervical spine are helpful in the interpretation of radiographs for clinical instability. The observed maximum is 2.7 mm. With radiographic magnification, the suggested upper limits of normal are 3.5 mm.
- In the cervical spine there is a strong pattern of coupling such that axial rotation is associated with lateral bending. When bending to the left, the spinous processes point to the right. This fact is important in the analysis of mechanisms of injury in the cervical spine.
- Excessive lateral bending with axial rotation of the cervical spine during clinical examination suggests C1–C2 fusion or fixed subluxation.
- There is no grossly abnormal motion in the cervical spine under "physiologic loads" as long as either all the anterior elements or all the posterior elements are intact.
- The coupling characteristics in the middle and lower portion of the thoracic spine may have some significance in the pathologic biomechanics that lead to scoliosis. In some instances, the coupling pattern has an axial rotation associated with lateral bending, which is the same as that seen in the deformity of scoliosis.
- In the lumbar region, a good deal of attention has been directed to the possibility that in some states the range of motion, the pattern of distribution of the instantaneous axes of rotation, or some other kinematic measurement is altered by disease processes. At present, none of these have been proven to be evidence of disease. Well-presented documentation of *in vitro* changes in the instantaneous axes of rotation associated with disease has not resulted in widespread clinical usage. However, future studies may demonstrate a distinct correlation between abnormal patterns and clinical instability and pain.
- Analysis and comparison of the characteristic kinematics in different anatomic regions of the spine are helpful in the analysis and evaluation of spine trauma. The hypermobility of the cervical spine has been implicated in whiplash and related problems. The less mobile thoracic spine, with the abrupt tran-

sition between it and the lumbar spine, has important mechanical influences in the fracture patterns in these regions with regard to axial rotation and general mobility.

- Recent biomechanical studies of the cervical and lumbar spine have shown kinematic and kinetic differences in regions of the spine that have previously been considered homogeneous. The clinician should consider that, biomechanically, the middle and the lower parts of the cervical spine are different; the upper, middle, and lower portions of the thoracic spine are different; the L4–L5 region is different from the upper lumbar spine; and the lumbar and the lumbosacral spines are different.
- Larger segments of the spine and the entire spine have been studied extensively to determine the effects of age and sex on mobility. There is generally a decrease in motion with advancing age, but there is no uniformity of opinion concerning this issue.
- Except in patients with ankylosing spondylitis, the clinical measurements of mobility of large segments of the spine have not been particularly useful.

NOTES

[A] The problem of determining the instantaneous axis of rotation and the angle of rotation for a certain rigid body undergoing plane motion may be solved by several different methods. A simple graphical method is described in Figure 2-4 on p. 89. This concept is also described in Chapter 9.

If the telephone were simply rotated about point C, through an angle equal to $\theta°$, the point A_1 is displaced to A_2, and similarly the point B_1 to B_2. Thus, a pure rotation about the point C has displaced the telephone from position 1 to position 2. We call the point C the instantaneous axis of rotation of this body motion, and θ is the angle of rotation.

Since the motion discussed takes place in a plane, and the instantaneous axis of rotation is a line, it has a point of intersection with this plane. This point of intersection is called the center of motion and is synonymous with the instantaneous axis of rotation.

[B] The state of the science of spine kinematics is really in a rather curious situation in regard to the instantaneous axes of rotation and helical axes of motion. The strict concept of IAR is really applicable only to motion that is limited to one plane. We know that vertebral segment motion is rarely uniplanar. Nevertheless, it has been studied in that way for several reasons: the anatomic tradition of describing motion in three planes—flexion/extension, lateral bending, and axial rotation; the use of radiography, which provides only uniplanar analysis of complex motion; and the clinical tradition of thinking of the motion in terms of rotation in three perpendicular planes. Thus, when we speak of IAR, we are generally talking of a complex motion and studying it in a simplified uniplanar view.

On the other hand, the helical axis of motion may be used to precisely describe the three-dimensional movement of one vertebra in relation to its subjacent fellow. However, at present this is of virtually no practical importance because the information is not available, nor is the clinical setting such that it would be accepted and used at the present time. However, with sophisticated computer graphics, three-dimensional imaging methodologies, and robotics on the clinical scene, the helical axis of motion may soon find its way to clinical relevance.

[C] Although the top angle is a satisfactory term, the radius of curvature may be a preferable method of indicating and quantitating a curve or arch. Radius of curvature is described in Chapter 9.

[D] Even studies from which instantaneous axes of rotation have been calculated have varied considerably in principle and technique. Here is a review of some of the salient considerations.

A given instantaneous axis of rotation depends upon the structure as well as the type of loading. In other words, it is not sufficient to say that, for example, vertebra T3 has an instantaneous axis of rotation, with respect to its subjacent fellow, located 3 mm anterior and 23 mm caudal to the anatomic center of its body. Although it is a precise physical location, it is not sufficient. In addition, we must also specify the type of loading, which in itself is an ambiguous term. Let us take, for example, flexion loading. Although everyone understands what is meant by flexion (i.e., forward bending), when it comes to simulating this loading in cadaveric experiments, different researchers use different types of loading. Figure 2-16 shows some versions of flexion loading (see p. 101). A compressive force placed anteriorly was used by Rolander,[102] when he studied the lumbar spine, and White,[124] who studied the thoracic spine. Flexion can also be created by applying a moment about a horizontal axis, as shown in Figure 2-17. This method was used by Markolf[69] and Panjabi.[84] Figure 2-16 also shows a method in which a horizontal force is applied to the center of the vertebral body and directed forward. This method was used by Panjabi and colleagues.[84] Another type of loading has been used by Lysell[87] and White[124] in experiments where multiple segments are employed with the lowest vertebra fixed. A transverse load is applied to the top vertebra to produce flexion. In this type of loading, all intermediate vertebrae are subjected to varying amounts of load. The top vertebra is loaded through a pure transverse force. All intermediate vertebrae are loaded with a combination of the transverse force, which is the same at all levels, and bending moments, which are increasing in magnitude in a cephalocaudad direction. Since the instantaneous axis of rotation is a function of the type of load applied, it is clear that, for the load depicted in Figure 2-16, the instantaneous axis of rotation calculated for each vertebra is going to vary because of the different combinations of force and moment at different levels. This concept applies equally to extension and lateral bending. With these loading considerations in mind, one should be quite circumspect in comparing results when there are such crucial differences in experimental techniques. Studies of the instantaneous axes of rotation in axial rotation have been treated more consistently by various investigators. However, it has been documented that axial rotation and lateral bending are coupled and are part of three-dimensional motion. Therefore, such a motion does not have an instantaneous axis of rotation but, instead, a helical axis of motion.

[E]Other things being equal, rotatory motion can be expected to vary directly with the height of the disc, with the disc material, and with the fourth power of the disc diameter, and inversely with its stiffness. In lateral bending, the disc height is therefore relatively less important than it was in the case of flexion/extension.

For the moment, disregarding the effect of the posterior elements and representing a motion segment by two rigid bodies connected by a cylindrical disc, the following trends can be projected for the motion as a function of the height, diameter, and material of the disc. According to the simple engineering theory of strength of materials, the rotation θ produced by application of a bending moment M (flexion/extension and lateral bending) is

$$\theta = \frac{64M}{\pi}\,\frac{H}{ED^4}$$

where H = the disc height
D = the diameter
E = the modulus of elasticity of the disc material
$\pi = 3.14$

This formula represents axial rotation θ produced by the axial moment (torque) M if G replaces E (G = the modulus of shear of the disc material) and 32 replaces 64.

[F]If the reader examines the study by White,[124] note that the coordinate system in the experimental investigation is not the same as the system used here.

[G]The ranges of motion are often used clinically to evaluate the patient for spinal instability.[22,23,24,51,94] Therefore, it is necessary not only to know the average values for these quantities but also to have a good measure of the variation present in the population. Several methods have been used to obtain these measurements. It would seem that the *in vivo* measurements would provide the most clinically useful data. However, such measurements are less reliable for several reasons. First, the measurements are obtained from roentgenograms taken of the spine in two postures. To obtain the baseline data for comparison with patients, normal subjects must be exposed to x-rays,[89,93] thus limiting the quality of the data base. Second, the accuracy of the in *vivo* measurements is often lower than what could be achieved in a laboratory setting using a more invasive, and often a more accurate, technique. Third, the *in vivo* measurements may significantly depend on the cooperation of the subjects. This is most important in the case of active measurements made of patients. This method happens to be the most common. The patient is asked to position himself in a fully flexed and then in a fully extended position. Lateral x-rays are taken in each position. Because of pain or other reasons, the subject may not fully extend or flex the spine, thus compromising the quality of the data. Recently, passive measurements have been advocated.[22,23,24] Here, the examiner, using appropriate protective measures against radiation exposure, bends the spine manually, either by hand or by applying a defined load, until the spine achieves full flexion or extension. This passive examination may be facilitated by giving a muscle relaxant to the patient. It has been shown that the passive examination is more reliable and sensitive than the active examination, especially for determining the hypo- and hypermobility of the spine.[24]

Against this background there is the in *vitro* method in which fresh cadaveric spine specimens are obtained and loads are applied to produce natural spinal movements.[78,80,81,85] The main advantage of this method is its high accuracy of load application and motion measurement. However, in the past, the methodology has involved applying preconditioning loads (e.g., applying full flexion load, removing the load, and then again applying full load). Three load/unload cycles are often used. The motion measurements are made after the specimen has been preconditioned. The process is repeated for the extension load. The two motions are then added to obtain the total range of motion. There are certain inherent problems with this type of approach. Because the flexion and extension loads are preconditioned, a certain movement between the two preconditioned states is not accounted for. Thus, most of the older *in vitro* data show ranges of motion that are smaller than the *in vivo* measurements. However, if the additional motion between the two preconditioned states, also called the neutral zone is added, then the *in vitro* ranges of motion are more closely matched to the in *vivo* measurements. Therefore, it is recommended that *in vitro* experiments be conducted in which the neutral zones are measured and accounted for while the ranges of motion are determined (see Chap. 1). Table 2-5 is provided as a synopsis of the various kinematic data published on the cervical and lumbar spine. The authors used this data and analytical judgment to arrive at the "representative" rotations presented in the kinematic tables (Tables 2-1–2-4) in this chapter.

[H]Translation motion of a vertebra is often given imprecisely. Unlike the angle of rotation, which is the same for all parts of a rigid body, the translation measurement is different for different points of the moving body. This is easily explained considering the instantaneous axis of rotation (IAR) concept. By definition, an IAR has zero translation (see Chapter 9, Biomechanics A to Z). A point on the rigid body has translation that is directly proportional to the distance of the point from IAR. As different points on a moving body have different distances from IAR, the points will also have different amounts of translations. This implies, now turning back to the vertebra, that the anterior-inferior corner of the vertebra will have a different translation from that of either the inferior-posterior or superior-posterior corners. Thus, to state that a vertebra translates anteriorly, say 3 mm, is not justified. It is essential that we state which point of the vertebra translates 3 mm anteriorly.

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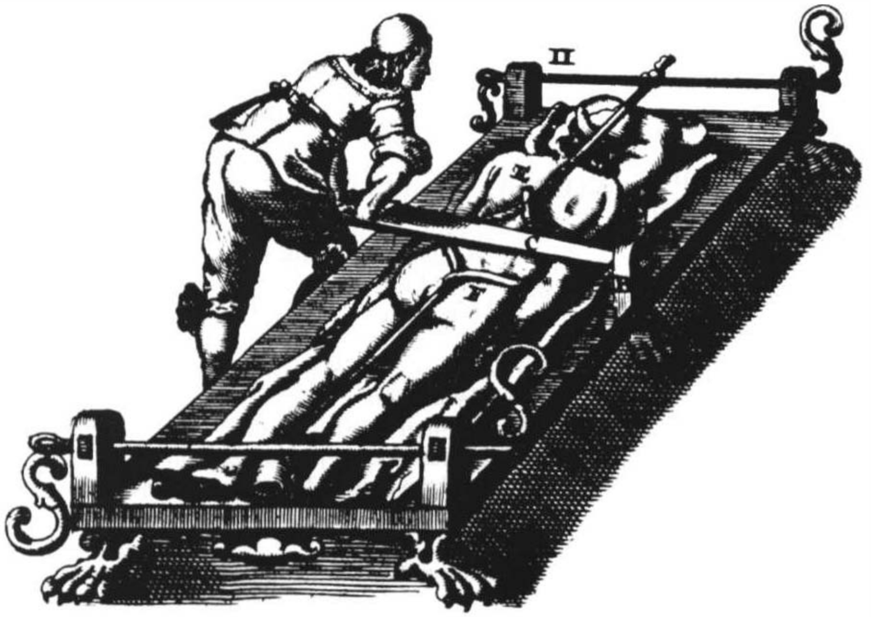
Practical Biomechanics of Scoliosis and Kyphosis

Figure 3-1. This illustration, taken from Scultetus' The Surgeons Store-House, shows that "biomechanics" has been involved in the correction of spine deformities for several centuries. Axial traction, pelvic traction, transverse loading, three-point bending, viscoelastic creep, and relaxation are all well demonstrated in this illustration. (Scultetus: The Surgeons Store-House. p. 63. Printed for Starken at The Miter on Fleet Street, near Temple Bar, London, 1674. Courtesy of Yale Medical Library.)



3

PART 1: SCOLIOSIS

ANATOMIC CONSIDERATIONS

Normal Curves

In the frontal plane, the normal spine appears straight and symmetrical, with the exception of a very slight right thoracic curve. This may be due to the position of the aorta.[111] Other investigators suggest that it is due to increased use of the right hand.[26, 39]

The Facets

Humphry pointed out that movements in the spine are possible mainly because of the shape and position of the articulating processes of the diarthrodial joints.[52] It is the orientation and position of these joints in space that influence the mechanics of the spine (see Figs. 1-19, 2-22). This is important to keep in mind when the phenomenon of coupling is considered. In the thoracic spine, the superior facet is almost flat and directed backward, a little laterally, and upward. The inferior facet is directed forward, slightly downward, and medially.[45] The orientation of the facets in the thoracic spine may be related to the irregular pattern of coupling found in this region. The significance of this is discussed on page 135.

The Ligaments

Little is known about the physical properties and activity of the ligaments in scoliosis. Walters and Morris carried out *in vitro* studies to compare mechanical properties of the interspinous ligaments in subjects with idiopathic scoliosis to those in subjects with scoliosis of known cause. No differences were observed.[121]

Nordwall compared mechanical properties of interspinous ligaments and tendons of the erector spinae muscles in patients with idiopathic scoliosis, patients with scoliosis of known cause, and patients with spondylolisthesis only. No significant differences were found.[82]

A good deal has been written concerning the mechanical importance of the ligamenta flava (yellow ligaments). Rolander credited them with a major role in restricting or dictating the kinematics of normal motion.[101] The following considerations involving the yellow ligaments may have clinical relevance to scoliosis. The yellow ligaments and facets have been shown experimentally to limit the amount of axial rotation in the normal thoracic spine.[126] Also, hemilaminectomy, which releases the "checkrein" force of yellow ligaments, can result in experimental scoliosis.[76]

More recently, the biomechanical characteristics of the disc in scoliosis have been studied. Brickley-Parsons and Glimcher[14a] observed differences in the distribution of Type I and II collagens on the convex and concave regions of the intervertebral disc in scoliosis patients. Because of expected differences in compressive loads on the concave and convex sides of the disc, the investigators speculated that this may be a reflection of Wolf's law in the collagen chemistry of the disc.

NORMAL KINEMATICS

> . . . It is as if one undertook, for example, to investigate a railroad accident solely from a study of the wrecked cars. Much could be learned as to the effect and direction of the destructive forces, the amount of force expended, and the kind of damage done, but more could be learned and future accidents could be better prevented by a study of the normal running time of the trains, their proper relation to each other at the time of the accident, and by an investigation of the signal system and the routine precautions adopted.
>
> (LOVETT. THE MECHANICS OF THE NORMAL SPINE IN RELATION TO SCOLIOSIS, 1905)

Lovett termed his study "mechanics." This discussion focuses primarily on the kinematics of the thoracic spine. The term *kinematics* is defined in Chapter 2.

Kinematics of the thoracic spine has been studied using two well-developed experimental methods. One method analyzed motion segments by applying known forces and measuring displacement with electrical recording devices. This technique provided two-dimensional analysis. Controlled studies were performed with this technique to compare the motion with and without the posterior vertebral elements. A detailed account of this experimental method is available in the literature.[126] The other method employed steel balls as markers and analyzed larger segments of the spine. Radiographs were taken of the vertebrae at different points in the characteristic ranges of motion. From this experiment it was possible to calculate precise three-dimensional kinematics of the spine. A detailed explanation of this technique may be found in a thesis by Lysell.[67]

Degrees of Freedom

This concept, which is basic to the understanding of kinematics, is defined in Chapter 2. A vertebra in motion thus has six degrees of freedom for its movement. It may translate along or rotate about any of the three axes described in Chapter 2. Until recent years, for clinical and experimental purposes, the motion has been measured in only two dimensions. The tendency has been to describe this motion in each of the three traditional planes—frontal, sagittal, and horizontal. Most analyses have dealt with only those components of the translation or rotation vectors which lie in one of these planes. This has been done most often with radiologic studies, taking into consideration only the rotary components. Such an analysis belies an accurate description, because there is almost always an element of translation along with rotation. Moreover, various motions occur in planes other than the traditional anatomic planes. This oversimplification is present in clinical measurements of the relative positions of the vertebrae in scoliosis; the deformity and the measurements are depicted in the single plane of radiographic film.

Coupling

This is applied to motion in which rotation or translation about or along one axis is consistently associated with rotation or translation about or along a second axis. Some interesting questions concerning this phenomenon in the normal and scoliotic spine follow.

In the cervical spine and the upper portion of the thoracic spine, there is a relatively marked and consistent coupling of *axial rotation with lateral bending*. The direction of coupling is such that axial rotation of the vertebral body causes its anterior aspect to point toward the concavity of the lateral bending curve. In other words, the spinous processes point more to the convexity of the physiologic curve.[A] In the middle and also in the lower regions of the thoracic spine, this same pattern is still present and probably predominant. However, in these areas it is not as marked, nor is it consistently present. Furthermore, the direction of the coupled axial rotation in the middle regions is in some cases the reverse of that of the cervical and upper thoracic spine.[B] These observations relate to some possible etiologic mechanisms (see p. 135).

Posterior Elements

Under controlled conditions, the mechanics of the functional spinal units were studied with and without posterior elements.[126] The removal of these structures resulted in significant increases in the amount of axial rotation. The posterior elements set a limit on axial rotation in the normal spine. Most probably, their release allows for considerably more correction of the abnormal axial rotation in the scoliotic spine. It is of interest to note that other investigators, using a computer model of the spine, showed that surgical ablation of the intervertebral disc (the anterior elements) significantly reduced the resistance of the spine to correction.[105] All of these points are mechanically sound. When or if they should be used clinically depends upon surgical judgment and technology. Various transverse loading devices have been developed that assist in the correction of scoliosis and do not require surgical release of any anterior or posterior elements.

Helical Axis

A truly three-dimensional analysis involves a description of the motion in terms of the equivalent helical motion of the vertebrae in space. This equivalent helical motion is a superimposition of rotation and translation about and along the same axis—the helical axis of motion. This axis has the same direction as that of the rotation axis (see Fig. 2-5). It is important to include accurate three-dimensional considerations, which allow the relationship of each vertebra to its subjacent fellow to be precisely defined. Although this concept as applied to scoliosis is valid from an engineering perspective, it does not appear to be necessary in regard to current clinical practice. The practical relevant clinical parameters do not involve this level of precision.

BIOMECHANICAL DEFINITION OF SCOLIOSIS

Scoliosis is defined as an appreciable lateral deviation in the normally straight vertical line of the spine. Since the ultimate effect of the disease is an extensive alteration in the mechanical structure of the spine, a biomechanical definition of the disease is necessary. There is abnormal deformation between and within vertebrae, too much curvature in the frontal plane, too much vertical axis rotation in the wrong direction, and not enough curvature in the sagittal plane (*i.e.*, a loss of normal kyphosis or a relative lordosis). It should be emphasized that there is also deformity that may not be recognized in the analysis of the traditional orthogonal planes.[125, 126, 127]

In other words, the relative position of vertebrae in regions of the spinal column is abnormal, and deformation within an individual vertebra is abnormal. There is too much curvature. Instead of a straight spine in the frontal plane (x, y plane) or the subtle, right physiologic curve, there is an exaggerated curvature in the frontal plane. The curves are in the wrong plane. Generous curves in the sagittal plane are normal. (There is normal cervical and lumbar lordosis and thoracic kyphosis.) The axial rotation is in a direction opposite of what would be expected from the physiologic coupling between lateral bending and axial rotation. In scoliosis, there is considerable deformation within a given vertebra. There may be wide pedicles on one side and a short pedicle on the other. The transverse processes may be asymmetrical in their spatial orientation. The spinous process may be deformed and bent out of the midline. The laminae and the vertebral bodies are asymmetrical (Fig. 3-2).

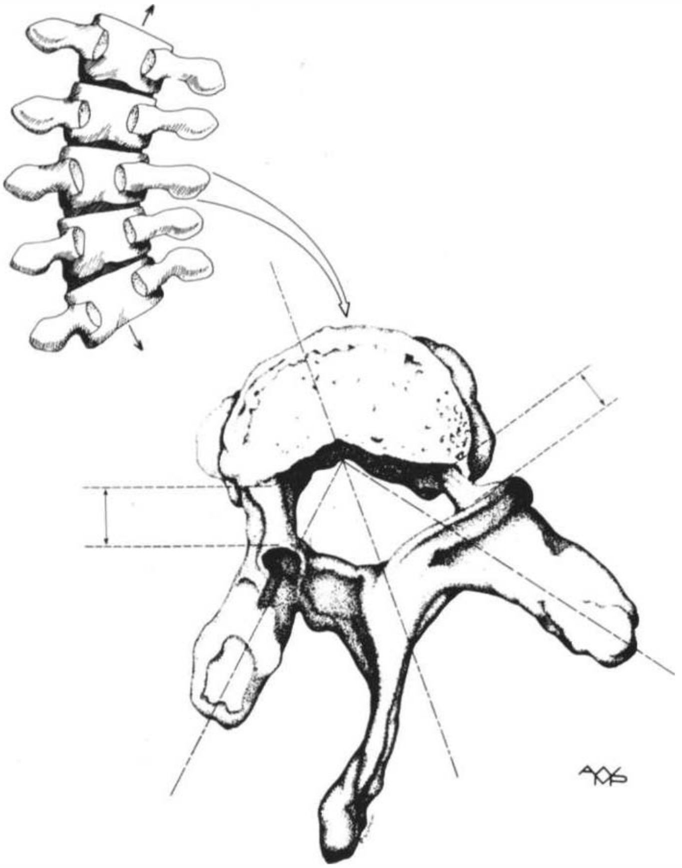


FIGURE 3-2 This diagram emphasizes that in scoliosis there is deformity *within* as well as *among* vertebrae. Recent studies by Sevastik, et al.,[107] show that deformation within the vertebra does not occur in curves with Cobb angles of less than 40°. Note the abnormal configuration and spatial orientation of the pedicles. Consider the potential value of studying this in the preoperative preparation and planning activities in which transpedicular fixation is to be employed.

ETIOLOGIC CONSIDERATIONS

Biomechanical Classification

There is a long list of known causes, conditions, and diseases that are associated with scoliosis. There are several methods of classification. A biomechanical classification is provided here and may best be appreciated in the following context. The spine remains normal because of the maintenance of a delicate and precarious balance. This balance depends on a precise functional status and dynamic symmetry, the key elements being the bony structure, the ligaments, the intrinsic neuromuscular mechanics, and, finally, the general balance and symmetry of the body. Scoliosis can result from either gross or subtle disruptions of the delicate balance. The diseases listed are not exhaustive for each category. There is also some overlap; a given disease may contribute or be presumed to contribute to the imbalance through more than one mechanism.

BIOMECHANICAL CLASSIFICATION OF SCOLIOSIS

Alterations of Intrinsic Osseous Structures

Abnormalities of Material Properties of Support Structure

Rickets (primary and secondary)
Osteogenesis imperfecta
Neurofibromatosis
Infections or tumors

Abnormalities of the Geometry of the Support Structure

Hemivertebrae
Maldeveloped vertebrae
Myelomeningocele
Asymmetrical spina bifida
Asymmetrical lumbosacral vertebral structure and articulation
Fractures and dislocations
Various surgical procedures
Postirradiation (vertebral end-plates)

Abnormal Regional Kinematics

Congenital unilateral bars
Partial failures of segmentation
Asymmetrical sacralization of fifth lumbar vertebra
Fractures and dislocations
Surgery

Alterations of Intrinsic Ligamentous Structures

Marfan's disease
Mucopolysaccharidosis
Myelomeningocele
Arthrogryposis
Surgery

Alterations in Static or Dynamic Balance

Neuromuscular Static Balance

Polio
Myelomeningocele
Syringomyelia

Neuromuscular Dynamic Balance

Cerebral palsy
Friedreich's ataxia
Muscular dystrophy

Postural Dynamic Balance

Abnormalities of vestibular apparatus
Visual disturbances
Torticollis
Leg-length discrepancies

Thoracic Static Balance

Rib removal (thoracoplasty): ipsilateral convexity
Excessive thoracic scarring: contralateral convexity

Congenital Scoliosis (Deformity Intrinsic to Body)

Infantile type
Sprengel's deformity
Klippel-Feil syndrome
Multiple congenital anomalies

Idiopathic Scoliosis

Miscellaneous Forms of Scoliosis

Experimental and Clinical Studies

The cause of 85–90% of cases of scoliosis is unknown. Idiopathic scoliosis occurs in an otherwise healthy child, often associated with a familial history of the disease. Numerous hypothetical, etiologic explanations are offered. From a mechanical point of view, the hypothesis should explain the cause of the abnormal curvatures, the abnormal rotation, and the forces necessary to cause deformation within a vertebra.

One of the major experimental thrusts has been to establish some imbalance in the neuromuscular and osseous ligamentous structures of the spine in experimental animals, the assumption being that imbalances that result in a scoliotic pattern may be sought as potential etiologic factors in idiopathic scoliosis. This presumes that scoliosis is caused by the weakness or absence of a structure on the convex side of the curve or an overactivity of its antagonist on the concave side. A large number of anatomic elements have been studied in rabbits and pigs. Table 3-1 is a comprehensive list of various surgical procedures that have produced scoliosis. All operations were unilateral, performed at five levels.

Michelsson listed operations that most consistently induce scoliosis.[76] They were (1) resection of the dorsal ends of the ribs, (2) hemilaminectomy, and (3) transection of the posterior costotransverse

TABLE 3–1 Michelsson's Experimental Surgical Scolioses

	Rabbits			Pigs		
Operation	No. of Animals	Funct. Scoliosis	Struct. Scoliosis	No. of Animals	Funct. Scoliosis	Struct. Scoliosis
Transection or Resection of Muscles and/or Ligaments						
Transection of the costotransverse ligaments	59	+	±	7	+	+
Transection of the ligaments of the heads of the ribs	38	+	+			
Transection of the ligaments attached to the dorsal ends of the ribs	17	+	+	2	+	+
Transection of the intercostal muscles	9	+	±	3	+	+
Transection of all muscles and ligaments between the transverse processes and between these and the arches in the lumbar spine	8	+	+			
Operations on Nerves						
Transection of the intercostal nerves	13	+	±			
Operations on Bones						
Hemilaminectomy in the thoracic spine	56	+	+	2	±	±
Hemilaminectomy in the lumbar spine	9	+	+			
Transversectomy in the thoracic spine	7	+	±			
Transversectomy in the lumbar spine	2	+	+			
Resection of the heads of the ribs	14	+	+	1	+	+
Provoked epiphysiolysis of the heads of the ribs or transection of the necks of the ribs	16	+	±			
Resection of the dorsal ends of the ribs	125	+	+	25	+	+
Transection of the thoracic wall lateral to the tubercles of the ribs	8	+	+			
Resection of the ribs laterally of their tubercles	25	+	+			

(Modified from Michelsson, J.: The development of spinal deformity in experimental scoliosis. Acta Orthop. Scand., 81 [Suppl], 1965.)
+ = Scoliosis usually resulted.
± = Persistent structural scoliosis sometimes resulted.

ligaments. The factor that is common to both (1) and (2) is (3). Michelsson suggested that the posterior costotransverse ligaments are crucial in maintaining equilibrium and symmetrical growth in the spine.

When one finds a consistent unilateral alteration resulting in scoliosis, the assumption is that with a growing spine, the initial, functional scoliosis ultimately develops into a structural deformity. This is explained on the basis of what has been traditionally called Heuter-Volkmann's[119] law. The theory suggests that increased pressure across an epiphyseal growth plate inhibits growth, whereas decreased pressure across the plate tends to accelerate growth. This theory purports that, on the concave side of the curve, the epiphyseal plates have abnormally high pressures that result in decreased growth, whereas on the convex side of the curve the pressures are less, resulting in accelerated growth. These two factors contribute significantly to vertebral asymmetry. Work by Stillwell on monkeys nicely supports this hypothesis.[112]

Two experiments involved the fixation of the spine in a curved position and fixation of the spinous processes of the vertebrae. The first resulted in occasional scoliosis, and the second resulted in severe scoliosis with lordosis and rotation.

Another etiologic consideration related to mechanics is asymmetrical radiation of the spine, resulting in curvature due to changes in the epiphyseal growth plates, with either unilateral stimulation or unilateral ablation.

Experimental scoliosis has also been produced by radiologic exposure of the growing spine, induction of lathyrism, oxygen deficiency, unilateral labyrinthine stimulation, and unilateral labyrinthine ablation.[76,94] The very broad variety of experimental variables that have resulted in a "scoliotic" deformity suggests that the maintenance of a normal spine in a growing animal is dependent upon a delicately balanced, easily disrupted equilibrium.

This general experimental approach is questionable, because the goal is to explain idiopathic scoliosis *in man*, an erect, biped organism. The frequency, duration, direction, and magnitude of the loads are significantly different in the pig, dog, rabbit, and mouse. There are obvious differences in the anatomy of the spine in animals. Subtle anatomic factors, such as facet orientation, can significantly alter the mechanics of the spine. Thus, there is yet another limiting factor in the use of animals to study scoliosis in man. Using quadrupeds as experimental prototypes is, no doubt, valuable but should be viewed in this perspective.

Experimental Studies Update

This section includes the author's views of some of the more cogent experiments that relate to the etiology of scoliosis.

Lawton and Dickson[62] have studied New Zealand white rabbits to develop an experimental scoliosis. They created in these animals pure frontal plane deformities, pure sagittal plane deformities, and a combined sagittal and frontal plane deformity that was called biplanar. They noted that neither pure scoliosis (frontal plane deformity) nor pure lordosis (sagittal plane deformity) resulted in progressive scoliosis. However, all 20 animals given the combined deformity involving both planes developed scoliosis. Moreover, they noted that if the two-plane deformity was released before maturity, the deformity spontaneously improved. The investigators state that this experiment supports the view that the etiology of idiopathic scoliosis is the anterior elements growing faster than the posterior elements, causing a loss of normal kyphosis and a buckling of the anterior elements (vertebral bodies) outward laterally. This creates the scoliosis. The investigators purport that partial correction of the frontal plane deformity alone does not address the important loss of normal kyphosis (lordosis deformity). This is an important clinical biomechanical consideration that requires additional thought and investigation.

In some support of the work of Lawton and Dickson[62] and the hypothesis of Roaf[99] is the work of Öhlén and associates.[85] They studied 127 patients with idiopathic scoliosis and noted that they did have less thoracic kyphosis than normal.

Hakkarainen[46] studied 253 growing rabbits in an experimental design that involved immobilizing the animals in a cast that produced a scoliotic curve of the spine. The animals were left in the cast for 2–5 weeks. The results were that 85% of the animals became scoliotic, and 52% of those animals had either permanent or progressed scoliosis. The curves that were less than 30° at the time of cast removal did not progress. On the concave side of the curve there were shortened intercostal muscles and evidence of growth disturbance.

These two studies show that a variety of changes in the normal balanced mechanical relationships can result in either temporary or permanent deformity. Attempts to compensate for or re-establish that delicate balance are sometimes successful and sometimes not. The type, degree, extent, and duration of the imbalance are also factors that influence the probability of return to a normal or new balance, either *spontaneously* or as a result of some *intervention*.

There has been considerable interest in a variety of neurophysiologic abnormalities associated with scoliosis. These have been comprehensively summarized by Yamada and colleagues.[132] The work of Pincott and Tafts[92] is very interesting because, unlike most animal studies, theirs was done with primates. This experimental scoliosis was actually an incidental finding. Monkeys were injected intraspinally with live *attenuated* oral poliomyelitis vaccines. Some of the animals developed scoliosis. These animals were found to have damage on the convex side of the spinal cord, particularly in the posterior horn and posterior central gray matter (Clarke's column). The scoliosis was thought not to be due to poliomyelitis, because the anterior horn was not involved. Thus, these data are interpreted as supportive of the theory that asymmetrical weakness of the paraspinal muscles can be due to loss of proprioceptive innervation.

Another relatively rare study in primates by Pincott and associates[91] has shown that resection of the dorsal spinal nerves can create a scoliosis convex to the side of the resection. This supports the hypothesis that scoliosis can be created by asymmetrical spinal muscle weakness due to loss of proprioception.

Yamada and co-workers[132] emphasized that virtually any disruption of the postural reflex system can result in scoliosis. The imbalance may be in the afferent system either primarily or secondarily, as suggested by the preceding experiment. Also, the disruption may be due to disruption of the afferent system. The authors also indicated that there is clinical and experimental evidence that brain stem dysfunction may contribute to the etiology of scoliosis.

These studies show that (1) one can produce scoliosis through the creation of pure sagittal plane lordosis, (2) scoliosis can also be produced through the production of a frontal plane deformity, (3) subtle muscle imbalance secondary to the alteration of sensory input can create scoliosis, and (4) disruptions of postural reflexes probably can produce scoliosis. At the time of this writing, the experimental data support the trend of many possible causes of "idiopathic" scoliosis rather than one common cause.

Clinical Studies Update

The work of Yamada and colleagues[132] summarizes well the clinical evidence for disruption of postural reflex as a cause of scoliosis. The publication by Sevastik and associates[107] provides a thorough review of the literature. This work emphasizes two factors that may subtly alter the delicate balance that can yield to what sometimes seems like a propensity to develop scoliosis. The asymmetric growth of the ribs (*i.e.*, increased longitudinal growth on the concave side) may be the cause of right convex thoracic scoliosis. It was also suggested that the more pronounced vascularization of the often larger right breast may stimulate enough growth of the underlying right costal cartilage to upset the delicate balance of forces acting on the normal spine. This study also noted with computerized tomography (CT) studies that with Cobb angles of <40° there was no evidence of asymmetry of vertebral bodies, pedicles, laminae, or transverse processes in transverse sections of the *apical* vertebra.

Yarom and Robin,[133] in a study of spinal and peripheral muscles in patients, noted some abnormalities in the concave side of the curve. They reported a mild Type I fiber atrophy and a generalized tendency toward small myofibrils. This suggests a generalized muscle disorder with asymmetrical changes upsetting the delicate balance.

A not-so-subtle form of imbalance is the scoliosis seen in myelodysplasia. Here, there is gross muscle imbalance and paralysis, as well as incomplete and asymmetrical posterior element structural imbalance.[90]

Another not-so-delicate disequilibrium of the delicate balance of a normal growing spine is created by chest wall resections in children. Derosa[30] noted that children with *posterior*, not anterior, chest wall resection developed progressive curvature with the convexity always to the normal side. This clinical study fits neatly with the animal investigations of Langenskiöld and Michelsson,[60] who found that among the 15 operations that create experimental scoliosis, the one best able to do so was paravertebral rib resection.

Mayfield and associates[70] have completed clinical studies that show another mechanism for the development of scoliosis. Children treated with orthovoltage radiation (>3,000 rads) for neuroblastoma may develop scoliosis or kyphosis from various patterns of location of epiphyseal arrests. This can result in a variety of patterns of progressive deformities.

The issue of leg length difference as an etiologic factor in scoliosis has been studied.[42, 89, 9] ciated lumbar scoliosis was observed in a study of 23 young adults to be compensatory and nonprogressive. The scoliosis was minor in patients with less than 2.2-cm discrepancies.[89] In a study of 15 patients with leg length differences averaging 3 cm (1.4–5.5 cm), the development of scoliosis was noted.[91] Neither of these studies noted significant back pain. Both reported some residual scoliosis following correction of discrepancy with lifts.

These updated clinical studies support the observations from experimental studies that disruption of postural reflexes, both subtle and dramatic muscle imbalance, resection of the posterior portion of the thoracic cage, radiation therapy, and leg length discrepancies are all capable of producing or causing scoliosis.

Etiologic Theories

A review of the salient theories that have a basis in biomechanical principles follows.

Roaf suggests that the basic problem in scoliosis is relative lengthening of the anterior components of the spine compared with the posterior elements.[99] Such a situation in an unyielding anterior musculoskeletal wall of the body results in lateral deviation of the spine and the subsequent development of scoliosis. The theory does not explain why the deviation is so predominantly to the right. Also, there is unconvincing evidence that the muscles of the anterior abdominal wall cannot stretch, yield, or accommodate the long anterior elements. Moreover, the muscles are not particularly tense in patients with scoliosis.

MacEwen produced scoliosis experimentally in animals by transection of the dorsal nerve root and suggested that the result may be due to a loss of sensory input.[68] The convexity of the resultant curve was to the side of the disrupted neural sensory elements. Alexander, Bunch, and Ebbesson showed with histologic staining techniques and examination of the anterior nerve cells that the ablation of the dorsal sensory roots also caused an associated motor root impairment.[3]

A theory proposed by White is as follows. The observation that occasional coupling of axial rotations of the vertebrae causes the anterior aspect to point toward the convexity of the lateral curve in normal bending in the middle portion of the thoracic spine must be considered. It is generally acknowledged that scoliosis frequently starts in this midthoracic area. There is already a physiologic, slight, right thoracic curve in this region. If some precarious balance of the normal thoracic motion should be disturbed, vertebrae in the physiologic, right thoracic curve might somehow rotate too much into the convexity of the curve (Fig. 3-3). Such an occurrence could set off a chain of events leading to asymmetrical loads on the epiphyseal plates and muscle and ligamentous imbalance, with ultimate progression to scoliosis. The precipitating condition may be an abnormal or malaligned facet, a discrete traumatic episode, a chemical hormonal change, extreme handedness of the individual, or any number of other possible embarrassments that upset the delicate balance. The crucial variable may be whether or not the thoracic vertebrae of the normal curve rotate toward the concavity or the convexity of the lateral curve.[127]

The relation of handedness to scoliosis has been supported by the observation of the left convex curvature in left-handed individuals.[44] It is tempting to speculate about the great proportion of right thoracic curves in idiopathic scoliosis. Do left-handed individuals with scoliosis tend to have left thoracic curve deformities? McCarver and colleagues reviewed left thoracic and related curve patterns.[71] Of 14 patients with left thoracic curves, ten were right-handed, only two were left-handed, and two were ambidextrous. One left-handed patient and one of the two ambidextrous patients had infantile id-

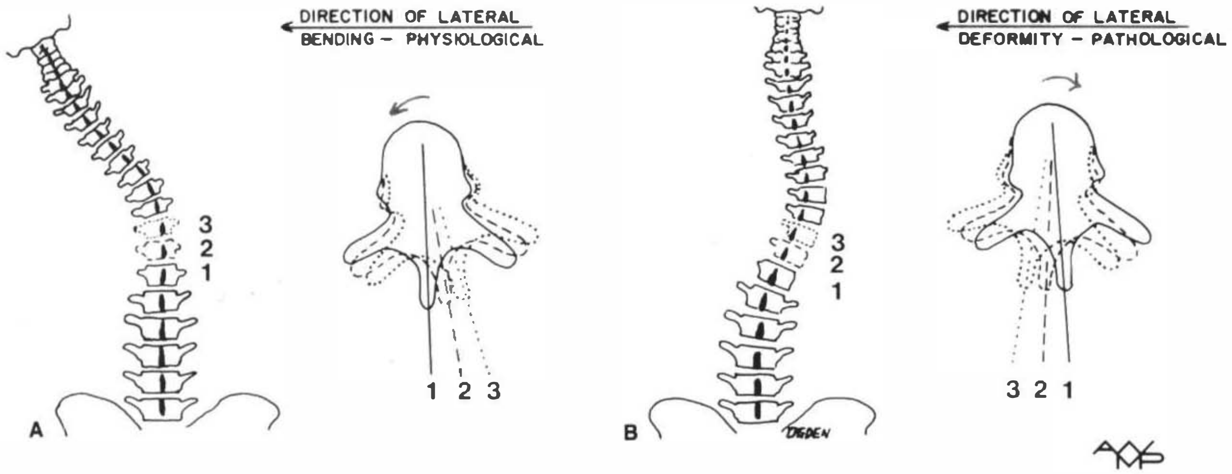


FIGURE 3-3 Axial rotation into the normal (physiologic) lateral curve (**A**) and the scoliotic curve (**B**) is represented. (**A**) The normal curve generally shows axial rotation of the anterior portion of the vertebra into the concavity of the physiologic curve. However, *sometimes* the normal curve may rotate into the convexity. (**B**) In the scoliotic spine, the associated axial rotation is *always* into the convexity of the lateral curve. *(White III, A.A.: Kinematics of the normal spine as related to scoliosis. J. Biomech., 4:405, 1971.)*

iopathic scoliosis.* These data do not support a hypothesis suggesting some association of left thoracic deformity with left-handedness.

Another possible neurologic basis for the etiology of scoliosis has been proposed. This is based on epidemiologic studies carried out by Yamada and colleagues. These investigators found that of 100 patients with scoliosis, 99 had abnormal equilibrium. This malfunction progressed with the severity of the scoliotic curve. At full growth, the findings gradually diminished and disappeared. The dysfunction was noted in the proprioceptive and optic reflex systems.[131] This observation shows an association, but not necessarily a cause. Nevertheless, these neurologically based clinical studies provide important information to help understand scoliosis.

Loynes carefully reviewed 241 patients who had thoracoplasty and removal of three to ten ribs. A convex scoliotic curve to the side of the operation developed in 99% of these patients. Scoliosis tended to progress with time.[66]

Ponseti[95] suggested that a shift in the position of the nucleus pulposus toward the convex side of the curve might be the cause of scoliosis. The normal physiologic shift of the nucleus pulposus is toward the concavity of the curve.

Occasionally a patient is seen who has a single or double curve over a tilted, asymmetrical, or malformed fifth lumbar vertebra. The situation can also exist with pelvic obliquity. Such asymmetry may result in a moment about the z-axis that would tilt the entire spine off to one side. In order to keep the center of gravity over the sacrum, a physiologic curve develops in the lumbar spine. With time, the unbalanced forces acting on the epiphyses lead to structural changes. If compensation is then needed above the lumbar curve, and the epiphyses are young enough to respond, a similar process may occur in the thoracic curve. That is, a functional curve that was initially compensatory may become structural (Fig. 3-4). Our update of the experimental and clinical studies of scoliosis provided no salient new hypotheses regarding the etiology of scoliosis. We believe that the normal spine in a growing person has a precise, precarious, delicate mechanical balance. Asymmetrical changes in primary structures, support structures, growth centers, the position of the spine, and related neural or muscular components can result in the development of scoliosis.

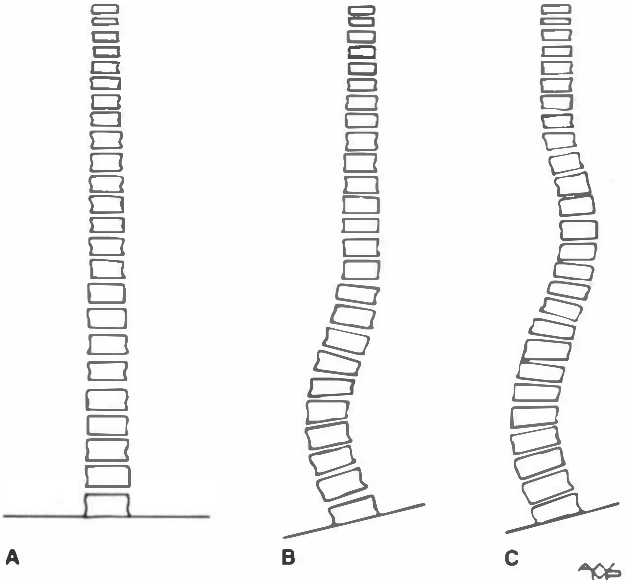


FIGURE 3-4 (A) A normal balanced spine. (B) "Tilt" or asymmetry in lumbosacral area, with a functional or structural lumbar curve. (C) A superimposed functional and structural thoracic curve.

BIOMECHANICAL CONSIDERATIONS INVOLVED IN PROGNOSIS

Mehta[74] suggested that measurement of rib vertebral angles at the apex of the curve would be useful in determining the prognosis of idiopathic scoliosis.

There have been several publications that show some correlation between a high rib vertebral angle (RVA) at the apex of the curve and progression of the curve (Fig. 3-5).[58,74] Ceballous and associates[17a] noted that in infantile idiopathic scoliosis, the rib vertebral angle difference (RVAD) was >20° in 80% of those curves which progressed and <20° in the remaining curves. In the resolving curves, 80% had an RVAD of <20°, and the remaining 20% had RVADs of greater than 20°. Thompson[114a] reviewed 86 patients with infantile idiopathic scoliosis and also noted that those curves with RVADs <20° had a more favorable outcome. These two studies did not find that age of onset and extent of initial angular deformity as measured by Cobb's angle were related to prognosis for progression. Kristmundsdottir and colleagues[58] also studied RVADs in patients with infantile idiopathic scoliosis. This work confirmed that the RVAD was a useful prognosticator in infantile idiopathic scoliosis. These investigators studied

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* Personal communication, D. Levine.

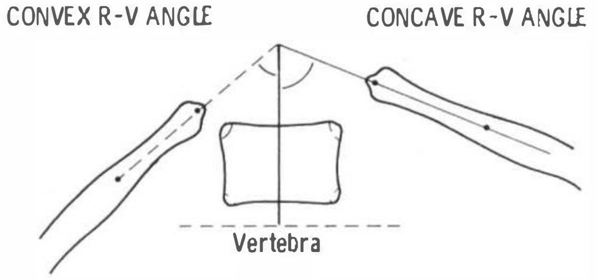

FIGURE 3-5 The method of measurement of the rib vertebral angle. *(Kristmundsdottir, F., Burwell, R. G., James, J. I. P.: The rib-vertebra angles on the convexity and concavity of the spinal curve. Clin. Orthop. 201:205, 1985.)*

several measurements retrospectively in 169 children with infantile idiopathic scoliosis. They found that the best and only measurement necessary is that of the RVA at the convexity of the *apical* vertebra. If that angle measures less than 68° on the initial radiograph, progression of the curve is likely.

The expectation of a greater incidence of progression of curves in skeletally immature patients is supported by the work of Rogala and colleagues.[100a] They noted in patients with idiopathic scoliosis that skeletally immature girls with a 10° curve had a 15.4% incidence of progression, while skeletally mature girls with a 10° curve had only a 6.8% incidence of progression.

Two studies by Weinstein and associates from the University of Iowa are useful in developing prognoses for curve progression as well as other problems associated with scoliosis.[123,124] They noted slightly more backache in scoliosis patients, although it was related neither to severity of curve nor to osteoarthritic changes. Thoracic curves between 50° and 80° were found to progress in adult life. Only thoracic curves were associated with a decrease in pulmonary function. Significant effects were noted in vital capacity and forced expiratory volume in 1 second only after curves reached 100–120°. The authors suggest fusion of progressive thoracic curves of 50° at skeletal maturity. They recommend fusion of thoracolumbar curves that reach 50–60° because these may develop marked deformity or back pain associated with a translatory shift.

In another study from the same center,[124] 102 patients were followed for an average of 40.5 years, and some useful guidelines for determining progression were confirmed. The investigators noted that with *thoracic* curves, progression was likely if the Cobb angle was >50–75°, the apical vertebra was axially rotated >30%, or the Mehta angle was >30°. They noted that large *Mehta* angles were associated with extreme degrees of axial vertebral rotation. In the *lumbar* spine, progression was related to a Cobb angle >30° and apical vertebral rotation >33%; right lumbar curves were twice as likely to progress as left curves. Also, the relation of L5 to the intercrestal line and translatory shifts played an important role in predicting progression.

The work of Lonstein and Carlson[64] is a major study that emphasizes some different guidelines and provides a nomogram to predict probability of progression. In a review of 727 patients with idiopathic scoliosis, they found that progression was related to: the magnitude of the curve, the patient's chronological age, and the Risser sign. Unlike other investigators, except for the above, these authors found no radiologic criteria to be useful predictors of progression.

Haderspeck and Schultz[47a] developed a biomechanical model to study the possible role of muscle malfunction in the progression of idiopathic scoliosis. The authors point out the limitations of their biomechanical model; nevertheless, the study suggests that if there is a role, it is more likely related to the *control* mechanisms than to the muscles. This observation certainly fits with the numerous studies that show the importance of neural and other postural control mechanisms.

FACTORS IN PROGNOSIS FOR INFANTILE IDIOPATHIC SCOLIOSIS (IIS) AND ADOLESCENT IDIOPATHIC SCOLIOSIS (AIS)

IIS—Progression Likely

RVAD > 20°*
RVA on convex side < 68°*

AIS—Progression Likely

Skeletal immaturity
Thoracic curves 50–80°
Apical vertebra rotated > 30%
RVAD angle > 30°
The Risser sign < 2

AIS—Lumbar spine progression

Apical vertebral rotation > 33%
Cobb angle > 33°

* Measured at apex of curve.

PREVENTION

There is not a great deal of information on prevention of scoliosis. Severe scoliosis can be prevented by early detection and surgical intervention if progression cannot otherwise be prevented.

McMaster[73] has suggested that infantile idiopathic scoliosis may be prevented by care and attention to how the baby is placed and maintained in its place of rest. The hypothesis is that with improper positioning, gravitational forces may deform the skull, thorax, and spine. If the hypothesis is correct, this particular deformity can be prevented by laying the newborn in a prone position and encouraging free movement.

School screening has proved effective in reducing severe scoliosis. Moiré fringe analysis[129] and a specially designed inclinometer[16] have been used as adjuncts in scoliosis screening. Both methodologies can be correlated with an appropriate level of frontal plane deformity as measured by the Cobb angle. Deviation of two or more Moiré contour lines was associated with a Cobb angle of 14–25°. Children with less than two Moiré contour lines had Cobb angles of <10°.[129] Five or more degrees of trunk rotation about the longitudinal axis (y-axis) measured with the inclinometer correlated with about a 20° Cobb angle, and this indicates the need for orthopedic consultation.[16]

BIOMECHANICAL CONSIDERATIONS INVOLVED IN TREATMENT

The mechanics of treatment attempt to return the spine to a normal configuration. There are basically two types of deformation that must be corrected. The first is called the functional curve. This is an abnormal curvature that is always present, except when some force is applied to correct it, such as active muscular strain by the patient. The patient may also bend toward the convexity of the functional curve. This curve is maintained by muscles and gravitational forces. Compensatory curves are usually functional curves. However, long-standing functional curves may become structural if not corrected.

In contrast to the functional curve is the structural curve, which is more rigid and cannot be corrected by active muscle forces. This curve usually involves some deformation within vertebrae; there is wedging and distortion of the osseous structure, and the ligamentous components of the curve are stiff. Either curve may have some component of axial rotation.

The correcting loads may be applied through a variety of different techniques. The loads vary in frequency, amplitude, duration, and mode of application. Basic engineering principles are involved in the correction of scoliosis.

Creep and Relaxation

Creep is an important concept in the treatment of scoliosis. The phenomenon is due to the viscoelastic properties of the muscles, ligaments, and bones. Creep is the deformation that follows the initial loading of a material and that occurs as a function of time without further increase in load. When a force is applied to correct a spinal deformity, and the force continues to work after the initial correction, the subsequent correction that occurs over a period of time as a result of the same load is due to creep (Fig. 3-6). When a load is applied to a viscoelastic material and the deformation remains constant, the observed subsequent decrease in load with time is *relaxation*. There are a number of clinical examples of the use of either creep or relaxation: the use of halo femoral traction (creep); the pause of several minutes between distraction increments with a Harrington rod (relaxation); and reoperation 10–14 days following implantation of a rod in order to gain additional distraction (relaxation).

In an experimental study, a Harrington rod with a pressure transducer was constructed with the ability to reflect axial loading. With the use of a wireless telemetry system, it was possible to obtain information about axial forces on the rod associated with different activities. Nachemson and Elfstrom found that the distraction force is decreased 20–45% in the first hour following surgery.[79]

Comparison of Axial, Transverse, and Combined Loads for Scoliosis Correction

True simulation of the scoliotic spine mathematically requires a complex, three-dimensional model. However, there is some merit in studying the behavior of the spine by highly simplified models in order to test a basic concept. A simplified model is employed here in order to study the comparative efficiency of different types and combinations of loads applied to a scoliotic spine for correction. The

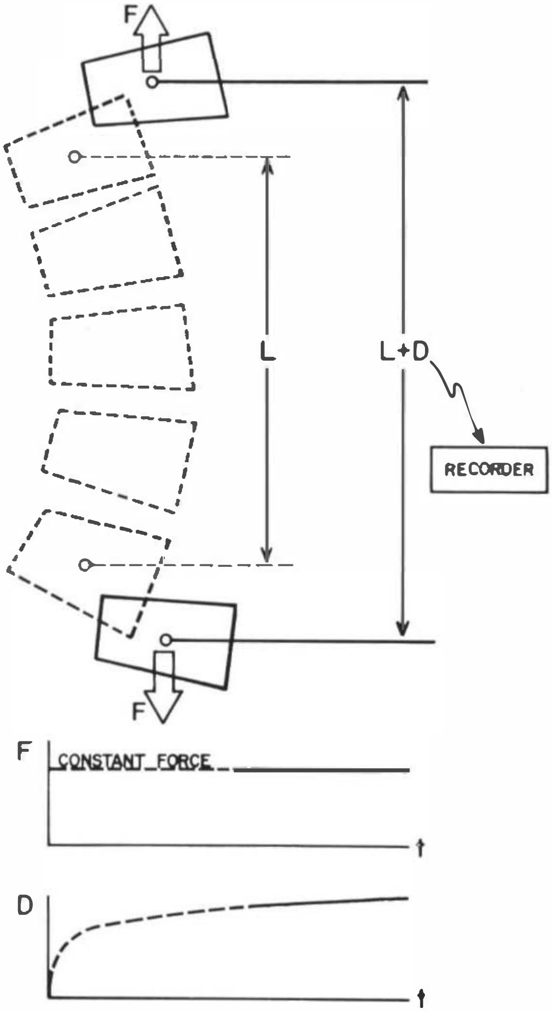

FIGURE 3-6 Creep in scoliosis. *F* is a constant force applied with axial traction. The original length of the scoliotic segment *L* corrects and increases to *L*+ D as a function of time. *D* is the deformation or change in the length of the curved segment of spine.

scoliotic spine is modeled by three components: two rigid links AC and BC, connected by way of a torsional spring C (see Figs. 3-7–3-9). The components lie and move in the frontal plane. The links are oriented to simulate spine deformity in θ degrees as measured by Cobb's method. The static behavior of this model is studied under three separate loading conditions—axial force, transverse force, and a combination of axial and transverse forces.

The principle of axial loading in correcting a scoliotic spine is used frequently. Examples include skeletal traction, the Milwaukee brace, and Harrington rods. Figure 3-7A shows the spine being stretched by the axial force. A three-component model for this loading is shown in Figure 3-7B. An axial force is applied at the two ends of the spine segment, represented by points A and B in the model, to elongate and straighten the spine. The mechanism of angular correction by elongation is due not to tensile stresses in the spine but rather to the bending moments (stresses) created at the various disc spaces. It is these bending moments that correct the angular deformity.

In contrast to axial loading, transverse loading has not been used as extensively, until recently. In the Milwaukee brace and on the Risser table, use of the lateral pad applies transverse loading. Figure 3-8 shows the spine being subjected to lateral loads. A three-component model for this load type is shown in Figure 3-8B. The lateral force is applied at C, and reactive forces half its size are taken up at points A and B. The angular correction is again obtained by creating corrective bending moments at the disc spaces. Simple expressions for the bending moments produced at the disc space, represented by point C in Figure 3-8B, for axial and transverse loads separately, may be derived.

Studying Figure 3-7A in more detail, one notices that the corrective bending moment at the apex of the curve is the axial force F multiplied by its perpendicular distance D to the apex of the curve. It is easily seen that the greater the deformity, the greater is the distance D. In other words, the correctional ability of the force *increases* with the severity of the deformity.

A different situation occurs when the spine is subjected to transverse loading. Figure 3-8A shows that the corrective bending moment at the apex of the curve equals half of the force at the apex (the other half works on the other half of the spine) multiplied by D, the perpendicular distance to the apex of the curve. In contrast to the axial force, the corrective bending moment for the lateral force *decreases* as the deformity of the spine increases.

From this discussion it becomes apparent that a combination of axial and transverse loads is most beneficial for all situations. In other words, the axial component provides most of the corrective bending moment when deformity is severe, and the transverse component takes over the correcting function when deformity is mild. Figure 3-9 shows what the load situation for the combined case might look like. This fits well with the findings of Schultz and

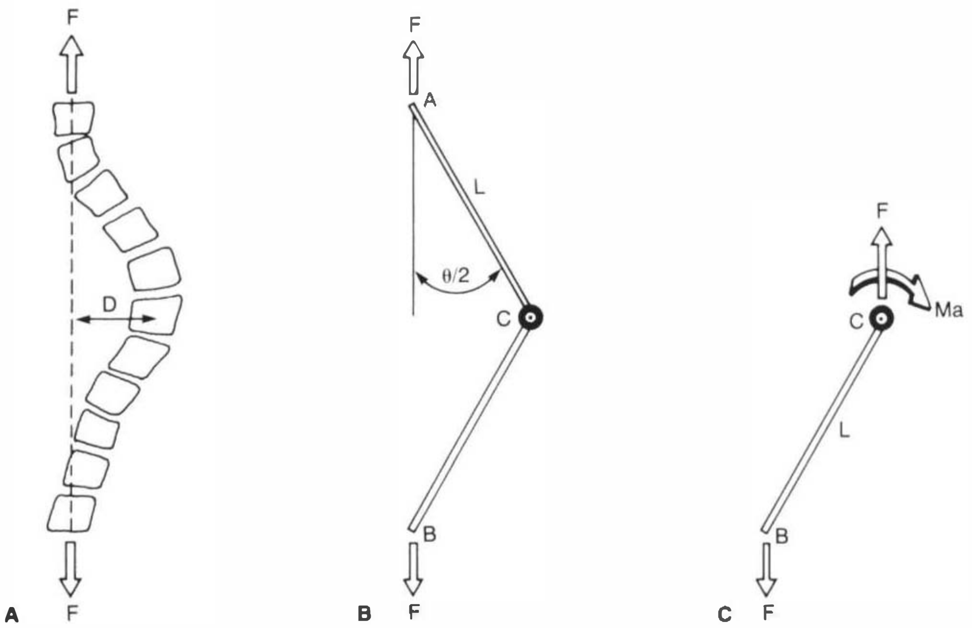

FIGURE 3-7 Axial load. **(A)** The scoliotic spine under axial load. **(B)** A simplified model of the spine being subjected to axial distraction force F. **(C)** Free body diagram of the model link BC and the joint C.

Hirsch, which were based on computer-generated models of the human spine. They found that, with the mild curves, axial loading with the distraction rods provided relatively little incremental correction and required relatively large forces.[105] Using equal loads at the three loading points, the two end forces will have to be tilted 30° toward the center force for the equilibrium of the spine. These arguments are supported by engineering analysis of the simple models shown in Figures 3-7 through 3-9.[C]

Comparison of the efficiency of the three loading types can be made on the basis of the corrective bending moment produced at the disc space. The greater the bending moment, the greater is the angu-

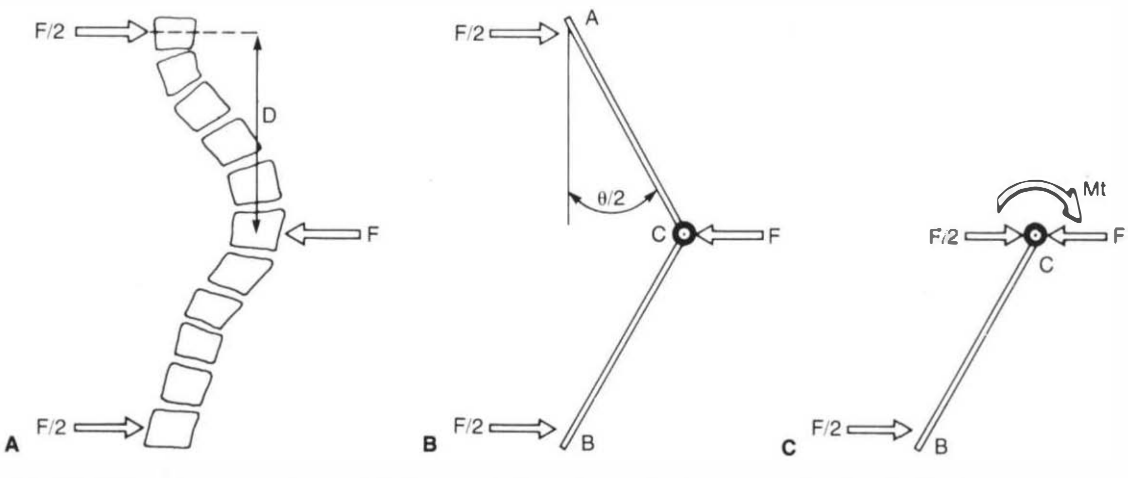

FIGURE 3-8 Transverse loads. **(A)** The scoliotic spine under transverse loads. **(B)** A simplified model of spine being subjected to three-point transverse forces. **(C)** Free body diagram of the model link in BC and the joint C.

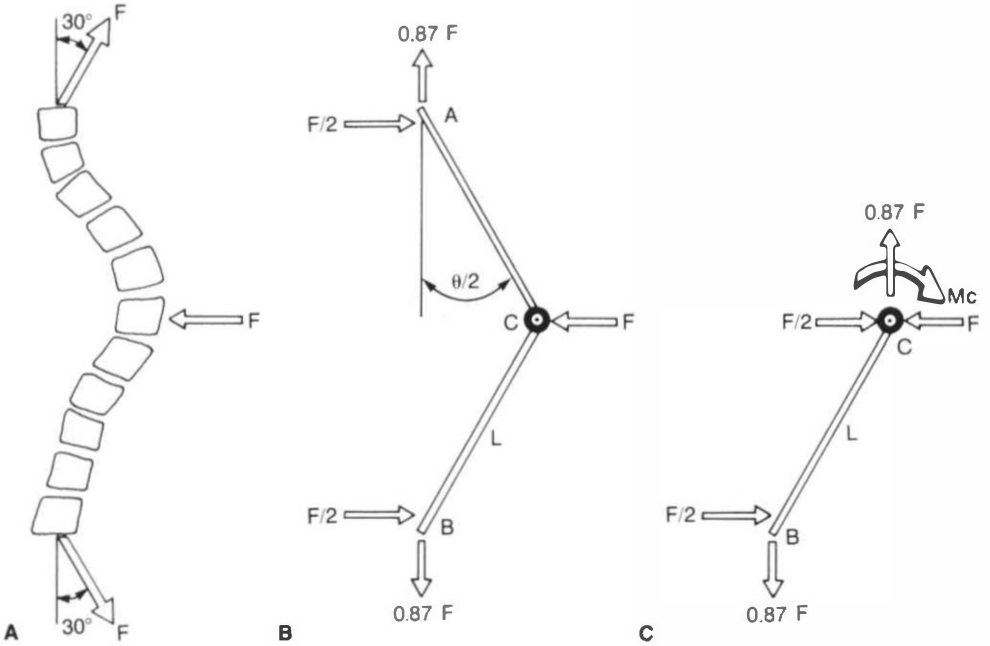

FIGURE 3-9 Combined loads. **(A)** The scoliotic spine under combined axial and lateral loads. **(B)** A simplified spine model being subjected to combined loading. **(C)** Free body diagram of the model link BC and the joint C.

lar correction obtained. Figure 3-10 shows the comparative results in graphical form. The diagram shows, on the horizontal axis, the angular deformity θ in degrees as measured by Cobb's method and, on the vertical axis, a factor M/FL, which represents the amount of corrective bending moment obtained at the apex of the curve as a result of any of the three load types. If F and L have the same values in the three methods (a given scoliotic spine is loaded to the same load level), then M/FL represents the *relative* corrective moment.

Comparing the graphs of the axial load with those of the transverse load, it can be seen that these two curves cross at about where the angle θ is 53°. Therefore, based on the analysis of this theoretical model, axial loading is more beneficial for severely deformed scoliotic spines, while transverse loading is ideal for correcting milder curves. However, when all three graphs are compared, it becomes clear that the combined load is the most efficient load type for all degrees of deformity.

Use of the three graphs in Figure 3-10 determines the most efficient treatment for a given patient. Suppose there are two patients whose curves measure 30° and 70°, respectively, by Cobb's method. Assuming that all three load types are feasible and available for treatment, the problem is to choose the most efficient. Plotting the value of θ = 30° on the horizontal axis, for the patient with mild deformity (see Fig. 3-10), one can read off the vertical axis the three values of the quantity M/FL provided by the axial, transverse, and combined load types. These are 0.26, 0.48, and 0.71, respectively. Considering the M/FL value provided by the axial load as 100%, then the corresponding values for the other two load types are 185% and 273%. Thus, for this patient, the transverse load is more efficient than the axial load, while the combined load is the best. Similarly, one can plot and read off the values for the patient with severe deformity, 70°. From Figure 3-10, the M/FL values are 0.57, 0.41, and 0.91, respectively. Again counting on the percentage basis and considering the M/FL value provided by the axial load type as 100%, the values are 72% and 160%, respectively, for the transverse and the combined loadings. Thus, for the patient with severe deformity, the axial load is more efficient than the transverse load, but again the combined load is superior. Based on these theoretical considerations, patients with severe deformity should be treated by axial loading in the begin-

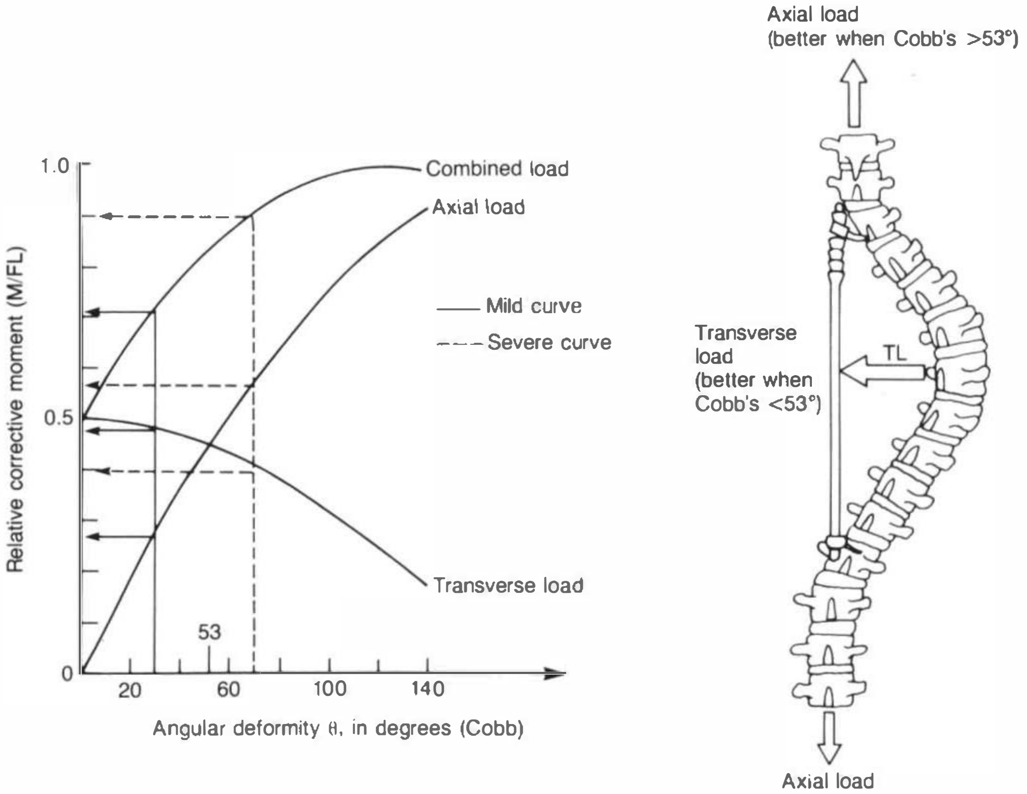

FIGURE 3-10 A graphic representation of "relative corrective moment M/FL" as a function of spine deformity in degrees (Cobb's method) for the three loading types. According to the theoretical model studied here, we note the following: The combined load is the most efficient for any degree of deformity; the axial load efficiency increases with the angular deformity; and the transverse load efficiency decreases with angular deformity. The deformity angle of 53° is a break-even point for the axial and transverse loads. Examples of two theoretical patients with mild (30°) and severe (70°) curves are shown.

ning, and as the deformity decreases, the loading should be changed to the transverse type, assuming that axial and transverse loadings cannot be combined and applied simultaneously.

Connock and Armstrong have made an innovative contribution in the development of the apparatus that allows for the direct application of transverse loading to scoliotic vertebrae. Other devices have also been developed.[97] The principle of the technique is shown in Figure 3-11. By applying tension between the rod and metal devices attached to the posterior element, a transverse load TL is applied.[22] There is an advantage, however, in applying the transverse load at a different point (Fig. 3-11C). This tends to correct the abnormal axial rotation of the scoliotic vertebra. The element of abnormal frontal plane curvature is nearly corrected with the current treatment armamentarium. A method of applying transverse loads to a point to *correct* the axial rotation is being developed.

Update of Biomechanical Considerations in Treatment

In this section, the authors will review some of the most salient clinical biomechanics in scoliosis correction. Actual surgical procedures are discussed elsewhere in this chapter (p. 148) and in Chapter 8. A discussion of *vertebral rotation, transverse loading*, the *correction of axial rotation*, and *quantitative biosurgical* measurements follows.

Vertebral Rotation

In a study by Zagra and associates[134] designed to examine fusion, an interesting observation was made. The fusion mass was most developed on the

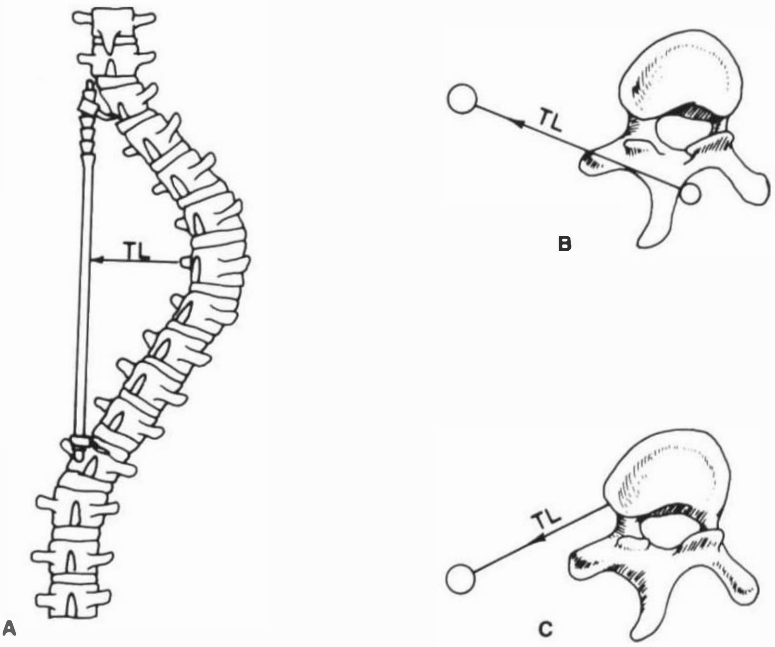

FIGURE 3-11 A comparison of the effects of two types of transverse loading (TL) on the axial rotation of a scoliotic vertebra. **(A)** Both types of transverse loading correct the lateral curvature. **(B)** Transverse loading attached *posterior* to IARs tends to increase the abnormal θy rotation. (C) Transverse loading attached *anterior* to IARs tends to decrease the abnormal θy rotation.

concave side of the curve at the apex. This fits with the traditional interpretation of Wolff's law. Bone is thought to be laid down in regions of maximal stress. Of particular interest here is the observation of bone resorption in the central portion of transverse planes through the fusion mass. The central resorption of relatively unstressed bone surrounding the neutral axis suggests that there may be torsional stresses in this region. In other words, there may be a significant component of y-axis torsional stresses in scoliosis. The torsional loads are hypothesized to develop secondary to vertical forces that have a shear component, a portion of which is resolved into an axial torque.

This shear loading axial torque may also be the explanation for the abnormal y-axis rotation that occurs in scoliosis. As the distance increases cephalad and caudad away from the apical vertebra, there is relatively less torque and therefore less rotation of the vertebra.

Cobb[18] stressed the importance of vertebral rotation in 1948 when he described a technique for measurement of axial rotation of the vertebra based on the location of the tip of the spinous process in relation to the vertebral body. This method was somewhat limited in accuracy because of the extensive variation in the deformation of the vertebra and the spinous process in particular (see Fig. 3-2). Subsequently, Nash and Moe,[81a] in 1969, described a relatively more accurate method in which the degree of rotation was estimated based on the position of the pedicle on the convex side of the curve in relation to the vertebral body. Mehta[74a] described a system for determining vertebral rotation in 1973. This system, based on an image-matching method, was developed to use the appearance or profile of the vertebra in 15° incremental positions between 0° and 90° of rotation. While this did not eliminate the problems and was not more precise than the other methods, it did provide the possibility for evaluating vertebral rotation up to 90°.

In the early 1980s, Aaro and Dahlborn used CT to study axial rotation of the vertebrae and rib cage deformity in scoliosis.[1a, 1b, 2] This method was found to be highly accurate, with CT vertebral axial rotation measurements only 0.3° different from actual measurements. However, the accuracy is lost if the vertebra is tipped (rotated) 20° in either the frontal or sagittal plane.[1a] These investigators discovered some cogent findings in their studies. They noted a strong correlation between apical vertebral axial rotation and the lateral scoliosis curve and the development of the rib hump. The rib hump was also further accentuated by increased lordosis.[1b] The correlation between apical vertebral axial rotation and the lateral curve and rib hump was also noted by Armstrong and colleagues.[3a] More recently Stokes and associates have developed a method for measuring axial rotation of the vertebrae in scoliosis.[113]

Aaro and Dahlborn also studied with CT the effects of Harrington instrumentation on axial rotation of the vertebra, the kyphosis/lordosis, and the rib cage deformity in 33 patients.[2] It is important to note that no significant derotation was observed. The rib cage deformity was significantly reduced only by the Harrington distraction rods.

In summary, apical axial vertebral rotation can be roughly estimated on plain films by several techniques. For accurate measurement, CT scans are required. This technique has limited accuracy if the vertebra is also rotated in either the sagittal or frontal planes, or both. Axial rotation is correlated with both the frontal plane and the rib hump deformities. We know from the section on progression that axial rotation is associated with progression in idiopathic scoliosis.[124] Apical axial vertebral rotation is an important component of the scoliosis deformity in terms of characterization, prognosis, rib hump deformity, and treatment.

Inflection Point

The work of Dabney and co-workers[25] confirms with experimental biology the concept described in previous models and in Figure 3-12. The investigators corrected experimental scoliosis in rats with a distraction rod that measured forces. They observed a kind of *inflection point* whereupon there was a rapid increase in forces with each increment of distraction. In addition to being similar to the models, these experimental distraction forces, we believed, were quantitatively and qualitatively similar, on scale, to those observed in human scoliosis. This study also showed that the distractive forces were greatest for curves above 50°.

Biomechanics of Transverse Loading and Axial Derotation

Transverse loading tends to convert the pure bending moments created by the distraction rod to two couples. This improves the total corrective forces and decreases the loads on the rods and on the posterior elements of the vertebrae to which the rods are attached. Also, the additional fixation point at the completion of full correction adds to the rigidity of the total construction.

Wolf and associates[130] used a transverse loading device in the treatment of 19 patients with scoliosis. The transverse loading device added another 16%

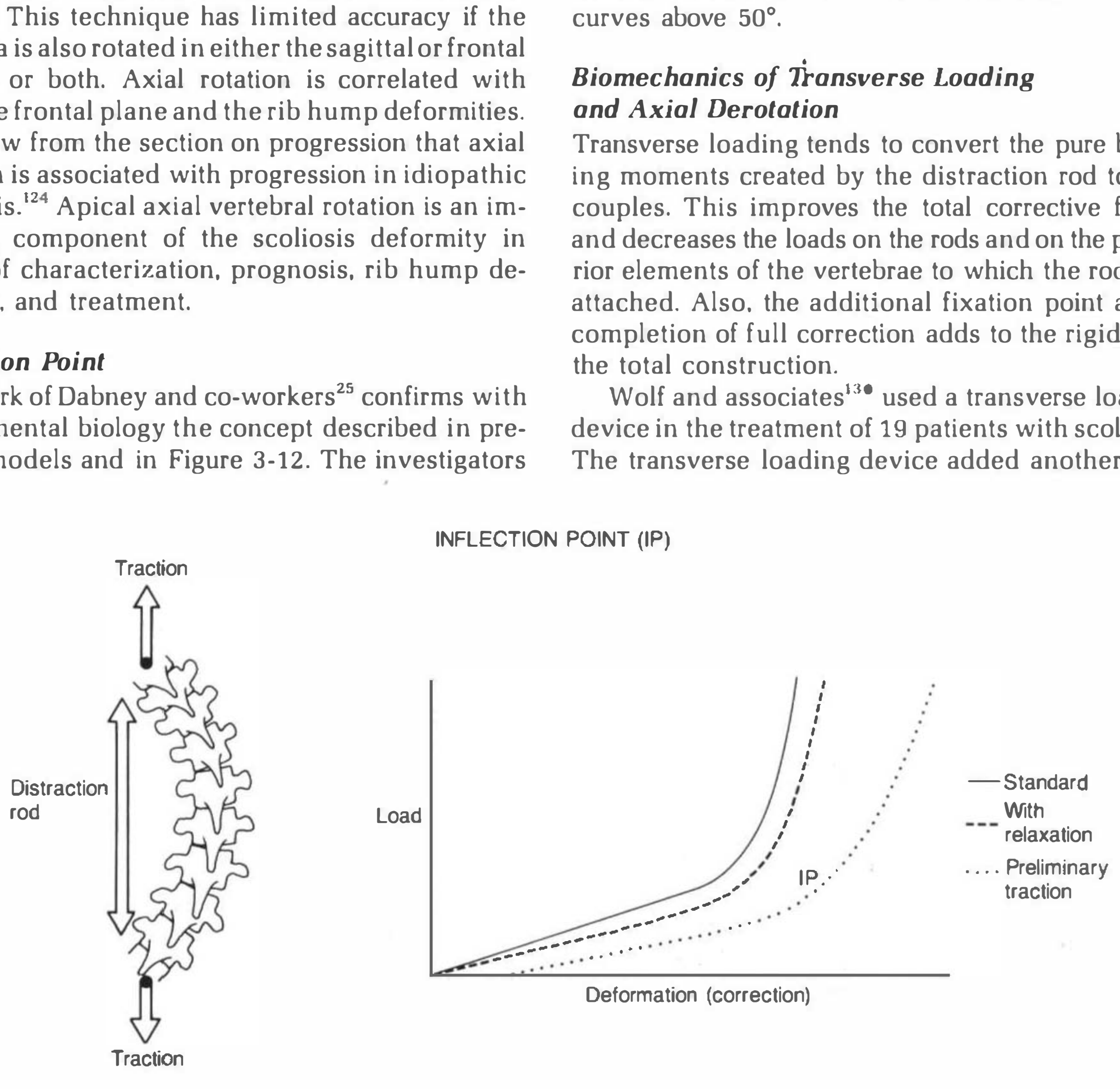

FIGURE 3-12 The use of a traction force or a distraction force to a scoliotic spine creates stresses within the spine. This *theoretical* diagram and graph show the relationship between the correction of the scoliotic spine and the load required to achieve it. There is a phase in the load-deformation curve in which there is precipitous increase in the slope of the curve, indicating increasing resistance. This point is called the *inflection point*. This occurs with or without relaxation between distractions, and with or without preliminary traction. The use of relaxation between distractions may, however, allow more correction before the inflection point is reached. The inflection point signals that a yield or failure point is being approached.

correction to the 50.3% maximal attainable correction of the Harrington distraction rod alone. Ransfort and Edgar[97] described a transverse system to supplement the Harrington distraction rod. Subjectively, there was no adverse effect upon the rib hump. The transverse loading device was thought to have alleviated the load on the Harrington system. Ogilvie and Millar[83] studied Harrington rods with either sublaminar wires or convexity compression rods and transverse loading wires in the treatment of idiopathic scoliosis. The sublaminar wire construction generated more favorable corrective vectors. Both further rotated the scoliotic spine in the direction of the abnormal axial vertebral rotation. This basic problem is depicted in Figure 3-11. We know that the Harrington distraction rod alone does not correct axial rotation.[2] There is clinical confirmation that posteriorly applied transverse loading, although it helps to correct the frontal plane deformity and has other biomechanical advantages, does not correct, and may in fact contribute to, the abnormal axial rotation. This may not be important clinically. That is, at the practical level, this additional abnormal axial rotation may not perceptibly affect rib hump, cosmesis, progression, pseudarthrosis rate, or anything of clinical importance.

Quantitative Clinical Approaches

Medications are carefully prescribed and monitored to achieve the ideal dose response. Yet we rarely quantitate the forces or deformations we use surgically to achieve the desired mechanical response. Thus, we applaud those instances in which our surgical and/or engineering colleagues have begun to quantitate intraoperative therapeutic forces.

Jacobs[54] described a dynamic outrigger with a dynamometer that allowed forces to be monitored during the initial intraoperative correction of scoliosis. With this device it was possible to recognize the *inflection point* described in Figure 3-12. This point of rapid increase in force with slight additional distraction was taken as the point of maximum allowable distraction without damage to the spinal column, lamina failure, or hook cutout. The load–deformation curve and the inflection point are shown in Figure 3-12. The outrigger is used to distract to the *inflection point*, the axial distance is measured, and the Harrington rod is placed and distracted that same linear distance. More recently, Ashworth and colleagues[4] have described a force-indicating spreader as a teaching tool.

We believe it is useful to mention several clinical biomechanical points. After one reaches the point of rapidly increasing loads, the incremental correction that occurs with distraction is minimal, while the incremental loads incurred are very high. Thus, the risk/benefit ratio is also very high for further distraction once this point is reached. The high risk/benefit ratio can be somewhat alleviated by using threaded Harrington distraction rods, because the incremental loads may be applied at much lower magnitudes.[115] Another point in the use of incremental distraction is to take advantage of the viscoelastic properties by allowing for *relaxation* of the tissues. After a given incremental distraction, the initial loads on the rod and spinal column will decrease with time. Dunn and co-workers[34] showed that in a few minutes (2–4 minutes) the magnitude of the load may drop 50% as a result of *relaxation*.

MECHANICS OF DIFFERENT TREATMENT METHODS

Exercises

It is doubtful that exercises alone correct scoliosis. A vigorous and thoroughly supervised exercise program may re-educate patient and muscles so as to correct a functional curve. The muscle forces that can be applied are of relatively low amplitude and frequency and are usually of short duration. Exercises should not be relied upon to hold or correct a curve when used alone.

It is interesting to speculate as to why exercises alone have not been found beneficial in the treatment of scoliosis. Muscular forces are rarely working at a significant mechanical advantage for correction of the scoliotic spine. The erector spinae muscles are not able to function at the most efficient mechanical advantage. The other factor, which is perhaps more important, is the need for prolonged force application in order to take advantage of the viscoelastic creep. It is difficult to maintain voluntary muscle contraction and apply forces to the spine long enough for the resisting viscoelastic structures to yield. These two factors limit the capacity of exercises in achieving correction.

Electrical Stimulation Update

Electrodes powered by intermittent cyclical currents have been implanted in the erector spinae muscles on the convex side of the curve to stimulate muscle contraction. A totally implantable unit with its own power supply has been developed. Transcutaneous stimulation of muscles is also being used in humans to correct scoliosis. Preliminary reports indicate that this may prove beneficial in some cases.[9]

Biomechanical modeling studies by Schultz and colleagues indicated that trunk muscle stimulation can be expected to effect substantial changes in spine configuration. This was thought to be possible even with contractions of modest intensity.[104] Brown and associates,[15] in a multicenter study (54 principal investigators), evaluated the use of noninvasive muscle stimulation in 548 patients with idiopathic scoliosis. The conclusion was that the treatment was acceptable to patients and was a viable alternative to orthotic treatment. Axelgaard and Brown used surface stimulation in the treatment of progressive scoliosis. They reported success in the treatment of single and double curves. Overall success was 84% in single curves and 83% in double curves. If only the compliant patients are considered, the success rate goes up to 97% and 93%, respectively.[5] Bradford and colleagues[14] used surface electrodes in the treatment of 30 patients with idiopathic scoliosis. These investigators, like Goldberg,[43] more recently concluded that the treatment does not work.

Orthotic Treatment

The Milwaukee Brace

This orthosis may be advantageous in the treatment of scoliosis. A *normal* mold is constructed into which the deformed patient is fit. Active exercise is performed while in the brace, and corrective forces are applied to a growing spine. The spine is being supported, splinted, and stretched between the throat and occipital molds on one end and the pelvic girdle on the other. Additional forces are applied through corrective pads attached to the uprights of the brace and the axillary sling. The brace takes advantage of growth potential, so that it has the possibility of correcting structural deformity. Active correctional forces are being applied to the spine by the patient both consciously (the active exercise program) and unconsciously (occasional moving away from the pads, the axillary sling, and the throat mold and occipital piece). The amplitude of the active forces is limited by the patient's muscle strength. The passive correctional forces are limited by the stiffness of the tissue transmitters (ribs, muscle, skin, fat), biological tolerances (pain threshold, organic functions), and the psychosocial tolerances of the patient (see p. 478).

A special exercise program is a crucial aspect of the use of the Milwaukee brace. Exercises in breathing and pelvic alignment are an important adjunct. These are thought to resist lumbar lordosis and enhance the active correctional forces of the patient's own intrinsic muscle groups. When the brace is constructed, it must be "made straight." That is, it must not yield or be fit in any degree to the deformity as it exists. The brace should thus be made on a level pelvic support, with the uprights erect and perpendicular to the horizontal plane of the pelvis. In other words, the deformed body must go into a normal mold. It is important to hold the correction, once it is achieved, until maturity and, finally, to wean the patient gradually.

The Milwaukee brace has been used as a holding device both pre- and postoperatively.[102] However, it must be kept in mind that, following completion of treatment, some patients can progress as much as 1–2° yearly, up to age 25 or 29.[75]

Nachemson and Elfstrom implanted telemetrized Harrington rods in patients with idiopathic scoliosis and obtained data about the effects of several pertinent factors on forces exerted upon the rods.[79] Much of this information is of clinical importance. Information from this study suggests that it is probably worthwhile to avoid allowing patients to lie on the convex side of their curve during the first 2 weeks after rod implantation.

When forces on the rod with a Risser cast, a Milwaukee brace, and no support were compared, Nachemson and Elfstrom found the brace to be superior in terms of reducing the axial forces on the rod (Fig. 3-13). They also showed that when different parts of the brace are removed, the axial forces increase.[79] When the patient coughs or vomits coming out of anesthesia, the forces are increased tremendously, and the patient may fracture the lamina and have the rod pull out. A smooth recovery period from anesthesia is therefore crucial to the use of the Harrington rods. In lifting the patient from the bed to the frame, the forces were also noted to increase significantly. There is argument about whether or not a Milwaukee brace should be kept on when the patient is recumbent. If one lies down and takes off a Milwaukee brace, there is an increase in the axial force upon the Harrington rod. Therefore, the Milwaukee brace does diminish the load even in a supine position.

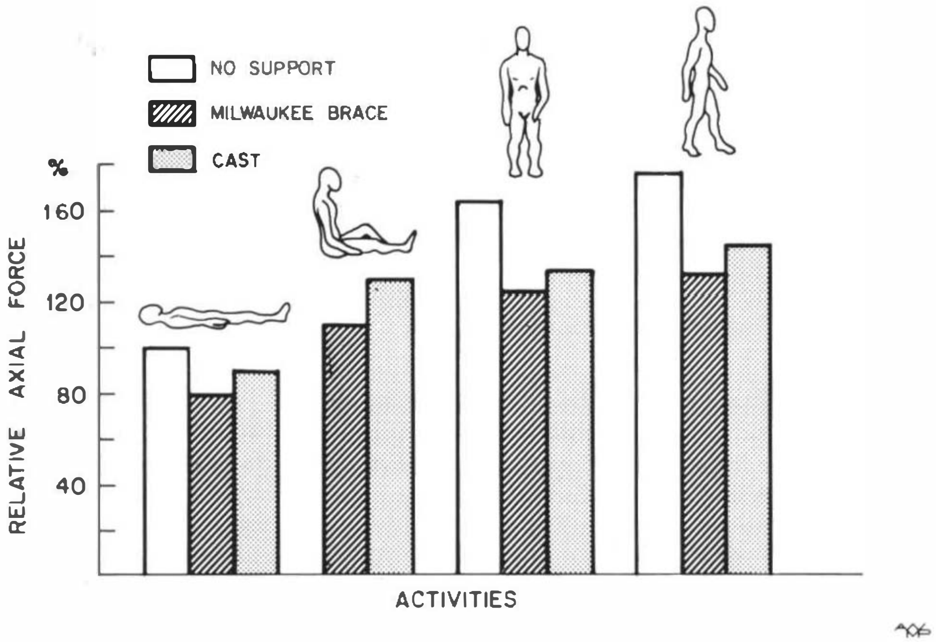

FIGURE 3-13 Relative axial force in the telemeterized Harrington rod in patients wearing no external support, a Milwaukee brace, and a Risser cast. The force in a patient with no support and lying supine has been assigned a value of 100%. The bar graphs show patients under these three conditions performing different activities. *(Data from Nochemson, A., and Elfstrom, G.: Intravital wireless telemetry of axial forces in Harrington distraction rods in patients with idiopathic scoliosis. J. Bone Joint Surg., 53A:445, 1971.)*

Galante and associates carried out an investigation using a strain gauge device under the occipital and mandibular pads of Milwaukee braces. They found that if the axillary pad or the sling was removed, there was increased axial loading. This implies that a loss of correction results in increased load on the brace. The Milwaukee brace provides both active and passive correction, as shown by this experiment. The removal of the lateral forces from the axillary pad or the thoracic pad caused increased axial loading; therefore, these pads were presumably applying corrective loads to the spine. The patient is corrected passively. Furthermore, Galante and colleagues found that when a patient moves away from the thoracic pad, which actively corrects the lateral curve and axial rotation, there is also passive stretching of the spine. This experiment showed that the average force on the brace in a standing position was 19 N (4.25 lbf). However, if the patient pulled away from the thoracic pad, the average force increased to 63 N (14 lbf), slightly greater than three times as much. This shows that the brace gives passive distraction when the patient moves away from the pads.[41]

Update on Milwaukee and Boston Brace Treatment

It is, we believe, accurate to state that clinical studies of the use of orthotics facilitate treatment of scoliosis and teach us that we may not have been doing quite as much as was thought or hoped.[17,61,77,81,102] Carr and associates[17] studied 133 young patients (8–16 years), mostly females. There was a 5-year follow-up of 74 of those patients. Eighty percent showed some increase in curves after the brace was discontinued. The *average* correction at 5-year follow-up was 2° (−18°–29°) for thoracic curves and 4° (−11°–17°) for thoracolumbar curves. The best indicator of a good prognosis was a 50% correction of the curve at the time of bracing. A poor prognosis is suggested by a curve of >40° and/or a high thoracic curve with neck curvature and asymmetrical shoulders.

Rudicel and Renshaw[102] showed that the Milwaukee brace was ineffective in the correction of decompensated curves. Cochran and Nachemson[21] indicated that although patients survived the brace from a functional and social perspective, there was not yet scientific evidence of the efficacy of the method. Probably the best designed and controlled study on the Milwaukee and Boston braces was done by Miller and colleagues in 1984.[77] This was a study of 255 females, ages 8–17, matched for maturity and curve severity. One group was treated with the Milwaukee or the Boston brace; the other was followed. There was a *slight*, but not statistically significant, trend of less curve progression in the treated group.

The Boston brace was noted by Aaro and coworkers,[1] in a study using computerized tomography, to correct axial rotation 38% and Cobb angle 54%. Laurnen and colleagues[61] studied the Boston brace and suggested that it seemed to work with curves with apices at or below the T6 level, but for curves above this, the Milwaukee brace is preferred.

Careful monitoring of good clinical studies is important to definitively determine the cost/benefit ratio (*i.e.*, risks of brace, psychosocial and time expenditure of patient and family, and professional care costs, versus possible avoidance of surgery and ultimate cosmetic and cardiorespiratory results).

Traction

There are a variety of methods that may be employed. Head halter and/or ankle pelvic straps can be used. Greater forces can be applied for longer periods of time using skeletal traction. Halo traction is combined with the pelvic hoop or the more common femoral pins. The halo-hoop has proved valuable in the treatment of a variety of different types of scoliosis. It is especially valuable for straightening out the usually recalcitrant pelvic obliquity. The distractive forces of this apparatus can be measured by placing dynamometers in the four uprights. This is desirable wherever feasible. DeWald and associates showed that forces of 360–400 N (80–90 lbf) were readily tolerated by patients. Kyphotic deformities may also be corrected by the apparatus. Any neurologic signs were reversible by reducing the traction. The investigators also proved by objective *in vivo* monitoring that one of the four uprights may be removed for surgery without loss of correction.[32]

There are certain advantages of preoperative traction in the treatment of severe scoliosis. There is the maximum opportunity to benefit from gradual correction, which takes advantage of the creep phenomenon. This minimizes the possibility of spinal cord damage through correction that is too rapid. Moreover, the patient is awake, and any neurologic change can be readily recognized. This minimizes the need for intraoperative maneuvers, such as monitoring spinal responses and awakening the patient during the procedure to check for motor or sensory function. Since the resisting structures are significantly corrected preoperatively, the final force on the Harrington rod is presumed to be diminished. This should reduce the possibility of failure of the thoracic lamina and "pull out" of the rod. Another advantage of preoperative traction is found in the correction of double curves with different stiffnesses. Radiographs taken during full traction give an excellent idea of the relative final correction for each curve that is compatible with good balance.

Recent studies have shown, however, that axial traction is probably not advantageous in patients under 20 years of age with idiopathic curves with a Cobb angle of less than 90°.[80]

Edgar and colleagues[36] studied 175 patients with adolescent idiopathic scoliosis distributed into the following three preoperative management groups: (1) Risser localizer casts, (2) Cotrel dynamic traction (3 weeks), and (3) Cotrel dynamic traction (1 week). There was essentially no difference in overall correction in the three groups. This study showed that double curves were corrected better in the group that had Cotrel dynamic traction for 3 weeks. Generally, the Cotrel traction was as good after 48 hours as it was after 3 weeks. Given the fact that most (90%) viscoelastic creep occurs in the first 3–5 minutes, this is not surprising.[79] The time frame of correction may thus be significantly less than 48 hours. If muscle contraction resistance is playing a role, we should assume that it would take more than 3–5 minutes to achieve 90% of correction. The authors considered traction to be unnecessary in curves less than 70°.

Viviani and associates[118] have applied incremental distraction forces to the heads of scoliosis patients. From this data they have constructed finite element models that determine the segmental stiffness from spinal geometry derived from the patient's radiographs. These data can be further analyzed with the help of software programs. This would allow actual simulation of the corrective potential of various implants on the specific curves of the specific patient. We have, as a basic theme of this text, applauded the measurement of forces and other parameters as a basis for more precise quantitative control and monitoring of our therapeutic interventions. This type of approach fully developed will allow for more idealized selection of instrumentation and improved capacity to maximize correction and avoid complications, particularly with powerful maneuvers such as that involving the derotation of the Cotrel–Dubousset instrumentation (see Fig. 3-15). Our preceding considerations remain applicable, however, in treating severe scoliosis deformities in other circumstances.

Harrington Instrumentation

Here the forces are applied directly to the two end vertebrae of the curve. These distraction forces are applied manually and are maintained by a locking mechanism in the rod. However, they are not measured. The forces are applied in small increments. Their amplitude is limited by the strength of the bony structure to which the rod is attached. The limiting factor is the lamina of the thoracic vertebrae, which may fail if the distraction rod has a force of 370 N (83.3 lbf).[122] The force may be applied indefinitely. This rod is always used in conjunction with a spinal arthrodesis of the involved vertebrae. The surgeon can take some advantage of relaxation by waiting at least 5 minutes between the last three increments of force applied to the distraction rod. The corrective value of the last notches on the rod is low, and the forces involved are high.[105]

Waugh conducted studies on the biomechanics and technique of the Harrington rod. He found that there is considerable stress concentration of the hook in the upper lamina, which is the weak point of the system. The tolerance limit was found to be equal to a force of about 200–300 N (43.9–65.8 lbf). These studies showed that in flexible curves, 200–300 N (43.9–65.8 lbf) of force was adequate for 60–70% correction. Over 300 N (65.8 lbf), the rod is likely to pull out. Waugh suggested that for greater contact area between the hook and the lamina, a different hook design should be used.[122] He has experimented with the use of methylmethacrylate to reduce stress concentration by increasing the contact area. He also suggested that the steps on the Harrington rod be placed closer together, since the increment of one additional step augments the force on the rod from 200 N (43.9 lbf) to above the crucial 300-N (65.8 lbf) load. If the steps are not as great, the force may range from 200 to 280 N (43.9–61.4 lbf), gaining more correction and still remaining within the range of the tolerance of the lamina.

Axial Forces

Since we know that the limiting factor in the amount of corrective force that may be applied to the scoliotic spine with the distraction rod is the load-bearing tolerance of the thoracic lamina, *in vivo* measurements of axial loads on the Harrington rod become quite important.

The experiments of Nachemson and Elfstrom generated some relevant data on this subject.[79] Forces during operation and immediately after surgery are shown in Figure 3-14. On the ordinate is the force in newtons, and on the abscissa is the time in minutes after operation. One sees that there is a marked decrease in the axial force with time as a result of the viscoelastic properties of the ligaments (relaxation). Within the first hour, the average axial force had fallen to 78% of the original value. Therefore, distraction is best applied with some time lapse between application. The use of the outrigger distraction in surgery, with occasional small increments of axial load, can expand the time range over which the maximal axial load is applied. Thus, greater correction may be obtained by the same load.

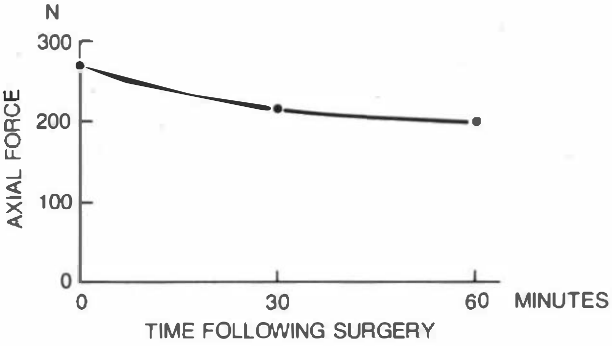

FIGURE 3-14 Axial force in newtons in the Harrington rod in a patient immediately following surgery. Immediate relaxation of force with time is shown. N = 2.21 lbf. *(Data from Nachemson, A., and Elfstrom, G.: Intravital wireless telemetry of axial forces in Harrington distraction rods in patients with idiopathic scoliosis. J. Bone Joint Surg., 53A: 445, 1971.)*

It is important to handle the patient carefully. Lifting and turning the patient in bed produces a great increase in the axial force. Some of the highest forces are produced with coughing. If bucking and coughing are factors in patients awakening from anesthesia, large forces can be created. One strong cough or buck can tear the upper hook from the thoracic lamina. The importance of a smooth, early recovery phase from anesthesia cannot be overemphasized.

Compression Rod

This is a corrective device that is used on the convex side of the curve. The rod is in tension and thus applies two colinear compressive forces to the spine. The change in the positions of the vertebrae between the ends of the rod is dependent upon the bending moments produced at the centers of rotation for the vertebrae. If the centers of rotation are between the rod and the concavity of the curve, these vertebrae tend to be corrected. However, if the centers are between the rod and the convexity of the curve, the bending moments produced may tend to exaggerate the deformity.

The total corrective effect depends upon the balance between the two and upon the distribution of the centers of rotation of the vertebral bodies with respect to the direction of the compressive forces. In general, the corrective effect is small and may even be negative. Waugh attached a compression rod to the convex side of a scoliotic curve after applying a distraction rod to the other side. There was a 9.8-N (2.2 lbf) reduction of the axial forces with the compression rod in place.[122] It does not appear that this small difference contributes significantly to correction.

Recently, more attention has been paid to the sagittal plane deformity in scoliosis. Gaines and Leatherman[40] have suggested that the Harrington compression rod has helped in the correction of the sagittal and transverse plane components of the scoliosis deformity.

Segmental Instrumentation

It may be fair and accurate to state that segmental sublaminar wire fixation has come and almost gone since the first edition of this text in 1978. The efficacy of the fixation in regard to rigidity and corrective usefulness has been well recognized and is presented in Chapter 8. The problem has been the risks associated with the placement and removal of wires within the spinal canal. Because of the development of other forms of segmental fixation (spinous process wiring and pedicle fixation), we have elected not to review the role of sublaminar segmental instrumentation in the treatment of scoliosis.

Cotrel–Dubousset Instrumentation

Drs. Cotrel and Dubousset have recently developed a system designed for the correction and fixation of scoliosis. The basic components of the system involve longitudinal rods and accompanying pedicle screws, hooks, transverse loading rods, and the necessary instrumentation to apply distraction, compression, and transverse loading as well as to contour the rods.

Because the system is relatively new at the time of this writing, we have available only the beginning biomechanical and clinical studies of the use of the system. However, an update on the most cogent studies that are available at this time follows.

Clinical Studies of Cotrel–Dubousset (C-D) Instrumentation for Scoliosis

The treatment has been described by the developers and is offered as a universal instrumentation system.[23] The authors report on approximately 250 patients with scoliosis, lordosis, and kyphosis treated with C-D instrumentation. They used no postoperative external support. The correction averaged 66%. They also reported improvement in the associated sagittal plane deformity as well as apical longitudinal axis derotation. In the paralytic curves there was a mean correction of 77%. The surgeons reported some complications, including one paraplegic, one monoplegic, six upper hook dislodgments, two fractured transverse processes, five postoperative hematomas (associated with the use of the device for transverse traction—DTT), and two wound infections.

Denis[29] has provided a well-explained and superbly illustrated exposition of the use and rationale of C-D instrumentation. He concludes that it is more difficult than Harrington instrumentation and should be performed by a well-trained surgeon devoting meticulous attention to detail. He describes nicely the simultaneous correction of frontal plane convexity and sagittal plane lordosis achieved by rotating the appropriately attached and contoured rod 90°, as shown in Figure 3-15. The tendency for the rib hump deformity to be corrected by these same forces is demonstrated in Figure 3-16.

A preliminary report of 37 patients treated with C-D by Birch and colleagues[8] adds to the clinical documentation of this instrumentation. Twenty-five of these patients had juvenile or adolescent idiopathic scoliosis. The follow-up period was only 1–7 months. The surgeons observed improvement in rib deformity that was thought to be superior to that observed with the Harrington device. Here, too, no postoperative immobilization was used. They noted greater complexity and longer operating time, but noted also the advantages of the versatility of the system and the elimination of the need for external postoperative support and the necessity for rib resection. They note also approximately a ninefold increase in the cost of the C-D device as compared with the Harrington or Luque device.

In a preliminary report by Gurr and McAfee,[46] nine patients with adult scoliosis had C-D instrumentation. The results were thought to be compatible with standard techniques. The correction was proportional to preoperative flexibility, and there was no loss of correction. There was no failure of fusion. Blood loss and hospitalization time were comparable to those of previous techniques. There was no difference in regard to the number of levels fused when compared with previous techniques.

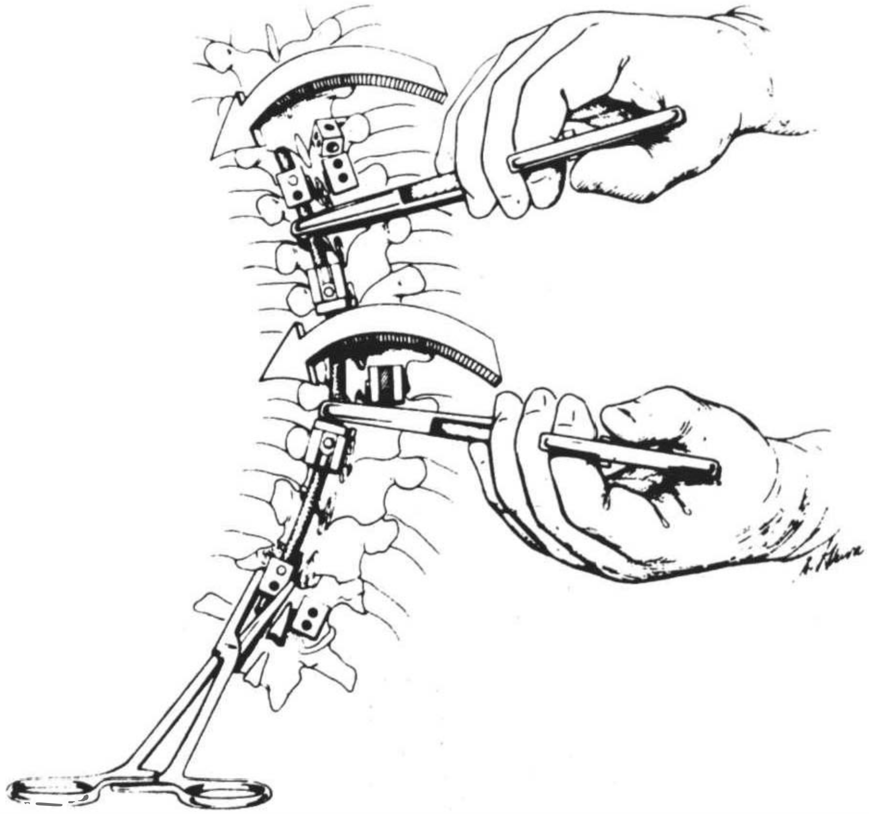

FIGURE 3-15 Rotation of the rod is done very carefully and progressively, about 10 to 20° at a time. Observation of all concave hooks is of major importance because of the risk of sudden fracture of a lamina or of hook displacement. If everything appears under control, the rod is rotated to about 90° where its initial scoliotic configuration is replaced by a kyphotic configuration. The net effect of this rotation is: (1) to correct the sagittal contour, (2) to pull the apex of the curve back to the midline, and (3) to reduce the rotational deformity of the scoliosis by posterior displacement of the concave side of the apical vertebrae.

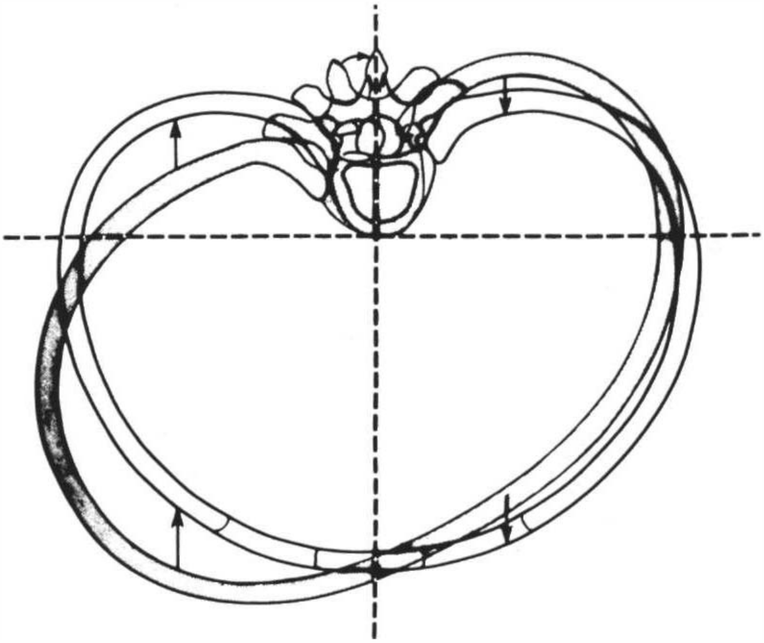

FIGURE 3-16 Correction of the rib hump is permitted by two vectors of forces: (1) pulling the concavity posteriorly at the apex, and (2) pushing the convexity anteriorly. This will be done by the convex rod. (*Denis, F.: Cotrel–Dubousset instrumentation in the treatment of idiopathic scoliosis. Orthop. Clin. North Am., 19:291, 1988.*)

Operation time was longer for the initial cases. Early results suggest a high level of versatility of the C-D system. The potential for three-dimensional correction is an advantage in scoliosis surgery. The authors point out, however, that there must be flexibility to achieve this, and attempts to correct a rigid deformity with this powerful system may lead to failure at the weakest link—hook "pull out." The satisfactory fusion rate and the rarity of instrumentation failure suggest that the theoretical consideration of "stress protection" by a too highly rigid system may be an unwarranted concern. Of course, additional studies are required to fully place this system into its appropriate clinical perspective, particularly in regard to the related risks and benefits.

The importance of the flexibility of each individual curve in the treatment of scoliosis brings forward the biomechanical studies of Viviani and coworkers.[118] Certainly, as suggested, it would be useful to predict preoperatively, by a study of spinal geometry and a mathematical model, an unexpected high stiffness deformity. This can be further refined and used by measuring intraoperative stiffness during distraction. This type of information becomes relatively more important when one uses very high force-generating instrumentation and the associated maneuvers (rod rotation) of the C-D system.

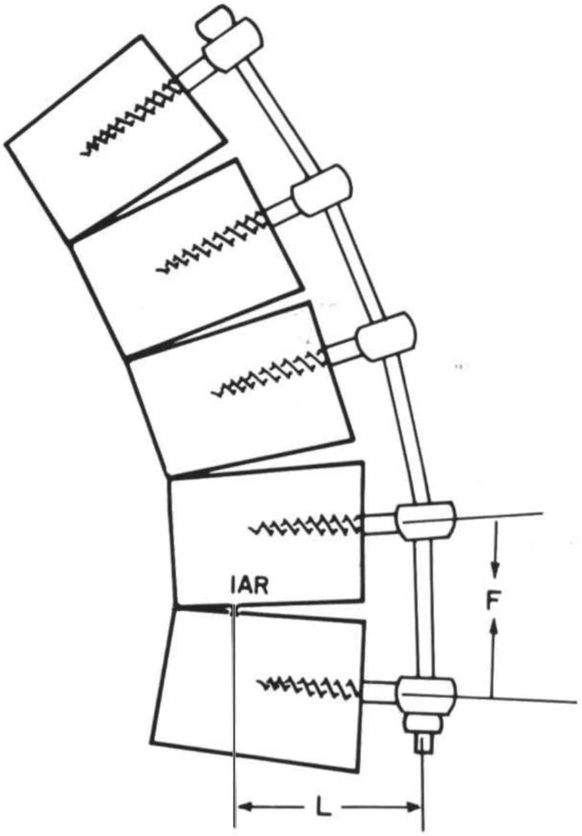

FIGURE 3-17 This frontal plane representation of a scoliosis demonstrates the biomechanical principle of the Dwyer technique—the application of compressive forces on the convex side of the curve. The force F is created by applying tension to the wire on the convex side. The correction is produced by the bending moment F × L.

Dwyer's Technique

The Dwyer technique is another surgical procedure for the correction of scoliosis.[35] The technique is especially valuable where there is a large anterior curvature (lordosis) or where there is absence of the posterior anatomic structures, as with meningomyelocele. The biomechanical principle of Dwyer's method is shown in Figure 3-17. Compressive forces are applied to the spine on the convex side of the curve at each segmental level. This consists of removing the discs and inserting screws into the vertebral bodies on the convex aspect. A braided wire is passed through holes in the screw heads. By applying tension to the wire, corrective bending moments are created at the intervertebral spaces. The magnitude of the force is limited only by the fixation of the cable holes to the vertebrae and is reported to be in the range of 450 N (101 lbf) of tensile force in the braided wire. The correction is maintained by swaging all screw heads onto the wire. Looking at one functional spinal unit, the corrective bending moment provided at each disc space is the tension in the wire multiplied by the distance between the juncture of the screw head and the wire (the lever arm) and the instantaneous center of the functional spinal unit. Since the length of the lever arm is small, a large amount of tension is required in the wires to produce a significant correction. Thus, the spine is subjected to a very large force for a given bending moment. To understand what these forces do to the epiphyseal plates, it is necessary to look at the two vertebrae. In Figure 3-17, the two vertebrae are shown with the instantaneous axis of rotation when a pair of axial forces is applied at the screw heads. With the application of these forces, motion takes place. From the definition of the instantaneous axis of rotation, this motion is a rotation about the instantaneous axis of rotation. Figure 3-17 shows how the application of tension to the wire on the convexity of the curve tends to correct it.

Hall suggested in a review article[49] that the indications for this procedure are: congenital or acquired absence of the posterior elements, severe paralytic scoliosis, thoracic lordosis, or severe lumbar lordosis. With the subsequent development of posterior segmental instrumentation, some of these indications may be relatively less imperative.

Anterior Epiphysiodesis

Anterior unilateral epiphysiodesis on the convex side of the curve in a growing child has been employed by Tylman as a form of treatment in scoliosis.[117] The rationale is that if growth can be arrested on the convex side of the curve, then the concave side can continue to grow, balance the asymmetry, and achieve correction. Tylman also employs posterior arthrodesis on the convex side. Timing and analysis of potential growth to the untreated areas are crucial and difficult to estimate. This technique and concept has been used by Bradford and colleagues,[10] who modified it by adding distraction rods on the concave side of the curve.

Rib Resections

Rib resections on the concave side of the curve have been shown by Halsall and associates[50] to be the most effective mechanism for improving flexibility in scoliosis. Owen and co-workers[86] reported their clinical experience with 42 children and recommend costectomy before Harrington instrumentation. This (1) gives a better cosmetic correction of the rib hump, (2) makes the deformity more flexible, and (3) provides material for bone grafting. Ribs should not be resected in patients with respiratory compromise.

Scoliosis and Respiratory Physiology

The values for respiratory function in young patients with scoliosis may not increase after surgery; they do not decrease either.[135] Kumano and Tsoyoma[59] studied pulmonary function in patients treated with Harrington rods and in patients treated with Dwyer instrumentation. The key findings that related to respiratory compromise were (1) curve located in the thoracic region, (2) rib cage deformity, and (3) thoracic lordosis.

Common to these three factors are reduction and/or distortion of the normal space for the cardiopulmonary organs. The investigators noted a significant gain in pulmonary function in those patients treated with Harrington rods who had a curve of <90° and a 30% correction of their curve. The patients treated with the Dwyer system had no improvement in respiratory function after more than 2 years of follow-up.

More recently, Ogilvie and Schendel[84] studied thoracic volume as it relates to scoliosis correction. They found that thoracic volume was increased most when a thoracic kyphosis was made more normal (20–40°).

Shufflebarger and associates reported improvement in vital capacity and maximum voluntary ventilation 3 months following Cotrel–Dubousset instrumentation in adolescents with thoracic idiopathic scoliosis.

Selection of Fusion Levels

King and co-workers[55] have offered some distinct and well-presented guidelines for determining fusion levels in patients with thoracic idiopathic scoliosis (Figs. 3-18 and 3-19). The authors studied 405 patients and divided the curve into five types. The recommended treatment, based on their work, is as follows.

Type I—Fusion of both curves to the lower vertebra. *Type II*—Selective thoracic fusion to the lower vertebra that is both neutral and stable. When the neutral vertebra and the stable vertebra are not the same, the stable vertebra appears to be more reliable. *Type III*—Fusion to include the measured thoracic curve, with the lower level of fusion ending at the first vertebra that is most closely bisected by the center sacral line. *Type IV*—Fusion to include the measured thoracic curve, with the lower level of fusion ending at the first vertebra that is bisected by the center sacral line. *Type V*—Fusion of both thoracic curves. The lower level should include the vertebra that is most closely bisected by the center sacral line. These guidelines appear sound, based on reason and rational hypotheses about mechanical balance of the erect scoliotic patient. The original publication requires careful study for a full understanding.

Other experience has strongly advocated limiting

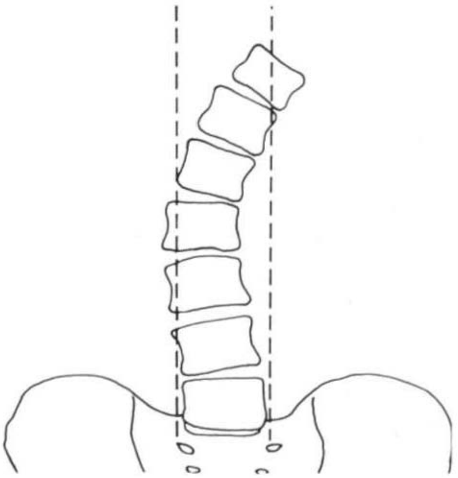

FIGURE 3-18 The stable zone of Harrington, defined by parallel lines drawn through the lumbosacral facets. The vertebral bodies within the lines are in the stable zone. *(King, H.A., Moe, J. H., Bradford, D. S., Winter, R. B.: The selection of fusion levels in thoracic idiopathic scoliosis. J. Bone Joint Surg., 65A:1302, 1983.)*

the extent of the caudad fusions in the lumbar spine[28] and avoiding the lumbosacral junction if at all possible.[38,56] Cochran and associates reviewed a large series of patients in Sweden and found a higher incidence of low back pain in those who had fusions extending down to L4 or L5.[20]

Failure of Fusion and Other Complications in Scoliosis Patients

Failure of fusion may be associated with pain, loss of correction, or dyspnea. There should be adequate extent of fusion initially. Pseudarthroses should be recognized early and treated. Treatment may require a two-stage operation with multiple osteotomies. The goals of surgery are to get the head and thorax over the sacrum and to relieve pain.[24] Direct surgical exploration of the fusion at 6 months postfusion has been suggested by McMaster to definitively recognize pseudarthrosis and treat it early.[72]

In *adult* scoliosis, surgery complications are relatively frequent and significant. Swank and associates reported a 53% rate in 222 patients. These included (1) pseudarthrosis, (2) urinary tract infections, (3) wound infections, (4) instrumentation problems, (5) pulmonary disorder, and (6) loss of lumbar lordosis.[106] Sponseller and colleagues[108] followed 45 patients treated surgically for adult scoliosis. They found that surgery was not helpful in eliminating pain, although self-image was improved. The complications were significant. There was a minor complication rate of 40% and a major complication rate of 23%, including one death from pulmonary embolism.

Erwin and co-workers[37] reviewed records and x-rays of 2,016 patients in a study of the complications of Harrington rod breakage in scoliosis patients. The patients were divided into groups A and B. The breakage rate in group A was 12.5% and in group B was 2.1%. There were three notable differences in the low failure rate of group B. These patients had autogenous bone grafts, facet joint fusions, and were immobilized longer postoperatively (2–4.5 months vs. 6 months). It is difficult to determine the relative impact of these three factors on the Harrington rod failures. Other potentially important mechanical factors, such as rod length and severity of the curves, in the failed rods were not studied.

A synopsis of some key clinical biomechanical

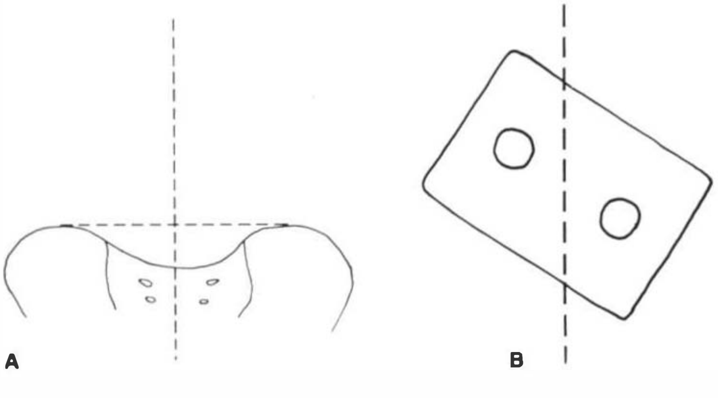

FIGURE 3-19 Center sacral line. **(A)** First draw a line between corresponding points at the top of the iliac crests in the AP x-ray, then erect a perpendicular to that line which also goes through the center of the sacrum. **(B)** The vertebra that is most closely bisected by the line is the stable vertebra. *(King, H. A., Moe, J. H., Bradford, D. S., Winter, R. B.: The selection of fusion levels in thoracic idiopathic scoliosis. J. Bone Joint Surg., 65A:1302, 1983.)*

CHECKLIST: CLINICAL BIOMECHANICS OF SCOLIOSIS TREATMENT

Maintain lumbar lordosis
Correct thoracic lordosis
Use creep and relaxation
Measure forces whenever feasible
Respect the inflection point
Correct two planes and the third will follow
Fuse no further down than L3
Resect the rib hump for best cosmesis
Transverse loading is valuable, especially below 50° curve

principles in the management of scoliosis is in the chart above.

CONCLUSIONS

Scoliosis probably constitutes one of the most challenging and complex clinical problems in the field of orthopedic biomechanics. In some elusive and insidious manner, biological and mechanical factors combine to produce a disease process that can develop into a major cosmetic deformity and, ultimately, ill health as a result of cardiorespiratory changes.

Part 1 of this chapter has sought to bring together the important biological and mechanical considerations that constitute a current understanding of scoliosis.

PART 2: KYPHOSIS

ANATOMIC CONSIDERATIONS

Normal Curves

In the sagittal plane there is a physiologic cervical curve with anterior convexity, a thoracic curve with posterior convexity, a lumbar curve with anterior convexity, and a small sacral curve with posterior convexity (see Fig. 2-30).

With use of a Cobb angle measurement in the sagittal plane, the range of normal thoracic kyphosis supported by the literature is 20° to 50°.[7,11,13,78,96,98,110,120] It goes without saying that all of these references do not report the same range. The range presented is a composite. Nevertheless, it is appropriate as offered, given the best information available currently.

An analogous composite representation of lumbar curvature, based on a review of the literature, is 20° to 70°.[7,31,78,96,110,120]

Voutsinas and MacEwen[120] have appropriately pointed out that the use of the Cobb angle alone does not give a complete sense of the magnitude of the curve and may in fact be misleading. They suggest that the length and width of the curve be observed and quantitated (see Fig. 3-20).

Bernhardt and Bridwell[7] have thoroughly stud-

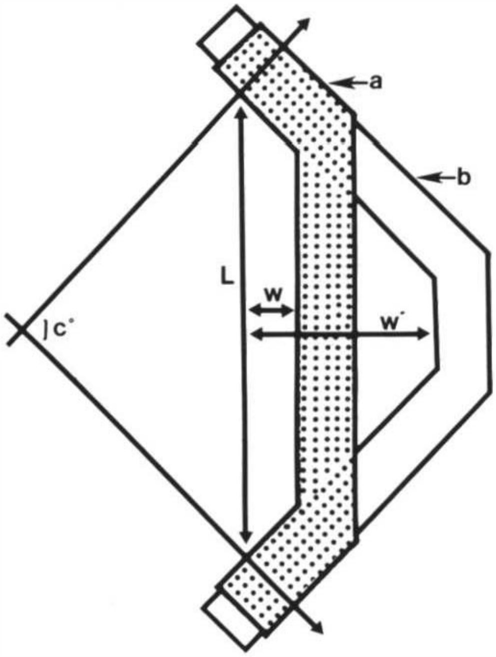

FIGURE 3-20 The two spinal curvatures (a and b) represented by this schematic drawing are obviously quite different in magnitude. However, using Cobb's method to measure the deformities, the degrees of curvature (C°) are identical. The differences in the curves are more accurately reflected when the length of the curves (L) and their respective widths (W and W') are taken into consideration. *(Reproduced with permission from Voutsinas, S. A., MacEwen, G. D.: Sagittal profiles of the spine. Clin. Orthop. 210, 235, 1986.)*

ied the *thoracolumbar junction* and have emphasized their observation that it is essentially straight.

BIOMECHANICAL DEFINITION OF THORACIC KYPHOSIS

Kyphosis is defined in the medical dictionary as "abnormally increased convexity in the curvature of the thoracic spine as viewed from the side; hunchback."[33]

We offer the following biomechanical definition of thoracic kyphosis. There is mainly a sagittal plane curvature with a posterior convexity that measures a Cobb angle greater than 50°. There may be wedging of the vertebrae at one or more levels. There may be cosmetic deformity, pain, or neurologic involvement. The associated disease may be trauma, tumor, osteoporosis, Scheuermann's disease, or some congenital or developmental process. Curves that are within several degrees of the 50° measurement and are not associated with pain or any particular disease process may be difficult to define as abnormal. This is because the opinion as to whether or not there is a cosmetic problem is highly subjective and depends upon the observer. If the patient realistically views curves in the 45–55° range as a cosmetic problem, then by this definition an abnormal kyphosis would be present.

ETIOLOGIC CONSIDERATIONS

The causes of thoracic kyphosis are trauma, tumor, infection, and congenital and developmental conditions. The biomechanics of the initiation and progression of kyphosis in association with trauma are presented in Figure 3-21. There is loss of the anterior support structure, which normally sustains significant compression loads. There may also be a failure of posterior components, which are primarily loaded in tension. This combination may result in kyphotic deformity. Anterior support may also be lost because of osteoporosis, tumors, or infection (usually tuberculosis) of the vertebral body either directly or as a result of treatment. Scheuermann's disease may be caused by abnormal growth in the anterior margin of several of the thoracic vertebral bodies, resulting in kyphosis. Anterior bars and various relatively rare congenital defects may also lead to abnormal thoracic kyphosis.

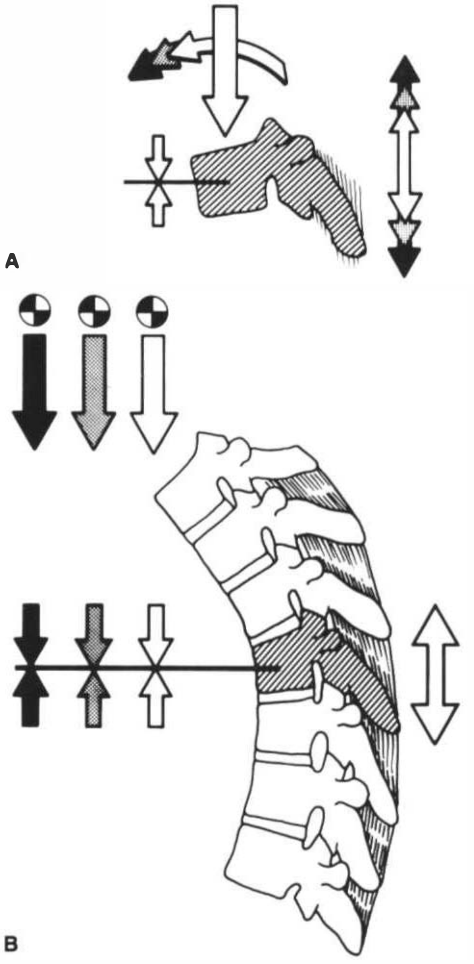

FIGURE 3-21 The various structural and mechanical factors that may contribute to kyphotic deformity are shown here. **(A)** The anterior elements are in compression and the posterior ones in tension. **(B)** Progressive kyphosis increases the moment arm, which adds relatively more compression and tension. Various structural changes anteriorly and posteriorly may contribute to or facilitate progression of deformity.

Analysis of Kyphotic Deformity

The thoracic spine is subjected to compression and flexion (bending moment) as a result of the location of the center of gravity, which is anterior to the thoracic vertebrae. Because of angulation or wedging of a vertebra within the thoracic curve, the flexion bending moment may be significantly affected, even though the compression does not change. The changes in the stresses in the spine are directly proportional to the bending moment acting on it; thus, we present a simple biomechanical analysis that provides some estimates of these increased stresses.

A decrease in the anterior part of the vertebral body height, caused by injury or disease, results in angulation of the vertebra and, therefore, an increase in kyphosis. The biomechanical consequence of this deformity is an increase in the lever arm of the center of gravity line with respect to the injured vertebra. The resulting *increase* in the flexion bending moment is a function of the angulation of the injured vertebra or wedging, as well as the original curvature of the kyphosis. The vertebral wedging may be expressed in degrees or in terms of the anterior body height ratio. The latter is defined as the ratio of the anterior to posterior vertebral body heights as seen on a lateral x-ray. Both the wedge angle and the anterior body height ratio can be directly measured from an x-ray and are not affected by radiographic magnification.

We constructed a simple mathematical model of a thoracic spine (Fig. 3-22). The center of gravity (c.g.) of the trunk (more precisely, the body parts above the vertebra) is anterior to and above the vertebra being investigated (Fig. 3-22A). The left figure shows the trunk–vertebra relationship *before* the wedging, while the right figure shows the same situation but *after* the wedging of $\alpha°$.

The spatial relationship between the center of the vertebral body and the center of gravity is defined by (1) the line L joining the two centers, (2) the angle θ formed by the line L and the center of gravity line *before* wedging, and (3) the wedge angle α. By changing the angle θ, one may simulate thoracic spines of varying degrees of curvature. A few other parameters were defined to complete the model. The angulation of the injured vertebra (the wedge angle α) was simply the angle formed by the two end-plates of the affected vertebra (Fig. 3-22B). The anterior body height ratio was defined as the anterior body height H_A divided by the posterior body height H_P.

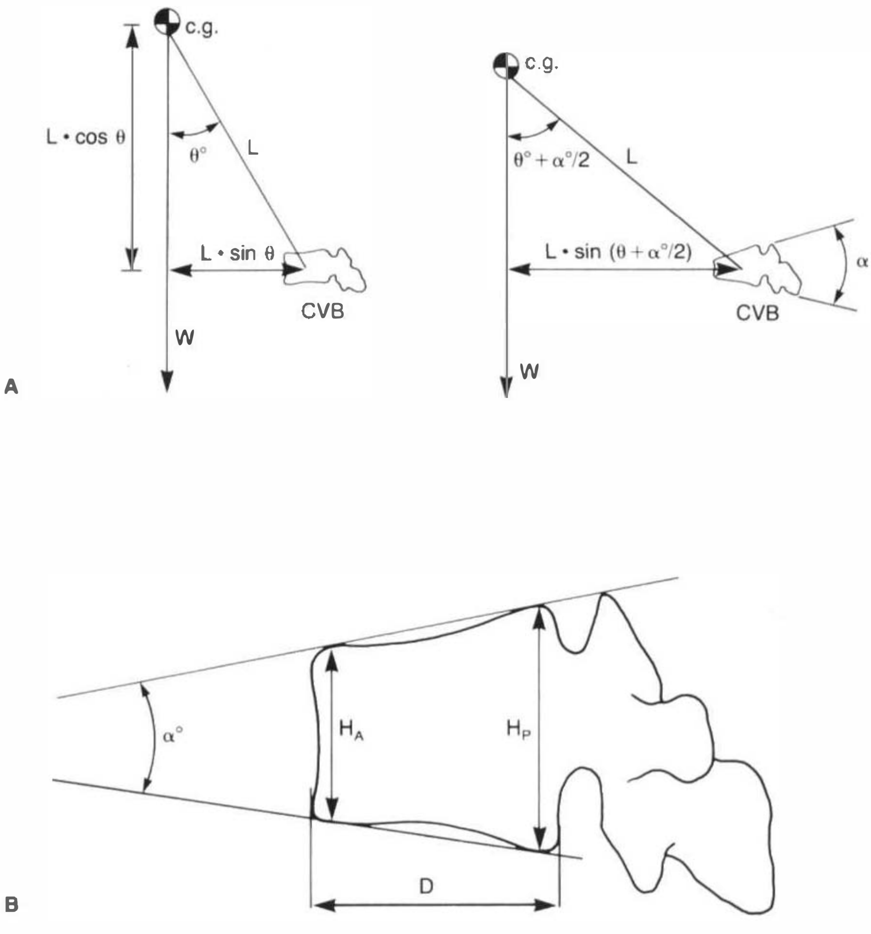

FIGURE 3-22 A model to analyze the effect of vertebral wedging on the bending moment in the thoracic spine. **(A)** On the left is the model before injury, defining relationship between the center of gravity (c.g.) of the trunk and center of the vertebral body (CVB). L is the distance between the two centers, and θ is the angle formed by the c.g.–CVB line with the vertical line of gravity. The model after wedging of $\alpha°$ is seen on the right. Note that there is increased lever arm of the trunk c.g. with respect to the vertebra. **(B)** The vertebral wedging may be defined by the wedge angle α or by the anterior body height ratio. The relationship between these two parameters is depicted. Results of the analysis are shown in Figure 3-23. See text and Note D for details.

Finally, D was the AP diameter of the vertebral body, as measured on the lateral x-ray. (For the present analysis, we assumed $H_P/D = 0.75$.)

Using the equations developed,[D] we computed the resulting increases in the flexion bending moment due to the vertebral wedging. We chose two different thoracic curvatures to study their effect on the bending moment increases. The results, in the form of a relative increase in bending moment, defined as a percentage increase over the normal bending moment due to wedging, are depicted in Figure 3-23. The increased bending moment versus the wedge angle is shown in Figure 23A, while the same increase as a function of the anterior body height ratio is shown in Figure 23B. In each figure there are two curves representing thoracic spines of two different curvatures. The upper curves depict the results for the case of a thoracic spine of mild kyphosis (angle $\theta = 50°$). The lower curves are for a relatively more kyphotic spine (angle $\theta = 80°$).

In general, we found an increase in the relative bending moment due to wedging of the vertebral body. The increase was greater for a less kyphotic thoracic spine. Thus, the vertebral wedging, after injury, for example, carries a risk of further wedging with time. This risk is greater for a relatively straight spine (before injury). It should be emphasized that our biomechanical analysis is rather simple. Its main purpose was to bring to attention the risk of progressive wedging. However, it is a theory with no clinical documentation at the present time but one that perhaps can be looked at in the future.

Experimental Studies

Although there are numerous experimental studies of the etiology of idiopathic scoliosis, there is little in the way of contemporary studies of the cause of kyphosis.

Clinical Studies

Mayfield and associates[70] showed that children treated with >3,000 rads of radiation for neuroblastoma were prone to develop either kyphosis or scoliosis. Of the patients surviving for 5 years, 70% had spinal deformity. Of those with deformity, 16% were kyphotic and 50% were scoliotic. The presumed mechanism is growth arrest of the anterior portion of the vertebral growth plate. Laminectomy (disruption of the tensile load-bearing posterior elements) is also thought to be a contributing factor in the development and progression of kyphosis. The kyphotic deformities progressed 3° per year, and the scoliotic deformities progressed 1° annually.

Ippolito and Ponseti[53] completed histochemical studies of the growth plates of a 16-year-old male with juvenile kyphosis who was killed in an auto accident. Abnormal histological and histochemical changes were noted. The growth was thought to be stunted because of an abnormal cartilage in the growth plate. The abnormal cartilage appeared to have a paucity of glycoproteins, a different type of collagen, and some qualitative and/or quantitative abnormalities of the proteoglycans.

Etiologic Theories

These clinical studies suggest that when the growth plate is interfered with, an imbalance occurs, and alterations in the loading patterns tend to advance the deformity. Certainly, structural changes associated with tumor, trauma, and surgical procedures can alter the mechanics so as to initiate or perpetuate the development of a kyphotic deformity.

The initiation and progression of all types of kyphotic deformities may be thought of as a dynamic equilibrium in which there is normally a balance between factors that may increase the bend of the thoracic spine and those which tend to resist or maintain it. There are structural and mechanical considerations to be considered. These can be evaluated through an analysis of the anterior elements, which are loaded primarily in compression, and the posterior elements, which are loaded primarily in tension. The effect of the increasing bending moment, which occurs with progression of the deformity, should also be considered. This method of analysis is presented in Figures 3-22 and 3-23, along with the specific biomechanical role played by various clinical factors and conditions.

PROGNOSIS

Although there was little to be found in the way of experimental or clinical studies related to prognosis, some guidelines can be extrapolated from biomechanical analysis. The simple model provided in Figure 3-22 helps one to associate approximate increases in bending moments that occur with decreases in anterior vertebral height (increases in ver-

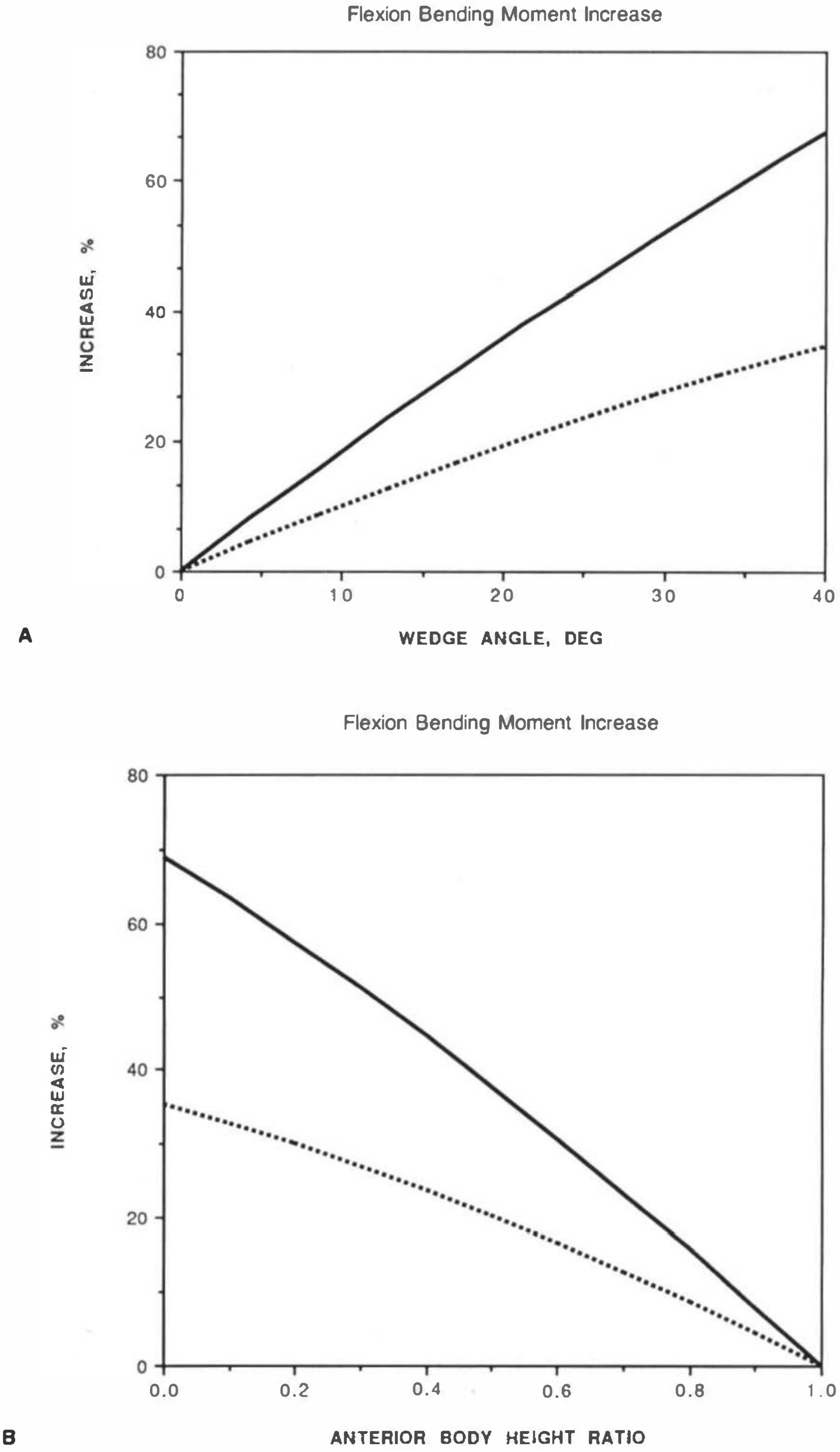

FIGURE 3-23 The results of a simple biomechanical analysis of increased risk of kyphotic deformity with time, after a vertebral wedging has occurred. **(A)** Flexion bending moment (%) increases as a function of the wedge angle (degrees). The above curve is for a more straight and the lower curve is for a more kyphotic thoracic spine. **(B)** Results similar to those of A, but expressed as functions of the anterior body height ratio. For details, see text, Figure 3-22 and Note D.

tebral wedging). Increased wedging and increased moment arms both result in increased kyphotic angulation.

TREATMENT OF THORACIC KYPHOSIS

Update of Biomechanical Considerations in the Treatment of Thoracic Kyphosis

We suggest that the clinical biomechanics of the deforming factors as outlined in the preceding section be reviewed. With that background, the application of various therapeutic modalities can be readily evaluated and strategically selected. The specific therapies are discussed in the subsequent paragraphs and in other chapters in more detail.

The major goals are: (1) to correct deformity, (2) to diminish deforming forces, (3) to compensate for structural alterations with biological and/or nonbiological implants, and (4) to achieve adequate fixation with implants and bone grafts. Nonoperative treatments attempt to correct deformity and maintain the correction.

Exercise

Active exercise is not usually offered as a treatment of kyphosis. However, transcutaneous muscle stimulation has been used in the treatment of kyphosis.[6] The investigators reported an average correction of 13° of kyphosis deformity in 18 patients. In view of the dismal results in several studies that evaluated the role of electrical stimulation in correcting scoliosis, it is highly unlikely that this will prove effective in the treatment of kyphosis.

Orthotic Treatment

Please see Chapter 7 for the use of the Milwaukee brace in the treatment of kyphosis. This remains the orthotic treatment of choice in our opinion. However, it is important to know the following. In a study of 75 patients using the Boston brace (for which patient compliance is better), the device worked well in compliant patients with curves $<70°$.[47] A theoretical advantage is that this brace corrects lumbar lordosis and thus causes some compensatory correction of the thoracic kyphosis.

Surgery

The surgical procedures for kyphosis may be categorized into anterior, posterior, combined, and miscellaneous surgical considerations (Fig. 3-24). Our view of the most salient updated information on surgery for thoracic kyphosis follows.

Anterior Procedures

Several devices have been developed and made available for anterior spine surgery. Some have the capability of serving as a kind of internal distractor of the vertebral bodies in order to reverse the kyphotic deformity. The one designed by Pinto and colleagues,[93] which operates on a turnbuckle principle, is not used as an implant but only to distract at the time of surgery.

Posterior Procedures

A preliminary report of the use of Cotrel–Dubousset (C-D) instrumentation in kyphosis and other conditions has been published by Gurr and McAfee.[46] Three patients were treated for post-traumatic kyphosis. Two of these were treated by anterior releases prior to posterior instrumentation. The C-D system was technically helpful in correcting the kyphosis, and solid bony union was achieved in all three patients.

Combined Anterior/Posterior Procedures

Bradford and co-workers[12] reported their experience with the anterior and posterior approach in the treatment of 24 patients with Scheuermann's disease. An anterior fusion was performed and was followed within 2 weeks by a posterior fusion. The authors expressed the opinion that the combined approach was superior to the posterior technique alone. Two staged combined approaches have been described by other surgeons.[63]

There are several published experiences concerning the management of post-traumatic kyphosis with a combined anterior and posterior approach.[69] The surgery was done for treatment of pain, progressive kyphosis, increasing neurologic deficit, or any combination of the three. Here, too, as suggested in Figure 3-24, simple laminectomy was not desirable. Nonoperative treatment was unsatisfactory. The experience of the preceding investigations was essentially confirmed in a subsequent work by Roberson and Whitesides.[100] Operating on post-traumatic tho-

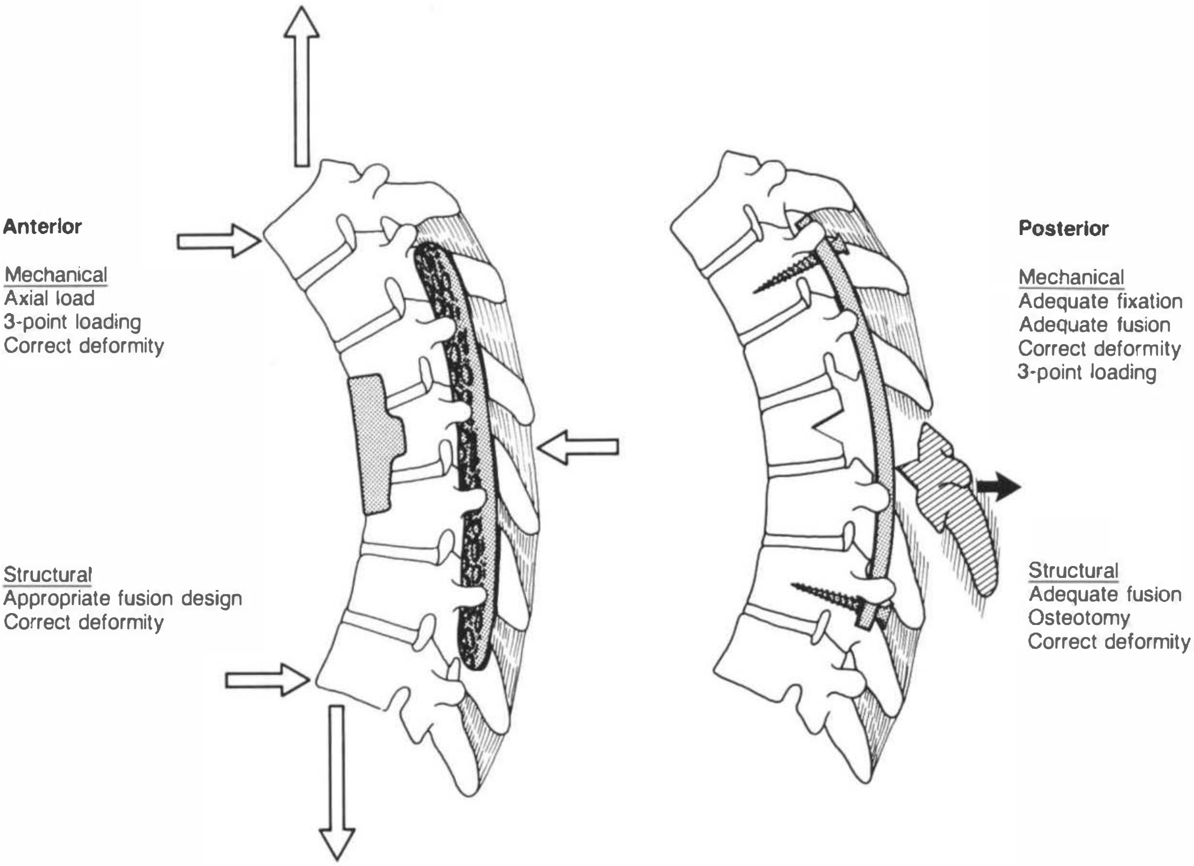

FIGURE 3-24 This is a summary of some key biomechanical factors in kyphosis correction. The three-point loading concept is helpful in orthotics. Correction of deformity decreases bending moments. The bone graft, whether anterior, posterior, or both, must be adequate. Instrumentation should be rigid enough to maintain position. Osteotomy may sometimes be needed to correct the deformity.

racolumbar kyphosis for essentially the same reason, these investigators employed anterior, posterior, or combined procedures. The authors reported no superiority in any of the three basic approaches.

Miscellaneous Surgical Considerations

Ryan and Taylor[103] reported three patients who had short, sharp, thoracic kyphosis due to Scheuermann's disease with a sharp kyphosis in the lower thoracic region. The key factors involved in neurologic compromise were: (1) an acute angle of the kyphosis, (2) the number of segments involved, (3) the rate of change of angulation at the apex of the curve, (4) local anatomic variations, (5) trauma, and (6) impairment of the vascular supply of the cord.

Kostuik and colleagues[57] presented their experience with one staged anterior and posterior approach for the management of iatrogenic lumbar kyphosis. This "iatrogenic flat back syndrome," as it was called, was treated with an anterior opening wedge osteotomy with Kostuik–Harrington instrumentation followed by a posterior closing extension osteotomy. Dwyer cables and screws were used posteriorly. The report was based on 44 patients and an average follow-up of 4 years. The major factor indicating surgery was the complaint of pain. Pain was relieved in 90% of the patients. The importance of maintaining some lordosis at the time of initial surgery was emphasized.

SPINE DEFORMITIES—TRADITIONAL AND FUTURE

We suggest that it is reasonable that we measure scoliosis mainly in the *frontal* plane with appropriate attention to measuring it in the *horizontal* plane (vertebral axial rotation) and in the *sagittal* plane (kyphoscoliosis and lordoscoliosis). Kyphosis and lordosis we continue to measure fairly exclusively in the sagittal plane, and this seems to be satisfactory.

Because the clinical biomechanics of the spine is an evolving field that has demonstrated its capacity to become more scientific and quantitative, and because deformities of the spine are *truly* three-dimensional deformities, we wish to submit the following for the reader's consideration. Continuing to think of scoliosis as a combination of three snapshots in the anteroposterior, lateral, and horizontal traditional planes is an oversimplification that belies contemporary knowledge and sophistication. Perhaps it is time to further recognize the spinal deformity as occurring in all components of three-dimensional space. The helical nature of the deformity is better appreciated as depicted in Figure 3-25. It is possible to miss significant components and aspects of the deformity if it is viewed, analyzed, and thought of in just the three traditional planes (i.e., sagittal, coronal, horizontal). Since it is not clinically practical to routinely view the deformity in full three-dimensional space, we submit for your consideration two selected additional planes for viewing, analyzing, and describing scoliosis. Two additional views, namely a right anterior oblique and a left anterior oblique view are suggested. It may prove useful to measure the additional views for research purposes as well as for evaluating the efficacy of various methodologies of internal correction and fixation.

Deacon and colleagues[27] showed the following with 11 articulated scoliotic spines. A Cobb angle of 70° was measured on anteroposterior radiographs. Then the Cobb angles were measured on true anteroposterior radiographs of the *deformity*. This measurement revealed a mean Cobb angle of 99°, which was 41% greater than the initial measurement on the traditional anteroposterior plain radiographs. There were also discrepancies between the measurement of kyphosis (41°) on a true lateral radiograph and the measurement of 14° true apical lordosis when the true lateral projection was measured. These investigators have shown that the three-dimensional nature of scoliosis definitely can "change" in both magnitude and character, depending on the plane of its radiographic projection.

DeSmet[31] and associates presented the top view analysis of the spine to further describe the scoliosis deformity. The spine is viewed from above and the vertebrae are schematically represented by a series of overlapping rectangles. From the horizontal plane view, both the frontal and sagittal plane curves can be visualized (Fig. 3-25).

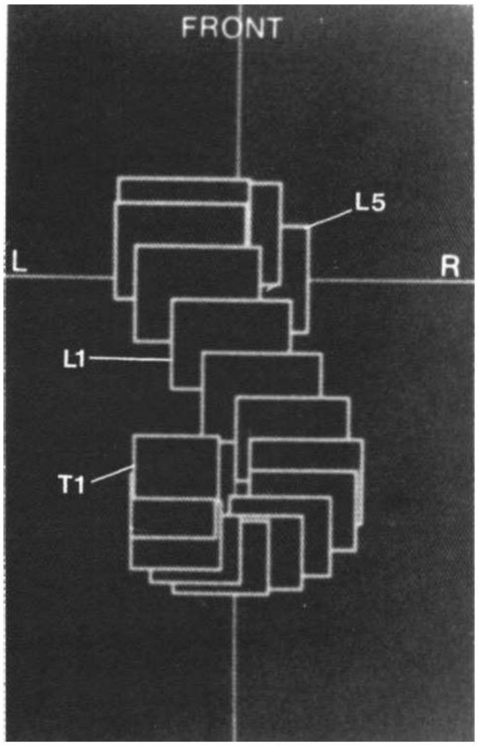

FIGURE 3-25 Top view as if the spine is viewed from above with overlapping of the rectangles representing vertebrae from T1 above to L5 on the bottom. The spine is seen not to deform in one plane but to curve like a helix. The lumbar spine has mostly lordosis without much side-to-side swing. The thoracic spine sweeps far to the right before ending out of balance to the left at T1. (DeSmet, A. A.: *Radiographic evaluation*. In DeSmet, A. A. (ed.): *Radiology of Spinal Curvature*, pp. 23–58. St Louis, C.V. Mosby, 1985.)

The development of a system for the practical quantitative three-dimensional analysis and description of spinal deformity constitutes an important challenge for the clinical biomechanics of the future.

■ CLINICAL BIOMECHANICS

Scoliosis

■ Removal of posterior elements allows increased axial rotation in the thoracic spine.

■ Biomechanical theories suggest that removal of the disc may also improve correction potential.

■ Three-dimensional analysis of scoliosis is crucial to a comprehensive understanding of the disease. The sagittal plane component may have been underemphasized.

■ The rib vertebral angle difference (RVAD) at the apical vertebra of >20° is a useful predictor of progression of infantile idiopathic scoliosis.

■ Skeletal immaturity and thoracic curves of 50–80° are useful predictors of progression of idiopathic scoliosis.

■ Axial rotation is important in terms of prognosis, rib hump deformity, and treatment.

■ The surgeon may take advantage of the creep and relaxation characteristics of the tissues primarily by allowing ample time to pass between application of various correctional forces.

■ The surgeon may take advantage of the analysis of the relative value of axial, transverse, and combined loading in the correction of scoliosis. Theoretically, axial loading is more efficient for severe curves and moderate curves, and transverse loading is more efficient for less severe curves. Combined loading is always more effective than either type alone.

■ The *inflection* point is an important concept in the treatment of scoliosis. When there is a rapid increase in incremental forces associated with correction, this is a sign that the weak link in the system (usually the spine) is approaching its tolerance limits.

■ Whenever possible, it is good clinical biomechanics to measure the therapeutic forces that we exert on the spine.

■ Certain factors are important in using the Milwaukee brace. Evidence suggests that the brace is as effective as a cast in reducing axial forces adjacent to a scoliotic curve. Removal of axillary supports and thoracic pads increases the axial forces and thus reduces the effectiveness of the brace. The brace continues to resist axial loads in the spine when the wearer is recumbent. Thus, the brace should also be worn when the patient is in this position.

■ The major practical considerations regarding the use of Harrington instrumentation are as follows: The distraction force that may be applied with the Harrington rod is determined by the tolerance of the thoracic lamina; 295 N (65.8 lbf) is the upper limit of this force; coughing or bucking during the stages of recovery from anesthesia can apply dangerously high forces to the Harrington rod; greater surface contact of hook to lamina and smaller increments between notches on the rod increase the tolerance limits of the mechanism; compression rods on the convex side have little or no correctional value.

■ The Dwyer technique is a biomechanically sound and effective technique that has the additional advantage of applying asymmetrical loads to the epiphyseal plate.

■ Initial reports suggest that the Cotrel–Dubousset instrumentation system is useful for the rigid fixation of several spine conditions. The system also has the capability of correcting scoliosis deformity in all three planes simultaneously.

Kyphosis

■ The clinical biomechanics of kyphosis involve an equilibrium between the compressive forces borne by the anterior elements and the tensile forces borne by the posterior elements.

NOTES

[A] Using the coordinate system, when axial rotation is coupled with lateral bending, a negative θz is associated with a positive θy.

[B] In the middle and lower regions of the thoracic spine, a negative θz is associated with a negative θy.

[C] Figure 3-7C is a free-body diagram of the link BC and the spring C under axial load. The equilibrium condition for the link BC gives the expression for the bending moment M_a at the junction C:

$$M_a = FL \text{ Sin } (\theta/2) \qquad (1)$$

where F is the maximum safe load applied to the vertebra in the axial direction. L is the link length, and θ is the angular deformity. Equation (1) shows that, for given values of F and L, the bending moment M_a varies as a sine function of angle (θ/2). A free-body diagram for the transverse loading is shown in Figure 3-8C. Again, the equilibrium condition at the junction C gives the expression for the bending moment:

$$M_t = \frac{FL}{2} \text{Cos } (\theta/2) \qquad (2)$$

Here F is the transverse force applied to the middle of the curve and F/2 are the reactive forces. In this case, for given values of F and L, the bending moment M_t is proportional to half of the cosine of angle (θ/2). One may take this analysis a step further and study the effect of combining axial and transverse loads. Such a combined situation is shown in Figure 3-9. Using equal loads at all three loading points, A, B, and C, the axial components of forces at A and B are 0.87F, while the horizontal components are F/2. (The end force vectors are tilted 30° toward the center force.) From the free-body diagram, Figure 3-9C, and the equilibrium conditions for the link BC, we can write the expression for the bending moment at C caused by the combined loading. This is

$$M_c = FL \text{ Sin } (30° + \theta/2) \qquad (3)$$

Let us assume that the center of gravity of the trunk is a certain distance (L • Sin θ) anterior to and a certain distance (L • Cos θ) above the injured vertebra (Fig. 3-22A, left side). Here, L is the distance between the vertebral body center and the center of gravity of the body parts above the vertebra, while θ (theta) is the angle between the line L and the line of gravity. If W is the weight of the trunk (more precisely, the weight of the body parts above the vertebra), then the flexion bending moment M acting on the vertebral body, before the injury, is given as

$$M_{\text{NORMAL}} = W \cdot L \cdot \text{Sin } \theta \qquad (4)$$

Now let us assume that, as a result of injury or disease, there is vertebral wedging of α° (alpha degrees) (Fig. 3-22A, right side). The new bending moment is

$$M_{\text{INJURY}} = W \cdot L \cdot \text{Sin } (\theta + \alpha/2) \qquad (5)$$

Thus, the percentage increase in the bending moment is

$$M_{\text{RELATIVE INCREASE}} = 100 \times [(\text{Sin}(\theta + \alpha/2)/\text{Sin } \theta) - 1] \qquad (6)$$

As we see from this equation, the relative bending moment increase is both a function of the angulation α° of the vertebra due to injury as well as a measure θ° of the original thoracic curve.

The above relationship may be expressed in the form of the anterior body height ratio. First, the wedge angle, depicted in Figure 3-22B, is related to the anterior body height ratio in the following manner:

$$\text{Anterior Body Height Ratio} = \frac{H_A}{H_p} = \left(1 - \frac{2\,D}{H_p} \text{Tan } \alpha/2\right) \qquad (7)$$

where D is the AP diameter of the vertebra and H_A and H_P are the respective anterior and posterior body heights. By combining the above two equations and eliminating the variable α, the $M_{\text{RELATIVE INCREASE}}$ is expressed in the form of the anterior body height ratio.

Results for two different thoracic curvatures, defined by θ equal to 50° and 80°, are shown in Figure 3-23.

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Practical Biomechanics of Spine Trauma

A bricklayer injured in a work-related accident was asked by his insurance company to explain how the accident happened. He filed the following report, which "says it all" in terms of clinical biomechanics and trauma. Here is the letter:

In response to your request for additional information in block number 3 of the accident reporting form, I wrote "poor planning" as the following details will be sufficient. On the day of the accident, I was working alone on the roof of a new 6-story building. When I finished my work, I discovered that I had about 500 pounds of bricks left over. Rather than carry the bricks down by hand, I decided to lower them in a barrel by using a pulley which, by luck, was attached to the side of the building at the 6th floor.

Securing the rope at ground level, I went up on the roof, swung the barrel out and loaded the bricks into it. I went back to the ground and untied the rope holding it tightly to insure a slow descent of the 500 pounds of bricks. You will note in block number 11 of the accident reporting form that I weigh 135 pounds.

Due to my surprise at being jerked off the ground so suddenly, I lost my presence of mind and forgot to let go of the rope. Needless to say, I proceeded at a rather rapid rate up the side of the building. In the vicinity of the 3rd floor, I met the barrel coming down. This explains the fractured skull and the broken collar bone. Slowed down only slightly, I continued my rapid ascent, not stopping until the fingers of my right hand were two knuckles deep into the pulley. Fortunately by that time, I had regained my presence of mind and was able to hold tightly to the rope in spite of my pain.

Approximately at the same time, however, the barrel of bricks hit the ground and the bottom fell out of the barrel. Devoid of the weight of the bricks, the barrel now weighed approximately 50 pounds. I refer you again to my weight in block number 11. As you might imagine, I began a rapid descent down the side of the building. In the vicinity of the 3rd floor, I met the barrel coming up. This accounts for the two fractured ankles and lacerations of my legs and lower body.

The encounter with the barrel slowed me down, however, and fortunately when I landed on the pile of bricks only three vertebrae were cracked. I am sorry to report, however, that as I lay there on the bricks in pain, unable to stand, and watching the empty barrel six stories above me—I again lost my presence of mind—and I let go of the rope.

—Adapted from Neil Chayet, "Looking at the Law," WEEI radio, Boston, Massachusetts. June 29, 1985.

4

A good deal is known about the mechanism of injury in spine trauma. However, there are also a number of assumptions that can lead to copious, sometimes erroneous, conclusions in the analysis of spine trauma. It is not uncommon for physicians to look for 2 to 3 seconds at a lateral radiograph of the cervical spine and then to embark upon detailed deliberations about the directions and magnitudes of the forces responsible for the observed injuries. In many of these situations, the "learned dissertations" constitute inappropriate, unfounded speculations. Given the current thrust in the direction of a more scientific engineering study of clinical phenomena, this type of speculation is no longer acceptable. However, such an analysis cannot yet be completely replaced by valid scientific data. The current state of knowledge lies somewhere between gross speculation and precise science. This chapter seeks to go beyond explanations based on speculation and imprecise understanding of mechanics to offer a framework for a more sound analysis.

GENERAL CLINICAL CONSIDERATIONS

In the following discussion of the suggested treatment of spine injuries, we assume that the fundamentals of emergency treatment and basic care of the patient with an injured spinal cord are being carried out. These include the usual attention to the airway and treatment of hemorrhage and shock. We assume that adequate fluid replacement is initiated and the splinting of injured parts is carried out whenever necessary. The patient should be treated with due consideration of early drainage of the bladder and careful monitoring of vital signs and neurologic status, along with the recording of a medical history and physical examination. Initial care should also include radiologic evaluation (with a high index of suspicion looking for skip fractures), baseline laboratory studies, and the indicated consultations. Collaboration among orthopedist, bioengineer, neurosurgeon, and radiologist can be invaluable in piecing together the possible mechanism of injury and evaluating the damage to important, stabilizing, anatomic structures (see Chapter 5).

MECHANISM OF INJURY

The mechanism of injury is of great importance in the complete understanding of spine trauma. An analysis of injury mechanisms is beneficial and practical and can be helpful in choosing the technique of reduction and management of certain injuries. A recognition of the resultant injury as opposed to the mechanism of injury is also important. There are "families" of injuries that result from identical or similar mechanisms of injury (Fig. 4-1). With a thorough understanding of the biomechanics involved, the association of certain injuries may come to be expected.[49]

To build a system for evaluating injury mechanisms, it is necessary to challenge, if not eliminate, some of the current "ways of thinking" that may create difficulties with terminology. Physicians speak of flexion injuries, extension injuries, rotation injuries, lateral bending injuries, and compression injuries. Studies of kinematics and analysis of forces in the functional spinal unit have shown that these types of traditional descriptions of patterns of motion in the spine are no longer adequate to completely explain spine mechanics and injury mechanisms. The traditional concept of flexion or extension is not one simple motion but involves two types of motion, such as translation and rotation in the sagittal plane. Depending upon the force vector, markedly different patterns of motion as well as vertebral deformation may be produced. Another case in point is lateral bending, which involves rotations about the horizontal and vertical axes, respectively, as well as translation perpendicular to the sagittal plane. In other words, lateral bending may cause any combination of transverse shear in the horizontal plane, rotational shear about the vertical axis, and tensile and compressive stresses in the vertebral bodies. Therefore, to assume that a mechanism of injury involves only lateral bending is an oversimplification. In order to develop a system for describing mechanisms of injury, the analysis may be carried out with the functional spinal unit as the analytical unit. The load–displacement and failure patterns may then be analyzed and described in that context.

Any of the six degrees of freedom of a functional spinal unit can carry with it an associated force or bending moment (torque). Kinematics explains that the coupling patterns are inherent to the functional spinal units in most regions of the spine (see Chapter 2). Orientations of different functional spinal units in three-dimensional space at the time of injury modify these coupling patterns.

Crucial to an analysis of the forces involved is an understanding of the instantaneous axis of rotation (IAR). This axis dictates the pattern of deformation that occurs in the functional spinal unit as a function of its relationship to the force applied. For example, a vertical force applied anterior to the IAR causes flexion. The same force applied posteriorly to the IAR produces extension. Therefore, in analyzing the mechanism of injury, the functional spinal unit or units involved must be evaluated, considering the six possible degrees of freedom and the various IARs due to application of different force vectors.

After taking into consideration the varying degrees of freedom of movement of the vertebra, it is then important to analyze some of the characteristics of the forces and moments that may be applied. Generally, in civilian injuries, forces come in at some slightly inclined angle, off one of the three orthogonal axes, x, y, or z. As a result of clinical habit, patterns, and reality, the large majority of analyses have been involved with sagittal plane motion, with the injuring force vector presumed to be somewhere in the sagittal plane. In other words, vertical compression and flexion/extension mechanisms tend to predominate in the injuries physicians observe. In airplane ejection injuries there is nearly vertical force, with minimal eccentricity close to the y-axis in the sagittal plane. Even here, the various regions of the spine are subjected to flexion, compression, or extension because of varying curvature of the spine. Civilian flexion/extension injuries include whiplash, bilateral facet dislocations, and hangman's fractures.

The forces vary tremendously in the rate of application and magnitude. Because of the difference in stiffness and energy-absorbing capacities of the ligamentous structures and the bone, it is theoretically possible that the failure point of the bone–ligament complex of the functional spinal unit may vary as a function of the magnitude, direction, and rate of application of these forces. This factor may explain some of the discrepancies and disagreements in the literature concerning whether the bone or the liga-

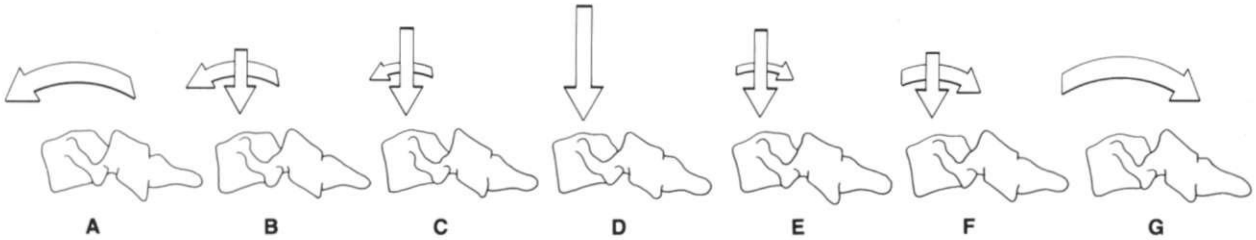

FIGURE 4-1 Load spectrum for mechanism of injury. These diagrams represent a fundamental theoretical construct upon which vertebral mechanisms of injury may be analyzed, and focuses on a sagittal plane (y–z plane) analysis. The view is that most injuries in the sagittal plane are the result of some combination of a compression force (−y-axis) and a bending moment (±x-axis). The middle vertebra (**D**) is subjected to a pure compression force. Just to the left the vertebra (**C**) is subjected to a large compression force and a modest bending moment. The next vertebra (**B**) is subjected to a somewhat modest compression force and a relatively large bending moment. To complete the spectrum, yet another vertebra (**A**) to the left would have a pure bending moment. These +x-axis moments would constitute injuries in the flexion mode. In these loading conditions there may also be an element of shear. This is associated with the bending moment. This shear component is not represented by an arrow.

Conversely and analogously, we have the same spectrum of mechanisms going to the right, beginning with the first vertebra (**E**) and ending with the vertebra (**G**) subjected to pure extension bending moment. These injuries would be in the extension mode.

The pure compression force gives a compression fracture with central concavity and possible comminution. As we go more to the left we see progressively more wedge compression of the vertebral body. At the far left, with a pure bending moment, one expects extensive tensile failure of the posterior elements.

As we move to the right of the middle compression example, one expects fractures of the lateral masses and the lamina. There is, of course, some element of shear induced in these injuries; this, however, is not depicted in the diagram.

Another aspect of the hypothesis is that because of this spectrum of load combinations there may be patterns of injuries such that when one sees one there is a warning that other associated injuries are to be suspected. An example is the flexion injury in the athlete in which there is a major bending moment with a relatively small compression force. This results in a modest wedge compression fracture and extensive disruption of the posterior ligamentous structures, having failed in tension. The families or patterns of injury suggest that when we see wedge compression fracture we suspect disruption of the posterior elements.

mentous structure fails first. An example in point is the controversy of the fracture of the dens versus the rupture of the transverse ligament of the dens. Although there have been no experiments designed specifically to test this in the spine, it may be that the rate of application of the load determines whether the ligament or the bone fails first.[49] The work of Noyes and colleagues shows that such a phenomenon is operative elsewhere in the body.[178] There is another possibility to be considered. For a given individual, one structure may be stronger than the other. There is adequate basis to assume that there is a broad variation in the strength of the dens, given the numerous congenital variations in the ossification centers and the complex lines of potential weakness due to failure of normal maturation and fusing together of these centers. These considerations exemplify the complications that may be involved in an analysis of the mechanism of injury.

An understanding of normal kinematics in the various regions of the spine is essential. For example, the amount of rotation that is permitted about the y-axis (in the horizontal plane) in the cervical spine is significantly greater than that observed in the lumbar spine. The alignment of the facets in the lumbar region does not allow much rotation about the y-axis. When the normal range of rotation about the y-axis is forcefully exceeded, there is fracture or disruption of the lumbar facets and/or posterior elements (see Fig. 5-54).

The spine has certain characteristics that predispose different areas to injury. The classic example is the thoracolumbar area. Predisposition to injury here is thought to be due largely to the stress concentration imposed by the abrupt decrease of the stiffness from the thoracic to the lumbar spine. This is due to a rather abrupt change of the alignment of the facet joints, from the thoracic type of orientation to that of the lumbar, which involves a rotation of the facet joint planes of almost 90° about the y-axis (see Fig. 2-23). Most studies of injury have shown this region to be the most frequent site of spine fractures. However, Griffith and colleagues have published an article that describes the changing patterns of fractures in the dorsal and lumbar spines. They noted that there were two peaks of incidence of injury. The highest was in the midportion of the thoracic spine, the second being at the thoracolumbar junction.[102] The authors can offer only a theoretical explanation for this area of high incidence. Since it is a relatively stiff area, it is less able to absorb energy before failure. Moreover, because of its location, large bending moments are readily exerted.

Effects of Arthrodesis on Threshold for Injury of Adjacent Segments

This is an important question for which there is no definitive answer. The issue of hypermobility, disc degeneration, and spinal stenosis adjacent to iatrogenic fusion has been studied. There have been occasional case reports but no definitive documentation of a causal relationship.

A closely related issue is that of stress concentration and a lowered threshold of injury due to the large difference in stiffness between the fused and the mobile segment of the spine. Drennan reported a C6–C7 dislocation after an automobile accident in a 16-year-old boy who had a spine fusion at T1–L1 for scoliosis.[65] With an automobile accident as the cause of the injury, it is impossible to know whether or not the threshold for injury was lowered.

There is also the report of a patient with Klippel-Feil syndrome who sustained a fracture dislocation adjacent to the congenital fusion as a result of a simple fall. This is, of course, suggestive but not definitive.[76,231]

UPDATE OF RESEARCH STUDIES OF SPINE TRAUMA

There have been a number of experimental studies designed to evaluate the mechanics of spinal injuries. Studies on the application of forces to the spine have been reviewed in Chapter 1. Here, some of the salient experiments and findings regarding spine trauma are discussed.

The goal of this section is to review the cogent basic science and clinical studies that improve the knowledge and understanding of mechanisms of injury (MOI). Research on MOI can be grouped into *clinical observations*, *laboratory experimental studies*, and *biomechanical* and *mathematical models*. The material is therefore organized accordingly.

Clinical Observations

Generally, these presentations are based on interpretations of clinical histories and radiographs. From that data, plausible conceptualizations regarding magnitude of forces and displacements are made, and various hypotheses are derived.

Recently, studies of fracture of the occipital condyles have been reported.[8a] Compression and shear due to a vertical load associated with axial rotation and/or horizontal translation is the author's and our hypothesized mechanism for this injury. Fractures of the ring of C1 have generally been considered vertical (y-axis) compression MOI. Fractures of the dens, when anteriorly displaced, are thought of as flexion injuries with an anterior shear component. When posteriorly displaced, they are considered extension injuries with a posterior shear component. Injury to the axis is most commonly a traumatic spondylolisthesis with the characteristic failure through the region of the pedicles (isthmus). This is generally viewed as a hyperextension MOI.

Extreme cervical compression may occur during a wide variety of activities. Similar external loading patterns occur in diving injuries, American football, trampoline injuries, auto crashes, and emergency aircraft egress (ejection seat injuries). Compression forces are responsible for a number of nonfatal fractures in egress injury. These fractures have been classified by Kazarian[133] as (1) fractures of the vertebral body margins, (2) anterior wedge fractures, (3) lateral wedge fractures, and (4) cleavage fractures of the centrum. The anterior and lateral wedge fractures occur when flexion forces are combined with compression forces. The anterior wedge fracture is the most common nonfatal egress injury. This injury is characterized by collapse of the anterior part of the vertebral body. It is typically found in the region of C5–T1. This injury is clinically benign, and complete recovery is possible. Pain and discomfort, however, may be significant and may cause two or more months of disability.

The tear drop fracture is readily identified on lateral radiographs as the apparent anterior fracture of the vertebral body with an anteroinferior fragment adjacent to the end-plate resembling "a drop of water dripping from the vertebral body."[223] The term "tear drop" reflects the sadness often associated with the neurologic sequelae. The mechanism of this injury has been described as hyperflexion[223] and as combined flexion and compression loading.[92] It is important also to recognize another more important fracture plane oriented in the sagittal plane that often splits the vertebral body into left and right halves (Fig. 4-2). These halves may rotate with backward displacement and neural canal encroachment. This often unappreciated component of vertebral injury, which frequently accompanies the tear drop fracture, has been described by Torg[261] and Fuentes and colleagues[92]. Also, Fuentes described an associated lamina fracture in 48% of these "tear drop" fractures.

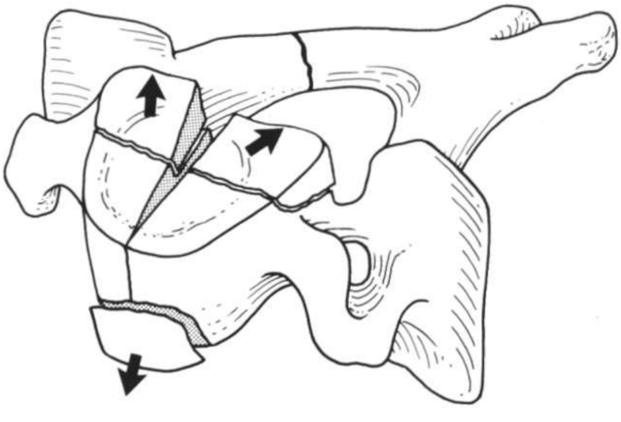

FIGURE 4-2 Orientation of common fracture planes and directions of fragment displacements commonly occurring in the tear drop fracture. Theoretically, this injury would be in the C–D–E range of vertical load and bending moment of relative magnitude that is shown in Figure 4-1. When this injury is in the D–E range, the 48% incidence of a lamina fracture is more likely to be a part of the injury. In both situations, the tear drop component is likely due to shear loading secondary to the vertical component of the load (see Fig. 4-8).

Flexion injuries are presumed to result when the head and neck are forced forward on the trunk beyond the normal limits. Braakman and Penning[28] have classified flexion injuries into compressive hyperflexion and distractive hyperflexion. Compressive hyperflexion mechanisms are those caused by forces that cause flexion of the neck combined with axial compression forces. Flexion of the head may accompany the flexion of regions of vertebral segments, but this head rotation is not necessary.[117] During this combination of forces and moments, various types of fractures may occur. Wedge compression fracture, as described above, is assumed to occur under relatively mild compressive forces. It can occur with disruption of the posterior elements if the flexion moment is large enough.

Distractive hyperflexion may occur during deceleration, where the flexion moments tend to separate the vertebral segments. Three types of injury result—hyperflexion sprain and unilateral and bilateral facet dislocations. Hyperflexion sprain is defined as a temporary or partial luxation of the intervertebral joints following traumatic hyperflexion under moderate forces, with rupture of the posterior ligaments and joint capsules but without dislocation. This type of injury is assumed to be relatively rare, because major forces are required to rupture the posterior ligaments. These major forces would then likely result in interlocking of the facets and facet dislocation. Only rarely would the hyperflexion movement stop prior to interlocking.[28]

Unilateral facet dislocation is assumed to occur as a result of flexion (x-axis) and axial (y-axis) rotation displacements with the associated moments. These injuries and unilateral facet fracture dislocations are presented in more detail on page 220.

The definitive etiology of bilateral facet dislocation has not been well established. This injury involves the caudal facets of the superior vertebra being displaced anterior to the cephalic articular facets of the inferior vertebra. Extensive ligamentous disruption is produced, including tearing of the interspinous, intertransverse, and capsular ligaments, the ligamenta flava, and a variable portion of the annulus (Fig. 4-3).[17] The flexion injury mechanism has been assumed,[17,177] although other authors have advocated extension,[255] vertical compression,[16] and axial rotation[209] mechanisms. The experimental production of these injuries will be discussed in the following section.

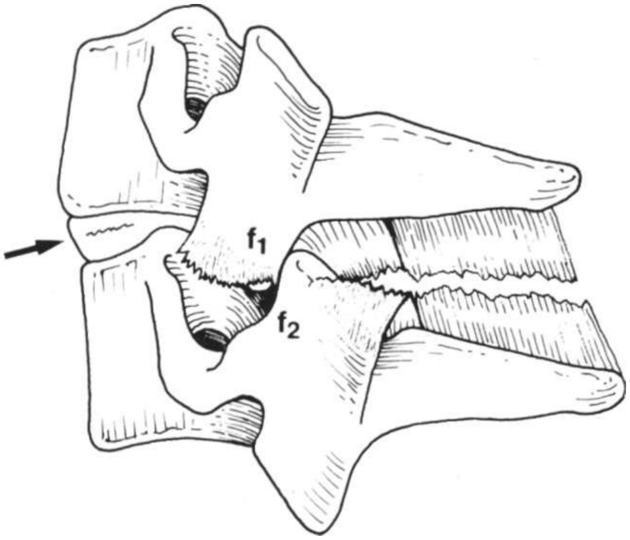

FIGURE 4-3 Unilateral facet dislocation. The inferior facet (f_1) of the superior vertebra is displaced and "locked" anterior to the superior facet (f_2) of the vertebra below. The arrow points to a tear in the annulus. There is also rupture of the interspinous ligament, the yellow ligament, and the facet joint capsule. As described by Beatson,[17] these are the structures typically damaged in unilateral facet dislocation.

The key information from these studies is as follows: many fracture patterns demonstrate axial compression as a component of their mechanism of injury, including the more severe cleavage fractures; clearly, wedge fractures also have a significant flexion component; flexion mechanisms may include anterior vertebral compression with posterior ligamentous structures loaded in tension; the mechanisms of dislocations remain controversial, and nonmidsagittal motions, such as axial rotation, may play an important role.

Hypothesis and Conceptual Frameworks

Based upon the principles of biomechanics, we propose the following hypothesis. An injured functional spinal unit (FSU) received its injury as a result of being subjected to a set of defined axial forces, shear forces, bending moments, and axial torques. There is a direct and reproducible relationship between the injuries sustained and the loads applied. It is further hypothesized that the loads causing the injuries to an FSU are determined by the external loads acting on the body and the specific head–neck–thorax position at the time of trauma. To investigate and interpret this hypothesis, quantitative descriptions of vertebral positions, loads, displacements, and point and rate of load application are required. Six components of displacement (translations and rotations) and the corresponding six components of load (forces and moments) at the level(s) of injury are needed to describe the biomechanical environment (see Fig. 1-32). The major injuring vector(s) causing a specific cervical spine injury may be regarded as the net force vector and/or moment vector resulting from these components.

Laboratory Experimental Studies

Much of the experimental work reporting specific cervical spine injuries with specific loading patterns exists within the automotive engineering literature. A number of studies of whole-body cadaveric testing and isolated spine segment testing have been reported in the Proceedings of the STADD Car Crash Conferences.[3,117,161,179] Generally, compressive loadings of the cervical spine in varying degrees of forward and lateral flexion and extension have been performed. Marked variability in the load sustained to failure has been noted within and among the studies. Although a one-to-one correspondence between major injuring vectors at cervical segments and external patterns of loading has not been established, important observations can be made. The load magnitude, the rate and duration of load application (the time history), the point of load application, the direction of loading, and the initial cervical spine position and relative positions of the head and thorax all determine the individual cervical spine injury.

During extreme compression, the entire cervical spine may bend or buckle in a manner not consistent with the patterns and range of normal coupled physiologic motion. The initial positions of the head, neck, and thorax are perhaps most important in determining the mechanical response to a given load combination.

Alem and co-workers,[3] in cadaver drop and pendulum impact studies (Fig. 4-4), noted that "buck-

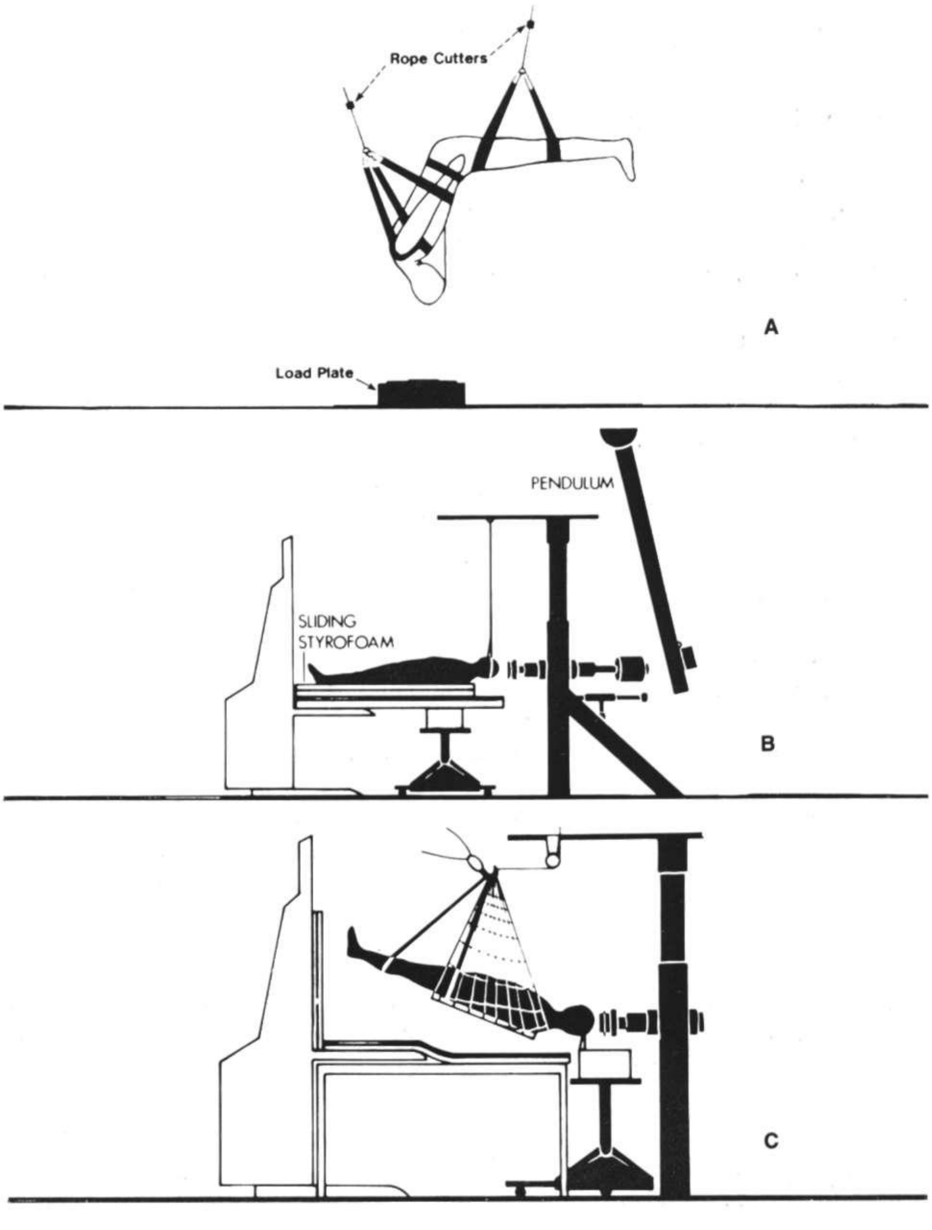

FIGURE 4-4 Various techniques of cadaver impact testing. Although all three methods of loading the cervical spine are dynamic and seem to load the spine in axial compression, there are significant differences. In this realistic trauma situation, (**A**) the inertia forces of the body apply compression force and a flexion/extension bending moment, which is dependent upon the body configuration and cervical spine posture at the time of impact. (**B**) By resting the body on a table and suspending the head, the cervical spine forces are better controlled. (**C**) A compromise between techniques **A** and **B**. *(Huelke, D. F., and Nusholtz, G. S.: Cervical spine biomechanics: a review of the literature. J. Orthop. Res., 4:232.)*

ling" of the cervical spine occurred with axial loading when the neck was positioned to simulate the natural attitude of normal sitting or standing persons. In virtually all tests, cervical spine damage was noted when the neck "buckled" under the impact load. In contrast, with similar impact loading, no cervical spine damage was apparent when the spine was aligned along the line of action of the impact force. However, autopsies were not performed, and time histories were not reported for this group. It should be noted that these investigators used the term "buckling" to describe the displacements of the spine from an initial curved position, rather than the buckling of a straight segmented column, as used by Torg.[261]

Maiman and colleagues[150] studied cervical flexion/extension and its relation to failure load magnitude and cervical damage. Cadaver head-to-torso segments and cervical spine segments were subjected to axial loading in either a neutral upright position or prepositioned in 25° of flexion or extension. The point of load application for the neutral position was at the vertex, except for two specimens with loads applied 2 cm posterior or 1 cm anterior to the vertex. All specimens prepositioned in flexion or extension failed in an apparent flexion or extension mode, respectively. The cadavers with posteriorly applied load failed in extension, and those with anteriorly applied load failed in flexion. Of the neutral specimens loaded directly at the vertex, only two failed in apparent direct axial compression, one failed in extension, and two failed in flexion. Those cadavers axially loaded sustained much greater loads to failure (mean, 3567 N) than those prepositioned in flexion or extension (means, 1823 N and 1089 N). In flexion and extension cases, the failure was most probably due to the stresses produced by the bending moments during trauma. However, the bending moments were not recorded.

The significance of the initial cervical spine position at the time of axial impact loadings was also studied by Hodgson and Thomas[117] in cadaver impact tests. Cervical spine strains measured at the C3, C5, and C7 vertebral bodies were evaluated for various neck and thorax positions. The highest vertebral body strains were seen with axial loading at the crown of the head with relative flexion of the neck and chest, which will straighten the lower cervical lordosis.[261] Of the varying positions of load application, axial loading at the vertex with the cervical spine in neutral posture demonstrated the highest vertebral body strains, in comparison with positions of more anterior force application and more cervical extension.

Results from the above three studies suggest that the straight (as with mild forward flexion) cervical spine can *withstand* the highest axial compressive loads. But, perhaps even more important, the straightened cervical spine will *sustain* the highest load (and strain) magnitudes on external axial impacts of all positions studied.

Yoganandan and co-workers[298] measured peak forces produced by vertical impact of restrained and unrestrained cadavers. Eight of fifteen heads were restrained to simulate muscle forces supporting the head in a neutral position. The unrestrained cadavers demonstrated more rapid cervical flexion upon impact than did the restrained cadavers. Peak forces measured at the head were higher in the restrained (9800–14,700 N) than in the unrestrained (3000–7100 N) cadavers. The forces at the site of injury were considerably less (1880N and 1570N respectively). A predominance of anterior ligamentous and posterior bony injuries occurred in the restrained group, and posterior ligamentous and anterior bony injuries occurred in the unrestrained group. This study may also have theoretical implications in regard to the question of the relative threshold of injury of the muscularly relaxed versus the muscularly splinted spine.

Nusholtz and colleagues[179] suggest that cervical spine "flexion-type" damage from vertical impact of the mildly flexed spine positioned in the midsagittal plane is unlikely. However, they found that vertebral body compression and posterior ligamentous disruption were readily produced in axially rotated or laterally flexed spines. Thus, this study emphasizes the importance of the non-midsagittal plane loads. The importance of non-midsagittal motions was also emphasized by Roaf.[209] He was able to produce purely ligamentous injury in cadaver spine segments by flexion or extension alone.

Panjabi and co-workers took a different, more comprehensive approach toward understanding the mechanics of trauma.[185a] Using fresh porcine cadaveric three-vertebrae cervical spine specimens, they produced high-speed trauma (4.6 m/s). Flexion, extension, and compression traumas were produced by varying appropriately the eccentricity of the point of compressive load application with respect to the geometric center of the middle vertebra. The event, which took less than 20 ms, was monitored by a three-component load cell (capable of measuring compression force, shear force, and flexion/extension moment) placed under the specimen and a high-speed 16-mm camera. The load readings were recorded by a computer at the rate of 3000 per second, while the camera recorded the trauma at 1000 frames per second. After the trauma, the specimen was studied radiographically, biomechanically, and morphologically. All three load components (*i.e.*, flexion/extension, shear force, and compression force) were present in each of the trauma modes, although proportions were different. The flexion moment was highest in the flexion trauma (154 Nm), and the extension moment was highest among the specimens subjected to extension trauma (105 Nm). Compression force was highest among the compression trauma specimens (10,140 N). It was also present in the extension trauma (8920 N) and flexion trauma (6360 N). These results support the concept that it takes less compressive force to produce spinal fractures in the flexion or extension trauma mode than it does in the compression trauma mode. But it must be emphasized that the injury results because of the combined effect of all loads: compression force, shear force, and bending moment. For example, in flexion trauma, although the compression force was lower than that in the compression trauma (6400 N vs. 10,100 N), it was accompanied by a larger flexion bending moment (154 Nm vs. 130 Nm). A similar argument holds for the extension trauma versus compression trauma. In other words, while studying the mechanisms of trauma, it is necessary to measure all the load components present. Thus, for the sagittal plane trauma, one should measure anterior/posterior and superior/inferior forces and flexion/extension bending moment (±FY, ±FZ, and ±MX; see Fig. 1-32). For a completely three-dimensional trauma, it is necessary to measure all six load components.

Roaf applied static loads to fresh frozen autopsy functional spinal units using an experimental apparatus that applied compressive loads.[209] He recognized that this was not the most characteristic rate of application of loading found in injuries of the spine. The method was thought to be justified by the fact that on a few occasions a more rapidly acting force was applied, and the same pattern of failure in the functional spinal unit was observed. There are no values given for either of the rates of application. This basic work provides some reasonable guidelines to the sequence of events that may be involved in the compressive loading of vertebrae in trauma. There is the initial deformation. The disc is relatively less compressible than the vertebra when tested in a normal functional spinal unit. Therefore, there is bulging of the vertebral end-plate. The process continues to subsequent failure of the end-plate, which is probably the first structure to fracture. As deformation proceeds, there is a fracture of the cortical shell and compression of the cancellous bone. This study did not demonstrate the phenomenon suggested by Armstrong, which involves pressing of the nucleus material on the annulus, causing a localized protrusion.[11] However, in the older specimens where the nucleus was no longer fluid, Roaf recognized that there was asymmetrical compression and sometimes pressure transmitted to the annulus, resulting in either tearing of the annulus or general collapse of the vertebra due to the buckling at the sides. This study also included experiments in which the resistance of the annulus *with* the presence of a fluid nucleus pulposus was compared with the resistance of the annulus *without* the presence of a fluid nucleus pulposus. After denucleation (the removal of the nucleus pulposus), compressive loading produced typical annulus prolapses of the type seen in operations involving disc protrusions. Roaf suggested that this may have considerable medicolegal significance, because a vertebral end-plate fracture with extrusion of the nucleus into the vertebral body and loss of disc turgor may result in the subsequent development of typical annulus herniation.

There were other fundamental facts related to spine trauma that came out of this work. Roaf indicated that he was not able to produce a "hyperflexion injury" of a normal intact spinal unit because vertebral body crush always occurred prior to the rupture of the posterior ligaments. Similarly, he recognized that it was not possible to rupture the anterior longitudinal ligament in the cervical spine by pure hyperextension force prior to producing crushed fractures of the neural arch. He noted that the ligamentous structures, while significantly resistant to compression and tensile loading of the functional spinal unit, were quite vulnerable to rotation. He also noted that the disc was subject to injury because of the horizontal shear forces produced by the rotation. Thus, when there are extensive ligamentous ruptures clinically, the possibility of rotation (moments about the y-axis) should be strongly considered in an analysis of the mechanism of injury.

Crowell and co-workers[49] noted both posterior ligament injuries and compression fractures in cadaveric spine segments from the middle and lower regions that were subjected to pure flexion loads. Such failures occurred in six of seven middle-segment specimens and in two of seven lower-segment specimens subjected to flexion of 21–33° over two spinal segments. Much more extensive ligament injuries, including posterior longitudinal ligament tears, occurred in two other specimens that were axially rotated 16° prior to flexion loading. This further emphasizes the importance of non-midsagittal positions, such as axial rotation, in lowering the threshold of loads causing injury in flexion loading.

Farfan and colleagues carried out experiments that led them to the hypothesis that the annulus does not tend to fail with compressive loading, but rather with shear loading.[81] This is no reason, however, to assume that compression and shear, as well as bending, may not be contributing factors. In this same study it was observed that, with compressive loading of the functional spinal units, blood was squeezed out of the cancellous bone of the vertebral body. This phenomenon led to the rather appealing hypothesis that the fluid component of the blood in the spongy elastic vertebral body serves as a shock-absorbing mechanism. Such a mechanism is assumed to work only at high rates of deformations, which is generally the situation in clinical spine trauma.

The study of Gosch and colleagues provides some interesting information about spine trauma. This investigation examined the effects of impact loading applied to the heads of monkeys. The animals were anesthetized, placed on a track, and accelerated into a fixed metal barrier. A transducer load cell mounted on the impact block measured the energy delivered to the head at the time of impact. High-speed cinematography was used to analyze the movement following the impact. Lateral radiographs of the spine were taken after the injury, and specially prepared midsagittal cuts of the frozen vertebral column were made in order to study spinal cord compression by displaced bone fragments. The experimental design included three groups of animals. The magnitude of the force was varied in each of the three groups. In addition, the three groups were distinguished by loading the necks at the time of trauma in flexion, in extension, and under vertical compression ($-y$-axis). The investigators found that in addition to flexion or hyperextension, it was necessary to induce an axial rotation to the spine (y-axis rotation) in order to produce dislocation.[98] This, of course, corroborates the observations by Roaf and is especially significant because it employs living animals and more clinically realistic impact loading. Gosch and colleagues emphasized that the anterior and posterior compressive forces, especially those occurring in hyperextension, produced the greatest damage to the central portion of the spinal cord.

They also indicated that the muscle tone at the time of impact had a profound effect on the threshold of forces required to produce cord damage. Muscle tone was necessary to produce the cord injury. This is compatible with and somewhat supportive of the oft-stated assumption that, all other things being equal, the more relaxed individual (an infant or a semiconscious adult under strong pharmacologic influence) is less likely to be injured in a situation involving physical trauma.[A] A qualifying point should be added here. The assumption holds unless the magnitude of the traumatizing forces is low enough to be checked by the intrinsic splinting power of the subject's muscles.

Mechanisms for Some Specific Injuries

One purely ligamentous injury, the bilateral facet dislocation, was consistently produced by Bauze and Ardran[16] at relatively low loads. Axial load was applied to a flexed cervical spine specimen with the end below the dislocation stiffened by insertion of a metal rod into the neural canal. The combination of axial loading of the flexed cervical spine and stiffening of the segment below the level of injury created a sufficient local forward flexion moment and anterior shear force to consistently produce this injury. The clinical significance of these specialized loading conditions remains unknown. Bilateral facet dislocation was also produced in vertical unrestrained cadaver drop by Yoganandan and colleagues,[298] although non-midsagittal position data are not reported.

Mechanisms creating other specific cervical spine injuries have been demonstrated by McElhaney and co-workers[161] in cadaver spine specimens. In these tests, a very small anterior to posterior distance variation in the point of application of axial cervical spine loading produced great variation in the resulting type of fracture, as shown below. Several fracture patterns were associated with specific loading conditions.

Extension fractures: Buckling of the straight cervical spine in extension was noted with only 1 cm of posterior eccentricity of axial loading.

Jefferson fractures: These were produced with direct axial loading of the generally straight cervical spine, with slight extension (magnitude not reported) of the specimens noted on review of pre- and post-test x-rays.

Burst fractures: These fractures were produced by direct axial load of the straight cervical spine segments, which were found to be slightly flexed on x-ray review. These specimens required larger loads and strain energies prior to fracture than did the Jefferson fractures.

Anterior wedge fractures: Application of the axial load less than 1 cm anterior to neutral position produced compression of the anterior vertebral body. Axial loads applied at a distance greater than 1 cm anteriorly produced buckling rearward and subsequent disc and end-plate failure.

The "tear drop" fracture: This fracture has been experimentally produced by axial compression of the neutral and minimally flexed cervical spine.[3, 150] The presence of a sagittal fracture plane in addition to the more horizontal plane is demonstrated in careful analyses of clinical injuries of this type.[92, 261]

Disc ruptures: These injuries have been produced in many specimens subjected to axial impacts in various degrees of cervical flexion/extension.[3, 298] Disc ruptures were most common (six out of a total of eight) in tests that also subjected the specimens to axial rotation and lateral flexion at the time of impact.[179]

Experimental Overview

These studies demonstrate that the initial head–neck–thorax position and loading conditions dictate the cervical spine response to impact. There appear to be three patterns of response: bending of the curved spine under axial load, buckling of the straightened spine under axial load, and deflection of the spine out of the load path. Varying the initial position altered the type of response, maximum load at failure, and extent of injury. Thus, in potentially subinjurious impacts, the initial position may determine the presence or absence of cervical spine damage. In addition, non-midsagittal loads will lower the injury threshold.

Physical and Mathematical Models

Parameters for developing models of the spine that define the mechanical response of the spine include several key factors:

- Static and dynamic mechanical properties of bone and the interconnecting ligamentous structures
- Vertebral, disc, and ligament geometry
- Load–displacement history
- Boundary conditions for the region of interest
- Mass and mass moment of inertia of the vertebral components and associated body parts

These inputs define the core of the models and to a large extent define limitations that can be placed on the results. A brief review of the status of modeling follows. The purpose is to provide the reader with a sense of the state of development of the science as well as its level of reliable applicability to clinical and medicolegal issues.

Anthropometric Data

The most fundamental aspect of a biomechanical model of the spine is the representation of the geometric relationships between the interconnecting components. Anthropometric data have been collected to describe the cervical spine. Virtually all biomechanical analyses of the spine are begun with an appropriate geometric description of the region to be studied. Panjabi,[184] Belytschko and co-workers,[18, 19] and others have advocated the refinement of geometric data.

Anthropometric data have been published by several authors and are provided in Chapter 1.

Physical Models

A number of anthropomorphic physical models have been developed that have progressively increased in complexity and accuracy of representation of the human neck in loading.[55, 88] These test devices ("crash dummies") are generally used to predict the dynamic human response to impact testing of magnitudes unacceptable for human volunteers. Such data obtained may be validated by cadaveric testing and dynamic human testing under lower impact conditions. Newer physical models[55] estimate head accelerations, cervical disc pressures, and muscle strains. The accurate simulation of such parameters through these models for a wide range of injurious and subinjurious conditions is appealing, although further validation of the current models will be required before one can be confident in clinical conclusions drawn from these simulated structures.

Mathematical Models

Over the last 30 years, mathematical models have been used to investigate head–neck kinematics and the biomechanics of injury. The complexity and degree of sophistication of the models have increased steadily. Summaries of the investigations of the load–displacement response of the cervical spine have been presented by King and Chou,[139] McElhaney and co-workers,[162] and Goldsmith.[97] Comparisons of analytical results with experimental data have been presented by Prasad and King[201] and Landkof and colleagues.[140] Such comparisons are important as a means of establishing the validity of analytic models.

Early continuum models were obtained by Liu and Murray[148a] and Shirazi,[234] which provided closed-form solutions. These models represent the cervical spine as a continuous beam structure, with solutions satisfying equations of equilibrium.

Finite difference solutions of the load–displacement response of the cervical spine were obtained by Terry and Roberts,[259] Rybicki and Hopper,[218] Li and co-workers,[146] and Landkof and colleagues.[140] Such solutions are numerical procedures that fit experimental gross motion data with modeling constants describing the cervical response characteristics. These early models are limited by their depiction of the cervical spine as a simplified structure, such as a viscoelastic beam, that does not consider individual segmental motion or loads.

"Lumped parameter" and "discrete parameter" model systems are capable of representing the spinal geometry and physical properties of individual anatomic structures much more realistically. Such models subjected to forward accelerations have been developed by Toth,[267] Hopkins,[121] Kaleps and co-workers,[130] Orne and Liu,[180a] Panjabi,[184] Prasad and King,[201] and Huston and co-workers.[123a] These models have been extended to three dimensions by Merrill,[164a] Williams and Belytschko,[290] and Deng and Goldsmith.[55] Parameters such as masses, springs, and damping elements are estimated for anatomic components of models of varying complexity and anatomic accuracy. The prediction accuracy of the models may be expected to depend largely on the accuracy of the modeling parameters. The advancement of model complexity with continued refinement of "lumped parameters" into increasingly anatomically accurate and increasingly mechanically discrete elements continues (Fig. 4-5). The potential uses of accurate mathematical models are many. Applications may include continued improvement in the design of protective devices for injurious conditions, modification of activity techniques, prediction of injurious activities for the normal and pathological spine, and prediction of dynamic responses to therapeutic interventions.

Biomechanical Modeling Parameters

The development of mathematical models that describe the influences and responses of individual anatomic components of the neck requires a description of the mechanical properties of such components (the modeling parameters). A limited amount of strength data for individual components of the neck can be found in the literature. Messerer[166] reported a series of compression tests on human vertebral bodies from all levels of spines of cadavers ranging in age from 25 to 80 years. For the cervical region, the maximum compressive breaking load of the vertebral bodies is 2800–4200 N for the 20- to 40-year-old.[166, 238, 297] The tensile failure stresses of the spinal ligaments are about 10–20 MPa, based on studies of the lumbar ligamentum flavum[171a] and the anterior and posterior longitudinal ligaments.[260] In terms of breaking load, the transverse ligament of the atlas has a breaking load of about 1000 N *in situ*, when C1 is rapidly translated anteriorly (+z-axis) with respect to C2.[84] The tensile breaking mass of other lumbar spine ligaments are 3400 N for the anterior longitudinal ligament and 1800 N for the posterior longitudinal ligament.[260] Studies by Dvorak and colleagues[69] showed that the *in vitro* strength of the alar ligament was 200 N and that of the transverse was 350 N. (For more detailed data on spinal ligament strength, see Chapter 1.) The ultimate tensile stress of human muscle has been estimated to lie between 0.2 and 0.6 MPa for stretching of the passive muscle.[132] The estimated stress of active contracting muscle ranges from 0.4 to 1.0 MPa.[83, 111]

The intervertebral disc and its interactions with the vertebral bodies play an important role in determining the strength of the functional spinal unit. Sonoda[238] reported strength tests on human vertebral columns from all major regions of the spine in compression, tension, and torsion. Here we provide data for the cervical spine. The compressive breaking load is about 3200 N for the 40- to 59-year-old.[238, 297] The tensile breaking load is about 1000 N and the torsional breaking moment is about 6 Nm.

Biomechanical properties of the intervertebral

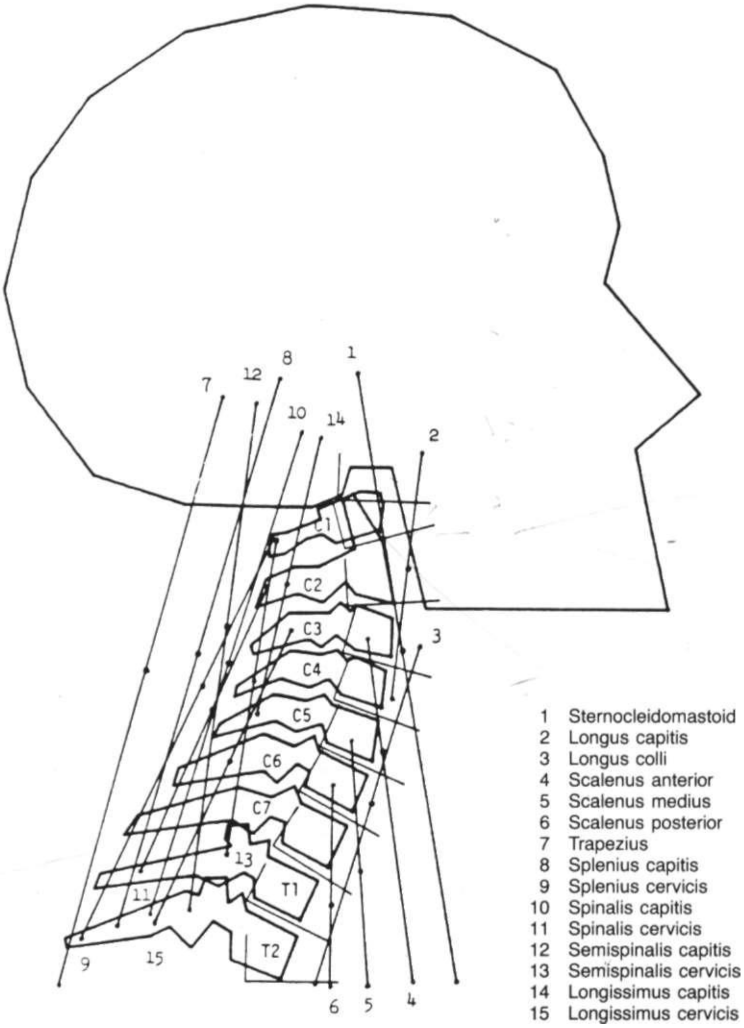

FIGURE 4-5 Diagram of the parametric model of Deng and Goldsmith. A realistic model of the neck should include the physical and geometric properties of the spinal column, muscles represented by their passive and active physical characteristics and their attachment point coordinates, and inertia loads of vertebrae and skull, in the case of dynamic simulations. *(Deng, Y. C., and Goldsmith, W.: Response of a human head/neck/ upper torso replica to dynamic loading-II. Analytic/numerical model. J. Biomech., 20(5):487, 1987.)*

joint have been measured by groups of researchers for each of the three mobile areas of the spine. All of these studies have been conducted at low levels of load, far below conditions associated with vertebral fracture and ligamentous injury. Measurements of the load–displacement response of spine segments at larger load magnitudes by Panjabi and co-workers,[190] Miller and colleagues,[167] and Edwards and co-workers[70] have shown that the properties of the spine vary significantly as load magnitudes increase.

Experimental stiffness data for the cervical spine have been reported by Liu and co-workers,[148] Moroney,[169] Panjabi and colleagues,[187] and Coffee and co-workers.[45] The stiffness values reported by Panjabi and co-workers were measured at a relatively low load magnitude (25 N), below the higher stiffnesses encountered prior to failure. The Moroney stiffness data, used in the Deng and Goldsmith[55] mathematical model, do not represent variation in segment stiffnesses at different cervical levels. In this model, individual segmental stiffnesses were assumed to be proportional to the disc cross-sectional area for the cervical spine.

The stiffness data of Coffee and co-workers[45] show a distinct difference in cervical spine stiffnesses between the middle (C2–C5) and lower (C5–T1) cervical spine. Similar differences between the middle and lower segment *rotational* stiffnesses were reported by Liu and co-workers,[148] although translational stiffnesses in their two specimens were similar in the two segments. Consideration of such individual segmental variations and recognition of the three mechanically distinct regions of the cervical spine (the upper C0–C1–C2 region plus these two) may be important in accurate cervical spine modeling and the understanding of injuries (see Chap. 1).

Coupling of cervical spine motions and coupled force–displacement data have been reported by Panjabi and colleagues.[187] These force–displacement data, which take three-dimensional coupled motion into consideration, enable the formulation of stiffness matrices for each cervical level, which predict three-dimensional responses to forces and moments of various directions and magnitudes. Recent sophisticated mathematical models[55,56] attempt to predict such coupled responses. Incorporation of accurately coupled response estimates into models that predict complex three-dimensional load–displacement relations may significantly improve the accuracy of the models.

Validation of Mathematical Models

Validation of these models has been performed by comparison of load responses predicted by using experimentally determined load responses of instrumented humans, other primates, and cadavers. Studies performed by Ewing and co-workers[79,80] have quantified the human head and neck response to forward and lateral impact accelerations employing accelerometers and high-speed cameras. These results have been compared with recent modeling predictions of the responses to such impacts,[55,56,290] with improving agreement in these later studies.

Cervical spine responses and injuries predicted by modeling techniques may also be validated by comparison with clinical observations. Modeling estimates of the response to impact may be compared with careful analysis of automobile accidents, including comparison of the vehicle paths and internal and external automobile damage with soft-tissue and bony injuries. Similar comparisons may be possible for sports injury mechanisms, for which preimpact velocities and positional data may be obtained from American football game films, such as in the work of Torg.[261]

Proposed Injury Criteria for the Neck

A reliable quantitative load-to-failure criterion has not been established in these studies. A wide range of failure loads were observed within tests using a consistent methodology, and an even wider range was observed among methodologies.

A proposed flexion/extension injury criterion of the neck involves the equivalent moment at the occipital condyles. This equivalent moment consists of the moments produced by neck forces and chin–chest contact forces (if chin–chest contact occurs), taken with respect to the occipital condyles.[165] The forces acting on the head are: the force of gravity, an axial force produced by the neck structure directed along the axis of the vertebral column parallel to the dens, a shear force that produces a distributed bending moment along the cervical spine, and a chin–chest contact force. Mertz and Patrick[165] conducted volunteer tests on an average male for horizontal deceleration at various gravitational force (g) levels, including a maximum of 9.6 g. These tests were used to determine the effects produced by varying the mass, center of gravity, and mass moment of inertia of the head by the addition of a helmet. The center of mass was varied from above the head's center of mass to below.

In static tests on volunteers, maximum bending moments of 35.4 Nm in flexion and 23.8 Nm in extension were obtained.[165] There is no contribution from the chin–chest contact in the static tests. In dynamic voluntary tests, a maximum value of 47.6 Nm was reached in extension. In hyperflexion, the chin–chest reaction resulted in a maximum equivalent moment at the condyles of 88.4 Nm sustained without injury. This exposure produced pain extending from the posterior neck to the posterior midthorax and resulted in a stiff neck that lasted for several days.[165]

According to Mertz and Patrick, the maximum voluntary static neck reaction is about 1130 N in tension and 1110 N in compression. For shear, the neck can withstand about 845 N. These static strength values apply to low g conditions where the viscous resisting forces produced by the muscles are not a large part of the resisting load. In a high g situation, the viscous contribution of the muscle reaction is comparable to the static strength, resulting in a higher resisting load than that predicted by a static strength analysis. In other words, the static strength values quoted are a low estimate of the neck strength in the high-acceleration environment.

In cadavers tested in the horizontal direction, maximum bending moment levels of 177 and 190 Nm were achieved without apparent damage, as determined by x-ray analysis. A maximum anteroposterior force of 2100 N was developed at the occipital condyles of one cadaver without producing any apparent neck trauma. However, these values for the Mertz and Patrick cadaver studies should be viewed with caution. Not all fractures are visible on x-rays, and ligamentous damage is impossible to identify on radiographic examinations alone.[49]

Alem and co-workers,[2] like other investigators, noted a wide range of forces producing cervical spine damage and found *force* to be a *poor* predictor of injury. However, they found the *impulse*, the integral of force over time, to be a *useful* predictor of cervical spine injury in their cadaver study. In their series of 14 cadaver axial impacts for which impulse values are given, no cervical spine injuries were noted in 5 cadavers with applied impulses of 24–35 N-seconds, and cervical spine injuries were noted in 9 cadavers with applied impulses of 35–49 N-seconds.

In addition to structural bony or ligamentous force-to-injury criteria, forces producing neurologic sequelae have been studied. Just as for the spinal column, the force and impulse have been found to be good indicators of cord injury (seen histologically)[60] and functional loss in animal models.[190a] The association of increased risk of neurologic injury with greater vertebral injury has been recognized in clinical evaluations.[208] The possibility of neurologic deficit resulting from cervical spine injury is an area of research that is difficult to investigate using models, but it can be studied experimentally. Studies of nonhuman primates have identified the neural and physiologic effects of axially applied tension loads. Studies conducted by Cusick and co-workers[49a] in 1982 using male *Macaca mulatta* monkeys have shown that axial loads of 556–1555 N produced immediate reduction of the somatosensory evoked potential amplitude. When these loads were maintained, these changes were followed by changes in heart rate, blood pressure, and perfusion. Spinal cord trauma is discussed in one of the following sections. It is clear that the concept of force magnitude and time dependency of injury is valid for quantifiable neurologic changes as well as for bony and ligamentous injuries. Such improved quantitation of time, cervical spine position, force and moment direction, and biologic variables will improve the accuracy of injury prediction.

SUMMARY AND CONCLUSIONS

What do these studies mean? Obviously, the answer to the question "What does it take to break one's neck?" is not simple. The preceding studies, in our view, justify the following general comments. The conditions required to injure the neck include several key variables. They are: impact magnitude, direction, point of application, and rate of application. These will be different for different relative positions of the head, neck, and thorax at the time of impact. The *rate of application* of the impact load is a critical variable. Other variables being equal, the magnitude of required load to failure tends to vary directly with the rate of application of the load. Impulse (the integral of force over time) will likely be shown by subsequent research to be the most useful predictor of injury.

The relative positions of the head, neck, and thorax are major factors in both the threshold of failure and the patterns of failure. Here, patterns of failure refer to which structural components of the spine are injured. In considering threshold and pattern of failure, there are two salient situations to observe. They are the presence or absence of preset axial rotation of the spine and the buckling potential. Position is important in regard to the likelihood of injury with true mechanical buckling (straight spine under axial load) or bending of a curved spine in or out of the sagittal plane.

Additional cadaveric testing will help define the critical initial conditions and critical loading associated with spine injury. However, the supply of cadavers is limited, and experimental cadaver testing is costly. Furthermore, from the experimental studies it is apparent that there are several independent variables that determine the load magnitude and type of kinematic and time response to impact loading. Accurate mathematical modeling analyses may play a significant role in defining such responses, with cadaver testing useful for validation of modeling results.

The continued improvement in three-dimensional descriptions of mechanical characteristics of cervical spine components obtained from human cadaveric testing, which identify level-specific differences and define the range of variation within the population, will refine the accuracy of models. Such models, both mathematical and physical, will then be able to improve response prediction for the wide range of physiologic and injury-producing loads encountered in life.

Clinical studies that clearly, and preferably quantitatively, identify and clarify injury patterns and the associated biologic responses are necessary for true validation of the models. The expansion of modeling and experimental mechanical studies to pathologic mechanical states and their operative and nonoperative treatments will enable more accurate assessments of the necessity for and efficacy of such treatments.

More communication and collaboration among investigators from varied disciplines and different academic and industrial backgrounds will accelerate progress in the understanding of the biomechanics of spine trauma. Such collaboration of workers attempting to solve similar problems for diverse purposes will avoid redundancy of efforts and enable the optimal interaction of various areas of expertise for the mutual benefit of all.

SOME COGENT STUDIES OF SPINAL CORD TRAUMA

This section presents some basic information about the pathoanatomic changes that occur in the spinal cord following trauma. Most of the points presented here have been selected from a comprehensive review by Dohrmann.[58] The studies are organized according to three basic questions: What happens to the spinal cord following mechanical impact? When does it happen? What, if anything, can be done to therapeutically alter the sequence of events?

Pathoanatomic Changes

A number of investigators have reported post-traumatic alterations in nerve cells and myelinated fibers and demonstrated hemorrhage within the gray matter, which is generally followed by edematous changes in the white matter.[271–275] The amount of hemorrhage was grossly related to the magnitude of the contusing forces. Dohrmann's electron microscopic studies revealed that the hemorrhages in the spinal cord contusion were probably related to tears in the walls of the muscular venules that are located in the gray matter.[61] Turnbull noted that the central vessels in the spinal cord are oriented so as to be stretched and compressed by a posteroanterior force.[268] This implies a greater vulnerability of the cord to forces in the sagittal plane of the z-axis. In other words, abnormal anterior or posterior translation of the vertebrae is more likely to cause damage to these vessels and therefore to the spinal cord.

Temporal Considerations

It has been shown that when the intrinsic vessels of the spinal cord of the cat are injured to the point of paraplegia, the perfusion of the gray matter is reduced as early as 15 minutes after experimental trauma, and after 1 hour the flow is virtually nonexistent.[59–63] However, it is injury to the major sensory and motor tracts, located in the white matter, that is responsible for the devastating functional loss following contusion of the spinal cord. After trauma, the perfusion of the white matter decreases and begins to stabilize by approximately 1 hour. Then it either returns within 24 hours after injury (transitory traumatic paraplegia) or continues to decrease (permanent traumatic paraplegia).[61] This decrease in blood flow patterns in the white matter is due to vasospasm and is probably secondary to the post-traumatic subarachnoid hemorrhage.[59]

Investigators have observed a centrifugal progression of edema in feline spinal cords during the first 8 hours following trauma.[100, 272] At 1 hour following contusion, edema primarily involved the gray matter. At 4 hours, it had progressed to the immediately adjacent white matter. By 8 hours, the entire contused spinal cord had become edematous. As expected, there is a correlation between the magnitude of the force applied to the spinal cord and the associated damage.

Magnitude of Trauma and Degree of Damage

Allen, in 1911,[6] first quantitated experimental trauma applied to the spinal cord. He found that by dropping a 30-g mass approximately 10 cm perpendicularly onto the thoracic spinal cord of dogs, temporary paraplegia was produced.[B] If the impact was increased by dropping the 30-g mass from a height of 14 cm, the dogs were rendered permanently paraplegic. Intramedullary hemorrhage and edema were the changes noted within 4 hours of impact. In a recent paper, correlation has been established, at least in cats, between the magnitude of trauma (as measured by impulse) and the resulting spinal cord lesion.[B,60] Another method of quantitating spinal cord trauma is to place a small hydraulic balloon in the epidural space and apply measured compressive forces. These studies were done by Tarlov and colleagues, who observed degeneration at the level of the balloon and also extension of pathology in both directions away from the site, commensurate with the amount of pressure applied.[249–252] This observation is significant because it probably explains the frequently observed clinical finding of spinal cord damage a variable distance from the location of the recognized trauma.

The observation of coalition or extension of damaged areas of the spinal cord has other clinical significance. Dohrmann and colleagues showed experimentally in monkeys that a "transitory traumatic paraplegia," which appeared to be complete, could resolve spontaneously in 1 to 2 weeks after injury.[59] This was the case provided that bleeding was local, central, and did not coalesce to form a large hemorrhage involving the entire gray matter. This implies that recovery is at least partially dependent on the size and extent of the damage caused by the initial trauma and also fits with the generally good prognosis in the clinical central spinal cord syndrome. In some instances, recovery may be related to collateral circulation. Kamiya carried out studies in which he experimentally compressed the anterior region of the cervical spinal cord in dogs. Through his studies he was able to recognize a fair amount of compensatory circulation between the anterior spinal artery, the central arteries, and the longitudinal anastomotic vessels near the central canal.[131]

Effects of Treatment

What, if anything, can be done to prevent the pathologic changes associated with spinal cord trauma? Richardson and Nakamura applied local compression that produced intramedullary edema but no hemorrhage.[207] Interestingly, these investigators were apparently able to reverse all of the electron microscopic alterations associated with the low-grade trauma with a combination of local hypothermia and steroids. White and colleagues reported that local hypothermia beginning 6 to 8 hours after the contusion did not significantly alter the pathologic changes in spinal cord morphology.[285] Albin and colleagues contused spinal cords and then applied local hypothermia. They observed that if the hypothermia was instituted within 4 hours of contusion, the animals regained most of their sensory and motor function. However, if this was not initiated until 8 hours following contusion, then there was a failure to demonstrate recovery.[1] Other studies demonstrating recovery included those by Ducker and Hamit, who were able to demonstrate recovery using either local hypothermia or steroids within the first 3 hours of injury.[66]

Hartzog and colleagues and Kelly and colleagues, following contusion on the spinal cords of experimental animals, were able to demonstrate functional recovery with the use of hyperbaric oxygen.[110,136]

Black and Markowitz observed that steroids enhanced functional recovery, while incision of the dura mater had deleterious effects in monkeys with contused spinal cords. They also noted that recovery rates were the same for monkeys decompressed with a one- or a three-level laminectomy.[21]

In general, most studies evaluating various treatments in experimental spinal cord contusion have used small control (untreated) groups. In a group of over 100 unrelated animals with spinal cord contusions, the lesions produced were found to be more clinically variable than initially believed. This suggests that some of the "improvements" observed with various treatments may not have been caused by the treatments. Therefore, to make definite conclusions apropos of treatment efficacy, large control groups are necessary.*

Osterholm and colleagues described an increase in intramedullary norepinephrine after trauma. Excessive accumulation of norepinephrine and hemorrhage into the gray matter of contused spinal cords was prevented for up to 6 hours by administering alpha-methyl tyrosine within 15 minutes of the experimental trauma.[182,183] However, other investigators have not been able to confirm this reported increase in norepinephrine.[53,66,113,172,204]

In order for these factors to demonstrate any effect in the experimental situation, they must be operative soon after the injury. It is rarely possible in the clinical situation to achieve this. Nevertheless, based on these and other experimental studies, clinical practices continue to include the use of a number of such variables, so far with inconclusive results.

Summary of Spinal Cord Trauma

The major points regarding spinal cord trauma are as follows. The primary pathoanatomic changes are hemorrhage and edema. The severity is related to the magnitude of the injuring force, or more precisely the energy, the momentum, and the impulse. The hemorrhage is usually in the central gray matter. The

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* Personal communication, G. J. Dohrmann.

source of the hemorrhage is probably small tears in the walls of the muscular venules. Initially, the lesion is more central and then moves peripherally where it involves the white matter and causes most of the clinical damage. There are sometimes associated areas of hemorrhage rostrad and caudad to the site of impact. Steroids, local hypothermia, hyperbaric oxygen, and other agents to some degree have been shown in experimental animals to diminish or eliminate the hemorrhage and/or associated neurologic deficit. The modalities must, however, be applied as soon as possible, at least within the first 4 hours after injury. Neither laminectomy nor any of the other experimental treatments has had any degree of success in the clinical situation.

The functional size of the spinal canal may be further reduced as a result of physiologic movements of the spinal column. In extension of the spinal column, the canal length decreases, thus increasing the cord cross-sectional area. This is a manifestation of Poisson's effect, defined as the increase in the cross-sectional area with decrease in length, or vice versa, the total volume remaining the same. In addition, the cord is compressed by posterior protrusion of the intervertebral disc annulus and buckling of the ligamentum flavum. Another mechanism is active in flexion of the spine. In this physiologic spinal motion, the cord tends to stretch. It may be stretched further over an osteophytic spur, producing local compression and bending.

Alf Brieg pioneered the biomechanical research of the spinal cord and nerve roots.[30,31] Bohlman and colleagues[26] conducted experiments using a canine model in which controlled pressures to the spinal cord over a set of time intervals were applied and the functional output of the spinal cord over time was monitored. Raynor and Koplik[206] published a biomechanical analysis of trauma of the spinal cord. They hypothesized that it is the shear stress in the spinal cord, produced by anterior or posterior compression, that results in spinal cord dysfunction. This conceptual engineering analysis correlated nicely with the findings observed clinically and experimentally by Schneider and co-workers.[221,227] Finally, Tencer and his colleagues, in a series of three papers, have presented *in vitro* experimental results concerning the pressure applied to the spinal cord as a result of various changes in the spinal column by simulating spinal deformity, canal occlusion, functional postures, and surgical procedures.[256–258] These papers have produced reliable biomechanical and clinically relevant data using modern techniques. These investigations employed fresh cadaver spines (C0–S1) including the spinal cord and dura. They used a specially constructed transverse load (force) applicator and a displacement transducer to measure translation. This was designed to apply a force to the spinal cord and continuously monitor the displacement of the anterior and posterior walls of the dura. They studied the effects of several variables on the force and displacement that affected the spinal cord.

Two important findings of the study are depicted in Figure 4-6. In Figure 4-6A, the horizontal axis represents canal occlusion, and the vertical axis indicates the applied load. The relationship is nonlinear. With *increasing* occlusion there is relatively greater increase in the load. The simulation of a kyphotic deformity by removal of a 75% anterior body wedge or performance of a laminectomy did not alter the force–occlusion graph. In Figure 4-6B,

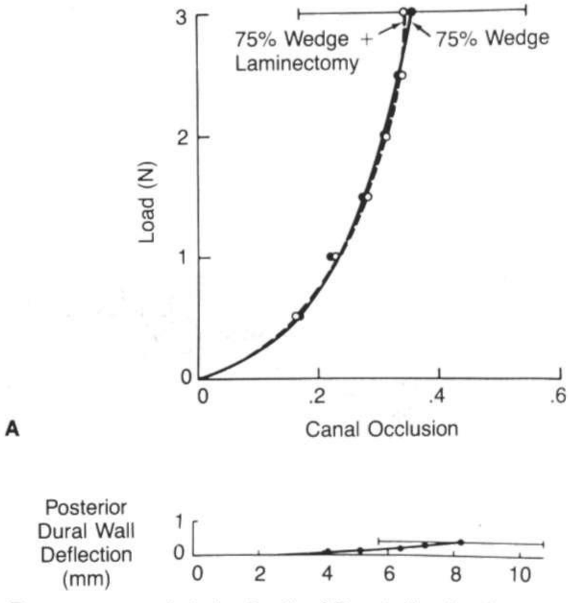


FIGURE 4-6 **(A)** Canal occlusion plotted against applied load. Note the nonlinear relationship. The effects of 75% anterior wedge (producing kyphosis) and laminectomy are not significant. **(B)** Displacement of the posterior wall of the dura as a function of displacement of the anterior wall. *(Adapted from Tencer, A., Allen, B., and Ferguson, R. L.: A biomechanical study of thoracolumbar spinal fractures with bone in the canal: I. The effect of laminectomy. Spine, 10:580, 1985.)*

the relationship between the displacement of the posterior wall of the dura and the displacement of the anterior wall of the dura is depicted. Note that the posterior wall hardly moves when the anterior wall has been displaced up to 6, 8, or even 10 mm. These data support the opinion that anterior impingement does not significantly affect the posterior wall of the dura. Therefore, posterior decompression (laminectomy) would have little effect in decreasing anterior spinal cord pressure in the thoracic spine.

For an understanding of the biomechanics of the pathological conditions of the spinal cord, it is important to analyze the normal functioning of the spinal cord and column. Brieg[30,31] pioneered this type of research in a series of experiments. He used fresh whole cadavers appropriately prepared to show by direct visual observation the movements and deformations of the spinal cord and the nerve roots. Physiologic movements of the spinal column were produced by bending the spine at different regions and in different planes. Several important findings have come out of these studies. Brieg's work showed that the spinal canal length changes are accompanied by spinal cord length changes without producing any significant stresses in the spinal cord. The mechanism is that of unfolding and folding of the spinal cord, much like an accordion (Fig. 1-42). In flexion, the spinal canal increases in length because the centers of rotation of the vertebrae are anterior to the canal. There is stretching of the ligamentum flavum. In extension, the spinal canal decreases in length. The disc protrudes posteriorly into the canal,[33] while the ligamentum flavum protrudes anteriorly.[197] Lesser changes take place during lateral bending and axial rotation. In normal anatomy, these physiologic movements of the spinal column do not produce any abnormal stresses or strains in the nervous tissue. In pathological situations, however, such as hypertrophy of the ligamentum flavum and abnormal disc bulge, osteophytic formations, ossification of the posterior longitudinal ligament, and congenitally narrow spinal canal, these same functional changes may result in abnormal stresses and strains in the spinal cord, leading to neurologic problems (Fig. 4-50).

An osteophyte, a bulging disc, and a hypertrophic ligamentum flavum all apply direct compressive forces to the spinal cord in the transverse plane. This results in specific and well-defined stress patterns in the spinal cord. Although the stress patterns are three-dimensional and extremely complex, a simplified biomechanical analysis is helpful. Raynor and Koplik[206] conceptualized such a situation in biomechanical terms and provided a qualitative analysis of the stress patterns in the spinal cord due to a direct compression applied in the transverse plane.

It is known from engineering analysis of long structures with elliptical cross-sections that when a transverse force, also called shear force, is applied to the structure on its flat side, it results in well-defined stress patterns in the structure. (Stress is defined as force per unit area.) These stresses are of the shear type and have a distribution with the maximum value in the center and zero along the periphery. Raynor and Koplik, in 1985, applied this concept to the spinal cord. They argued that the stresses produced in the spinal cord by the application of transverse force adequately explain the neurologic dysfunction seen during trauma in which the central areas of the spinal cord are preferentially damaged as compared with the peripheral areas. Although the analysis was conducted for the trauma of the spinal cord, the results are equally valid for the application of a chronic transverse force (e.g., one caused by an osteophyte).

A Qualitative Biomechanical Analysis of Stresses in the Spinal Cord

The purpose of this simple analysis of a complex phenomenon is to look at the qualitative stress patterns generated in the spinal cord as a result of a clinically relevant situation (e.g., an osteophytic impingement resulting in compression, stretching, and bending of the cord) (see Fig. 1-44, p. 72). The analysis builds on the earlier work of Brieg[30,31] and Raynor and Koplik.[206] For this biomechanical analysis, the spinal cord is considered a long homogeneous structure with an elliptical cross-section. Although we have used the term *stress*, the term *strain* (percentage change in length) is equally valid in this analysis. Because of the qualitative nature of the analysis, both terms are considered equivalent.

The situation depicted in Figure 1-44 may be seen biomechanically as consisting of three simple loads applied simultaneously to the cord. The loads are the compression load applied by osteophytic impingement, the tensile load due to flexion of the spine, and the bending load due to the change in the direction of the spinal cord over the osteophyte. The stress patterns due to each of the three loads are analyzed separately and then combined.

The first of the three loads considered is the direct compression (impingement) applied on the anterior surface of the cord and directed posteriorly. This force produces local compressive stresses that decrease in magnitude away from the point of contact. In addition, such a compression load produces shear stresses that have a zero value at the point of load application, increase in value toward the middle, and are maximum in the center of the spinal cord. The compressive and shear stress patterns depicted are in the form of iso-stress contours (*i.e.*, each contour represents a single constant stress value).

The second load studied is the tensile (tension) load. The tensile load applied to a long structure produces uniform tensile stress across the cross-section. The third load is the bending of the cord over an osteophyte. When a long structure is bent, there are differential stresses across the bent cross-section. On the concave side of the bent structure, compressive stresses develop, while on the convex side there are tensile stresses. The magnitudes of both of these stresses are highest on the outer surface of the structure. The stresses decrease toward the center and reach a zero value in the middle of the structure. The posterior region of the spinal cord is loaded in both tension and compression. The tensile forces are exerted in flexion, and the compression forces, from invagination of the yellow ligament, are exerted during extension.

Now the effects of all three loads acting on the spinal cord are combined. This simulates a clinical situation in which an osteophyte impinges on the spinal cord while the spinal column is flexed, resulting in combined compression, stretching, and bending of the spinal cord. The resulting stress patterns are summations of the stress patterns shown in Figure 1-44. The compressive and tensile stresses are both normal stresses (*i.e.*, perpendicular to the cross-section of the structure), and therefore they may be combined to provide a joint distribution of the normal stresses. The resulting stress pattern for the normal stresses is as follows. Starting from the anterior part of the cord, we have high compressive stresses that decrease toward the center of the cord. A little beyond the center of the cord, the stresses may be nearly zero. Continuing in a posterior direction, the stresses again increase but are now tensile in nature. It should be emphasized that our analysis is only qualitative; therefore, the precise location of the point where the stresses are zero may vary. In addition, we have the shear stresses (Fig. 1-44). The shear stresses are maximum in the center of the cross-section and decrease toward the periphery. It is interesting to note that the maximum shear stresses occur in the region where the central venules are located. These venules are thought to be the structures in the spinal cord least resistant to mechanical damage.[58]

Current Perspective and Future Needs

Biomechanics can be helpful in providing an understanding of the cervical spondylotic myelopathy (CSM). When external forces are applied to the spinal cord, they result in internal stresses and strains. To determine the basis of neurologic dysfunction due to the loads applied to the cord, it is necessary to obtain precise relationships between the anatomic structures of the spinal cord (gray and white matters and blood supply) and the stresses produced. To obtain such relationships, two items are necessary: detailed quantitative anatomy of the spinal cord and spinal column, and the specific stress patterns generated in the spinal cord. To determine the second item, it is necessary to know the physical properties of the spinal cord and dura and to develop mathematical models to simulate the spinal cord structures.

Although there are several studies providing descriptive anatomy of the spinal cord and column, there is a complete lack of detailed quantitative anatomy. Concerning the mechanical properties of the spinal cord and dura, there are only a few studies.[30,247,256–258] All of these biomechanical studies have been conducted by applying tensile forces to the spinal cord specimen. We have seen that the stresses produced in the spinal cord are of three types: shear, compressive, and tensile. Therefore, studies should be conducted to determine, in addition to the tensile properties, the shear and compressive properties of the spinal cord. Such data are presently not available. To our knowledge, there are no mathematical models available that could relate the external loads applied to the cord to the internal stresses produced in it. However, it is quite feasible to develop such models with the use of contemporary techniques of finite elements.

Experiments using animal models are needed to relate the computed stresses in the spinal cord to the neurologic dysfunction. Many *in vivo* experiments have been conducted in which mechanical pressure or certain compression was applied to the cord. However, the mechanical input was seldom quantified in precise biomechanical terms. These experiments have contributed significantly to our knowledge about the neurologic dysfunction caused by compression, but they are not suitable for relating the mechanical input to the neurologic dysfunction in a precise manner. New experiments must be designed and conducted. Specifically, one should precisely measure: (1) the mechanical input, in the form of the magnitude of cord compression or pressure and its distribution on the spinal cord, and (2) the quantitative functional output of the animal using various neuroelectric monitoring methodologies. Sometime in the future, with the use of biomechanical techniques, it may be possible to relate specific bony pathology to internal stresses in the cord. The stresses, in their turn, may be related to the neurologic dysfunction, with use of the knowledge gained from quantitative animal experiments. Thus, a predictive algorithm may be developed to relate the spinal column pathology to neurologic dysfunction.

A SYSTEMATIC APPROACH TO ANALYSIS OF MECHANISM OF INJURY AND CLASSIFICATION OF SPINE TRAUMA

The problem of accurately analyzing mechanisms of injury in clinical spine trauma is a formidable one. In any given situation, for a mechanism of injury to be identified, a systematic analysis must be carried out. With thorough analysis of cumulative cases, accuracy and understanding are improved. In this section it is suggested that the thorough evaluation of a case of spine trauma, to determine mechanism of injury, include analysis in the following categories: detailed clinical history, detailed physical examination, application of research data, and biomechanical interpretation of radiographs.

Clinical History

The information here is often very unreliable for a variety of reasons. There may be amnesia, with no information or with confabulation. Often, in the excitement of the mishap, there is failure to observe, or there may be inaccurate observation. Sometimes the history is quite definitive, as in seat ejection injury or hangman's fracture. Injuries occur in unfortunate individuals on both sides of the law. Sometimes, especially in athletic injuries, the history is vivid and accurate to the smallest detail and can be confirmed and reexamined on slow-motion film. The clinical history includes information that can be obtained from police records, direct examination, and photography of the accident site. Careful observations of skid marks and associated property damage in conjunction with expert consultation may generate useful and accurate information. On-site examination or photographic study of vehicles involved in an accident can provide clues or hard data. In a diving injury, where possible, a site visit can often contribute significantly to piecing together the complete analysis. One should gather as much relevant data as possible and weigh it according to circumstances, sources, and internal consistency.

Physical Examination

Bruises, lacerations, and associated fractures about the face and head can be helpful in determining the possible direction of force vectors. However, impact trauma may occur on *rebound*, and a particular laceration may not be indicative of the direction of the force that caused the injury to the spine. Associated fractures in other areas of the spine, as well as in the long bones, may be helpful in determining the magnitude and direction of vectors involved in the injury.

Application of Research Data

There are basically three types of research information that may be usefully applied—clinical, experimental, and mathematical. There are a number of series of cases in the literature in which the mechanism of injury is reasonably well defined and the pattern of failure of the spine is well documented. Examples include hangman's fractures, seat ejection fractures, seat belt fractures, and some compression fractures of the thoracic and lumbar spine. A number of experimental studies have subjected animals and cadavers to various types of trauma in controlled situations. The injuries were studied anatomically and through movies, cineradiography, and radiography. These data can be reasonably extrapolated and applied to clinical situations in a manner that provides additional analytical insight and some standard with which to check "deduced" mechanisms of injury.

Biomechanical Interpretation of Radiographs

Here, the clinician analyzes the radiographs, taking into consideration the information from the history and physical examination. In addition, the radiographs are interpreted in the context of current biomechanical knowledge and principles. We know from the work of Crowell and co-workers that radiographs are insufficient to display all injuries. These investigators noted that experimentally produced bony and soft-tissue failure was sometimes not recognizable on x-ray or even direct observation of the anatomic specimen unless the mechanism of injury displacement was reproduced[49] (Fig. 4-7). Some guidelines for the biomechanical interpretation of radiographs of the spine after trauma follow. These are not absolute rules, but they are acceptable maxims based on current knowledge.

Guidelines for Biomechanical Interpretation of Radiographs

Bone tends to fail first along lines of tensile stress, then it may fail as a result of either shear or compression. Presumably, it is best designed to withstand compressive loads.

One may assume that, in the cervical spine, the anterosuperior or the anteroinferior triangle of bone, seen frequently on lateral radiographs, is pulled off by the peripheral annulus fibrosus fibers. Thus, the triangular portion of the vertebral body (actually the anterosuperior rim) may be pulled off in an extension type of injury, where the triangular fragment stays with the annular fibers, which remain attached to the intact vertebra, and the fractured vertebra is pulled away by the tensile loads. However, it is also possible in a flexion injury for compressive loading to result in high shear stresses, causing failure along approximately the same lines in which tensile failure would occur (Fig. 4-8).

When this triangular fragment or "tear drop" is observed, beware the possibility of an occult comminuted fracture of the same vertebral body (see Fig. 4-2).

In compression of a functional spinal unit, the vertebral body end-plate generally fails first; however, recent experimental work by Crowell and co-

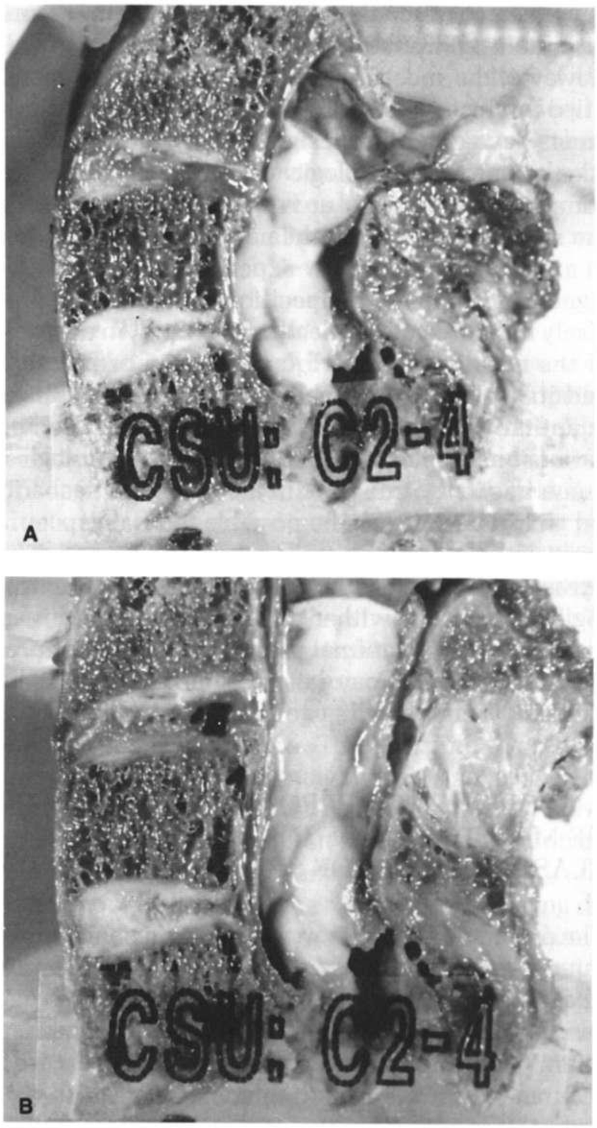
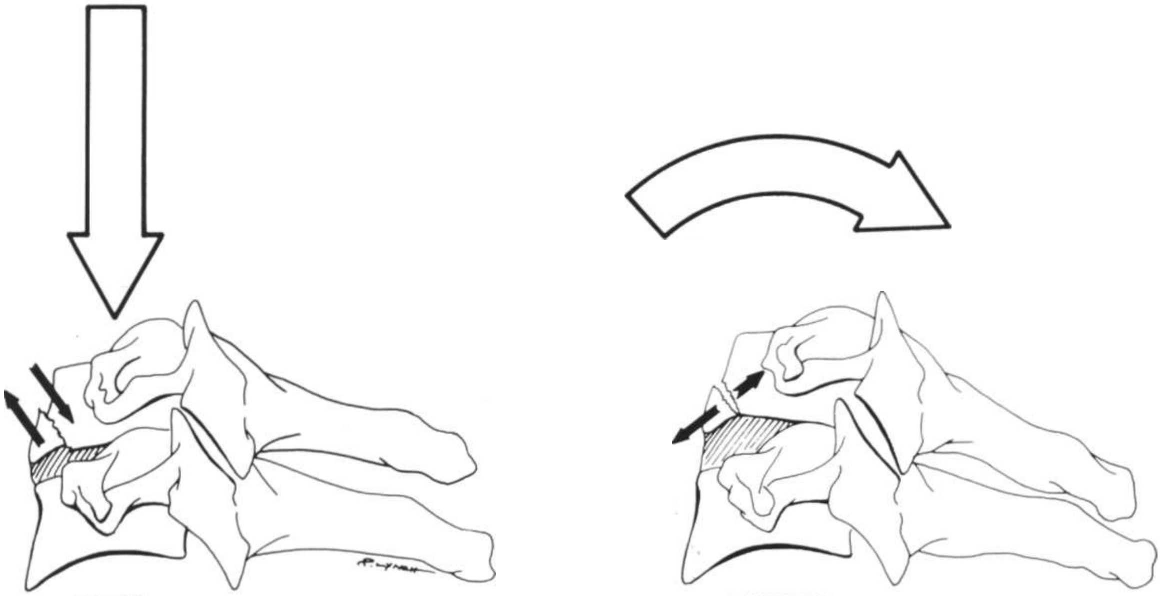


FIGURE 4-7 **(A)** This injury was occult even on direct observation. **(B)** Upon reproduction of the displacement involved in this *experimental* mechanism of injury, it is possible to recognize the disruption and failure of the posterior annulus fibrosus, posterior longitudinal ligament, and the capsular and interspinous ligaments. Dynamic loading, as with a stretch test, is likely to bring out such an injury in the *clinical* setting. *(From Crowell, R. R., Coffee, M. S., Edwards, W. T. and White, III, A. A.: Cervical ligament healing under three dimensional loading. Proc. Cervical Spine Research Society, 1987.)*



SHEAR

TENSION

FIGURE 4-8 A triangular fragment of bone at the anteroinferior portion of the vertebral body may result from either shear or tensile failure. In compressive loading there are shear stresses along a line about 45° to the force vector. In extension loading, the same region of bone is subjected to tensile stresses.

workers[49] has revealed occult disc injuries in the absence of vertebral end-plate fractures.

Wedging of the configuration of vertebral bodies occurs as a result of compression from eccentric forces.

In the normal anatomy of the functional spinal unit, with most loading vectors, the bone tends to fail before the ligamentous structures. There are exceptions.

Where there is wide separation between the anterior and posterior elements, indicating ligamentous rupture, most probably a significant element in the mechanism of injury is axial rotation (torque about the y-axis).

Narrowing between vertebral bodies at a given interspace where there is other evidence of trauma is suggestive of failure of the annulus or its attachment, implicating a mechanism involving shear or tensile loading. Abnormal separation at the disc space has the same implications.

There is an important variable that is time-dependent and must be taken into consideration. In any injury to a complex structure such as the spine, there are a number of changes that occur between the onset and completion of the injury. The structures change in geometric and physical properties, and the force vectors of injury change in direction and magnitude. As a consequence, the clinician is faced not with just the analysis of one isolated injury, occurring instantaneously at one structure, but also with a series of rapidly changing injury mechanisms occurring at a series of rapidly changing structures. Current analyses need not seek this level of complexity. However, the oversimplification involved in the use of a static two-dimensional representation of a complicated series of dynamic, three-dimensional events should be kept in mind.

The radiographic evidence is interpreted and analyzed along with available information from the clinical history, physical examination, and research data. A major injuring vector (MIV) is determined. This is challenged by the evaluator, to see that there is internal consistency and that the information from all four categories corresponds. Rarely will everything fit perfectly, but the extent to which this or some other systematic approach is applied determines the accuracy of analysis and the level of understanding.

Designation and Analysis of Injury Using MIV

In 1972, Roaf suggested an international classification for spinal injuries.[211] Here, an independently determined system, similar to the one suggested by Roaf, is used. The coordinate system has a different orientation and has been submitted for consideration as an international standard.[189, 283] Concluding that an injury is merely the result of flexion or extension is not an adequate analysis of the forces acting on the functional spinal unit. The text is written so that the reader who does not, so to speak, get down to the xyz of it can progress smoothly without the classification. However, the reader is encouraged to consider the system, because its use is a step toward a biomechanical understanding of the spine.[C]

The system is as follows: Dislocations are described cephalocaudally; the more cranial unit is described as dislocated in relation to its normally positioned subjacent fellow. Examples include occipital-atlantal, atlanto-axial, C5 on C6, T8 on T9, and L3 on L4. The dislocations are described as posterior, anterior, lateral, or rotary.

The term *atlantooccipital* dislocation is seen in the literature. However, the term *occipital-atlantal*, or C0–C1, dislocation seems preferable because it is consistent with the tradition of naming the more cephalad unit first. One uses C5–C6 dislocation or lumbosacral dislocation rather than C6–C5 dislocation or sacrolumbar dislocation.

Occasionally the coordinate system is used in parentheses to indicate the displacement. The anteroposterior radiographs are viewed and described as though the examiner is behind the patient, looking at the back of the radiograph. The examiner's right is the patient's right. Open-mouth views of the dens are viewed with the examiner and patient face to face.

The term *major injuring vector (MIV)* is used to describe the mechanism of injury. Any given injury results from a complex series of forces and moments applied to the body in a variety of different ways. Ultimately, the loads are transmitted to the functional spinal unit in the region of the spine where the injury takes place. The most important complex of forces that causes the injury may be summarized and represented by the vector MIV. The coordinate system and a three-dimensional diagram to show the most representative orientation in space are used. The diagrammatic spatial orientation is occasionally supplemented by designating where the MIV is pointing in the coordinate system. No attempt has been made to estimate or indicate vector magnitude. Where certain bending motions are of particular importance they have been indicated on the diagram.[D]

REVIEW OF SOME SPECIFIC CERVICAL SPINE INJURIES

Fractures of the Occipital Condyle

A recent review of these fractures has provided a useful update on the pathoanatomy, mechanism of injury, and potential instability of some of these fractures.[8a]

Mechanism of Injury

The investigators suggested three classifications of injury as follows: Type I—impacted occipital condyle fracture; Type II—occipital condyle fracture associated with a basilar skull fracture; and Type III—an avulsion fracture of the occipital condyle. The mechanism of injury (MOI) suggested for Types I and II is vertical (y-axis) compression. For Type III, a combination of anteroposterior (z-axis) translation and axial (y-axis) rotation with the associated force and movement is the cause of injury (Fig. 4-9A).

Discussion

Fracture of the occipital condyle is a rare, or at least not frequently diagnosed, injury. The MOIs are based on rational theoretical analysis. If the MOI for Types I and II is correct, one would expect to see the injury more frequently in association with Jefferson's fractures. Montesano and Anderson do indicate that the diagnosis is difficult to make. Thus, the injury may be more common than we have thought. The MOI offered for the avulsion injury is also feasible. We believe that there may be a significant element of horizontal shear involved in the Type III injury. The pathoanatomy and biomechanics of the alar and capsular ligaments in this injury are keys in the determination of whether shear or avulsion is the MOI here. It appears that it is an avulsion of that portion of the occipital condyle to which the alar ligament is attached (see Fig. 4-9B).

The authors, basing their opinion on an appropriate contemporary analysis of the anatomy and biomechanics of the transverse and alar ligaments,[69] have warned of the possibility of instability in association with Type III occipital condyle fractures. This is because the alar ligament, which attaches to

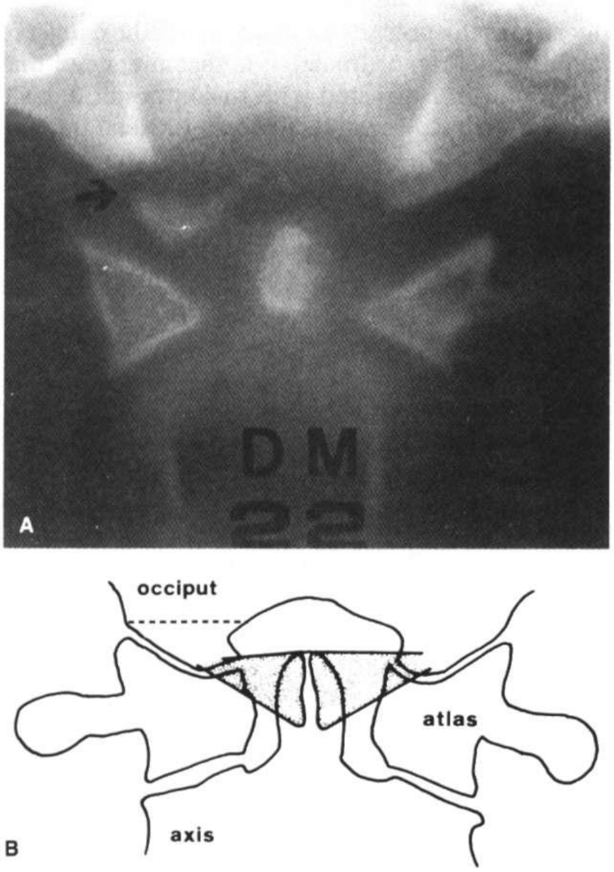


FIGURE 4-9 This 22-year-old man sustained an occipital condyle avulsion fracture in a motor vehicle accident. **(A)** The tomogram demonstrates the avulsed condyle. This is an example of a type III occipital condyle fracture that is possibly unstable. **(B)** The attachment of the alar ligaments to the dens and to the occipital condyle is shown. The dotted line is the site of the avulsion fracture. (**A** *from Anderson, P. A., and Montesano, P. X.: Morphology and treatment of occipital condyle fractures. Spine, 13:731, 1988;* **B** *modified from Dvorak, J., Schneider, E., Saldinger, P., and Rahn, B. A.: Biomechanics of the alar and transverse ligaments. J. Orthop. Res., 6:452, 1988.*)

the occipital condyle, is lost as a stabilizer. This and other issues about this fracture need further study before one can be certain of these reasonable hypotheses.

Suggested Treatment

The authors warn that there must be a high index of suspicion and that the diagnosis cannot be made on plain x-rays. The suggestion is that computerized tomography (CT) scans, with sagittal and coronal reconstructions, or tomograms are required for thorough evaluation. Types I and II may be treated with a Philadelphia or a SOMI-type orthosis for 6–8 weeks and followed. The Type IIIs, according to the investigators, should probably be treated with a more rigorous orthosis and followed with flexion/extension views or a stretch test (page 318, Chap. 5). These recommendations are preliminary, and more work is needed.

Occipital-Atlantal (C0–C1) Dislocation

This often fatal injury usually consists of an abnormal anterior displacement of the skull in relation to the atlas (Fig. 4-10).

Mechanism of Injury

The mechanisms in these injuries are speculative. Presumably, there is a major force vector of high magnitude along the + z-axis. The skull is translated anteriorly in relation to the atlas. Mainly, this causes a shear type of loading at the atlantooccipital joint, rupturing the articular capsule and causing the dislocation. It should be noted, however, that Eismont and Bohlman[72] and Bucholz and Burkhead[35] studied 112 postmortem victims and suggested a hyperextension mechanism. This was based on the associated findings of submental lacerations, mandibular fractures, and an intact atlantooccipital membrane. The investigators noted that the C0–C1 dislocations are more common in children. It may be that the capsule of the occipital-atlantal articulation and/or occipital-dens portion of the alar ligament is less well developed in the child.

Discussion

In this injury, the central nervous system is usually damaged or transected at the medulla oblongata or at the spinal medullary junction. The injury usually causes death and may not be recognized at autopsy unless specifically looked for with flexion/extension radiographs. Alker and colleagues studied 146 accident fatalities and found that 21% had demonstrable neck injuries. Most of these injuries involved C0–C1 or C1–C2.[5]

There is a reported case of a patient who survived with an occipital-atlantal dislocation.[93] This involved a 41-year-old male who was in an automobile accident. There was a lateral displacement of the occiput in relation to the atlas. The MIV was oriented along the x-axis. The injury was treated with a

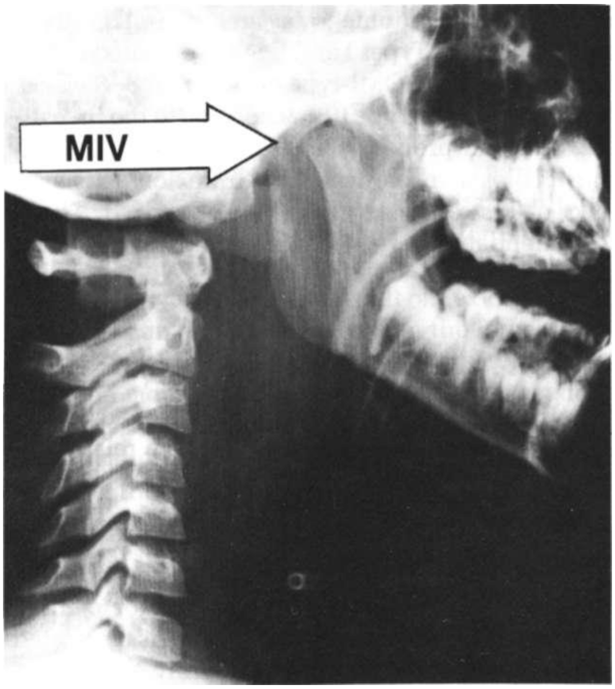


FIGURE 4-10 A radiograph of a 9-year-old girl who sustained an occipital-atlantal dislocation when hit by an automobile. The MIV is a large magnitude force vector, directed mainly in the positive direction along the z-axis (anterior translation in the sagittal plane). There is also an increased space between the spinous processes of C1 and C2 suggestive of disruption of the posterior ligaments and/or a fractured dens. The patient died shortly after this injury, which is usually fatal. With improvements in training of emergency medical technicians and in equipment some of the patients are surviving. With tensile failure of the posterior ligaments, including the atlantooccipital membrane, extension is an unlikely MIV in this particular patient. We note in the text, however, that several investigators have considered C0–C1 dislocations an extension injury. *(Courtesy of Teaching Library, Department of Diagnostic Radiology, Yale University School of Medicine.)*

small amount of traction in a cervical brace for 5 months. Following this, the patient had a sensation of precariously balancing his head on his neck and had a clicking sound on moving his head. He was treated with a posterior craniocervical fusion.

An excellent review of the literature on this subject by Van Den Bout and Dommisse is available.[269]

It should be mentioned that occipital-atlantal dislocation has been reported in patients who have not suffered trauma.[77,154,277,288]

Suggested Treatment

Because of the precarious stability and the danger of fatality with minor injury, it is recommended that these injuries be treated with fusion from C0 to C2 following 2 to 3 weeks of moderate skeletal traction with 3–4 kg (6–9 lb) of weight.[H]

Fractures of the Posterior Arch of C1

Such a fracture occurs just behind the lateral masses, where the ring is grooved by the vertebral artery (Fig. 4-11).

Mechanism of Injury

Fractures of the ring of the first cervical vertebra are thought to occur primarily as a result of vertical compression on the posterior aspect of the arch of C1.[232] There may also be some acute rotation of the skull counterclockwise about the x-axis (some element of extension). In addition, the lateral masses of C2 may serve as a fulcrum at the site of injury. The anterior arch of C1 is locked, buttressed, or fixed by these lateral masses, while the posterior ring is displaced in the caudad ($-y$) direction. This results in a tensile failure of the bone on the cephalad surface of the ring, where these fractures generally appear to start. This is compatible with the presumed mechanism of injury and the generally accepted maxim that bone fails in tension. The cephalad surface of the ring is in tension in this injury mechanism. This is the weakest point of the ring because it is the thinnest portion and has the lowest area moment of inertia against the type of loading involved. This is due to grooving at the site bilaterally, where the vertebral artery courses from its osseous vertebral canal across the ring of C1 and into the foramen magnum.

Since this fracture is sometimes seen along with the so-called hangman's fracture, presumably the two injuries have similar mechanisms (Fig. 4-12). In both situations, some element of extension is involved. However, in this fracture the major associated force is compression, whereas in the other injury, bending is a significant factor.

Discussion

The symptoms associated with this injury are headaches, suboccipital pain, or nonspecific pain associated with stiffness. The diagnosis can be readily made by a lateral radiograph of this area.

In rare instances, the vertebral artery may be in-

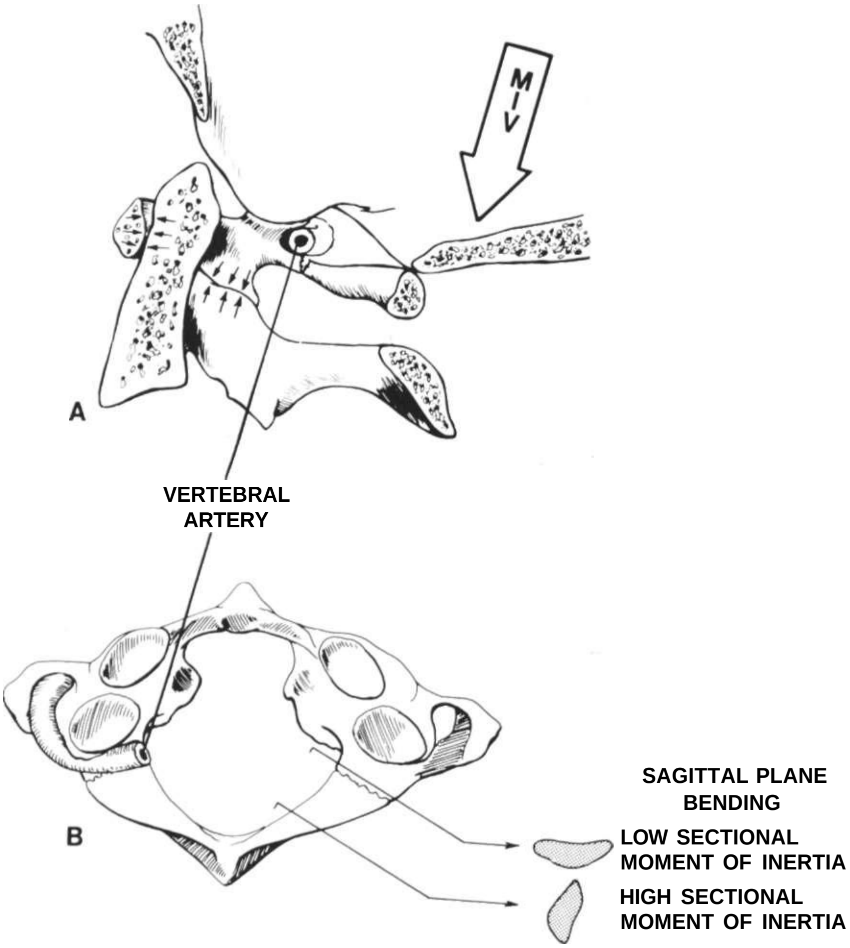


FIGURE 4-11 This diagram demonstrates how biomechanical and anatomic factors work together to result in a typical fracture of the posterior arch of Cl. (A) A midsagittal section of the occipital-atlanto-axial complex shows a possible mechanism of injury, one that causes fracture of the ring of Cl. A force causing extension (—x-axis rotation) results in fixation of the anterior ring of Cl against the dens and fixation bilaterally at the articular condyles between the ring of Cl and C2. With this fixation, it is possible for the impingement of the occiput against the posterior ring of Cl to cause a bending moment about the ring of Cl, as shown. Note the position of the vertebral artery, at which point the ring of Cl is grooved and weakened. (B) The top view of the ring of Cl shows the vertebral artery on the left in the region of the fracture. On the right side, the sectional moment of inertia against bending in the sagittal plane is much smaller where the fracture occurs than it is in the more posterior area, where the resistance to bending in that plane is much greater. The areas are shown by the cross-sections at these two points. In addition to the considerations of the effects of the sectional moment of inertia on the location of the fracture site, there is another factor. Forces are applied at the tip of the ring of Cl by the occiput; thus, the maximum bending moment is also at the site where the fracture occurs.

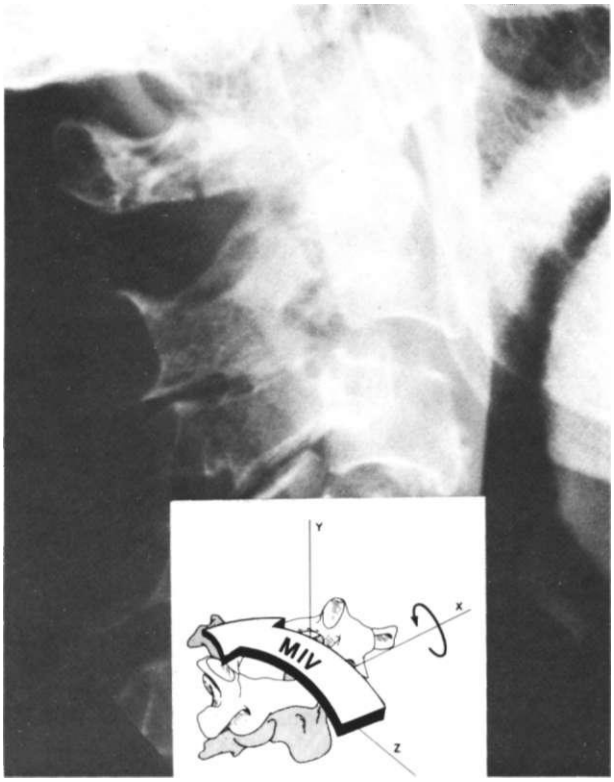


FIGURE 4-12 A radiograph of an elderly woman who fell down a flight of stairs and sustained a fracture of the posterior ring of C1 and a traumatic spondylolisthesis. This exemplifies the association of the fracture of the ring of C1 with traumatic spondylolisthesis. The ring of C1 is thought to be bent such that the convexity points caudad. Note that the fracture appears to begin on the caudad side of the ring. The MIV is a negative bending moment in the sagittal plane about the x-axis, shown diagrammatically in the inset. This injury is interesting to contrast with the one shown in Figure 4-11 where the fracture starts on the cephalad side of the ring. In this fracture, there is a traumatic spondylolisthesis and failure of the annulus anteriorly. Note the abnormal separation between the bodies of C2 and C3. As a result of these two injuries, C1 is *not* fixed by impinging against a stable C2, as in Figure 4-11. Consequently, it rotates and impinges on the posterior elements of C2, creating a bending moment in the same plane but in an opposite direction. Thus, the fracture starts at the caudad portion of the ring. Such a fracture of the ring of C1 should alert the clinician to the probability of disruption of the anterior elements of C2.

volved, which may cause symptoms of basilar insufficiency. In the classic review by Jefferson in 1920,[128] he referred to two patients who had vertebral artery involvement, mentioned by Delorme in 1893.[54] In the review by Sherk and Nicholson, arteriovenous fistula involving the vertebral artery was discussed as a complication of this injury.[232] Arteriography should be considered in cases where there is a question of vascular involvement.

Suggested Treatment

When these fractures are not displaced or are only moderately displaced and not comminuted, they may be presumed to be stable. This should be confirmed with flexion/extension films, taken with caution after local symptoms subside, provided there is no neurologic injury. These injuries may be treated for 3 months with an orthosis that offers intermediate control of the cervical spine. This means a cervical appliance that has shoulder and thoracic support, in addition to a cervical collar (see Chap. 7 for a discussion of orthoses).

Comminuted Fracture of the Ring of C1

This so-called Jefferson fracture is classically described as a four-part fracture of the ring of the atlas.

Mechanism of Injury

The mechanism was first described in 1920. The fracture is a result of compression, usually due to a direct blow on the vertex of the head. The spatial orientation of the occipital condyles is such that when they are driven axially in the caudad direction, they act as a wedge, causing a bursting effect and separating the ring of the atlas, usually into four parts. This is shown in Figure 4-13. The mechanism of this injury is not unlike that of a simple fracture of the ring of C1. There are two differences. The MIV in the Jefferson fracture has a greater magnitude, and it is more in line with the y-axis.

Discussion

Alker has indicated that, despite the traditional description of the "classical" Jefferson fracture shown in Figure 4-13, he has never seen it in just that form. It is usually a two-part or a three-part fracture (Fig. 4-14). His experience is extensive and includes numerous CT studies of this region following trauma.[5]

The salient radiologic characteristic of this fracture is bilateral displacement of the articular facets of the ring of C1, seen on an open-mouth view with the head in a neutral position (Fig. 4-15).

Suggested Treatment

In the large majority of cases, this is a stable injury without neurologic involvement and may be treated by protective immobilization with a cervical orthosis of intermediate control. If there is gross lateral displacement of the lateral joints, their position may be improved by traction. In the less common situations where instability can be demonstrated, fusion of C0 to C2 should be considered. The surgical constructions for this procedure are discussed in Chapter 8.

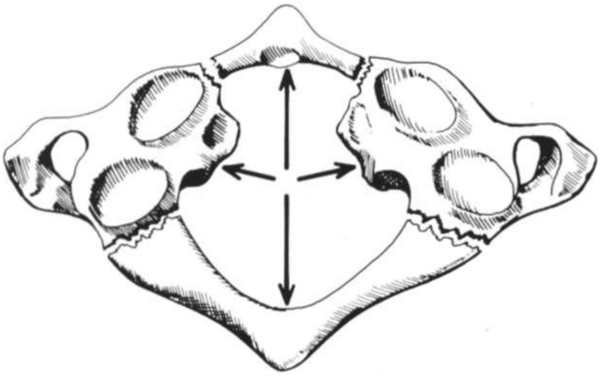


FIGURE 4-13 This is the classical fracture described by Jefferson. Separation of the ring of C1 into four fragments is shown at the usual sites of failure. This fracture is rare, according to Professor G. Alker, M.D.[5]

Fractured Dens (Odontoid)

The authors respectfully submit that the correct anatomic name for this structure is *dens*. Fractures in this region have been classified by Anderson and D'Alonzo into three types, shown in Figure 4-16.

Type I is an oblique fracture in the isthmus of the dens. Type II occurs at the junction of the isthmus with the vertebral body of C2. Type III is actually a fracture not of the dens but of the body of C2 in the region of the base of the dens (Fig. 4-17). These three types are important in the clinical biomechanics of this injury.

Mechanism of Injury

Previous hypotheses have suggested hyperextension, hyperflexion, and horizontal shear as possible MOIs of dens fractures.[219] This hypothesis stimulated the analysis and analogies of mechanism of injury depicted in Figure 4-18.

Well-designed and controlled experimental studies by Altoff[7] showed that fractures could not be produced by pure horizontal shear. He was able to produce fractures of the dens by a combination of horizontal shear and vertical compression. When the combined compression/shear force vector was

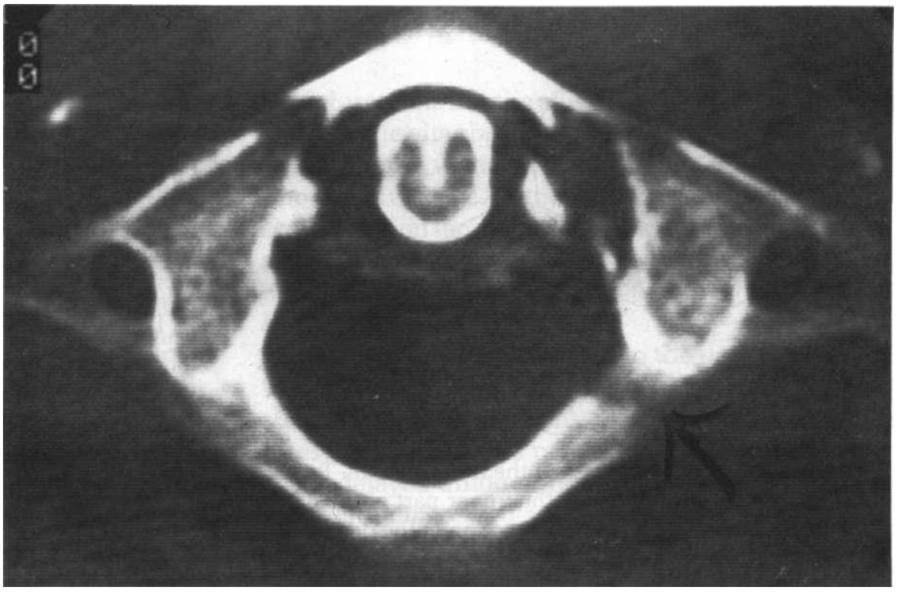


FIGURE 4-14 This three-part fracture is much more common than the four-part fracture that carries the Jefferson eponym. *(Radiograph courtesy of Professor George Alker, M.D.)*

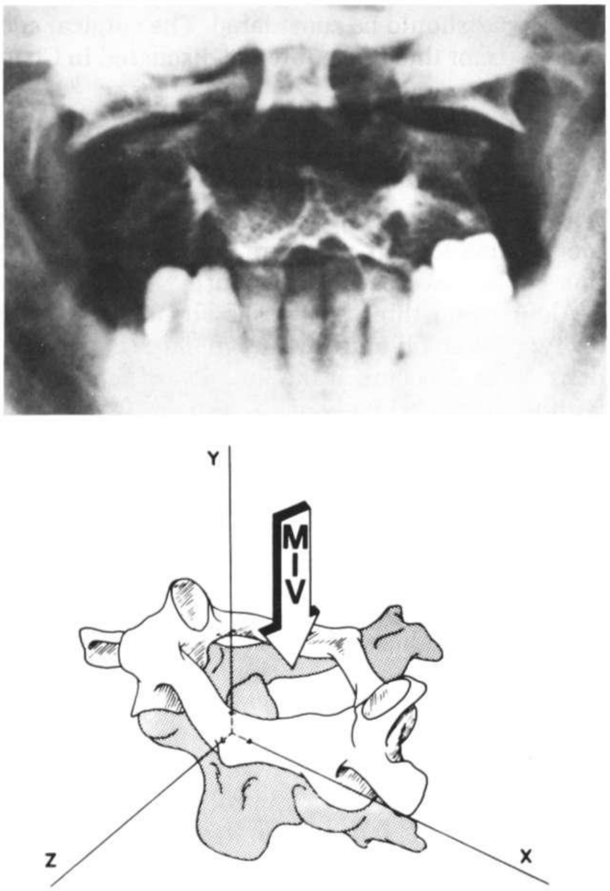


FIGURE 4-15 A radiograph of a 47-year-old male with a Jefferson fracture. Note the wide separation of the lateral masses on the open-mouth view of the dens. The MIV is shown on the inset. The orientation is not always exactly parallel with the vertical (y) axis.

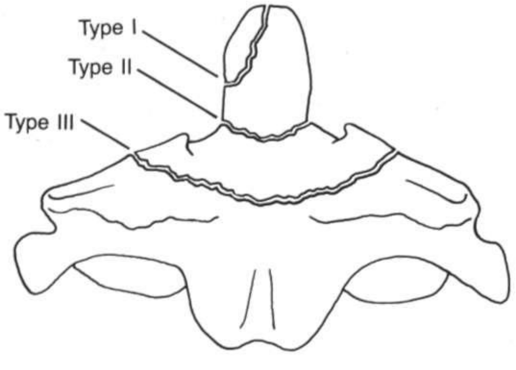


FIGURE 4-16 Dens fractures in the frontal plane. Type I is an oblique fracture through the upper part of the dens itself. Type II is a fracture at the junction of the dens with the vertebral body. Type III is actually a fracture into the body of the atlas, which anatomically is not literally a dens (odontoid) fracture.

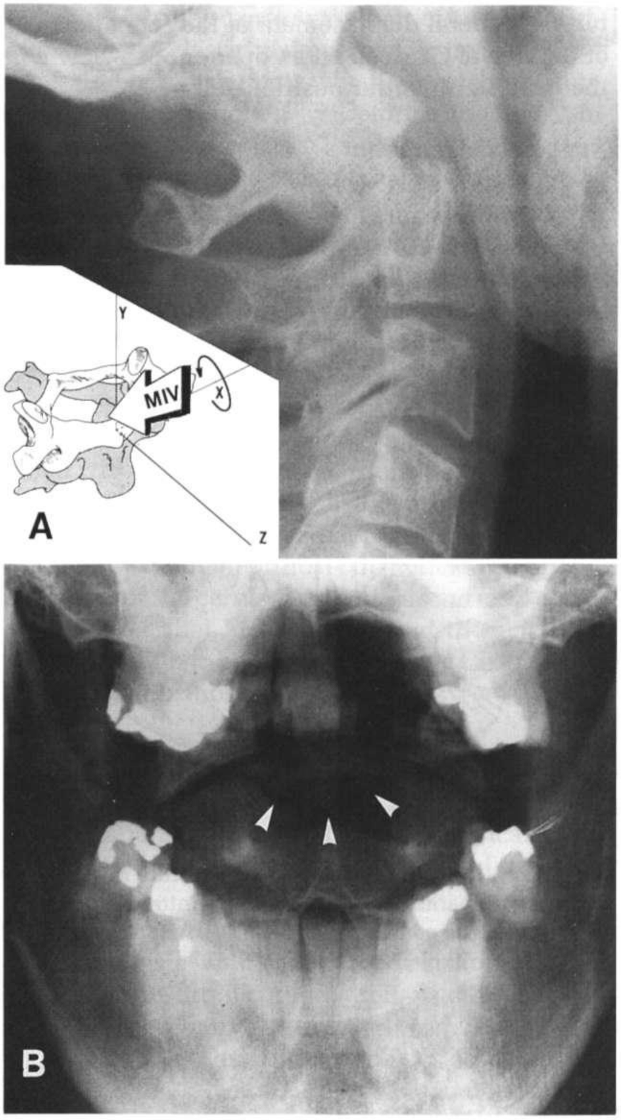


FIGURE 4-17 Radiographs of a 33-year-old man who was struck in the front of the head with a heavy weight. He sustained a Type III fracture of the dens, which is the fracture that is most likely to heal. **(A)** With the posterior dislocation and a history of a blow to the front of the head, we assume an MIV as shown. The vector is primarily in the sagittal plane. The impact produces shear as well as an element of vertical compression. In comparison, there is relatively more shear here and relatively more compression than shown in Figure 4-12 (traumatic spondylolisthesis and fracture of the ring of C1). **(B)** The true fracture, documented on laminagrams, is indicated by the three small arrows. *(Courtesy of Teaching Library, Department of Diagnostic Radiology, Yale University School of Medicine.)*

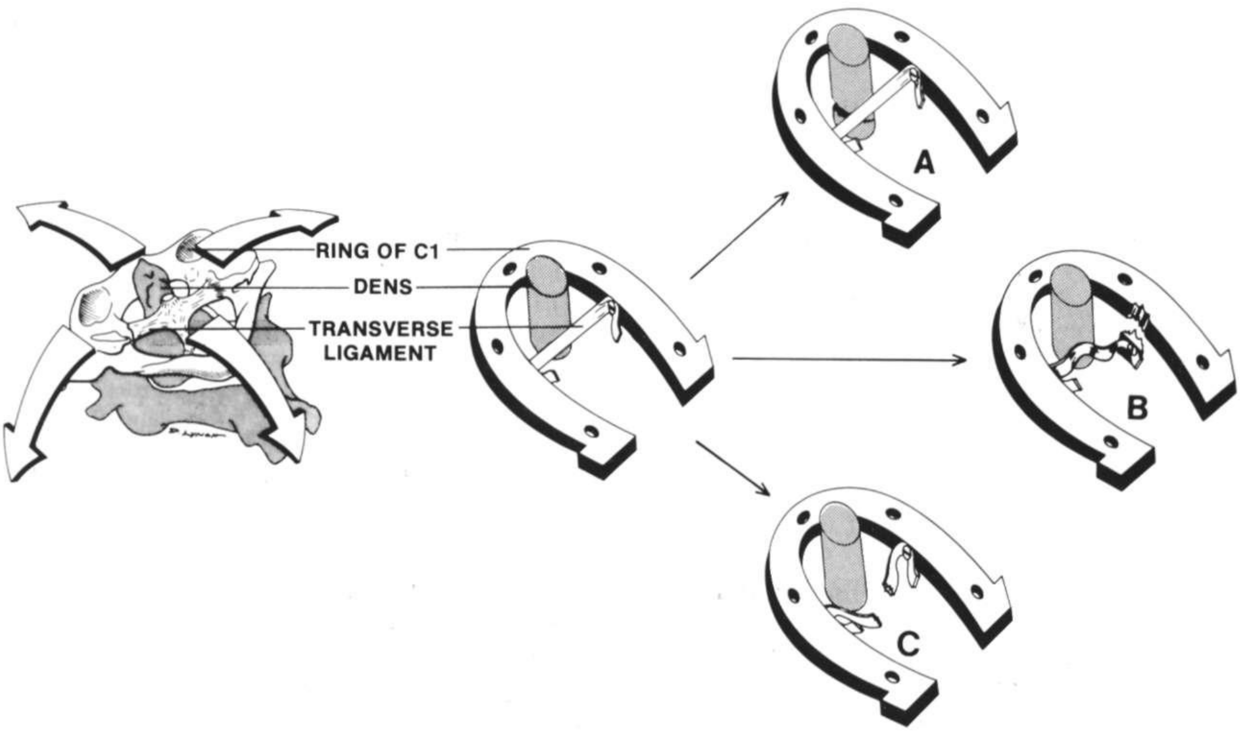


FIGURE 4-18 Representation of the relationship between the dens and the transverse ligament in the ring of C1. A stake represents the dens, a horseshoe represents the bony ring of C1, and a bridle strap represents the transverse ligament. With various types of loading on these structures, three basic kinds of injury can occur. **(A)** There can be a fracture of the dens (a failure of the stake) with displacement in any direction. This may be due to impingement of the horseshoe against the stake, or impingement by the bridle strap. **(B)** There can also be a failure of the attachment of the stake or transverse ligament to the horseshoe. When this occurs, it is sometimes possible on the open-mouth view of the dens to see a small fragment of bone lying between the dens and the lateral mass. **(C)** This shows the simple failure of the transverse ligament (bridle strap).

applied directly in the sagittal plane, fractures analogous to an Anderson and D'Alonzo[8] Type III fracture were produced. Moving the experimental force vector 45° out of the sagittal plane (*i.e.*, applied from an anterolateral direction), a fracture analogous to the Type II fracture was produced. When the vector was applied from a direction 90° to the sagittal plane (along the x-axis), a fracture analagous to Type I was produced. A liberal interpretation of this experiment is as follows: as the vector producing compression and shear moves out of the sagittal plane and is directed more laterally, the dens fracture produced is relatively more cephalad. The various MIVs suggested in Altoff's work are depicted in Figure 4-19.

If, in fact, immobilization proves to be the key issue, it may make the more aggressive approach of anterior screw fixation relatively more appealing in the future. This has the added advantage of not eliminating the important axial rotation of C1–C2, which occurs with the posterior fusion techniques. The disadvantage is that, at present, the anterior screw fixation tends to be associated with more complications.[42]

Discussion

The issue of the blood supply to the dens is crucial. There is a question as to whether fractures of the structure in certain regions, particularly Type II fractures, interfere with a precarious blood supply and cause delayed union or nonunion. The work of previous clinicians[219, 220] and anatomists[231] justifies the conclusion that the dens has a rich arterial supply top and bottom with ample anastomoses. The effect of experimental fractures of the dens on its blood

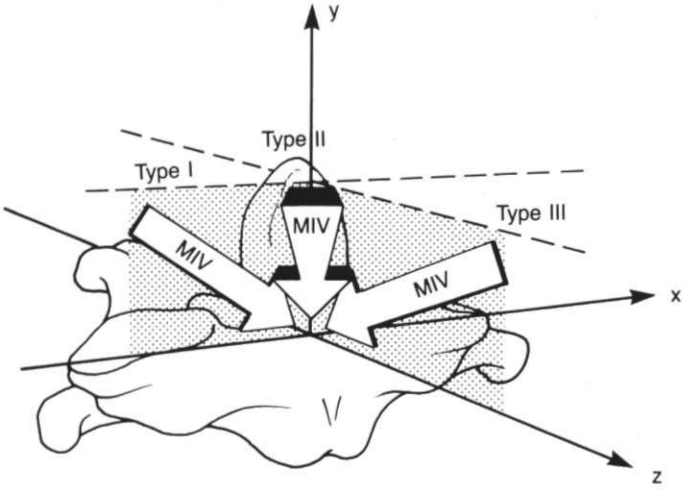


FIGURE 4-19 Based on the experimental work of Altoff,[7] a significantly different MIV could be determined for each of the three types. We have presented these in an attempt to show the major injuring vector for each of the three fractures. As shown, they all come in at about a 45° angle to the vertical. The MIV for Type I is mainly in the x,y plane, Type III is mainly in the y,z plane, and Type II is in an oblique plane halfway between the other two.

supply was studied *in vitro* by filling the arteries in the region with contrast medium either before or after fracture. It was noted that arteries entering at the base anastomose with those entering the apex. In some instances, the experimental fractures of the base of the dens involved injury to the ascending arteries. In all these *in vitro* experimental fractures, arteries within the dens were filled. Type II fractures of the isthmus filled from arteries entering the apex.[7] This information makes it *unlikely* that failure of union of Type II dens fractures is due to loss of blood supply to one of the fracture fragments.

These[7,219,220] investigators and others hold that inadequate immobilization is more likely than inadequate blood supply to be the cause of nonunion in these fractures.

If it is not an issue of blood supply, what is it? In a study of 48 patients, Altoff made the following cogent observations: (1) the rate of bone union was increased directly with the extent to which the fracture involved the body of C2 ($p < 0.05$); (2) fractures displaced anteriorly had a better prognosis for union than those displaced posteriorly ($p < 0.05$); (3) the union rate was higher in those patients treated in traction for 6 weeks than it was in those treated in a collar or Minerva jacket.

Displacement is important for the obvious reason that it decreases contact surface and is associated with less effective fracture healing.[188] It is also useful to remind ourselves that displacement in two planes greatly reduces contact surface.[217]

Clark and White[42] completed a multicenter study of dens fractures sponsored by the Cervical Spine Research Society. It was noted that Type II fractures that were either significantly angulated or displaced, or both, had an increased probability ($p < 0.05$) of malunion or nonunion, or both. In general, the Type II fractures were described as the most troublesome, but it was noted that Type III fractures can also progress to nonunion.

Dunn and Selijeskog, in a review of 110 patients, considered age greater than 65 years, a Type II fracture, a major posterior subluxation >2 mm, and a late diagnosis (>7 days' delay) to be indicators of a poor prognosis.[68]

The factors important in prognosis of dens fractures are listed in Table 4-1.

Suggested Treatment

Current knowledge suggests that treatment of the undisplaced dens fracture is based on a classification into three types.[8,239] The Type I fractures are rare in the strict sense of the classification.[42] Treatment with a cervical orthosis of intermediate control is satisfactory. Even failure of union should not be a

TABLE 4–1 Factors That May Detract from an Excellent Prognosis in Dens Fractures

- Type II fractures
  - $>$Age 65
  - With angulation
  - With displacement
- Posterior displacement >2 mm
- Displacement in more than one plane
- Delay in diagnosis
  - Particularly >7 days
- Inadequate immobilization

problem if the fracture is above the level of the transverse ligament and does not cause any loss of stability.

Type II fractures are probably the most difficult and controversial. In this type, the rate of failure of union is 32% with halo treatment.[42] The alternative approach is to proceed with elective fusion of Type II fractures in adults who are not undue surgical risks. There are other considerations here also. With surgery, at least 10–15% of neck rotation is lost.[8] This figure may be higher, since normally about 50% of axial rotation in the neck is at C1–C2. On the other hand, a good deal of inconvenience and time and monetary loss may result when there is residual instability following conservative treatment. If the patient recovers without surgery, then of course there is no exposure to the unique and usual surgical risks involved. It is probably advantageous to present the pros and cons of surgery to the patient. When several of the factors listed in Table 4-1 are applicable, perhaps one should lean toward surgery. If surgery is elected, the authors recommend Brooks' technique (see Chap. 8).[103]

In the authors' opinion, surgical intervention is relatively more appropriate for dens fractures that have several characteristics listed in Table 4-1.

Type III fractures (see Fig. 4-17) may be expected to heal satisfactorily with conservative treatment, which consists of 10–14 days of traction, followed by a cervical orthosis that offers the most effective control for an additional 18–20 weeks. The healing rate is over 90%,[8] but stability should be demonstrated by flexion/extension radiographs. Note that in the Altoff thesis[7] there was a statistically significant increased incidence of union in those patients treated with 6 weeks of skull traction.

There is an important point to be made about Type III dens fractures. The multicenter study by Clark and White[42] indicated that this fracture is not as benign as has been previously reported. It is better to slightly overtreat than to undertreat this injury. In other words, a halo device should be considered in this type of injury. Available information does not justify being more definitive at this time.

Traction is applied primarily to anchor the head and thereby splint for soft-tissue healing. It is not a particularly effective mechanism for reducing a displaced dens fracture. In fact, one should take special care not to distract the fragments.[217] This can be done by appropriate x-ray monitoring. When there is no neurologic deficit and fusion has been elected, complete anatomic reduction of the dens fracture is not necessary.

Summary of Treatment Suggestions Because of the complexity and controversy regarding this topic, a brief synopsis is offered. Type I fractures and most Type III fractures may be treated without surgery—Type I with a less aggressive program and Type III with a more aggressive immobilization program. The decision for initial traction is optional and is based on each individual case; if elected, this treatment is followed by a halo vest for 8–12 weeks. The 8 weeks of halo vest treatment is followed by an additional protection in a Philadelphia collar or a SOMI orthosis for 3–4 weeks. Although Type II fractures may be treated with aggressive *nonoperative* treatment, posterior fusion is recommended. Certainly, with Type II and Type III fractures, if two or more of the conditions listed in Table 4-1 apply, then surgical treatment is a strong consideration. Direct anterior screw fixation may be included in these considerations.

Os Odontoideum

This entity, which is difficult to explain on an embryologic basis, is presented here because it is considered to be traumatic in origin.[85,86,123] Most cases probably are; however, it is possible that some may be developmental in origin.[231]

Mechanism of Injury

Most probably, those of traumatic origin are the result of the same MOIs proposed for frank dens fractures, which were discussed in the preceding section.

Discussion

Spierings and Braakman[240] reported on 37 patients with os odontoideum. This work appropriately addressed the measurement of the distance between the posterior border of the body of C2 and the front of the posterior arch of the atlas. This distance is measured in flexion and should not be less than 13 mm (see Fig. 5-16).

Suggested Treatment

Treatment should be nonoperative symptomatic management except in the following circumstances: patients who show a distance less than 13 mm, as described above, and have any combination of intractable pain, neurologic signs, or significant neurologic symptoms. Patients with progressive neurologic deficits should have C1–C2 posterior fusions. Positioning and control during and after surgery are crucial considerations.

Traumatic Atlanto-Axial (C1–C2) Dislocations and Subluxations

This is a group of conditions in which trauma results in an abnormal relationship between the ring of C1 and the ring of C2. There may be either an anterior or posterior abnormal displacement of C1 on C2, or there may be a rotary subluxation of C1 on C2. These are discussed in detail in Chapter 5.

Mechanism of Injury

Obviously, the mechanism of injury in the three situations is not the same. For the anterior dislocation, the MIV is primarily along the +z-axis, causing an abnormal anterior translation and the dislocation. This is associated with a significant amount of flexion ($+\theta x$ rotation). The equilibrium between the magnitude of the MIV and the "strength" of the bone is such that in addition to a fractured dens there is also enough force to cause a displacement. The orientation of the MIV is similar to that of the undisplaced dens fractures and in addition consists of a force of sufficient magnitude to cause displacement as well as fracture.

When the fracture dislocation is posterior, the same analysis applies, except for a change in the direction of the MIV. The resultant translation and rotations are in the opposite directions. Two very unusual posterior dislocations of C1 on C2 have been described, in which the dens is spared and the ring of C1 is carried up over the tip of the dens.[109,195] An example is shown in Figure 4-20. In both of these case reports there was no significant fracture of the dens. There was a small fragment of bone anterior to the second cervical vertebra in one of the cases. The MIV in these injuries probably had a significant +y-axis component in order to raise the ring of C1 up over the dens. Thus, the MIV is presumed to have been primarily in the sagittal plane, acting at the *origin* of the coordinates and directed 30–40° between the −z-axis and the positive y-axis (Fig. 4-20).

The rotary subluxation of this joint presumably occurs when the force vector is not directly along the z-axis but is off center enough to create a torque about the y-axis and cause rotary dislocation or subluxation.

Discussion

In considering the clinical biomechanics of the atlanto-axial complex, what are the normal relationships at the ring of C1, and how are they maintained? What kinds of forces and structural failures are necessary to disrupt this relationship? What is necessary to reestablish and maintain the normal relationship?

A basic understanding of the functional anatomy at the ring of C1 is crucial to any study of atlanto-axial pathology. Three anatomic perspectives of the ring of C1 are shown in Figure 4-21.

The normal relations at the ring of C1 are maintained by a complex of the normal intact structures. The spatial relationship of the cord and the dens at the ring of C1 has been popularized as Steele's rule of thirds.[244] The anteroposterior diameter of the ring of C1 is approximately 3 cm. This space is allocated such that there is about 1 cm for the dens, 1 cm for the cord, and 1 cm of free space. In a detailed study by Wolf and colleagues, the spinal canal at the level of the atlas in 200 subjects was found to have a sagittal diameter ranging from 16 mm to 33 mm.[294] Carella[37] carried out measurements in this area and found the sagittal diameter of the ring of the atlas to be consistently greater than that of the axis, sometimes by only a few millimeters. However, more significant perhaps was the observation that in patients who experienced hypoplasia or failure of fusion of the components of the ring of the atlas, there was a significant reduction in the sagittal diameter.[226] This should be kept in mind when making clinical judgments concerning patients who have congenital irregularities of the atlas. The relative allocation of space into three equal parts presumably remains the same throughout that range of variation of sagittal diameter. However, proper radiologic studies to evaluate the absolute and relative magnitudes of the dens, the spinal medulla, and the free space can be employed if required.

A review of some of the basic anatomic considerations may be helpful. Detailed discussions of the functional anatomy of this region are available and can be reviewed in the work of Martel and Werne.[154,279] The important relationship of the three anatomic perspectives of the ring of C1 (Fig. 4-21) on

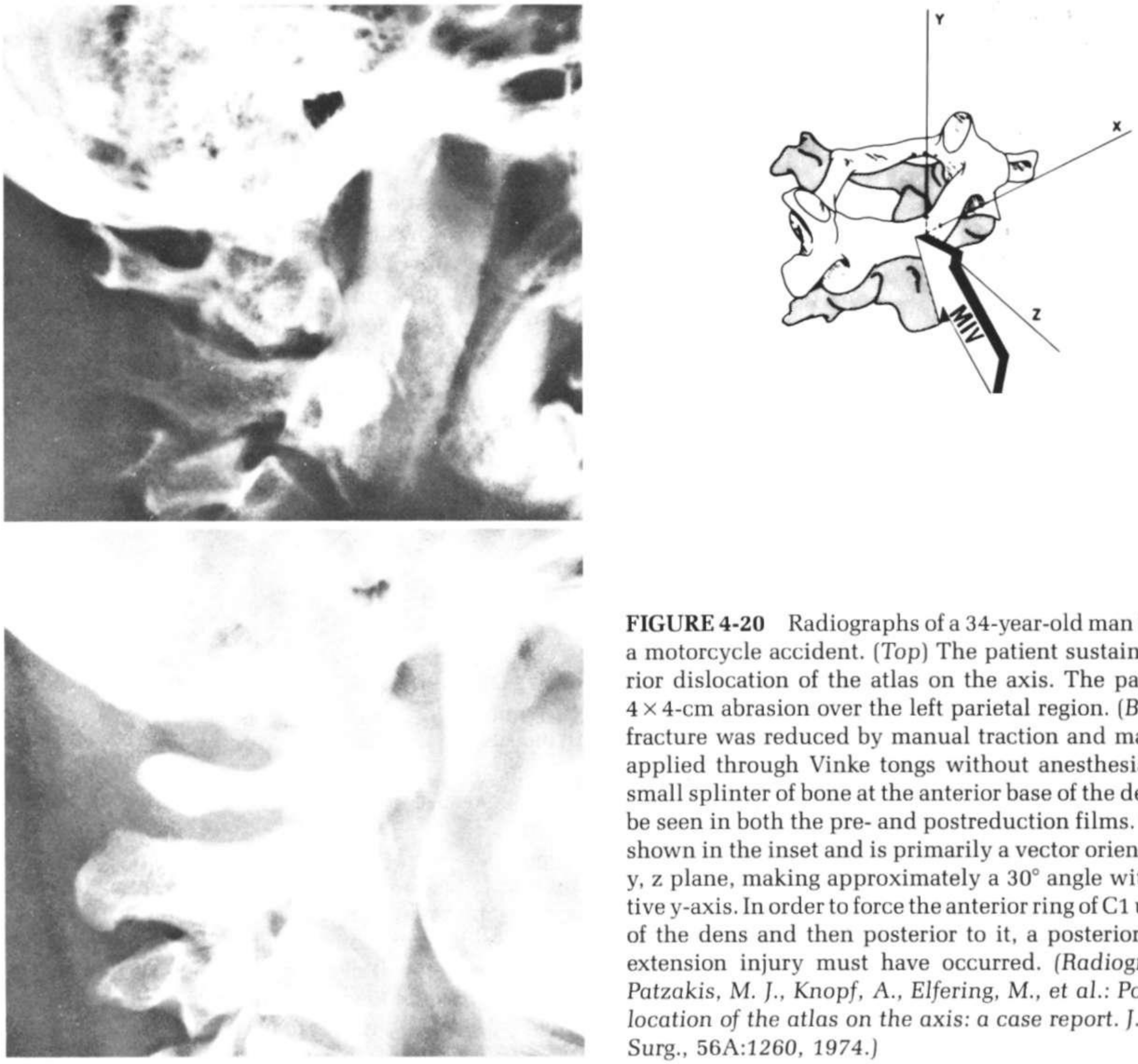


FIGURE 4-20 Radiographs of a 34-year-old man involved in a motorcycle accident. (*Top*) The patient sustained a posterior dislocation of the atlas on the axis. The patient had a 4 × 4-cm abrasion over the left parietal region. (*Bottom*) The fracture was reduced by manual traction and manipulation applied through Vinke tongs without anesthesia. Note the small splinter of bone at the anterior base of the dens that can be seen in both the pre- and postreduction films. The MIV is shown in the inset and is primarily a vector orientated in the y, z plane, making approximately a 30° angle with the positive y-axis. In order to force the anterior ring of C1 up to the tip of the dens and then posterior to it, a posteriorly directed extension injury must have occurred. *(Radiographs from Patzakis, M. J., Knopf, A., Elfering, M., et al.: Posterior dislocation of the atlas on the axis: a case report. J. Bone Joint Surg.,* 56A:*1260, 1974.)*

radiographic analysis has been summarized in the work of Shapiro and colleagues.[230] In the adult, the distance between the anterior arch of the atlas and the dens in flexion or a neutral position should not exceed 2–3 mm. In the child, this distance can be up to 3–4 mm. When these distances are exceeded, there is either rupture or laxity of the supporting ligaments, primarily the transverse ligament and secondarily the alar, apical, and capsular ligaments of the atlanto-axial joint. The other mechanism through which these spatial relationships may be disrupted is of course a displaced fracture of the dens.

It has generally been accepted that significant traumatic displacement occurs only with fracture of the dens. Transverse ligament rupture has been thought to occur only secondary to other disease processes, or it is considered to be a spontaneous rupture.[279] In the past, it was believed that with loading that causes anterior translation (+z-axis), the bone generally fails rather than the ligament, unless there is some active disease process involving the ligament or its bony attachment. However, recent experimental studies have shown that traumatic rupture of the transverse ligament in horizontal loading (+z-axis translation) can occur in the absence of dens fractures (Fig. 4-22).[84] The experimental work is supported by several case reports.

It is possible that traumatic rupture of the transverse ligament in patients with multiple trauma causes sudden death and is not recognized at autopsy; the cause of death is assigned to one or more of the other more readily recognized injuries. This assumption is well supported by the work of Alker and colleagues.[5] They completed postmortem radiographic examinations of 146 traffic fatalities. Twenty-one percent had neck injuries, most of which were at the C0–C1–C2 region.

Another possible mechanism of displacement of the ring of C1 is reported by Patzakis and colleagues.[195] If the odontoid is not fractured, and the transverse ligament is not ruptured, then displacement can occur by raising the ring of the horseshoe and bridle strap up over the stake (tip of the dens). Following this, it either relocates by falling back down anteriorly or dislocates by falling posteriorly. Obviously, this displacement can be resisted or facilitated by the size and/or shape of the dens. Figure 4-23 readily shows that some dentes are more likely to allow the ring of C1 to ride up and over the tip of the odontoid. This may be caused by a force great enough to overcome the tensile forces imposed by the anterior longitudinal ligament. Figure 4-23 further shows that other anatomic characteristics can play a role in determining the type of injuries that result from a particular spatial orientation of a force vector.

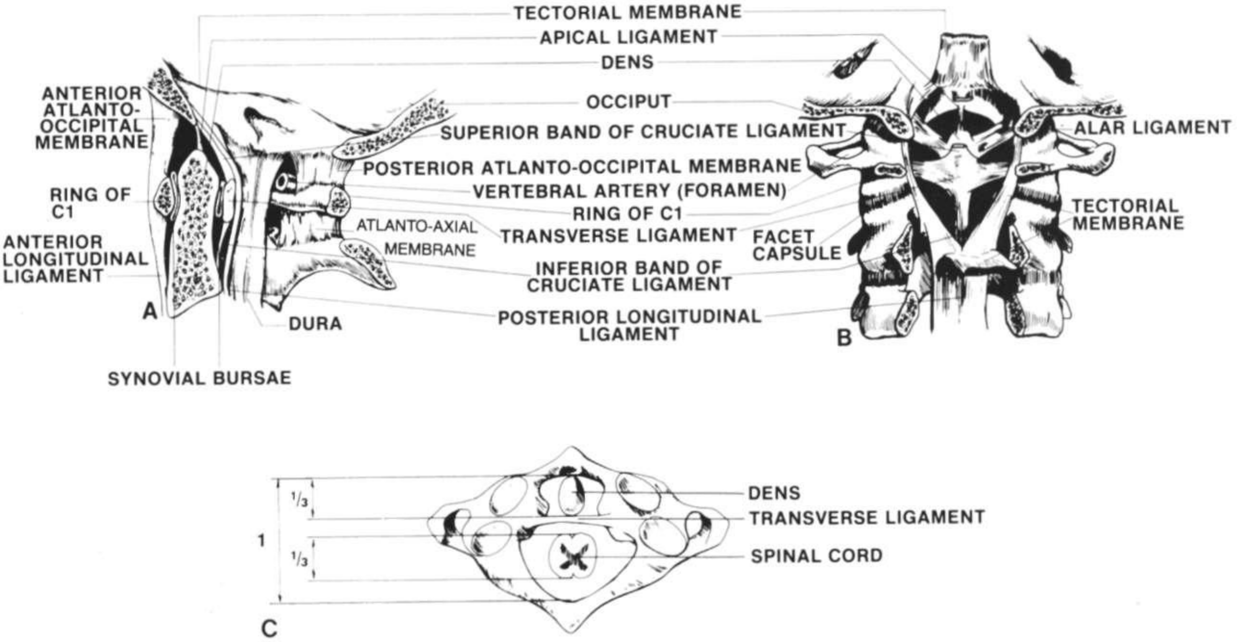


FIGURE 4-21 Three anatomic perspectives at the ring of C1, including the functional anatomy of this region relevant to the biomechanics of spine trauma. **(A)** In the lateral view, note the bursa between the ring of C1 and the dens anteriorly, and the transverse ligament and dens posteriorly. There is also a bursa between the transverse ligament and the tectorial membrane. These bursae contain synovial tissue. They may be involved in abnormal displacements related to inflammatory processes resulting from either erosion of the dens or laxity or weakening of the transverse ligament. The apical ligament is seen in this view. It is one of the accessory ligaments of the occipital-atlanto-axial complex that, in conjunction with the alar ligaments, can offer some resistance to horizontal displacement after rupture of the transverse ligament. **(B)** This view shows the tectorial membrane folded down and up, with the superior portion of the cruciate ligament transected to view the apical and the alar ligaments and the tip of the dens. The posterior longitudinal ligament is a well-developed structure that offers considerable stability to this area. Partial stability is offered by the apical and alar ligaments. **(C)** This view emphasizes Steele's rule of thirds. It shows the upper view of the ring of C1 along with the dens and the spinal cord. One third of the total anteroposterior diameter is constituted by the dens, one third by the spinal cord, and one third by free space.

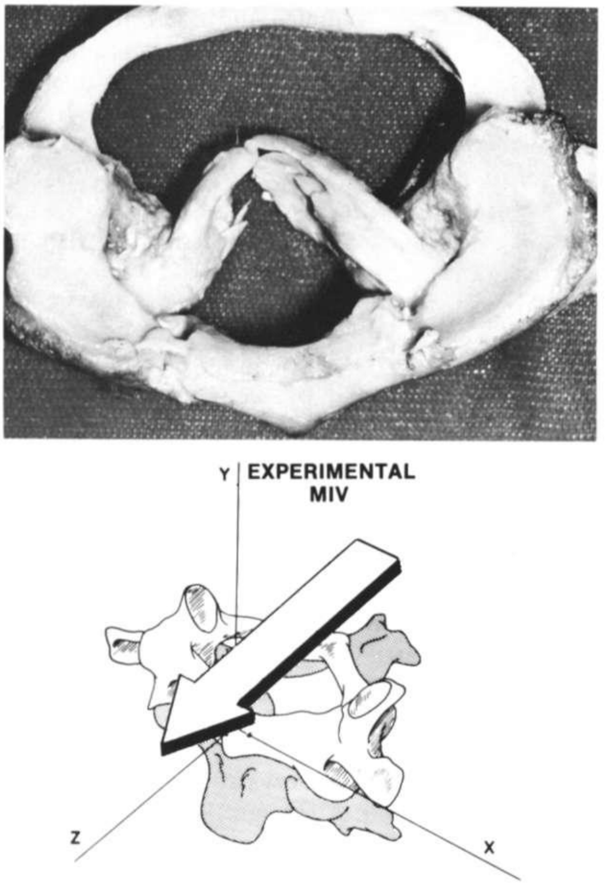


FIGURE 4-22 A photograph of experimental rupture of the transverse ligament. In cadaver studies this pattern of failure was observed in the absence of fracture. The experimental MIV is shown. This shows that it is mechanically possible to have transverse ligament failure and a C1–C2 dislocation without a fractured dens. *(Fielding, J. W., Cochran, G. V. B., Lawsing, J. F., and Hohl, M.: Tears of the transverse ligament of the atlas, a clinical biomechanical study. J. Bone Joint Surg., 56A:1683, 1974.)*

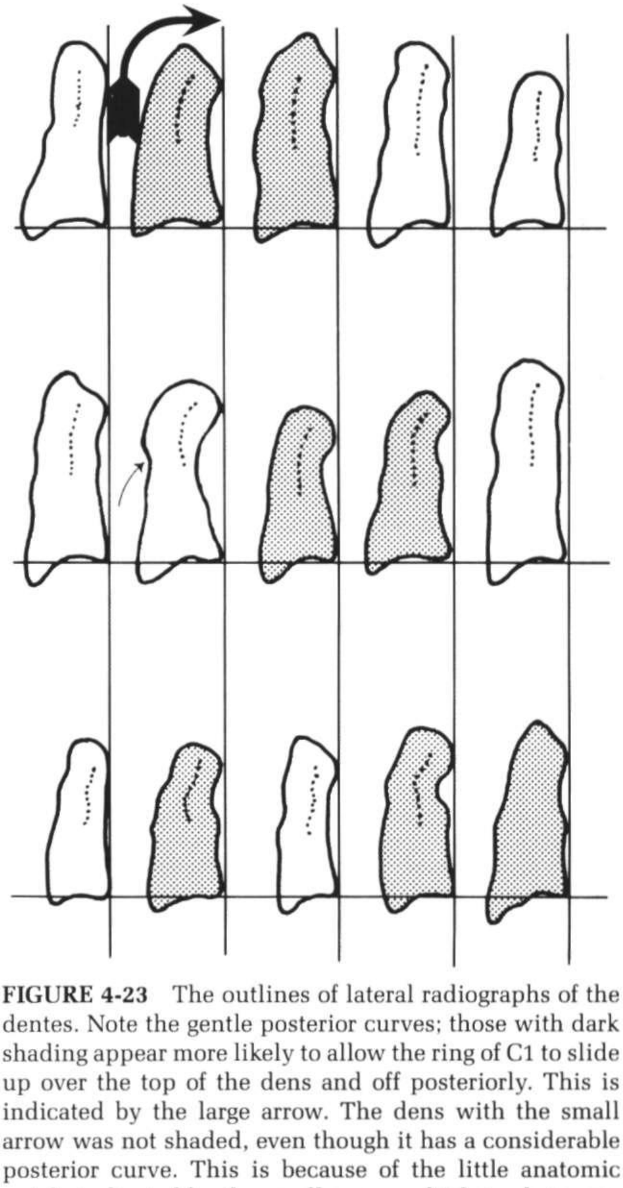


FIGURE 4-23 The outlines of lateral radiographs of the dentes. Note the gentle posterior curves; those with dark shading appear more likely to allow the ring of C1 to slide up over the top of the dens and off posteriorly. This is indicated by the large arrow. The dens with the small arrow was not shaded, even though it has a considerable posterior curve. This is because of the little anatomic notch, indicated by the small arrow, which tends to prevent posterior dislocation. *(Modified from Werne, S.: Studies in spontaneous atlas dislocation. Acta Orthop. Scand., 23[Suppl.]:35, 1957.)*

Probably the most common atlanto-axial dislocation occurs with fracture dislocation of the dens. Elliott and Sachs[74] were among the first to focus on the problem in this perspective. They carried out a clinical and anatomic analysis in conjunction with a case presentation. They report a fascinating saga of a Russian carpenter who lived and worked for 32 years with an anterior atlanto-axial fracture dislocation. The patient experienced five subsequent injuries following his original fracture dislocation. On several occasions he recovered from formidable neurologic deficits of paralysis and sensory loss. He finally succumbed to an injury that involved an irreversible neurologic deficit. Examination showed the encroachment of the spinal cord through a "tonsillotome" mechanism (Fig. 4-24). The posterior ring of C1 serves as the fixation fulcrum. Elliott and Sachs observed the relatively large space for the spinal cord at this level and suggested that this was probably the reason that the patient was able to live and work as long as he did.[268]

In discussing the atlanto-axial subluxation, it is

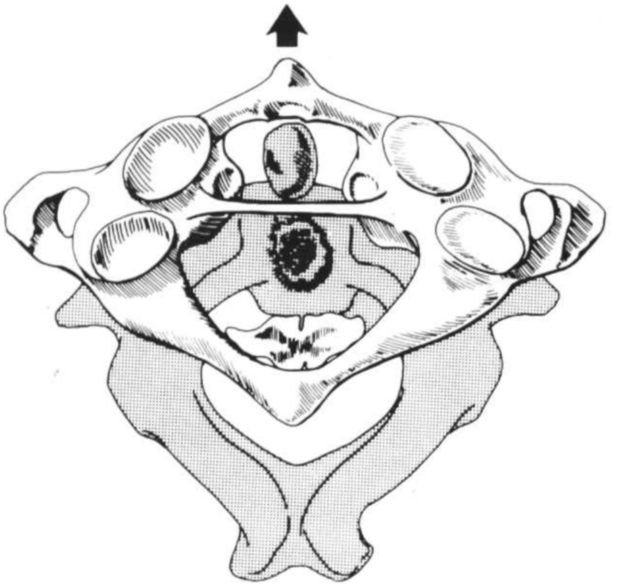


FIGURE 4-24 Guillotine mechanism. This is a top view of the ring of C1 and C2. It demonstrates trauma in the upper portion of the cord through what may be called a "guillotine" mechanism. With this, various anatomic structures allow a displacement of the ring of C1 in relation to the ring of C2 to occur following a grossly displaced dens fracture. A relative horizontal translation of the ring of C1 on C2 applies shear forces of large magnitude to the cord. This is somewhat analogous to the mechanism by which a guillotine operates.

important to mention the clinical significance of occipital pain due to the irritation of the sensory fibers of the occipital nerve that emerges between C1 and C2. This nerve may be irritated as a result of the abnormal displacement between the two vertebrae.

Suggested Treatment

The mechanical goal in the treatment of these patients is to reestablish, as much as possible, the normal anatomic relationships at the level of the ring of C1 and maintain them. The suggested treatment of traumatic C1–C2 subluxations and dislocations is reduction by skeletal traction, followed by fusion of C1 to C2. When there is no neurologic deficit, probably perfect anatomic reduction is not that important, in view of Steele's rule of thirds. Thus, if reduction is not readily achieved, it need not be vigorously pursued. A solid fusion is the most important factor. The most compelling argument for fusion is that it is good insurance. There is evidence that a significant portion of atlanto-axial injuries can result in prolonged neurologic sequelae and death if not stabilized by fusion. Some physicians have suggested nonsurgical treatment,[48] and others have suggested fusion, if osteosynthesis is not achieved with conservative therapy.[91,191,212] Most have advocated some type of fusion.[4,8,52,67,114]

Deciding which levels to fuse and how to fuse them is open for discussion. If the ring of C1 is not fractured and is intact, a posterior fusion of C1–C2 is completely adequate to stabilize the joint. Including either the occiput or C3 further restricts the already compromised motion and is not necessary for stability. When there is a compromise of the ring of C1, the occiput should be fused to C2. The Brooks' posterior fusion works well and has certain biomechanical advantages.[103] These are described in Chapter 8.

Most probably, some of the failures reported with fusion of C1–C2 are related to the choice and the execution of the surgical technique employed. Removal of the ring of C1 is rarely necessary as even a partial reduction, and adequate stabilization is usually sufficient to decompress and protect the cord.

Conservative treatment with prolonged immobilization, using either Minerva casts or fixation, has been recommended. Although the reported results in most cases have been satisfactory, the potential risk of depending on healed ligaments to maintain stability in this vital situation does not seem to be warranted.

"Spontaneous" Atlanto-Axial Dislocations

Abnormal displacement of the ring of C1 occurs either without trauma or with quite trivial trauma. These lesions have been given a long list of names, including torticollis nasopharyngium, malum suboccipitale rheumaticum, spontaneous hyperemic dislocation of the atlas, dislocation nontraumatique, distension/subluxation, maladie de Grisel, and nontraumatic subluxation. In the group of "spontaneous" atlanto-axial dislocations, a broad variety of other conditions are included, in which there may be abnormal displacement with little or no trauma. Figure 4-25 shows a spontaneous anterior subluxation of the atlas secondary to rheumatoid arthritis.

Mechanisms of Injury

There are a variety of different situations that predispose an individual to this injury. They are listed below. The final, common factor is that normal or

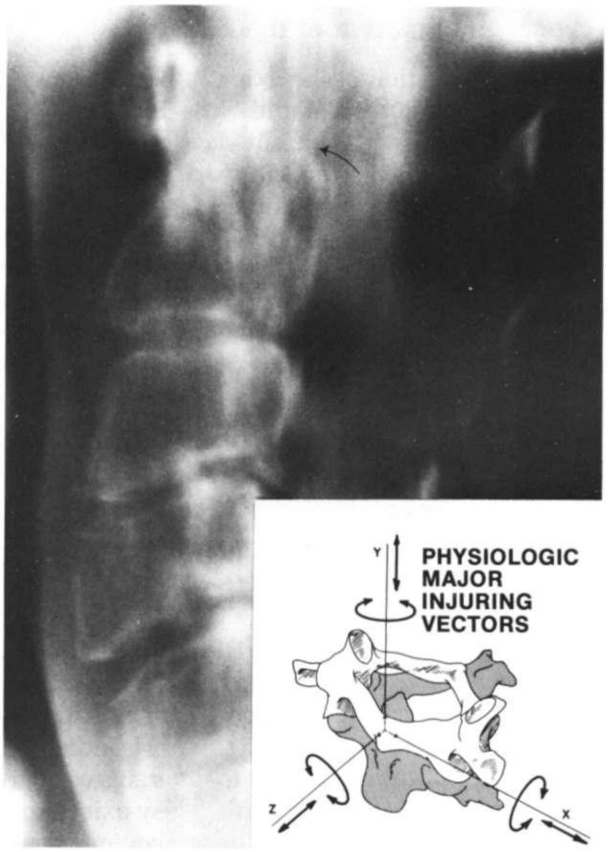


FIGURE 4-25 Laminagram of an elderly man with severe rheumatoid arthritis. There is an abnormal separation between the dens and the anterior ring of C1. The MIV consists of the normal complex of physiologic forces acting on the weakened structures. The transverse ligament is lax; moreover, there has been some erosion of the posterior portion of the dens (note the bursa in Fig. 4-21), which allows for even more of an abnormal anterior (+z-axis) translation. *(Courtesy of Teaching Library, Section of Neuroradiology, Yale University School of Medicine.)*

CONDITIONS ASSOCIATED WITH "SPONTANEOUS" ATLANTO-AXIAL DISLOCATIONS

- Primary synovitis
  - Rheumatoid arthritis
  - Ankylosing spondylitis
  - Psoriatic arthritis
- Regional infections (viral, bacterial, spyrochetal, granulomatous)
  - Nasal
  - Oropharyngeal
  - Cervical
- Primary or metastatic neoplasms
- Congenital anomalies of the dens
- Congenital laxity of the ligaments
  - Down's syndrome
- Poliomyelitis*
- Previous, forgotten and/or undiagnosed traumatic injuries

* In 1934, Coutts described true subluxation of the atlas in a patient with unilateral polio of the stabilizing muscles.[48]

only slightly greater than normal forces acting in the region cannot be tolerated by the anatomic structures, and abnormal displacement occurs. The transverse ligament of the atlas is the most commonly involved "weak link." Presumably, it stretches, ruptures, or pulls out from its osseous attachment, allowing the displacement.

Discussion

An old, unrecognized, forgotten, or unsuccessfully treated injury can cause an apparent "spontaneous" atlanto-axial subluxation. Griswold and Southwick pointed out that a review of the literature showed an overall rate of failure of union for fractured dentes of 20–40%.[103] Although this is not generally considered to be a cause, when the diagnosis of "spontaneous" atlanto-axial subluxation is suggested, the patient should be questioned carefully, and when there is a history of injury, old radiographs should be reviewed.

Parke and associates have identified a previously undescribed system of veins that may play a role in the pathoanatomy and biomechanics of Grisel's syndrome. The pharyngovertebral veins have frequent lymphovenous anastomoses with the posterior pharyngeal region and also connect with the periodontoidal venous plexus and the suboccipital sinuses. In other words, a pharyngitis could cause hyperemia in the region of the dens and the attachment of the transverse ligament. This could result in ligament laxity and the characteristic C1–C2 subluxation of Grisel's syndrome.[193]

Suggested Treatment

The subluxation should be reduced with appropriate traction, and the associated infection should be treated. Stabilization by posterior arthrodesis of C1–C2 is then carried out.

Rotary Subluxation of C1–C2

In this entity there is a fixed abnormal rotation between C1 and C2. C1 is rotated about the y-axis in relation to C2 and is fixed in that position.

Mechanism of Injury

Presumably, the mechanism of this injury is acute axial rotation resulting in disruption of the articular capsules of the joints articulating with the lateral masses. The MIV is a torque about the longitudinal axis. The injury can occur in the absence of fracture and is not generally associated with neurologic deficit.

Discussion

The diagnosis of this lesion can be elusive. Many patients may be seen with the head rotated to one side and tilted away from the side toward which it is rotated. This has been referred to as "a bird with his head cocked, *listening* for a worm." Based on the fact that the bird's eyes are at the sides of its head and are among its most precise sensory organs, and the assumption being that worms make little noise, the authors suggest that maybe the bird is *looking* rather than listening. In any case, a patient with head held in this position may have a C1,C2 rotary subluxation or spasmodic torticollis. The differential diagnosis of this entity constitutes a sizable list, including psychiatric disease.

The key to the diagnosis of this condition is radiologic. Several articles have been written suggesting complex radiologic analysis, with one or two open-mouth views to make the diagnosis.[125, 126, 230, 276] Actually, this is unduly complicated, and the normal anatomic variants make those evaluations difficult to interpret.* There are two radiologic signs that strongly suggest the diagnosis. One is unilateral superimposition of the lateral mass of C1 on one of the articular facets of C2, seen on the open-mouth view. The other is anatomic anterior displacement (greater than 2 mm in an adult, greater than 3 mm in a child)[230] of the ring of C1 in relation to the dens, seen on a lateral film. Both of these findings are demonstrated in Figure 4-26.

Suggested Treatment

Reduction should be attempted with adequate axial traction. The evaluation and management of these injuries are discussed in detail in Chapter 5.

Vertical Subluxation of the Axis

Vertical subluxation of the axis can occur as a secondary process in rheumatoid arthritis. Figure 4-27 is an example of vertical subluxation with rheumatoid arthritis. This y-axis translation is due to the repeated gravitational loading on the lateral articular masses of C1 and C2. These weakened structures allow for the vertical cephalad (+y) displacement of the dens into the foramen magnum. Displacement can occur as a result of other diseases, including tuberculosis, Paget's disease,[175] and osteogenesis imperfecta.[205] It should be noted that there is no consistent correlation between neurologic status and radiographic evidence of destruction and displacement.[248] Nevertheless, guidelines obtained from radiographic measurements are useful in the clinical evaluation (see Fig. 5-8).

Suggested Treatment

Treatment of these patients is highly individualized. When neurologic symptoms cannot be arrested or treated by conservative measures, such as rest, traction, and orthotic support, then surgery is indicated. Fusion of the posterior occiput to C3 is suggested. A construct designed to resist vertical (−y-axis) forces should be employed. For postfusion fixation, a halo apparatus is best able to protect the neural elements while the graft matures.

Traumatic Spondylolisthesis of the Axis, or "Hangman's Fracture"

> You have to hang Mr. A. He is 5 ft., 10½ inches in height, and weighs 12 st., 2 lbs., 6 oz., 1 dwt. His neck from the sternocleidomastoid to the sternohyoid measures 6¾ in. The neck is strong and 17 in. in diameter. Calculate to three places of decimals the drop necessary to hang this man thoroughly, without risk of giving pain to on-lookers. Also give the diameter and quality of the rope you would employ, in terms of pounds "avoirdupois of strain."
>
> CHARLES DUFF: A HANDBOOK ON HANGING
> LONDON, 1938

There is a very important biomechanical consideration related to the phenomenon of judicial hanging. A considerable amount of information has been generated that relates radiographs and autopsy findings to a known mechanism of injury in a living human being. When the clinician observes similar failure patterns in different patients, a similar mechanism can be presumed to have been operative.

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* Personal communication, R. Shapiro.

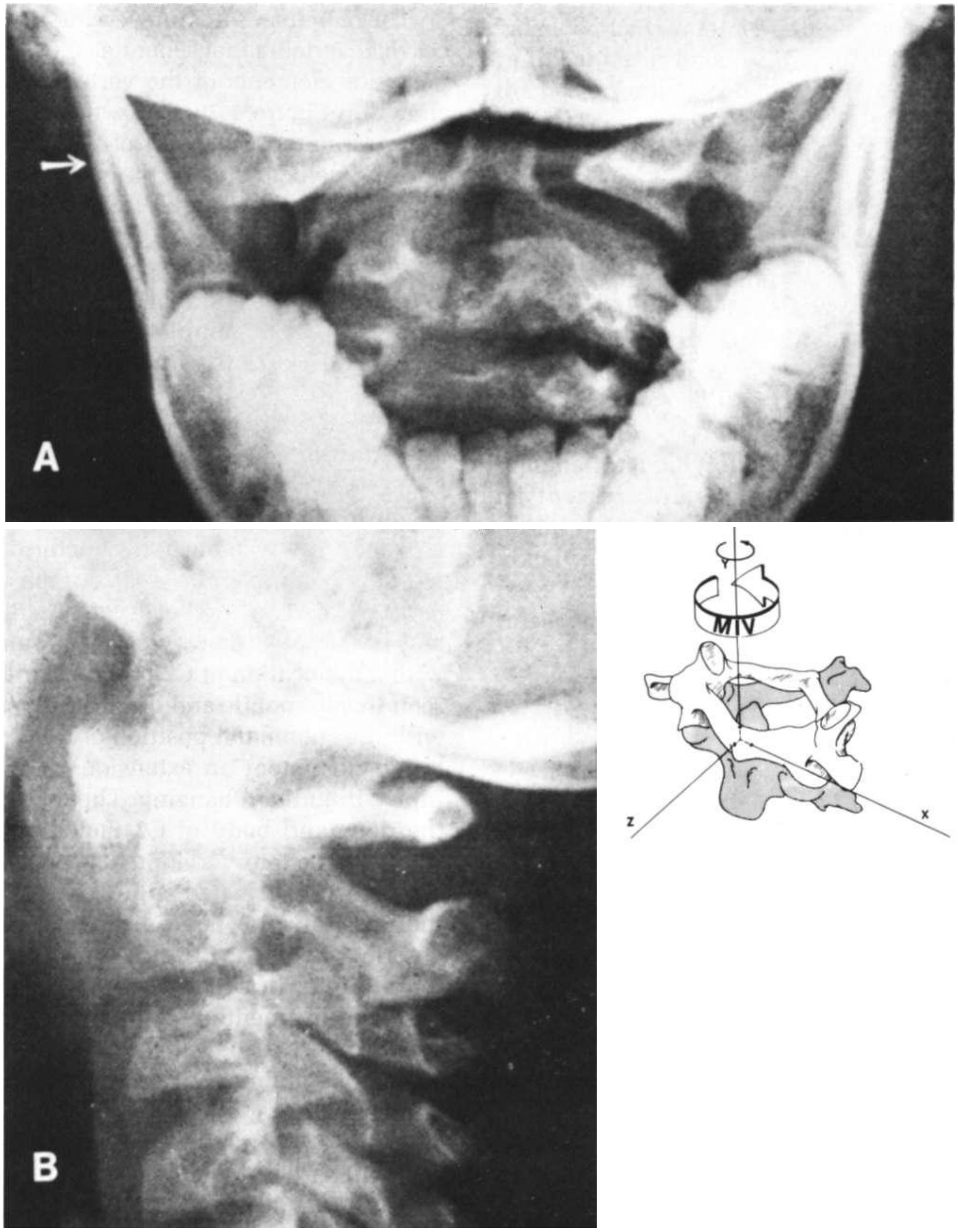


FIGURE 4-26 A rotary subluxation of C1 on C2. **(A)** The arrow indicates an overlap of the lateral mass of C1 in relation to the articular facet of C2. This overlap occurs as a result of lateral as well as anterior displacement. **(B)** On the lateral view, one can see an abnormal displacement between the anterior portion of the dens and the posterior portion of the anterior ring of C1. The MIV for this type of injury is primarily a moment about the y-axis, as shown in the inset. *(Radiographs from Garber, J. N.: Abnormalities of the atlas and the axis: vertebral, congenital and traumatic. J. Bone Joint Surg., 46A:1782, 1964.)*

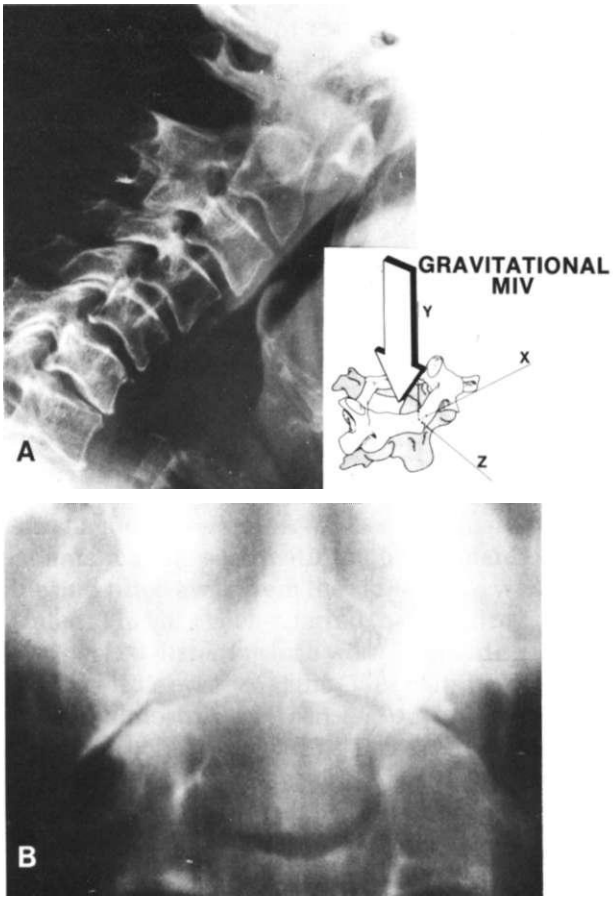


FIGURE 4-27 Radiographs of an elderly male with severe rheumatoid arthritis. **(A)** Vertical subluxation of the axis. This is best described as a vertical ($-$y-axis) subluxation of the occiput. The dens is at the level of the occipital bone and just at the mouth of the foramen magnum. There is anterior subluxation of C1 on C2 and spontaneous fusion of the lateral masses of C2 and C3. **(B)** Lateral displacement of the dens and asymmetrical erosion of the lateral mass of C1. The vertical subluxation is largely due to the inadequate supporting structure of the lateral masses. The MIV is primarily the vertical component of the gravitational forces of the head acting over a long period of time on the weakened bone. *(Courtesy of Teaching Library, Section of Neuroradiology, Yale University School of Medicine.)*

The hangman's fracture is a fracture of the second cervical vertebra that separates the anterior from the posterior elements of the vertebra. An example is shown in Figure 4-28. Thus, the fracture occurs in the most anterior portion of the lateral masses, or into the pedicle area of the vertebra. This fracture may be associated with fractures of other spinous processes or fractures involving the vertebral body of C3. There may be no associated neurologic findings, or there may be symptoms ranging from nerve root irritation to complete flaccid paralysis. There are several works that thoroughly describe this injury.[29,75,153,224,291,295]

Mechanism of Injury

In judicial hanging with the submental knot, a number of observers have confirmed the lesion commonly known as hangman's fracture. In association with the fracture of the area between the two articular facets of C2, there are several other injuries. There may be complete disruption of the annulus fibrosus, with a dislocation of C2 on C3. Similar injuries are seen in automobile and diving accidents. Certainly, with the submental position of the rope, it is possible to document an extension–distraction type of injury in judicial hanging. The moment exerted on the dens and body of C2 may also create tensile forces on the intervertebral disc and may sometimes result in an associated failure of that structure, with the possibility of large displacement between the anterior elements. Cornish observed disruption of the annulus in specimens that he studied.[47] The posterior ligamentous structures are thought to remain intact as they are compressed. Figure 4-29 gives a diagrammatic representation of the hangman's fracture.

There are some interesting anatomic considerations that are relevant here. The transverse foramen for the vertebral artery is in the region of the pedicle (isthmus) of C2.[47] Because of this foramen and the configuration of the neural arch, the structure of C2 in this region has a relatively low area moment of inertia to bending in the sagittal plane. This may be a factor in determining the site of failure. In addition, there is another structural consideration. The occipital-atlanto-axial complex has other characteristics that indicate failure at the pedicle (isthmus). The large extension force creates a bending moment on the dens so that it rotates in the sagittal plane about the $-$x-axis (Fig. 4-30). This bending moment is balanced by two forces—the tensile force produced in the anterior longitudinal ligament, the disc, and the posterior longitudinal ligament on one side, and a compressive joint reaction force between the facet joints of C2 and C3 on the other side. These two equal and opposite forces create the balancing bending moment. The effect of all of these loads is the production of maximum bending moment in the region of the pars interarticularis. Because the cross-section of the bone is small, this site is the weakest and thus the most susceptible to fracture during the type of load that results in a hangman's fracture.[E]

Discussion

Traumatic spondylolisthesis is an appropriate name for this fracture, because the defect occurs in the posterior elements as a result of trauma. The emotionally charged appellation of "hangman's fractures" will no doubt continue, despite the fact that this form of execution is not very common, and it is

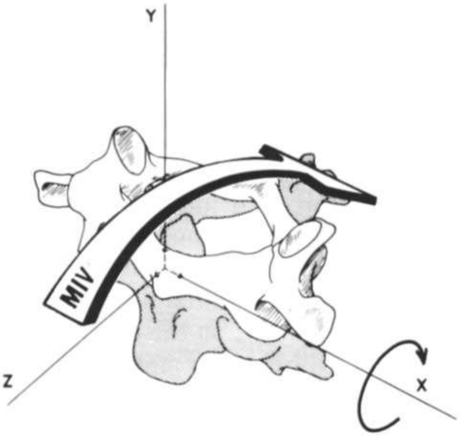


FIGURE 4-28 Radiograph of an adult male who sustained in a motorcycle accident a fracture of the parietal bone and a hyperextension injury that resulted in a traumatic spondylolisthesis of C2. The separation of the posterior elements from the anterior elements is well demonstrated in this particular lateral view. The MIV is shown. A major factor is a negative torque about the x-axis. The vector creating the torque may be oriented so that it points more in the + y-axis, as in judicial hanging, or it may tend to point more in the direction of the − z-axis, as in the case of an automobile accident. *(Courtesy of Teaching Library, Department of Diagnostic Radiology, Yale University School of Medicine.)*

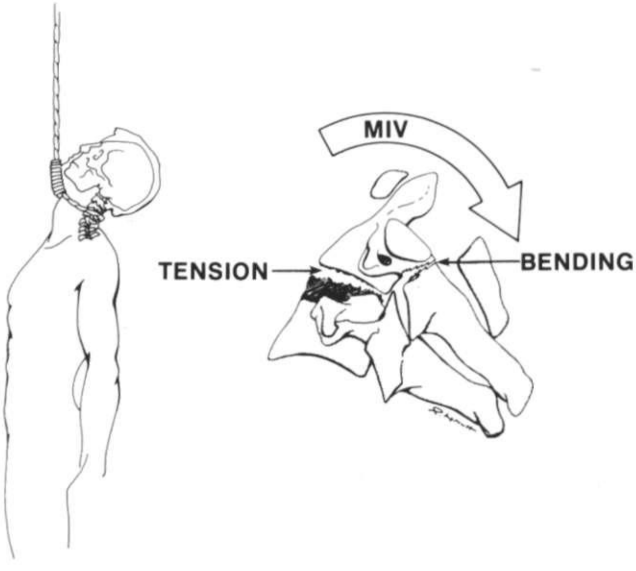


FIGURE 4-29 The judicial hangman's fracture. With the submental knot there is a negative torque created about the x-axis. There is failure in the posterior osseous elements and separation anteriorly at the annulus fibrosus. The pattern of this injury varies with the relative magnitude of the component of the vector in the + y direction as well as the − z direction.

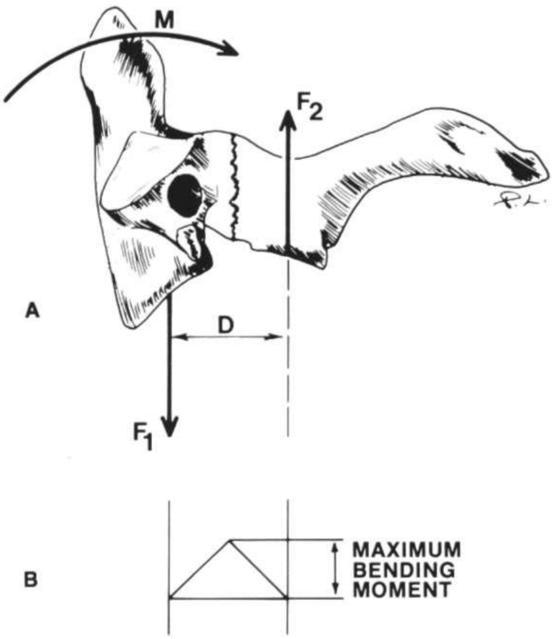


FIGURE 4-30 **(A)** Vertebra C2 is acted upon by the bending moment M, applied by the anterior ring of C1, and a couple of equal, parallel, and opposite forces, F_1 and F_2, applied by C3. The bending moment and the couple balance each other. **(B)** The bending moment distribution along the pedicles is represented by the triangular bending moment diagram, which shows that in the middle of the pedicle there is maximum bending moment.

really the "hangee" rather than the "hanger" who owns the fracture.

The use of this name and the presumably similar mechanism in auto accidents and other injuries have been questioned by some physicians. They point out that many individuals who sustain this injury do not have even transient neurologic symptoms. The difference is explained by the magnitude and *direction* of the MIV and the duration of load application in the judicial victim, who must "hang by the neck until dead." With the prolonged duration of the load, viscoelastic instability becomes operative, and the critical failure load of the soft tissues is reached, causing separation of vertebrae and neurologic death. Should this somehow fail to happen, strangulation ensures demise.

These injuries, when secondary to diving or auto accidents, are often without neurologic symptoms. Perhaps this occurs because spondylolisthesis creates a loose neural arch that can further accommodate the cord in an area where it normally has more than ample space.[196]

Actually, there is a large family of injuries that have in common a fracture in the region of the pedicle (isthmus) of C2. The complex of injuries that occurs depends on the specific force vectors involved, the magnitude, direction, point of application, and duration of application. In addition, the position of the structures of the spine at the time of impact and the individual mechanical properties of the structures in that particular patient all determine the particular injury, the elements destroyed, and the amount of displacement. When one observes a traumatic spondylolisthesis of C2, $-\theta x$ bending moment is the major component of the injuring forces. Therefore, when this type of fracture is found, we know from judicial hanging that the most likely mechanism of injury is extension.

In auto accidents, where the extension may be due to the forehead hitting the steering wheel or a slanted windshield, a vertical axial component of compression may be involved, along with the rotary bending moment (Fig. 4-31). Rogers has noted considerable crushing of the third cervical vertebra in addition to the C2 fracture.[214] He also noted other injuries that did not fit neatly with a simple extension mechanism. One of his patients had facet fractures between C7 and T1, which strongly suggests a compressive force. In judicial hanging there is a bending moment creating extension with *tensile* forces on the cervical spine, whereas in the auto accident there is a similar bending moment but with *compressive* forces on the cervical spine.[291] A clinical study by Bucholz of motor vehicle accident victims supports the concept that these are mainly hyperextension injuries that are sometimes accompanied by a vertical compression component.[34]

Suggested Treatment

It is obvious that not all hangman's fractures should be treated in the same manner. The treatment depends on what, if any, associated injuries are present. Although the posterior ligaments are generally intact, they make no contribution to stability because of the defect of the pedicle. Clinical experience has shown that the defect usually heals with effective protection and immobilization.[29]

Some surgeons vigorously advise against traction.[47] This is because of the large distractions that may occur (Figs. 4-31, 4-32). Such distractions document that the annulus fibrosus and other anterior elements have been disrupted and that the situation

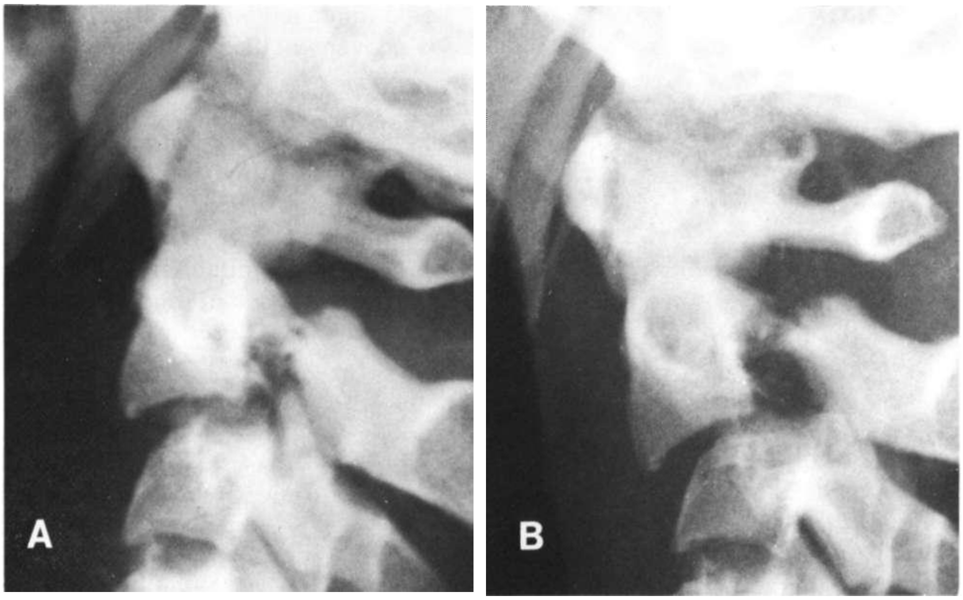


FIGURE 4-31 Radiographs of a young male who sustained a traumatic spondylolisthesis in an automobile accident. **(A)** Fracture dislocation without traction. **(B)** The instability of the lesion and the effect of 3 lb of axial traction. In this situation the annular and longitudinal ligament fibers between C2 and C3 were obviously destroyed in the injury. It is assumed that the MIV has a significant +y-axis component. In addition to the major y- and −x-axis torque, there is a tensile or shear load applied to the annulus. This does not always occur in such an injury. Note that the posterior elements of C2 had been pulled anteriorly, which was, no doubt, injurious to the cord. This patient died soon after admission to the hospital. *(Courtesy of Teaching Library, Section of Neuroradiology, Yale University School of Medicine.)*

is unstable. Carefully monitored axial traction with close observation may be used to safely elicit this information[34] (see p. 318). Traction of 30–40 N (7–9 lbf) may be used prior to surgery or orthotic immobilization to improve reduction, relieve muscle spasm, and rest the tissues. Several recent publications have offered refinements in the classification and treatment of these injuries.[34,71,89,145] The authors' attempt to provide some useful guidelines for the evaluation and treatment of traumatic spondylolisthesis of the axis follows. We suggest that there are four key considerations in determining the management of these injuries:

1. Stability
2. Neurologic status
3. Facet joint status
4. Positioning (alignment and distraction)

Each of these *four keys* is discussed and a classification is suggested.

Stability In this fracture, because the posterior elements are separated by the traumatic spondylolysis, stability depends on the integrity of the anterior elements. If they are grossly or completely disrupted, there will be anteroposterior or z-axis translation and abnormal sagittal plane or x-axis rotation, as seen in Figure 4-31. There may also be abnormal distraction or y-axis translation in addition to abnormal sagittal plane or x-axis rotation. This is shown in Figure 4-32, in which there is total disruption of the anterior elements. There may be less dramatic instability. This may be evaluated as suggested in Chapter 5 (p. 314) and by Francis and co-workers.[89]

If there is no neurologic problem and the injury is *stable* (anterior elements intact or almost intact), then the patient may be treated by an intermediate-range cervical orthosis.

If the patient is *unstable* with no neurologic deficit, then there should be traction for alignment if needed. This is then followed by 10–12 weeks of

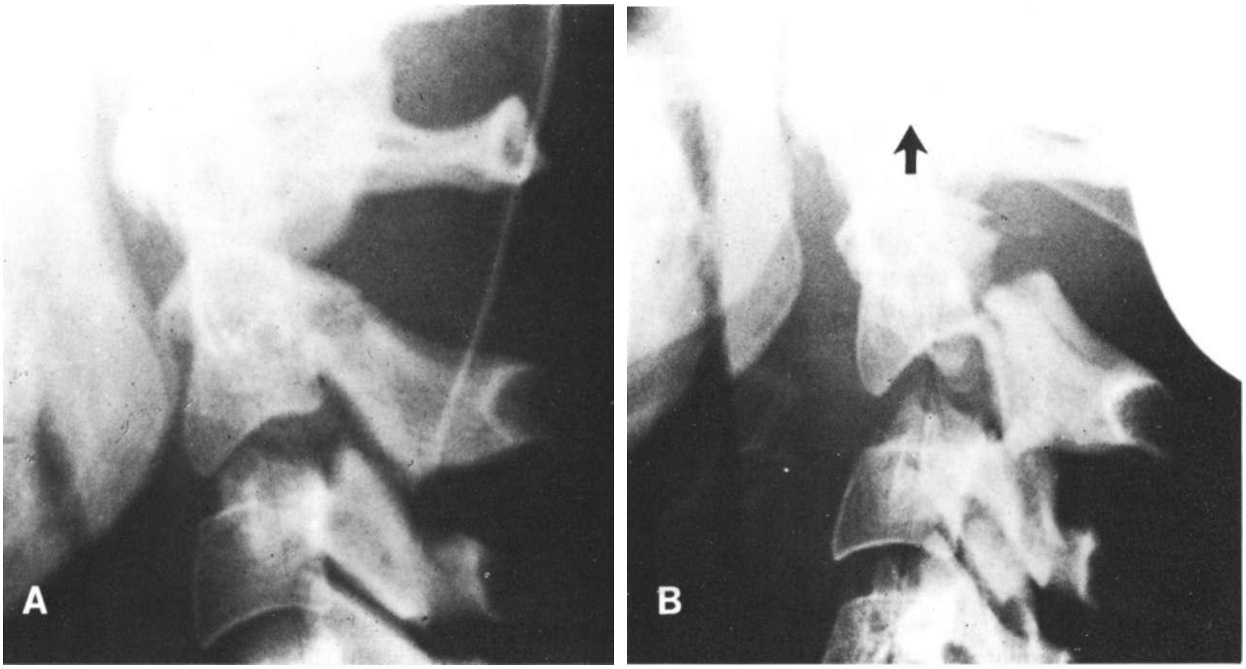


FIGURE 4-32 Radiographs of an adult in an automobile accident. **(A)** The patient in a resting position. **(B)** The patient after the application of 8 kg of axial traction. The separation of the anterior elements is dramatically demonstrated. Note that there is little change in the relationships of the posterior elements. The implication is that the ligamentous structures are intact posteriorly. They were not ruptured by tensile loading at the time of injury, as were the anterior elements. This fits with the presumed mechanism of injury; the posterior elements were under compression at the time of failure. *(Courtesy of Dr. Mason Hohl, Beverly Hills, Calif.)*

treatment in a halo vest. The halo position is manipulated to maintain good position and alignment, as suggested by Levine and Edwards.[145]

When chronic instability is associated with severe pain and/or neurologic deficit, anterior C2–C3 fusion may be required.[89, 282]

Neurologic Status These fractures are usually not associated with a neurologic deficit.[145] Francis and co-workers reported neurologic problems in only 6.5% of 123 patients. If there is a neurologic deficit, appropriate imaging studies, such as computerized tomography and/or magnetic resonance imaging (MRI), should be done to identify the lesion and its position.

If there is evidence of a space-occupying lesion within the canal, it should be removed surgically, and an appropriate reconstructive procedure should be followed. If there is no evidence of material in the canal, or about to go into the canal, the vertebrae should be carefully aligned with traction. If, perhaps, there is a disc fragment that could go into the canal with realignment, it should be removed surgically prior to reduction. Care is taken not to overdistract the spine with either traction or the halo device.

Facet Joint Status An unusual occurrence associated with traumatic spondylolisthesis of the axis is bilateral or unilateral facet dislocation.[71, 145] This may cause difficulty in terms of (1) interfering with reduction and correct alignment, (2) contributing to neurologic deficit through canal compromise, or (3) chronic pain. Closed reduction may be difficult, particularly if there is anterior instability and if traction causes distraction. Also, axial traction to reduce the dislocation must pull through the relaxed, relatively delicate atlanto-axial membrane, as compared with the more hardy and elastic yellow ligament. Open reduction and some internal fixation may be required for these injuries.

Positioning When all or most of the anterior elements are out, there may be significant malalignment and/or distraction between the bodies of C2 and C3. When this is excessive, it may cause neurologic irritation or prevent reduction of the fracture. This can be problematic, both in bed traction and in halo pelvic or halo jacket traction. It is important to look for this and correct it by altering whatever traction is being used.[145]

Classification and Treatment Even though there are already several classifications in the literature, we elected to present the one in the accompanying display because it is thought to be simple and useful.

Some authors consider analysis of the MOI important in determining treatment.[71,145] With this injury, positioning of the bodies of C2 and C3 and reduction of the fracture are desirable ends. However, assumptions about the MOI (short of judicial hanging) are highly speculative, and repositioning is desirable regardless of how the malpositioning came to be. Therefore, the MOI has not been considered in the classification and treatment system proposed here.

The use of seat belts with shoulder harnesses would greatly reduce this injury in auto accidents. Also, staying on the right side of the road and the law is of considerable help.

Cervical Compression Fractures

This group includes several fractures. They all have to do with vertebral body failure and include simple compression fractures, vertical compression fractures, and comminuted or "tear drop" fracture dislocations.

Mechanism of Injury

These lesions are thought to be flexion injuries.[286] A major component of the force vector is exerted along the y-axis in the negative direction and primarily in the region of the anterior elements. The type of fracture that results is a function of the magnitude, the spatial orientation, and the location with respect to the spine of the force vector on one side, versus the physical properties of the various anatomic structures of the vertebra on the other side.

A simple compression fracture with minimal deformation suggests a lower magnitude of force directed toward the midline and axially onto the vertebral body. A vertical compression fracture with central depression probably has a similar mecha-

OUTLINE FOR MANAGEMENT OF TRAUMATIC SPONDYLOLISTHESIS

Type	Characteristics	Suggested Treatment
I	Stable; no neurological deficit	Cervical collar intermediate 8–10 wks
II	Unstable; no neurological deficit	
	A. Good positioning	Halo vest or halo pelvic 12 wks
	B. Poor positioning	Traction Halo vest or halo pelvic 12 wks
	C. Facet joint(s) dislocated	Closed reduction, if successful; halo fixation (vest or pelvic), if failed; open reduction, internal fixation; halo vest or halo pelvic, 12 wks
III	Unstable; with neurological deficit	Imaging evidence of material in canal: *surgical removal,** then treat as Type IIA–IIB or IIC

* This is based on recent work by Arena and Eismont,[10] which showed that in some fracture dislocations of the cervical spine with cervical facet subluxation and dislocation, free fragments of disc material inside or just anterior to the canal found their way into the canal upon reduction of the dislocation or traumatic deformity.

nism, with a force of greater magnitude. In addition, the annulus fibrosus acts as a wedge, which is driven through the end-plate into the vertebral body.

Finally, there is the comminuted cervical vertebral body fracture, often referred to as the "tear drop" fracture dislocation. Presumably, this is due to a vertical force vector of high magnitude, which causes an explosive failure with a variable amount of cord compression. There is usually significant cord damage, which may be caused by posterior vertebral fragments being driven into the spinal canal (Fig. 4-2).[92, 261]

Discussion

The simple compression fracture is generally straightforward but can sometimes be difficult to distinguish from the normal slight wedging that appears in lateral radiographs of the cervical spine in young individuals. Usually, significantly more or less relative wedging of one particular vertebra, along with the pattern of the entire clinical picture, leads to a correct diagnosis. The vertical compression fracture without neurologic deficit is not significantly different from the simple compression fracture.

The "tear drop" fracture is of interest for several reasons. First of all, consider the appellatory semantics. Most would agree that the term is catchy, poetic, and popular. The accord probably ends here. There are different theories explaining how the name was chosen and to precisely what it refers. Is it the radiograph, the mechanism, the clinical picture, or the prognosis? Rand and Crandall elected to classify "tear drop" fractures as hyperextension injuries, based on the triangular shape of the fragment of bone often seen at the anteroinferior border of the involved vertebra.[203] Penning suggests that the name comes from the shape of the anteroinferior wedge of bone that is sometimes seen in these injuries.[197] It has also been proposed that the name was given by Schneider because of the sadness resulting from the associated neurologic damage. Perhaps a quote from the article by Schneider and Kahn may be helpful:[223]

> This lesion is characterized by crushing of one vertebral body by the vertebral body superior to it in such a manner that the anterior part of the involved centrum is not only compressed, but often is completely broken away from its major portion. In most of these cases, this fragment has resembled a drop of water dripping from the vertebral body and it has been associated with dire circumstances so frequently that the terms "tear drop" and "acute flexion" fracture dislocation of the cervical spine seemed to describe the lesion and to suggest the mechanism of injury.

This particular group of fractures is of interest in yet another aspect (see Fig. 4-33). The picture of the residual deformation of an injury is seen on the initial radiograph. In other words, a given functional spinal unit or fragment may displace a certain amount at the time of injury, but on the initial radiograph, only the residual displacement is seen, which may be a good deal less than the initial displacement. There may be no residual displacement, leaving a normal alignment and causing perplexity concerning the presence of neurologic deficit. There is sometimes a residual subluxation at the level involved or an obvious residual fragment of vertebral body remaining in the region of the spinal canal. In some instances, however, plain radiographs show little or no infringement on the canal by a posterior fragment. The fragment presumably has been repositioned as a result of the restraining forces around the explosion, which push the fragments back into the region from which they came. This type of deformation of the vertebra at right angles to the MIV is an example of Poisson's effect. The original volume of space occupied by the vertebral body may now be relatively less as a result of impaction from the compressive forces. Figure 4-33 is offered as a diagrammatic explanation of these ideas.

A clinical example that demonstrates several of the factors that have been discussed is provided in Figure 4-34. This injury was sustained by a college football player following a head-on tackle. The patient was rendered quadriplegic at the level of C5. The lateral laminagram (Fig. 4-34B) shows very minimal posterior displacement of the body of C5. This is a small residual displacement. However, Figure 4-34A shows a complete block at the C5–C6 level, and the clinical picture was that of major spinal cord injury. Note the increased density, the decreased height, and the vertical fracture line involving the body of C5. These are all findings related to the compression injury, with the MIV oriented in the sagittal plane in the direction of the $-y$-axis.

Another clinical example is provided in Figure 4-35A, which reiterates these points, but more important, it demonstrates the more classic "tear drop" configuration of fragmentation. This is seen best at the anteroinferior portion of C4. There is compression of the vertebral body and minimal posterior

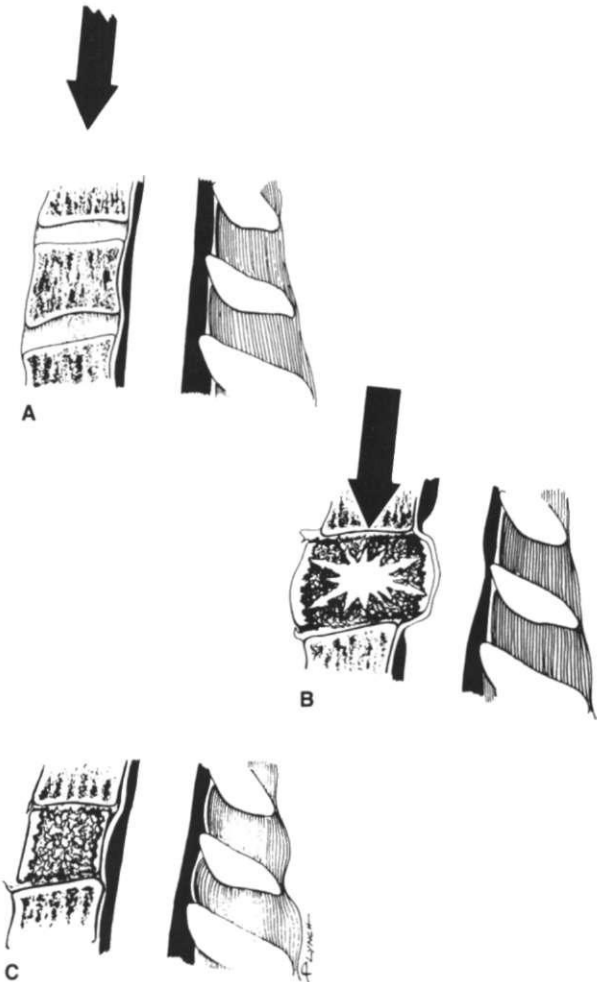


FIGURE 4-33 Representation of the presumed dynamics involved in both the compression and the "tear drop" fracture with cord damage. **(A)** A primary vertical force or a flexion moment occurs. **(B)** There is an explosion of the vertebral body, with deformation of the posterior shell of the vertebra, which pushes back and causes spinal cord impingement. **(C)** When the force is removed, the recoil phenomenon in tissues locally and through the body may leave only a very modest displacement of the vertebrae into the spinal canal, as shown in Figures 4-34 and 4-35, or no residual displacement in the canal, as shown here. The spinal cord injury may be severe, even though there is no residual impingement.

displacement. This patient sustained a diving injury and was quadriplegic at the C3–C4 level. The distinct block and a larger residual displacement are seen in Figure 4-35B, a cysternal myelogram.

The work of Fuentes[92] and Torg[261] brings new light to this well-recognized situation. In addition to the mechanism described, it may well be that a significant number of the neurologic problems associated with this fracture are due to posteriorly extruded fragments of vertebral body in the canal. The "tear drop" fracture of the cervical spine is analogous to the "burst" fracture of the thoracolumbar region. Thus, it will be more readily diagnosed and better evaluated and treated following analysis by CT scanning. The authors recommend a CT scan for all tear drop fractures.

Before discussing treatment, the characteristic neurologic picture that accompanies this fracture should be described. The classic findings indicative of *anterior spinal syndrome* are usually present in the typical "tear drop" fracture dislocation. There is immediate complete paralysis and cutaneous sensory loss below the level of the lesion; however, there is bilateral sparing of vibratory and position sense. The neurologic patterns associated with some of the different patterns of spine trauma are discussed in more detail in Figure 4-51.

Suggested Treatment

The patient should be evaluated to determine the presence or absence of clinical instability. The simple anterior compression fractures without neurologic deficit can be managed by bed rest in the acute phase, followed by a cervical orthosis of minimal or intermediate control for 3–6 weeks, depending on the clinical conditions. The comminuted compression fracture without any suggestion of instability or neurologic deficit may be treated in a similar manner. These latter injuries, however, are more likely to require additional bed rest with traction and an orthosis of intermediate control for a longer period of time.

It should be noted that in some cases of this injury, presumably when the posterior longitudinal ligament remains intact, axial traction completely reduces the posteriorly displaced fragments. Thus, skeletal traction of an appropriate magnitude is recommended as the first step in the treatment of this injury. Axial traction should always be applied incrementally and checked by lateral radiographs in order to recognize problems of overpull or a positive stretch test. In some patients, when the reduction is satisfactory, there is no clinical instability, and other aspects of the clinical picture are good, treatment may consist of 1 week of traction followed by another 5–6 weeks in a cervical orthosis of intermediate control.

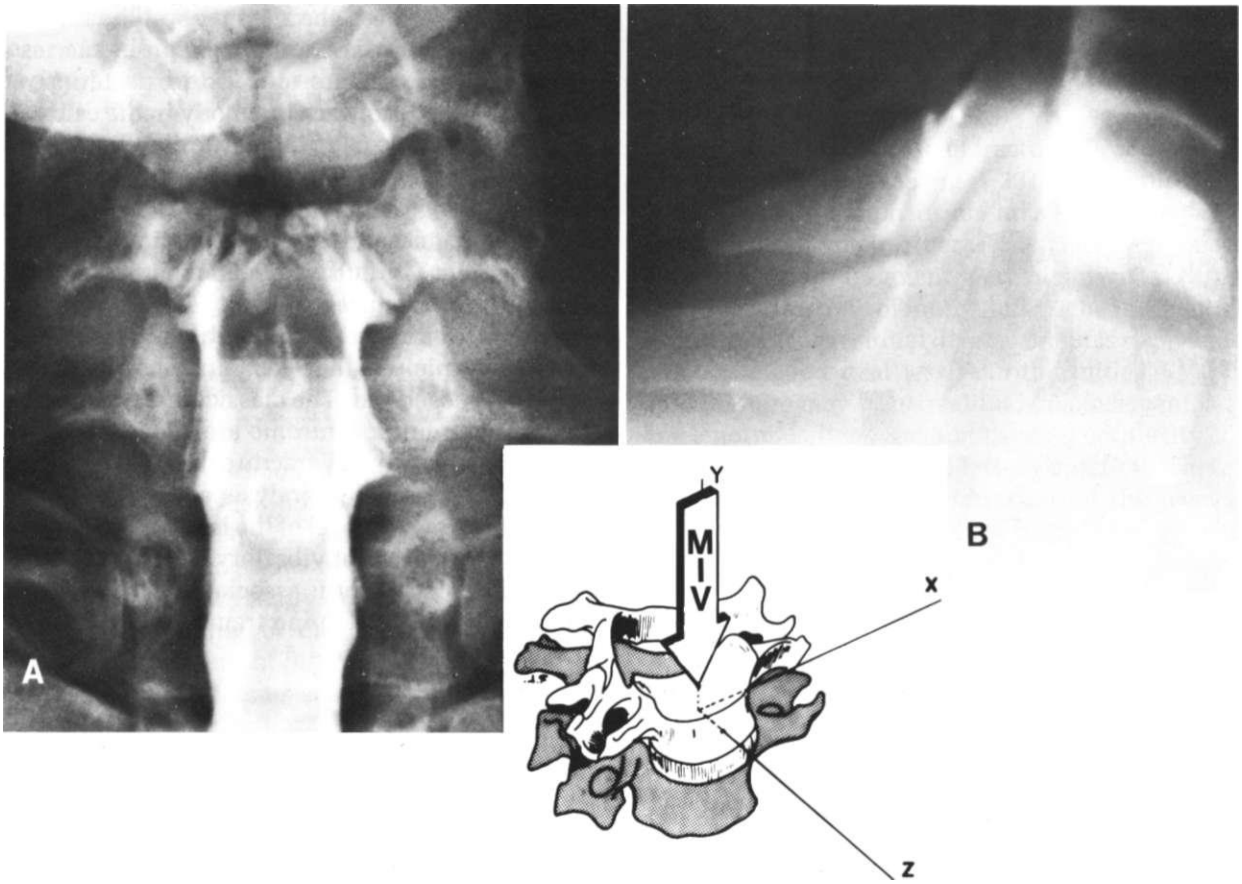


FIGURE 4-34 Radiographs of a college football player following head-on, spear tackle. The patient was immediately tetraplegic following the injury. **(A)** Moderate compression of the body of C5, with complete block on myelogram. **(B)** A laminagram that shows slight increase in radiodensity of C5 secondary to compression. There is also a vertical fracture and some posterior displacement of the body of that vertebra. This injury is thought to exemplify the diagram of Figure 4-33. The MIV is primarily a high-magnitude force exerted along the y-axis in the negative direction. *(Courtesy of Teaching Library, Section of Neuroradiology, Yale University School of Medicine.)*

In the "tear drop" fracture dislocation, the emphasis has been on early laminectomy with dentate ligament transection, followed either immediately or in 2–3 weeks by spine fusion. This was recommended by Schneider, on the basis that this was of benefit to the cord in both the immediate and advanced stages of the disease.[221,223] The anterior cord compression is presumed to be related to fragments of bone or disc material displaced into the canal, traumatizing the cord and/or compromising the anterior spinal artery. The authors' opinion is that decompression is indicated when there is incomplete neurologic deficit and imaging evidence of material in the canal. In our view, it should be by way of an anterior corporectomy and fusion. The site of the spinal cord embarrassment is anterior. There may be recovery of an additional root, and in addition the traumatized cervical spine, stabilized by fusion, prevents cervical kyphosis and may allow for better overall nursing care and patient rehabilitation. Therefore, we have decompressed and reconstructed the spine as shown in Chapter 8.

Lateral Bending Injuries of the Cervical Spine

There are injuries to the cervical spine that result in unilateral wedging or fracture of the vertebra, with or without neurologic deficit.

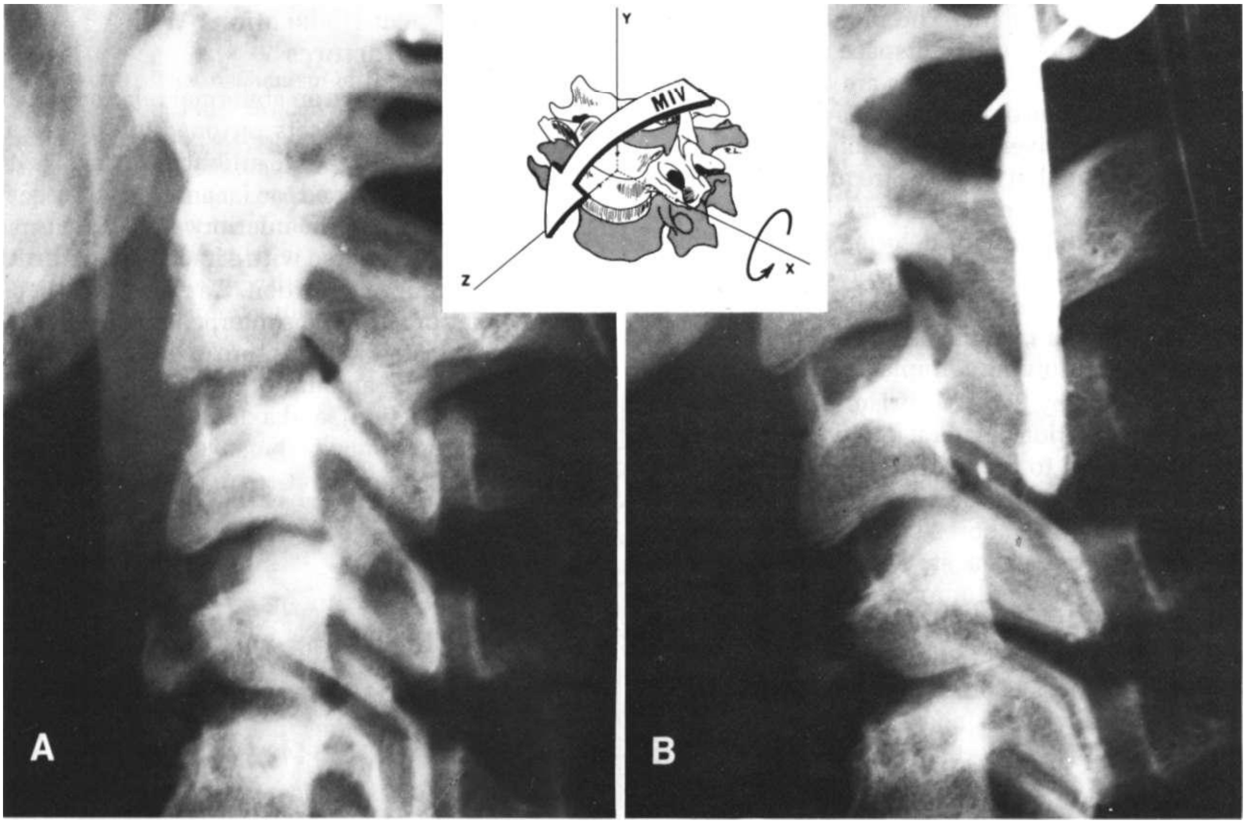


FIGURE 4-35 Radiographs of a classical example of the "tear drop" fracture. The patient was tetraplegic following a diving injury. **(A)** The triangular fragment off the anteroinferior body of C4. Compression is also seen here, as well as displacement of the posterior portion of the vertebral shell into the spinal canal. **(B)** The complete block at that level as a result of the injury. The MIV is shown. Because of the significant wedging of C4, C5, and C6, some +x-axis torque is hypothesized. Note also that in both *A* and *B* there is slight separation of the posterior vertebral elements at C4–C5. This is due to tensile loadings associated with the +x-axis torque. The triangular portion of bone ("tear drop") off the portion of the body of C4 is an example of shear failure shown in Figure 4-8. *(Courtesy of Teaching Library, Department of Diagnostic Radiology, Yale University School of Medicine.)*

Mechanism of Injury

The major injuring vector in this injury causes a significant degree of rotation about the z-axis. The vector, as seen in the coronal plane (y,x), generally is somewhere between the y-axis and the x-axis. Thus, the head is forced to tilt to either the left or the right. Because of physiologic coupling, there is a component of axial rotation ($\pm\theta y$) involved. As in the case of bending, there is compression on one side (the concave side) and tension on the opposite side (the convex side) of the neutral axis. Consequently, the injury involves wedging of the vertebral body and possible unilateral facet dislocation or fracture of the lateral masses on one side with ligamentous sprains, and ruptures on the opposite side. The forces that create these injuries are readily generated in motorcycle, automobile, and football injuries. During such an automobile injury, the person is thought to be hit from the side by some object when thrown from one side of the vehicle to the other. In the football injury, the mishap occurs in the process of tackling or blocking.

Discussion

The lateral bending football injury has been well documented by Chrisman and co-workers.[40] Radiographically, the injury may suggest tearing of the intertransverse ligaments and/or an acute angulation in the frontal plane. The radiograph may also show asymmetry of the interspace at the site of injury. These structural changes are often associated with either chronic or prolonged asymmetrical neurologic deficits. Given the mechanism of injury, it is readily apparent that it may be frequently associated with a type of avulsion of the brachial plexus on the convex side of the bending deformation at the time of injury. Neurologic deficits have ranged in severity from a rapidly resolving (15 seconds to several minutes or hours) pain, paresthesias, and paralysis, to a complete, unresolving tetraplegia, as in a football player reported by Roaf.[210] Chrisman noted in these patients a limitation of lateral flexion to the involved side in addition to the neurologic deficit.

It is interesting to speculate on the observation that players with a short "bull neck" and players with a long, supple neck are less frequently injured than those in the medium range. It fits with a biomechanical evaluation that the long, supple neck may bend over a longer radius of curvature, and thus less bending stress is produced. On the other hand, the short, bulky "bull neck" is stronger, and the short length offers a shorter moment arm and results in less of a bending moment, which causes less damage.

Suggested Treatment

In severe types of injury, similar to those described by Roaf, it is important to bear in mind that treatment with longitudinal traction may amplify the deformity and displacement of the vertebra. In severe injuries, the patient benefits from a cervical orthosis of intermediate control.

Chrisman noticed an increase in these injuries in college football players when their equipment was changed from the high type of shoulder pads to the lower, professional type of shoulder pads. The lower pads allowed a greater range of lateral bending of the cervical spine and therefore did not protect the spine against this type of motion when the high-magnitude forces involved in competitive football were applied. Consequently, prevention of these types of injuries is accessible through the use of the high type of shoulder pads or use of a heavily padded horseshoe-shaped cervical collar.

Because of recent observations[73] on the association between traumatic neurologic injuries and small cervical canal anteroposterior diameters, canal size should be taken into consideration in advising patients about returning to contact sports.

Unilateral Facet Dislocations With and Without Fractures

These injuries involve an abnormal relationship between the articular facets on one side, at the involved level. The caudad articular surface of the superior vertebra is fixed too far anterior to the cephalad articular facet of the inferior vertebra. This may or may not be associated with a fracture of the articular facet or lamina. A portion of a ruptured articular capsule may or may not be interposed. This injury is usually associated with rupture of the posterior ligaments, a variable amount of anterior displacement (+z-axis translation), and axial rotation ($\pm\theta y$). The annulus fibrosus may be damaged to a variable degree. The anatomic changes with this injury are well described by Beatson.[17]

Mechanism of Injury

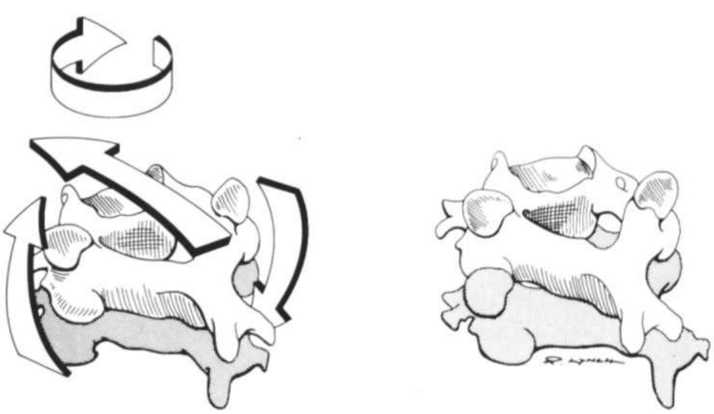
An appreciation of this injury is intimately linked with an understanding of the normal kinematics of the cervical spine. A unilateral facet dislocation is caused by an exaggeration of the normal kinematics of the spine. Physiologic lateral bending is coupled with axial rotation, such that the spinous processes tend to move toward the convexity of the physiologic curve (see Fig. 2-15). When this is exaggerated in trauma, the facet on one side goes too far caudad, and the one on the opposite side goes too far cephalad and dislocates (Fig. 4-36). An example of this type of dislocation without fracture is shown in Figure 4-37.

In unilateral facet dislocation where there is a fracture, a different mechanism is involved. Figure 4-38 shows an example of such a fracture. When there is a fracture along with dislocation, a large joint-reactive force has been created at the fractured facet articulation. There is an impingement of the surfaces that results in failure through the base of the more cephalad facet, which sometimes goes into the lamina. This can develop in one of two ways. There is either a significant element of axial compression involved with rotation and lateral bending, or there is an *unphysiologic* association of axial rotation with lateral bending. In the example presented in Figure 4-36, a unilateral facet fracture dislocation on the left can occur, with either a large component of axial compression or unphysiologic bending to the left.

Discussion

The relationships of the mechanisms of injury involving facet dislocations and fracture dislocations

FIGURE 4-36 The mechanism of injury in a unilateral facet dislocation. Basically, this injury results from an exaggeration of the normal coupling of axial rotation and lateral bending. If a significant vertical compression ($-y$-axis) component is added, then there may be a *fracture* of the *left facet*. If there is flexion and lateral bending, as shown here, but with a severe torque in the *opposite* direction ($+\theta y$), then there may be a fracture of the right facet.



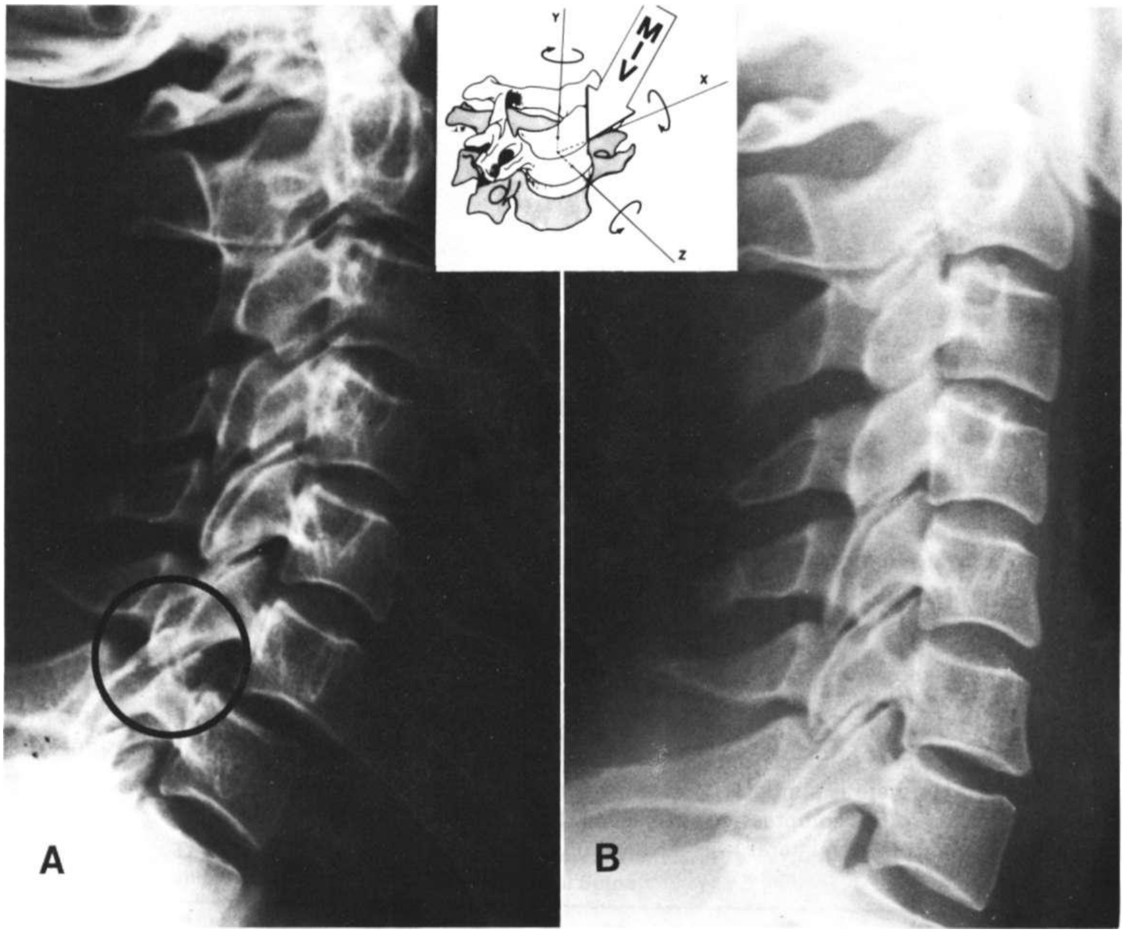


FIGURE 4-37 Radiographs of unilateral left facet dislocation. **(A)** The dislocation prior to reduction. There is an abnormal anterior translation of C6 on C7. Note the "interdigitation sign." The peaks of the lateral masses seen on this lateral view are not in their normal, shingled relationship, but there is an interdigitation of peaks shown in the circle. This is not always as readily seen, but when present at the same level where there is a moderate abnormal anterior translation, the findings are diagnostic of unilateral facet dislocation. Essentially, this injury is an exaggeration of physiologic flexion, lateral bending, and axial rotation. **(B)** The reduced state and the normal "shingling relationship" that the lateral masses exhibit on the lateral radiograph. *(Courtesy of Teaching Library, Department of Diagnostic Radiology, Yale University School of Medicine.)*

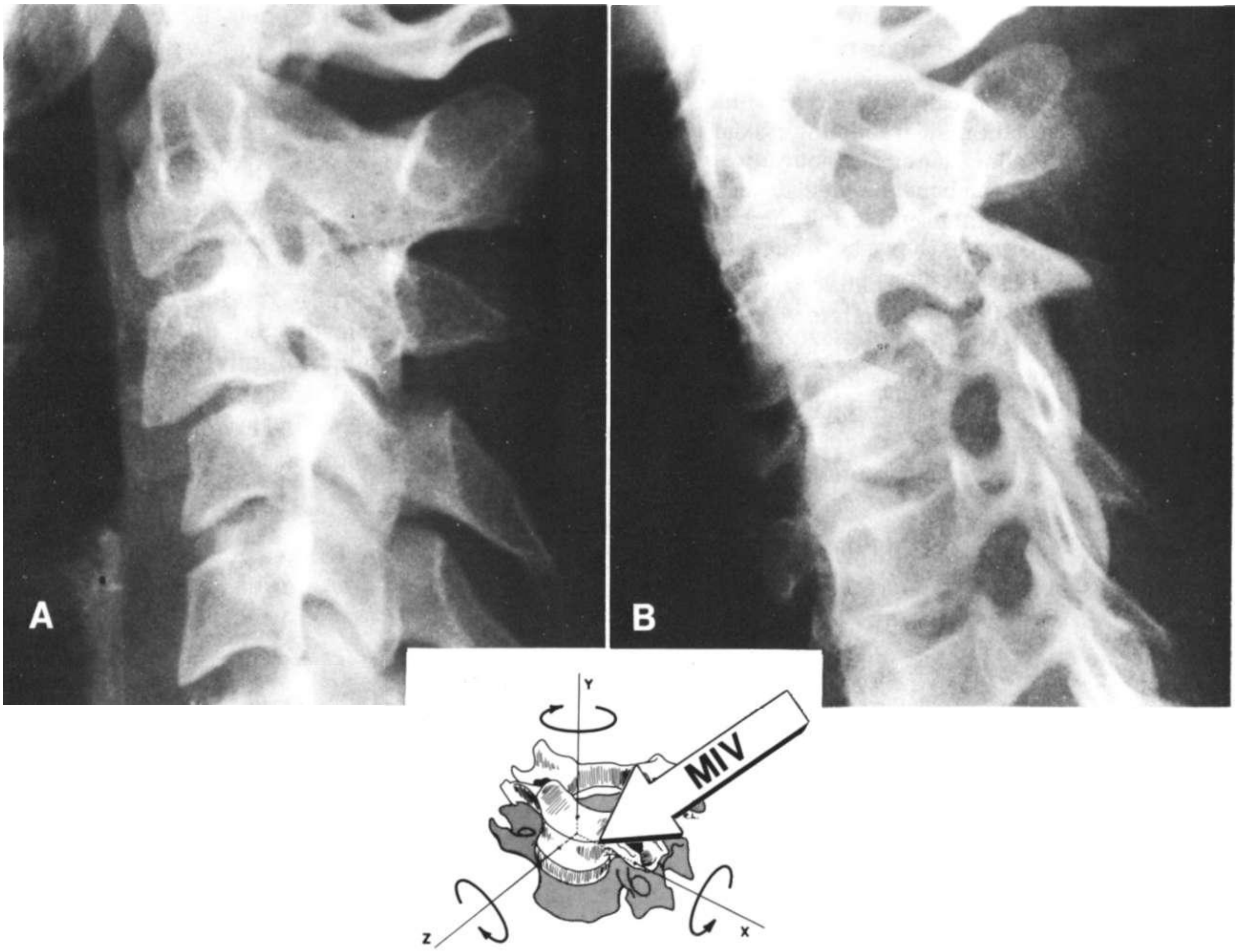


FIGURE 4-38 Radiographs of young male who sustained a unilateral fracture dislocation of the left facet. **(A)** The anterior subluxation, which includes a significant element of rotation. **(B)** This oblique view shows the fracture of the caudad portion of the left articular facet, which actually involved a small portion of the lamina. This was observed at the time of surgical exploration. The MIV for this injury has caused flexion, compression, and axial rotation. (There is respective +x-axis rotation, −y-axis compression force, and −y-axis torque.) The mechanism here is similar to that shown in Figure 4-36, but here there is a greater compressive force. The compressive force accounts for fracture of the facet (fracture dislocation) rather than pure unilateral facet dislocation.

result in an intriguing biomechanical analysis. The importance of a knowledge of normal kinematics in the clinical biomechanics of the spine has been pointed out. The relationship between physiologic and unphysiologic coupling patterns and unilateral facet dislocations and fracture dislocations provides a neat example for such a biomechanical analysis. Table 4-2 shows the relationships between abnormal forces causing certain movements and the injuries to be expected. It is also of interest to note that this combination of injuries found in the lower cervical spine is due to the special orientation of the facet joints. Their 45° angle is just right to permit either dislocation or fracture dislocation, depending upon the factors previously discussed. Table 4-2 does not include considerations of the effects of axial compressive loads on these patterns of injury. The axial compression load (−y-axis) increases the probability of a facet fracture in all the patterns, but more so in the situations where there is unphysiologic coupling.

These injuries may be readily diagnosed on a routine lateral radiograph when there is abnormal sagittal plane rotation, abnormal anterior translation, and the "interdigitation sign," as seen in Figure 4-37A. Sometimes, however, the injury can be difficult to recognize and is seen only on oblique views, pillar views, or laminagrams.

In the pure, isolated, unilateral facet dislocation, there is usually no neurologic deficit. The spine is clinically stable but painful. When there is associated disruption of the annulus fibrosus and fracture involving the facet joints, neurologic complication may also be present.

Suggested Treatment

The treatment may well be divided into pure unilateral facet dislocation and dislocation with variable amounts of associated injuries.

In situations where there is no evidence of neurologic involvement or fracture, and there is little residual displacement, much of the ligamentous complex of the functional spinal unit may be presumed to be intact. It is possible with considerable difficulty to reduce this dislocation and treat the patient with an orthosis of minimum or intermediate control, with a satisfactory result. There may be a late sequela of degenerative arthritis, but it can be managed as indicated.

The safe, effective reduction of such an injury constitutes a challenging and interesting problem in the clinical biomechanics of the spine. Skull traction with Vinke or Gardner-Wells tongs are recommended. These are thought to be the most useful because the site and mechanism of fixation to the skull allow for the application of very large loads without the tongs being pulled out. When there is an isolated dislocation, the ligaments effectively assist the muscles in holding the dislocated facet in the overlapped position. Considerable axial ($+y$-axis) displacement is required to permit the dislocated facet to snap back into place. When the reduction is successful, it often occurs with an audible pop and an immediate dramatic reduction in pain.

When the tongs are in place, an initial weight of 7–10 kg (15–22 lb) is applied, and lateral radiographs are taken. The surgeon looks carefully for abnormal axial separation between vertebral bodies, indicating disruption of the annulus and a much more severe injury. If this is not present, weights are then added in increments of 4–5 kg (9–11 lb), up to 28 kg (61 lb). After each increment of weight, radiographs are taken, and 1–2 hours should pass before the next increment, to take full advantage of creep. The patient may need some analgesics and sedation, with encouragement and a positive attitude on the part of the physician. Usually, the reduction is visible on the radiograph or heard by the patient before the maximum load is reached. If there is still no reduction in the 28–29-kg range (61–64 lb), the following procedure should be tried before abandoning closed reduction. With the traction reduced to 20–23 kg (45–50 lb), the head is gently flexed laterally away from the dislocated side approximately 60° and rotated toward the dislocated side approximately 60°. (First there is z-axis rotation followed by y-axis rotation. To reduce a left facet dislocation, a $+\theta z$ should be followed by a $+\theta y$; to reduce a right facet dislocation, a $-\theta z$ should be followed by a $-\theta y$.) While the head is still held in this position, 50% of the traction is removed to allow the facet to slip back into place. If subsequent radiographs show persistent dislocation, open reduction is required.

TABLE 4–2 Hypothesized Mechanisms of Unilateral Facet Dislocations and Fracture Dislocations in the Cervical Spine

		$+\theta z$ Right Lateral Bending	$-\theta z$ Left Lateral Bending
$+\theta y$	Head turns to left	Unphysiologic coupling ↓ Right facet fracture dislocation	Physiologic coupling ↓ Right facet dislocation
$-\theta y$	Head turns to right	Physiologic coupling ↓ Left facet dislocation	Unphysiologic coupling ↓ Left facet fracture dislocation

These maneuvers are based on an analysis of the kinematics of the spine and the presumed mechanism of injury.

At the time of open reduction, a portion of the cephalad articular process of the facet of the inferior vertebra on the side of the dislocation is removed. Because of the mechanical and anatomical disruption as a result of injury and treatment, as well as the opportunity presented by the open reduction, fusion is advisable. Posterior wiring and fusion of the two involved vertebrae are sufficient. For unilateral facet dislocations that have posterior element fractures and are clinically unstable by other criteria, open reduction, internal fixation, and fusion are suggested.

FLEXION INJURIES

Although it is somewhat of an oversimplification to describe injuries as "flexion injuries," we have chosen to group several injuries together. The first is a documented "full nelson" injury, another is "perched facets," another is the occult flexion injury in the athlete, and the final one is a bilateral facet dislocation. We suggest that these injuries are on a theoretical continuum in which there is primarily a bending moment in the sagittal plane (z,y) with a relatively small component of vertical force. The various injuries in this group as listed above are thought to be the result of increasingly greater bending moments.

Full Nelson Injury

Innocent "horseplay" provides a clinical experiment with the real-world quality that is superseded only by the judicial hanging phenomenon. Short of judicial hanging, the following clinical injury represents our best documentation of mechanism of injury.This important clinical event was reported by Thiel and Staudte in 1988.[259a]

A healthy 17-year-old male engaging in horseplay was put in a "full nelson" by one friend while each of two other friends held an arm in abduction and full extension (Fig. 4-39A). He was able to resist the flexion forces on his neck initially, but then he fatigued. When this occurred, he heard a creaking sound in his neck followed by a ripping feeling in the anterior part of his throat and the sensation of electric shocks in both arms.

A lateral view of the patient's cervical spine is seen in Figure 4-39B. There is a subluxation of C6 on C7, with a compression fracture of C7 involving a fracture of the anterosuperior rim of C7. This is probably due to avulsion by the anterior attachment of the annulus fibrosus, which pulled out of the rim of bone as a result of shear loading directed anteriorly. There is anterior translation of C6 on C7, separation of the posterior elements, and subluxation of the facet joints.

This is an unstable lesion because the posterior ligamentous elements are out and the disc is disrupted, allowing anterior translation and vertical +y-axis displacement or separation of the vertebral bodies of C6 and C7. This is best seen in the posterior elements. Had the anterior fracture not occurred, this could well have been one of the occult instability problems described in Chapter 5, that is, the occult instability following an acute flexion injury in an athlete, whereupon there may be only a slight compression fracture and minimal separation of the posterior elements.

Mechanism of Injury

This is the exciting aspect of this injury. We know the exact mechanism and we know precisely what the resulting injuries are. The major component of the injury is "flexion." However, a more detailed analysis indicates that there is primarily a large bending moment about the +x-axis. This involves a major component of tension on the posterior elements, some shear forces in the horizontal (x,z) plane, and some compression on the anterior elements. This is shown diagrammatically in Figure 4-40.

Discussion

This unfortunate accident has provided convincing documentation for some of the presumptions about mechanisms of injury in the cervical spine. In our theoretical analysis of mechanisms of injury presented on page 171 in the preceding part of this chapter, this injury would be more on the left portion of the spectrum, at about the A or B part of the illustration (see Fig. 4-1). There is relatively more of a bending moment and relatively less of a vertical (−y-axis–directed) load. This fits with the expected loading that would be produced by a "full nelson." This particular documentation of mechanism of injury and the availability of x-rays for pattern analysis provide a superb basis for the further development of a hypothesis to explain and analyze the occult instability following flexion injuries.

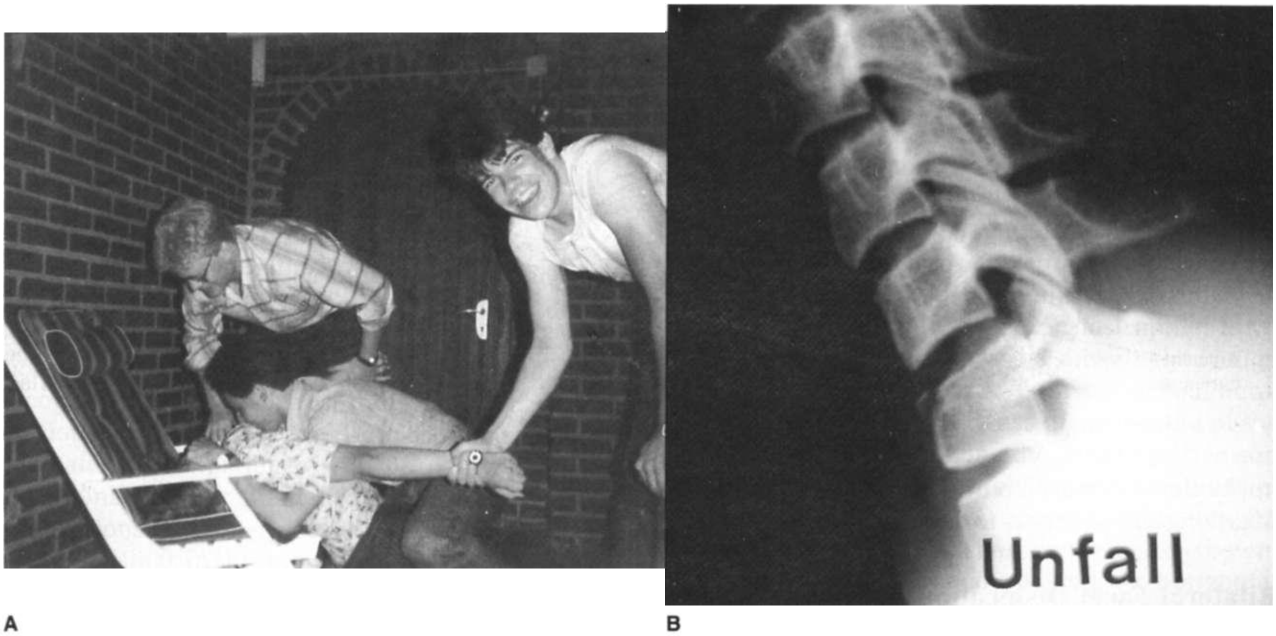


FIGURE 4-39 **(A)** This landmark photo was taken just at the moment that the young man's neck yielded, cracked, and sustained the injuries shown in the roentgenogram. **(B)** There is slight separation of the posterior elements and slight compression of the body of C7. The small triangular portion of bone has occurred in association with shear failure of the bone as the well-attached peripheral fibers of the annulus (Sharpey's fibers) remain attached to the bone. Note also the slight anterior translatory displacement of the body of C6 on C7. *(Photo and roentgenogram courtesy of Professor H. W. Staudte, University of Aachen, West Germany.)*

Suggested Treatment

This patient would be clinically unstable by the criteria proposed in Chapter 5, page 314. He was appropriately treated with posterior stabilization employing cerclage wire and bone graft.

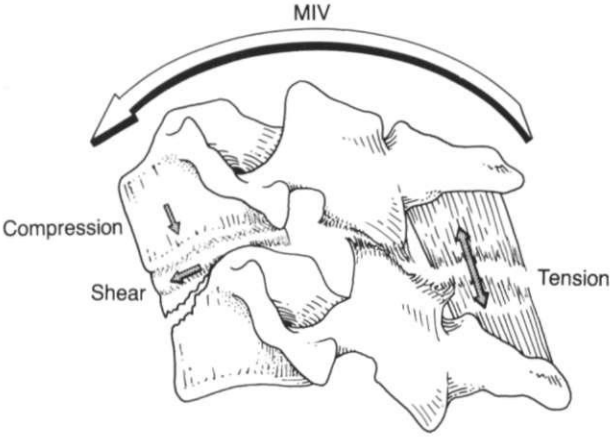


FIGURE 4-40 Diagrammatic representation of MIV mechanism of injury and the specific structures involved.

Note on the So-Called Perched Facets

In the clinical setting one sometimes encounters the phenomenon of either unilateral or bilateral "*perched facets*" in the cervical spine. At a given level, one or both facets on x-ray appear to be just touching at the tips (*i.e.*, the joint or joints are neither reduced nor partially reduced). They are not overlapped in the reversed abnormal relationship that occurs in a true unilateral or bilateral facet dislocation. Yet the facet(s) remains in the perched position and is not readily reduced, particularly in the case of a unilateral perch.

The authors submit for your consideration the following *hypothesis*. The facet(s) is, in fact, *not* perched but dislocated. The "perched" idea comes from a radiologic illusion. The facets are dislocated to the extent that their respective cartilaginous portions (and or invaginated capsule) are in fact overlapped, but because the cartilage is not seen on x-ray, the osseous components of the joint appear "perched" (see Fig. 4-41). The solution is to simply treat the situation as a full facet dislocation and

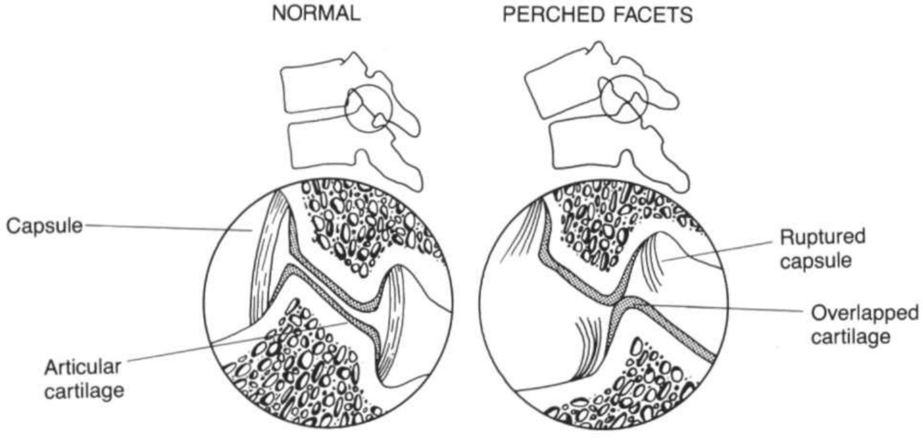


FIGURE 4-41 We hypothesize that the so-called perched facet is a true dislocation. The cartilaginous components are overlapped and locked, as shown here. It appears on the radiograph as "perched" because the overlapped cartilage cannot be seen.

apply the necessary controlled traction, muscle relaxation, and sedation to reduce it.

Bilateral Facet Dislocation

This injury involves the caudad articular facets on both sides of the superior vertebra being displaced anterior to the cephalad articular facets of the inferior vertebra. These injuries are generally associated with considerable disruption of the ligamentous structures of the functional spinal unit.[17]

Mechanism of Injury

In order for the facets to ride up, as they do in this injury, and displace superiorly (+y-axis) and anteriorly (+z-axis), there has to be considerable tensile loading on the posterior elements. This is most likely due to a flexion injury, involving little compression. The forces involved are probably of considerable magnitude. The MIV is presumed to be a flexion bending moment very close to the sagittal plane, as shown in Figure 4-42. Any significant asymmetrical application tends to result in lateral bending and axial rotation and unilateral rather than bilateral dislocation (see Fig. 4-36).

Discussion

These injuries are readily diagnosed, and usually the criteria of instability are suggested by a systematic evaluation of clinical stability in the cervical spine (see Chap. 5, p. 314).

Suggested Treatment

Regardless of neurologic status, the overall rehabilitation and comfort of the patient may be facilitated by internal stabilization. This decision is best made

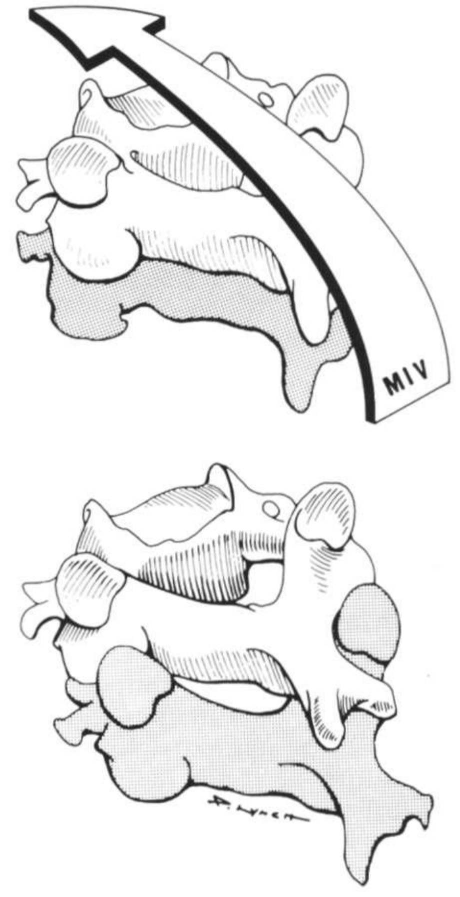


FIGURE 4-42 Mechanism of injury in bilateral facet dislocation. There is a major anterior displacement (+z-axis) with a significant element of force in the vertical direction (+y-axis). The MIV is as shown in the figure as a torque (about the +x-axis).

on an individual patient basis. In instances where there is partial neurologic loss and potential for considerable activity, reduction followed by internal fixation and fusion is more likely to be helpful.

ACUTE POST-TRAUMATIC CERVICAL DISC HERNIATION

Some recent work of Arena and colleagues[10] emphasizes the importance of the association of intervertebral disc extrusion with cervical facet subluxations and dislocations (Fig. 4-43). Failure to recognize extrusion may result in neurologic deterioration during the treatment process. The authors evaluate their experience with 68 patients having acute unilateral or bilateral facet subluxations or dislocations. There were six (8.8%) documented soft disc extrusions in this group. The first was a patient with bilateral facet dislocations diagnosed only after open reduction of the facets and posterior fusion converted the neurologically intact patient to a quadriplegic. Other patients in the series had similar diagnoses and clinical courses. Obviously, preoperative myelogram, CT scanning, or MRI should precede posterior cervical wiring and fusion (see Fig. 4-43). When a disc has been extruded, anterior disc excision should precede reduction and fusion.

Gross Dislocations and Fracture Dislocations of the Lower Cervical Spine

This grouping includes a large variety of injuries below the level of C2.

Mechanism of Injury

Gross dislocations and fracture dislocations of the lower cervical spine occur as a result of forces of large magnitude. Although there are a variety of possible injuries, each is discussed separately to distinguish it from the less severe but similar afflictions. The large majority of these injuries tend to occur in the sagittal plane. The MIV is either a large bending moment exerted about the x-axis, causing movement in the positive direction ($+z$) and resulting in a flexion injury, or a bending moment in the negative direction ($-\theta x$), resulting in an extension injury. The major injuring vector is thought to be a moment, because the hypothesized mechanism consists of a tensile failure of the intervertebral disc. The annular fibers are presumably ruptured or are more likely torn from the vertebral end-plate. These fibers are probably the most effective of the structures that resist translation in the sagittal and coronal planes. In order for gross dislocations to occur, the mechanical integrity of the disc must be destroyed, along with the anterior and posterior longitudinal ligaments and the posterior elements.

Discussion

Figure 4-44 is presented as a representative case for the analysis of mechanism of injury. Gross dislocations and fractures are usually catastrophic; however, occasionally they can be amazingly benign with respect to cord damage. An instantaneous autodecompression results in gross displacement of the posterior elements, allowing the cord to go free and unscathed by any portion of its osseous canal. In Figure 4-44, the patient fell approximately 6 m (18 ft) and struck his face, sustaining the injury shown. The force vector has two components—a horizontal force directed posteriorly ($-$z-axis) and a vertical force ($-$y-axis). The horizontal force, because of its distance from the site of injury, produces an MIV that has a large bending moment about the $-$x-axis. In all probability, the disc, including the upper cartilaginous end-plate of C7, translated forward with the body of C6. Note the triangular configuration of bone located beneath the anteroinferior corner of the body of C6. This is the anterosuperior rim pulled off the body of C7 by the peripheral, circumferential annular fibers, which are firmly affixed to the bone through Sharpey's fibers. The vertical compressive force accounts for compression of the lateral mass, the fracture of the pedicles, the laminae, and the base of the spinous process of C6. On the anteroposterior view, it can be seen that the lateral masses are not symmetrically compressed, indicating that the MIV is not acting purely in the sagittal plane. However, there is some component of the force acting around the z-axis (some element of lateral bending). Apparently, the trauma to the face was not purely in the midline but slightly to one side, creating this component. Finally, there is little or no disruption of the relationship between the spinous processes of C6 and C7. This fits with an extension mechanism.

Suggested Treatment

These severe fracture dislocations with or without neurologic deficit are probably best treated with reduction using skeletal tong traction followed by fixation and fusion. The traction is gradually applied in

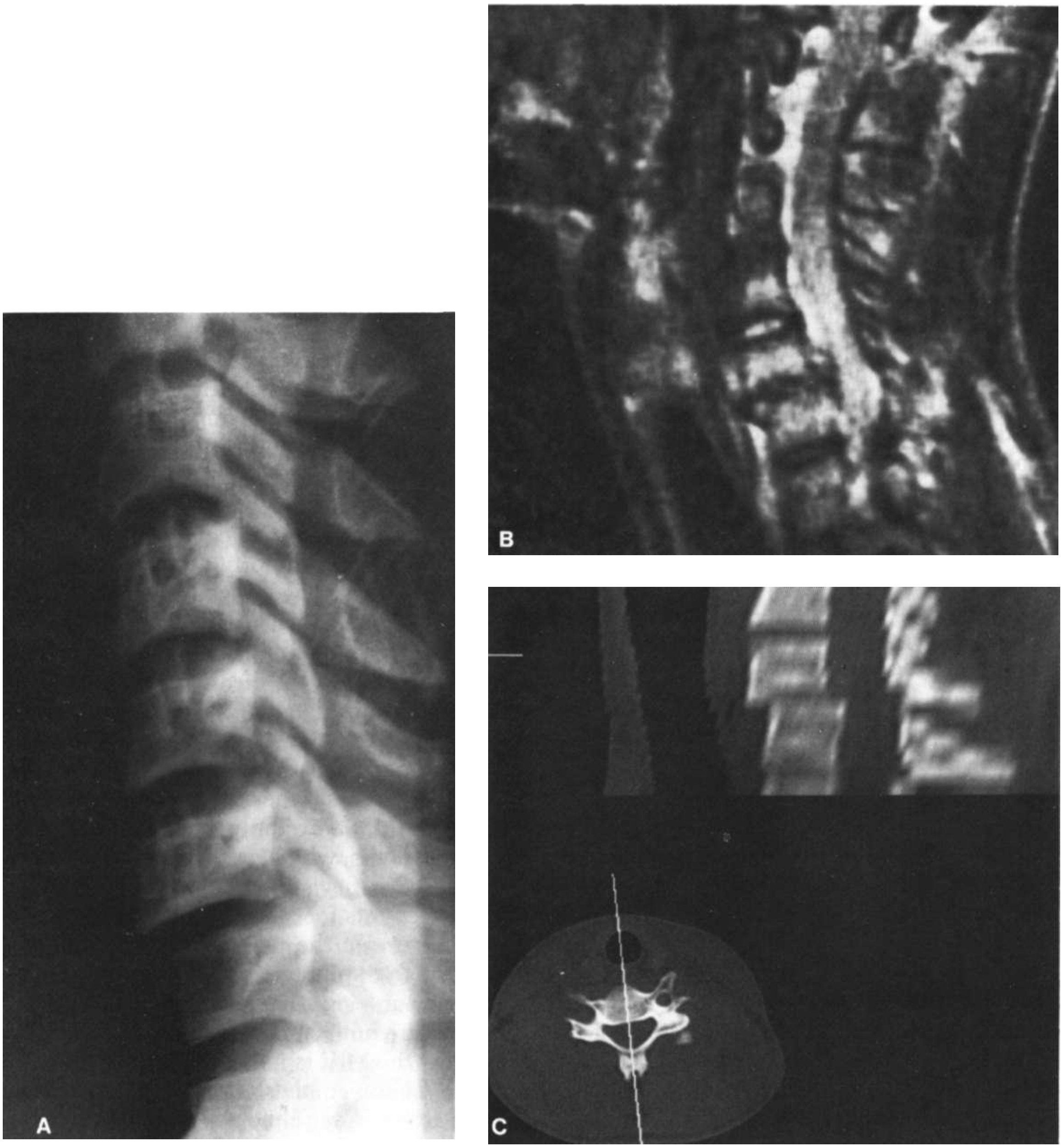


FIGURE 4-43 This is the important observation of Eismont and associates. **(A)** A post-traumatic unilateral C6–C7 facet fracture dislocation. The patient was neurologically intact. However, the patient was at risk to develop neurological deficit had there been an attempt at reduction. **(B)** An MRI image shows a large fragment of disc tissue in the canal and behind the body of C6. **(C)** The disc herniation may not have been visualized on CT scan. Unilateral and bilateral facet dislocations and fracture dislocations should have an MRI prior to attempts at reduction, unless there is already a complete neurological deficit. This patient was treated with anterior discectomy and fusion. Because this anterior approach further increases the instability, it must be followed with either a posterior fusion or, if reduction can be achieved, an anterior plate with screws. *(CT and MRI images courtesy of Professor Frank Eismont, M.D.)*

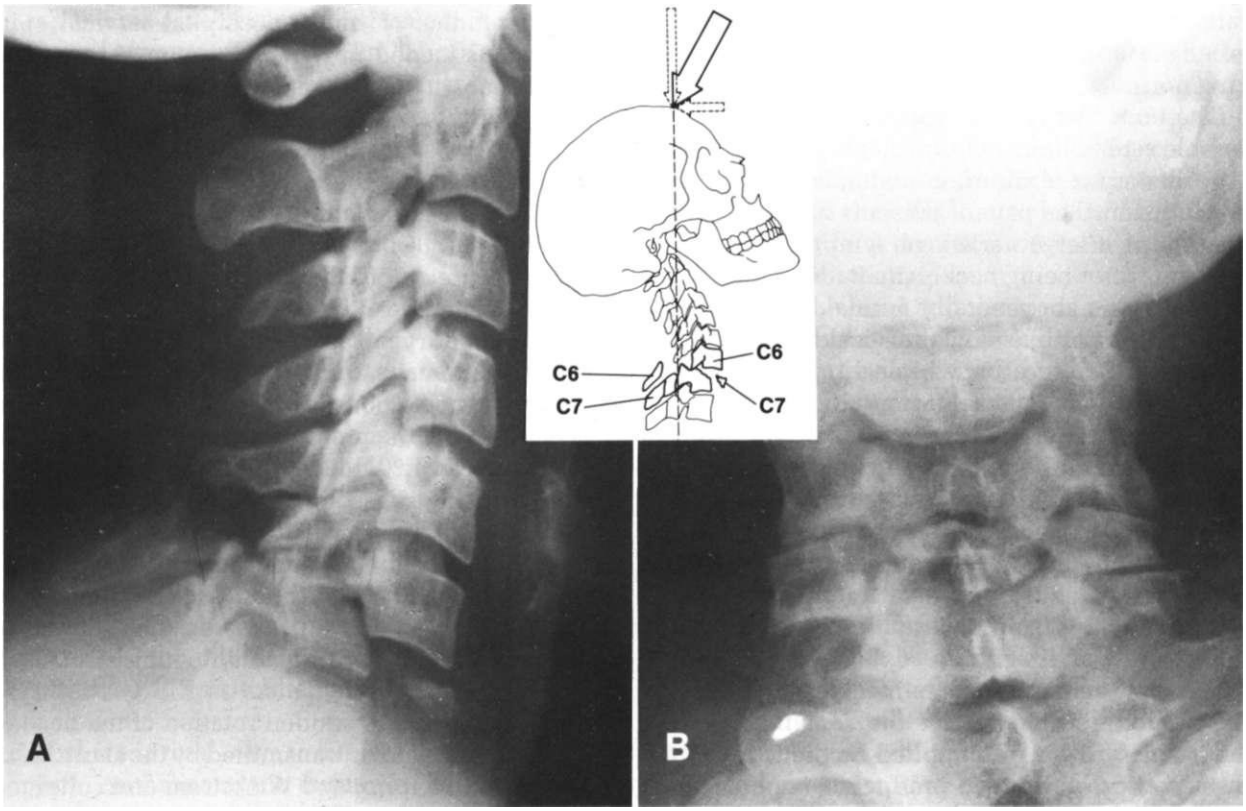


FIGURE 4-44 Radiographs of a 38-year-old male who fell and struck his face. There is an anterior dislocation of C6 on C7. There is fracture and wide separation of the laminae and spinous process of C6. The MIV is primarily a negative torque about the x-axis, created by the posteriorly directed horizontal force, with a significant −y-axis force vector, as indicated by the compression of the lateral masses of C6, shown in **B.** Note the triangular fragment of bone in **A.** This is probably a shear failure due to the vertical compression force. The fragment probably remains attached to the annulus, which is still attached to the inferior body of C6. *(Courtesy of Teaching Library, Department of Diagnostic Radiology, Yale University School of Medicine.)*

increments of 40–50 N (9–11 lbf) with radiographs and clinical neurologic monitoring. Fusion may be carried out through an anterior or posterior approach, depending upon the nature and character of structural damage and any site of decompression. Laminectomy has a high morbidity rate and loss of neural function. Anterior decompression and fusion for vertebral compression and neurologic deficit offered the best chance for neural recovery and restoration of stability.[23] Postoperative fixation can include internal wiring or plating, with or without halo fixation, which is determined by the stability required.

EXTENSION INJURIES

Again this is an oversimplification. However, it helps to organize the material this way. The hypothetical analysis of mechanisms of injury for the regions we will now discuss is depicted in parts E, F, and G of Figure 4-1 (p. 171).

Whiplash

Although it has been suggested that this term be abandoned, such a project may not be desirable or feasible, because it is thoroughly descriptive and well entrenched. However, when one attempts to define the term, this suggestion seems worth considering.

The term "whiplash" applies to a complex and variable set of clinical circumstances. There is usually a history of a minor or moderately severe rear-end collision. The patient presents with some combination of a large variety of symptoms, the only common factor being neck pain. Radiographs of the cervical spine are generally normal, except for the possible loss of physiologic cervical lordosis. There is enough literature on whiplash to fill an entire book of this size. Selected information on the topic and an overview of the current state of confusion are presented here.

Mechanism of Injury

The exact mechanism of injury is not certain, although it is generally considered to involve hyperextension. There is too much rotation about the x-axis in the negative direction ($-\theta x$). The inertia of the head tends to hold it in a resting position following the sudden acceleration of the remainder of the body. The forward pull applied by the trunk to the lower portion of the head produces a moment and a rotation of the head in the negative direction around the x-axis, causing extension of the cervical spine. The injury can also be caused by a phenomenon in which the head undergoes positive rotation around the x-axis as a result of sudden deceleration of the body, followed by a negative rotation around the x-axis as a result of recoil.

Discussion of Experimental Studies

Macnab simulated whiplash employing monkeys as experimental subjects. He dropped the animals from varying heights and they were suddenly decelerated, with the head left unsupported and free to move. The gravitational forces were varied by altering the height from which the animals were dropped. The anatomic lesions created were studied. The damage ranged from minor tears in the sternocleidomastoids to partial avulsions of the longus colli muscle. There were retropharyngeal hematomas and damage to the sympathetic nerves. Hemorrhages in the muscle layers of the esophagus were also noted. The most frequent and reproducible lesions were ruptures of the anterior longitudinal ligament and separation of the annulus fibrosus from the associated vertebra.[149] This is supported by recent studies of human occipital-cervical spine specimens loaded in hyperextension. As observed by Macnab, in the monkeys these spines failed at the disc or at the anterior longitudinal ligament.[45a]

Work by Patrick showed that if hyperextension or hyperflexion in either direction ($\pm\theta x$) describes a certain critical angle, then damage in the area may result.[194]

A discussion of whiplash injuries must include the associated head injuries. With the usual type of whiplash injury (an acceleration of the body leaving the head and neck to its own inertia), often there are associated contusions of the brain. These come from essentially two sources. The first is a contrecoup phenomenon in which there is movement of the skull, causing subsequent trauma to the cortex and cerebellum. There are also occasional injuries at the base of the brain due to the sudden angular acceleration of the skull. In addition to these injuries, an external blow to the skull may occur when the patient's body is thrown forward into some structure in the automobile. Thus, there is the possibility of brain injury due to sudden rotation of the head, as well as impact forces transmitted by the skull. Pruce, Torres and Shapiro, and Wickstrom and colleagues have reported significant electroencephalographic abnormalities in patients following whiplash injuries.[202, 266, 287] In some instances these findings have been similar to those observed in other types of closed head injuries. There is no doubt that some of these intracranial lesions may account for, or at least contribute to, some of the bizarre local symptoms, as well as some of the psychoneurotic problems that are often associated with this disease complex. Such information deserves due consideration in maintaining a humane, professional attitude toward patients with this injury.

It is interesting to estimate the magnitude of collision impact necessary to result in a concussion. Ommaya and Hirsch studied this problem using scaling techniques and extrapolating from data obtained in experiments on chimpanzees and monkeys. The results indicated that a head rotation acceleration of about 1800 rad/s^2 would result in a 50% probability of cerebral concussion.

An angular acceleration of the head of 1800 rad/s^2 (100,000 deg/s^2) is reached when a car is hit from behind, producing 5 g* horizontal acceleration of

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* g is the acceleration due to earth's gravity, typically 9.81 m/s^2 (32.2 ft/s^2).

the car.[180] This acceleration of 5 g is equal to attaining a speed of approximately 18 km/h from standstill within 0.1 second. In other words, if a car is hit from behind, causing it to move at a speed of 18 km/h (10.8 mph) within 0.1 second, there is a 50% probability of cerebral contusion for the occupants.

Studies by Clemens and Burow of unembalmed cadavers between the ages of 50 and 90 years provide some cogent data on rear-end impacts.[44] These studies were done at about 19 km/h and 13–16 g. Fifteen tests were completed. There were intervertebral disc failures in 90% of the cadavers, torn anterior longitudinal ligaments in 80%, tears of facet joint capsules in 40%, some bone fracture in 30%, tears of the yellow ligament in 10%, and posterior longitudinal ligament tears in 10%. The injuries were located mainly at the C5–C6 or C6–C7 region.

A considerable amount of work has been done on various experimental studies and mathematical modeling of the phenomena involved in whiplash injuries.[155] In the bibliography, some of the salient mathematical and experimental studies on this topic are included.

There are several ways to explore the mechanism of injury. Anthropometric dummies that have been instrumented to provide data on complete motion of various body parts may be used. Mathematical models may simulate the occupant–car system; this method has the advantage of great flexibility. In addition, animals can be used in real-life experiments. Unfortunately, the anatomy is much different from that of the human.

Figure 4-45A shows the experimental results obtained by Severy and colleagues when they simulated the whiplash injury mechanism by using well-instrumented anthropometric dummies and human volunteers (at slow speeds) during controlled experimental collisions.[229] Typically, a 13-km/h (8 mph) rear-end collision produced a 2 g acceleration of the vehicle and a 5 g acceleration of the head after a lapse of about 0.25 seconds.[229] This magnification of the acceleration for the head and the resulting forces are the result of the unrestrained head.

Motion of the head following collision is documented in Figure 4-45B. A typical curve of head rotation versus time is shown. The head first goes into flexion and then into extension within 0.2 seconds. The maximum injuring forces, however, occur in extension and are found mainly in the region of C6 and C7. The results were obtained by McKenzie and Williams using a mathematical model in computer simulation of whiplash.[163] The question of whether or not the head goes directly into extension or is preceded by some flexion, as depicted above, is debatable.[36, 94] Although most evidence may suggest that the major injuries are due to hyperextension, it is important for the clinician to bear in mind that there may also be a significant flexion component involved in the mechanism.

Motion of the head and the loads causing injury to the spine are dependent upon the seatback stiffness. This important effect is shown in Figure 4-45C. The two curves represent horizontal acceleration of the head, responsible for shear loads in the neck, versus time after collision. The stiffer seatback produces less acceleration and therefore less shear stresses. Similar results were obtained for the angular acceleration of the head, indicating that less bending stresses occur in the neck when the occupant is sitting in a stiffer seat. Obviously, with regard to whiplash injury, the harder seat is safer.

A study of Figure 4-46 demonstrates three basic safety factors related to whiplash injuries. First, there is the headrest, which limits the possible amount of extension that is allowed in the case of rear-end collision. Headrests should be at least as high as the level of the ears, which approximates the center of gravity of the skull. Unfortunately, in many designs they are below the center of gravity, in which case they serve as a fulcrum and accentuate injury. Second, there is the restraining capacity of the shoulder strap. By restraining the motion of the chest, there is a decrease in the amount of inertial forces exerted on the cervical spine. The importance of wearing seat belts from a preventive medical point of view cannot be overemphasized. A review of the facts should convince any reasonable person.[64] Third, there is the spring, which represents the stiffness of the seatback. The stiffer the spring, the safer the seat.

Recent work by Dvorak and associates has addressed carefully the anatomy and biomechanics of the craniovertebral complex. Although there are no specific data relating to whiplash, the investigators speculate about the possibility that C0–C1 instability secondary to failure or elongation of the alar ligaments may be involved.

In independent work by Worth in Australia on the C0–C1–C2 region, he studied patients with neck pain syndromes following automobile acci-

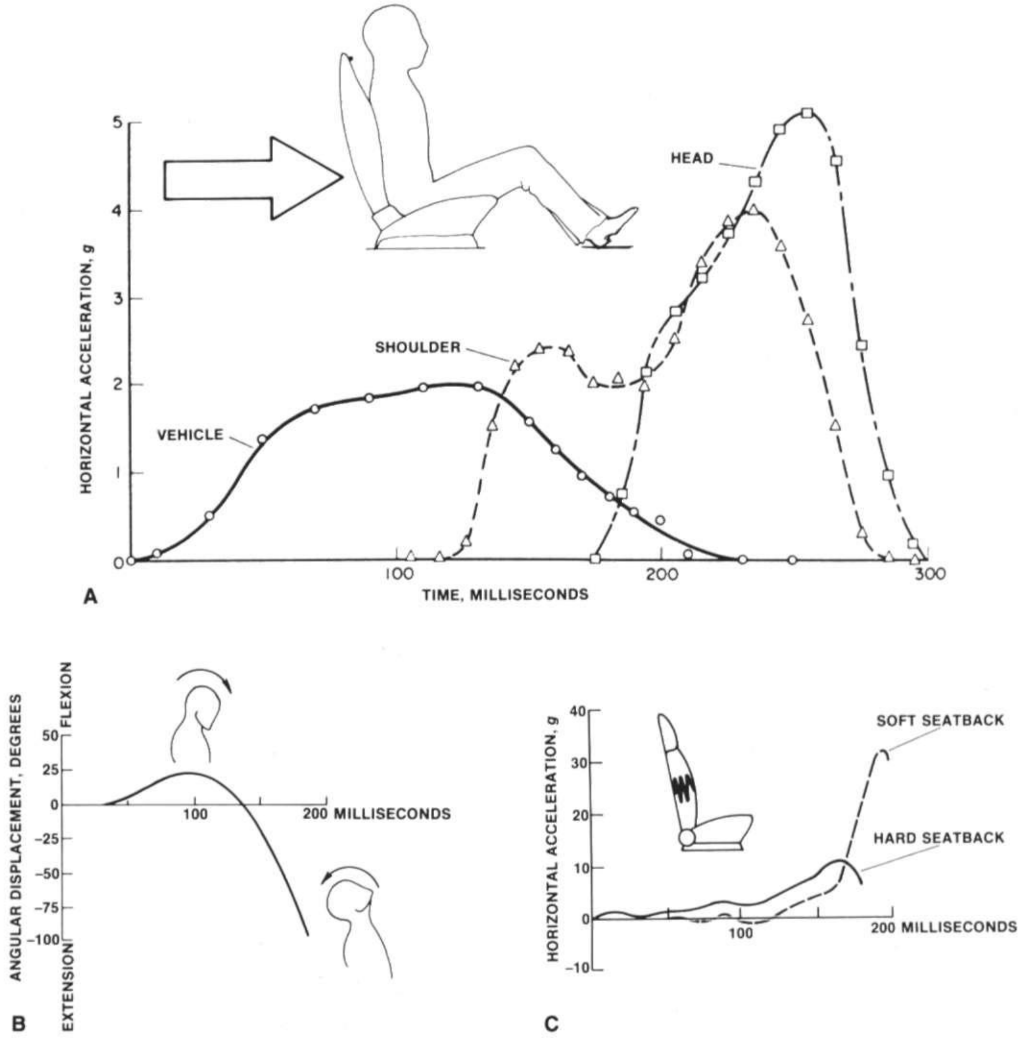

FIGURE 4-45 **(A)** Note that the shoulder and the head lag behind as the vehicle is accelerated when hit, but they catch up and within 0.3 seconds they reach accelerations 2 to 2.5 times the maximum vehicle acceleration. **(B)** The head first goes into flexion and then into hyperextension within the first 0.2 seconds. **(C)** The horizontal acceleration of the head for two different stiffnesses of the seatback. The harder seatback has a lower acceleration and therefore is associated with less injurious loading. *(**A** based on results from Severy, D. M., Mathewson, J. H., and Bechtol, C. O.: Controlled automobile related engineering and mechanical phenomena. Medical aspects of traffic accidents. Proceedings of Montreal Conference, p. 152, 1955; **B** and **C** based on results from McKenzie, J. A., and Williams, J. F.: The dynamic behavior of the head and cervical spine during "whiplash." J. Biomech., 4:477, 1971.)*

dents.[296] By manual analysis he reported decreased motion in the C0–C1 region and hypothesized that this might somehow be related to injury of the occipital-atlanto-axial complex. Two divergent hypotheses have emerged from these detailed studies of the kinematics and biomechanics of the C0–C1–C2 complex. Additional studies are needed to establish correlations of injuries in this region with the whiplash phenomenon. The idea of a ligamentous injury followed by excessive motion and the irritation of high spinal cord or brainstem elements seems to be the more appealing hypothesis.

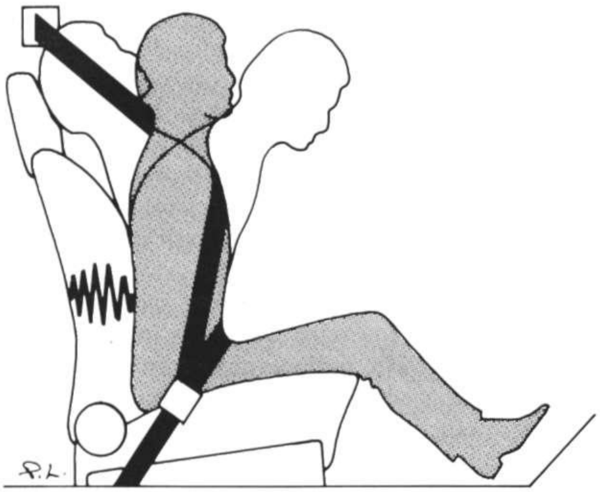

FIGURE 4-46 Representation of the headrest, the shoulder strap, and seatback stiffness, three factors related to the extent of injury in whiplash.

Discussion of Clinical, Psychiatric, and Medicolegal Considerations

The extensive attention that this group of injuries has received in the medical literature is partially due to the fact that the symptoms associated with it are much broader and less specific than in other neck injuries. Consequently, there is an abundance of medicolegal studies, psychiatric studies, and dissertations on the topic.

Typically, an individual is involved in a rear-end automobile accident. Either immediately following this, or after several symptom-free days, the patient develops some part of a broad symptom complex. There may be associated frontal or occipital headaches, numbness or weakness in one or both upper limbs, and vertigo or tinnitus. There may also be dysphagia and blurring of vision or nystagmus. Several clinicians have noted the most common findings to be neck pain with muscle spasm, limited motion, and loss or reversal of the normal cervical lordosis. The accompanying list gives another clinician's view of the findings that may result from a whiplash injury.[20] Gay and Abbott offer a comprehensive review of the major clinical presentation of this problem and the management of this disease.[94]

There has been much speculation about the possible influence of psychiatric factors and monetary compensation on the severity of the patient's symptoms. Gotten pointed out that there is a dramatic difference in the time lost from work *before* and *after* settlement. He reviewed the work record of a group of 100 patients with whiplash injuries. Prior to settlement, 41% of the patients lost 3 months or more from work. Following settlement, only 7% of the patients had lost that same amount of time. He noted that a number of the symptoms were of psychological origin and were refractory to treatment. Moreover, the symptoms usually resolved following settlement of the litigation. His hypothesis stated that the illness was used as a means of implementing psychological adjustments that had been postponed or unfulfilled because of financial difficulties.[99]

In contrast, there is the work of Macnab, who purports that these patients are not simply a group of hysterical, neurotic, or dishonest people. He notes that the symptoms are much more frequent in patients who have had hyperextension injury as opposed to flexion or lateral neck injury. He reasons that this strong predilection to extension injuries would not exist if it were based simply on psychiatric and compensation factors. He noted further that some of the concomitant injuries, such as broken ankles and other limb injuries, generally heal without prolonged refractory histories. Finally, he points out that if compensation and litigation are major factors, one would not expect 45% of the patients (as he found in his sample) to continue to have symptoms for 2 years or more after settlement.[149]

A similar study that does not accept the "psychoneurotic litigation" origin of whiplash symptoms is

CLINICAL FINDINGS FOLLOWING WHIPLASH INJURY

Narrowing of one or more intervertebral discs upon radiographic examination
Chip fractures or compression damage of vertebral bodies
Compression fractures
Tear of facet joint ligaments shown by subluxation on radiograph
Hyperalgesia over the cervical dorsal area
Pain, loss of motion, and involuntary muscle spasm in the neck and upper part of the dorsal region
Electromyographic evidence of motor nerve involvement
Sensory pain of a cervical, dermatomic distribution
Disturbance of cervical postural reflex
Blurred vision

(Adapted from Billig, H. E.: Traumatic neck, head, eye syndrome. Journal of the International College of Surgeons, *20*:558, 1953.)

that of Schutt and Dohan. These investigators reviewed a large series of women with neck injuries due to automobile accidents. They noted that there was no association between delay of onset of symptoms and persistence of symptoms and litigation. From this observation, they rejected the hypothesis that, after a symptom-free interval, patients develop symptoms as a result of "coaching" to build a case for litigation. Moreover, they found that symptoms persisted for approximately the same amount of time in the same percentage of individuals in two groups of patients, those involved with litigation and those with no litigation pending.[228] It is very difficult to know what role psychological and sociological factors play in whiplash injuries.

Hohl[116] reviewed 146 patients and was able to identify certain clinical characteristics in the early, post-injury stages that were associated with a poor prognosis.

Norris did a similar study.[176] The findings of the two investigations are presented in the chart below.

Discussion of Radiologic Considerations

Wagner and Abel carried out clinical and experimental studies that led them to the conclusion that the whiplash phenomenon may include occult injuries. With specialized radiographic techniques they identified several lesions: interarticular isthmus fractures, with or without lamina fractures; fractured transverse process of C1; rotary subluxation of C1 with respect to the occiput and the axis; and Luschka joint fractures. This work suggests that some patients thought to have normal radiographs following a whiplash injury may well have some of the above injuries, which are occult on routine radiographs.[275]

Zatzkin and Kveton compared radiologic findings in 50 patients with whiplash injury with 35 normal adults. The radiographs taken and the relative frequency of "abnormal" findings in the two groups are included in the accompanying list and in Table 4-3, respectively.[300] A radiograph should not be called abnormal unless several of the findings in Table 4-3 are present.

CLINICAL FACTORS ASSOCIATED WITH POOR PROGNOSIS FOLLOWING SOFT-TISSUE INJURIES OF THE NECK

Numbness and/or pain in an upper extremity
Radiographic visualization of sharp reversal of cervical lordosis
Restricted motion at one interspace (flexion/extension radiographs)
Need of cervical collar for more than 12 weeks
Need of home traction
Need to resume physical therapy more than once because of symptom exacerbation

Objective Neurologic Signs

Neck stiffness
Muscle spasm
Pre-existing cervical spondylosis

Upper factors: Adapted from Hohl, M.: Soft-tissue injuries of the neck in automobile accidents—factors influencing prognosis. J. Bone Joint Surg., *56A*:1675, 1974. *Lower factors:* Adapted from Norris, S. S.: The prognosis of neck injuries resulting from rear-end vehicle collisions. J. Bone Joint Surg., *65B*:608, 1983.

RADIOGRAPHIC STUDIES

Anterior view of cervical spine
Open-mouth view of the dens
Lateral view of cervical spine, with chin relaxed and in maximum flexion
Lateral view in maximum extension
Right and left oblique view of 45°

(Adapted from Zatzkin, H. R., and Kveton, F. W.: Evaluation of the cervical spine in whiplash injuries. Radiology, *75*:557, 1960.)

TABLE 4–3 Abnormal Findings in Whiplash Victims and Their Relative Frequency Compared with Normals

Radiographic Findings in Order of Importance	**Frequency Ratio**	
	Whiplash Patients	Normals
Marked decrease in ability to flex cervical spine	10	1
Marked straightening of normal curve	5	1
Scoliosis curvature in coronal (y, x) plane	5	1
Reversal of cervical lordosis	3	1
Slight to marked inability to extend cervical spine	2	1
Wedging or narrowing of one or more interverterbal discs	2	1
Encroachment on one or more intervertebral foramina	2	1

(Adaped from Zatzkin, H. R., and Kveton, F. W.: Evaluation of the cervical spine in whiplash injuries. Radiology, *75*:557, 1960.)

Suggested Treatment

Treatment of these injuries involves support for the neck during the phases of severe muscle spasms and stiffness. This can be achieved with a cervical orthosis of intermediate control. Various modalities of physical therapy have also been employed. The primary consideration, however, seems to be time, which allows the soft-tissue lesions to heal. The patient may be symptomatic for 2 or 3 days, months, or years. Certainly, appropriate analgesics are also useful, and some physicians may choose to use some of the so-called muscle relaxants, although these have not proved to be advantageous. Traction has also been suggested in the treatment of these patients.

When symptoms persist for 3 months or more, the physician may want to perform an anterior cervical spine fusion at a moderately suspicious level.[245] If the usual indications for the procedure are present, then it is reasonable to go ahead with it. However, a mere history of whiplash should not alter the usual conservative indications for anterior cervical spine fusion.[284] In the broad spectrum of spine problems, this is one of the most difficult syndromes to manage, and considerable supplementary objective information is required to solve this problem.

Careful examination; understanding; accurate, empathetic communications; and reassurance carry even more than their usual importance in the care of patients with whiplash injuries.

In the prevention of this injury, the most important consideration is that a proper headrest be available. This prevents the excursion of the neck to a degree of hyperextension that might be injurious to the intrinsic structures of the spine. It has been noted by Wickstrom and colleagues that some of the headrests in current auto models are not high enough and may actually serve as a fulcrum accentuating rather than preventing an injury (Fig. 4-46).[287] Moreover, the work of Patrick demonstrated that when it is possible to anticipate an injury, a potential victim can protect himself by clasping his hands together and placing them behind his head, and falling in this same position across the seat of the car. This maneuver makes use of the upper limbs and the adjacent seat to protect against hyperextension of the neck.[194]

Other preventive methods include designing a seatback with some type of attenuating mechanisms that would absorb some of the forces and diminish the initial acceleration of the body in relation to the head and neck. A damping mechanism included in the shoulder and lap restraint system may also be helpful. Patrick suggested that a primary consideration in seat design for protection against this injury is to limit the extension angle of the neck to below 80° and preferably below 60°. Meanwhile, observant motorists can protect themselves by paying attention to the vehicles that approach from behind.

Hyperextension Injuries Beyond Whiplash

This group of injuries is said to comprise 50% of serious cervical spine injuries, and they may be associated with significant cord or root damage. They may have an associated posterior element fracture and/or fracture of the anteroinferior tip of the dislocated vertebra. If we refer to the theoretical spectrum of loading, these injuries will be in the E, F, and G range of the spectrum.

Mechanism of Injury

Hyperextension injuries differ from the usual whiplash in clinical appearance and prognosis. It is tempting to hypothesize that in the hyperextension injury, the amount of rotation is greater than in whiplash. This may be true but is unproved. In contrast to the mechanism of hangman's fracture, the major forces are applied lower in the cervical spine. This may be due to the instantaneous axes of rotation created by the location of major force vectors. Another factor is the relative ability of the vertebrae in the two regions, in any given patient, to resist the applied loads. In other words, major injuring vectors that are very similar in two patients may produce different injuries. There is enough variation in the mechanics of the complex structure of the human spine that an extension injury could cause a hangman's fracture in one patient and a hyperextension injury involving C5–C6 in another.

Taylor,[254] in an article on the mechanisms of injury to the spinal cord in the absence of damage to the vertebral column, proposed the following. There are hyperextension injuries in which there is considerable extension of the spine without any temporary dislocation. However, the damage to the cord is a result of impingement of the forward bulging of the ligamentum flavum. This hypothesis is supported by five cadaver studies in which there was hyperextension of the spine and observable impressions on the posterior portion of the cord at the level of the interlaminar spaces. It was noted in these studies that the thecal space for the cord was narrowed as much as 30%. Figure 4-47 is supportive of this interesting hypothesis.[129] However, in cadaver spines, in which these experimental tests were carried out, the ligamenta flava have probably lost some of their elasticity. When normal elasticity is present, these ligaments shorten with hyperextension and thus do not normally impinge upon the spinal cord.

Forsythe suggested a mechanism of injury in which hyperextension is associated with shear loading that develops at the site of dislocation. The injury results in rupture of the anterior longitudinal ligament, separation of the intervertebral disc at the vertebral end-plate, separation of the posterior longitudinal ligament from the vertebral body, and fracture or dislocation of the facets. The overall residual deformation may sometimes have a normal or near normal appearance. The residual lesion might show only slight *anterior* subluxation of the involved vertebra, a fracture, or no changes on routine radiographic examinations.[87] These considerations may contribute to the confusion over whether encroachment during hyperextension that is due solely to the yellow ligament, without fracture and/or dislocation, can result in cord damage.

In a more recent publication, Marar submitted some new considerations on this subject.[152] His study reviewed 45 patients with severe hyperextension injuries and noted the primary soft-tissue failure shown in Figure 4-48A. In addition, he studied four patients who died from their injuries. Careful autopsy evaluation was carried out, and a rather dramatic observation emerged from the investigation. In each of the four autopsy studies, a complete transverse fracture was found in the vertebral body at the level of the inferior portion of the pedicles (Fig. 4-48B). The cord had been compressed between the upper portion of the fractured vertebra and the lamina of the subjacent vertebra. Spontaneous reduction had taken place, and the fracture could not be visualized on routine radiographs.[151] The failure in these hyperextension injuries can occur through the osseous structures, as shown in this study, through the annulus fibers, or more likely at the attachment

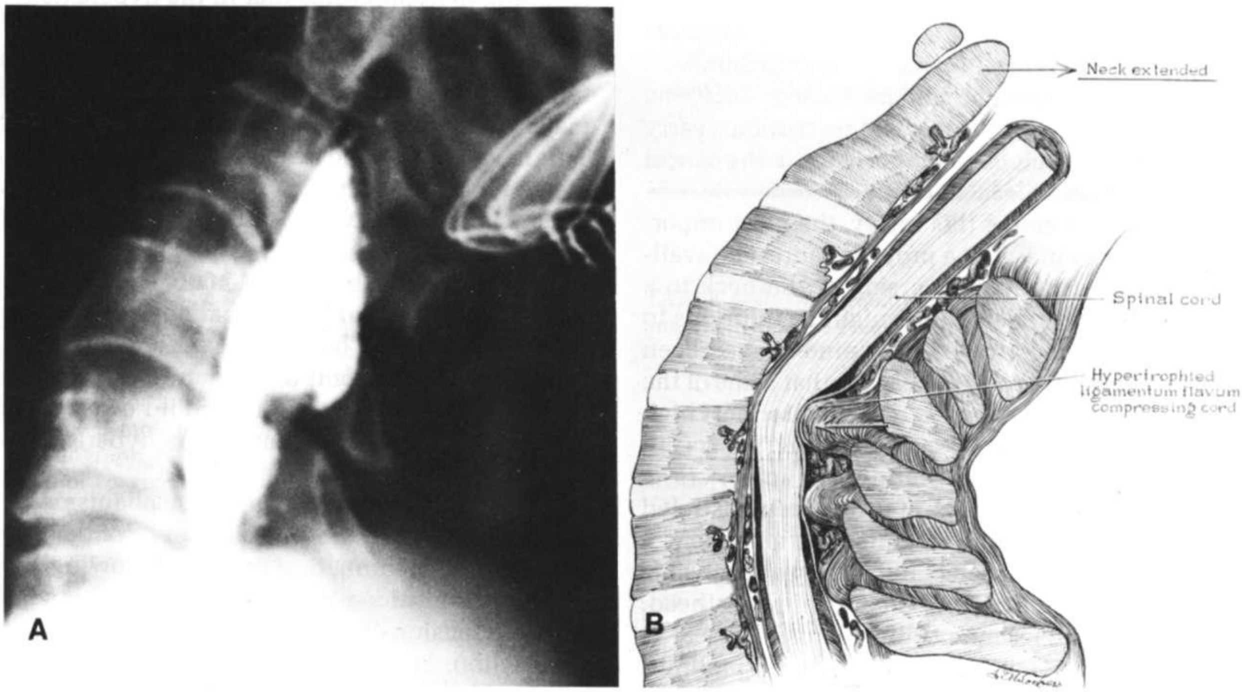

FIGURE 4-47 **(A)** Myelogram shows significant encroachment upon the spinal canal by an extradural structure in a patient who has Brooks' C1–C2 fusion. This encroachment is due to the infolding of the yellow ligament. **(B)** The situation diagrammatically. Such findings are sometimes associated with symptoms. *(Johnson, R. M., Crelin, E. S., White, A. A., and Panjabi, M. M.: Some new observations on the functional anatomy of the lower cervical spine, Clin. Orthop., 111:192, 1975.)*

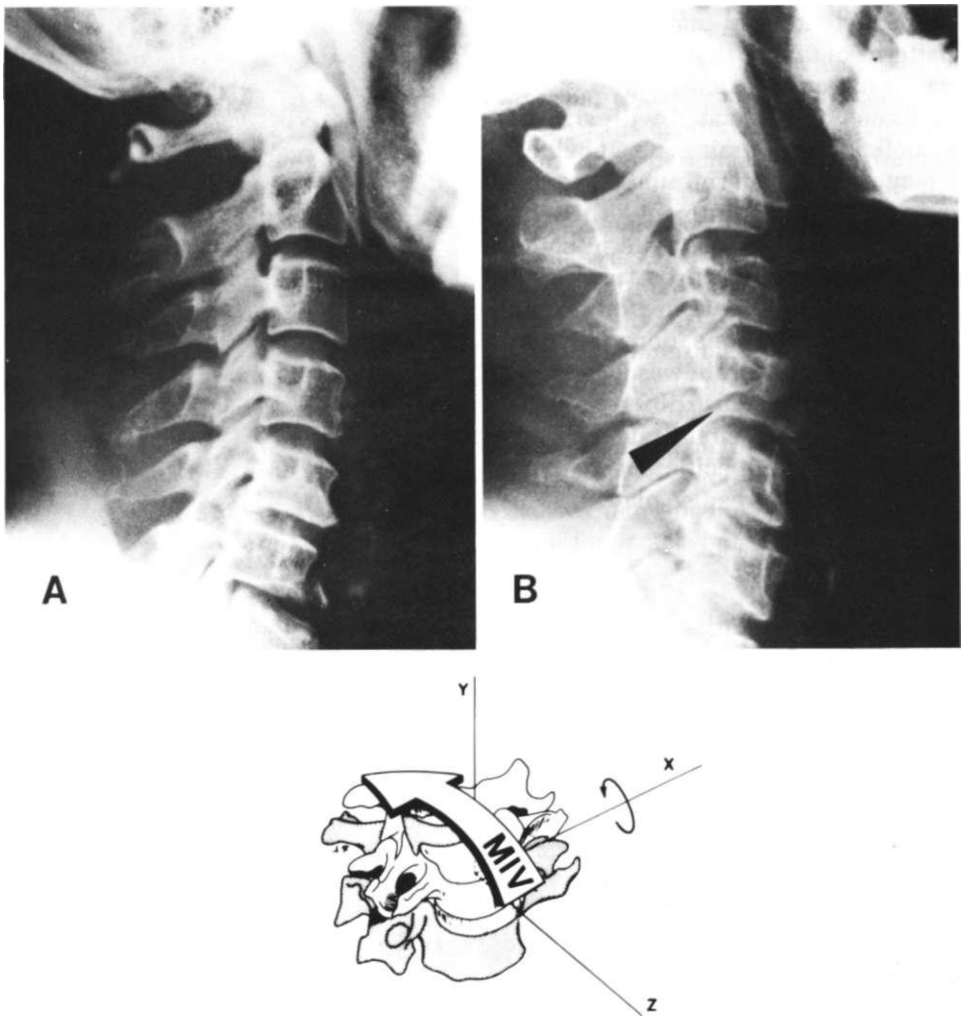

FIGURE 4-48 **(A)** Posterior displacement of C6 on C7 following a hyperextension injury. The patient had an incomplete tetraplegia. The triangular fragment from the anteroinferior lip of the body of C6 results from tensile failure of the bone, which has remained attached to the annulus through Sharpey's fibers and not the delicate anterior longitudinal ligament. The remainder of the separation has occurred at the cartilaginous end-plate, where the attachment is less secure. **(B)** This fascinating radiograph shows a transverse fracture of the body of C4 below the level of the pedicle. There is backward subluxation of the upper fragment. These findings suggest the presence of large shear forces. There was an incomplete tetraplegia. This fracture most likely would be missed on a routine radiograph should it spontaneously reduce, from either elastic recoil or moving the patient. The MIV for both injuries is shown in the inset. The major factor is a torque of high magnitude about the x-axis in the negative direction. Both of these injuries would be to the right of our theoretical loading spectrum shown in Figure 4-1. *(Radiographs from Marar, B. C.: The pattern of neurological damage as an aid to the diagnosis of the mechanism in cervical-spine injuries. J. Bone Joint Surg., 56A:1648, 1974.)*

of the annulus to the vertebral end-plate. Here, then, are additional patterns of failure that have the same basic MIV as that of hyperextension injuries of the cervical spine and may be associated with significant neurologic deficit. The strain rate at which trauma occurs may be the determining factor in whether the tendon or the bone fails under tensile loading. The work of Noyes on failure patterns of the anterior cruciate ligament of the knee of monkeys showed that tensile loading may either rupture the tendon or avulse a fragment of bone, depending on the strain rate at which the structure was loaded. In other words, the same MIV can cause different types of anatomic damage, depending on the rate of deformation or rate of loadings. Specific mechanical properties of individual structures also play a role in such determinations.[178]

Discussion

Forsythe and Marar emphasized that this unstable hyperextension injury may spontaneously reduce and have a normal radiographic appearance.[87,151] The stretch test (p. 318) may be helpful in the diagnosis of an occult transverse fracture of the vertebral body or an undisplaced failure at the disc-endplate interface.

The mechanism of cord lesion in many of these injuries is probably due to the "pincher phenomenon" described by Penning[197] and Taylor and Blackwood.[255] There is a significant compromise of the anteroposterior diameter of the spinal cord between the posteroinferior lip of the vertebra above and the anterosuperior portion of the subjacent lamina. The threshold for cord damage through this "pincher phenomenon" is greatly accentuated when there are osteophytes or when yellow ligament encroachment is present at the level of the translatory displacement. This is shown in the radiograph in Figure 4-49. A summary of the various pathoanatomic conditions that can traumatize the spinal cord through abnormal translation in extension injuries is presented in Figure 4-50. The yellow ligament, the osteophytes, and the pincher effect are the three main considerations. If any combination of these three factors is operative, the injury to the cord for a given displacement is certain to be greater.

Suggested Treatment

Carefully monitored axial traction should be used to treat these injuries. Posterior fusion with wiring and bone grafting has been recommended.[87] If the posterior longitudinal ligament is intact and there is good spontaneous reduction, the injury may be stable. The clinician cannot always be sure of the condition of the posterior longitudinal ligament. Thus, in order to avoid the catastrophe of initial or subsequent neurologic damage, it is suggested that these injuries be systematically evaluated for clinical stability; the unstable cases should be treated by posterior fusion, followed by an orthosis of intermediate control.

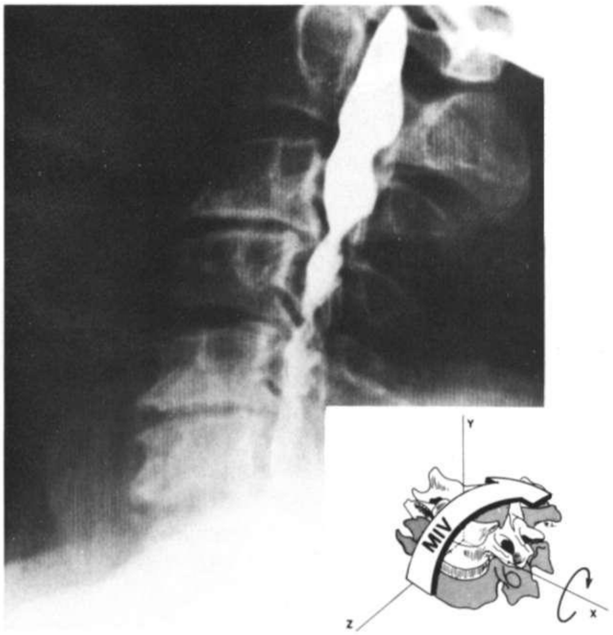

FIGURE 4-49 Radiograph of a 50-year-old male with cervical spondylosis who suddenly became paraplegic following a moderate fall off of a curb due to intoxication. The patient struck his head and extended his neck at the time of the fall. The cord damage here is due to the encroachment of the spinal cord at the C4–C5 level by midline osteophytes anteriorly and invaginated ligamentum flavum posteriorly, both shown by the partial block on myelogram. A radiologic estimate of the true sagittal anteroposterior diameter of the spinal canal in this patient is only 6 mm at the C4–C5 level. This is an example of minor trauma or possibly even a physiologic motion causing damage when the margin of safety and space for the cord has been significantly reduced. *(Courtesy of Teaching Library, Section of Neuroradiology, Yale University School of Medicine.)*

If no posterior elements are fractured, a posterior approach is the procedure of choice. This is because posterior wiring to an intact vertebra above and below the level of the ligamentous or the bony injury will provide some degree of immediate stability. In

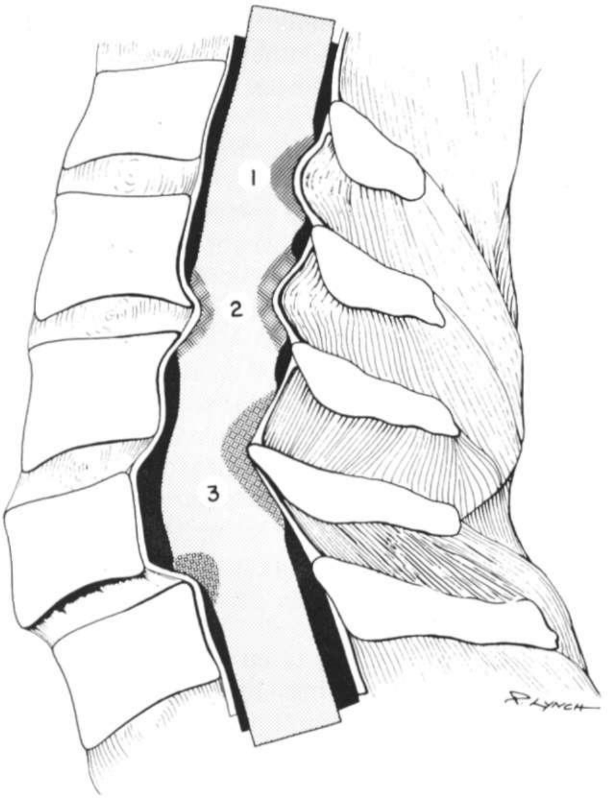

FIGURE 4-50 Anatomic factors that may compromise the space for the spinal cord and cause neurologic problems. *(1)* Encroachment by the yellow ligament. *(2)* Encroachment by the midline osteophyte anteriorly and the yellow ligament posteriorly. *(3)* The "pincher phenomenon." This can occur from anterior translation of the vertebra, as shown. It can also occur from posterior translation of that same vertebra, in which case the spinal cord is damaged by the posteroinferior edge of the vertebral body and the superior edge of the subjacent lamina (see Fig. 8-6). The effects of these factors are summative, and any combination of the three may occur at the same level. They either lower the threshold at which trauma can cause neurologic damage or they cause such damage.

the case of posterior element fractures or ligamentous disruption with this injury, the anterior approach may leave the patient unstable both anteriorly and posteriorly in the immediate prolonged postoperative period.[243] However, this may be compensated for by halo fixation. If the surgeon can be sure of intact posterior elements, both anterior and posterior fusion are nearly equally suitable. However, satisfactory *immediate* internal postoperative stability may be achieved with posterior fusion and wiring.

Central Spinal Cord (Medulla) Syndrome

In individuals with cervical spondylosis, hyperextension injuries of the cervical spine without fracture or dislocation sometimes produce a pattern of neurologic deficits known as the central spinal cord syndrome. The syndrome classically consists of motor deficit in the upper limbs and a segmental level of impaired sensation. Because the fibers in the spinothalamic tract conveying pain and thermal sensation are located posterolaterally and the touch fibers are grouped anteromedially, a dissociated sensory loss of pain and temperature sensation may occur depending on the size of the central lesion. In other words, sensation to pain and temperature is lost, but sensation to touch remains. If the central lesion involves more of the surrounding white matter, various long tract signs and bladder dysfunction may be noted.

Mechanisms of Injury

Taylor studied the effect of hyperextension of the neck on the distribution of Pantopaque between C3 and C7 in normal cervical spines. He found a decrease in the anteroposterior diameter of the canal of up to 30%. Presumably, this is due to the invagination of the yellow ligament. If, in addition to the space encroachment by the yellow ligament, there is also a transverse osteophyte protruding posteriorly, obviously considerable damage to the spinal cord may occur. Figure 4-50 shows how a combination of these factors can result in a compromise of the anteroposterior diameter of the cord and subsequent damage.

Schneider and colleagues constructed a crude model of the cord out of foam rubber and drew on it the cross-sectional anatomy, covered it with rubber to represent the meninges, and compressed it in what would be the sagittal diameter of the cord. This study suggests that in addition to compressive forces, there are associated longitudinal tensile forces that cause damage proximal and distal to the site of compression.[F] This may partially explain the axial progression of neurologic damage cephalad and caudad to the site of maximum compression.[222] The primitive biomechanical model does not take into consideration the role of the anterior spinal artery or the differential susceptibility of the various cord elements to the different strains they must encounter in anteroposterior compressive loading. These general findings are supported by the work of

Dohrmann and Panjabi on experimental spinal cord trauma in cats. They found the traumatic lesion to spread axially in both directions from the site of impact. The spread is related to the magnitude of trauma or more precisely the impulse of the transverse impact.[B,60]

Discussion

Rand and Crandall have suggested several possible neuropathologic sequences that may result from the mechanical changes described above:

A small local area of hematomyelia with associated edema and swelling in the central portion of the cord may develop.

There may be central local recess of the gray matter that occurs secondary to hemorrhage. This is somewhat analogous to the lesions produced experimentally by Wagner and colleagues through direct impact loading of the exposed spinal cords of cats.[272]

Simple central concussion with surrounding edema may occur.

There may be infarction with central softening secondary to compression or damage of the anterior spinal artery.[203]

There are a number of possible mechanisms that may produce the final neuropathology once the external forces are generated. Some of the major investigations on this subject have been reviewed on page 184.

It should be noted that a midline posterior osteophyte alone may compromise the cord enough to cause damage even without discrete trauma. Murone emphasized the importance of the role of the initial normal sagittal diameter of the cervical spinal canal as related to the development of myelopathy. In his study of Japanese males with cervical spondylosis, he found that those who developed myelopathy had initial sagittal diameters that were less than normal, and those without myelopathy had sagittal diameters greater than normal.[171] Thus, cord injury can also occur gradually or as a result of a mild or severe flexion injury. A crucial variable in either situation is the initial sagittal diameter of the cervical canal for the particular patient.

The injuries may not have obvious associated fractures, and this may be difficult to diagnose radiographically. There are several characteristics that may help piece together the complete clinical picture. One obvious finding is cervical degenerative disease with one or more transverse bars (osteophytes).

An avulsion fracture of an anterior osteophyte may be present, suggesting a hyperextension injury. The old, reliable sign of retropharyngeal soft-tissue swelling or hemorrhage may be manifested. Other clues include any clinical consideration in the history suggestive of an extension injury and the presence of bruises or lacerations of the forehead or face.

Table 4-4 is presented here, but in fact it is relevant to the entire section on cervical spine trauma. Marar presented a classification of neurologic injury patterns in cervical spine trauma and attempted to correlate them to specific injury mechanisms.[152] Table 4-4 comes from this study. The classifications

TABLE 4–4 Associations of Gross Neurologic Patterns and Broad Injury Mechanisms in the Cervical Spine

	Neurologic Damage	Injury
Group I	Total motor and sensory loss to all four limbs. Total transection of cord. No recovery occurred.	Burst fracture or bilateral facet dislocation, flexion injury.
Group II	Motor loss of varying degrees, either in all four extremities or in the upper limbs only. Sometimes segmental or patchy transient sensory loss was associated. *(Central spinal cord damage)*	Hyperextension injuries.
Group III	Complete motor loss in the extremities, with hypoesthesia and hypalgesia up to the level of the lesion. No loss of position and vibratory sense. *(Anterior spinal cord damage)*	Vertical compression, bursting injury, "tear drop" fracture dislocation; possibly some associated flexion or extension.
Group IV	Motor power in all four limbs or the upper extremities alone with no sensory loss.	Unilateral facet dislocation, fractured arch of atlas, and a variety of injuries.
Group V	Brown-Séquard syndrome.	Unilateral facet dislocation or a burst fracture.

(Marar, B. C.: The pattern of neurological damage as an aid to the diagnosis of the mechanism in cervical-spine injuries. J. Bone Joint Surg., 56*A*:1648, 1974.)

and correlations are not sharply delineated. However, the approach is novel and interesting, and the analysis constitutes a noteworthy endeavor to impose some order to an extremely complicated and difficult clinical area. To a large degree, progress in this area can be measured by the extent to which it is possible to collect, analyze, and interpret observations. Physicians can then more clearly develop, define, and delineate the entities in Table 4-4, placing them in precise, functional categories. A good deal is still to be learned.

Figure 4-51 is provided as a summary and review of the pathoanatomical considerations important in the understanding of the anterior spinal cord syndrome and the central spinal cord syndrome.

Suggested Treatment

Several investigators, Bailey and Rand and Crandall, have emphasized that surgical intervention is not indicated for the central spinal cord syndrome.[13, 203] Schneider and co-workers suggested that it is probably contraindicated.[222] Thus, the diagnostic consid-

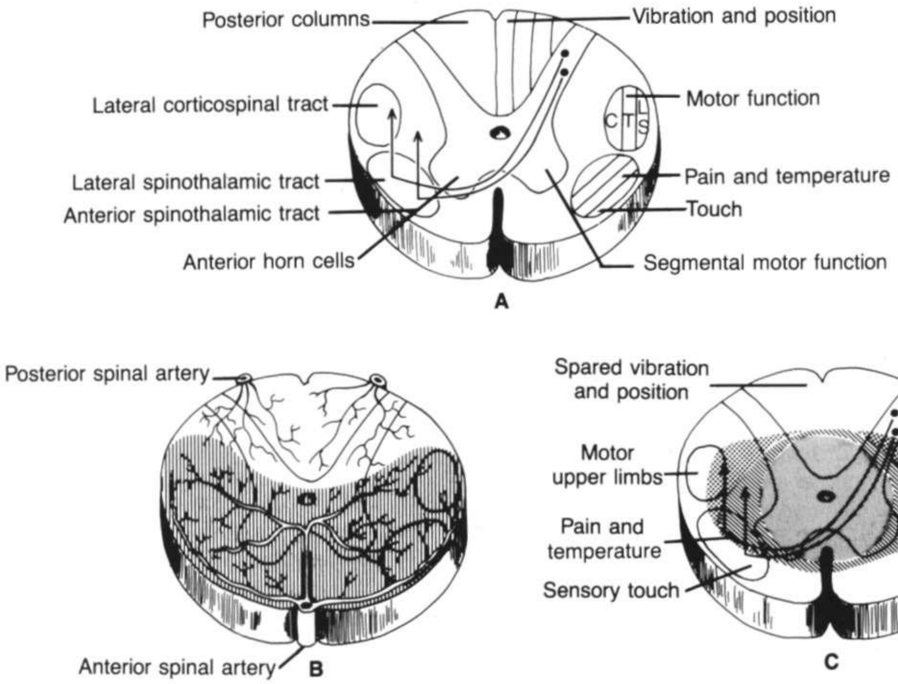

FIGURE 4-51 The pathoanatomic considerations in spinal cord trauma. **(A)** A cross-section of the cervical portion of the cord. On the left, the important anatomic structures are labeled. On the right, their functions are indicated (S, L, T, C: sacral, lumbar, thoracic, cervical). **(B)** The anterior spinal cord syndrome. The arterial supply is shown in this diagram. The anterior spinal artery supplies the shaded area, which includes the corticospinal tracts, the anterior horn cells, and the spinothalamic tracts. The posterior columns supplied by the posterior spinal arteries are not involved. Therefore, the resulting clinical neurologic picture is one of preservation of vibratory and position sense, with complete motor and sensory loss (pain, temperature, and touch) up to the level of the lesion. **(C)** The central spinal cord syndrome. The hemorrhage and edema located centrally involve the anterior horn cells and the segmental sensory fibers with sparing of the more peripheral structures. This gives a clinical picture of motor deficit in the upper limbs with a segmental sensory loss. Depending upon the degree and progression of the lesion, there may be paralytic involvement in the upper and lower limbs, with more involvement of the upper limbs. This is thought to be due to the lamination of the corticospinal tracts in which the cervical tracts are more centrally located. In addition, there may be more involvement of the lateral than of the anterior spinothalamic tracts, resulting in loss of pain and temperature sensation with preservation of touch. The different shading suggests the variable patterns that are possible.

erations leading to recognition are important. Surgery contributes nothing. The emphasis has been to protect the cord and to attempt to reduce edema. The patient should be systematically evaluated for the presence of clinical stability. If there is no stability, then after 3–4 weeks in traction, when the neurologic picture is stable, the patient may progress to a cervical brace of intermediate control. The patient may be freed of all immobilization by 12 weeks postinjury.

CLAY SHOVELER'S FRACTURE

These injuries consist of fractures of one or more of the spinous processes in the lower cervical or upper dorsal spine. They occur in individuals who have been working at some form of shoveling or similar activity. Many of the patients have been undernourished, unhealthy, out of condition, or unaccustomed to heavy labor. The classic example is that of an unhealthy or unconditioned worker who has had symptoms of pain between the shoulders for several days or weeks when he lifts a shovel full of clay. Some of the clay sticks to the shovel, and there may be the sound of a crack, followed by severe pain in the back between the shoulders. Lateral radiographs show the fracture of one or more spinous processes (Fig. 4-52).

Mechanism of Injury

Presumably, injury is due to forces transmitted to the spinous processes from the shoulder girdle by the muscles attached to them.

The trapezius, rhomboid, and the posterior serratus muscles originate in the spinous processes. The fracture can be one, sudden, staccato type of force, or it may be a fatigue type of phenomenon. Many, if not most, of these fractures occur as a result of a fatigue mechanism. Support for this is suggested by the fact that it frequently occurs in the uninitiated shoveler and is associated with a repetitive activity, usually preceded by symptoms in the lower cervical or upper dorsal spine for a period of time prior to the actual fracture.

The major vector of force of these muscles in relation to the spine is horizontal (±x-axis; Fig. 4-53). Their function during shoveling is to firmly support the shoulder girdle in relation to the spine and the thoracic cage. This allows for the most efficient transmission of force from the trunk by way of the shoulder girdle and arms to the shovel. Repeated forces are transmitted from the shovel through arms, shoulders, and the muscles to the spinous processes. These bony structures, because of the state of health or lack of biomechanical adaptation from previous loadings, may not be able to withstand the forces. Although individual loads may be within the load-bearing tolerance of the bone, repeated loads result in a type of fatigue failure.

FIGURE 4-52 Radiograph of a fracture of spinous process of C6, characteristic of the "clay shoveler's" fracture. The MIV may be a bending moment applied to the spinous processes in the horizontal (x, z) plane. The failure occurs either as an isolated episode or, more likely, through a fatigue type of mechanism. *(Courtesy of Teaching Library, Department of Diagnostic Radiology, Yale University School of Medicine.)*

Discussion

There are some additional biomechanical concepts that are related to this injury. One of the theories of injury suggested in a report by Hall is the possibility of failure due to pull through the ligamentum nu-

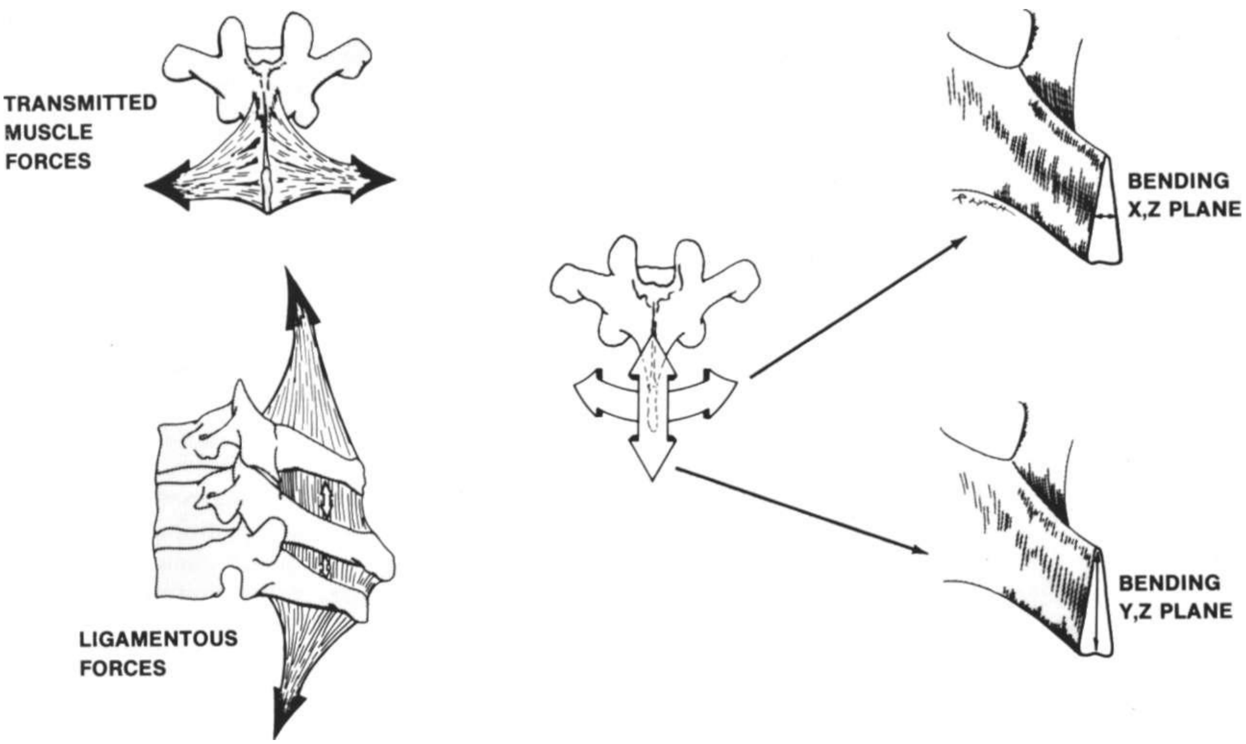

FIGURE 4-53 Representation of the mechanism of spinous process fracture in clay shoveler's injury. Large forces are transmitted to the spinous processes through the origin of the muscles that stabilize the shoulder. Relatively smaller forces are transmitted through the interspinous ligaments in flexion and extension of the spine. The spinous process is constructed such that there is a much smaller sectional (area) moment of inertia to resist bending in the x, z plane than there is in the y, z plane. Thus, the most likely mechanism of failure is through muscle forces transmitted to the spinous processes causing bending in the x, z plane. There is an additional biomechanical factor of fatigue failure operative here.

chae and the supraspinous ligaments.[104] This is not likely in view of the rather delicate, almost nonexistent nature of the ligamenta nuchae in the human and the rather thin interspinous and supraspinous ligaments in the cervical and thoracic spine.[129, 281] There are other interesting anatomic facts that relate to the mechanics of this injury. The sectional moment of inertia per area of the spinous process is greater against bending in the sagittal (y,z) plane than in the horizontal (x,z) plane. The site of the fracture within the spinous process occurs at the smallest cross-sectional area of that structure (Fig. 4-53). The observation that most of the fractures are at C7, followed by C6,[9] may well be related to the length of the spinous processes, since C7 and C6 have the two longest process. The fractures probably occur at these sites because the longer processes allow for the application of a greater bending moment.

A clay shoveler's fracture is also recognized in whiplash injury. The mechanism is supported through experimental studies. Gershon-Cohen and colleagues carried out some crude experiments on cadavers, in which acute forces simulating hyperextension and hyperflexion were found to be associated with avulsion injuries of the spinous processes. These investigators suggested that the ligamenta nuchae and the intraspinous ligaments may be responsible.[95]

Suggested Treatment

Hall, based on his experience with 13 patients, suggested *early* removal of the fractured fragment.[104] Annan and Watson-Jones were in favor of conservative treatment.[9, 278] This consisted basically of analgesics and rest, followed by gradual rehabilitation and rebuilding of muscle strength back to the level of the preinjury performance. Neck movements, flexion/extension in particular, are very painful, especially in the early convalescent phase. Presumably, this is due to motion transmitted to the fracture site through the normal musculotendinous attachments. A cervical orthosis of intermediate control with a thoracic attachment may be of considerable value for the first 3–4 weeks, followed by active exercise.

REVIEW OF SOME SPECIFIC THORACIC AND LUMBAR SPINE INJURIES

Vertebral End-Plate Fractures

There are three types: (1) those which involve the central portion of the end-plate, (2) those which are quite peripheral in the end-plate and involve the cylindrical cortical shells, and (3) fractures that produce transverse fissures extending across the entire end-plate.[198] In some instances there may be extensive disruption, with displacement of the annulus fibrosus and cartilaginous end-plate into the cancellous bone of the vertebral body. In Figure 4-54, the upper end-plate of L3 is fractured but cannot be seen on regular radiograph or on the tomograms. The L2 vertebra shows an end-plate fracture of Type 2 described above.

Mechanism of Injury

In the experimental situation there is bulging of the annular fibers and vertical displacement of an intact vertebral end-plate with a gradually applied vertical load. Failure occurs when the stress on the end-plate exceeds the maximum allowable stress. If the forces continue, the initial crack propagates in different directions and eventually fragments. As this happens, a nonfragmented cartilaginous end-plate and the annular fibers are displaced into the cancellous bone area. Vertebral end-plate fractures are due to vertical compression loads, primarily on the vertebral body. The particular configuration depends on the magnitude and direction of the injuring vector and the mechanical characteristics of the existing

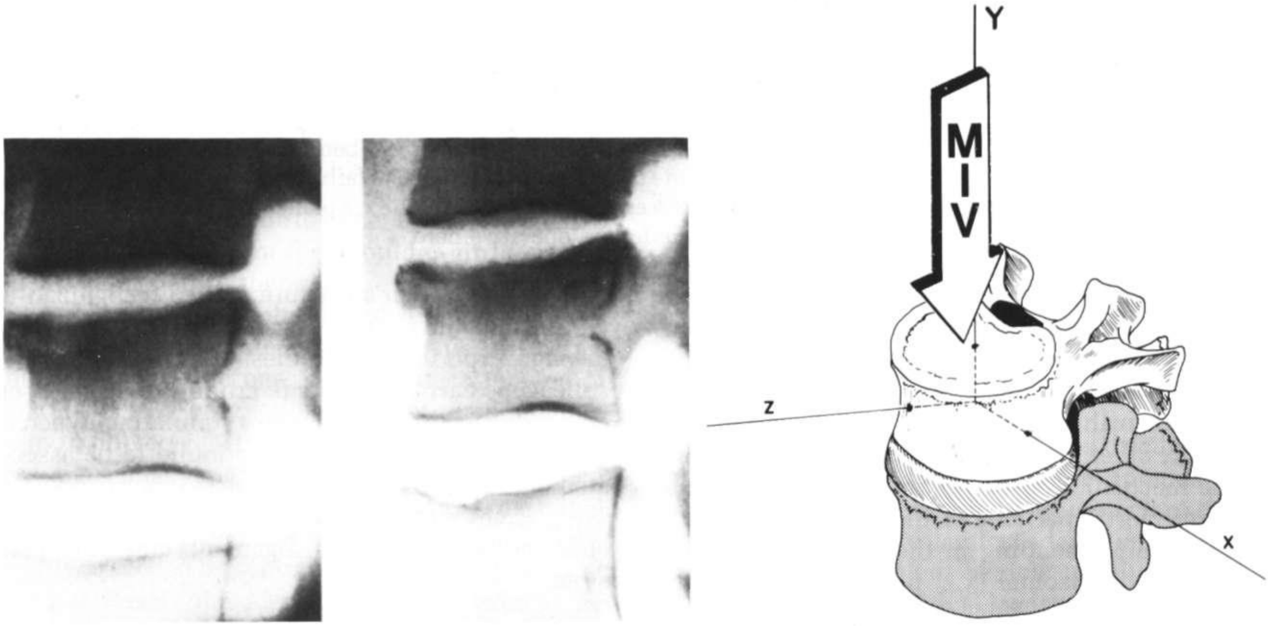

FIGURE 4-54 Laminagrams showing two vertebral end-plate fractures. The middle vertebral body L2 shows a fracture of its upper end-plate involving the cortical shell. The vertebra beneath it shows a fracture of the neutral portion of the end-plate. This was not apparent on the plane radiograph nor on the adjacent laminagrams, 0.5 cm away. These fractures frequently go undiagnosed. The MIV is shown on the inset as a force of moderate or severe magnitude along the −y-axis. *(Radiographs from Perey, O.: Fracture of the vertebral end-plate in the lumbar spine—an experimental biomechanical investigation. Acta Orthop. Scand., 25 [Suppl.], 1957.)*

anatomy in the particular vertebra and individual. It has been shown that the load-bearing capacity varies with age. A group of vertebrae loaded in individuals over 60 failed at an average load of 4200 N (950 lbf), whereas those under 40 sustained loads of up to 7600 N (1710 lbf).[198]

Discussion

These fractures frequently go undiagnosed. Laminagrams show a minimally displaced vertebral end-plate fracture that is not observed on routine radiographs. However, even this technique is unlikely to show an undisplaced fracture, unless healing has progressed with some visible callous formation. The authors make the following recommendation. When diagnosis is important for legal or medical reasons, patients with injuries presumed to result from vertical compression loads should have postinjury laminagrams followed by subsequent studies in 3–4 weeks.

Rolander and Blair have observed that even in normal nontraumatic lumbar spine mechanics there can be deformation or even fracture of the vertebral end-plate.[215] Perey found in experimental compressive loading of functional spinal units that end-plate fractures were relatively more frequent than compression fractures.[198] We submit that in vertical loading the end-plate is most probably the first osseous structure to fail.

Suggested Treatment

The treatment is aimed at the relief of symptoms. Bed rest with analgesics is employed until the acute symptoms subside and the patient is able to ambulate. An appropriate spinal orthosis of minimum or intermediate control may be employed in the convalescent period if the symptoms so indicate.

Ejection Seat Injuries

Compression fractures and vertebral end-plate fractures of the spine often occur when pilots must be catapulted from aircraft in dangerous situations.

Mechanism of Injury

The mechanism of this injury is the tremendous force that is applied to the spine in the sitting position as a result of acceleration of the seat up and out of the aircraft. The determination of injury and the extent of the same depends upon a number of variables. Some of the important variables are as follows: the magnitude of the acceleration and its rise time, the stiffness and damping quality of the base of the seat, the design of the seat, the spacial orientation of the vertebral column (*i.e.*, flexed, extended, or straight, the latter being the most desirable), and the training and the readiness of the pilot.

Discussion

Because these injuries are intimately associated with modern technological martial arts, a good deal of investigation has gone into their prevention. It has been suggested that the seatback contour be designed to maintain, as closely as possible, a normal vertebral alignment, while others consider a straight spine to be the most desirable. Ideally, head flexion should not be induced at the time of ejection. The recommended trunk–thigh angle is 135°. It has also been suggested that a rocket-propelled ejection system might be safer, since there would be less *impact* at the time of the initial acceleration. The usual mechanism is acceleration imparted to the seated pilot through the explosion of gunpowder charges. Laurell and Nachemson reported that accelerations up to 20 g acting on the long axis of the spine could be safely tolerated provided that the rate of increase of acceleration does not exceed 300 g per second.[141] In the report by Chubb and colleagues,[41] it was indicated that sitting in the erect position with hips and head firmly against the seat was the most significant factor in prevention of compression fractures.

The middle portion of the thoracic spine was found to be the most frequently involved in the report by Hirsch and Nachemson. Generally, these compression fractures did not cause neurologic disorders. The pilots were able to return to active service as early as 2 months following injury.[115]

Treatment

These fractures may be treated according to the usual guidelines for the management of compression fractures (outlined below).

Compression Fractures

These fractures occur most frequently in the lower thoracic and upper lumbar regions. Generally they show varied amounts of wedging of the vertebral bodies. Such fractures may be associated with an end-plate failure, which may go unnoticed in the presence of the more obvious wedging and comminution.

Mechanism of Injury

This fracture is most probably caused by a vertical force exerted either as an axially directed vector (±y-axis) or by a moment about the x-axis, or a combination of the two. These concepts are demonstrated diagrammatically in Figure 4-55. The effect of these loads on the middle vertebrae is shown. The dot represents the assumed position of the axis of rotation at this level. Therefore, when an axially directed compressive force is applied directly in line with the axis of rotation, the result is a direct compression of the vertebra with end-plate fractures. However, when the compressive force is anterior to the axis of rotation, the vertebra in question is subjected to a compressive force and a bending moment. This causes wedging of the vertebra, as shown in Figure 4-56.

Discussion

It has been pointed out that in pure vertical loading of the vertebral functional spinal unit the posterior elements bear a significant portion of the load. Thus, one would expect to see fractures or disruptions of the posterior elements associated with vertebral body fractures in those compression fractures in which vertical loading is the main component. Ewing and colleagues have shown that the fracture threshold can be raised from 10 to 18 g in embalmed human cadavers by moderately hyperextending the vertebral column.[78] This points out the role of the facets in resisting compressive loads (see Chap. 1) and leads to the recommendation that in the presence of a predominantly vertical compression fracture, one should make a careful radiologic evaluation of the posterior elements, where additional fractures are likely to be found.

Most vertebral body comminutions come from some combination of vertical forces and bending moments. The end-plate fracture is a result of relatively pure vertical loading, and a fracture with extensive compression and anterior wedging, as shown in Figure 4-56, is a combination of both types of loading with high magnitude.

McSweeney has noted several cases in which he has observed cervicodorsal injuries associated with sternal fracture dislocation.* There have been

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* Personal communication, Mr. T. McSweeney.

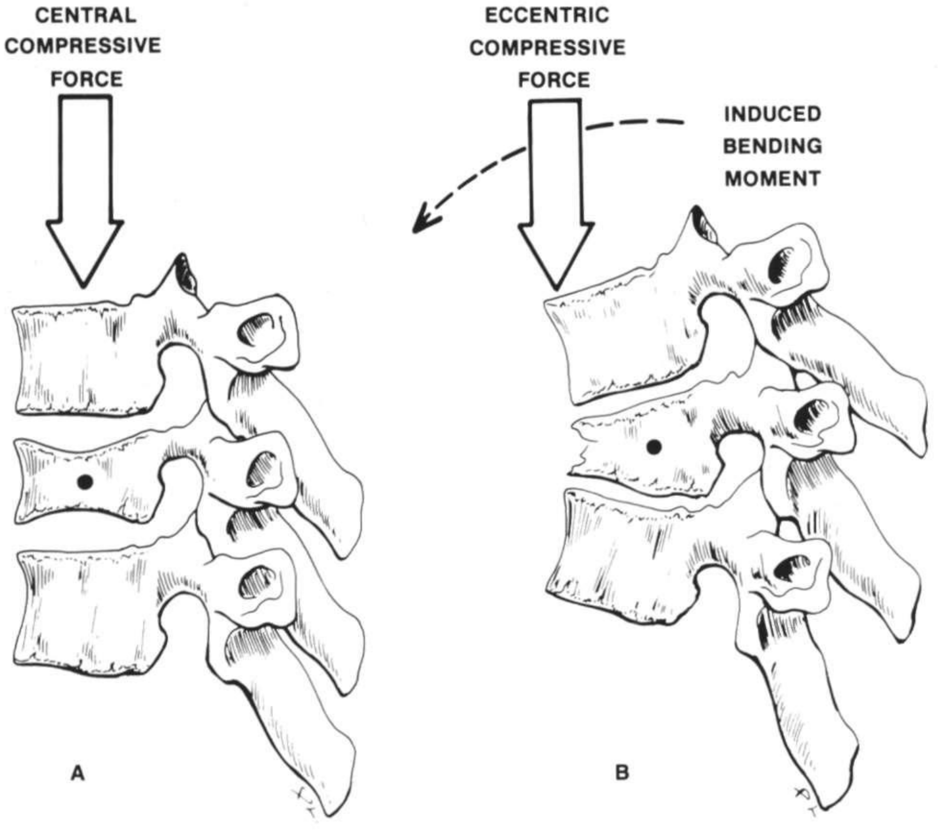

FIGURE 4-55 Different configurations of fractures produced by compression. **(A)** A centrally located axial compressive force close to the neutral axis (represented by the black dot) produces biconcave deformities of the end-plates. **(B)** An eccentrically located force away from the neutral axis results in a greater bending moment and produces a compressive fracture of the body, with characteristic wedging.

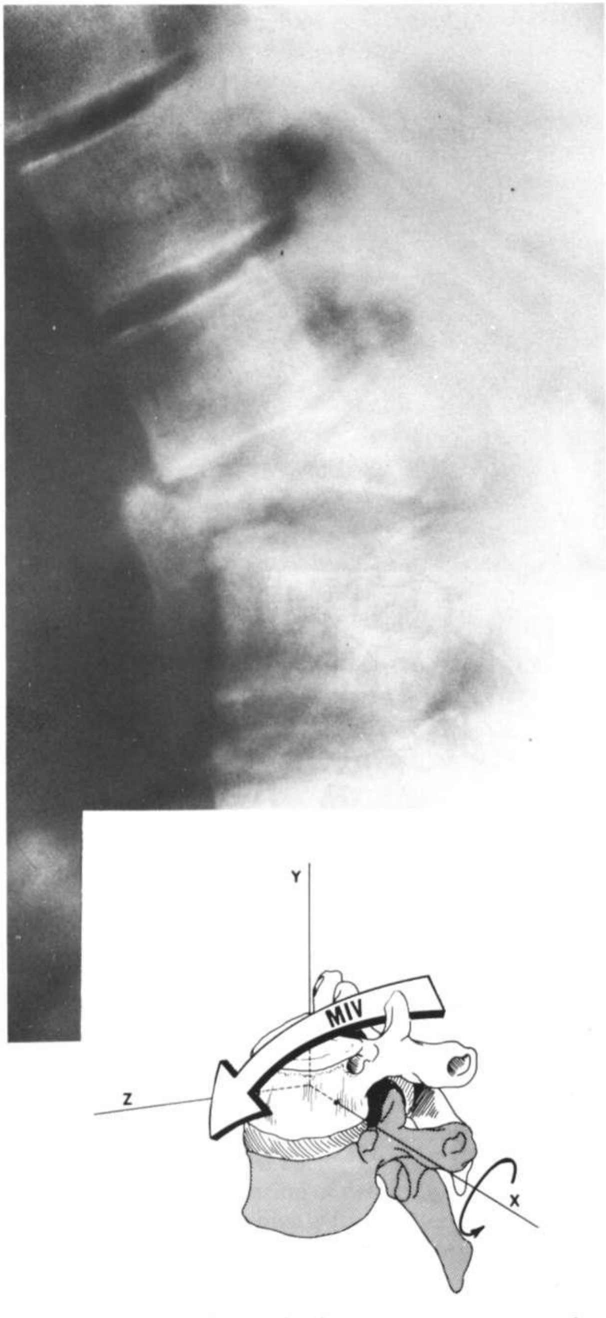

FIGURE 4-56 Radiograph of a severe compression fracture of D7, which resulted in paraplegia with a complete block, which was shown with a myelogram. The MIV shown is vertical load, due largely to torques about the x-axis. This is evidenced by the anterior wedging. Also, there is considerable vertical loading, which has resulted in the central compression. This is a "burst" type fracture, and, as reported by Bohlman,[26] in this region, is generally associated with complete paralysis. *(Courtesy of Teaching Library, Department of Diagnostic Radiology, Yale University School of Medicine.)*

wedge compression fractures that were sometimes not readily recognized because of their location in the low cervical or very high thoracic areas. We suggest that these injuries have a large bending moment involved, and the point of application of the force is well anterior to the central axis. There is a large vertical compressive force anterior and in the region of the manubriosternal joint. There is wedging due to the anterior forces and fractures of the spinous processes due to the large tensile forces exerted posteriorly (Fig. 4-57). The processes in the lower cervical and upper thoracic spine are long; thus, large bending moments are applied to them. There is an important clinical point to be learned here. In the presence of a manubriosternal dislocation, the physician should look carefully for injuries of the cervicodorsal junction, and the converse is also true.

Plaue pointed out that much of the experimental work on the failure of vertebrae in compression has not adequately emphasized the sequence of events after the initial failure. In an experimental study of autopsy material, he noted that vertebrae, after compression fracture, maintained 60–70% of their original load-bearing capacity. Moreover, vertebrae compressed to one-half the original height began to approach their *original* load-bearing capacity.[199] This is presumably due to an increase in load bearing as a result of a "new," impacted vertebra. Thus, consideration should be given to the complex and controversial question of the postfracture management of these patients. If this phenomenon is operational in the clinically compressed vertebra, then it may well be the case that any observed additional angulation of the spine that occurs during the treatment is due to longitudinal strain of the posterior soft tissues as a result of tensile loads applied rather than further compression of the fractured vertebra (Fig. 4-58). Narrowing of intervertebral discs in the kyphotic curve and excessive fragmentation of the vertebral body may also be contributing factors in the progression of deformity.

Suggested Treatment

The treatment of these injuries is a controversial subject, and the guidelines are not well delineated. The patients should be systematically evaluated for clinical instability. Severe pain and slow healing have been reported in unreduced fractures. Certainly, the presence of an unsightly gibbus is not desirable. In fractures with unsightly deformity,

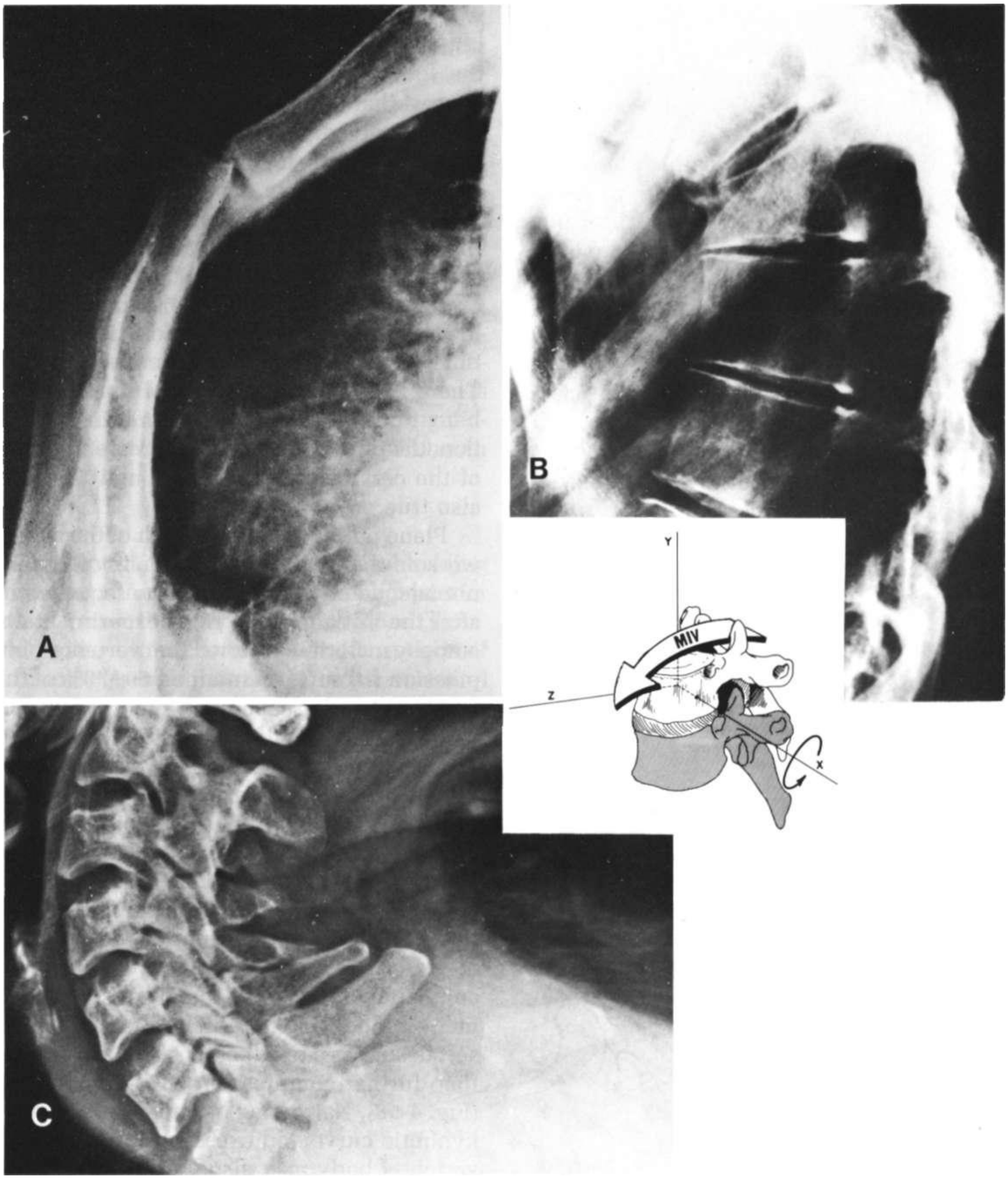

FIGURE 4-57 Lateral radiograph of the sternum. **(A)** There is a fracture dislocation of the manubriosternal joint. The patient also had a pneumothorax associated with a fracture of the second rib. **(B)** Lateral radiograph of the dorsolumbar spine shows wedge compression fractures of the fifth and sixth thoracic vertebrae, resulting in a 90° kyphosis. This is due to a large +x-axis bending moment. **(C)** A view of the cervicodorsal junction shows that the spinous processes of the three upper thoracic vertebrae have been completely avulsed. This is due to tensile loading associated with the MIV. Cervical lordosis is exaggerated because of the thoracic kyphosis. *(Radiographs courtesy of Mr. T. McSweeney, The Robert Jones & Agnes Hunt Orthopaedic Hospital, Oswestry, U.K.)*

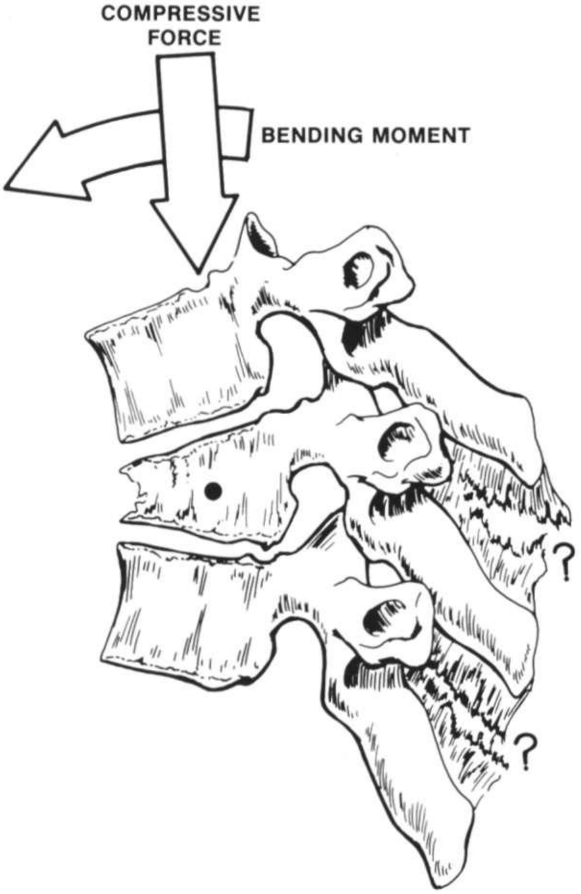

FIGURE 4-58 Following a fracture from any combination of compression and bending moment, the tendency for subsequent clinical angulation and kyphosis deformity depends largely on what happened to the posterior elements at the time of injury. If they are intact, the ability of the compressed vertebral body to withstand the loads most probably is great enough so that there will not be extensive clinical progression. On the other hand, if the posterior ligaments have been damaged, then the possibility of clinical progression of deformity or instability is significantly greater. A highly fragmented vertebral body fracture may also be likely to allow progressive kyphosis.

treatment with reduction and a hyperextension cast is desirable. Other fractures may be treated with bed rest, exercises, and careful follow-up with radiographs.

Patients with mild compression loss of one-third or less of presumed original height of one or more vertebrae can be treated with active exercise and mobilization after a period of bed rest to allow acute symptoms to subside and to permit any slow elastic recoil. Kazarian and Graves showed in the laboratory that at low strain rates trabeculae may bend and subsequently unfold and recover part of their original height. Specifically, vertebrae compressed to 50% of their original height will gain up to 72% of their original height in the first 72 hours.[134] There is then good biomechanical reason for 3–4 days of bed rest. We are not aware that this has been clinically documented. Based on the work of Plaue,[199] this should be satisfactory. Patients with more than this degree of compression are probably best treated with reduction and application of a well-molded plaster jacket. Even though there is considerable disagreement in the literature, Westerborn and Olsson have shown that in some wedged compression fractures it is possible to reposition the vertebral body, regain a large portion of its original height, and maintain the improvement for 5 years following reduction.[280] The duration of immobilization should be 3–6 months, depending on the severity of the injury. We recommend mobilizing these patients sooner under careful observation for evidence of progression of deformity.

The Burst Fracture (Comminuted Vertebral Body Fracture with Bone Fragments in the Canal)

Comminuted fractures of the vertebral body have come to be called "burst" fractures. These are often associated with bone fragments in the canal and varying degrees of neurologic deficit. With the more widespread availability of CT scanning and instrumentation, these fractures have received considerable attention.*

Mechanism of Injury

Because of the comminution, these are presumed to be high-energy impact fractures, and the configuration of the failed vertebra implies the vertical compressive loading pattern depicted in Figures 4-54 and 4-55A.

Discussion

These injuries generate two basic questions: (1) Should the canal be decompressed, and if so, how (*i.e.*, anteriorly or posteriorly)? and (2) With or with-

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* References 24, 25, 50, 57, 90, 124, 135, 147, 156, 157, 158, 181, 241, 256–258, 289.

out decompression, is the spine unstable, and if so, how should it be stabilized?

The nonsurgical approach, reduction with application of cast, has been evaluated and reported as effective as surgery.[50,289] The series was carefully compared with the experience of Dickson and coworkers, who used Harrington instrumentation.[57] The neural recovery rate was the same in the nonsurgically treated group. The hospitalization time was less in the surgically treated group.

In regard to surgery and conservative treatment, the work of Keene and colleagues is relevant. This study observed that 5 of 14 patients with thoracolumbar fractures initially had normal rectal tone and sacral sensation but were subsequently found to have urinary dysfunction involving hypotonic bladders documented by cystometrograms.[135] The presumption is that chronic instability can cause recurrent cord injury and persistent bladder dysfunction. This study implies that any comparison of operative and nonoperative treatment of these burst fractures should include appropriate monitoring of the urologic as well as neurologic status.

We offer a theoretical formula that may be a useful concept in describing the neurologic deficit in these fractures—cervical spine fractures as well as cervical spondylotic myelopathy:

$$\mathrm{UND} = \mathrm{II} + (\mathrm{CC} + \mathrm{M}) \times \mathrm{T}$$

UND = Ultimate neurological deficit
II = Initial impulse to neural elements
CC = Canal compromise (involves physical pressure and vascular compromise)
M = Motion
T = Time

We submit that analyzing and monitoring the clinical situation in this theoretical construct will help in decision making about management of these patients with vertebral body fracture and neurologic compromise. If the patient is not definitely improving, then the ultimate neurologic deficit (UND) can be reduced by decreasing canal compromise (CC) and the motion (M) with surgical decompression and fusion.

There are several surgical solutions to the problem. In the most complex burst fracture there is bone in the canal, an associated neurologic deficit, and instability either initially or following decompression. There are numerous approaches: (1) laminectomy, (2) reduction with Harrington rods, relying on an intact posterior longitudinal ligament to push the fragments out of the canal and back into the vertebral body,[57] (3) decompression of the canal through an anterior approach,[156,158] and (4) decompression posteriorly through the pedicle or posterolaterally with these approaches the fragment is directly visualized and pushed out of the canal and back into the vertebral body. There are other important considerations, including the choice of implants and the number of vertebrae to be involved in the fusion.[124]

Most of these considerations are thoroughly reviewed by Stauffer[241] and Bohlman.[24] McAfee and co-workers suggested that the choice of instrumentation be based on the character of injury due to a presumed mechanism of injury. With a primary compression injury, Harrington distraction rods are used; with a primary distraction injury, Harrington compression rods are used. In injuries where the major displacement is translation, segmental spinal instrumentation is employed.[157] This approach can be helpful but may be a bit more refined than is absolutely necessary. Certainly, one must avoid overdistracting an injury in which the major ligamentous continuity of the spine has been disrupted (see Fig. 8-66).

Most likely because of the relationship of the relative vascularity of and space available to the spinal cord in the upper portions of the thoracic spine (Fig. 6-25), there is a higher portion of complete paralysis and partial neurologic deficit. Bohlman documented the incidence of partial and complete paralysis associated with injuries in this region.[25] There was no recovery of neurologic function no matter how these injuries were treated. Laminectomy was deemed to be contraindicated, and anterior transthoracic decompression and fusion was thought to offer the best chance for recovery of neurologic function.[25,124]

There is not, in our view, a distinctly superior surgical solution. It is not yet known for certain exactly what is best as regards when, where, and how to decompress; what levels to arthrodese; and what levels, if any, beyond those should be permanently or temporarily immobilized with internal fixation. Nevertheless, given the current level of uncertainty, certain decisions must be made. We address here the problem of a burst fracture with or without neurologic deficit and imaging evidence of bone fragment(s) in the neural canal.

Suggestions for Treatment

The surgical solution has an advantage at least for psychological rehabilitation and socioeconomic reasons.[124,181] Harrington instrumentation attached to the second vertebra above and one vertebra below the fracture is recommended in association with a bilateral posterolateral fusion. Other posterior instrumentation methods are reasonable alternatives. If there is no improvement in canal space available and the neurologic deficit persists, then anterior decompression and bone grafting is suggested.[24, 156]

This addresses the fracture fragment in the canal with a neurologic deficit. The criteria for stability (pp. 338 and 351) can be applied to determine the status of stability for any of these fractures with or without a decompression procedure.

The following is a review of the various considerations and some suggestions which are thought to involve the fewest risks and most benefits to the patient. The authors have not recommended some of the newer techniques. This is not because of disagreement or disapproval, but because the incremental risks have not yet been shown to be outweighed by distinct benefits. This may change in the future. New instrumentation is discussed in Chapter 8. It is appropriate to recognize that virtually all of these injuries can be treated nonsurgically.[50]

We have chosen to use the term *neurologic deficit* here to mean partial neurologic deficit. Certainly, there is argument against any attempt to decompress a complete deficit. Judgment must be exercised. In cauda equina defects there is almost always some reasonable chance for recovery.

When there is no neurologic deficit, the patient can simply be evaluated for clinical stability (see pp. 338 and 351) and managed accordingly. If unstable, fuse; if not, treat with an orthosis and get the patient moving. Care should be taken not to overdistract, and the strategic use of compression rods can be considered in grossly unstable situations in which there is no ligamentous continuity.

The fusion length suggested is one or two levels above and below the injured vertebra or vertebrae. The problem of which levels to fuse and the placement of rods remains controversial and unresolved. We recognize the potential liability in the rod-long, fuse-short principle because of the possibility of damage to the cartilage of immobilized facet joints. We embrace the concept of fusing the least amount of the functional spine that will maintain clinical stability in both the short-term and long-term postoperative period. In regard to the long term, at the thoracolumbar junction and lumbar spine, one level above and below should suffice. Moving up in the lumbar spine, fusion of two above and below is indicated. With greater anterior vertebral body wedging (*i.e.*, anterior cortex equal to or less than one-third the posterior cortex), fusion of two above and two below the injured vertebrae is recommended.

In regard to short-term stability, if the fracture is grossly unstable and distraction rods alone are being used, one above and below for anchoring is desirable. If compression rods, locking hooks, or posterior element wiring is involved, the vertebral attachment for anchoring is not needed.

If there is imaging evidence of canal compromise, one approach is to employ Harrington distraction instrumentation and bilateral posterior lateral fusion. This requires a treatment opportunity within approximately 10 days of injury. The goal is to reduce the posteriorly protruded material in the canal by axial (y-axis) traction on an intact posterior longitudinal ligament. If this results in partial or complete decompression of the spinal canal and neurologic recovery, success is achieved. The canal is decompressed, any clinical instability present is treated by the fusion, and the patient needs only follow-up and rehabilitation.

If there is no satisfactory improvement in neurologic status and the fragment(s) in the canal is not completely cleaned, then an anterior lateral decompression by partial vertebral body resection (VBR) and interbody fusion (plus or minus anterior instrumentation) is indicated. The patient is then followed and rehabilitated. Surgical teams facile and highly experienced with the anterior VBR and decompression may elect to skip the posterior distraction decompression option. The posterior approach may prove inadequate in 30–40% of patients; however, the anterior decompression is more difficult and has more complications. A current and future goal is to work toward a more universal system that, through one surgical approach, can reliably address both the decompression and the instability problem.

We have begun to treat patients with canal compromise and neurologic deficit with transpedicular decompression, Harrington distraction rods, and Drummond wires. We fuse one or two levels above and below the fracture, and we seek to keep the fusion and instrumentation levels the same. Currently, this is our best advice for maximizing the risk–benefit ratio for these patients.

The Crush Cleavage Fracture

This is an interesting fracture because it has been described relatively recently and also because it may be analogous to the recently updated description of the so-called *tear drop* fracture of the cervical spine (see p. 173). The mechanism of injury is likely to be a high-energy impact with a significant vertical load and bending moment. Lindahl and colleagues described 14 patients with this injury. The upper half of the vertebral body is *crushed* and the lower half has a sagittal plane *cleavage* fracture. There were bone fragments in the canal, associated lamina fractures, and a neurologic deficit in 5 of the 14. This fracture may be evaluated and managed essentially as the burst fracture, except that it is good to keep in mind that this is more likely to be unstable.

The Burst Fracture of the Fifth Lumbar Vertebra

This is a somewhat unique variant of the burst fracture, mainly because of its location and the probability of its being satisfactorily treated with posterior decompression and fusion. The MOI, like the others, is thought to be a high-energy compression load mainly along the vertical axis (y-axis). There is a tendency for these to be associated with calcaneal fractures. This first publication[90] is based on four patients. Future studies will, no doubt, substantiate these current recommendations and/or improve knowledge and treatment of the injury.

Lateral Wedge Fractures

Such fractures are virtually the same as compression fractures, except that the component force vector is such that there is asymmetrical collapse of the vertebra about the sagittal plane. Thus, there is wedging in the frontal (y,x) plane.

Mechanism of Injury

The injuries occur, as in the cervical spine, with a force that causes severe lateral bending ($\pm \theta z$) associated with some flexion. There may be unilateral injury of the articular facet, the pedicle, or the lamina on the concave side, and fractures of the transverse processes on the convex side (Fig. 4-59). Fractures of the transverse process are due to tension in the intertransverse ligaments. The bone has failed in tension at a lower threshold than the ligamentous structures; therefore, instead of rupture of the intertransverse ligaments, the transverse processes have fractured.

Discussion

Nicoll found these fractures to be clinically different from anterior wedge compression fractures in both treatment and outcome. In his overall series he reports complete recovery in 40% of the anterior wedge compression fractures but only 21% of the lateral wedge fractures. He indicated several important clinical points about this type of fracture. The residual pain in lateral wedge fractures is greater. This may be related to the relative amount of soft-tissue damage in the two types of fracture. Sometimes there is iliopsoas injury, which may be accompanied by significant retroperitoneal hemorrhage and a clinical picture of an "acute abdomen." Two patients in Nicoll's series had a laparotomy with negative findings before injury to the vertebral complex was diagnosed.[174]

Suggested Treatment

The guidelines suggested for treatment of anterior wedge compression fractures are also recommended for lateral wedge fractures.

Gross Fracture Dislocations of the Thoracic and Lumbar Spine

This group of injuries is separated from end-plate fractures, compression fractures, lateral wedge fractures, and posterior element fractures. Any combination can occur in association with a dislocation. There are distinguishing mechanical characteristics, such as causative factors, damage incurred, and treatment. This section discusses trauma of the thoracic and lumbar spine that involves dislocation and possible neurologic involvement of the cord or the cauda equina.

Mechanism of Injury

The intrinsic anatomic structure of the spine in these regions is quite stable; thus, large forces are required to cause fractures and dislocations. The strength and stability of this area is related to several biomechanical factors. As a result of the size of the disc and the vertebra in this region, there is a large component of direct, osseous, annular fiber attachment to the periphery of the end-plate. In this region, the anterior and posterior longitudinal ligaments are strong and well developed.[260] The posterior elements include rigid osseous stability through the spatial orientation of the facet joints

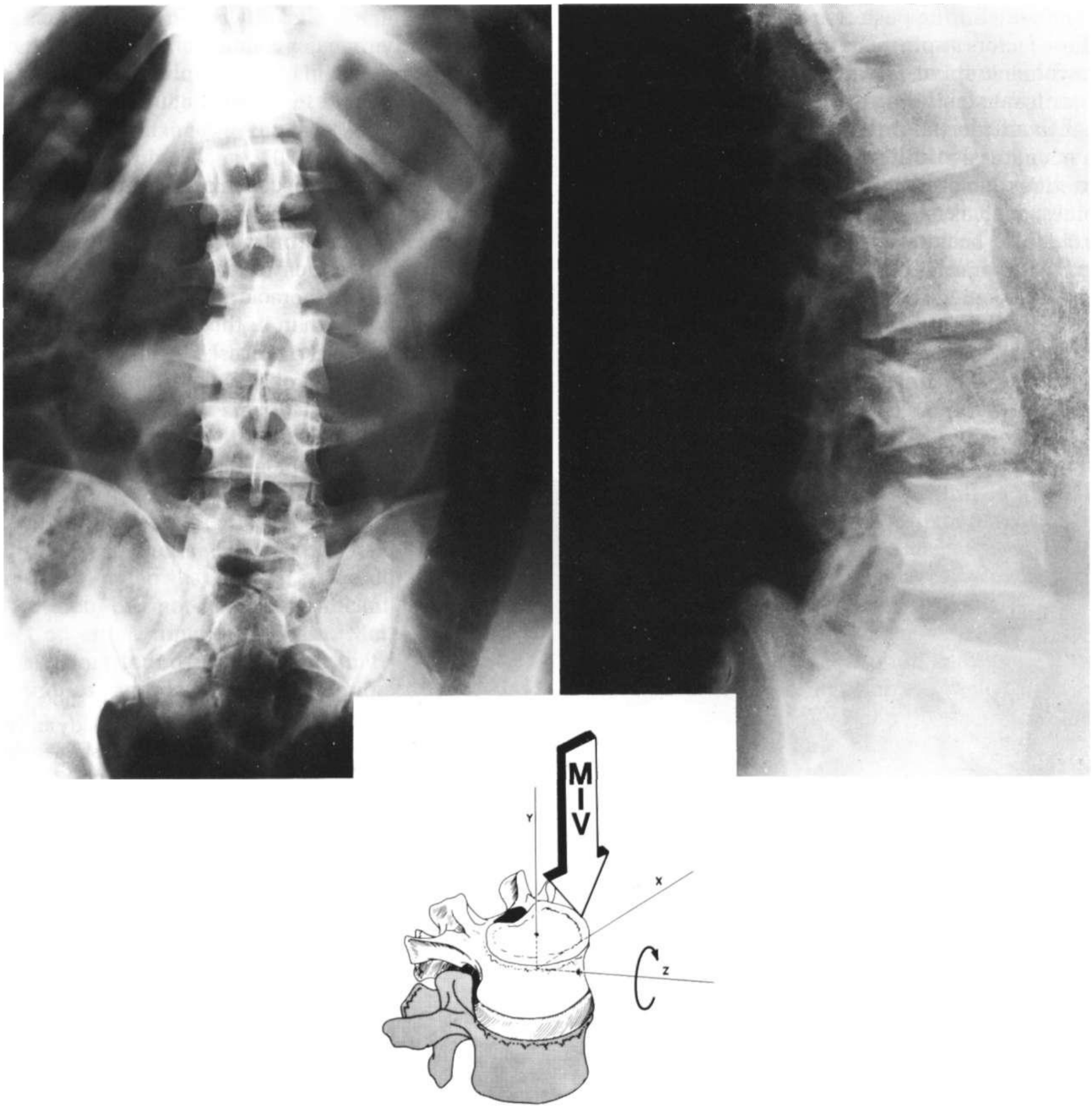

FIGURE 4-59 Radiographs of a patient who sustained a high-magnitude lateral wedge injury. There is asymmetrical compression of the left side of the third lumbar vertebra. There is slight compression of the pedicle and articular facet, indicating compressive loading. The right transverse process of L3 is fractured because of tensile loading on the opposite side. These factors show that the spine has been subjected to a left ($-$z-axis) bending moment in the coronal plane. Just to the right of the base of the spinous process of L3 there is a displaced fracture of the lamina. This may be a shear fracture; in a beam subjected to bending, there is maximum shear stress around the neutral axis. The radiograph on the right (the lateral view) shows the vertebral end-plate compression fractures.

along with strong posterior ligamentous structures. These factors apply more to the lumbar spine than to the thoracic spine. However, the thoracic spine has a significant stabilizing influence through the attachment and inertia of the rib cage. Since there is so much natural stability in these regions, fracture dislocations generally are associated with very high forces, such as falls from considerable heights, auto accidents, heavy weights falling on the back, or "cave-in" injuries of the type sustained by coal miners. Pure flexion is usually not enough to rupture the posterior ligaments and facet complexes. In order to disrupt these structures and cause dislocation, an element of shear loading is usually necessary in addition to the large normal loads that are required. Thus, the fracture dislocations in these regions generally involve a major element of axial rotation, flexion, and some primary or coupled lateral bending. An example is presented in Figure 4-60.

Discussion

Most of the injuries occur in or near the thoracolumbar junction.[102] This may be related to a phenomenon of stress concentration due to several mechanical differences in the two regions. There is generally less motion in the thoracic spine,[281] and the stiffness coefficient is less than in the lumbar spine[185, 186] for most of the motion patterns. The attachment of the ribs in the region probably contributes to the stiffness and certainly adds to its inertia. Finally, the

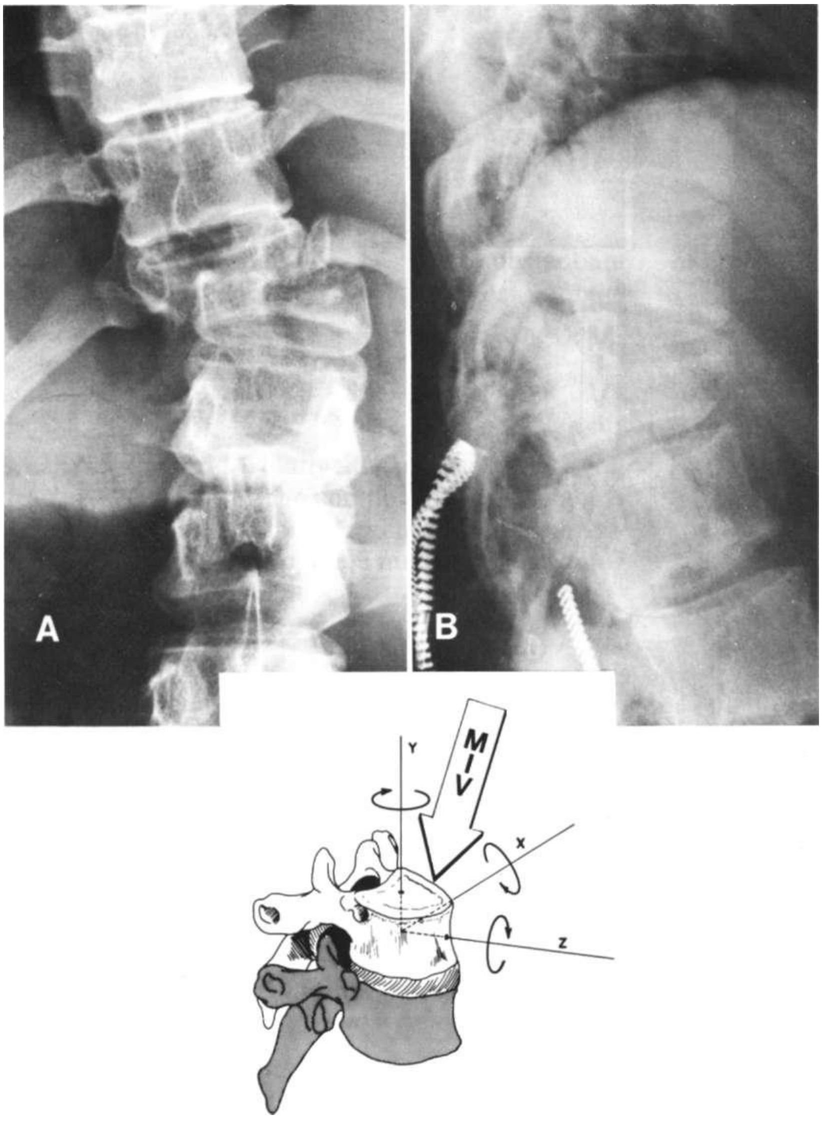

FIGURE 4-60 Radiographs show a displaced fracture through the body and posterior elements of D12 sustained by a young woman in an automobile accident. **(A)** Malalignment of the spinous processes above and below the fracture site indicate abnormal axial rotation in the negative direction about the y-axis. There are fractures of both transverse processes of L1 as well as the right transverse process of L2. The thoracic spine above the fracture is tilted to the left (−z-axis rotation). **(B)** This lateral view shows compression of the anterior portion of the vertebra, with the thoracic spine tilted forward (+x-axis rotation). Fracture of the pedicle is also shown here. The MIV is shown in the inset. There is an element of torque about all three axes (−y-axis, −z-axis, and +x-axis). There was a high magnitude of force with a significant element of torsional shear involved at the fracture site. This type of complex mechanism of injury is best explained in the helical axis of motion, a concept that is more complicated than the MIV and awaits a better understanding of the biomechanics of spine trauma. *(Courtesy of Teaching Library, Department of Diagnostic Radiology, Yale University School of Medicine.)*

patterns of movement are different in the two areas. There is a rather abrupt change in the physiologic range of axial rotation ($\pm \theta y$). This change occurs at the level where the facet joint orientation of the thoracic spine changes to the facet joint orientation of the lumbar spine (see p. 106). The anatomic level of that rather abrupt change varies among individuals, from T9 to L1. These anatomic facts are no doubt contributing factors to the tendency for a higher incidence of fracture at these levels.

Suggested Treatment

The treatment of these fractures is complex, complicated, and controversial.

Bohler, Davis, Key and Conwell, and Watson-Jones support the view that *anatomic reduction* is important in the management of these fractures.[22, 51, 52, 138, 278] On the other side of this controversy, Leidholdt and colleagues, in a retrospective study of 204 patients with fracture dislocations and paraplegia, noted that sagittal plane deformities of 30° or more did not have a significant deleterious effect on either function or pain. Moreover, these investigators found no evidence that these major deformities caused any further deterioration of neurologic function where incomplete neurologic lesions were present.[144] Nicoll and Westerborn and Olsson have not emphasized the anatomic reduction but rather the importance of determining stability and treating the patient according to that determination.[174, 280] Nicoll considered early ambulation, exercises, and activity to be crucial in the successful rehabilitation of his patients. Westerborn and Olsson treated some with hyperextension and plaster jackets, but in other instances they elected posterior fusion.

The topic of clinical stability is dealt with in detail in Chapter 5. Most writers agree that when there is neurologic involvement there is usually instability. Nicoll considered a stable fracture to be one in which the interspinous ligaments, the facet joint complex, and the vertebral discs are intact. He suggested that unstable fracture dislocations should be treated in a plaster cast, with the spine in neutral position.[174] Westerborn and Olsson treated some in hyperextension plaster jackets and others with posterior fusion.[280] Holdsworth suggests reduction and anterior or posterior fusion. In instances where the site of the fracture dislocation is through the anterior and posterior elements of the bony structure of the vertebra, he suggests conservative treatment of this most unstable injury. This recommendation was based on his expectation that stability is achieved with healing through osseous unions.[119]

Rules and guidelines about the treatment of these injuries are limited both by the clinical uniqueness of each injury and by the present state of knowledge. The following approach is our synthesis of current information, with anticipation of more definitive information in the future. We suggest that the guidelines for the evaluation and treatment of clinical instabilities presented on p. 351 be followed. When there is a fracture dislocation with no neurologic involvement, one reasonable attempt at closed reduction with skeletal traction should be made. If this is successful, an appropriate orthosis of maximum control for the region involved is applied. The vigorous orthotic treatment is to avoid redislocation. If the closed reduction is not successful, an appropriate orthosis with intermediate control is applied. The orthosis, in either case, is worn for 6 weeks with the patient in bed. During the next 6 weeks, the patient is ambulated with the device. Following this, activities continue without the orthosis and progress toward normality. Flexion/extension radiographs and clinical follow-up are used to monitor the progress. If disabling pain, neurologic symptoms, or abnormal motion should occur, then an appropriate fusion should be considered.

For protection of residual and recoverable neurologic function, fusion is generally required in the unstable fracture dislocations. Generally, some simple posterior technique suffices, except in cases where there is marked kyphosis or following extensive laminectomies. In these instances or in cases where anterior decompression is indicated, the anterior approach with interbody fusion is considered. Kelly and Whitesides and others have advised osseous stabilization of these injuries for better overall rehabilitation.[137]

Isolated Posterior Element (Neural Arch) Fractures

These fractures include any and all fractures posterior to the posterior longitudinal ligaments.

Mechanism of Injury

Such fractures are thought to occur with a flexion and axial rotation type of injury. It is also probable that vertical loading with the spine in extension can cause fractures of the posterior elements. When a large torque is applied to the spine, with a spatial orientation of the lumbar vertebra, there may be damage in this region. This is because the facets in the lumbar region are aligned primarily in the sagittal plane, and significant y-axis torques will fracture them.[G] Axial (θy) rotation of a lumbar functional spinal unit of greater than 3° can cause either intra-articular damage or posterior element fracture (see Figs. 2-22 and 5-54).

Discussion

These fractures are frequently overlooked; thus, careful examination and good radiographic techniques are required to make the diagnosis. Computerized tomography is often useful in discovering these fractures. They may also be associated with fractures of the transverse processes. This should alert the physician to look more carefully at the posterior elements at or near the level of a transverse process fracture.

Suggested Treatment

It has been suggested that above L4 the injuries may be treated functionally.[174] Patients are given bed rest until acute symptoms subside and are then mobilized in a spinal orthosis of minimal control for 6–8 weeks. Following this, they may gradually resume normal activity as tolerated.

When there is a lesion at the fourth or fifth lumbar level, the treatment depends on the presence or absence of spondylolisthesis. If there is clinical instability, then spine fusion to the normal adjacent vertebra above and below should be carried out. If there is no evidence of instability, then functional treatment of 6 weeks' bed rest followed by gradual mobilization in an orthosis of intermediate control should suffice. In some situations, intractable chronic back pain may develop. This should be treated with bilateral posterior lateral fusion, uniting the normal vertebrae cephalad and caudad to the spondylolisthesis.

Lap Belt Injuries

This type of fracture was first described in 1948 by Chance, purposefully.[39] These injuries occur in the region of the upper lumbar spine. They may be purely ligamentous or they may consist of fractures or fracture dislocations of various elements of the spinal column. Sometimes only the facet complexes are involved in the injury. Ordinarily, there is a varied amount of associated vertebral compression fracture. Often there is a residual deformation from the MIV resulting in a deformity that shows extensive separation of the posterior elements of the spine, with widening of the intervertebral foramen seen on lateral radiographs. On some occasions, the failure may occur entirely through osseous material, resulting in an almost perfect slice through the bony structures of the vertebrae (through the spinous processes, transverse processes, pedicles, and the body of the vertebra).[12]

Mechanism of Injury

The mechanism of this injury is one of rapid deceleration of the passenger, in which the patient is virtually wrapped around the seat belt, causing hyperflexion centered in the upper, middle portion of the lumbar spine. Because of the instantaneous axis of rotation imposed by the seat belt in this particular situation, there is relatively little vertebral compression. As a result, the entire structure of the vertebral column is subject to relative amounts of tensile loading. This hypothesis is well described and convincingly supported in the preliminary and comprehensive reports by Smith and Kaufer.[235,236]

Discussion

This particular pattern of injury offers some provocative questions for discussion, especially in view of the limitations of looking at a radiograph and describing the mechanisms involved in injury. In a sizable series of patients reported by Smith and Kaufer, the mechanism of injury was reasonably well documented. Although the exact magnitude and direction of forces involved were not known, the basic clinical situations were quite similar. Nevertheless, the investigators recognized a wide variety of patterns of failure in the vertebral functional spinal unit. This spectrum included a purely osseous injury with all of the ligaments apparently intact. In this situation, as shown in Figure 4-61, the fracture went through the spinous process, the transverse process, and the pedicle, almost as a slice. With this same mechanism, however, there were two injuries in which there was absolutely no osseous failure. Instead, the entire ligamentous structure failed, including all of the posterior ligaments, the facet capsules, and the annulus fibrosus. Between these two extremes, there were varying patterns of facet fractures and avulsions and moderate amounts of vertebral compression fractures.[235]

At this point it is tempting to present a *law* of mechanism of injury. *Similar force patterns do not*

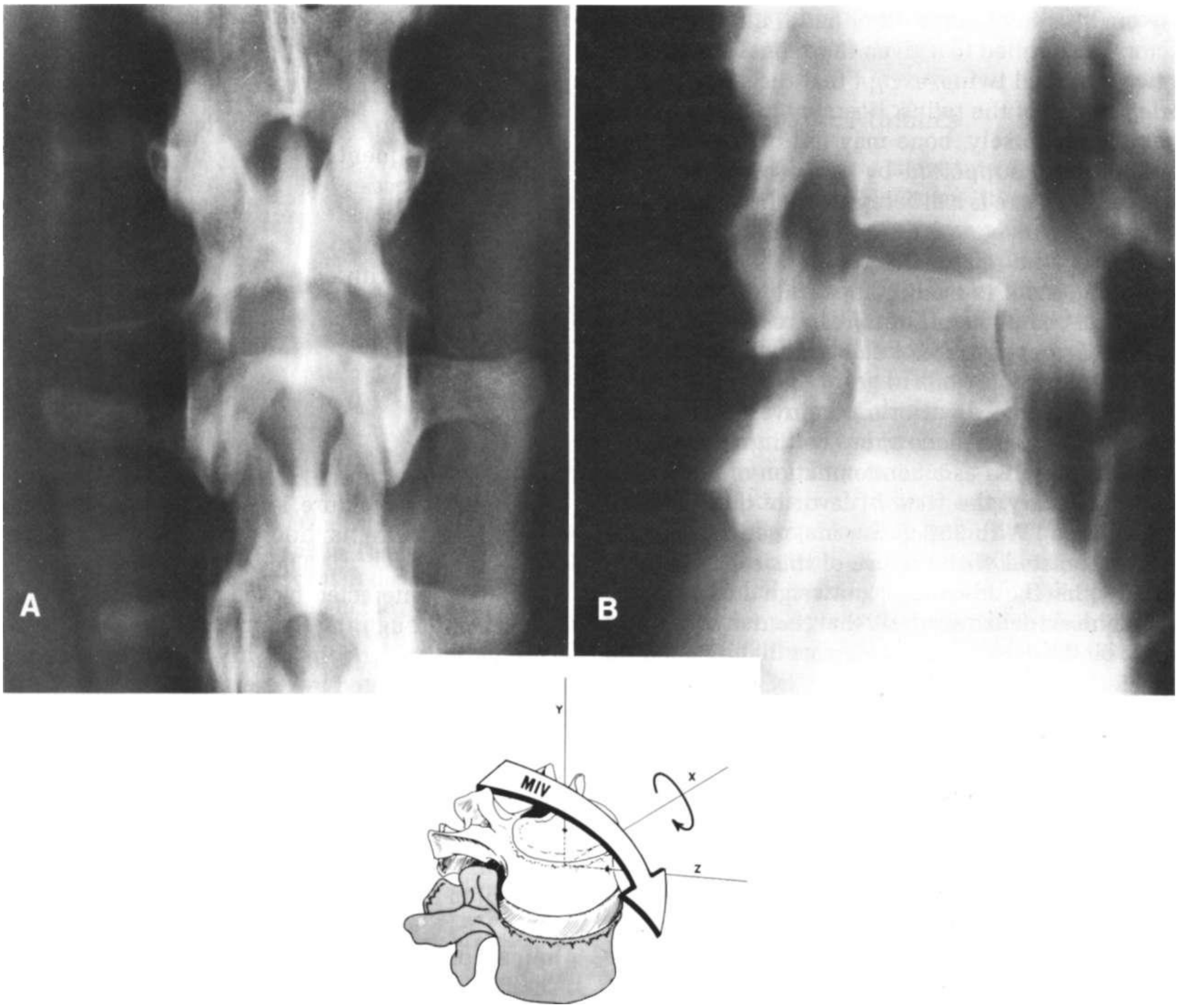

FIGURE 4-61 Radiographs of a 16-year-old girl who was riding in the front passenger seat of an automobile when it suddenly decelerated as a result of an impact with a telephone pole. **(A)** Lateral laminagram shows moderate compression fracture of the most anterior portion of the bodies of L3 and L4, with a horizontal fracture through the posterior elements of L3. **(B)** An anteroposterior view of the fractures of the posterior elements. Note how the clean horizontal slice cuts through the spinous process, the lamina, the pedicle, and actually the right transverse process. The MIV is a positive torque about the x-axis. There is compressive loading anterior to the axis of rotation and tensile loading posterior to it. In this particular case, the axis of rotation is somewhere in the central portion of L3. *(Radiographs courtesy of Dr. E. M. Rhodes)*

necessarily produce identical failure patterns. In addition, it is tempting to reject any laws that categorically state that in some particular injury ligaments will fail before bone or that the converse will occur. There are two important variables that lead to a number of exceptions to any such law. One has to do with *individual* biological differences, and the other has to do with subtle but significant *mechanical* differences. In other words, if the exact mechanical situation is applied to a given functional spinal unit in two different individuals, the thresholds for failure of tendon versus bone in the two individuals may be different. On the other hand, if similar load vectors are applied to a given functional spinal unit in two identical twins, except that the strain rate is varied in one of the twins, ligament may fail before bone, or conversely, bone may fail before the ligament. This is supported by an investigation that showed that there is a differential in the threshold of ligament versus bone failure in tensile loading, depending on the rate of deformation.[178] From these considerations it is readily apparent that clinically similar mechanisms of injury can result in significantly different patterns of damage to the functional spinal units.

We are in accord with various investigators who believe that the presence of lap belt injuries should not be interpreted as a condemnation of seat belts. On the contrary, the facts in favor of their use are compelling![64] With 35% fewer major and fatal accidents associated with the use of this safety device, we think that the advantages outweigh the disadvantages. The evidence suggests that the risk of lap belt injury be thought of as a necessary liability that is especially high when the belt is not properly worn.[101,235]

The belt should not be worn across the abdomen or up above the rim of the pelvis. For correct use of the belt, it is to be strapped across the lower portion of the pelvis at the hip joints. The ideal placement is directly over the anterior capsule of the hip joint. In this situation, the torque is applied closer to the hip joints, which have a greater degree of mobility and intrinsic osseous ligamentous stability than is present in the lumbar spine. Also, the proper use of shoulder straps in conjunction with the lap belt virtually eliminates the possibility of this injury by preventing rotation of the thorax about the lap portion of the seat belt.

Suggested Treatment

Treatment should be based on a careful evaluation of clinical stability. In most instances with severe injuries, at least all the posterior elements are destroyed, and internal fixation and fusion is the treatment of choice. Chance reported lap belt fractures with failure through the osseous tissues. He emphasized that good results could be achieved by treatment in hyperextension.[39]

In the management of these patients, physicians should be suspicious of associated intra-abdominal or retroperitoneal hemorrhage.

Unilateral Lumbar Facet Dislocation

Unilateral lumbar facet dislocations have rarely been reported in the literature.[25,124,170] These are probably flexion rotation injuries. The dislocations are not infrequently accompanied by unilateral transverse process fractures. The problem can be treated with reduction and a 3-month period of immobilization (thoracolumbar cast with one or both thighs) either with or without fusion.

REVIEW OF SPECIAL INJURIES TO THE SPINE

Sacral Fractures

The transverse fracture, sometimes called a suicidal jumper's fracture, is thought to occur as a result of shear loading.[38,82,216] The weight of the torso going down is counteracted by the ground reaction force going up through the femur, the acetabula, the ilia, and the sacroiliac joints. This hypothesis has actually been supported by experimental studies.[216] Another possible injury mechanism is that of a very high impact force against a fixed pelvis.[82,112]

These fractures may be missed because of difficulty seeing them on x-ray and also because they are often accompanied by injuries demanding more attention. CT scans or anteroposterior tomograms are helpful in making the diagnosis. There is a "clinical pearl" to be shared here. When observing the regular anteroposterior x-ray of the sacrum, always look carefully at all the sacral foramina. Check that they have smooth, even arches and are symmetrical in both position and size. Irregularities in these parameters may represent a fracture or tumor.

Fractures of the sacrum are generally associated with high magnitudes of force. Thus, there is usually multiple trauma, and the pelvic fracture itself may incur extensive hemorrhage.

The pelvis is an osseous ring, and rarely is there a fracture at only one segment without an associated fracture or dislocation elsewhere in the ring. Vertical fractures of the alae of the sacrum may be overlooked on radiographs. Often, an associated fracture of the pubis, ilium, or ischium will serve as a "tip-off" for a more careful examination of the sacrum. The failure seems to occur more frequently through the bone of the sacrum than at the sacroiliac syndesmosis. Injuries to the pelvic rim are shown in Figure 4-62.

FIGURE 4-62 A 52-year-old male with fracture dislocation of sacrum and fracture of pelvis. The pubic fracture is more obvious and should always alert the clinician to the probability of a fracture elsewhere in the pelvic ring. The MIV in these types of injury is always of very high magnitude. *(Courtesy of Teaching Library, Department of Diagnostic Radiology, Yale University School of Medicine.)*

Coccygeal Fractures

Coccygeal fractures can also be difficult to diagnose. The normal anatomic variation in this area makes it almost impossible to make a diagnosis solely on the basis of the angulation at the sacrococcygeal joint. These injuries occur as a result of a direct blow or a fall on the "tail bone." Coccydynia is notoriously difficult to treat and often has a significant functional component with or without demonstrable organic disease.

Post-traumatic coccydynia in the presence of a demonstrable fracture is treated with bed rest during the acute phase of 5–7 days. Following this, the patient may be allowed activities as tolerated and a large sponge donut for sitting. Symptoms that persist after 6 or 8 weeks may fall in the category of chronic functional problems.

A recent study of 51 patients operated on for coccydynia probably for the first time shows very good results.[200] The patients were treated with either partial or total coccygectomy, and a good or excellent result was obtained in 88% of the patients. The history of most of these patients involved a fall, usually with associated leg or back pain. It is possible that the psychological aspects of this condition have been overattended.

Spinal Injuries in Athletics

Spinal injuries in sports constitute only about 3% of all athletic injuries. However, it is thought that 50–75% of the fatal injuries involve the head and neck, and cervical spine trauma accounts for about 25% of those injuries.[143] Practically any sport may be involved. A partial list includes the better-known sports, such as football, soccer, lacrosse, rugby, baseball, judo, skiing, riding in snowmobiles, tobogganing, jumping on trampolines, and diving.

The informative, comprehensive, internationally based review by Torg in 1985[262] provides a useful update on sports injuries to the cervical spine. The importance of axial loading in the MOI in American football is described. A typical situation is that of a defensive back making a tackle involving contact with the other player with the top of the helmet. The injury as hypothesized by Torg[264] probably involves axial loading with an element of buckling. (This simulates a significant flexion component in the injury; Fig. 4-63). There is a characteristic clustering of injuries in the middle part of the cervical spine,[264] which may have some basis in the intrinsic mechanical properties of the cervical spine.

We know that rugby, which is not too unlike American football, is also plagued with what is thought to be a flexion injury. These injuries, however, cluster more at the C5–C6 (lower cervical) level. They, too, are often sustained by spear tackling and have stimulated recommendations for rule changes.[160] (This is discussed in the basic science section of the chapter, p. 175.) Appropriate rule changes based on a recognition of the MOI have reduced neck injuries in American football by two-thirds, from 32 in 1976 to 11 in 1983.[265]

Another major source of cervical spine injury has been gymnastics, specifically the trampoline. Because these injuries seemed essentially unpreventable by rule changes alone, trampolines have been outlawed or severely restricted in some European countries. However, trampoline activities are not restricted in the United States to the extent that the data would seem to indicate. In a retrospective questionnaire by Clarke in 1977,[43] 185 high schools, 683 two-year colleges, and 1125 four-year colleges were surveyed to determine the incidence of permanent

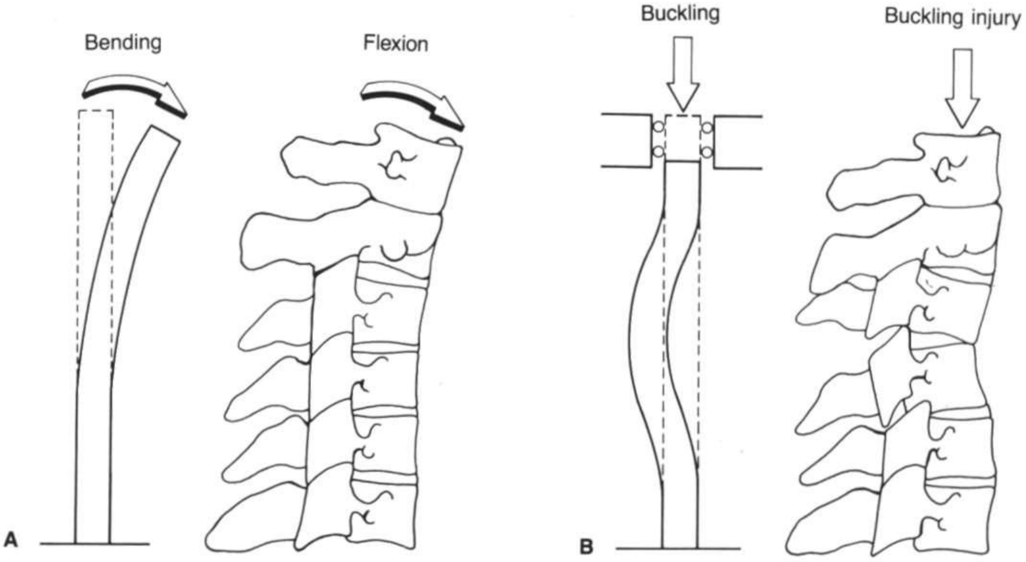

FIGURE 4-63 This figure depicts the buckling hypothesis that is proposed as an explanation for injury in the middle region of the cervical spine. This hypothesis is based on an analysis of the cervical spine as a beam, even though it is recognized that the spine is not a beam. **(A)** Shown on the left is the ordinary bending of a beam due to bending moment application, which is analogous to regular flexion of the cervical spine seen on the right. **(B)** Shown on the left is the beam model with vertical compression loading, which results in buckling. The convexity of the buckling could go in any direction. However, the structure (anatomy) of the cervical spine is such that the buckling is as shown on the right. The anterior elements, with the disc as the major structure, are more compressible than the posterior elements, which involve the less compressible lamina and facet joints. Therefore, the convexity of the buckling injury is posterior. *(Courtesy of Joseph Torg, M.D.)*

paralysis resulting from sports participation during the period 1973–1975. Eight sports accounted for all of these injuries. Football came first with 64%, followed by gymnastics with 26%.

Diving accidents are a frequent cause of cervical spine injuries.[233] Shields and co-workers looked at 152 recreational injuries in patients admitted to the spinal injury service at Rancho Los Amigos Hospital between 1964 and 1973. This 152 represented 9.5% of the total of 600 such patients. The point is that 118 (or 78%) of these recreation-related injuries were due to diving. Because this is not related to organized sports, rules cannot be helpful. Studies by Albrand and Walter[2] of experienced and inexperienced divers have helped to provide the useful guidelines presented in the accompanying chart.

Hockey is another sport in which cervical spinal cord injury is a significant risk. The mechanism of injury, similar to that in football, involves the collision of the helmeted head into the boards with axial loading of the flexed neck in compression. A check of the injured player from behind is usually involved.[253] There are plans in progress to enforce rules against boarding and cross-checking, and rules against checking from behind are also being consid-

DIVERS' GUIDELINES TO AVOID NECK INJURY AND PARALYSIS

1. *Don't* dive in water less than twice your height.
2. *Don't* dive into unfamiliar water.
3. *Don't* assume that the water is deep enough. "Familiar" bodies of water (rivers, lakes, bays, swimming holes) change levels.
4. *Don't* dive near construction work.
5. *Don't* dive until the area is clear of other swimmers.
6. *Don't* drink and dive.
7. *Don't* permit or indulge in horseplay while swimming or diving.
8. *Don't* dive into ocean surf or lakefront beaches.
9. *Do* dive in a properly maintained and supervised pool where the depth has been measured and marked.

ered. Players will be educated about the risks of neck injury in the sport.

Hypothesis for MOI

Torg and associates have developed an interesting hypothesis to explain most of these cervical spine injuries as related to sports. We cannot present it better than it is expressed in this quote.

> With the neck in the anatomic position, the cervical spine is extended owing to normal cervical lordosis. When the neck is flexed to 30 degrees, the cervical spine straightens. In axial-loading injuries, the neck is slightly flexed and normal cervical lordosis is eliminated, thereby converting the spine into a straight, segmented column. Assuming the head, neck, and trunk components to be in motion, rapid deceleration of the head occurs when it strikes another object, such as another player, trampoline bed, or lake bottom. This results in the cervical spine being compressed between the abruptly decelerated head and the force of the oncoming trunk. When maximum vertical compression is reached, the straightened cervical spine fails in a flexion mode, and fracture, subluxation, or unilateral or bilateral facet dislocation can occur.[263]

There are lessons to be learned from the Acapulco divers. In a beautiful setting in Mexico, on the Pacific Ocean, there is a cove bordered on each side by towering, rocky cliffs. The high cliffs have two diving platforms, 100 and 135 ft above the water. On the other side are the low ones with a platform for spectators 40 ft above the water. There is a tradition in which young native men dive off these cliffs to entertain tourists. The base of the rocky cliffs extends 21 ft into the water from the diving points. Therefore, the diver must have a powerful push-off in order to thrust himself away from the cliff enough to clear the rocks below. He must move horizontally away from the diving point a total of 27 ft in order to clear the rocks by a margin of 6 ft. The distance between the cliffs at the place of entry is 15 ft. The depth of the water at the point of entry is 15 ft. Depending on the weight of the diver, wind velocities, and so forth, the momentum at the time of impact from the 135-ft platform is 11,000–16,000 foot-pounds per second.

In 1962, Schneider, Papo, and Alvarez completed a superb clinical study of six of these dives.[225] The study included evaluation of the individual diving techniques, a physical examination, and x-rays of the cervical spine. The histories and neurologic exams of the well-trained divers were normal; however, x-rays of those divers who use an entry technique with their hands over their heads, but apart, showed distinct changes in the cervical spine. The changes included deformation of the vertebrae suggestive of compression, mild fractures, and spontaneous anterior fusion of the vertebral bodies. It was also noted that the radiographic change tended to occur in those divers with longer, less well-muscled necks. The leader and instructor, who had the most dives, used a technique in which the water is entered with thumbs and hands locked and arms straight above the head to protect the head from impact. The well-trained divers using this technique had essentially normal cervical spine plain x-rays. This information may have some application in the training of Olympic-style high-platform divers, although the impact loads are significantly lower. The study does have useful implications in regard to the potential changes in the cervical spine as a result of recurrent impact loads to the vertex.

Football, a sport that is frequently filmed and carefully studied, has been analyzed often with respect to injury mechanisms.[265] The lateral flexion injury of football has been well documented. Melvin and colleagues and Schneider and colleagues have pointed out the role of the face guard in cervical spine football injuries.[164, 226] The lever arm created by the face guard attached to the helmet, which has an excellent grip on the head, supplemented by a snug chin strap, has the potential to apply tremendous torques to the cervical spine. The mechanism can involve either flexion or extension ($\pm\theta x$). Face mask designs that create a smaller moment lever because of the shorter arm (extending only 2.5 cm beyond the tip of the nose) are helpful in diminishing the force applied to the spine in the case of a mishap.

Other spine injuries in football result from the tremendous energy involved in the deceleration of large masses (players of 100–150 kg) colliding with each other (as with spear tackling and blocking) or with the earth. A recent cinematographic kinetic analysis of neck injuries in football estimated the average force acting on the neck of the injured tackler to be between 3120 N and 7120 N (700–1600 lbf).[299] However, these are crude estimates, largely because the films were taken not for precise analysis but simply to record the game. Sometimes these forces exceed the nonfailure energy absorption capacity of the spine and related structures and result in spine failure. Prevention of these injuries can be maximized by good conditioning to take full advantage of biomechanical adaptation and to decrease the probability of failure; good training to teach players the habit of protecting themselves by splinting their muscles for over 2 seconds after the stop-action whistle; and the elimination of spear tackling and some of the plays in which players are highly prone to injury. The best example of the latter is the "kick-off." The role of the trunk muscles in contributing to the support of the spine has been discussed in Chapter 6, and they certainly should be strengthened along with the cervical muscles, which have the possibility of offering considerable splinting protection.

Stauffer and Fox presented 14 patients with quadriplegia as a result of football injuries. All but one of the injuries occurred in teenagers who were either scrimmaging or playing a game. Eleven of the 14 injuries were compatible with a flexed position of the neck at the time of impact. Also, eleven of the injuries occurred in a defensive back involved in open-field tackling.[242]

A flexion injury producing a "tear drop" fracture is not uncommon. Many such injuries occur as a result of spear tackling or diving into unfamiliar waters. Others are due to faulty diving techniques. This fracture and its mechanism are discussed on page 216.

Recent work by Fuentes and colleagues,[92] Nichols and colleagues,[173] and Lee and colleagues[142] has documented some important considerations regarding the tear drop fracture. Based on a study of 32 cases of quadriplegia associated with vertebral body compression fractures documented by the National Football Head and Neck Registry, the investigators arrived at some significant conclusions: (1) the typical "tear drop" fracture is a three-part, two-plane fracture (see Fig. 4-2); (2) this three-part, two-plane vertebral body fracture is present in no less than 50% of the football injuries that result in quadriplegia; (3) the sagittal vertebral body fracture is an internal but previously little-recognized feature of this fracture pattern; (4) axial loading of the cervical spine is the responsible mechanism of injury; (5) the fracture is unstable and is associated with catastrophic neurologic injury; and (6) complete radiologic examination is essential in evaluating patients with cervical spine trauma in whom this lesion is suspected. Although there are no major new revelations here, the key points are now well documented.

We have several comments. Although axial loading is a major factor, there may be a significant sagittal plane bending moment (flexion or buckling) involved in this injury.

Computerized tomograms, magnetic resonance imaging, and possibly a myelogram may be necessary to fully characterize and evaluate this injury. These fractures probably are unstable, and 48% will have an associated lamina fracture.[92] Certainly, in the presence of an incomplete neurologic lesion, or for the purpose of gaining an additional set of nerve root functions, these injuries may be treated with vertebrectomy and anterior fusion.

Snowmobile and tobogganing accidents usually cause fractures in the lumbar spine. The most common is a type of compression fracture due to sudden vertical loading secondary to hitting a bump. In snowmobiles, these could be damped to some extent by the seat construction. The toboggan rider can generate a considerable damping effect through his own joints by riding on his knees and shins, with knees and hips flexed.

We as physicians should probably be more assertive in recommending rule changes in athletics and other aspects of preventive medicine, such as education of the athletes about risk as it relates to spine trauma.

Spondylolisthesis

In spondylolisthesis there is a defect in the pars interarticularis that is associated with anterior translation of the involved vertebra in relation to the subjacent one.

Mechanism of Injury

The mechanism is unknown. However, a feasible theory suggests that it is due to a fatigue fracture superimposed on a hereditary defect or predisposition.

Discussion

An excellent analysis of etiologic factors has been described by Wiltse and colleagues.[292,293] The possible causes are listed in the accompanying display.

The theory that seems to have survived best at present is that of fatigue fracture. This is based primarily on a number of observations, including radiographic documentation of an intact pars prior to vigorous athletic activity and/or discrete trauma,

SOME HYPOTHESIZED MECHANISMS OF DEVELOPMENT OF SPONDYLOLISTHESIS

Separate ossification centers (The lateral masses, which normally have one ossification center, have two that fail to fuse.)
A fracture that occurs at the time of birth
An ordinary fracture of that region
A stress or fatigue fracture
Displacement secondary to increased lumbar lordosis
Displacement secondary to a pinching mechanism of superior and inferior articular processes
Weakness of regional ligamentous and facial support structures
Aseptic necrosis of the pars interarticularis
Dysplasia of the pars interarticularis

followed by radiologic evidence of spondylolisthesis.[46, 293]

There are several factors about the overall clinical picture, however, that are not entirely compatible with the hypothesis of fatigue fracture. There is a significant difference in the incidence of the lesion across genetic barriers. Certain families have a documented predisposition. The disease is thought to have an incidence of 1.95% in American blacks, 5.8% in American whites, and about 60% in American Eskimos. Fatigue fractures usually occur *after* the age of 6.5 years. Yet this is when a precipitous increase in spondylolisthesis is observed. The radiographic changes are not typical of fatigue fractures elsewhere. The clinical history in patients with these lesions is not one of unaccustomed repetitive loading, as in the "military trainee" or the newly initiated "jogger" who sustains a fatigue fracture. These differences are possibly due to a different configuration of loading involved in the lumbar segment, resulting in a clinical picture that is atypical of the fatigue fracture. A definitive etiology remains to be proved.

Suggested Treatment

In young individuals, 21 years of age or less, plaster immobilization in a body jacket from the nipples down to the knee, on one side for 3 to 4 months, should allow for healing. In order to rest the lumbosacral joint, it is necessary to include one thigh (see Fig. 7-24). In adults, and in situations where conservative treatment does not eliminate symptoms, bilateral posterolateral spine fusion to a normal vertebra above and below the deficit is indicated. In cases where there is nerve root irritation, removal of the posterior elements of the involved vertebra, as described by Gill and co-workers,[96] is desirable. We believe that, except for unusual circumstances, this procedure should be done in conjunction with a spine fusion to provide clinical stability and prevent further anterior translation of the body of the involved vertebra.

Osteoporosis and Spine Trauma

Much work has been done on the cellular and metabolic aspects of osteoporosis. However, the critical clinical variable in this disease is the overall strength of the bone. It is estimated that 4 million people in the United States, for example, have osteoporosis that is advanced enough to result in vertebral fracture. Recently, more attention is being directed to the biomechanical aspects of the disease. In particular, investigators have begun to define the correlations of quantitative computed tomography with the apparent density of the vertebral trabecular bone and with vertebral body strength in compression. These investigators found statistically significant correlations between the quantitative CT scan and direct measurements of the vertebral trabecular apparent density ($R^2 = 0.89$, and $p < 0.0001$).[159]

Also, there was a high correlation of vertebral compressive strength with direct measurements of trabecular apparent density. However, the vertebral compressive strength did not correlate significantly with quantitative CT scan. Despite this last finding, there is good reason to anticipate that it will be possible through quantitative CT screening to predict vertebral fracture risk in osteoporotic patients. Subsequent work has shown a stronger correlation with vertebral compressive strength if the square of the average trabecular density in the anterior portion of the vertebral body is used.

The work of McBroom and co-workers showed the relative importance of cancellous bone as compared with the cortical shell in determining overall vertebral strength.[159] Contrary to the prior observations of Rockoff and colleagues,[213] who found that the cortical shell contribution could be anywhere from 45–75%, McBroom and associates found that with removal of the cortex there was only a 10% drop in overall vertebral body strength. This observation, indicating a relatively minor contribution by the cortex, explains the usefulness and accuracy of the apparent density of the trabecular (cancellous) bone as a predictor of vertebral strength.

Hansson and associates have studied the mechanical properties of the human vertebrae for the same clinical biomechanical purposes using different methodologies.[105–108] These investigations have demonstrated a correlation of ultimate vertebral strength with bone mineral content ($r = 0.86$).[106] They have also shown that bone mineral content decreases with age[107] and that the geometric configuration of the vertebral body is a factor that is to some degree related to the extent of disc degeneration and the amount of bone mineral content.[108] In order to estimate vertebral strength, these investigators measured material properties (using dual photon absorptiometry) and dry density of the trabecular bone.[105] They found that compressive strength, modulus, and stiffness were proportional to the square of the dry density and to the square of the bone mineral content. A clinical use of this information would employ dual photon absorptiometry to predict compressive bone strength.

The clinical value of either or both methodologies once developed will be to prevent fractures by alerting and protecting at-risk patients. Also, it will be possible to monitor the natural disease process and to evaluate various therapeutic programs.

Pathologic Fractures

When there are alterations in the quality or the quantity of the supporting structures of the vertebra, fracture can occur with loads far below those required to damage the normal vertebra. The location of the abnormality within the vertebra and the orientation of the damaging vector determine the type of fracture that occurs. The details of the mechanical effects of structural loss in osteoporosis are discussed on p. 40. Boukhris and Becker have studied the association between vertebral fractures and osteoporosis and have reported the incidence of these fractures with sex and race groups.[27]

Fractures in Ankylosing Spondylitis

The emphasis here is that morbidity and mortality associated with fractures in cervical spines affected with ankylosing spondylitis are much greater than what we see in fractures of cervical spines not involved with this disease. Barnes, Hollin and colleagues, Janda and colleagues, and Vicas reported differences in the clinical picture of trauma in the presence of ankylosing spondylitis. These investigators noted a poor overall prognosis; there was a higher incidence of spinal cord injury, and it was more severe and complicated. Even in cases where there was fracture without neurologic deficit, the mortality was greater.[15,120,127,270]

The difference is largely due to the tremendous loss of flexibility of the spine with ankylosing spondylitis. This allows for a much lower threshold for injury, and trivial trauma can result in a fracture. Neurologic deficit, if present, may be due to a hematoma[116] or to direct trauma associated with the fracture. Nature has carefully evolved the highly specialized construct of the normal spine, which is designed for considerable energy absorption through its flexibility. The more rigid spine of ankylosing spondylitis does not have this capacity and therefore fails after the absorption of much less energy. Consequently, it may fragment more and also exert more force onto the cord after fracture. Because of the rigidity of the spine on both sides of the fracture, the probability of large displacements and initial or additional spinal cord trauma is quite significant. Thus, we agree with Janda and colleagues[127] that it is desirable to consider halo fixation. Immobilization for 5–6 months is suggested before carefully controlled flexion and extension studies are conducted to check healing.

Suggested Treatment

The following recommendations are largely based on the work of Brown and Raycroft.[32] Consider all patients with ankylosing spondylitis who sustain head and neck trauma to have a cervical spine fracture, until proved otherwise. Lateral tomograms may help to make the diagnosis. These fractures should be considered unstable and on the brink of neurologic catastrophe (severe initial or additional neurologic deficit). It is recommended that fractures in these patients be treated with rigid immobilization in a halo cast in the preinjury position of deformity. Some clinicians think that the halo cast is preferable to the halo vest, if any halo traction is needed. The authors warned that halo *traction* may not maintain proper position. Other surgeons think that these patients are best managed with internal fixation of an arthrodesis.

Not all problems are solved by the halo, however, because the problem of maintaining alignment of one or two long stiff segments of spine remains. This may well be a situation in which internal fixation anteriorly and/or posteriorly is beneficial. The halo would still be useful, as the bone in ankylosing spondylitis is often weak or brittle. More well-documented clinical experience is needed to resolve this problem.

Klippel-Feil Syndrome and Minor Cervical Spine Trauma

The biomechanical principles of stress concentration and the inability of a rigid spine to absorb energy to avoid fracture are unfortunately demonstrated in Klippel-Feil syndrome. A case report exists of a patient with congenital fusion of several vertebrae who sustained minor trauma (fell back against a car) that resulted in quadriplegia.[76] This is compatible with our clinical experience and a somewhat similar case report by Sherk and Dawoud.[231] We agree with the idea that perhaps these patients ought to be screened and evaluated on an individual basis, and when there is hypermobility and instability, prophylactic arthrodesis is considered and discussed with the patient. More experience and clinical documentation will be useful.

Fracture Dislocations in Children

Any discussion of spine injuries in children should include a statement about interpretation of lateral cervical spine radiographs. In children under 10 years of age, there normally may occur at the C2–C3 or C3–C4 levels as much as 3–4 mm of anterior (+z-axis) translation. This should not be diagnosed as a dislocation or subluxation without other distinct clinical evidence.

Spinal injuries and fracture dislocations in children are traditionally thought to be relatively rare. A recent report by Aufdermaur suggests the possibility that their reputation for rarity may be due in part to failure of recognition. He reported on 12 cases of spinal injury in patients from 0 to 18 years. In all these patients, the fracture occurred through the epiphyseal plate. The upper plate was involved twice as frequently as the lower one. The fractures occurred in the growth zone, in either the columnar region or the zone of provisional calcification. These injuries were difficult to see on radiographs. This work has pointed out the fact that epiphyseal injuries occur in vertebrae in much the same manner as in the rest of the growing skeleton. Those regions of the epiphysis seem least able to resist loads in tension, under compression, or during shear because they have the least interstitial support structure and the greatest amount of cell protoplasm.[12]

■ *CLINICAL BIOMECHANICS*

- Our precise understanding of the mechanisms of injury in spine trauma is a good deal less developed than the usual casual deliberations imply.
- Mechanical buckling is an important concept in the evaluation of cervical spine trauma.
- Rule changes based on an understanding of MOIs can be effective preventative measures in sports injuries. Perhaps the medical profession should be more assertive in these matters.
- Careful handling and splinting are crucial in the immediate and intermediate care of patients with injury to the spine. This is particularly the case for young children, who have relatively large heads.
- Some basic understanding of the normal kinematics and biomechanics of the spine is crucial to a meaningful analysis and interpretation of injuries and their mechanisms.
- An understanding of the regional variations in the ranges of motion and the stiffness properties of the spine is fundamental to an understanding of mechanisms of injury. The cervical spine is probably the most flexible and has the widest range of mobility. It appears that local stiffness and other distinct mechanical characteristics in the middle cervical spine (C3–C5) may be related to certain mechanisms of injury. The thoracic spine is the least flexible and has the smallest range of motion. The lumbar spine lies somewhere in between in respect to these parameters.
- Horizontal shear forces are *probably* the most efficient mechanisms for disrupting ligamentous connections between vertebrae.
- In vertical compression loading, the vertebral end-plate fails first most of the time, but not always.
- When high magnitudes of force are applied to the spine, relaxed muscles seem to be associated with less cord injury than do tense or actively contracting muscles.
- With vertical impact loading, the relaxed, relatively unrestrained cervical spine may be less traumatized than a more restrained one (*i.e.*, one with contracted muscles).

■ The anatomic and clinical damage in spinal cord trauma is not based on an all-or-none threshold of force. It is related to the magnitude of the trauma in a variety of complex ways, with the *impulse* being of particular importance.

■ Hemorrhage and edema appear to be the major pathoanatomic result of trauma to the spinal cord.

■ The clinical history in evaluating the mechanism of injury of spinal trauma is valuable only after it has been carefully checked and collaborated with other evidence, and even then it is limited significantly

■ A variety of bruises and lacerations can occur on rebound impact or during other associated trauma. These should not be confused with the effects of the major injuring force that has caused the spine trauma and that is under evaluation. Therefore, in analyzing the mechanism of injury, these signs are to be weighed and checked along with other information.

■ The anterosuperior or anteroinferior triangle of bone that is frequently seen on lateral radiographs of the injured cervical spine may represent failure of that bone in either shear or tension. This means that such an injury may occur as a result of compressive or tensile loading.

■ The shear component of this injury is at about 45° to a presumed vertical force. This fits nicely from a theoretical point of view because the "tear drop" fracture has been discovered to be much like the burst fracture of the thoracic and thoracolumbar spine, which is largely a compression injury.

■ The site of failure of a number of cervical spine injuries is largely dependent upon the anatomic structures of the bone involved. Fractures of the ring of C1 generally occur in the region of the ring that is grooved by the vertebral artery. Here the structure is weaker and less resistant to certain bending loads. Also, in rare instances the vertebral artery may be involved at the site, causing symptoms of vascular insufficiency.

■ Dens (odontoid) fractures must be carefully evaluated and their treatment individualized. There are clear-cut instances in which they need not be fused. However, the mechanics and the risks involved are such that it is advisable to fuse when there is doubt about adequate stability. In injuries that result in an abnormal relationship between C1 and C2, careful biomechanical evaluation is crucial. While the authors do not recommend fusing all such injuries, if there is doubt after a systematic evaluation, particularly with Type II, surgical stabilization is suggested.

■ Evaluation of persistent post-traumatic torticollis should include careful radiographic analysis of the C1, C2 relationship.

■ The so-called hangman's fracture (traumatic spondylolisthesis) is of considerable biomechanical interest. It provides an opportunity for understanding the group of related injuries that can occur as a result of the forces involved in the process of judicial hanging. This group includes the following injuries: odontoid fractures, fractures of the ring of C1, and traumatic spondylolisthesis of C2.

■ An important consideration in the management of a patient with traumatic spondylolisthesis is the recognition of the possible association with failure of the anterior longitudinal ligament and the attachment of the annulus fibrosus between the bodies of C2 and C3. The defect may not show up on routine radiographs but may become evident catastrophically if too much axial traction is applied. Therefore, traction should be applied in increments and monitored by lateral radiographs and neurologic evaluations. These precautions will prevent the undesirable displacement that can occur.

■ The "tear drop" compression fracture causes trauma to the cord through the impact of the expanding posterior shell of the vertebra. The shell may or may not remain displaced in the spinal canal at the time of radiographic examination. If there is residual displacement of fragments, one must consider surgical decompression and fusion. Theoretically, on a fresh fracture with an intact posterior longitudinal ligament, reduction could occur with traction.

■ In persistent unilateral facet dislocations without fracture, an understanding of the mechanics of the injury and the kinematics of the cervical spine can be helpful in reduction of cases that are resistant to simple axial traction. The basis of the manipulation is explained and involves the following: With the head under about 28 kg (63 lb) of traction weight (maximum traction), a carefully controlled bending of the head laterally away from the dislocated side to 60° is followed by an axial rotation of the head of about 60° toward the dislocated side. Finally, with the head held in this position, 50% of the traction is carefully removed. This should achieve reduction.

■ Prior to any type of reduction of cervical spine facet subluxations and dislocations, imaging of the associated intervertebral disc is necessary to rule out disc protrusion and to prevent damage with reduction in the presence of a disc protrusion.

- Despite the frequently present psychiatric and medicolegal factors involved in whiplash, there is clearly good biomechanical and anatomic evidence for considering such injuries to be bona fide organic diseases.
- The patterns of failure in severe hyperextension injuries beyond whiplash may include unrecognized tensile failure either through the osseous mass of the vertebral body or at the attachment of the annulus fibrosus. The residual deformation after recoil may result in a spontaneous reduction, which can make recognition on routine radiographs impossible.
- Biomechanical studies of C0–C1–C2 and the transverse and alar ligaments provided a theoretical basis for some of the clinical findings in whiplash injuries. Clinical correlations are needed.
- Vertebral end-plate fractures can easily go unrecognized. Lateral laminagrams are helpful when the diagnosis is not evident.
- With manubriosternal dislocations, wedge fractures of the cervicodorsal region, and the converse, should be suspected.
- In determining the probability of progressive kyphosis following a compression fracture, the clinician should bear in mind that the compacted compressed vertebra may have 60–70% or more of its original preinjury ability to withstand vertebral compressive loads. Therefore, physiologic loads should be tolerated without additional deformity from further vertebral wedging and compression. The clinician must take into consideration the status of the posterior ligaments. If they are intact, then protected ambulation should be adequate to prevent subsequent deformity in most situations.
- It is suggested by a number of authorities that in severe unstable fracture dislocations of the thoracic and thoracolumbar spine, protection of any residual cord and nerve root function, as well as the general post-trauma care and rehabilitation of the patient, is greatly facilitated by secure internal stabilization.
- If the seat belt were worn properly across the hip joints and in conjunction with a shoulder harness, seat belt injuries would be nonexistent. However, in our opinion, even an improperly worn seat belt is better than no seat belt at all.
- Fractures of the spine with ankylosing spondylitis carry high mortality and morbidity and thus are best treated with a halo type of fixation, and may also need internal fixation.

NOTES

[A] The authors are not aware of any epidemiologic documentation of less injury among relaxed individuals. The role of muscle forces before, during, and after spine trauma is not well understood. It appears that there may be some protection for the athlete through splinting when trauma can be anticipated. The work of Soechting and Paslay showed that the reaction time required for the motorist to voluntarily splint his muscles virtually eliminates the possibility of any voluntary muscle activity at the time of injury.[237] These investigators observed that in a typical automobile moving at a speed of 10 km/h (6 mph), the head will hit a dashboard 38 cm (15 in) away from the head in about 0.15 seconds. With the muscles' reaction time being the same order of magnitude, it is not possible for the muscles to contribute any protection when collision happens at higher speeds. The critical unit of 10 kmph (6 mph) obviously eliminates the large majority of auto accidents. On the other side of this issue we have the evidence of Gosch and colleagues, who showed clearly that cervical spine injuries were less severe in anesthetized than in awake monkeys subjected to trauma.[98] It may be that normal muscle tone rather than splinting is the crucial factor. The important question of the effects of muscle forces in spine trauma needs and merits considerable study.

[B] When trauma is created by impacting a given mass M in kg (lb) from a height h in m (ft), there may be different mechanical parameters that are relevant to the trauma of the spinal cord. There are at least three that are readily discernible: the energy of the falling mass due to its height, the momentum of the mass due to its velocity, and the impulse that the mass delivers to the spinal cord. As can be seen from the equations below, these quantities are distinctly different.

$$\begin{aligned} \text{Energy} &= mgh \\ \text{Momentum} &= m\,v_1 = m\sqrt{2gh} \\ \text{Impulse} &= m(v_2 - v_1) \end{aligned}$$

Where g is gravitational acceleration = $9.81\ \text{m/s}^2$ ($32\ \text{ft/s}^2$)

v_1 is the velocity of the mass at the time of impact in m/s (ft/s)

v_2 is the velocity of the mass on rebound in m/s (ft/s)

In a recent paper, correlation between impulse, as defined above, and the traumatic spinal cord lesion produced has been shown.[60]

[C] In analyzing injury, the more traditional terms are maintained; however, in addition, our classification is used whenever the knowledge is adequate. The standard uses of the coordinate system have been described in Chapter 2. Nevertheless, a partial review is provided here. The traditional anatomic planes are shown in

Figure 2-2. These designations assume the anatomical position.

Planes
- y, z—sagittal plane
- x, y—frontal plane
- z, x—horizontal plane

Directions
- + x translation—toward the patient's left
- − x translation—toward the patient's right
- + y translation—cephalad or up
- − y translation—caudad or down
- + z translation—forward
- − z translation—backward

Rotation
- + θx—clockwise about x-axis, as in flexion
- − θx—counterclockwise about x-axis as in extension
- + θy—clockwise rotation about y-axis, as in axial rotation to the left
- − θy—counterclockwise rotation about y-axis, as in axial rotation to the right
- + θz—clockwise rotation about z-axis, as in right lateral bending
- − θz—counterclockwise rotation about z-axis, as in left lateral bending

[D] A more detailed look at the concept of the major injuring vector involves considering the vector to consist of the forces and moments acting on the vertebra in question. Because the traumatic force is seldom directly applied to the spine, there is always a bending moment and/or a torque at the site of injury, along with the original force. The moment or torque is the result of the lever arm between the force and vertebra. As an example, take the case of a man hit from the front on the center of the forehead. If the force is in the sagittal plane, then the MIV present at the vertebra consists of a force that is exactly equal and parallel to the original force on the head, and a bending moment that is equal in magnitude to the force times the shortest (perpendicular) distance from the vertebra to the line of action of the force on the head. Now if the person were not hit exactly in the center of the head, or if the force were not in the sagittal plane, then in addition to the force and the bending moment there would also be a torque about the y-axis of the vertebra. To complicate the problem even more, the resulting moments and torques at a vertebra at the time of injury are dependent upon the spatial relationship between the original force and the instantaneous position of the vertebra in space. The MIV presented in the various diagrams for a given injury is the most dominant force and/or moment vector *at the vertebra* responsible for causing the injury, according to the best biomechanical analysis of the injury-causing mechanism.

[E] In hangman's fracture, the major injury vector is a bending moment acting on C2 by means of the dens and the anterior ligamentous structure. Let this moment be M, as shown in Figure 4-30A. The free-body analysis technique involves isolating C2. Let us assume that the contribution from C1 onto C2 is a bending moment M and the contribution from C3 onto C1 is a pair of forces F_1 and F_2, as shown. Because the vetebra C2 must be in moment equilibrium, the two forces must be equal, opposite, and parallel. The magnitude of these forces is given by the equation

$$F_1 = F_2 = \frac{M}{D}$$

where D is the distance perpendicular to the forces. Force F_1 is a tensile force contributed by anterior ligamentous structures between C2 and C3, and force F_2 is a compressive facet joint reaction. The bending moment diagram for this loading situation is shown in Figure 4-30B. The bending moment is maximum in the middle of the two forces. Anatomically, this is also the section with the lowest moment of inertia. The probability of fracture at this section with this kind of loading is therefore quite high. If the force F_1 is not parallel to the axis of the disc C2–C3, then this disc will have shear forces in addition to the tensile force. Similarly, if F_2 is not perpendicular to the facet joint surfaces, shear forces will be present at the joint, thus causing stress in the facet capsular ligaments.

[F] The study of Schneider and colleagues is another example in which Poisson's ratio is operative. A compressive load across the cord produces tensile strains in the axial direction.

[G] One of the very few motion patterns in which the lumbar spine is stiffer than the thoracic spine is that of axial rotation. In axial rotation, the stiffness coefficient of the lumbar spine is about two times that of the thoracic spine.[186] Thus, the risk of fracture is greater than the risk of dislocation in the lumbar region when there is an axial (y-axis) torque.

[H] Traction is a force; the unit of measure is newtons (pound-force). However, it is generally applied by hanging weights that are specified by their mass in kilograms (pounds). Thus, a 1-kg weight applies 9.8 N (2.2 lbf) of traction. Similarly, a 1-lb weight applies about 4.5 N (1 lbf) of traction.

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The Problem of Clinical Instability in the Human Spine: A Systematic Approach

> *"I don't know what you mean by 'glory'," Alice said.*
>
> *Humpty Dumpty smiled contemptuously. "Of course you don't—till I tell you. I meant 'there's a nice knock-down argument for you'!"*
>
> *"But 'glory' doesn't mean a nice knock-down argument," Alice objected.*
>
> *"When I use a word," Humpty Dumpty said, "it means just what I choose it to mean, neither more nor less."*
>
> *"The question is," said Alice, "whether you can make words mean so many different things."*
>
> *"The question is," said Humpty Dumpty, "which is to be Master—that's all."*
>
> *—Lewis Carroll,* Alice in Wonderland

5

INTRODUCTION

During the time that one of the authors was preparing for his orthopedic boards, he asked himself the following question: "How do you determine when the spine is unstable?" This question was most anxiety-provoking, especially since the initial response was a recollection of a surgeon tugging up and down on the posterior elements of a lumbar vertebra and saying, " . . . yep, it's unstable, we'd better fuse it." The aspirant thus went frantically through the standard orthopedic references in search of the answer. This study tended to reduce the anxiety level with regard to the ensuing examination, but it clearly raised some other, serious doubts. No one seemed to have a clear and valid answer to the question. In addition, it became apparent that making such a crucial determination in the clinical situation is extremely difficult. In many instances, such a decision can very significantly affect patient care. Misjudgment of one type may result in death or major neurologic deficits. Misjudgment of another type may result in unnecessary surgery, again with death or other major complications. Evaluating a patient's condition erroneously can cause considerable needless inconvenience related to wearing complex encumbrances, such as Minerva casts or halo pelvic fixation devices. Correct judgment provides the patient with realization of the maximum recovery with an absolute minimum of risks and inconvenience. The problem of clinical instability is not yet solved, and this chapter does not purport to provide physicians with ideal judgment and wisdom. It does, however, endeavor to present a systematic approach to the problem, based on current clinical and biomechanical knowledge.

Definitions

Many physicians, to some degree, have been like Humpty Dumpty in the "mastery" of the word *stability* as it is used clinically. There are a number of stated and unstated definitions of the term,[49, 124] and if the definition of the condition is confusing, it can be expected that the diagnosis and treatment of the condition will reach progressively higher orders of ambiguity. One of the problems in the literature has been the absence of a clear definition. All physicians use the term *stability*, but they may have a variety of different concepts and definitions in mind as they use it. In this chapter, we have chosen to employ the term *clinical instability*. The working definition is given in the box below.

Clinical instability is the loss of the ability of the spine under *physiologic loads* to maintain its pattern of displacement so that there is no initial or additional neurological deficit, no major deformity, and no incapacitating pain.

The complexity of the subject matter demands a few qualifiers. *Physiologic loads* are those which are incurred during normal activity of the particular patient being evaluated. *Incapacitating deformity* is defined as gross deformity that the patient finds intolerable. *Incapacitating pain* is defined as pain unable to be controlled by non-narcotic drugs. Clinical instability can occur as a result of trauma, disease, surgery, or some combination of the three.

Unless the term *clinical instability* or *clinical stability* is used, we are not referring to the preceding definition. Stability or instability alone is used when the term found in the literature is repeated and when other statements about it are reported. The term has rarely been defined in previous publications. Its connotations, however, occasionally overlap, to some extent, with the definition we have offered. When the term *clinical instability* is used, it refers to the preceding definition.

Background and Organization

In the diagnosis of clinical instability in any region of the spine, several crucial factors come into play. Anatomy is significant in terms of position and space relationships between neural structures and potentially damaging structures. It is also important because various structures provide different magnitudes and types of forces that are helpful in preserving stability. Biomechanical studies and information on kinematics are presented whenever they are contributory. For each region of the spine, recommended methods of evaluation and management are discussed according to the outline in the accompanying display.

BASIC ELEMENTS OF A SYSTEMATIC ANALYSIS OF THE PROBLEM OF CLINICAL STABILITY IN THE SPINE

Anatomic considerations
Biomechanical factors
Clinical considerations
Treatment considerations
Recommended evaluation system
Recommended management

The various anesthesiology departments of the nine teaching hospitals of Harvard Medical School in 1986 proposed and agreed to use certain routine standards of patient monitoring during administration of anesthesia.[46] The standards were put forth in a list of certain monitoring procedures and other basic practices, such as anesthesiologists' or nurse anesthetists' presence in the operating room, breathing system disconnection monitoring, ability to measure temperature, and so forth.

In this as well as the first edition of this work, we suggest a systematic checklist approach to the evaluation of clinical stability. In our view, it is a method that assures a weighted consideration of the important factors and provides a reproducible standard. We realize that the "checklist protocol" approach may run counter to the strong tradition of independent physician judgment and decision making. However, the current medical, legal, economic, and social climate may direct the evolution of medical education and practice more toward a checklist or protocol type of approach for our common activities. Most probably, things will evolve toward establishing standards of care for problems of the spine. We believe that this will improve quality and yet leave many challenging opportunities for problem solving and individual physician ingenuity and creativity.

The checklists offered in this chapter to evaluate instability are intended as a guide to a consideration of the multifactorial clinical picture; they are not developed, nor are they submitted to represent a standard of care.

Radiographic Magnification

The major practical consideration in the determination of clinical instability is the evaluation of the patient's radiographs. By adopting standard distances, more meaningful measurements may be made from radiographs. Radiographic examination is the most often used objective means of determining the relative positions of the vertebrae in a potentially unstable spine. Therefore, it is important to give some consideration to the accurate interpretation of linear radiographic measurements.

Linear Measurements

The parameters measured on radiographs are either linear, such as the distance between two points, or angular, such as the angle between two lines. The relative position of the radiographic source, the spine, and the film are the only factors that affect the magnification. A three-dimensional vertebra is transformed into a two-dimensional image on the radiographic film. A line AB on the object (vertebra) in a plane parallel to the film is imaged as A′B′ on the film. The image is always bigger than the object. Let the magnification M be the percentage increase in length. The formula that shows the dependency of magnification on source, object, and film positions is

$$M = \frac{100 \times D_2}{D_1 - D_2}\ \%$$

where D_1 is the distance between the radiographic source and the film, and D_2 is the distance of the object (spine) from the film. These are shown in Figure 5-1.

The most commonly used value for the distance D_1 is 1.83 m (72 in). If the film is placed next to the shoulder, then the object-to-film distance D_2 in a lateral radiograph is half the shoulder width. Assuming this is 0.3 m (12 in) for the "average" person, the magnification, using the above formula, is 20%. But the distance D_1 has not been standardized. Assuming a variation of 0.3 m (12 in) below and above the value of 1.83 m (72 in), the corresponding magnifications are 25% and 17%.

This range of magnification is for the "average" person. However, there is considerable variation of shoulder widths among individuals. For a nominal value for D_1 of 1.83 mm (72 in), the magnification factors for different shoulder widths (D_2) can be calculated. The results are shown in Table 5-1.

This table clearly shows that there is large variation in magnification introduced by the physical size of the person. The technique of holding the film next to the shoulder is responsible for this. If the two kinds of variation described above (source-to-object and object-to-film distances) are allowed to operate

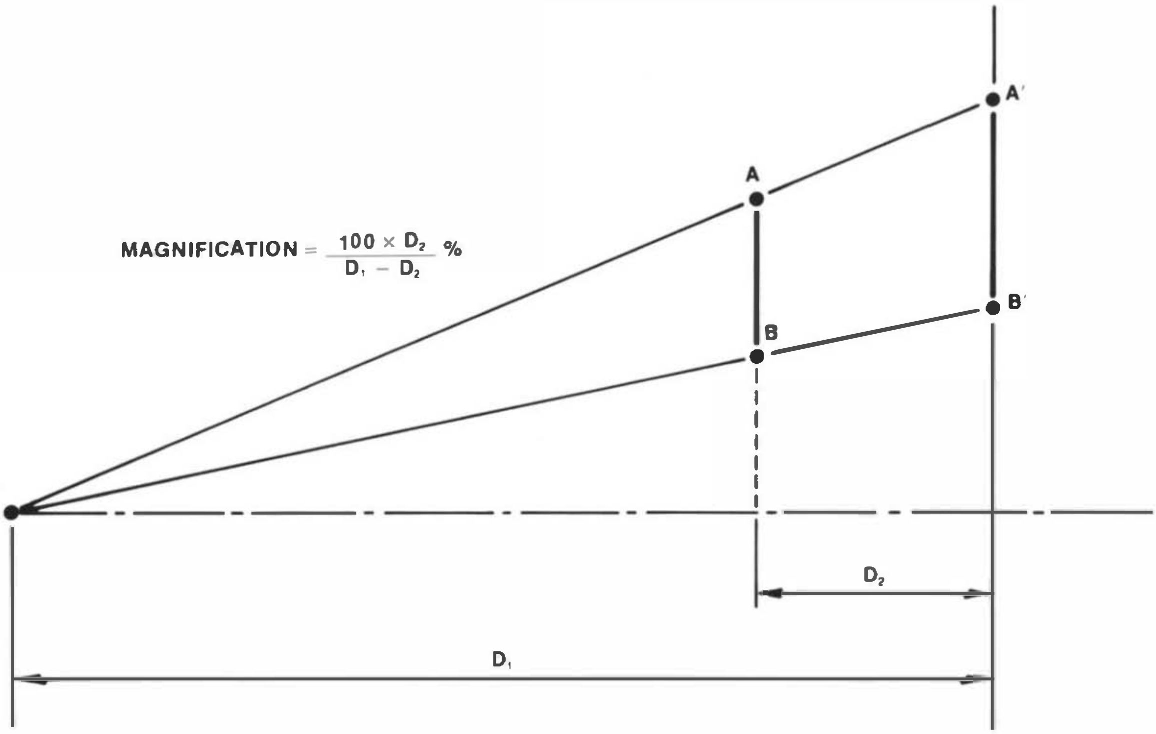


FIGURE 5-1 On an x-ray, the image is always bigger than the object: *A'B'* is greater than *AB*. The magnification depends upon the source-to-film and object-to-film distances, D_1 and D_2, respectively. D_2 is the distance that is the more sensitive in regard to magnification.

TABLE 5–1 Percentage Magnification of Image Associated with Different Spine-to-Film Distances*

Spine-to-film Distance m (in)	Magnification (%)
0.15 (6.0)	9
0.20 (8.0)	12.5
0.25 (10.0)	16
0.30 (12.0)	20
0.36 (14.0)	24

* Source-to-film distance = 1.83 m (72 in)

simultaneously, then the range of magnification widens to 7.5–30%.

The large variation in the magnification as shown above, of course, makes it nearly impossible to make any precise measurements from the radiographs when the distances D_1 and D_2 are unknown. An effort should be made to obtain radiographs in which the distances are known, so that, using the equation above, one may easily calculate the magnification.

Another approach is to *standardize* the distances. This has several advantages. Templates can be prepared to utilize the standard distances; repeated computations to determine magnification for different patients are not necessary; and, most important, it is then possible to compare different radiographs of the same patient, even if they were taken at different times. Based on these considerations, we suggest the following as standard distances:

Radiographic source-to-film distance,
D_1 = 1.83 m (72 in)
Spine-to-film distance,
D_2 = 0.36 m (14 in)

These two distances give a linear magnification, M = 24%. The value of 1.83 m (72 in) was chosen because it is the one in most general use. The value of 0.36 m (14 in) for the distance D_2 was chosen to allow inclusion of practically all possible body widths.[A] The standard distances and the corresponding magnification are depicted in Figure 5-2.

Perhaps the best solution to the magnification problem is to express the linear displacements as a

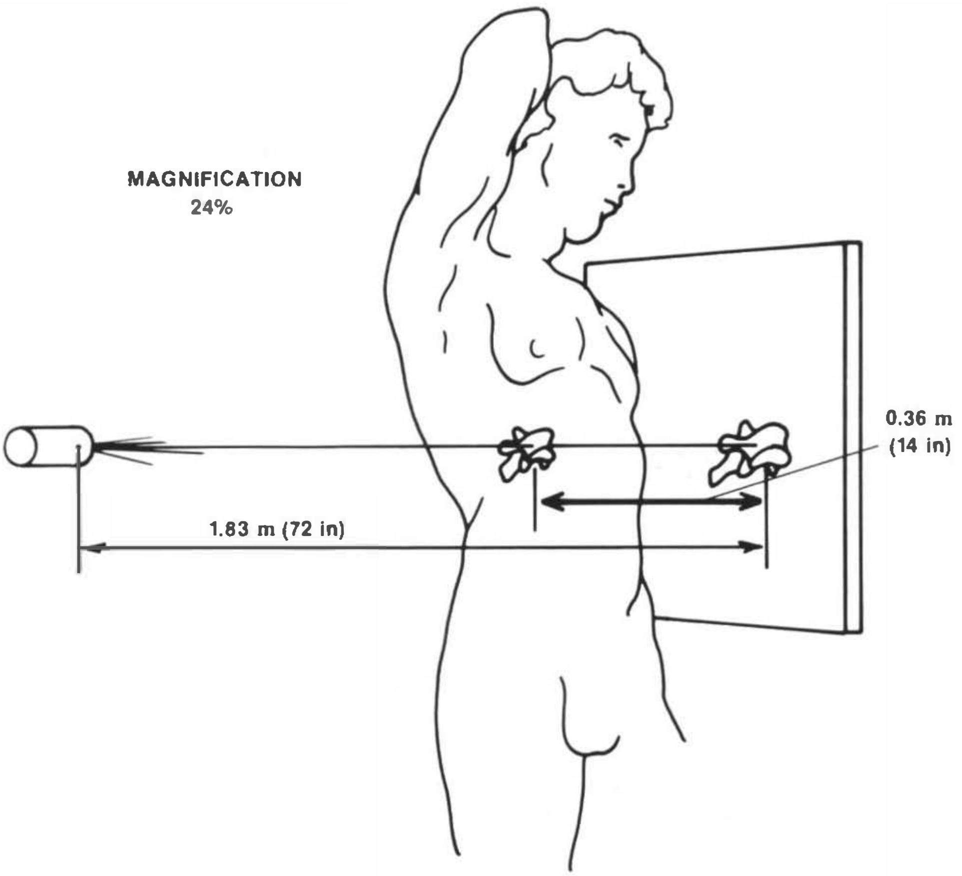


FIGURE 5-2 Suggested standard distances to minimize errors in linear radiographic measurements. With these source–film and spine–film distances, the linear dimensions of the vertebrae are magnified by 24%. The angular measurements are not magnified during radiography.

percentage of some other distance on the same x-ray (see Fig. 5-35).

Angular Radiographic Measurements

In contrast to linear measurements, angular radiographic measurements are true representations of the object. For example, the angle between two endplates in the sagittal plane is precisely the same as the corresponding angle measured on a *true* lateral.

Muscle Forces and Clinical Stability

The role of the muscles in clinical stability has not been fully evaluated. It is our view that the muscles offer a small amount of protection through splinting in the acute phases of injury. Furthermore, in the less acute situation, and against the normal range of physiologic loading, the muscles do not play a significant role. For example, in polio patients with total paralysis of cervical muscles, there is no loss of clinical stability as long as the bony and ligamentous structures remain intact.[122] Although it would be better to analyze and quantify the role of muscles in clinical instability, we feel justified in our endeavor to analyze clinical instability without full knowledge of the exact role of the muscle forces exerted. It is certainly possible that voluntary and reflex muscle activity in response to pain may be operative in the acute phase.

Biomechanics of the Spinal Cord and Nerve Roots

There are several points about the neural elements that are worthy of review prior to a review of the specifics of clinical stability in the various regions of the spine. Breig[20] showed that, contrary to popular belief, the cord does *not* slide up and down in the spinal canal during flexion and extension and other motions. The cord and its elements are elastic and deform like an accordion as the dimensions of its protective canal change with motion (Fig. 5-3). Dorsal extension tends to stretch the spine anteriorly and shortens it posteriorly, and with lateral flexion, it stretches on the convex side and shortens on the concave side. This characteristic pattern of deformation is true of all the structures, including the axis cylinder, the blood vessels, the glial membrane, and all ligaments and meningeal components.

A cogent work by Reid,[129] also published in 1960, essentially agrees with Breig's findings but has some minor differences. Reid studied 18 necropsy specimens in which he observed movement of the spinal cord and nerve root as well as measured anterior pressures. Like Breig, he noted stretching of the neural elements with flexion and extension. The maximum stretching occurred between C2 and T1 in general. The average stretch was 10%, and the specimen with the greatest change measured 17.6%. There was some up and down (piston) movement noted in the region of C8–T5, and the maximum amount was 1.8 cm. The anterior pressures were maximum at 2.0–2.7 MPa (30–40 lb/in^2). The author concluded that mechanical changes associated with movement probably do affect myelopathy and radiculopathy.

Injuries to the cord and the nerve result from loss of elasticity of the cord, pathologic displacement between two or more vertebrae, or protrusions into the canal or posterior fossa. The spinal cord shows good elasticity and compliance in the axial direction. It is relatively much less able to accommodate deformation in the horizontal plane, which accounts for its high vulnerability to translatory displacements of the spine in that plane (see Fig. 5-74). The physiologic effects of cord trauma are discussed in Chapter 4. The main consideration is that one episode of force application can initiate a chain of events that begins with rupture of arterioles and venules in the central gray matter followed by hemorrhage, edema, and loss of spinal cord function.

As the reader completes this chapter, he will note that the analyses of instability can be broadly grouped as outlined in Table 5-2. Kinematic instability focuses on *quantity* of motion (i.e., the observation of too much or too little motion). Other kinematic analyses focus on *quality* of motion. Examples are: changes in the patterns of distribution of the instantaneous axes of rotation (centrodes, instant centers of motion); changes in coupling characteristics, either their extent or direction; and other possible major directional changes, such as with paradoxical motion.

By contrast, in component instability the analysis addresses the clinical biomechanical role of the various anatomic components of the functional spinal unit (FSU). The effects of the loss or alteration of various portions of the anatomy are considered in the determination of instability. Obviously, anatomic components may be altered by trauma, tumor, surgery, and degenerative and developmental changes.

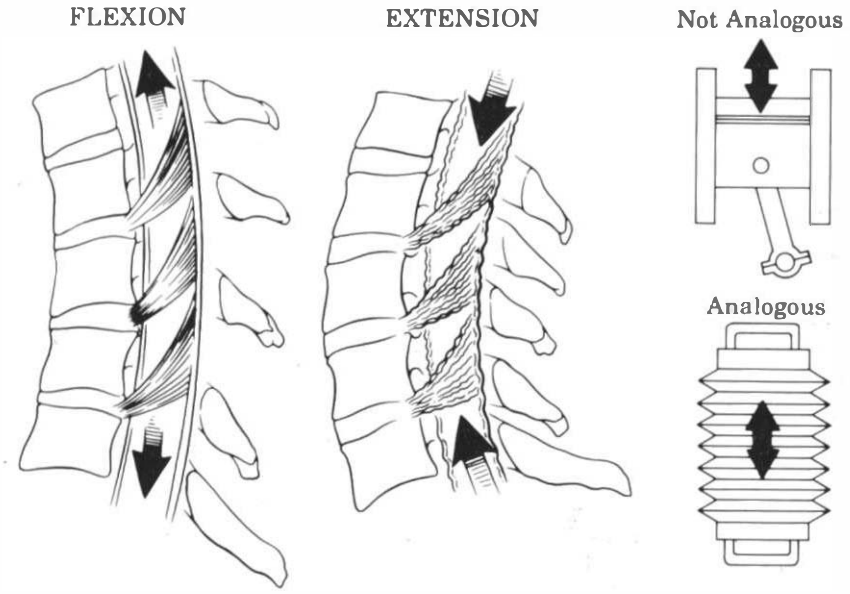


FIGURE 5-3 This diagram depicts the ability of the spinal cord and nerve roots to adjust to physiologic changes in distances within the spinal column. The capacity of these structures to withstand the accordion-like deformation in the axial direction is an important biomechanical characteristic of the cord and its associated structures. The literature shows that there is in fact a *small* degree of pistoning effect, i.e. cephalocaudal displacement of the neural structures within the canal.

TABLE 5–2 Conceptual Types of Instability

- *Kinematic instability*
  - Motion increased
  - Instantaneous axes of rotation altered
  - Coupling characteristics changed
  - Paradoxical motion present
- *Component instability*
  - Trauma
  - Tumor
  - Surgery
  - Degenerative changes
  - Developmental changes
- *Combined instability*
  - Kinematic
  - Component

Finally, instability may be analyzed in a combined approach considering both the *kinematic* and the *component* instabilities. This has been the authors' approach, as represented in the checklist.

PART 1: OCCIPITAL-ATLANTO-AXIAL COMPLEX (C0–C1–C2)

We have sometimes combined C0, C1, and C2 in our discussions. In other instances, the complex has been separated into C0–C1 and C1–C2. The separation generally has been done to make an analysis more comprehensible, or simply because it is the tradition. In fact, the complex is a 3-unit articulation, and it is preferable to study it that way, particularly as our knowledge and understanding improve.

OCCIPITAL-ATLANTAL JOINT (C0–C1)

With the possible exception of the terminal coccygeal joint, the occipital-atlantal joint has received less attention than any of the articulations in the axial skeleton. This generalization seems to hold for anatomic as well as biomechanical and clinical studies. However, Dvorak and co-workers recently completed some cogent research on the functional anatomy of this complex.[43-45]

ANATOMIC CONSIDERATIONS

The anatomic structures that provide stability for this articulation include the cup-shaped configuration of the occipital-atlantal joints and their capsules (Fig. 5-4), along with the anterior and posterior atlantooccipital membranes. The ligamentum nuchae should be included here, although its significance as a stabilizing structure in the human spine is controversial.[52,78,79] Additional anatomic stability is gained through the ligamentous connections between the occiput and the axis. This is achieved through the tectorial membrane, the alar ligaments, and the apical ligaments, which are of dubious mechanical significance.[68] Basing their opinion on the structural characteristics, the authors believe that the occipital-atlantal joint is relatively unstable, at least in the child. There may be some increase in stability in adult life due to a decrease in elasticity of the ligaments. The limited studies of the mechanics of this articulation can be reviewed in Chapter 2. Dislocations of this joint are usually fatal. It seems likely that a number of these injuries are followed by instant death and are not discovered at autopsy. The level of the anatomic lesion is such that, unless resuscitation is instituted in a very short period of time, the victim dies. However, there is little in the literature to document the clinical characteristics of instability at this joint.

A 14-year-old who was hit from behind by an automobile while riding his bicycle survived for over a year on a respirator (Fig. 5-5). Cephalad progression of medullary damage destroyed the lower cranial nerves, rendering him unable to swallow, talk, or chew. Another survivor has been reported by Evarts.[48]

BIOMECHANICAL FACTORS

Wiesel and Rothman have shown that the normal range of sagittal plane translation in flexion/extension does not exceed 1 mm.[169] This measurement is made between the basion of the occiput and the tip of the odontoid. Hypermobility may be seen in association with a congenital lesion of C1–C2 and C3–C4. Keeping in mind our present evaluations of the anatomy of this point and the dangerous risks involved in its displacement, we suggest that any dislocation or subluxation be considered unstable. A finding of more than 5 mm between the tip of the

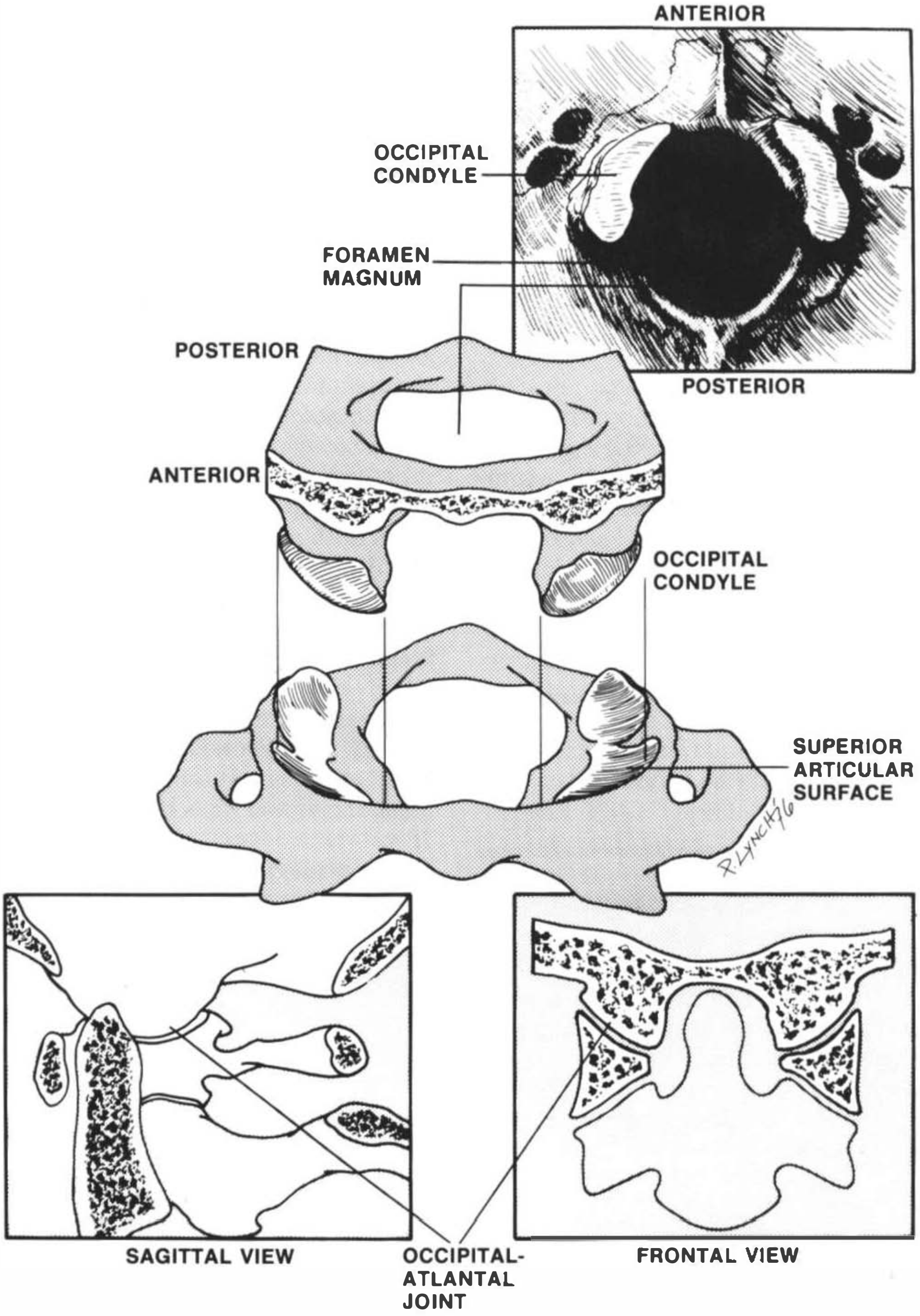


FIGURE 5-4 The three-dimensional anatomy of the cuplike articulations of the occipital-atlantal joints. The cup is relatively more shallow in the sagittal than in the frontal plane. Consequently, the joint is probably more unstable in anteroposterior displacement or dislocation than in lateral displacement.

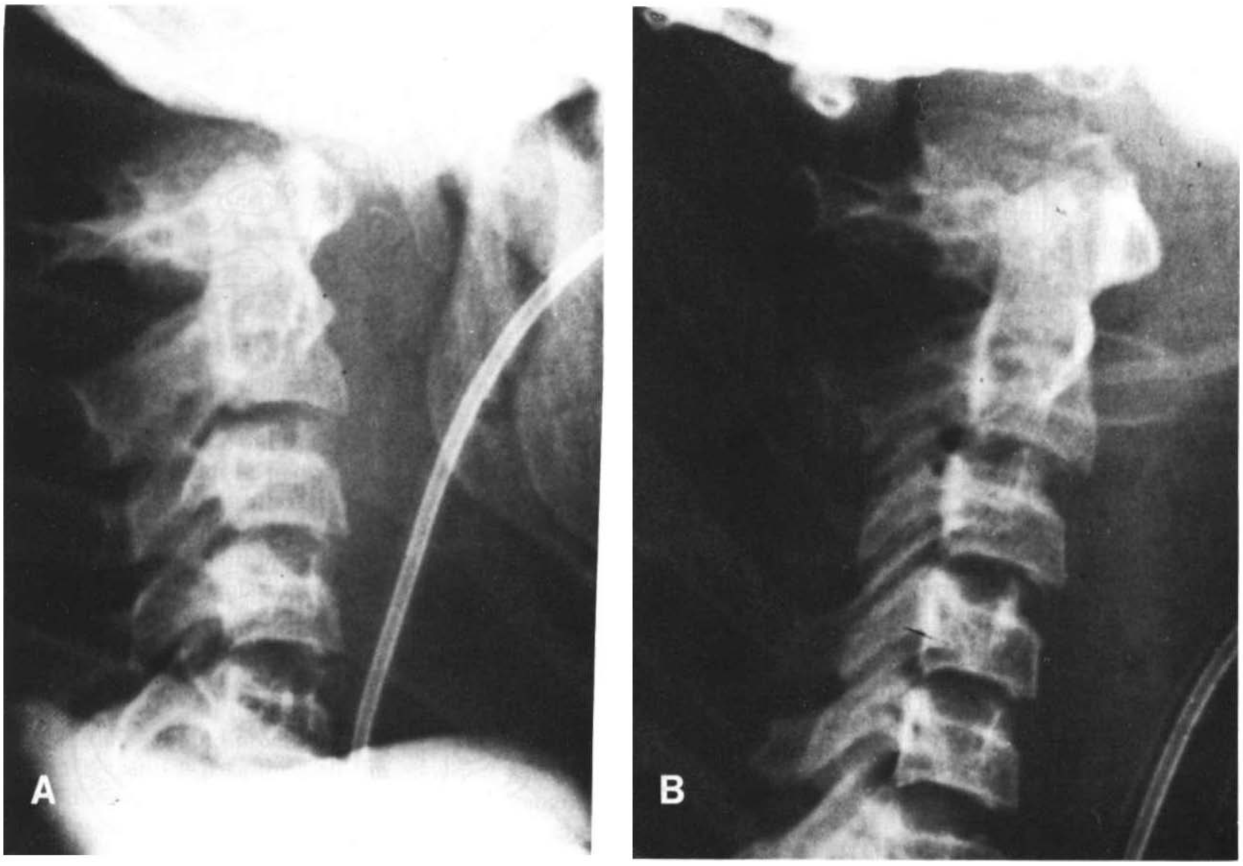


FIGURE 5-5 (A) Anterior dislocation of the occiput on C1. (B) Axial traction resulted in + y-axis displacement suggestive of total disruption of the intervening ligaments. *(Courtesy of Children's Hospital, Akron, Ohio.)*

dens and the basion of the occiput or more than 1 mm of translation in flexion/extension is an important and useful criterion. Symptoms of weakness of the limbs with or without associated neck and occipital pain provide additional indications of instability. The criteria are listed in Table 5-3, and the measurement is shown in Figure 5-6.

TABLE 5–3 Criteria for C0–C1–C2 Instability

>8°	Axial rotation C0–C1 to one side
>1 mm	C0–C1 translation (as measured in Fig. 5–6A)
>7 mm	Overhang C1–C2 (total right and left)
>45°	Axial rotation C1–C2 to one side
>4 mm	C1–C2 translation (as measured in Fig. 5–6B)
<13 mm	Posterior body C2—posterior ring C1 (as measured in Fig. 5–6C)
Avulsed transverse ligament	

CLINICAL CONSIDERATIONS

There is another measuring technique to show abnormal anteroposterior C0–C1 translatory displacement. For ease of recall, this can be called the BAOC. Shown in Fig. 5-7, the measurement is actually a ratio of BC/AO.[126] The distance from the basion to the posterior arch of C1 divided by the distance between the anterior arch of the atlas and the opisthion should be 1 or less. If the ratio is greater than 1, there is probably a C0–C1 instability or dislocation.

Basilar Invagination

Vertical or y-axis translation is also an important consideration, because too much cephalad displacement is indicative of basilar invagination.

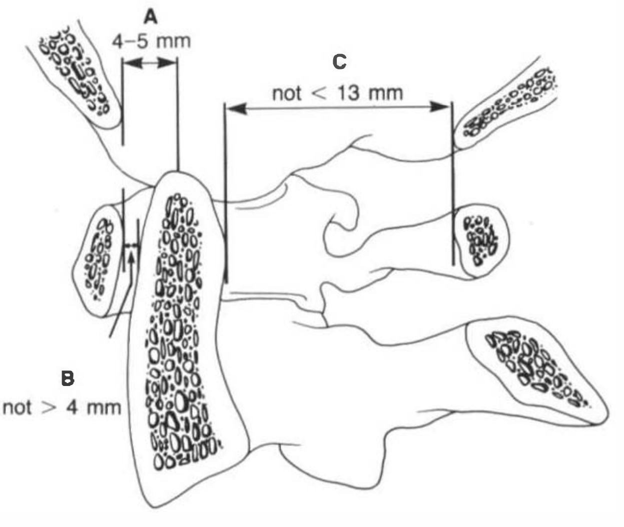


FIGURE 5-6 C0–C1–C2 lateral view. **(A)** The distance between the basion of the occiput and the top of the dens is 4–5 mm. An increase of more than 1 mm in this distance with flexion/extension views is believed to indicate instability of C0–C1, if one assumes that the transverse ligament of the atlas is intact. **(B)** The distance between the anterior border of the dens and the posterior border of the ring of C1 should not be greater than 4 mm. **(C)** There is another important measurement that we must consider. We refer to the distance between the posterior margin of the dens and the anterior cortex of the posterior ring of C1. This distance is of concern should it be less than 13 mm. *(Modified from White, A. A., Panjabi, M. M., Posner, I., et al.: Spinal stability: evaluation and treatment. AAOS Instructional Course Lectures. vol. 30. St. Louis, C. V. Mosby, 1982.)*

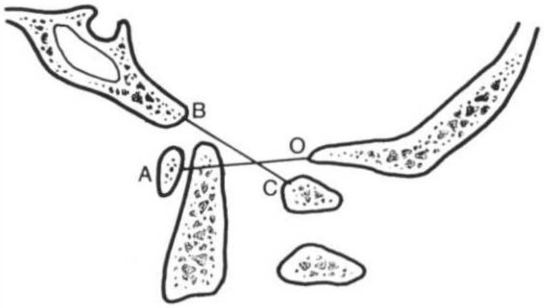


FIGURE 5-7 Diagrammatic representation of normal measurements that can be used to detect atlantooccipital dislocation. *A*, anterior arch of C1; *B*, basion; C, posterior arch of C1; *O*, opisthion. The ratio of *BC* to *AO* is normally less than 1. A ratio greater than 1 is positive for atlantooccipital dislocation. This is valid only in the absence of associated fractures of the atlas. This ratio may not be helpful in some cases of longitudinal or posterior distraction. *(Reproduced with permission from Kricun, M. E.: Imaging modalities in spinal disorders. Philadelphia, W. B. Saunders, 1988.)*

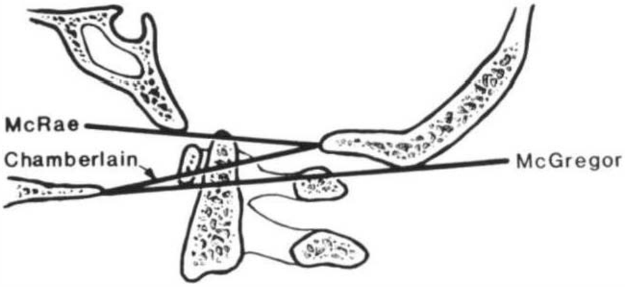


FIGURE 5-8 Diagrammatic representation for measurement of basilar invagination. *(Reproduced with permission from Kricun, M. E.: Imaging modalities in spinal disorders. Philadelphia, W. B. Saunders, 1988.)*

McGregor's line is drawn from the hard palate to the lowest point of the occiput.[98] More than 4.5 mm of protrusion of the tip of the dens above this line is considered abnormal (Fig. 5-8). If the dens is abnormally displaced above that line, there may be associated neurologic problems. This occurs most commonly in rheumatoid arthritis but may also be present with tumor or trauma. Chamberlain's line extends from the dorsal margin of the hard palate to the dorsal lip of the foramen magnum. The normal position of the dens in relation to this line is between 1 mm below it and 0.6 mm above it.[136] McRae's line extends[99] from the basion to the posterior lip of the foramen magnum. These lines are depicted in Figure 5-8. It is presumed that if the tip of the dens is *above* MacRae's line, there may be symptomatic basilar invagination. The digastric line is shown in Figure 5-9.[8] This line is thought to represent the limit above which the placement of the tip of the dens represents basilar invagination. There are a considerable number of variables that render the measurement and use of these lines less than completely reliable. The numerous considerations are reviewed thoroughly by Kricun.[87] Better clinical correlations with specific measurements are required to determine which of these measurements are most useful. At present, the use of all four combined with careful judgment and documentation of associated neurologic deficit is our recommendation.

With our present knowledge of the structure of this joint, and the dangerous anatomic risks in-

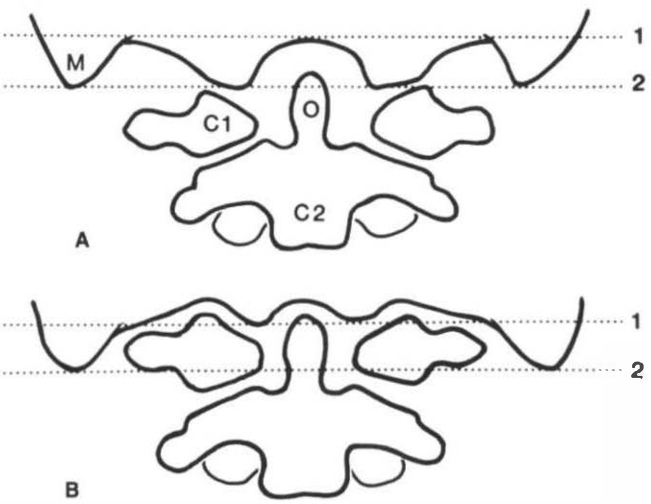


FIGURE 5-9 Digastric line. Diagrammatic representation of the atlantooccipital articulation and a measurement for basilar invagination. *1*, digastric line joins the digastric grooves; *2*, mastoid line joining the tips of the mastoid processes; *M*, mastoid process; *O*, odontoid process. **(A)** Normal appearance. The digastric line *1* is well above the odontoid process, and the mastoid line *2* passes through the tip of the odontoid process. **(B)** Basilar invagination. Both lines *1* and *2* pass through the odontoid process. *(Reproduced with permission from Kricun, M. E.: Imaging Modalities in Spinal Disorders. Philadelphia, W. B. Saunders, 1988.)*

volved in its displacement, we suggest that any dislocation or subluxation be considered clinically unstable. The treatment that we recommend is posterior fusion, C0 to C1 or C2, followed by immobilization for 3 months in a halo device attached to a thoracic jacket.

The time and extensiveness of external fixation can be reduced considerably if some form of stable internal fixation is employed. We found no documented clinical reports of isolated C0–C1 fusions; however, in principle, one should seek to fuse the least number of FSUs that will solve the problem. In the absence of definitive data, this is better left to the individual surgical judgment.

ATLANTO-AXIAL JOINT (C1–C2)

The C1–C2 articulation is the most complex and difficult one to analyze. Both the basic and clinical literature concerning this area are highly controversial, and sometimes confusing. The most valid information relating to the problem of clinical instability in this area is discussed here.

ANATOMIC CONSIDERATIONS

Various aspects of the anatomy of the atlanto-axial region are presented in Chapters 2 and 4.

One of the key variables in the problem of clinical instability is that of allowable displacement without neurologic deficit. This is partially dependent upon the normal sagittal diameter of the spinal canal, which has been studied in 200 normal adult subjects by Wolf and colleagues.[173] Eismont and associates have completed measurements of the anteroposterior diameter of the cervical spinal canal. This measurement is important in regard to clinical stability trauma, cervical spondylosis, disc disease, and ossification of the posterior longitudinal ligament. This measurement is a major variable in any mechanical or pathophysiologic situation in which neurologic signs and symptoms of the cervical region are involved. This is discussed in Chapter 8 in the clinical biomechanics of cervical spondylotic myelopathy.

There are two important measurements at the C1–C2 level that are significant in regard to clinical instability. The two measurements are essentially complementary. One measures the sagittal plane distance between the back of the anterior ring of C1 and the anterior portion of the dens. The upper limits of normal are 3–4 mm. The other measurement is the sagittal plane distance between the posterior dens and the anterior portion of the posterior ring of C1. This measurement is considered abnormal if less than 13 mm (Fig. 5-6).[145] Other sagittal plane diameters are presented in Figure 5-10. Similar data for children are also available.[71] The canal size can also be reduced by the presence of a bone spur or osteophyte at one or more levels (Fig. 5-11). This imposes a variety of mechanical loads on the spinal cord (see Fig. 1-44, p. 72). The osteophyte reduces space available in the canal and causes some impingement on it. These mechanical factors play a focal role in the clinical outcomes associated with trauma and cervical myelopathy.

The bony and cartilaginous articulations of the facet joints, with their biconvex configurations, are held together by a loose articular capsule designed to permit a large range of joint motion. Consequently, the capsule and osseous configuration contribute little to the clinical stability of the joint. Rather, the major mechanical stability of this articulation is provided through the dens and the ring formed by the anatomic structures surrounding it.

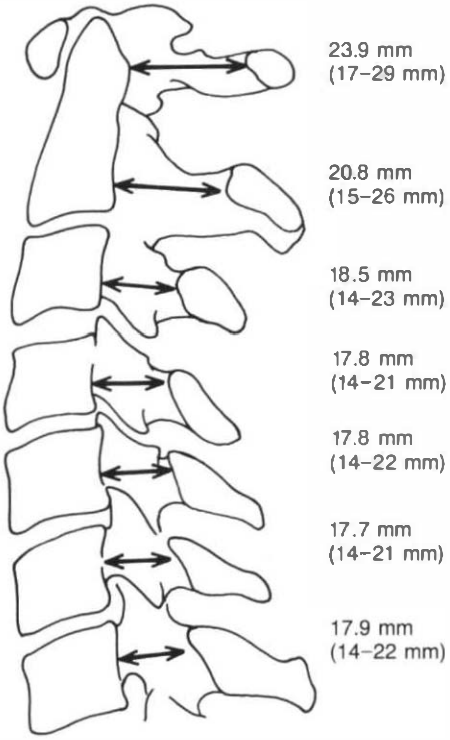


FIGURE 5-10 The true sagittal plane diameter of the cervical spine canal. The upper portion has relatively more space for the spinal cord, even though the cord is larger there. The means are presented for each level, with the range presented in parentheses. We must always consider that clinical x-ray measurements have a 20% to 30% magnification of the *real* linear distances. (*Measurements from* Eismont, *F. J.,* Clifford, *S.,* Goldberg, *M., and* Green, *B.: Cervical sagittal spinal canal size in spine injury.* Spine *9:663, 1984.* Other studies report *means 1–2 mm greater than these.*[90])

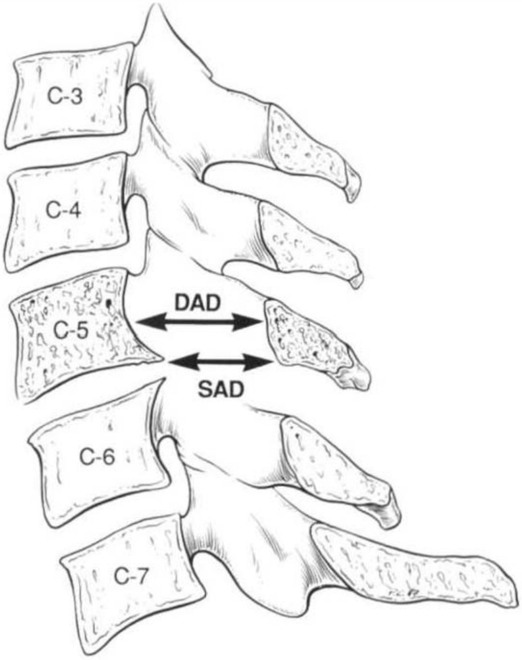


FIGURE 5-11 This is an illustration of two important concepts—developmental anteroposterior diameter (DAD) and the spondylotic anteroposterior diameter (SAD). These are generally measured as simple linear diameters. Because of differences in source-to-object-to-film distances, there is potential for variation. In general, 14 mm can be chosen as a representative for the lowest limits of normal. The SAD is an important measurement. The developmental diameter may not be the important functional diameter if there is an osteophyte (spur) that significantly reduces the canal space.

These consist of the osseous portion of the atlas anteriorly and laterally and the transverse ligament posteriorly. Virtually all of the other anatomic structures play a secondary role in the stability of this joint. The strong yellow ligament is not present between C1 and C2. Instead, there is the weaker atlanto-axial membrane.

Anterior Longitudinal Ligament

This structure is a continuation of the ligament that runs the entire length of the spine. It is well developed in the thoracic and lumbar regions and is described as a thin, translucent structure in the cervical region.[79] It begins at the anterior body of C2, attaches to the anterior ring of C1, and courses to the tubercle of C0. Little is known about the mechanical properties of this structure in this region of the spine.

The Anterior Atlanto-Dental Ligament and the Atlantooccipital Membrane

These structures are shown in Figure 5-12. The atlanto-dens ligament runs between the anterior portion of the dens and the caudal portion of the anterior ring of C1. The atlantooccipital ligament runs between the cephalad portion of the anterior ring of C1 and the tubercle of C0. This is thought to be a continuation of the anterior longitudinal ligament. The posterior atlantooccipital ligament or membrane

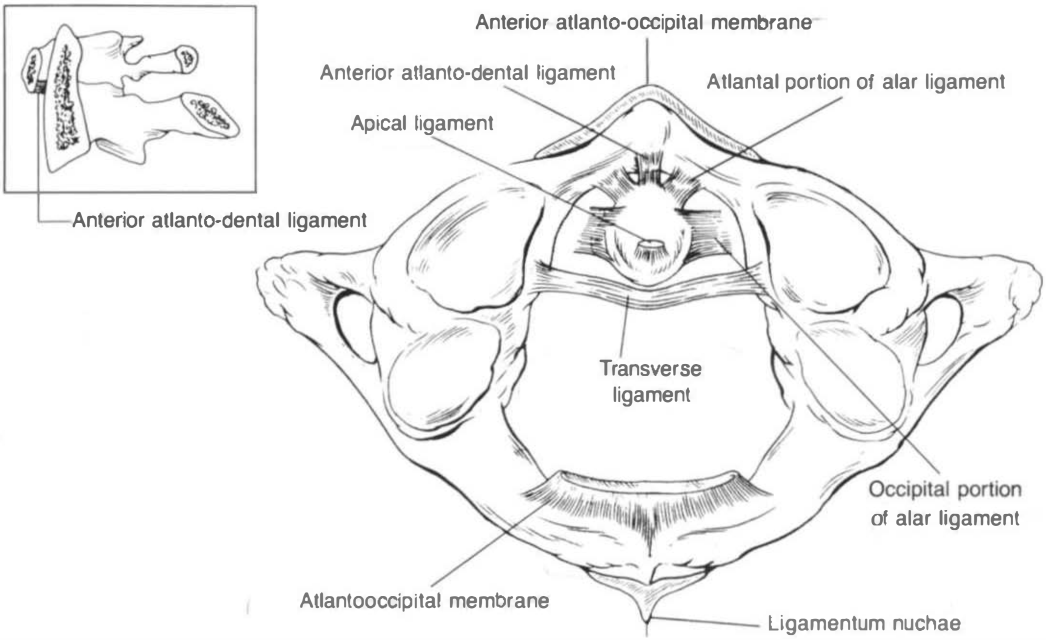


FIGURE 5-12 Schematic illustration of the major ligaments involved in the clinical stability of the upper cervical spine. The anterior atlanto-dental ligament has been described recently.[44] *(Dvorak, J., and Panjabi, M. M.: Functional anatomy of the alar ligaments. Spine, 12:183, 1987.*

connects the posterior ring of C1 to the occiput at the posterior ring of the intervertebral foramen. In a recent study, the atlantooccipital ligament was tested and found to fail at 265 N (±132 N) of tension.[123a] We can observe that when they are both intact, they offer some modest mechanical advantage in preventing anterior displacement of C1 and C2.

Dentate (Odontoid) Ligaments

These ligaments consist of the alar ligaments and the apical ligaments. The alar ligaments are a pair of structures that are attached to the dorsolateral surfaces of the tip of the dens, and each runs obliquely to the medial surfaces of the occipital condyles (Fig. 5-12). The two structures form an angle of 140–180°, the apex of which points caudally (see Fig. 5-15C).[88] The left alar ligament limits rotation of C1 and the head to the right (−y-axis), and the right alar ligament limits rotation to the left. The apical dentate ligament connects the apex of the dens to the anterior edge of the foramen magnum. It has been described by Hecker as a fairly strong structure of good development and of elastic consistency. Thus, it may be expected to contribute little to C0–C1 stability and nothing to C1–C2 stability.

The work of Dvorak and Panjabi on the functional anatomy of the alar ligaments includes some detailed observations.[44] Particularly significant was the identification of specimens in which the anterior atlanto-dental ligament was described for the first time. The structure shown in Figure 5-12 (inset) connects the base of the dens with the anterior arch of the atlas. Although the exact functional and clinical significance of these recently described ligaments has not been demonstrated, they may play a role in marginal instability of C1–C2. The "V sign" discussed on page 296 may be created by an intact anterior atlanto-dental ligament in association with a partially ruptured or elongated transverse ligament (Fig. 5-13). Fielding and co-workers[53] reported this "V sign" in 1974 and interpreted it as representative of a rupture of all but the lower fibers of the transverse ligament.

It has been hypothesized that the alar ligaments that limit the axial rotation and side bending of the C0–C1–C2 complex may be stretched or partially

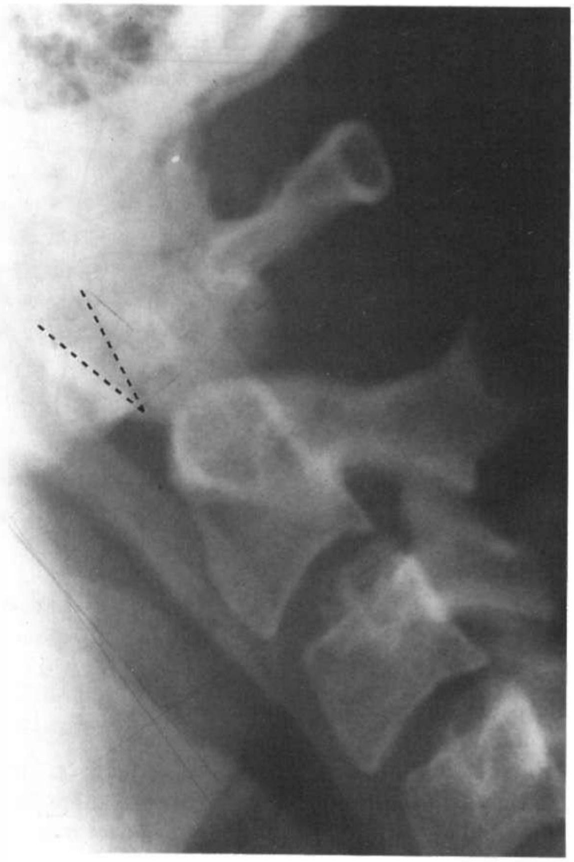


FIGURE 5-13 Roentgenogram of a 20-year-old male following an automobile accident in which he was the driver in a head-on collision. He hit his head on objects inside his auto, which forced his neck into extension. There were no neurologic symptoms, and the neurologic and orthopedic examinations were normal. He had persistent neck pain that subsided, and he gradually returned to normal activities. This "V sign," with tilting of C1 in relation to the dens, is thought by some writers to indicate instability. However, other writers disagree. We suggest that it may be due to congenital absence or laxity of the ligaments. See Figure 5-12 for the anatomy. We do not believe that this condition represents clinical instability.

ruptured in the whiplash-type mechanism of injury in which the head is axially rotated and flexed. This may lead to instability of the C0–C1–C2 complex.[44]

Dvorak and associates, in an *in vitro* computerized tomography (CT) study of the C0–C1–C2 complex, demonstrated the role of the alar ligament in limiting axial rotation to the opposite side.[45] Moreover, it was shown that transection of the left alar ligament resulted in a 30% increase in right (− y-axis) axial rotation of the occiput. This was rotation of C0 in relation to C2, divided equally at the two levels. Based on measurements of *in vivo* functional CT scans of C0–C1–C2 in 43 spine-injured patients and 9 normals, the investigations suggested guidelines for measuring and identifying normal and abnormal rotations.[43] The suggestion is that axial rotation between C0 and C1 greater than 8°, or between C1 and C2 greater than 56°, is abnormal. Also, a right–left difference in rotation at C0–C1 >5°, at C1–C2 >5°, or at C0–C2 >8° is thought to represent excessive motion. These findings may constitute a clinically significant rotatory instability; however, a complete clinical entity has not yet been fully defined.[43]

The Cruciate Ligament

The major portion of this ligament is the transverse ligament, which is the most important ligament of the C0–C1–C2 complex. This structure attaches to the two condyles of the atlas. There is an ascending and a descending band of this ligament. Both bands are triangular in shape. The ascending portion attaches to the anterior edge of the foramen magnum, and the descending portion attaches to the body of C2. The transverse band is the largest and the strongest. Its central portion has a thickness of 7–8 mm. The ascending and descending bands are 3–4 mm thick at the midportions (see Fig. 5-15B).[68]

Studies of horizontal translation showed that an anterior dislocation of C1 on C2 can occur with an insufficiency of the transverse ligament only. The biomechanical studies by Fielding and co-workers on the transverse ligament showed that although the structure was very weak in some subjects, when present it prevented more than 3 mm of anterior displacement of C1 on C2.[53]

Tectorial Membrane

This structure is a continuation of the posterior longitudinal ligament. It runs from the body of C2 up over the posterior portion of the dens and then makes a 45° angle in the anterior direction as it runs toward the attachment to the anterior edge of the foramen magnum (see Fig. 5-15A). The anterior portion of the dura and the spinal cord completes the description of the anterior elements.

Posterior Atlantooccipital and Atlanto-Axial Membranes

These structures are anatomically analogous to the yellow ligament. However, they are considerably different in physical properties. The posterior atlanto-occipital membrane attaches to the posterior ring of C1 and the posterior portion of the foramen magnum (Fig. 5-12). The posterior atlanto-axial membrane attaches to the posterior ring of C1 and C2. The yellow ligament is first present between the lamina of C2 and C3. It then continues to the sacrum.

Individuals and textbooks sometimes make the error of assuming that there is yellow ligament between the posterior elements of C0 and C1 and/or C1 and C2. A method for remembering that it is not there is to think about the large axial rotation (80°) that occurs between C1 and C2. The highly elastic but rather stiff yellow ligament would never allow this.

Nuchal Ligament

This is a triangular structure that is divided into a funicular and a triangular portion (Fig. 5-14). The funicular portion consists of a distinct band that runs from the posterior border of the occiput to the spine of C7. The lamellar portion divides the posterior neck into right and left halves. Its superior border attaches to the funicular portion, and anteriorly it attaches to the posterior tubercle of the atlas, the spinous processes of the cervical vertebrae, and the interspinous ligaments.[52] Although this ligament is thought by some investigators to have biomechanical significance in the neck, the precise role has not yet been clearly delineated.

This ligament may play an important role in the clinical biomechanics of the neck. One hypothesis is that it plays a major proprioceptive role in the appropriate functioning of the erector spinae muscles. Note Sherk and Dawoud's studies on progressive kyphosis following posterior surgical approaches to the neck.[142] Damage to this structure may be a factor in some whiplash-type injuries. Now that these ligaments can be well visualized with magnetic resonance imaging (MRI), perhaps these and other questions about their function and clinical significance will be answered.

Anatomic Interdependence and Clinical Stability

As stated earlier, C0–C1–C2 is a complex articulation that is oftentimes more appropriately "lumped together" than "split" into C0–C1 and C1–C2.

It is readily discernible from anatomic descriptions that there are a number of structures that provide some direct or indirect stability to both the occipital-atlantal joint and the atlanto-axial articulation. There are two groups of such structures. One group runs longitudinally, attaching to all three units, and includes the anterior longitudinal ligament, the tectorial membrane, the cruciate ligament, and the nuchal ligament. The other group offers some stability to both joints by skipping one segment. This includes the alar and the apical ligaments (Fig. 5-15).

Present knowledge does not permit a complete analysis of the role of interdependence in clinical stability. However, it is known that with failure of the structures that run between C0 and C1, at least some attachment of the occiput to the lower cervical spine remains through the apical and alar ligaments. Similarly, with the loss of only those structures between C1 and C2, some attachment of C2 to the occiput remains. Most clinically unstable injuries in this area probably destroy a number of structures in both articulations.

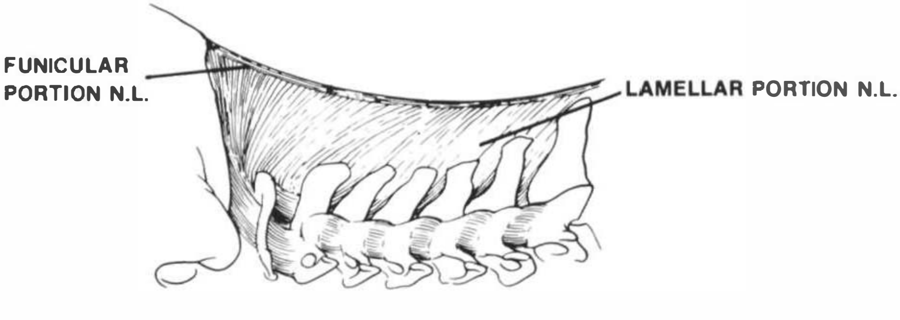


FIGURE 5-14 Diagram of the anatomy of the nuchal ligament. (*Fielding, J. W., Burstein, A. A., and Frankel, V. H.: The nuchal ligament. Spine, 1:3, 1976.*)

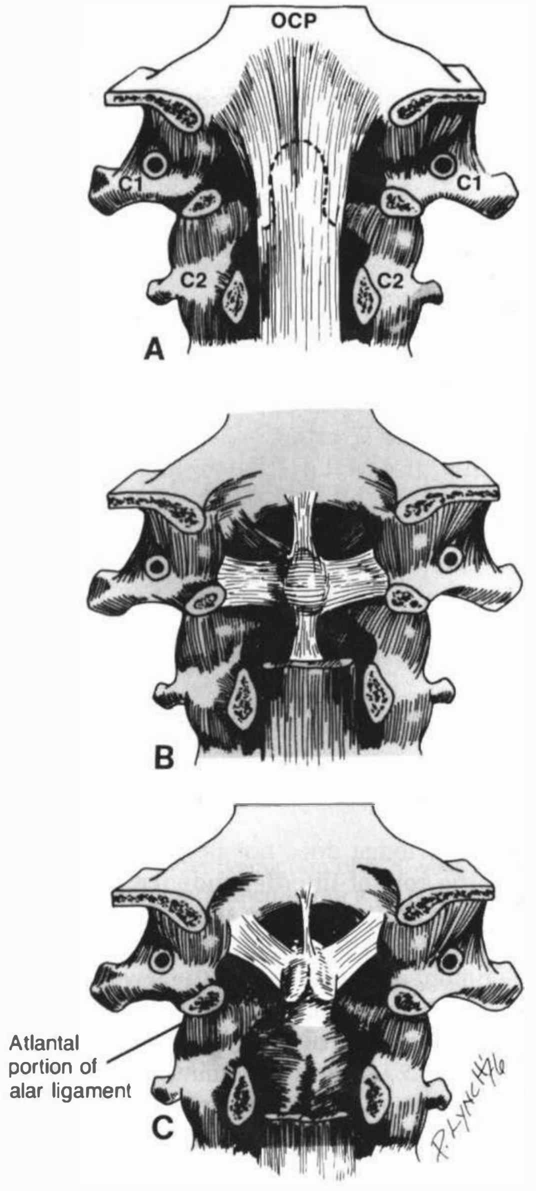


FIGURE 5-15 The major stabilizing ligaments of the occipital-atlanto-axial complex. The structures, seen here from the posterior, may be separated into three layers. **(A)** *Layer* one. When the posterior osseous and ligamentous structures are removed, the tectorial membrane is visualized. **(B)** *Layer two.* The next structure, moving anteriorly, is the cruciate ligament. The transverse ligament is a component of the cruciate and is the most important stabilizing ligament of the atlanto-axial complex. **(C)** *Layer three.* Anterior to the cruciate lie the apical and alar ligaments, which serve as secondary stabilizers. These ligaments contribute to the interdependence in the occipital-atlanto-axial complex by way of anatomic attachments.

BIOMECHANICAL FACTORS

The basic kinematics of the occipital-atlanto-axial joints are presented in Chapter 2. This section reviews other biomechanical data that may have relevance in a systematic approach to the problem of clinical instability.

In flexion of the C0–C1 joint, the limit is determined by impingement of the anterior margin of the foramen magnum on the dens. Additional flexion must then occur at the C1–C2 joint. This range is limited by the tautness of the tectorial membrane over the dens. Extension is also limited by the tectorial membrane.[160] Clearly, an additional amount of flexion would be expected to occur in this articulation upon failure of the tectorial membrane or failure of either the anterior portion of the foramen magnum or the dens. Werne confirmed these observations on a model that also showed little involvement of the tectorial membrane in inhibiting axial rotation.

Werne found that the cruciform ligament did not have any limiting effect on physiologic motion of the C0–C1–C2 complex. The ascending band is too delicate. The descending band was shown to allow 6–7 mm of vertical ($+y$-axis) translation before reaching its limit. The alar ligaments function together to check movements in axial rotation and lateral flexion.[160]

The C0–C1–C2 complex limits movement as follows. At C0–C1, flexion ($+\theta x$) is checked by osseous contact of the anterior ring of the foramen magnum on the dens. Extension ($-\theta x$) is restricted by the tectorial membranes, and lateral flexion ($\pm \theta z$) is checked by the alar ligaments. At the C1–C2 level, flexion is checked by the tectorial membrane, and extension is checked by the tectorial membrane and other posterior ligaments. Rotation is checked by the alar ligaments.[160] Although the cruciate ligament plays a small part in physiologic motion, Fielding showed that it is the most important structure in preventing abnormal anterior translation.[53]

Werne also carried out studies to evaluate the interdependence of the C0–C1–C2 articulation. He studied rotation of the C0–C1–C2 articulations in the sagittal plane before and after removal of the tectorial membrane. Although it was not true for all the specimens, the findings suggested that with loss of the tectorial membrane, there is an increase in flexion. Werne showed that if the alar ligaments were also transected, there was a "*luxation* of the *occiput*."[160]

Lateral flexion was studied in a similar manner, before and after transection of the tectorial membrane. Increased lateral flexion was noted here also, but it was less convincing than in the flexion/extension studies. Axial rotation was also studied, and it was concluded that the alar ligaments are mainly responsible for the limitation of axial rotation.

When vertical translation was studied, the findings showed that this movement was greater after division of the tectorial membrane. The alar ligaments did not play a role in resisting vertical translation, except in a few specimens.

The studies of horizontal translation showed that an anterior dislocation of C1 on C2 can occur with an insufficiency of the transverse ligament only. The alar ligaments and the tectorial membrane were not found to prevent dislocation after the transverse ligament was transected. If the alar ligaments happen to be short, as may be expected in persons over 25, they may possibly offer some restraint against gross dislocation. The tectorial membrane depends upon an intact transverse ligament to offer resistance to anterior translation. The biomechanical studies by Fielding on the transverse ligament showed that although the structure was very weak in some subjects, when present it prevented more than 3 mm of anterior displacement of C1 on C2. He also showed that the alar ligaments deform readily and are not capable of preventing additional displacement under loads that would rupture the transverse ligament.[53]

CLINICAL CONSIDERATIONS

Occipital-Atlantal Instability

Wiesel and Rothman[169] described a type of instability of C0–C1 indicating that >1 mm of sagittal plane translation of C0 in relation to C1 is abnormal (Fig. 5-6). This type of instability is probably rare but nevertheless is to be considered by the clinician evaluating a patient with pain and a variety of neurologic complaints, many of which may actually be bizarre. Most C0–C1 instabilities are post-traumatic and are associated with dire neurologic consequences and large displacements of the C0–C1 articulation (Fig. 5-5).

Os Odontoideum and Instability

Os odontoideum is thought by Fielding and colleagues to be traumatic in origin.[55] Whatever the origin, it can be associated with instability. Spierings and Braakman reported 37 cases with this condition.[145] When the distance between the posterior border of the body of C2 and the posterior arch of the atlas is less than 13 mm, these patients are considered clinically unstable (Fig. 5-16). If there are neurologic signs or symptoms associated with this measurement, surgical stabilization is indicated. The management here is somewhat analogous to that of the rheumatoid patient with C1–C2 subluxation. If there is an indication of neural irritation, surgical stabilization is necessary. If not, then it may or may not be desirable, depending on other considerations, such as anticipated loads, severe pain, progressive deformity, or subluxation above a Klippel-Feil anomaly.[142]

Comminuted Fracture of the Ring of C1 (Jefferson Fracture)

On the open-mouth view of the odontoid, with the head in neutral rotation, the normal radiograph does not show any overhang of the lateral masses of C1 in relation to the lateral border of the body of the second cervical vertebra. However, with a Jefferson fracture, there is overhang on both sides. If the total overhang from the two sides is as great as 7 mm, then there is presumably also a rupture of the transverse ligament (Figs. 5-17 and 5-18). When these conditions are present, there is clinical instability.[144]

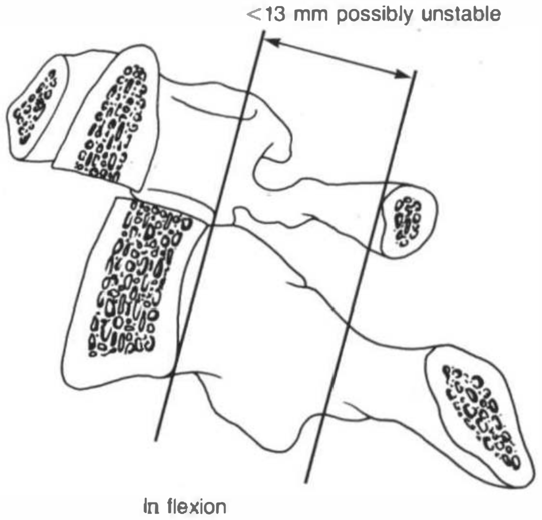


FIGURE 5-16 Os odontoideum with significant anterior displacement and less than 13 mm of space between line along posterior body of C2 and the anterior portion of the posterior ring of C1. (This illustration is not drawn to anatomic scale.)

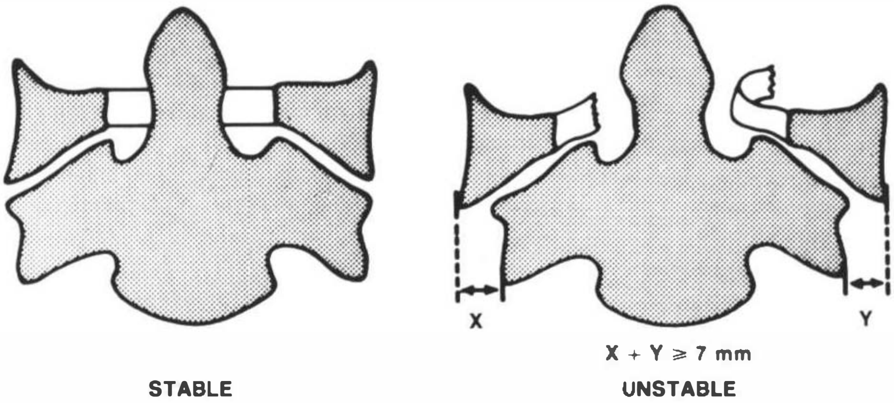


FIGURE 5-17 Jefferson fractures. When a comminuted fracture of C1 shows bilateral overhang of the lateral masses that totals 7 mm or more, a rupture of the transverse ligament has probably occurred, rendering the spine unstable.

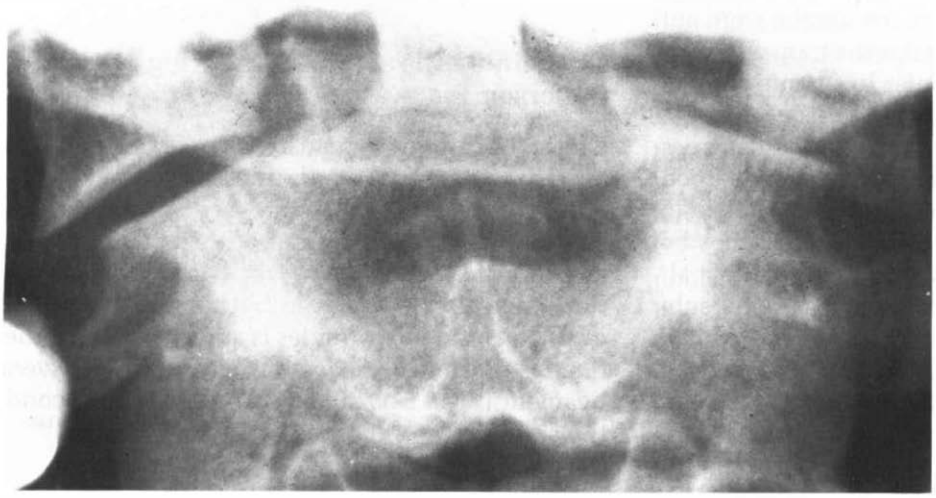


FIGURE 5-18 This is a three-part fracture of the ring of C1, which shows significant bilateral overhang of the lateral masses of C1 on C2. This is thought to be associated with rupture of the transverse ligament and instability. Some surgeons consider this fracture stable, however. *(Roentgenogram courtesy of George Alker, M.D.)*

Subluxations and Dislocations at the Atlanto-Axial Joint

With biomechanical experiments and a report on 11 cases, Fielding and colleagues showed that the clinical stability of the C1–C2 articulation depends upon an intact transverse ligament. This structure is, to some extent, supplemented by the alar ligaments, but in the absence of the transverse ligament the alar ligaments cannot be expected to prevent abnormal anterior translation of C1 on C2. All of the patients studied sustained severe head injuries in addition to or associated with injury of C1–C2. They all complained of neck pain for a variable period of time following injury. Three of the patients had neurologic symptoms. The average anterior displacement shown by radiographs of the patients was 7.2 mm. The investigators emphasized that fusion is probably the treatment of choice for this condition, because the potential risks of displacement in this area include quadriplegia and death.[53]

The clinical problem of subluxation and dislocation at C1–C2 is extremely complicated, controversial, and difficult to diagnose. The possible types of displacement have not been completely described and documented.

Based on our review of the literature, and our own analysis and evaluation, five patterns of abnormal displacement at the C1–C2 joint are submitted (see display). Two of the patterns are primarily translatory, and the other three are mainly rotatory.

Bilateral Anterior Displacement

This primarily translatory displacement may occur with a fractured or dysplastic odontoid, or an attenuated or ruptured transverse ligament (Fig. 5-19). There may be a history of a head or neck injury, or an

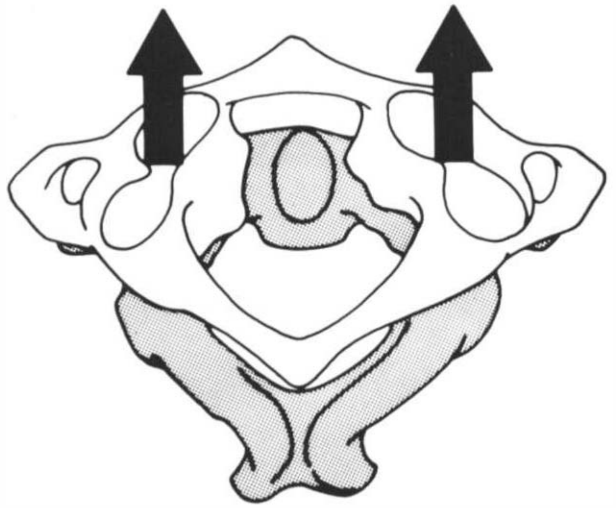


FIGURE 5-19 Bilateral anterior translatory displacement of C1 in relation to C2. This may occur with transverse ligament rupture or fractured, dysplastic, or aplastic odontoids.

PATTERNS OF SUBLUXATIONS AND DISLOCATIONS AT THE ATLANTO-AXIAL JOINT

- Translatory
  - Bilateral anterior
  - Bilateral posterior
- Rotatory
  - Unilateral anterior
  - Unilateral posterior
  - Unilateral combined anterior and posterior

---

impact to the trunk. In cases of transverse ligament inadequacy, there is usually a history of rheumatoid arthritis or an acute or chronic infection about the head or neck.[B]

The head may be in neutral or, because of muscle spasm, in the "cock robin" position, with some degree of flexion along with lateral and axial rotation. Truly lateral radiographs of C1, axial tomography, or computerized axial tomography shows an abnormal displacement of the ring of C1 anteriorly. These dislocations are clinically unstable when there is anterior displacement greater than 3 mm on radiographic examination with neurologic signs or symptoms. When they are due to transverse ligament disruption from pure trauma, they should be fused. If the laxity is secondary to infection, a trial of immobilization (halo apparatus) is recommended.[C] When the entity is associated with a fractured odontoid, either course is reasonable, depending on the probability of odontoid healing. The management of the various odontoid fractures is discussed in Chapter 4.

An unusual example of a type of bilateral anterior translatory displacement of C1 on C2 is shown in Figure 5-20. In this situation, the displacement occurred in the presence of a normal, intact odontoid. With a clear history of trauma along with a loud snap, the displacement was obviously the result of a rupture of the transverse ligament. This type of displacement should be considered clinically unstable because of very high risks of spinal cord damage

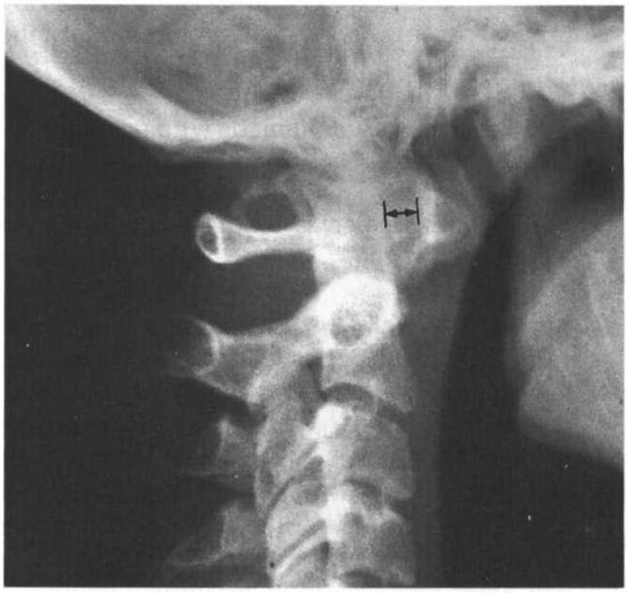


FIGURE 5-20 A clinical example of a bilateral anterior translatory displacement of C1 on C2. This is a radiograph of a healthy 21-year-old female who fell 4–5 ft onto her head from a climbing bar. She heard a loud snap and immediately had pain in her neck. There were no neurologic signs or symptoms. The displacement between the anterior dens and the posterior portion of the anterior ring of C1 measured 7 mm. This bilateral anterior translatory displacement should be considered clinically unstable. The radiograph shows another interesting finding. The little arch on the posterior ring of C1 has no known clinical significance but is frequently observed. It has at least two names, *posterior ponticle* and *foramen arcuale*. It is never seen in children but is partially or completely present in 12–16% of all adults.[163] The vertebral artery passes under it before entering the cranium. *(X-ray courtesy of John Wolf, M.D.)*

associated with additional anterior displacement in the presence of an intact odontoid. Moreover, the only remaining stabilizing structures are the accessory ligaments.[53]

The "V sign"

This finding most probably represents a normal variant and thus should not be considered unstable. Dvorak and colleagues[44] have recently discovered and described for the first time a set of ligaments, the *anterior atlanto-dental ligaments*. It may be that when this sign is noted following significant cervical spine trauma, the transverse ligament may be ruptured or elongated while the anterior atlanto-dental ligaments remain intact and prevent anterior displacement of the lower portion of the anterior ring of the atlas, causing the "V sign." An example of the "V sign" following trauma is shown in Figure 5-13.[49] If such a patient has no additional displacement with flexion and is asymptomatic with no history of related neurologic signs or symptoms, then careful follow-up is appropriate. If there is persistent pain or neurologic signs or symptoms, C1–C2 fusion should be considered.

Foramen Arcuale and Syndrome of Barre-Lieou

Barre-Lieou syndrome is one in which there are headaches, problems with phonation and swallowing, disturbances of vision, vasomotor face disturbances, and retro-orbital pain. The cause is thought to be alterations in flow of the vertebral artery and interference with the periarterial nerve plexus. The basic cause is thought to be spondylosis of the middle and lower cervical spine. Recently, Limousin[91] operated on 60 patients in whom vertebral artery kinking was considered to be caused by a foramen arcuale and was associated with Barre-Lieou syndrome. This radiographic finding is not uncommon (12–16% of adults), but Barre-Lieou syndrome is rare. Thorough documentation of mechanical kinking with arteriography is necessary to provide the important clinical documentation.

Bilateral Posterior Displacement

This is an extremely rare translatory dislocation (Fig. 5-21). It has been described in association with rheumatoid arthritis.[76] Such a dislocation is found only in the presence of a fractured dens, a dens destroyed by tumor or infection, a congenitally defective or absent dens, or a destroyed or congenitally absent anterior arch of the atlas. An example of this type of dislocation in the presence of a congenitally defective dens is shown in Figure 5-22. Sometimes patients with this condition provide their own clinical stability by holding their head with their hands. When this type of dislocation is seen with a fractured dens, it is treated according to the guidelines offered in Chapter 4. If the problem is due to an absent or destroyed dens, we suggest posterior fusion of C1–C2.

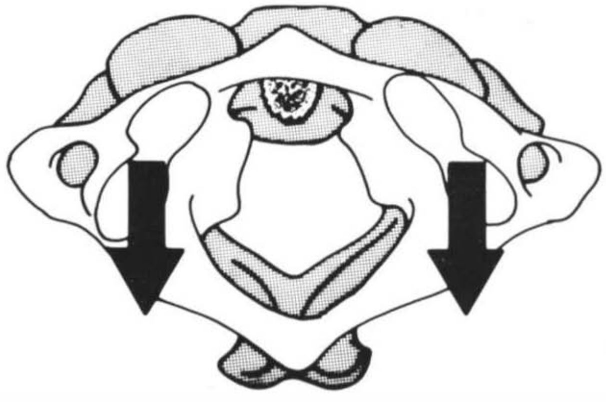


FIGURE 5-21 Bilateral posterior translatory displacement of C1 in relation to C2. This may occur with fractured, dysplastic, aplastic, or otherwise diseased odontoids.

Bilateral posterior displacement of C1 or C2 can occur in rheumatoid patients and can cause myelopathy. The instability resulting from erosion of the dens and the myelopathy is thought to be due to kinking of the spinal cord associated with its serpentine course into the foramen magnum. Shear loading of the spinal cord associated with motion in this region may be a factor in the myelopathy. Satisfactory treatment involves C0–C1–C2 arthrodesis following skeletal traction, although corrected alignment tends not to persist through the postoperative state.[92] MRI will probably be helpful in the further evaluation of this condition.

Unilateral Anterior Displacement

This type of abnormal rotatory displacement, between C1 and C2, is thought to be the most common. Either the left or the right articular mass moves anteriorly, and the axis of rotation is in the region of the articular facet that remains behind (Fig. 5-23). This usually occurs in association with various arthritic

FIGURE 5-22 This lateral radiograph shows a dysplastic odontoid. There is bilateral posterior translatory displacement of C1 on C2.

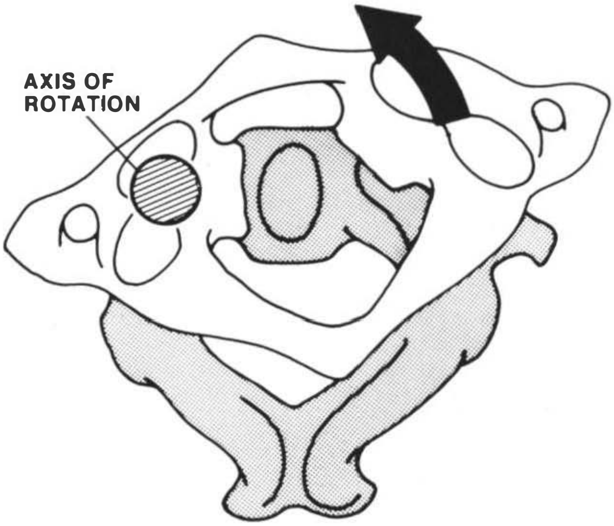
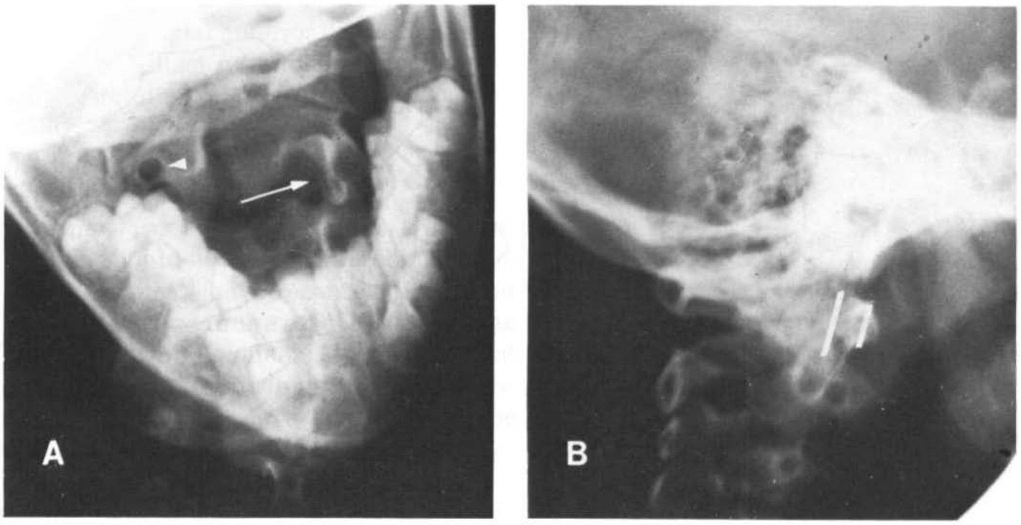


FIGURE 5-23 Unilateral anterior rotatory displacement of C1 in relation to C2. The presumed axis of rotation is at or near the relatively stable lateral mass articulation. In addition to damaged or abnormal odontoids and transverse ligament damage, this type of displacement, may have a unilateral articular capsule disruption on the anteriorly rotated side.

conditions or infections about the head and neck. Presumably, the pathology involves the transverse ligament as well as the articular capsule on the side of the subluxation.

The right unilateral anterior rotatory subluxation shown in Figure 5-24 would be expected to have a torticollis with the head "cocked like a robin, listening for a worm." There is lateral bending of the head and neck to the right and rotation of the head to the left. Rotating the head away from the direction that it faces, toward the subluxed side (in this case, rotation of the head to the right), is difficult, if not impossible. However, rotating it further in the direction that it faces, to the left, is not particularly difficult. The head is tilted toward the side of the subluxation or dislocation and rotated away from it. The muscles may or may not be in spasm. The syndrome is distinguishable from congenital torticollis by clinical history and by the absence of fibrosis of the sternocleidomastoid muscle. The anterior tubercle of the ring of C1 may be palpated on the posterior pharyngeal wall. In the case of a right unilateral anterior dislocation, as described here, the tubercle is displaced to the patient's left pharyngeal wall.

In the unilateral right anterior rotatory subluxation, there are findings on the open-mouth view that are compatible with normal rotation between C1 and C2. Specifically, there may be offsets or superimpositions of a lateral mass of C1 on C2. The key radiologic finding is seen on the true lateral view of the atlas (see Fig. 5-24B). Here, there is a large displacement between the dens and the anterior ring of the atlas. This is shown diagrammatically in Figure 5-24C. Unilateral offset is thought to be "normal."[37,72,113,141] Therefore, its presence alone is not diagnostic of abnormal displacement; there must also be an increased distance between the dens and the anterior ring of C1 in order to diagnose rotatory subluxation. Axial tomography or computerized axial tomography shows anterior displacement of C1. Cineradiography may show absence of movement between C1 and C2 on axial rotation.[51,54] Shapiro and colleagues have suggested that open-mouth views of the dens in neutral, with the head rotated



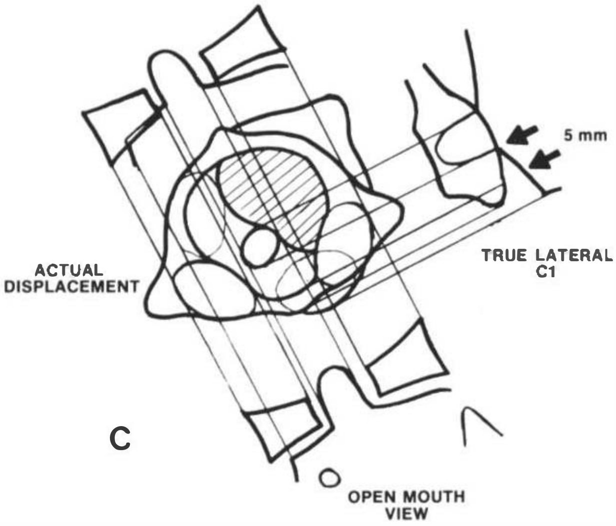


FIGURE 5-24 Clinical example of unilateral anterior displacement of C1 in relation to C2. The patient has a typical torticollis with lateral flexion of the neck to the right, and the head is rotated to the left and held in slight flexion. **(A)** The open-mouth radiograph shows deviation of the spinous process of C2 to the left *(arrow)*, $-\theta y$-axis, and apparent unilateral offset of the atlantal articular masses with respect to the foreshortened body of the C2. The foramen for the vertebral artery in C2 is readily visualized *(arrowhead)*, signifying marked rotation ($-\theta y$-axis) to the right. Because of the flexion of the head and the angulation of the central ray, the atlanto-axial articular surfaces appear to overlap. The same radiographic manifestations are seen in torticollis. Therefore, one cannot prove subluxation on the open-mouth view alone. **(B)** The key diagnostic findings in true rotatory subluxation are seen on the true lateral view of the atlas; the distance between the anterior margin of the dens and the posterior margin of the anterior arch of the atlas measures 5 mm. **(C)** The line drawing depicts the radiographic findings. The patient's thorax faces the bottom of the page. The axis is rotated to the right secondary to the right lateral flexion of the neck. The head (and therefore also the atlas) is rotated to the left maximally with respect to the axis and moderately with respect to the thorax. The central ray is centered anteroposterior and lateral to the atlas. The key finding again is the increased distance from the dens to the anterior arch of the atlas. *(Shapiro, R., Youngberg, A. S., and Rothman, S. L. G.: The differential diagnosis of traumatic lesions of the occipito-atlanto-axial* segment. *Radiol. Clin. North Am.,* 11:505, 1973.*)*

15° in either direction, allow the diagnosis to be made if the spine of the axis remains on the same side of the midline after the voluntary rotation.[141]

These lesions, in the absence of odontoid fracture and neurologic deficits, are thought to be clinically stable. However, if severe symptoms of pain persist and/or deformity is present, surgical therapy is indicated. We suggest a 2–3-day trial of correction with adequate traction in extension, followed by C1–C2 fusion.

Unilateral Posterior Displacement

This is probably the rarest type of rotatory injury.[54] It usually occurs with a deficient or a fractured odontoid. There may be neurologic symptoms, as with unilateral anterior displacement. One articular mass moves posteriorly, with the axis of rotation being in the region of the articulation on the opposite side (Fig. 5-25). Torticollis and pain are reported in the other types. In this entity, however, there is no abnormal separation between the dens and the anterior ring of C1. Lateral displacement of the dens may be seen on computerized axial tomography. Lack of C1–C2 motion may be apparent on cineradiography with axial rotation of the head.

The guidelines for treatment here are similar to those suggested for unilateral anterior displacement. We suggest that because this lesion is likely to be associated with a hypoplastic or fractured dens, it is more likely to be unstable.

Unilateral Combined Anterior and Posterior C1–C2 Subluxations and Dislocations

This situation occurs when there is abnormal rotatory displacement, with one lateral mass dislocating forward and the other backward. The axis of rotation is in the region of the dens (Fig. 5-26). If either mass rotates completely off the normal articulation, then the deformity may become fixed. Pain and torticollis are manifested. The entity may also be associated with neurologic deficit. Radiographic findings are the same as those of unilateral posterior displacement. This lesion need not be presumed to be unstable. If the dens, the transverse ligament, and the tectorial membrane are intact, the loss of functional integrity of the capsular ligaments may not render the FSU unstable. It can be difficult to recognize clinical instability in this type of subluxation. We suggest attempted reduction with traction. If this is not successful, and the symptoms are not tolerable, a C1–C2 fusion is recommended.

Atlanto-Axial Rotatory Fixation ("Atlas Spin-Out")

This lesion, described by Ono and colleagues in 1985[109] as atlanto-axial rotatory fixation (AARF), is essentially a y-axis rotatory subluxation of C1 in relation to both C0 and C2. In other words, the atlas has an abnormal axial (y-axis) rotation, leaving C0 and C2 in essentially normal relationship with each other.

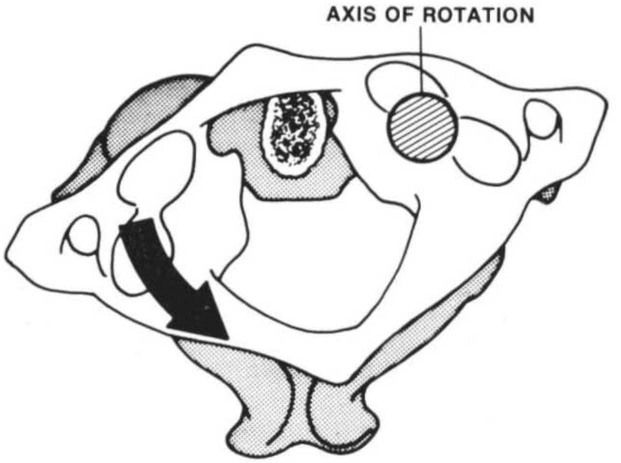


FIGURE 5-25 Unilateral posterior rotatory displacement of C1 in relation to C2. The presumed axis of rotation is at or near the relatively stable lateral mass articulation. In order for this type of displacement to occur, there must be some abnormality of the odontoid (either congenital or acquired).

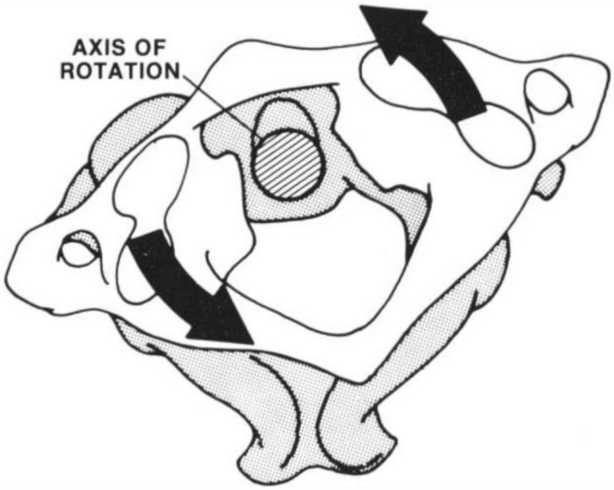


FIGURE 5-26 Unilateral combined anterior and posterior rotatory displacement of C1 in relation to C2. The presumed axis of rotation is at or near the odontoid. Both lateral mass articulations are disrupted. This type of displacement is expected to least compromise the space available for the cord, because the axis of rotation is closest to the cord in this situation.

The condition described by Ono occurs mainly in children. There is a y-axis rotatory displacement that usually subsides spontaneously or with treatment. However, in AARF, significant abnormal rotation of C1 remains in the presence of a realignment of C0 and C2.

This can be demonstrated on CT scans with cuts of (1) the foramen magnum and occipital condyles, (2) the ring of C1 and the dens, and, finally, (3) the lower body of C2. We have depicted it diagrammatically in Figure 5-27). This can be confirmed and/or suspected by anteroposterior and lateral radiographs of the C0–C1–C2 complex, in which the anteroposterior view will show an overriding lateral mass of C1 on C2, the "wink sign," and an "enlarged" anteriorly displaced projection of that part of the ring of C1 anterior to the dens (Figs. 4-26 and 5-24).

Computerized Axial Tomography of the Spine

This technique has provided great advances in the clinical evaluation of spine problems. It is especially useful in helping to elucidate some of the complex displacements of C1 and C2.

In Figure 5-28, a unilateral anterior rotatory subluxation of C1 is associated with a tumor involving the lateral masses and the facet articulation.

Discussion

The problems of the various subluxations and dislocations of the atlas need considerable work for documentation and clarification. These various entities have been presented as a complete theoretical listing of all possibilities. They have been summarized in Table 5-4. Werne and Fielding have presented their own groupings of these entities.[54, 160] In our opinion, the problem of definitive diagnosis and accurate analysis of the clinical stability has yet to be solved. We wish to reemphasize the limitations and speculative nature of information available on the subject.

Table 5-3 lists some measurements that have been proposed as criteria to indicate the upper limits of normal and/or instability. The indices have come from a variety of studies, including cadaver studies, x-rays of normals, and post-traumatic patients. Some are based on component ablation–type studies. It is suggested that the criteria be used in conjunction with at least three other important clinical considerations. These are: (1) the presence of

TABLE 5–4 Summary of C1–C2 Subluxations and Dislocations

Type	Causes and Displacements	Physical Findings
Bilateral anterior	Dysplastic dens, trauma, infection, +z translation	Neutral or "cock robin" position of head
Bilateral posterior (very rare)	Fractured, absent, or destroyed dens; −z translation	Patient may hold head in hands
Unilateral anterior (most common)	Arthritic conditions and infections; ±y-axis rotation, IAR at opposite joint	"Cock robin" position of head. Difficulty in rotating head away from direction in which it faces. No difficulty in moving further in that direction. Anterior tubercle of C1 may be shown to be displaced laterally by palpation of posterior pharynx.
Unilateral posterior (rare)	Usually associated with a deficient or fractured dens; ±y-axis rotation, IAR at opposite sides	"Cock robin" position of head
Unilateral combined anterior and posterior	Trauma; ±y-axis rotation, IAR at dens	"Cock robin" position of head

CT = computerized axial tomography; A-P = anteroposterior

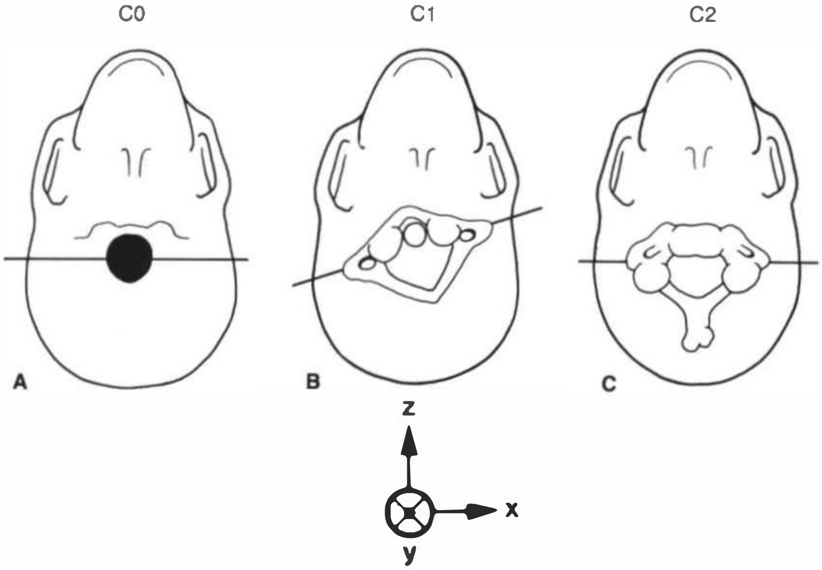


FIGURE 5-27 Diagrammatic representations of the atlanto-axial rotatory fixation. The diagrams represent cuts on an axial CT scan. **(A)** This cut is in the region of the foramen magnum and the occipital condyles and shows that a neutral position of the occiput and a line through the occipital condyles would be more or less perpendicular to the midsagittal plane. **(B)** An axial CT cut in the region of the atlas. A line drawn through the foramina for the vertebral arteries shows a rotation (−y-axis) of about 30°. This is a fixed rotation in the subacute stage of the condition. The dislocation is of the lateral masses of C1 and C2. **(C)** A fixed rotatory displacement of C1 in relation to C2, which is in neutral position. In **A** there is actually a compensatory reverse rotatory (+y-axis) displacement of C0 in relation to C1. In order to keep the head from being grossly rotated, there is approximately a 15° +y-axis compensatory rotation between C0 and C1. This compensates for the 15° −y-axis fixed rotatory subluxation of C1 in relation to C2.

Radiologic Studies	Clinical Stability	Treatment
Lateral of C1, CT scan: anterior displacement of C1 on C2	Anterior displacement of 3 mm, neurologic deficit—clinically unstable	Fusion or trial of conservative therapy
Lateral of C1, CT scan: posterior displacement of C1 on C2	Clinically unstable	Fuse C1–C2
Lateral of C1, CT scan: anterior displacement of C1 on C2. A-P open-mouth laminagrams C1–C2: lateral masses in different planes. Cineradiography or several radiographs of axial rotation: no motion of C1 or C2	With no neurologic deficit, probably stable	Trial of reduction and conservative treatment. If symptoms require, fuse C1–C2.
Lateral of C1, CT scan: no anterior displacement of C1 on C2. A-P open-mouth laminagrams, C1–C2: lateral masses in different positions. Cineradiography or serial radiographs of axial rotation: no motion of C1 or C2	With no neurologic deficit, probably stable	Attempt reduction and, if symptoms require, fuse C1–C2.
Same as unilateral posterior	With no neurologic deficit, may be clinically stable	Trial of reduction and conservative treatment; if not satisfactory, fuse C1–C2.

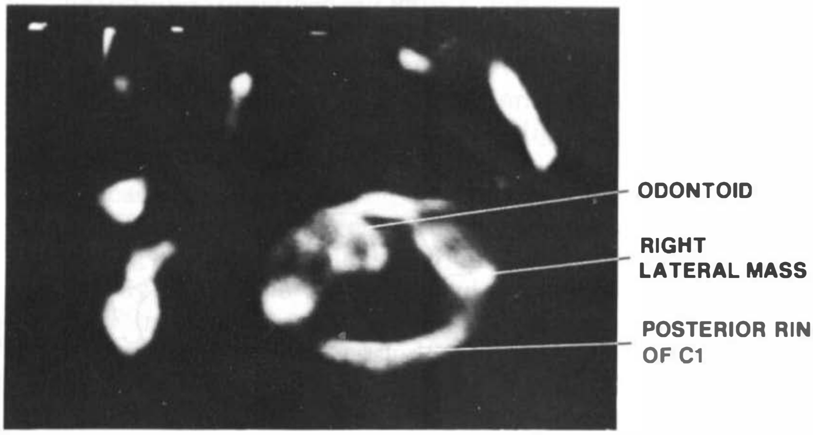


FIGURE 5-28 Computerized axial tomogram of the C1, C2 articulation. There is a unilateral anterior rotatory displacement of C1 in relation to C2 of the type shown diagrammatically in Figure 5-23.

neurologic signs or symptoms, (2) the anticipated loads based on the particular patient's lifestyle, and (3) other clinical considerations that are unique for the individual patient. If a patient has one or more of these abnormal measurements and evidence of neurologic involvement, then he or she should be considered clinically unstable and treated accordingly.

The guidelines so far discussed cannot be indiscriminately applied to rheumatoid patients. However, the guidelines offer a reference point as to when displacements may be approaching a range in which the patient is neurologically at risk. Pellicci and co-workers, in a review of 163 patients with both upper and subaxial cervical spine problems, emphasized that neurologic problems or "impending" neurologic problems are more important than radiographic changes in determining which instability should be surgically treated.[116]

RECOMMENDED EVALUATION SYSTEM

Little can be said about the clinical stability of C1–C2 subluxations and dislocations. We know that the transverse ligament, and one or more articular capsules, must be disrupted to allow abnormal rotatory displacement. If there is abnormal sagittal plane (+z-axis) translation by radiographic measurement of greater than 3 mm in an adult or 4 mm in a child, or if there are neurologic signs or symptoms suggesting irritation to the spinal medulla, the situation should be considered unstable.

A summary of the most cogent information for the evaluation of these conditions is provided in Table 5-4.

RECOMMENDED MANAGEMENT

We recommend treatment of these unstable injuries with traction in slight extension for 2–3 weeks, or until reduction is achieved. This is followed by fixation in a cervical orthosis of maximum control for an additional 10 weeks. If at that time there is no neurologic deficit, no torticollis, and no pain, the patient may progress to a cervical orthosis of intermediate control (four-poster brace with a thoracic support). If this is not the case, or should any of the above symptoms recur, a C1–C2 fusion is recommended.

PART 2:
THE MIDDLE AND LOWER CERVICAL SPINE (C2–T1)

This region of the spine that has received considerable attention with regard to the problem of clinical instability. At the cervical spine, neurologic deficit is most frequently associated with trauma.[130] This section reviews the past and current biomechanical and clinical factors that relate to the problem of clinical stability in this region.

ANATOMIC CONSIDERATIONS

What anatomic structures are necessary to maintain clinical stability in the lower cervical spine? A schematic representation of the anatomy of the lower cervical spine is presented in Figure 5-29.

According to Bailey, "the most significant of the anatomic structures providing stability to the cervical spine are the musculature and the firm bond between the bodies formed by the intervertebral disks."[7] He has emphasized the importance of the annulus fibrosus in other writings.[6] The role of the musculature is of considerable importance, but to our knowledge, its significance in clinical stability has not been determined.

Although Holdsworth emphasized the importance of the supraspinous and interspinous ligaments as well as the ligamentum nuchae, other investigators considered them to be much less significant.[73] Halliday and colleagues carried out anatomic dissections and observed that the interspinous ligaments are sometimes completely absent between one or more segments and that the ligamentum nuchae is quite delicate.[65] The latter observation was supported by the work of Johnson and colleagues, who carried out detailed anatomic studies of the ligaments of the lower cervical spine in 15 fresh specimens.[79] The more recent work of Fielding seems to offer considerable contradictory evidence, at least with regard to the anatomic structure of the ligamentum nuchae (see Fig. 5-14, p. 291).[52]

BIOMECHANICAL FACTORS

Experiments have been carried out on cervical spine FSUs in high-humidity chambers with the use of physiologic loads to simulate flexion and exten-

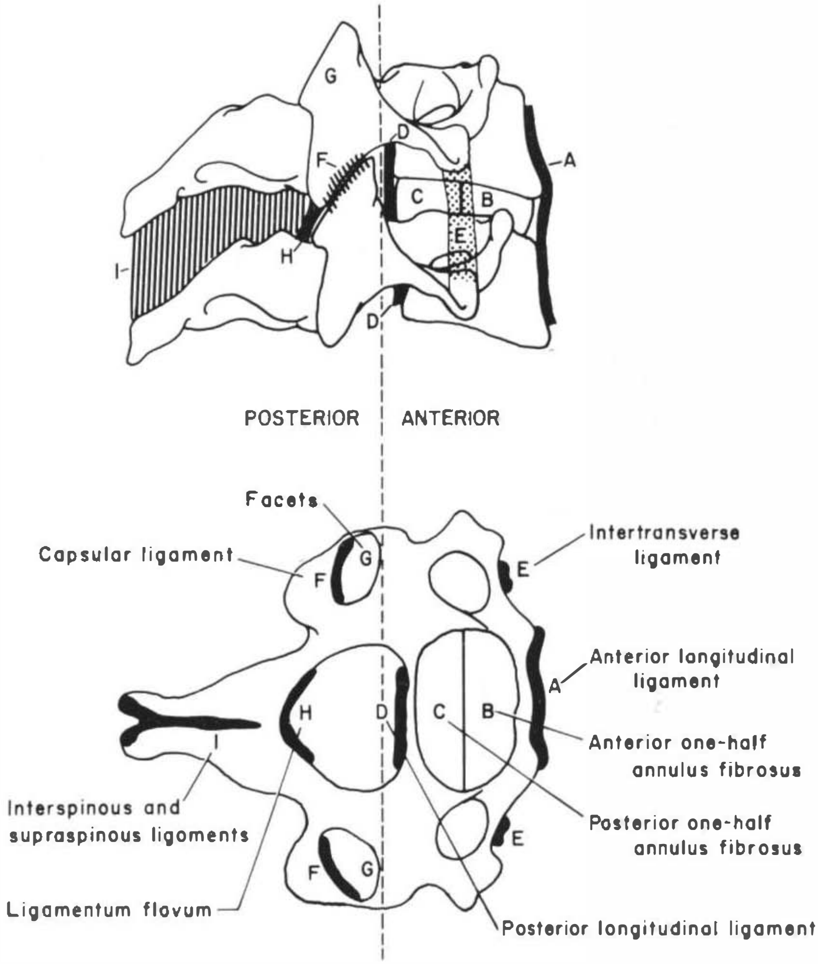


FIGURE 5-29 Schematic illustration of the ligamentous structures that participate in the stabilization of the middle and lower cervical spine. The components are divided into *anterior* and *posterior* elements. Anatomic components posterior to the posterior longitudinal ligament are defined as the posterior elements. The posterior longitudinal ligament and all the anatomic components anterior to it are defined as the *anterior* elements. In the experiments on clinical stability, ligaments were cut in the alphabetical order indicated in the diagram from anterior to posterior and in reverse alphabetical order from posterior to anterior.

sion.[111,162] The ligaments were cut in sequence from posterior to anterior in some FSUs and from anterior to posterior in others. The *failure point* was defined as the point at which the upper vertebra suddenly rotated 90° or was displaced across the experimental table. The *anterior elements* were defined as the posterior longitudinal ligament and all structures anterior to it. The *posterior elements* were defined as all structures behind the posterior longitudinal ligament (Fig. 5-29). At the conclusion of these studies, we suggested that if an FSU has all its anterior elements plus one additional structure, or all its posterior elements plus one additional structure, it will probably remain stable under physiologic loads. In order to provide for some clinical margin of safety, we suggest that any FSU in which all the anterior elements or all the posterior elements are either destroyed or are unable to function should be considered potentially unstable. Therefore, these studies show that the important anatomic structures for maintaining clinical stability are either all the anterior elements plus one posterior or all the posterior elements plus one anterior.

CLINICAL CONSIDERATIONS

There are a number of important clinical studies that have considerable bearing on the analysis of clinical stability. Several of these are discussed below.

Structural Damage and Neurologic Deficit

Is there a correlation between recognizable structural damage to the spine and neurologic deficit?

Barnes notes that "one of the most puzzling features of injuries of the cervical spine is the lack of correlation between the degree of vertebral displacement and the severity of the spinal cord lesion. There are cases with no radiographic evidence of bone injury in which the cord is irretrievably damaged; others, with gross dislocation, may have no paraplegia."[8]

Beatson was careful in comparing the amounts of displacement in the sagittal plane that occur with unilateral and bilateral facet dislocation. Even though there is more displacement with bilateral facet dislocation, he did not document any consistent difference in the neurologic deficits.[10] Unilateral facet dislocations usually consist only of root symptoms, while the bilateral dislocations are associated with serious spinal medullary lesions.[18]

An epidemiologic study of the incidence of cord injury with trauma to vertebrae in all regions of the spine resulted in some interesting data. When vertebral body fracture alone was present, the associated incidence of neurologic deficit was 3%. However, if there was malalignment of 2 mm or more, or vertebral body damage plus posterior element damage, then the incidence of associated neurologic deficit went up to 61%.[130] There are some other considerations that may relate to this question and account for some of the confusion. The experimental work of Gosch and colleagues on monkeys showed that it is possible for significant spinal cord damage to result from trauma without any fractures or ligamentous ruptures.[61] There is also the central cervical cord syndrome, described by Schneider and colleagues, in which there is paralysis of the upper limbs with function in the lower limbs and no radiologic evidence of fracture or fracture dislocation.[139] It was well documented by Marar that fracture dislocations in the lower cervical region can go unrecognized (Fig. 4-48).[94] He reported on autopsy studies of four patients with extension injuries and transverse fractures of the vertebral body. The fractures reduced spontaneously after compressing the medulla between the upper portion of the fractured vertebra and the lamina of the subjacent vertebra. The fracture was not visible on routine radiographic examination. The medullary damage in these four patients may have occurred in the presence of intact spinal elements.

When there is spontaneous reduction of such a dislocation and if there is no residual deformation at the time the radiograph is taken, the dislocation may go unrecognized. Sometimes the presence of retropharyngeal or retrotracheal soft-tissue shadows, increased space, or a prevertebral fat stripe may alert the astute clinician to the presence of an otherwise occult injury (see Table 5-5).

There is usually a discrepancy between the damaging displacement, which occurs at the time of impact, and the *residual deformation*, which is what is actually observed on the radiograph. The presence of residual deformation shows that the FSU as a whole was deformed into its *plastic range*. In addition, it is possible that the complex bony and ligamentous structure of the spine may conceivably deform enough to cause medullary damage but remain entirely within its *elastic range*. In this case, the FSU would recoil to its normal position and condition.

Therefore, although there are exceptions, there is

TABLE 5–5 Radiographic Signs of Cervical Spine Trauma

Soft tissues
Retropharyngeal space > 7 mm adult or child
Retrotracheal space > 14 mm adult; > 22 mm child
Displaced prevertebral fat stripe
Tracheal deviation and laryngeal dislocation

Vertebral alignment
Loss of lordosis
Acute kyphotic angulation
Torticollis
Widened interspinous space
Axial rotation of vertebra

Abnormal joints
Atlanto-dental internal > 4 mm* adults; > 5 mm children
Narrowed or widened disc space
Wide apophyseal joints

*Opinion varies as to whether this number should be 3 or 4. We have chosen 4 as a clinical "judgement call."

(Modified from Clark, W. M., Gehweiler, J. A., and Laib, R.: Twelve significant signs of cervical spine trauma. Skeletal Radiol., 3:201, 1979.)

some correlation between neurologic deficit and the radiographic appearance of the spine following trauma. Bursting fractures of the vertebral body, especially with horizontal displacement and posterior element fractures, are highly correlated with spinal cord injury. The contrast between unilateral and bilateral facet dislocations has shown that with the latter, more extensive injury, there is a greater neurologic deficit. Damage of the nerve roots, which may be independent of damage to the cord, has not been carefully distinguished and studied.

Neurologic Deficit and Clinical Instability

We believe that if the trauma is severe enough to cause initial neurologic damage, the support structures probably have been altered sufficiently to allow subsequent neurologic damage, and thus the situation is clinically unstable. Some exceptions should be noted. This important consideration is discussed in more detail on page 317.

The following clinical problem exemplifies several crucial points related to clinical instability.

Case Report

B. A. is a 19-year-old boy who was struck on the head from behind during a game of rugby. He developed an incomplete tetraparesis immediately following the injury. The radiographs on admission to the hospital showed an anterior subluxation of C4 on C5 (Fig. 5-30A). This was reduced with skull traction, and the tetraparesis resolved quickly. Note that the extension film in Figure 5-30B shows almost complete reduction, except for slight separation of the laminae and spinous processes between C4 and C5. The patient was treated with a Minerva jacket for 3 months. Radiographs immediately after removal of the jacket, which included flexion/extension views, were reported to be normal. However, some 4 months later, an obvious resubluxation of an extreme degree was noted (Fig. 5-30C). Fortunately, this was reduced after a week of recumbency on a plaster bed. Subsequently, a successful posterior fusion was carried out (Fig. 5-30D).*

Discussion

This case very nicely shows that even transient neurologic symptoms may suggest plastic deformation or complete failure of ligamentous structures. In this case, most likely there was extreme plastic deformation or rupture of all posterior ligaments, the posterior longitudinal ligament, and all or most of the annulus fibrosus, with little residual displacement.

The stretch test (see p. 318) may have been helpful in demonstrating the loss of continuity of the damaged ligaments in this case. Also note radiographically the decrease in height or abnormal narrowing of disc space, a sign indicating probable disruption of the annulus fibrosus.[6] This is especially important when seen in a young patient whose other disc spaces are normal following trauma. Webb and colleagues[159] and other investigators[89,97] have recognized a pattern for this type of situation. The findings are shown below. A review of Figure 5-30A shows the complete tetrad. This is a recognizable syndrome that has a high probability of being clinically unstable. The findings correlate very well with Penning's description of kyphotic angulation.[118] The clinician should also look for disc narrowing at the interspace under analysis. The initial treatment in the previous case report was perfectly adequate and would have been successful had there been spontaneous fusion or satisfactory healing of the ligamentous structures.

See also the discussion of "full nelson" in Chapter 4. Note the similarities and differences. This is a family of injuries.

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* Personal communication with T. McSweeney and W. Park, November 1975.

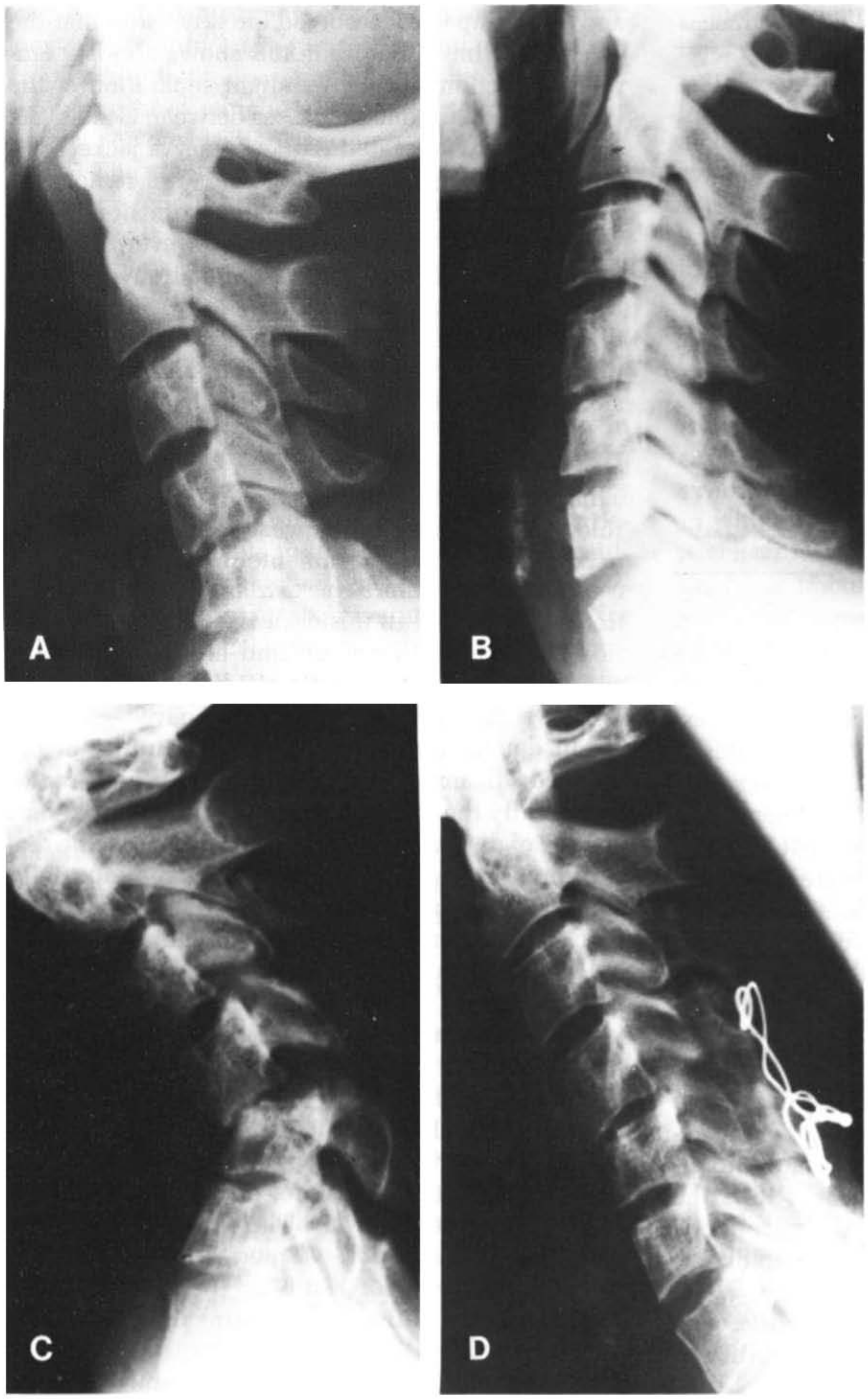

FIGURE 5-30 Radiographs of a flexion injury in a 19-year-old with transient spinal cord symptoms. (See p. 305 for a case report.) Knowing the nature of the injury, could the resubluxation shown in **C** have been predicted from radiographs **A** (taken on admission) and **B** (taken after 3 months of treatment)? This clinical picture and radiographic presentation should be learned so that the diagnosis can be made or suspected when the patient is first seen. *(***A***,* **B***, and* **D** *courtesy of T. McSweeney and W. Park; C from Webb, J. K., Broughton, R. B. K., McSweeney, T., and Park, W. M.: Hidden flexion injury of the cervical spine. J. Bone Joint Surg., 58B:322, 1976.)*

Clinical Stability of Unilateral Facet Dislocations

The clinical and experimental studies of Beatson are most instructive. He produced unilateral and bilateral facet dislocations and observed the sagittal plane displacement on lateral radiographs. These findings were correlated with anatomic studies of the associated ligamentous damage. On the lateral radiograph, a displacement of one-half or less of the anteroposterior diameter of the vertebral body indicates a unilateral facet dislocation. If the displace-

RECOGNIZING OCCULT INSTABILITY

History of a flexion injury
Widening of interspinous space
Subluxation of facet joint
Compression fracture of subjacent vertebra
Loss of normal cervical lordosis

(Webb, J. K., Broughton, R. B. K., McSweeney, T., and Park, W. M.: Hidden flexion injury of the cervical spine. J. Bone Joint Surg., 58B:322, 1976.)

ment is greater than one-half of the diameter, it should be diagnosed as a bilateral facet dislocation.[10] Bedbrook suggested that with a displacement of one-half of the anteroposterior diameter of the vertebral body, the spinal canal is encroached upon by one-third of its anteroposterior diameter.[11]

Beatson also observed that with unilateral facet dislocation there is rupture of the interspinous ligament and capsule of the involved facet joint. There is only minimal damage to the annulus and the posterior longitudinal ligament on the dislocated side. He noted that it would be difficult to reduce this dislocation with straight longitudinal traction because of the resistance of the intact disc and capsule on the undislocated side.[10] Unfortunately, there is no mention of the status of the yellow ligament in this study; information about the fate of this important structure in the various facet dislocations would be valuable.

Cheshire reported on three patients with unilateral facet dislocations and spinal cord involvement. All were manually reduced and treated for 6 weeks in traction. Redislocation occurred while they were wearing cervical collars. These patients were then treated by surgical fusion to ensure stability.[30] Braakman and Vinken reported on 37 patients with unilateral facet dislocations. Seven of these patients had nerve root involvement and 34 had medullary involvement. The investigators recommended conservative treatment and indicated a low incidence of late instability.[18]

Rorabeck and co-workers[134a] reviewed 26 patients with unilateral facet dislocations and made the following conclusions and recommendations:

1. Attempt reduction with traction.
2. If successful, treat with a halo thoracic orthosis.
3. If not reduced with traction, do an open reduction and fusion.

In our opinion, there should be enough intact ligaments that a closed reduction not associated with a fracture can be treated with a hard cervical collar with shoulder and/or thoracic support.

In reviewing the available clinical and experimental evidence, we make the assumption that only some unilateral facet dislocations are unstable. Generally, when there is neural involvement, especially with spinal medullary damage, enough displacement may have taken place to cause significant ligamentous damage. The observations of Beatson showed that, in at least some instances of unilateral facet dislocation, damage may occur to the anterior as well as the posterior elements.[10] Certainly, when there is an associated facet fracture there is less stability. In general, unilateral facet dislocations associated with neurologic damage or facet fracture must be considered clinically unstable. Other unilateral facet dislocations without neurologic deficit, especially when they are difficult to reduce (which implies a largely intact annulus and yellow ligament), may be considered clinically stable.

Clinical Stability of Bilateral Facet Dislocations

In several reported clinical series, the bilateral facet dislocation is considered to be an unstable injury.[11,22,30] Beatson's experimental observations showed that in order to create a bilateral facet dislocation, it was necessary to rupture the interspinous ligaments, the capsules of both facet joints, the posterior longitudinal ligament, and the annulus fibrosus.[10] This would result in a clinically unstable situation. These injuries are prone to undergo abnormal translation along the z-axis. Experimental studies have shown that when flexion is simulated after removal (fracture) of the articular facets, there is a good deal more anterior translation (Fig. 5-31).[111] This factor, in connection with the relative vulnerability of the cord to damage by displacement in the horizontal plane, makes such an injury extremely unstable.

The facets are important in another aspect of clinical stability. This has been well characterized by the work of Raynor, Pugh, and Shapiro.[126] The investigators studied *in vitro* 14 human FSUs loaded in shear before and after partial facetectomies. The specimens were tested and loaded to failure. Five of the FSUs were intact, five had 50% facetectomies (on each side), and four had 70% facetectomies bilat-

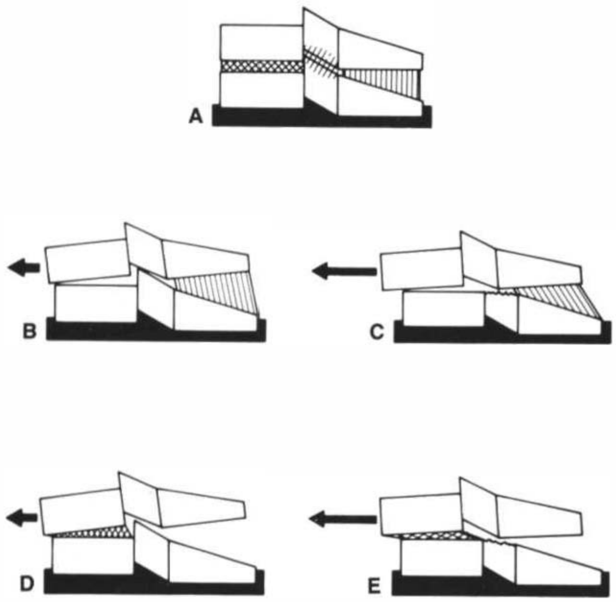

FIGURE 5-31 This diagram is designed to show the role that the articular facets play in the anterior translation of a given functional spinal unit. **(A)** The normal functional spinal unit. **(B)** With all anterior elements removed, **(C)** there is more anterior translation after destruction of the facet articulation. **(D)** With all the posterior elements removed, **(E)** there is also more anterior translation after the facets are destroyed. The practical significance of this is that with bilateral facet fracture or dislocation, the tendency for anterior translatory displacement is much greater.

erally. The 70% facetectomy specimens were significantly weaker ($p<0.05$) than those with 50% facetectomies. The authors suggested that an adequate portion of the root could be exposed (5 mm) with a 50% facetectomy, but if >50% was involved, stabilization should be considered. This, we believe, presumes that a laminectomy is also present.

Laminectomy and Clinical Instability

This is an extremely controversial topic. We believe that the procedure is done too frequently, with improper indications, and with an astounding lack of appreciation of its effect on the mechanical function of the spine. In children there is a tendency for kyphosis, anterior subluxation, and "goose-neck deformity" to develop as a result of laminectomies in the cervical spine.[27] A ten-year follow-up of 40 children who had laminectomies showed that 40% of the patients required stabilization for unstable injuries and progressive deformity.[96] Bell and associates[13a] reviewed 86 of 122 children who were operated on for cervical laminectomy. The mean age of the subjects at the time of surgery was 5.7 years (and the average number of years postsurgery at evaluation was 5 years). There was kyphotic deformity in 37% of the patients, and 13% developed a *hyperlordotic* or swan-neck deformity. Surgical fusion was recommended for the patients who developed this deformity. We suggest that children with laminectomy in the cervical region be x-rayed and measured at least annually until growth is completed. The growing spine is especially prone to deformity as the epiphysis responds to asymmetrical forces and causes wedging of the vertebral bodies.

Jenkins reported several cases of multiple laminectomies performed in the lower cervical spine without subsequent deformity or clinical instability.[77] Although there may not always be complications, we must emphasize that multiple laminectomies in the adult cervical spine may lead to clinical instability, with serious neurologic consequences. Clearly, the condition of the remaining structures following laminectomy is a significant factor in the outcome. The structure is less likely to be unstable if the facet articulations and their capsules are intact and the anterior elements are normal. If any of these remaining units are destroyed or nonfunctional, clinical instability is very likely to occur. The patient whose radiographs are shown in Figure 5-32 had laminectomies at C2 and C3, with the facet joints preserved and no anterior element surgery or injury. However, he developed a major neurologic deficit and severe kyphosis.

In Figure 5-33, some important points about spine structure and clinical stability are demonstrated. The patient is a 38-year-old female who had an ependymoma removed from the lower cervical spine. Figure 5-33A shows the spine after total laminectomy of C4, total laminectomy and partial facetectomy of C5, and removal of all the posterior elements of C6. It appears as though the annulus fibrosus between C5 and C6 is degenerated. It is presumably fibrosed, partially ossified, and actually better able to resist translation than a normal disc. At C6, relatively larger loads are applied, and the other interspaces are relatively more stable, because it was the only level at which *all* the posterior elements were removed. Predictably, the abnormal anterior translation and clinical instability occurred at the C6–C7 interspace, as shown in Figure 5-33A. Because of the spatial orientation and inclination of the interspace at this level, the abnormal translation was

FIGURE 5-32 This radiograph demonstrates clearly the role that laminectomy can play in contributing to the clinical instability of the spine. *(Courtesy of Harry Gossling, M.D.)*

anterior. (This is in contrast to the posterior translation of L2 or L3 seen in the somewhat analogous situation in Fig. 5-67). In Figure 5-33B, a bone graft in the C6–C7 interspace is shown. This was not a good choice of surgical construct because it required removal of the anterior longitudinal ligament and the annulus fibrosus in order to insert the graft. The immediate postoperative stability was dependent on the posterior longitudinal ligament alone. This may be expected to result in a grossly unstable situation. The bone graft was partially resorbed, and the predictable clinical instability is evidenced by the gross abnormal translation and rotation shown in Figure 5-33C, D. It was possible to reduce the severe subluxation and to achieve stability with the use of an anterior iliac trough graft and halo body cast fixation.

TREATMENT CONSIDERATIONS

Effects of Reduction on Prognosis

Does reduction of a fracture dislocation in the middle and lower cervical spine favorably affect the prognosis? Rogers stated that progression of spinal cord symptoms was more likely to occur in unreduced injuries.[134] Burke and Berryman stated that a large number of patients showed improvement after manual manipulation.[22] However, these statements were not documented.

Dall, in a report of over 200 cases, emphasized that the major determinants of overall prognosis were the nature and magnitude of the *initial injury* to the spinal cord. He offered follow-up data on a large number of patients that supported his assertions that the type of bone injury, the lack of reduction, and even *redislocation* had no influence on spinal cord recovery.[35] Dall did not separate his findings into categories of cord and root symptoms. Thus, the evidence does not rule out the possibility that root recovery is related to achieving and maintaining reduction of the traumatized spine. Braakman and Vinken, reporting on 37 patients, were affirmative about the value of reduction in providing nerve root recovery in patients with unilateral facet dislocations.[18] These studies corroborated the generally expressed maxim of "rigorous therapy to preserve the root."

The Role of Manipulations

What is the value of manipulation in dislocations and fracture dislocations of the middle and lower cervical spine? In the United States, there has apparently been a tendency to treat manipulation almost as though it were taboo, whereas in the United Kingdom it seems to be commonplace. Beatson has shown that axial traction is probably not the most efficient method of reduction of bilateral or unilateral facet dislocations. He suggested techniques for the reduction of both injuries.[10] Other investigators have described techniques and management of manipulations for the treatment of spinal trauma.[17,22,30]

When reduction of unilateral facet dislocations without medullary damage is not achieved with 50–60 lb of axial traction, manual manipulation seems to be a desirable approach. Detailed reviews of the various techniques are available in the literature. The studies of Taylor and Walton constitute

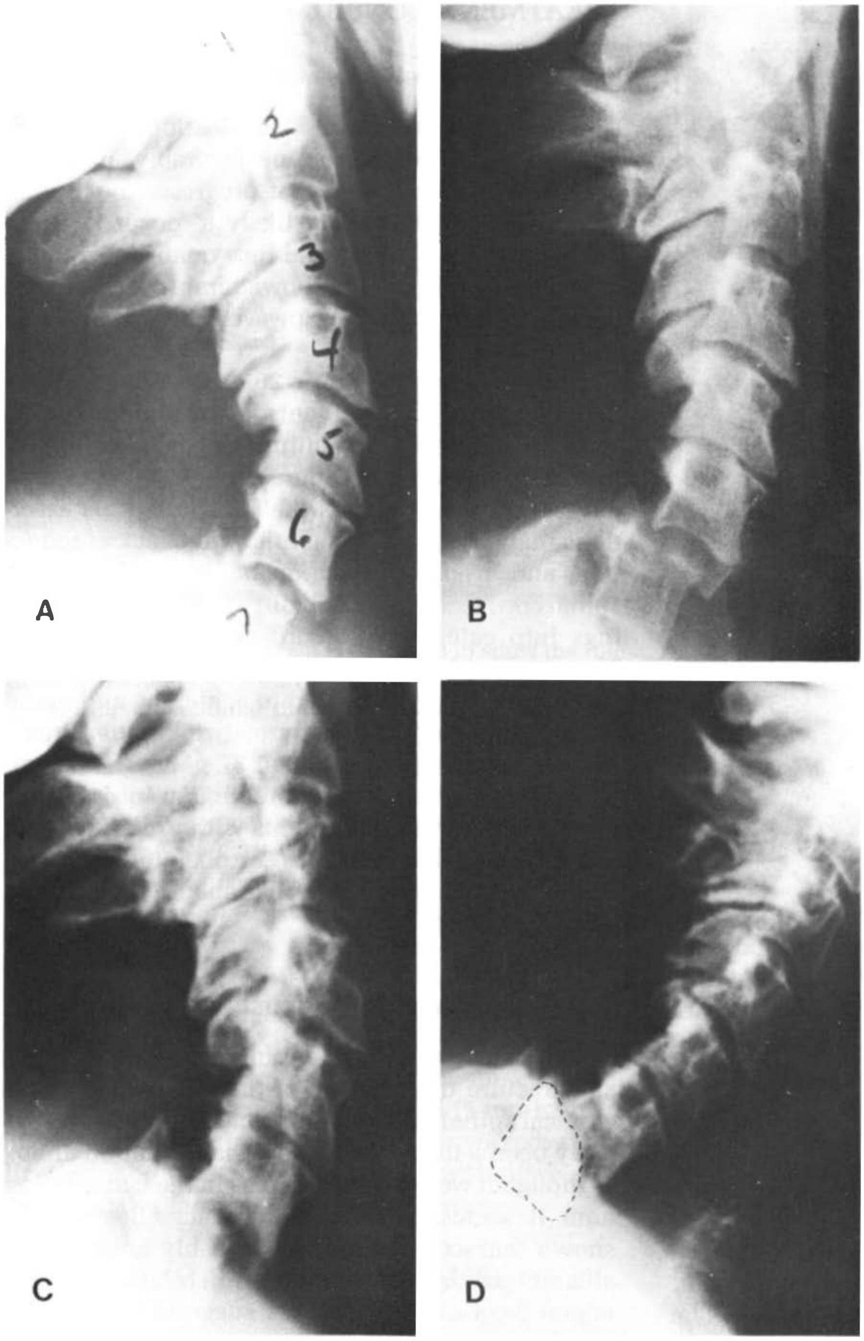

FIGURE 5-33 These radiographs demonstrate several important points about the problem of clinical stability. Laminectomy and facetectomy that include all the posterior elements render the spine clinically unstable; the subsequent disruption of the anterior elements in order to insert a bone graft makes the situation even more grossly unstable. The two components of instability, translatory, shown in **B** and **C**, and rotatory, shown in **D**, are graphically demonstrated by this case. *(Courtesy of W. O. Southwick, M.D.)*

some of the earlier descriptions.[151,158] There are basically three types of loads applied to the head in manipulations for reduction. There is axial loading, as with ordinary traction; axial rotation, in which a torque is applied in the horizontal plane; and lateral bending, in which a torque is applied in the frontal plane. Axial rotation and lateral bending are applied in sequence and in either order, depending on individual preference.

We suggest that there is probably a place for the manipulative reduction of the unilateral facet dislocation in which there is no neurologic involvement. The pathoanatomic aspects of this entity have been well studied, and the recommendations of previous investigators are available.[10,18,22,30]

The Role of Decompression

What about decompression? A detailed discussion of this controversy could fill as many pages as the topic of clinical instability. We are suggesting some guidelines and are aware of their limitations and controversial nature. The following are currently accepted indications for decompression: radiographic evidence of bone or a foreign material in the medullary canal, associated with medullary symptoms; evidence on CT, myelogram, or MRI of a discrete extradural block; and clinical judgment (e.g., in the presence of an incomplete progressive neurologic lesion, the surgeon believes that decompression would be beneficial). Generally, decisions about decompression should be made and carried out as soon as practical clinical conditions permit.

There are some cogent questions concerning decompression. Given the decision to decompress, should it be done from the front or from the back? This is determined primarily by the location of the pathology and secondarily by the area in which there is the most structural damage, if there is no imaging documentation. Decompression should be carried out whenever possible on the side of the defect. If that is impossible to determine, then most probably it is best to decompress where the major damage to the bony and ligamentous structure has occurred. Anterior decompressions for trauma are generally best achieved by total or partial excision of the vertebral body. The spine should be reconstructed with an appropriate bone graft (see Chapter 8). This is sufficient if the posterior elements are intact. If the posterior elements are not intact, they, too, should be fused and wired, preferably before anterior decompression. If the patient's condition and the surgeon's experience permit, the two procedures may be done in sequence under the same anesthesia.

With posterior decompression of one or two segments, leaving the facet joints intact, and with normal anterior structures, immediate fusion is not necessary. The patient should be followed carefully and observed for evidence of progressive instability with anterior displacement or for the development of posterior collapse (goose-neck) secondary to the loss of support structures. If decompression involves more than two levels, if there is any disruption of the facet joint integrity, or if the anterior elements are not intact, then bilateral posterolateral facet joint fusion and wiring should be carried out at the time of decompression.

A Biomechanical Hypothesis for Postlaminectomy Instability

There are several theoretical factors that may contribute to the development of postlaminectomy instability and kyphosis.

This procedure necessarily involves considerable alteration of anatomic structures that contribute to stability. There may be some denervation of the posterior erector spinae muscles, causing weakness, atrophy, and relative "imbalance" due to alteration in strength relative to the anterior cervical muscles. The removal of spinous processes and muscle attachments may result in a relative *anterior* displacement of the muscle attachments to scar tissue and facet structures. This anterior displacement of muscle attachment closer to the instantaneous axes of rotation (IARs) reduces the moment arm and thus further compromises the efficiency of the muscles in counterbalancing the flexion moment due to the weight of the head. The minimal force required to balance the head and neck in the neutral prone position is estimated with a biomechanical model to be 140 N (31.7 lbf). There is also disruption of the passive tensile forces normally exerted by the nuchal ligament, the supra- and interspinous ligaments, and the yellow ligaments, all of which are removed at the site of the laminectomy. These various posterior perturbations provide a propensity for progressive protrusion of the head and concomitant kyphosis. This concept is represented diagrammatically in Figure 5-34. The idea for the development of this concept was stimulated by the work of Nolan and Sherk,[107] who addressed the effects of stripping the extensor muscles of the neck on the biomechanics of the cervical spine.

Patients with progressive kyphotic deformity following extensive laminectomy may have a feeling of "the head being too heavy for the neck," as well as tension, strain, and pain in the posterior neck muscles. Presumably, some of the pain is due to increased loads on the disc and facet joints, as suggested by the free-body diagram model shown in Figure 5-3A. The facet joint forces increase because some of the posterior elements are removed by the

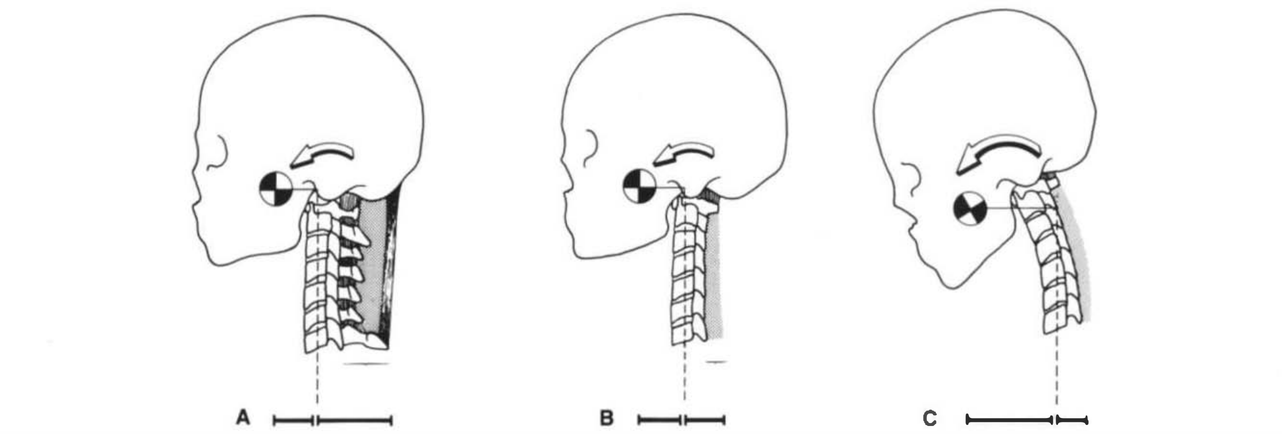

FIGURE 5-34 *Clinical biomechanical hypothesis* for the propensity to develop kyphosis following extensive laminectomy involving C2 through C7. The position of the center of gravity is assumed. **(A)** The posteriorly displaced ligamentum nuchae, spinous processes, and attached muscle mass (the dotted line represents the load-bearing axis). The ligaments and muscle masses provide a greater moment arm to resist forward bending and kyphosis. **(B)** Following laminectomy, the posterior moment arm is lessened by removal of the posterior elements, particularly those of C2 and C7. The posteriorly located ligamentum nuchae (a delicate structure, but well positioned to resist flexion) is removed, and the posterior cervical muscles have been denervated and weakened to some degree. There is less mechanical advantage (shorter moment arm) and less strength in the muscles. (C) The gravitational pull results in some flexion and kyphosis. As this occurs with the displacement of the center of gravity forward, the greater moment arm (mechanical advantage is anterior) is developed, as shown by the relatively larger anterior bending moment. This moment increases as the kyphosis increases. The forces generated by the anterior cervical muscles are presumed constant.

laminectomy. This is in contrast to improper lifting when the center of gravity is being shifted forward to cause the increased load at the fulcrum (see Fig. 6-5A). In this postlaminectomy hypothesis, the increased load is a result of the posterior lever arm being shortened because of removal of the more posterior structures to which they were initially attached. Despite this reasonable theoretical analysis, postlaminectomy kyphosis is uncommon in adults when no additional structures are compromised. Further assurance of postlaminectomy stability is gained if the posterior elements of C2 and C7 can be preserved at the time of laminectomy.

Traditional Indications for Surgical Arthrodesis

It is necessary to determine, as precisely as possible, just what the indications for fusion are. There are few works that attempt to present the indications for fusion following fractures and fracture dislocations. Beatson suggested that bilateral facet dislocations were an indication for fusion.[10] Braakman and Vinken stated that a unilateral facet dislocation with a neurologic deficit should be fused.[18] We believe that only a unilateral facet dislocation requiring open reduction, with or without neurologic deficit, should be fused.

This rationale is based on three considerations. First, the previously described work of Beatson shows that there can be significant disruption of the anterior elements associated with the injury.[10] Second, there is the possible complication of delayed arthritis and pain associated with the disrupted joint. Third, since the surgical exposure is performed in order to achieve reduction, it is possible to fuse and ensure against future pain and instability with little additional surgical risk.

Dall described satisfactory results in his large series of patients with fracture dislocations treated nonsurgically (3 months of skeletal traction and 3 months of treatment with a cervical collar). He considered progressive bone deformity with late pain or neurologic deficit to be an indication for fusion. In a series of 75 patients with fracture dislocations, only 3 were thought to need fusion. The work argues respectably for a nonsurgical approach.[36]

White and colleagues propose that if, upon examination of flexion, extension, or resting lateral radiographs, there is a relative translation (anteroposterior displacement in the sagittal plane) greater than 3.5 *mm*, a relative rotation (more than either adjacent vertebra) greater than 11°, or complete destruction and loss of function of all the anterior or posterior elements, then clinical instability after cervical spine trauma is present.[162] Although these criteria are precise and applicable, they, too, have limitations. They are based on *in vitro* experimental investigations rather than *in vivo* clinical observations. For this reason and others, we have not recommended that these three criteria alone serve as indications for fusion. However, a checklist has been developed that takes these criteria into consideration.

Incidence of Spontaneous Fusion

The high incidence of spontaneous fusion, which occurs without surgical intervention, leads to uncertainty about surgical fusion. Bailey and Bedbrook have both studied spontaneous fusions.[7,11] Table 5-6 shows some of the published figures of the incidence of spontaneous fusion reported by other investigators. Some are estimates, others are numerically documented. In any case, the incidence is high. To be able to predict which specific fracture dislocations will fuse spontaneously would solve the problem. Unfortunately, there is no published study that has attempted this.

TABLE 5–6 Incidence of Spontaneous Fusions Following Cervical Spinal Trauma

Investigator	Percent Spontaneously Fused
Rogers (1957)	36
Brav, et al. (1963)	42
Robinson and Southwick (1960)	50
Hørlyck and Rahbek (1974)	66

(White, A. A., Southwick, W. O., and Panjabi, M. M.: Clinical instability in the lower cervical spine. A review of past and current concepts. Spine, 1:15, 1976.)

Evaluation of Popular Arguments for Surgical Arthrodesis

The evidence in Table 5-6 argues well for the conservative approach of watching and waiting. Surgical enthusiasts deliberate about certain advantages of iatrogenic fusions. The most popular assertions are that surgical fusions reduce pain, improve stability, improve overall prognosis, facilitate nursing care, and reduce hospitalization time. Have fusions for fractures and fracture dislocations of the cervical spine been shown to reduce pain or improve stability? This question does not have a definitive answer, because pain is difficult to evaluate and stability has not been clearly defined. Munro reported that fusion reduced the incidence of pain in his patients by two-thirds, but there was no improvement in stability.[102] Rogers found no pain in a series of 39 patients who had undergone fusion.[134] Dall reported better reduction and stability in the few patients of his large series who were treated surgically.[36]

Effects on Nursing Care and Hospitalization Time

We found no studies in which this was investigated. The work of Brav and colleagues compared recumbency time of patients treated with and without fusion, and no difference was observed.[19] Norrell and Wilson suggested that early anterior fusions reduce the time required for bed treatment, but they give no figures to support this.[108] Durbin, Forsythe and colleagues, Petrie, and Rogers asserted that fusion reduces the hospital stay.[42,56,123,134] Munro reported that patients treated with fusion did not have shorter periods of hospitalization.[102] Brav and colleagues actually compared figures in fused and nonfused groups and observed no difference in the period of hospitalization.[19]

Effects on Overall Prognosis

Durbin, Forsythe and colleagues, Petrie, Rogers, and Verbiest contend that patients treated with fusion have a better overall prognosis than those treated nonsurgically.[42,56,123,134,155] Munro vigorously argues to the contrary in a paper that reviews and reinterprets the data of other investigators.[102] The series of Brav and colleagues, the only study that actually compares a number of fused and nonfused patients, showed the overall prognosis to be virtually the same in the two groups.[19]

The arguments for cervical spine fusion following trauma may be summarized as follows: There is virtually no convincing evidence in the literature to support this procedure. However, there is also no significant evidence to the contrary.

RECOMMENDED EVALUATION SYSTEM

The Checklist

This approach, like the pilot's checklist, is intended as a safety factor and to ensure that all pertinent factors are considered and appropriately balanced. We have "set" the sensitivity of the system at what in our opinion is the ideal level. The goal of this setting is to avoid *overtreatment* and *undertreatment* and to provide insurance against the development of additional sequelae from the basic clinical problem. Therefore, a total of 5 or more points on the checklist is necessary to make the diagnosis of clinical instability.

The Checklist for the Middle and Lower Cervical Spine

The system we propose is presented in Table 5-7. The patient is evaluated, and each item that applies is checked. If the numbers assigned to the checked items total 5 or more, then the spine should be considered clinically unstable. It is not assumed that the information available on all patients will provide a definitive answer for each item on the list. It is recommended that when the evaluation of a given element leads the clinician to a borderline decision that cannot be resolved, the value for that entity should be divided by 2 and added to the other points.

TABLE 5–7 Checklist for the Diagnosis of Clinical Instability in the Middle and Lower Cervical Spine

Element	Point Value
Anterior elements destroyed or unable to function	2
Posterior elements destroyed or unable to function	2
Positive stretch test	2
Radiographic criteria*	4
A. Flexion/extension x-rays	
1. Sagittal plane translation > 3.5 mm or 20% (2 pts)	
2. Sagittal plane rotation > 20° (2 pts)	
OR	
B. Resting x-rays	
1. Sagittal plane displacement > 3.5 mm or 20% (2 pts)	
2. Relative sagittal plane angulation > 11° (2 pts)	
Abnormal disc narrowing	1
Developmentally narrow spinal canal	1
1. Sagittal diameter < 13 mm	
OR	
2. Pavlov's ratio < 0.8†	
Spinal cord damage	2
Nerve root damage	1
Dangerous loading anticipated	1
Total of 5 or more = unstable	

* See Figures 5-35 and 5-36 for information on making these measurements.

† See Figure 5-35.

Anatomical Considerations

A schematic representation of the anatomy of the middle and lower cervical spine is presented in Figure 5-29. At the level of the intervertebral disc, the annulus fibrosus is the crucial stabilizing structure. Bailey emphasized the importance of this structure.[6] Munro carried out experimental studies on cadaver spines and concluded that cervical spine stability comes mainly from the intervertebral discs and the anterior and posterior longitudinal ligaments.[102]

White and Panjabi and colleagues performed experiments on cervical spine FSUs in high-humidity chambers using physiologic loads to simulate flexion and extension.[111,162] We defined the *anterior elements* as the posterior longitudinal ligament and all structures anterior to it. The *posterior elements* were defined as all structures behind the posterior longitudinal ligament. Based on these studies, it was suggested that if an FSU has all of its anterior elements plus one additional structure, or all of its posterior elements plus one additional structure, it will probably remain stable under physiologic loads. Therefore, in the checklist, in order to provide for some clinical margin of safety, we suggest that *any FSU in which all the anterior elements or all the posterior elements are either destroyed or unable to function should be considered potentially unstable.* Two points in the checklist are given for the loss of each of these anatomic elements.

One final anatomic consideration should be noted. If all other considerations are the same, patients with the anterior elements destroyed or unable to function are more clinically unstable in extension, while patients with the posterior elements destroyed or unable to function are more unstable in flexion. These considerations should be thought of during patient transfers and when immobilizing a patient's neck after injury.

Radiographic Criteria

The measurement of translation and displacement is shown in Figure 5-35. This method takes into account variations in magnifications and should be useful when there is a tube-to-film distance of 72 in.

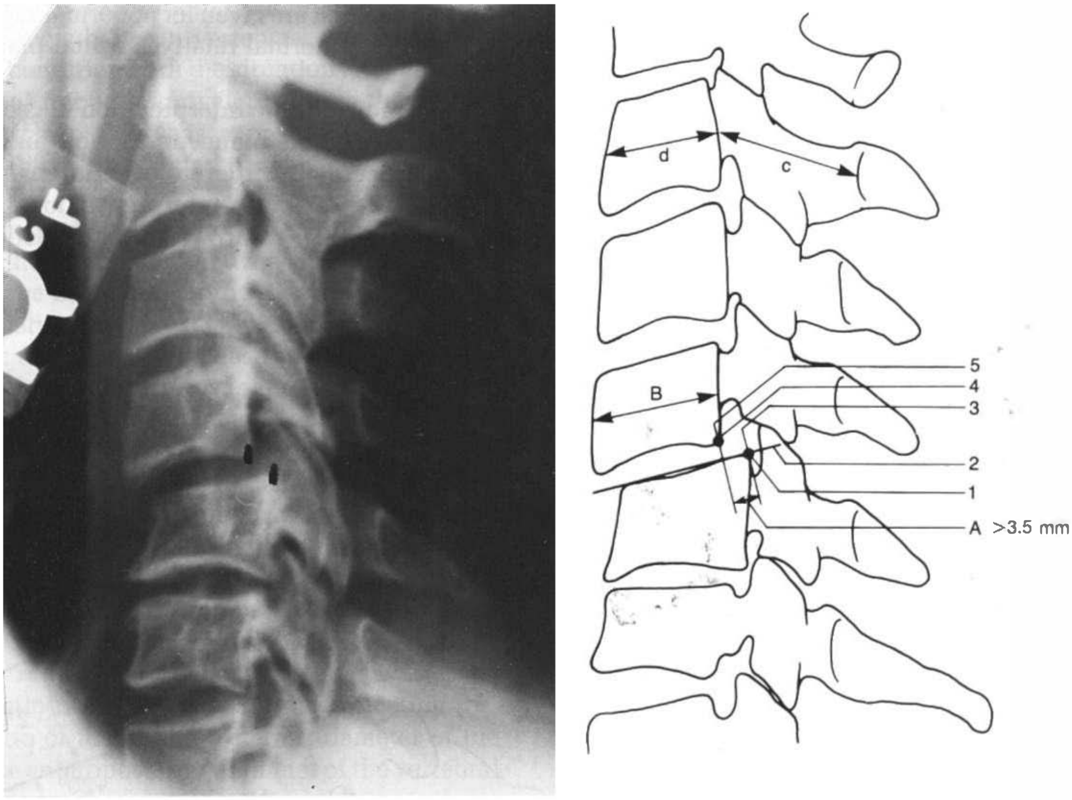

FIGURE 5-35 The method for measuring translatory displacement is as follows. (1) A point is marked at the posterosuperior angle of the projected image of the vertebral body below the interspace of the functional spinal unit (FSU) being evaluated. (2) A line is drawn along the upper vertebral end-plate of the vertebra below the interspace of the FSU under analysis. (3) At the point where this intersects the mark, at the posterior portion of the end-plate, a short perpendicular line is drawn. (4) Next, a mark is made at the posteroinferior angle of image of the vertebral body above the interspace of the FSU being evaluated. (5) A short line that goes through the second mark and is perpendicular to the line on the subjacent vertebral end-plate is drawn. The linear distance between the two perpendicular lines is measured. This can be called distance *A*. The anteroposterior sagittal plane diameter at the midlevel of the supra-adjacent vertebra is measured. This distance is called *B*. If distance *A* is >20% of distance *B*, then this is considered evidence of instability and should be so entered on the checklist. An *alternate* method is to simply measure the linear distance *A*, and if this is greater than 3.5 mm, it is considered to be suggestive of instability, and 2 points are entered onto the checklist.

Pavlov's ratio[113a] is a reliable, accurate method for recognizing a developmentally narrow canal without the variables involved in linear measurements. The measurement *c* is the distance between the midlevel of the posterior aspect of the vertebral body and the nearest point on the corresponding spinolaminar line. The measurement *d* is seen on lateral x-ray as the anteroposterior distance from the front to the back of the vertebral body measured at the midlevel. The ratio *c*/*d* is considered normal if 1 or greater and abnormal if less than .80. These measurements are used in conjunction with the checklist (Table 5-7).

Sagittal plane displacement or translation greater than 3.5 mm on either static (resting) or dynamic (flexion/extension) lateral radiographs, respectively, should be considered potentially unstable. This value was determined from an experimentally obtained value of 2.7 mm and an assumed radiographic magnification of 30%.[162] Two points in the checklist are given for abnormal sagittal plane displacement or translation.

Angular measurements are shown in Figure 5-36. There is no magnification problem in measuring rotation or angulation. Greater than 20° of sagittal plane rotation on dynamic (flexion/extension) radiographs should be considered abnormal and potentially unstable. This value was based on a review of the literature on *in vitro* and *in vivo* cervical spine ranges of motion. When dynamic radiographs are unable to be obtained (e.g., in an acute traumatic setting), a static (resting) lateral radiograph that shows greater than 11° of relative sagittal plane angulation should be considered potentially unstable.[E] Note that 11° of relative angulation means 11° *greater* than the amount of angulation at the FSU above or below the FSU in question. This standard of comparison takes into account the normal angulation between FSUs (*i.e.*, normal posture). Two points in the checklist are given for abnormal sagittal plane rotation or abnormal relative sagittal plane angulation.

The radiographic interpretation in general, especially for sagittal plane translation and displacement, is decidedly different in children up to 7 years of age.[28] It is risky to interpret radiographs of patients in this age group without knowledge of some

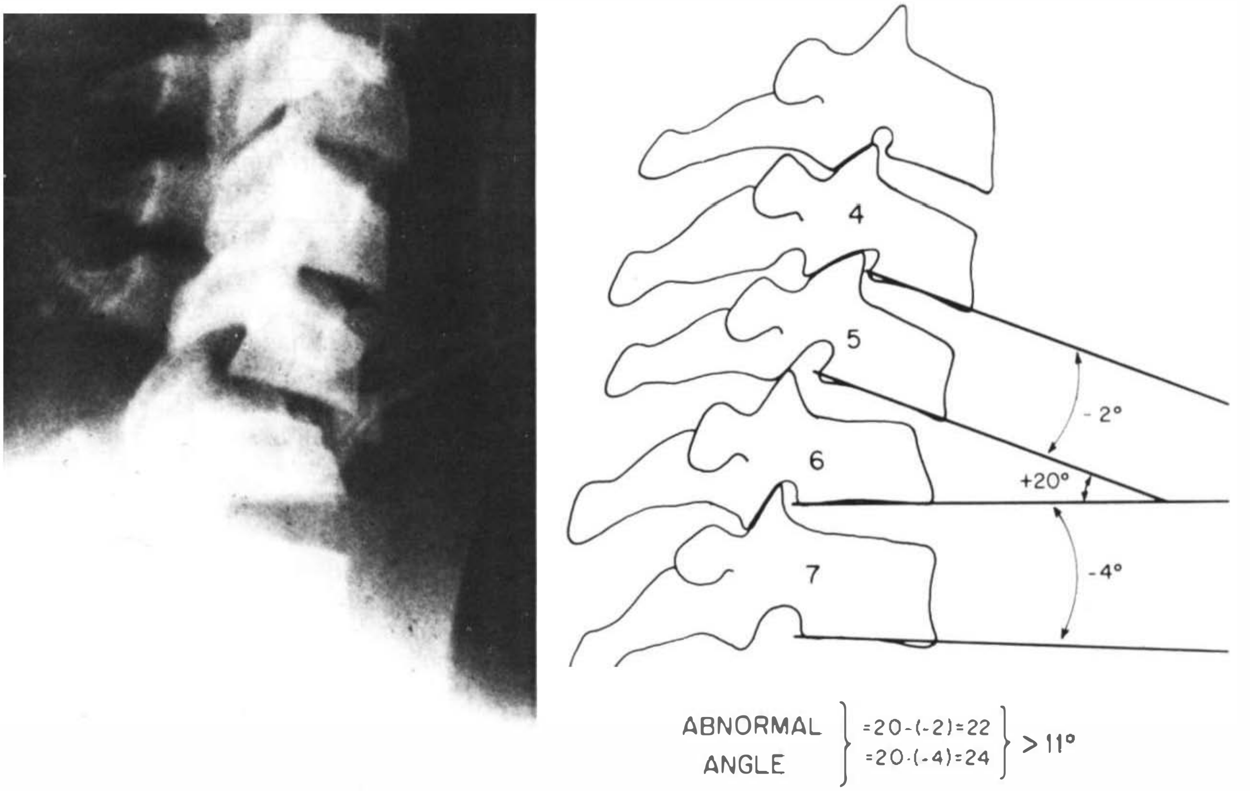

FIGURE 5-36 The angulation between C5 and C6 is 20°, which is more than 11° greater than that at either adjacent interspace. The angle at C4 and C5 measures −2°, and the one at C6 and C7 measures −4°. This finding of abnormal angulation is based on a comparison of the interspace in question with either adjacent interspace. This is to allow for the angulation that is present due to the normal lordosis of the cervical spine. We interpret a difference of 11° or greater as evidence of clinical instability. These measurements are to be used in conjunction with the checklist (see Table 5-7). *(White, A. A., Johnson, R. M., Panjabi, M. M., and Southwick, W. O.: Biomechanical analysis of clinical stability in the cervical spine. Clin. Orthop., 109:85, 1975.)*

of the normal findings that may appear to be pathological to the uninitiated.

Controlled, monitored axial traction (the "stretch test") may be helpful in the evaluation of the integrity of the ligamentous structures of the middle and lower cervical spine. (Refer to Fig. 5-39 and the boxed insert for a diagrammatic synopsis of this test and the details of the procedure.) An abnormal test is indicated by either differences of more than 1.7 mm interspace separation or more than 7.5° of change in angle between vertebrae, comparing the prestretch condition with the situation after application of axial traction equivalent to one-third body weight.[137] If a patient has a positive stretch test, 2 points should be added to the checklist (see Table 5–7).

Two final radiographic considerations should be noted. First, in the traumatized spine there may frequently be narrowing of the disc at the damaged FSU.[7] In patients under 35 years of age, post-traumatic disc narrowing is modestly suggestive of disruption of the annulus fibrosus and of possible instability. Second, if all other considerations are the same, patients with a congenitally narrow spinal canal are more apt to develop neurologic deficit because of less space available for the spinal cord. A congenitally narrow canal is defined as measuring less than 13 mm in its anteroposterior dimension on a lateral radiograph[50] or having a Pavlov's ratio of less than 0.8.[113a] The 13-mm absolute value accounts for some radiographic magnification, while the Pavlov's ratio need not consider magnification (being the ratio of the anteroposterior diameter of the canal to the anteroposterior diameter of the vertebral body). One point each in the checklist is given for abnormal disc narrowing or a congenitally narrow canal.

Neurologic Criteria

Is the presence of distinct medullary or root damage associated with spinal trauma or disease evidence of clinical instability? This consideration deserves some discussion with regard to our definition of clinical instability. We have said that clinical instability concerns the prediction of subsequent neurologic damage. Therefore, what is the significance of the presence of *initial* neurologic damage to the probability of subsequent neurologic damage? We believe that if the trauma is severe enough to cause initial neurologic damage, the support structures *probably* have been altered sufficiently to allow subsequent neurologic damage, and thus the situation is clinically unstable. It should be noted, however, that Gosch and colleagues have shown in animals that it is possible to produce medullary damage with intact supporting structures.[61] In general, despite a few exceptions, we believe that neurologic deficit is an important consideration in the evaluation of clinical instability. Evidence of root involvement is a weaker indicator of clinical instability. For example, a unilateral facet dislocation may cause enough foramenal encroachment to result in root symptoms and/or signs but not enough ligamentous damage to render the FSU unstable. Two points in the checklist are given for spinal cord damage and one point for nerve root damage.

Physiologic Criteria

The final checklist consideration involves the important individual variation in physiologic load requirements, especially with regard to differences in habitual activities. The clinician employs judgment in an attempt to anticipate the magnitude of loads that the particular patient's spine is expected to maintain after injury. Anticipating dangerous loads can be especially helpful when other available criteria are inconclusive. One point in the checklist is given if dangerous loading is anticipated. Examples of patients in whom one can anticipate heavy loads are heavy laborers, contact sport athletes, and avid motorcyclists.

Some Cogent Clinical and Experimental Points

Case Report

L. C. is a 23-year-old female who was involved in an automobile accident. She was unconscious for several minutes after the accident. When awake, she had neck and arm pain. However, the neurologic examination was within normal limits. Figure 5-37A shows a subluxation of C4 on C5. Figure 5-37B is a laminagram taken with only 17 lb of axial traction. When that amount of axial displacement was noted, the traction was reduced. The patient's neurologic status remained normal.

Discussion

This case report illustrates several important points. The diagnostic value of axially directed traction to establish the presence of ligament disruption is demonstrated. Traction revealed that there was total disruption of all the anterior and posterior liga-

FIGURE 5-37 Radiographs of a patient with disruption of all ligaments between C4 and C5. **(A)** The spine in resting position.

(continued)

ments, which rendered the spine grossly unstable. The radiograph in Figure 5-37B exhibits dramatically the observations of Breig, who found that the cord can withstand considerable axial displacement without structural damage and neurologic deficit.[20] The case report also shows the necessity of careful application of axial traction with close monitoring. While this patient had no additional damage with this degree of displacement, certainly it is not desirable to distract the cord and vertebral column to such an extent. Therefore, we recommend, especially in weak-muscled individuals, that the traction be applied in small increments with frequent lateral radiographs to check the position. The procedure in the case report was not a formal stretch test. The protocol for the stretch test is designed so that early minimal abnormal displacement can be recognized. Early recognition permits demonstration of loss of ligamentous continuity and prevents subsequent damage, which may be imminent in Figure 5-37B.

The Stretch Test

A detailed laboratory investigation of the patterns of displacement of the axially loaded cervical spine following ligament transections has been carried out. Figure 5-38 shows the experimental setup. The clinically relevant finding is as follows. Compared

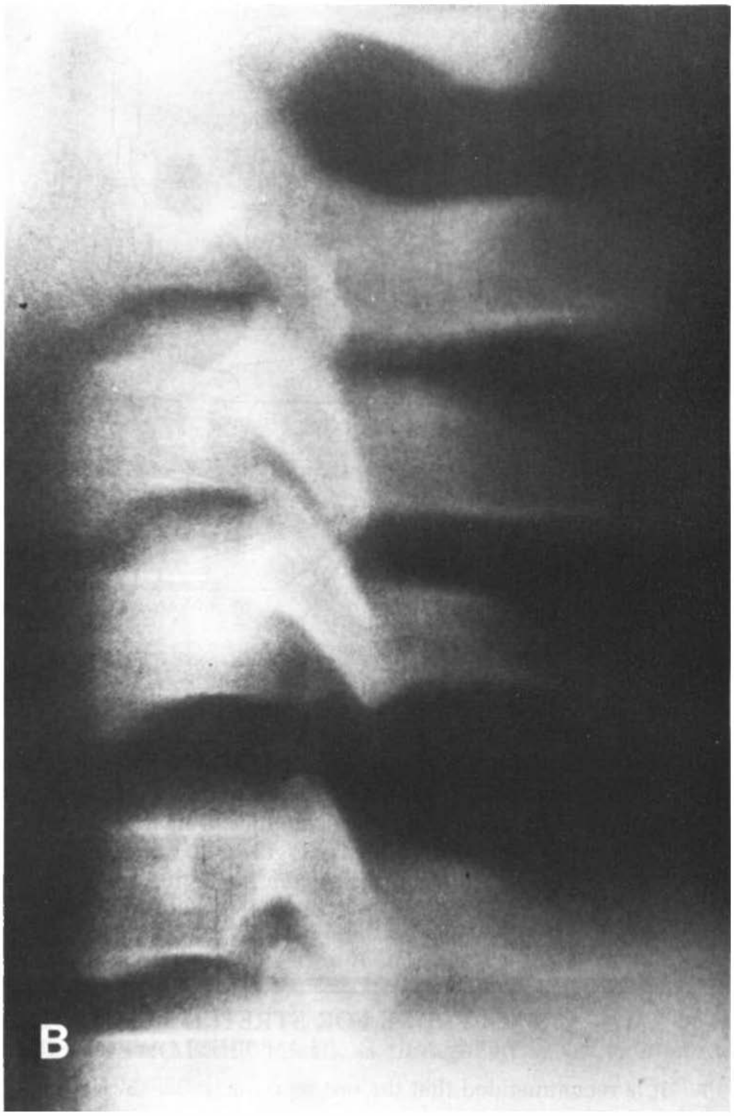

FIGURE 5-37 (continued)
(B) Laminagram of the spine after the application of 17 lb of traction. There was no neurologic damage or irritation.

with the pattern of motion observed in the intact FSU, an abnormal pattern of motion appears when all of the anterior structures or all of the posterior structures have been transected.[112]

The procedure for the stretch test is outlined briefly in the accompanying display and in Figure 5-39.

These guidelines, based on studies of eight normals, suggest that an abnormal stretch test is indicated by differences of >1.7 mm interspace separation or >7.5° change in angle between vertebrae, comparing the prestretch condition with the situation after the application of axial traction equivalent to one-third body weight.[F] These are guidelines that will be improved with additional experience. It can be expected that with some cases, abnormal separations at the level of injury may be seen and evaluated by clinical judgment.

Clinically, we have found the stretch test useful in providing some additional assurance in circumstances in which a decision is to be made on the advisability of allowing an athlete back to a contact sport following a neck injury with no obvious bony or ligamentous disruption. A negative stretch test adds an element of security to the decision to allow participation in contact sports.

In regard to the athlete and contact sports, the presence of a developmentally narrow canal implies

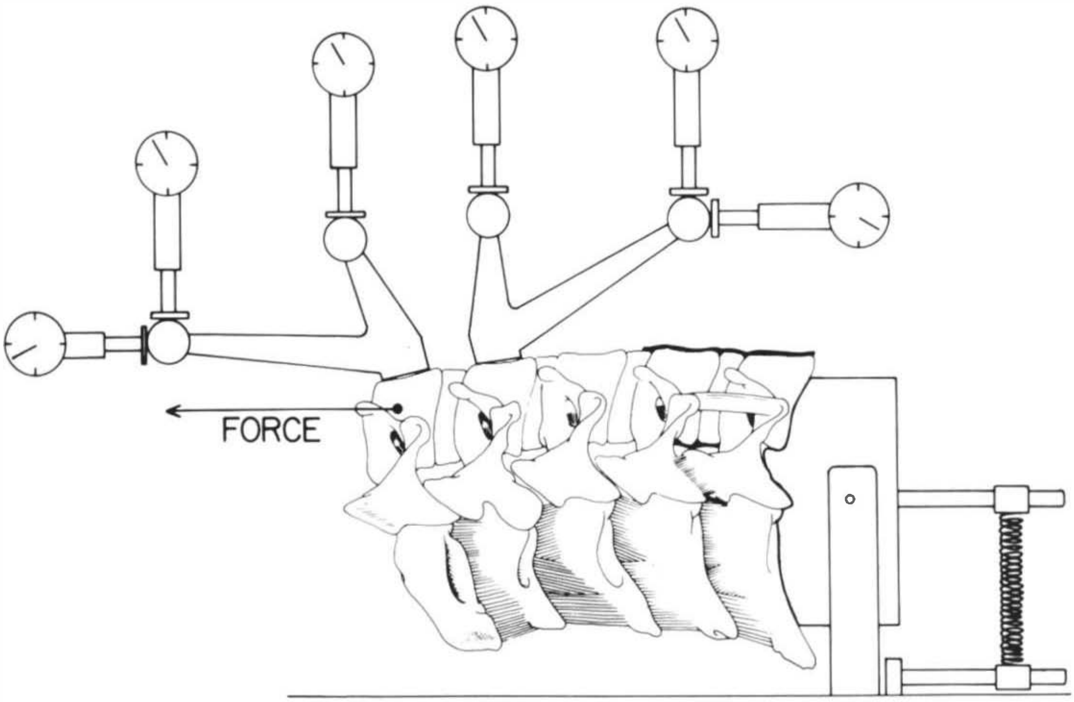

FIGURE 5-38 The experimental arrangement in the stretch test. A force equal to one-third the body weight is applied; changes in rotation and separation of disc spaces as a function of ligament transection are measured by radiographs, and the displacement gauges are attached to steel balls. The same schedules of ligament transections (anterior to posterior and vice versa) as those described in Figure 5-29 were used here. *(White, A. A., Southwick, W. O., and Panjabi, M. M.: Clinical instability in the lower cervical spine. Spine, 1:15, 1976.)*

PROCEDURE FOR STRETCH TEST TO EVALUATE CLINICAL STABILITY IN THE LOWER CERVICAL SPINE

1. It is recommended that the test be done under the supervision of a physician.
2. Traction is applied through secure skeletal fixation or a head halter. If the latter is used, a small portion of gauze sponge between the molars improves comfort.
3. A roller is placed under the patient's head to reduce frictional forces.
4. The film is placed 0.36 m (14 in) from the patient's spine. The tube distance is 1.82 m (72 in) from the film.
5. An initial lateral radiograph is taken. Review this, and the first film after traction is applied, carefully for C0–C1–C2 subluxations or dislocations. Abnormal displacements in this region can sometimes be difficult to identify.
6. A 4.5 kg (10-lb) weight is added. (If the initial weight is 4.5 kg (10 lb), this step is omitted.)
7. Traction is increased by 4.5 kg (10-lb) increments. A lateral film is taken and measured.
8. Step 7 is repeated until either one-third of body weight or 29.5 kg (65 lb) is reached.
9. After each additional weight application, the patient is checked for any change in neurologic status. The test is stopped and *considered positive* should this occur. The radiographs are developed and read after each weight increment. Any abnormal separation of the anterior or posterior elements of the vertebrae is the most typical indication of a positive test. There should be at least 5 minutes between incremental weight applications; this will allow for the developing of the film, necessary neurologic checks, and creep of the viscoelastic structures involved.
10. The test is contraindicated in a spine with obvious clinical instability.

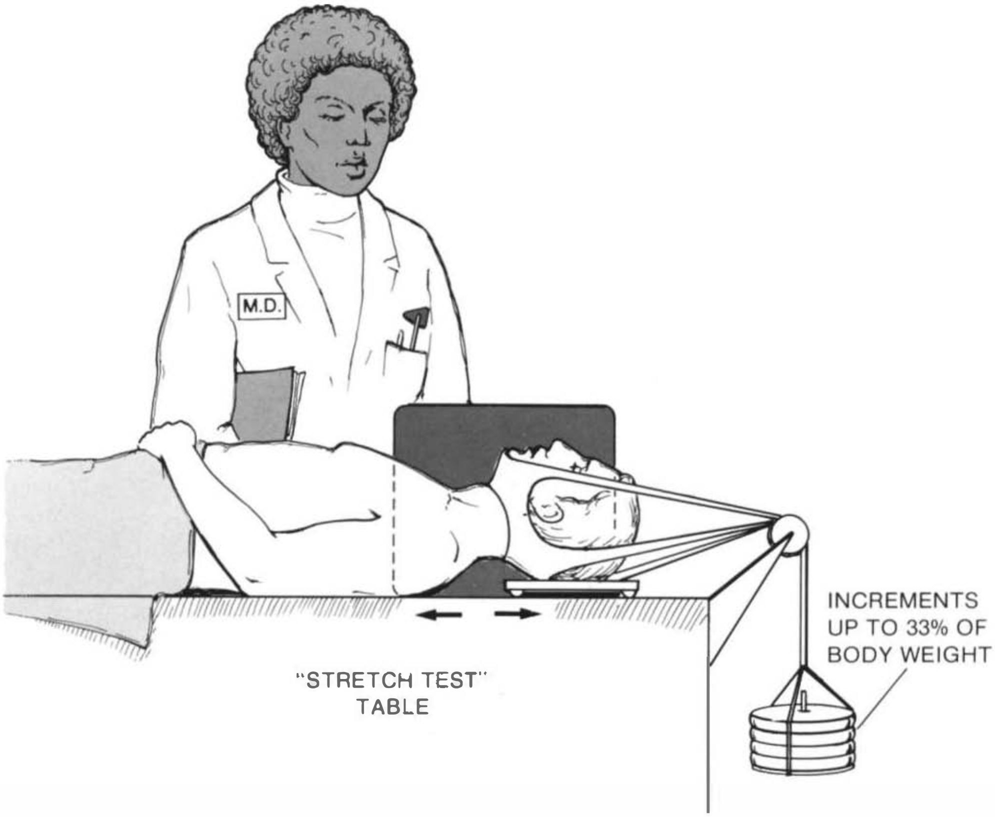

FIGURE 5-39 Diagrammatic synopsis of stretch test. A physician who is knowledgeable about the test is in attendance. The neurologic status is monitored by following signs and symptoms. Incremental loads up to 33% of body weight or 65 lb are applied. Each lateral radiograph is checked prior to augmentation of the axial load. Note the neurologic hammer to symbolize neurologic exam and the roller platform under the head to reduce friction.

a lower threshold for neurologic problems should there be a spinal injury.

Disc Space Narrowing or Widening

We have previously reported the observation by Bailey that in the traumatized spine there may be narrowing of the disc space at the damaged FSU.[6] We submit that this finding is modestly suggestive of disruption of the annulus fibrosus and possible instability.

A widened disc space (*i.e.*, perceptibly more gap in one space than in either adjacent one) may also be an indication of a disrupted annulus and instability.[31] This finding, which can be confirmed by a controlled extension view (perhaps more safely with a stretch test), is thought to be an extension injury.

Decreased Anteroposterior Diameter of Canal

There is statistically significant evidence to suggest that the size of the spinal canal is an important prognosticator of the likelihood of neurologic damage in cervical spine trauma.[47] Thus, we suggest that another point be added to the checklist for situations in which there is a narrow anteroposterior diameter in the canal (*i.e.*, <15 mm). The linear measurements are made as shown in Figure 5-35. Pavlov's ratio to evaluate canal size can also be used as a method of controlling for x-ray magnification (see Fig. 5-35).

FIGURE 5-40 Radiographs of patient with nontraumatic clinical instability. **(A)** Marked compression of the vertebral body of C6. **(B)** Lateral view showing angulation secondary to disruption of support function of anterior elements. **(C)** Myelogram showing myelographic block resulting from spinal canal encroachment due to vertebral collapse and angulation.

Example of Clinical Stability Evaluation

Patient N. K., a 25-year-old male graduate student from India, complained of posterior neck pain with radiation into both arms. There was numbness in the left hand in the distribution of the C5–C6 dermatome. Radiographs are shown in Figure 5-40. The anteroposterior view demonstrates collapse of the body of C6 and destruction of the lateral mass of C6. The lateral view shows the same, and in addition there is abnormal sagittal plane (z-axis) rotation of C5. The myelogram shows an incomplete block at the level of C5.

If the checklist is used to evaluate this patient for clinical stability, points would be given as follows: the anterior elements are destroyed and unable to function, 2 points; relative sagittal plane rotation is greater than 11°, 2 points; there is root damage, 1 point. The total is 5 points, and the patient was thought to be clinically unstable.

The diagnosis was tuberculous osteomyelitis. The patient had laminectomies of C5 and C6 at another hospital. This procedure was associated with some relief of pain but rendered the situation even more clinically unstable. He was subsequently treated with posterior facet fusion and anterior resection with iliac graft to replace the vertebra. These resulted in a clinically stable, pain-free patient with no neural defect.

Discussion

We have attempted to choose a number that would "set" the sensitivity of the system at just the proper level. In other words, if a score of 9 (4 above our "setting") were required in order to make a diagnosis of clinical instability, only those patients in grossly obvious imminent danger would have insurance against the problems of instability. Physicians would almost never overtreat a patient for clinical instability. Such a high setting would leave a number of patients inadequately treated. Conversely, a low diagnostic score, such as 1 (4 below our "setting"), would result in a large number of clinically stable spines being *treated unnecessarily*. However, rarely would there be a neurologic catastrophe due to a clinically unstable spine being left untreated. A score of 5 is thought to prevent unnecessary surgery or too vigorous treatment, yet at the same time it provides reasonable insurance against the unhappy development of additional root or cord damage (Fig. 5-41). The system is presented to capsulize our views, to stimulate others, to develop better systems, and to help physicians think about and use specific reproducible criteria for arriving at a diagnosis of clinical instability.

A recent study of 52 patients with cervical spine injury states that only one case of instability was detected and that three other patients who fulfilled the criteria of clinical instability were symptom-free.[3] It is concluded that the criteria (>3.5 mm of sagittal plane translation, >11° of rotational difference from either adjacent vertebra) alone cannot establish the indications for surgical intervention. The investigators state that the dislocations were reduced partially or completely in traction with Crutchfield tongs for an average of 7.9 weeks, and then a plastic collar was applied so as to continue immobilization for a total of 3.5 months. Therefore, all patients were treated adequately nonsurgically and then evaluated for instability. It appears from the information given that most of these patients were treated successfully with prolonged immobilization. Apparently, the investigators did not use the checklist as it is recommended to be used. As outlined in this chapter, the checklist is intended to be used to determine which patients need surgery or some form of prolonged immobilization.

As previously delineated, clinical instability in the checklist system is defined as one of the following: intractable pain, major deformity, or neurologic deficit. Thus, by definition, most of the patients in this study, having been treated with traction and immobilization, were not unstable, so the checklist need not be applied. One of the patients met two of the checklist criteria and had a complete neurologic recovery after a posterior cervical fusion. The assertion by the investigators that the "criteria" cannot establish the indications for surgical intervention is an example of the failure to use a common definition of instability and a failure to separate the treatment of instability from its diagnosis. An unstable or potentially unstable spine does not necessarily have to be treated surgically, but it must be adequately managed until it is no longer unstable.

TWELVE SIGNIFICANT SIGNS OF CERVICAL SPINE TRAUMA

The retropharyngeal space (pharyngeal air column to anteroinferior body of C2) should be considered abnormal if it is >7 mm in adults or children. The retrotracheal space (posterior tracheal air shadow to the anteroinferior body of T6) should be considered abnormal if it is >12 mm in children and >22 mm in

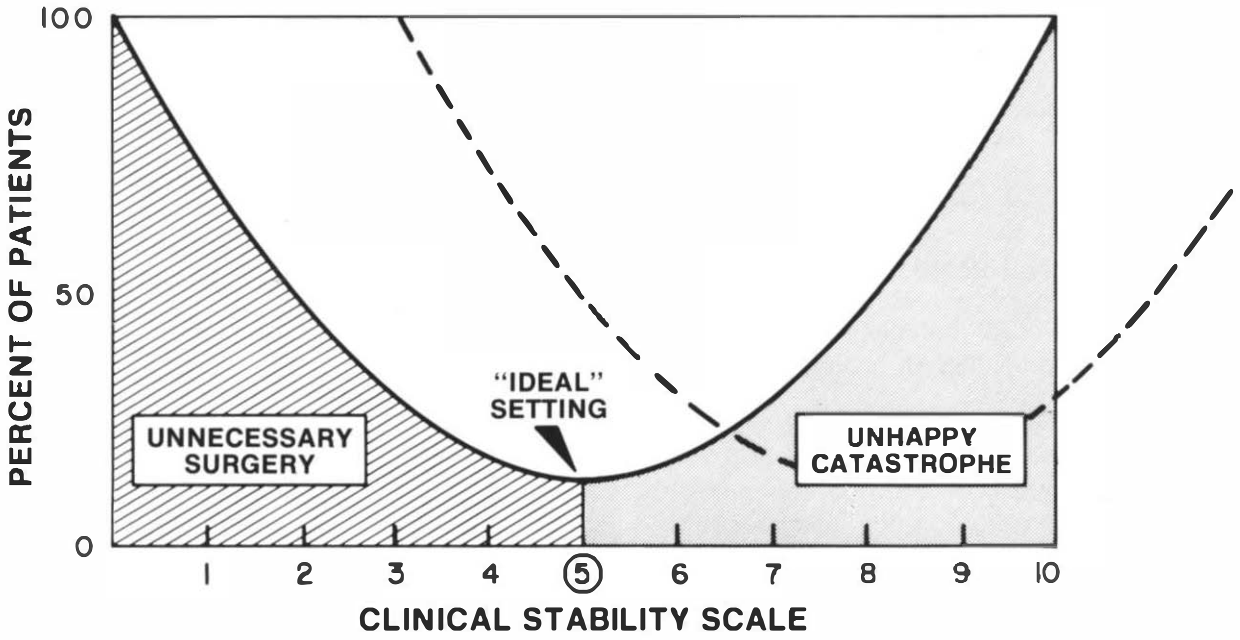

FIGURE 5-41 This is a theoretical graph depicting conceptually the choice of 5 as an "ideal" sensitivity setting. The ordinate shows percentages of *improperly treated* patients. The improper treatment could result in unnecessary surgery or unhappy catastrophe. The ideal setting should be the point at which the lowest percentage of patients are treated improperly. In our best judgment, that point is 5 on the *clinical stability scale.* We believe that this theoretical curve is correct; however, the real curve may be shifted to the right or to the left. For example, if it is shifted to the right, as shown by the dotted curve, then a cutoff of 5 on the stability scale would result in a significantly large percentage of improperly treated patients. Assuming the curve is as indicated on the graph, a setting of 1 would involve a large percentage of unnecessary surgery and no unhappy catastrophe. A setting of 9 would avoid any unnecessary surgery but would permit a large percentage of unhappy catastrophes.

adults. Measurements of the distance between the midpoint of the posterior margin of the anterior arch of the atlas and the anterior portion of the dens should not exceed 3 mm in the adult or 5 mm in the child.

Although there is some overlap with what has been presented, we considered it important to include Table 5-5 from the work of Clark and colleagues.[32] These radiographic signs, derived from a study of over 400 cases, do not necessarily indicate clinical instability, but they do alert the reviewer to the fact that there may be significant cervical spine trauma and that a thorough evaluation is in order.

Relative Clinical Instability in Flexion or Extension

Experimental studies have shown that the anterior ligaments contribute more to stability of the spine in extension than the posterior ligaments.[162] The posterior ligaments limit flexion more effectively than the anterior ligaments. This information makes it possible to ascertain in some clinical situations whether a given spine is likely to be more unstable in flexion or in extension. For example, in handling a patient in whom all of the anterior elements are destroyed, support of the patient to prevent extension is more important. This is also applicable to external support, as with transferring the patient and applying plaster immobilization or internal fixation. The converse is equally valid in the case of loss of the posterior elements. Figure 5-42 is a diagrammatic presentation of this basic concept.

RECOMMENDED MANAGEMENT

After defining and identifying clinical instability in the lower cervical spine, how is it to be treated? Even when a particular spine has been determined to be clinically unstable, fusion is not necessarily the best

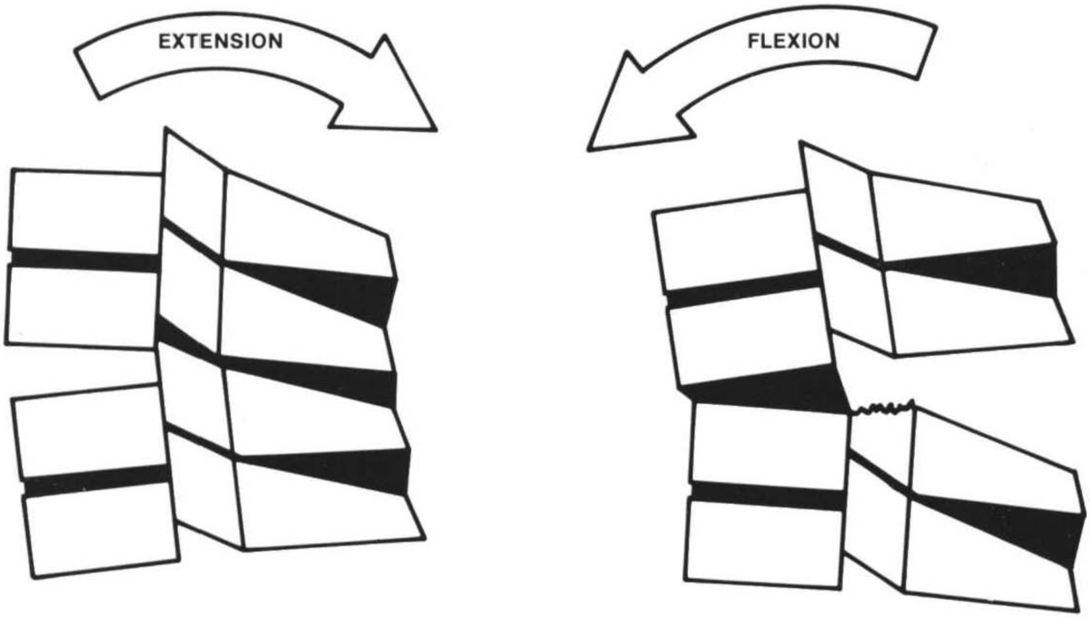

FIGURE 5-42 If all other considerations are the same, (**A**) when the anterior elements are ruptured or cut, the patient is more unstable in extension; (**B**) when the posterior elements are ruptured or cut, the patient is more unstable in flexion.

method of treatment. The evidence in the literature is inconclusive on the relative merits of surgical and nonsurgical treatment. There is a paucity of information on the use of the various halo devices for treatment of the unstable spine.

Management Flow Chart

A schematic organization of the management of clinical instability is presented in Figure 5-43.

Starting with patients with cervical spine trauma, initial treatment consists of bed rest for 1–7 days, in skeletal traction if there is evidence of severe injury. Patients with spinal cord involvement, with and without fractures or fracture dislocations, are considered to have major injuries. The patients without evidence of neurologic deficit who have either no fracture or only a minor compression fracture, without evidence of other damage, may be treated with head-halter traction. Minor injuries, such as strains, sprains, and muscle pulls, may be treated for symptoms and observed with subsequent radiographic examinations as indicated.

During the first week of traction, patients are given a thorough clinical evaluation and whatever supportive care is required. After the patient is stabilized physiologically and evaluated for decompression, closed reduction with traction may be attempted, and, if necessary, the various tests and maneuvers to rule out clinical instability may be performed.

It would be worthwhile to have a checklist to make a determination about decompression; however, at present, only the previously presented guidelines are available. Appropriate decompressions are carried out where indicated, and the patient is then evaluated for clinical stability with the checklist provided. If the decompression itself renders the spine clinically unstable, reconstruction and fusion may be carried out at the time of decompression.

All patients who are diagnosed as clinically stable can be treated with some modification of the following regimen. The assumption is that in 3–6 weeks, these patients will be comfortable, and whatever damage they sustained will be healing. A four-poster cervical collar with a thoracic attachment is desirable for adequate support and an effective intermediate range of control of movement. Such a device, along with the intrinsic stability of the spine, should be adequate to protect the patient from neurologic damage and to allow injured structures to heal. It is possible that clinical instability may develop at a later time.

We have recommended below a basic schedule for follow-up of patients treated for problems of clin-

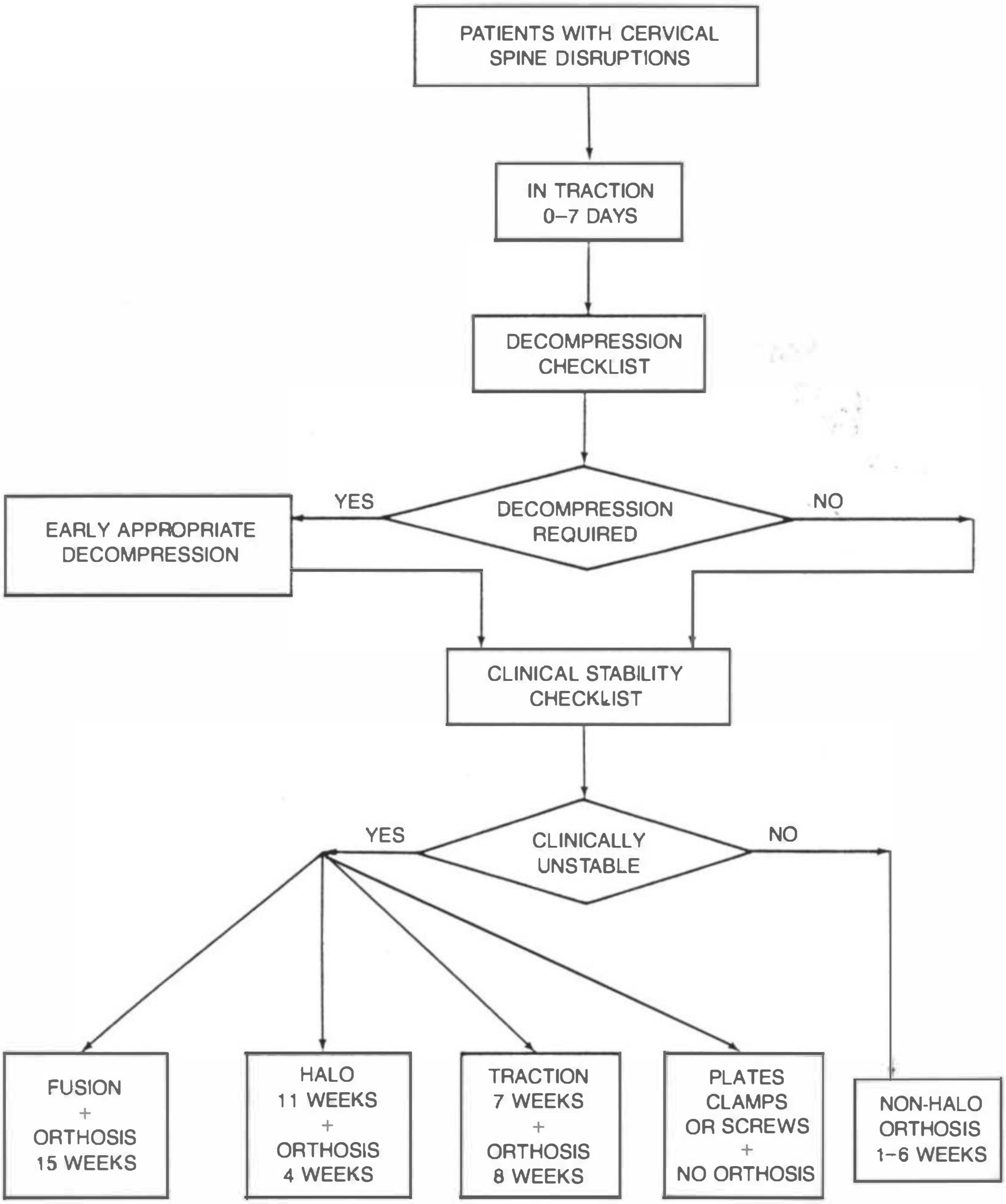

FIGURE 5-43 Recommended flow diagram for the management of patients with disruptions of the lower cervical spine. Treatment regimens include occupational and physical therapy, as tolerated, and a follow-up schedule as outlined on page 327. In the 1990s, another treatment can be considered. This involves the use of anterior and/or posterior instrumentation. The most appropriate indications and risks/benefits of this treatment have not yet been determined.

SUGGESTED FOLLOW-UP SCHEDULE FOR MANAGEMENT OF CLINICALLY STABLE AND UNSTABLE SPINE PROBLEMS

1. The schedule begins after the termination of initial treatment, surgical or nonsurgical (after removal of brace, cast, completion of bed rest, and initial physical therapy).
2. The patient is to be seen *3 weeks, 6 weeks, 3 months, 6 months,* and *1 year* after termination of initial treatment.
3. If the appropriate clinical evaluation is carried out on this time schedule, all early and late complications should be recognized in time for treatment. Whenever possible, radiographic techniques should be standardized for purposes of comparison.

ical instability in all regions of the spine. This schedule should allow cases of delayed instability to be recognized and treated before any complications occur. The visits are primarily for radiologic and neurologic evaluation. The frequency of the schedule is altered according to the individual patient's progress and prognosis.

Treatment of Clinically Unstable Conditions

The three alternatives suggested provide clinical stability in the majority of patients. There is no convincing evidence in the literature that any of the three approaches is superior. The considerations are complex and have not been thoroughly studied. A successful fusion constitutes the strongest reconstruction for the unstable FSU. However, it carries all the risks of spinal surgery in a seriously ill patient. Munro considered the risks to be significant.[103]

More recently, the halo apparatus has been used in the treatment of spinal trauma, with or without fusion.[85] This apparatus may have an important place in the treatment armamentarium of this disease. The use of the halo apparatus (a halo attached to an outrigger stabilized on the body) may serve effectively as the primary treatment for some of these injuries. This method has promise and merits careful study in the management of clinical instability in the lower cervical spine.

The halo apparatus offers the best immobilization for the facilitation of ligamentous healing, and it may shorten hospitalization for the patient who has no major neurologic deficit. However, there has been no extensive experience with it as a primary treatment for clinical instability. The halo apparatus has its own complications. In addition, there is the unanswered question of whether ligamentous healing in any given case of spinal trauma is of satisfactory strength to withstand physiologic loads. The use of skeletal traction for a total of 7 weeks, followed by an orthosis of intermediate control for 8 weeks, is probably a more conservative approach and can be expected to be effective. We recommend 7 weeks of traction because Brav and colleagues showed that patients treated for 6 weeks or more had a redislocation rate of only 2.3%.[19]

All of these regimens include careful clinical follow-up evaluations. Lateral views of flexion/extension films are helpful for recognizing failed treatment or delayed or progressive subluxation or deformity. There are, of course, other combinations and regimens. Some surgeons prefer early or late Minerva casts. We believe that the halo apparatus gives better immobilization. The usefulness of the halo apparatus in the treatment of cervical spine injuries is described in more detail in Chapter 8.

Each surgeon must choose from the various alternatives. The ideal would be to have some well-controlled prospective clinical studies comparing the effectiveness of the different regimens. The work of Hans Ersmark[47a] is a notable step in this direction. His work supports the superiority of the halo apparatus over both traction and surgery, in a well-executed study. Of course, it is essential to evaluate the various treatments in regard to clinical stability. In addition, patient attitude, cost benefits, complications, rehabilitation, nursing care, and hospitalization time are all crucial variables to be evaluated before the best treatment regimen can be determined.

Because our experience, and that of others, has included a long tradition of success with surgical fusions of unstable cervical spines, we recommend this procedure. Obviously, the halo apparatus is a strong contender, and more studies are needed to determine the best treatments for the clinically unstable cervical spine.

PART 3:
THE THORACIC SPINE (T1–T10) AND THE THORACOLUMBAR SPINE (T11–L1)

There are several unique considerations in the evaluation of clinical stability in the thoracic and thoracolumbar spine. This region of the spine is mechanically stiffer and less mobile than any of the other regions. There is less free space and a more precarious blood supply for the spinal cord in this region of the spine. It is well stabilized by the costovertebral articulations and the rib cage structure. Therefore, greater forces are required to disrupt it. The thoracic and thoracolumbar spine instability checklist represents our opinion about how to best diagnose clinical instability in this region, based on the currently available experimental and clinical information. Before we actually present the checklist, here is some relevant background information.

ANATOMIC CONSIDERATIONS

There are several anatomic characteristics that relate to the biomechanics of this region and therefore affect its clinical stability. There is a normal thoracic kyphosis, which is due to the slight wedged configuration of both of the vertebral bodies and intervertebral discs. Because of this physiologic kyphosis, the thoracic spine is more prone to be unstable in flexion.

The Anterior Elements

The anterior longitudinal ligament is a distinct, well-developed structure in the thoracic region (Fig. 5-44). Clinicians have noted that this structure is unusually thick in certain cases of abnormal thoracic kyphosis. This ligament is less developed in the cervical region.[79] The annulus, in this region as elsewhere, is one of the major factors in maintaining clinical stability. The posterior longitudinal ligament is also a distinct, well-developed structure that is of considerable importance in this region of the spine. The articulation of the ribs between vertebrae at the level of the intervertebral disc provides considerable additional stability to the thoracic spine FSUs. The radiate ligaments and the various costotransverse ligaments provide stability by binding adjacent vertebrae to the interconnecting rib (Fig. 5-45).

The Posterior Elements

The yellow ligaments are thicker in the thoracic region. The capsular ligaments in the thoracic region, in contrast to those in the cervical region, are thin and loose, which is significant anatomically. It

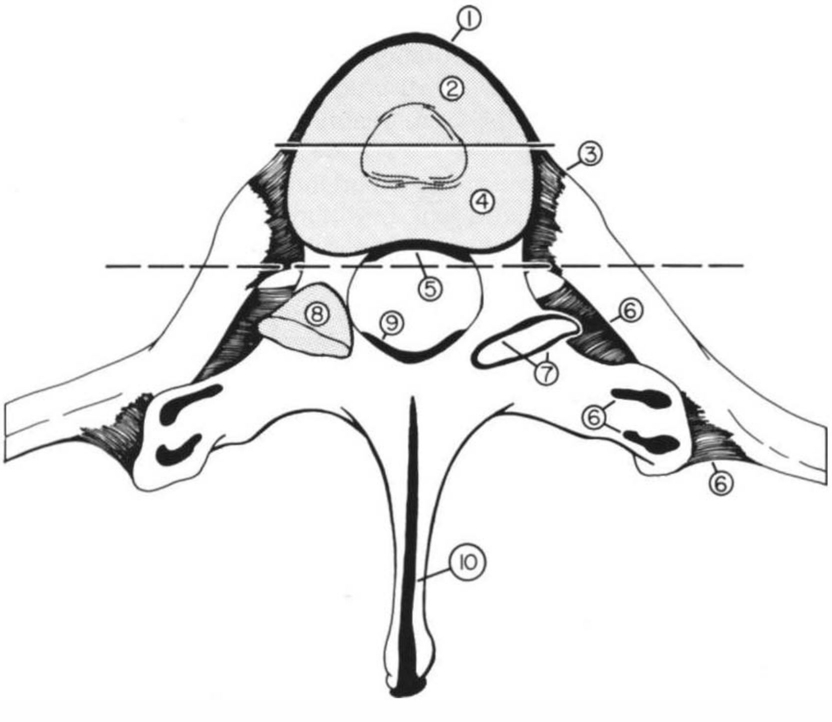

FIGURE 5-44 Schematic representation of the major ligaments involved in the thoracic spine. Ligaments are numbered according to the order in which they were cut in the biomechanics experiment referred to in the text. Anterior elements: 1, anterior longitudinal ligament; 2, anterior half of the annulus fibrosus; 3, radiate and costovertebral ligaments; 4, posterior half of the annulus fibrosus; 5, posterior longitudinal ligament. Posterior elements; 6, costotransverse and intertransverse ligaments; 7, capsular ligaments; 8, facet articulation; 9, ligamentum flavum; 10, supraspinous and interspinous ligaments.

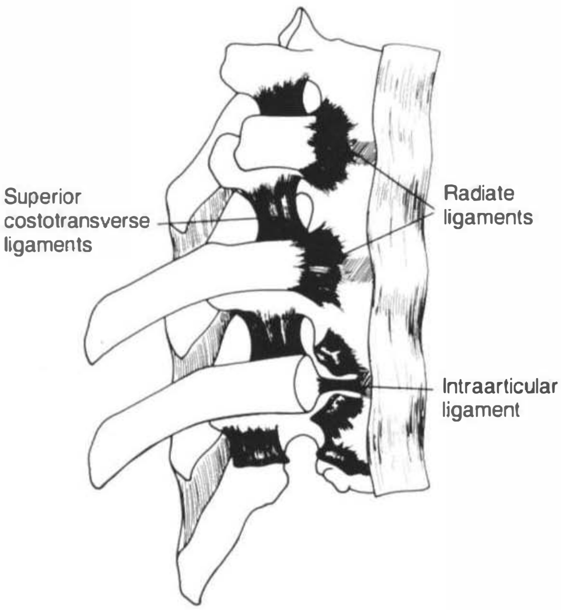

FIGURE 5-45 This diagram highlights ligamentous structures in the costovertebral articulation that make some contribution to the clinical stability of the thoracic spine. Note the radiate ligaments attaching to the head of the rib and to *both adjacent* vertebral bodies. There are also the costotransverse ligaments, which may offer some secondary stability.

means that the support provided by these structures against flexion is minimal and is much less than might be expected in the cervical spine, where there are well-developed capsules on the articular facets. Therefore, following multiple laminectomies in the thoracic spine, the usual support provided by the thick yellow ligament is lost, leaving the facet articulations, which can offer little support. This characteristic, along with the physiologic kyphosis, predisposes the thoracic spine to clinical instability after laminectomies at several levels. The rib articulation is able to provide some stability if it remains intact. The interspinous ligaments are not as thick in the thoracic as in the lumbar area. The supraspinous ligaments probably do not play a major role in the stability of this portion of the spine.

The osseous components of the facet joints and their fibrous capsules contribute significantly to clinical stability in the thoracic spine. The spatial orientation of the facet joints also has some significance. The kinematic factors are described in Chapter 2. In the middle and upper portion of the thoracic spine, the facets provide stability *primarily* against anterior translation. When the orientation changes, which may happen anywhere between T9 and T12, the facets provide less stability against anterior and posterior displacement. The joints are oriented more in the sagittal plane and therefore provide stability against axial rotation (see Fig. 2-22). Therefore, in the presence of excessive axial rotatory displacement in the lower thoracic and thoracolumbar region, facet dislocations or fracture dislocations must be present.

BIOMECHANICAL FACTORS

Effects of Thoracic Spine Stiffness on Clinical Stability

There are two mechanisms through which the ribs tend to increase the stability of the thoracic spine. The first involves the articulation of the head of the rib to the articular facets of adjacent vertebral bodies (Fig. 5-45). The second is related to the presence of the entire thoracic cage. The thoracic cage effectively increases the transverse (x,z plane) dimensions of the spine structure. This increases the moment of inertia, resulting in added resistance to bending in the sagittal and frontal planes, as well as resistance to axial rotation (Fig. 5-46). This has been well supported by studies of computer simulation of human spine models. The greatest increase in stiffness, as measured at T1 with respect to the pelvis, was 132% during the motion of extension. The percentage increase in lateral bending was 45%, and for flexion and axial rotation the increase was approximately 31%.[5]

Biomechanical Analysis of Clinical Stability

Biomechanical studies have shown that the range of motion between thoracic vertebrae is altered by removal of the posterior elements.[161] Flexion, extension, and axial rotation were compared in individual thoracic spine FSUs before and after removal of the posterior elements. There was a statistically significant increase in the degree of flexion, extension, and axial rotation following removal. These observations suggest that when the structures can no longer function as a result of surgery, trauma, or tumor, abnormal movement occurs. Such excessive displacement could be injurious to the cord.

We have completed some biomechanical investi-

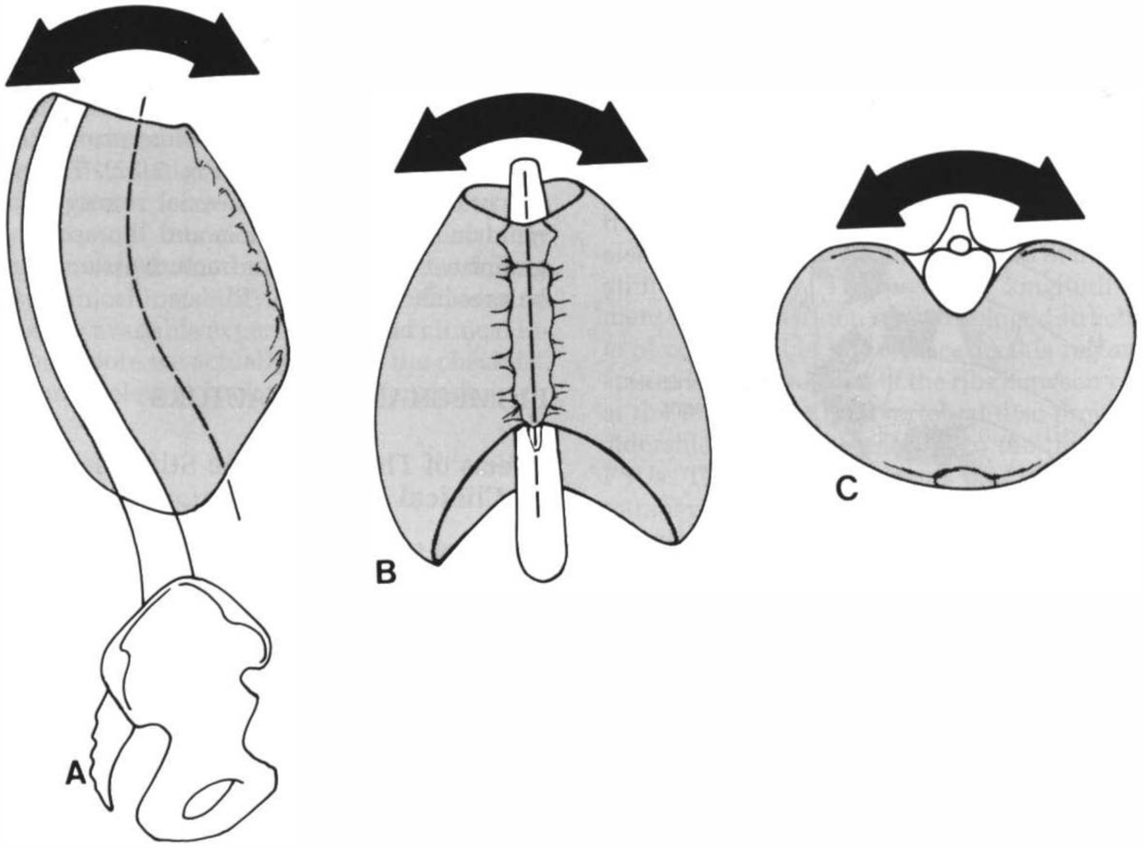

FIGURE 5-46 The thoracic cage effectively increases the transverse dimensions of the spine structure. This in turn increases its moment of inertia and, therefore, its stiffness and strength in all modes of rotation: (A) bending in sagittal plane (flexion/extension); (B) bending in frontal plane (lateral bending); and (C) axial rotation.

gations on the thoracic spine very similar to those described for the cervical region. In this study, thoracic spine FSUs with their rib articulations intact are examined. Flexion and extension are simulated with incremental loads up to 50% of body weight. Anatomic components are then cut from front to back and vice versa; the sequence is shown in Figure 5-44. Flexion or extension was simulated in several FSUs for each loading modality. The results suggest the following: with all posterior elements cut, the segment remains stable in flexion until the costovertebral articulation is destroyed; all anterior ligaments plus at least one posterior component must be destroyed to cause failure in extension; with the FSU loaded to simulate extension, stability can be maintained with just the anterior longitudinal ligament intact; and with the FSU loaded to simulate flexion, stability can be maintained with just the posterior longitudinal ligament and the other anterior elements intact. The maximum physiologic sagittal plane translation was 2.5 mm, and the maximum sagittal plane rotation was 5°.[66,164]

CLINICAL CONSIDERATIONS

Clinical Stability of Different Classifications and Fracture Types

Nicoll suggested that wedge-shaped fractures are unstable when there is rupture of the posterior interspinous ligaments.[106] There is an interesting clinical biomechanical concept that relates to the problem of instability, particularly in the naturally kyphotic thoracic spine. The concept is as follows. The greater the wedge of the vertebral body fracture, the greater the moment arm, and thus the greater the bending moment, which tends to produce additional kyphotic deformity and pressure on the spinal cord, particularly if there are disc or bone fragments in the canal. This concept is presented in Figure 5-47, and a more detailed biomechanical analysis appears in Chapter 3. Epidemiologic review of a large number of *fractures* suggests that the wedge fractures are generally stable because there is little risk of neurologic damage.[130]

The classification most frequently referred to in the literature is that of Holdsworth.

Holdsworth expressed the view that the rotational fracture dislocation occurs only at the thoracolumbar junction and in the lumbar spine and is the most unstable of all vertebral injuries; the cord and roots are in grave danger (Fig. 5-48). He states that 95% of all *paraplegias* at the thoracolumbar level have this rotational fracture dislocation.[73]

Although these classifications categorize various aspects of the problem, none of them deals comprehensively with all aspects. The definition of clinical stability with which we are concerned includes subsequent neurologic deficit and subsequent deformity and pain both short-term and long-term. There are other definitions and biomechanical rationales for defining and evaluating instability. These are presented in the lumbar portion of this chapter. The basic problems are identifying which injuries need to be treated to avoid instability and what that treatment should be.

HOLDSWORTH'S CLASSIFICATION OF STABLE AND UNSTABLE FRACTURES AND FRACTURE DISLOCATIONS

Stable	***Unstable***
Wedge compression fracture	Dislocations Extension fracture dislocations
Burst and compression fracture	Rotational fracture dislocations

(Holdsworth, F. W.: Fractures, dislocations and fracture dislocations of the spine. J. Bone Joint Surg., 45B:6, 1963.)

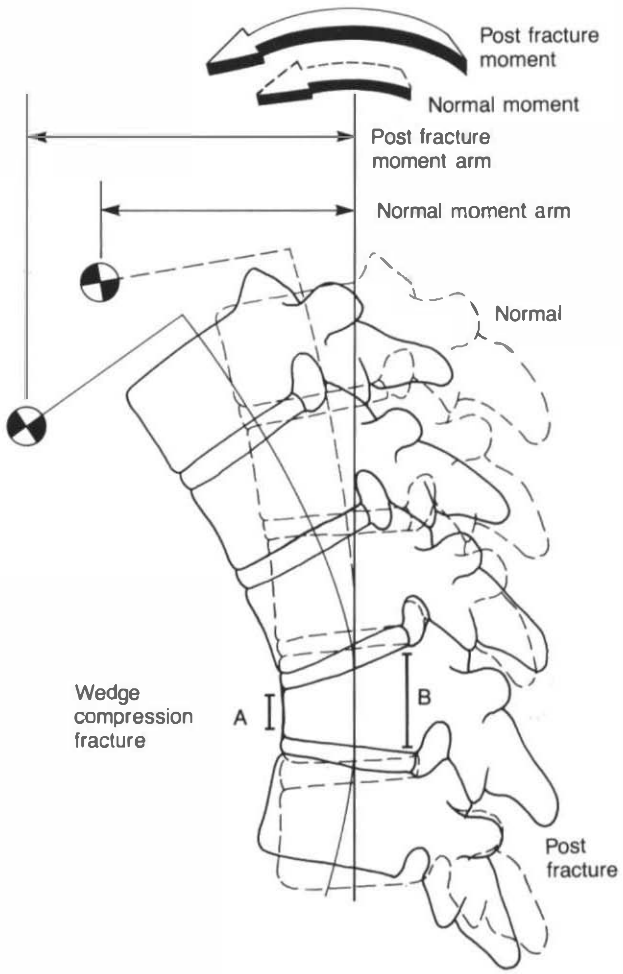

FIGURE 5-47 The "dashed" spine represents the normal thoracic spine with erect bending moment and center of gravity indicated. The moment arm is also shown. The "solid line" spine shows the injured spine with the wedge compression fracture, the increased moment arm, and increased moment as well as kyphotic deformity. The greater the wedge—that is, the greater the difference in the height of A and B—the greater the bending moment and the greater the deformity and instability.

Structural Damage and Neurologic Deficit

To what extent is structural damage to the vertebral elements correlated with neurologic deficit?

Riggins and Kraus classified vertebral fractures in all regions of the spine into seven mutually exclusive categories, depending upon the grouping of

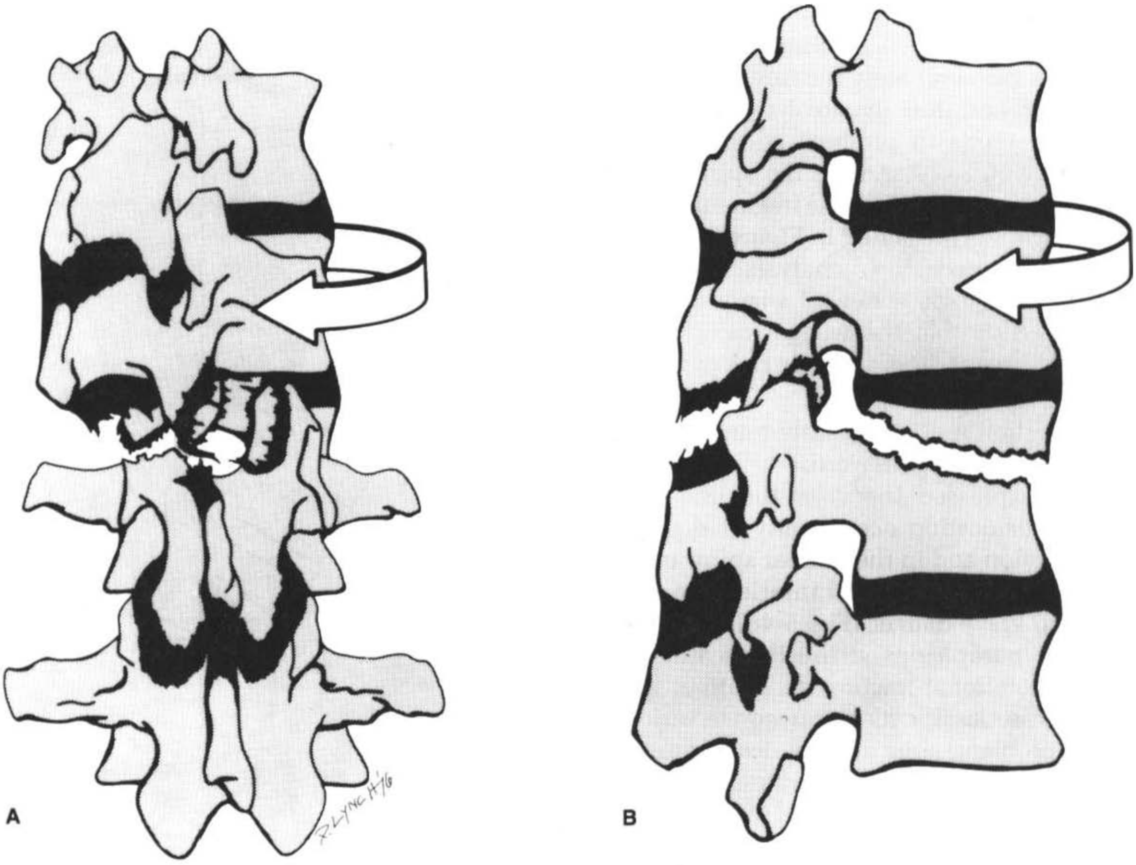

FIGURE 5-48 Rotatory fracture dislocation of Holdsworth. The primarily horizontal failure line can occur either through the annulus fibrosus attachment to the end-plate or through the vertebral body as shown here. **(A)** Posterior view of two lower vertebrae. **(B)** Lateral view.

three variables. The three variables are the condition of the vertebral body, the condition of the posterior elements (structures posterior to the posterior longitudinal ligament), and the presence or absence of dislocation (any malalignment of the vertebral column greater than 2 mm). The seven groupings, along with the percentage of each type that was associated with neurologic deficit, are given in Table 5-8.

Some relevant conclusions can be drawn from this work. Fracture dislocations have a higher incidence of associated neurologic deficits. The remaining group, which includes fractures or dislocations alone, has a lower incidence of associated neurologic deficits. Riggins and Kraus observed that a simple wedge fracture is generally not associated with neurologic damage, while the rotatory fracture dislocation described by Holdsworth is generally involved with a neural deficit.[130] The high incidence of neural problems associated with this injury is due at least in part to the very high forces required to produce it.

It seems, then, that there is a tendency for structural damage to be associated with neurologic defi-

TABLE 5–8 Association of Neurologic Deficit with Injury for All Regions of the Spine

	% With Neurologic Deficit
Dislocation only	17
Dislocation, posterior element fracture	27
Dislocation, body fracture	56
Dislocation, body and posterior element fracture	61
Posterior element fracture only	19
Body fracture only	3
Body fracture and posterior element fracture	11

(Riggins, R. S., and Kraus, J. F.: The risk of neurological damage with fractures of the vertebrae. J. Trauma, *17*:126, 1977.)

cits. However, in any given patient there may be extensive destruction and displacement of the vertebral column without neurologic deficit.[12,57,130] The converse is also true. In a northern California study it was reported that 13% of patients showed neurologic damage without recognizable evidence of spinal column injury.[130]

It can be assumed that with the advent and availability of computerized axial tomography, some of those cases included in this 13% would be discovered to have burst fractures with some compromise of spinal canal space. What kind of structural damage is likely to result in neurologic deficit? This is a crucial problem. It has been shown that sagittal plane kyphotic deformities of 30° or more are not associated with deterioration of neurologic function.[89] Roberts and Curtiss observed progressive deformity in their Type III fractures and some of their Type I fractures. In determining the subsequent deformity of the wedge compression fracture, the status of the posterior ligamentous structures is important.

Neurologic Deficit and Clinical Instability

The incidence of spinal cord damage associated with injuries to the thoracic vertebrae is approximately 10%. For the thoracolumbar region, the figure is 4%.[130]

In the thoracic and thoracolumbar spine, should one assume that all injuries that produce neurologic deficits are unstable? The rationale for such an assumption is as follows. If there is enough deformation at the time of injury to produce neural damage, there must also be sufficient structural damage to the vertebral column to render it clinically unstable. In the large majority of cases this is true. However, there can be neurologic deficit in situations in which there is either no structural damage or no recognizable damage.[61,94] The canal size in the thoracic region is relatively smaller in relation to the spinal cord size than it is in other regions. Therefore, deformation of ligaments within the elastic range can allow enough displacement to deliver a detrimental impact to the neural structures. Although the large majority of cases with neural deficit in the thoracolumbar spine are clinically unstable, it is important to keep in mind the possibility of exceptions to the rule as well as the presence of unrecognized structural damage (Fig. 5-49).

A common specific problem of this nature is that of a vertebral body fracture that is associated with canal space compromise and neurologic deficit. The problem may be analyzed in two parts, namely, management of (1) the neurologic situation and (2) the structural (stability) situation. Denis has used the three-column approach to evaluate instability.[38-40] This approach is described and illustrated in Figure 5-53. We have used the checklist systematic approach presented in Table 5-9 because it takes more factors into consideration and is designed to prevent overtreatment and to protect against neurologic complications.

Willen and associates[170,171] completed experimental and clinical studies designed to evaluate the development and correction of the local problem. They experimentally created, with dynamic axial loading, compression fractures with associated canal compromise. There was increased flexion/extension as a result of the injury. They were able to reduce the fracture with a Harrington rod distraction force of 400 N (90 lbf). In the clinical studies, they were able to reduce spinal canal encroachment to a point where the remaining anteroposterior diameter canal compromise was 26%. Distraction reduction of the fragment and stabilization with posterior instrumentation and arthrodesis is a good solution in a significant number of patients when done early (within 2 weeks).

TREATMENT CONSIDERATIONS

Effects of Reduction on Prognosis

Does surgical or nonsurgical reduction of dislocations and fracture dislocations in this region affect the prognosis? In North America there has been a trend to follow the dictates of Hippocrates, who believed that fractures should be anatomically reduced. Consequently, there has been advocacy of open reduction and instrumentation using various forms of internal fixation and surgical arthrodesis.[168] Nicoll and Leidholdt made observations on the prognosis of compression fractures. Nicoll studied patients with vertebral body compression fractures who were treated without plaster. A comparison of radiographs of the fractures on admission, discharge, and several years later revealed no significant increase in deformity. He concluded that there is no basis to assume that a good anatomic result is

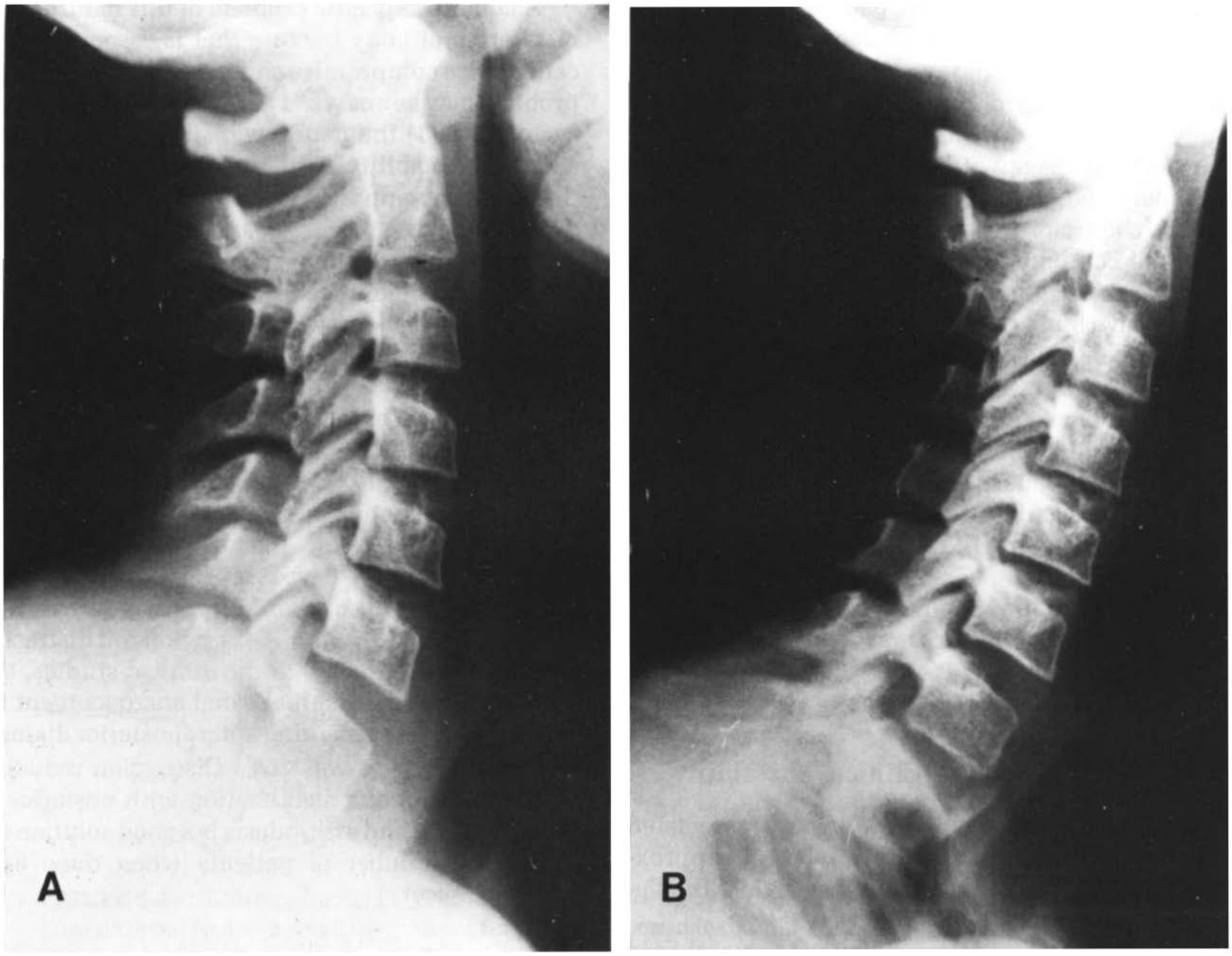

FIGURE 5-49 (A) A routine lateral cervical spine radiograph with a good view of C7 seems normal. However, the patient complained of back pain. (B) A radiograph taken with the shoulders pulled down. The fracture of the first thoracic vertebra is now obvious. These radiographs emphasize the necessity of careful and thorough examination of the cervical thoracic junction in any case of persistent undiagnosed neck or back problem following trauma to the spine. The fracture went undiagnosed for 3 months.

required for a good functional result, and he alluded to a patient who worked in a coal mine with an unreduced fracture for 30 years.[108] Leidholdt and coworkers reviewed 204 patients and concluded that sagittal plane deformity of 30° or even more *did not affect* prognosis. There was no suggestion that major deformities caused pain or decreased neurologic function.[89]

Nicoll and Leidholdt made another important observation. They noted that two paraplegic patients who had a deformity of 10° or more in the frontal plane experienced difficulty with ulceration of the ischial spine from sitting. This obviously should be considered in decisions regarding surgical reduction and fixation.

Young conducted a thorough study designed to ascertain any correlation between the severity of deformity of compression fractures of the thoracic and lumbar spine and the final clinical outcome in a group of workers. Severity of this fracture was measured by percentage of compression and loss of height of the vertebral body. Young observed no correlation of deformity with undesirable clinical outcome. There was no tendency for the original

TABLE 5–9 Checklist for the Diagnosis of Clinical Instability in the Thoracic and Thoracolumbar Spine

Element	Point Value
Anterior elements destroyed or unable to function	2
Posterior elements destroyed or unable to function	2
Disruptions of costovertebral articulations	1
Radiographic criteria*	4
1. Sagittal plane displacement > 2.5 mm (2 pts)	
2. Relative sagittal plane angulation > 5° (2 pts)	
Spinal cord or cauda equina damage	2
Dangerous loading anticipated	1
Total of 5 or more = unstable	

*See Figures 5-35 and 5-36. Measurement techniques are the same as for the cervical spine, except for Pavlov's ratio.

deformity to increase. Unlike the observations of Nicoll, who found that patients who were able to return to work in the coal mines showed spontaneous fusion,[106] Young found that spontaneous fusion occurred with the same frequency among patients with satisfactory and poor outcomes.[174]

Roberts and Curtiss, on the contrary, included in their report two patients who showed loss of recovery associated with progressive deformity of wedge compression fractures. In addition, they pointed out the *undesirability* of the use of the circle electric bed in the management of patients with severe thoracic and thoracolumbar spine injuries. They observed that progression of deformity may be associated with loading to which the spine is subjected during turning if the patient is in a bed that goes beyond the upright position.[132]

Most of the evidence suggests that the presence of deformity itself is not associated with a poor prognosis. The evidence presented here raises questions about the desirability of employing vigorous methods *purely* for the purpose of correcting sagittal plane deformity.

Operative and Nonoperative Reduction

Lewis and McKibbin conducted a retrospective comparison of conservative versus operative treatment in a series of patients with paraplegic thoracolumbar spine injuries. The following cogent quotation is from their work. "Support exists for both schools of thought and is sometimes expressed with great conviction; it is unlikely, however, that the relative merit of the two methods will ever be adequately appreciated unless they can be directly compared in a series of unselected cases.''[90] Lewis and McKibbin suggested that although surgery may offer little benefit with regard to the prognosis of neurologic deficit, it tended to help prevent pain and deformity.[90]

The significance of the status of the posterior ligaments in the probability of subsequent angulation is presented in Figure 5-50. The probability of posterior ligament disruption is greater in the wedged fracture. This is due to the fact that both the wedge and the posterior ligament disruption are likely to be the result of a significant flexion component in the injury mechanism.

Burke and Murry also compared two groups of patients, one treated conservatively and the other treated surgically. They found that 38% of the surgically treated and 35% of the conservatively treated patients showed neurologic improvement. These figures demonstrate no significant improvement in prognosis with surgery, despite the fact that the surgical group had more *incomplete* neurologic lesions, for which the prognosis is considered to be better. Significantly, the investigators also observed that of the ten patients who had significant spine pain, eight had undergone surgery.[23] Frankel and colleagues reported that out of 400 fractures of the thoracic and thoracolumbar spine, only two became unstable with nonsurgical treatment.[57]

The data from these studies do not demonstrate the superiority of either open or closed treatment of fractures and fracture dislocations in this region of the spine.

Harrington rods and a variety of other implant systems are currently in use for the reduction, stabilization, and arthrodesis of fracture dislocations of the dorsal lumbar spine. The general rationale is that a better result is achieved with less expenditure of time and medical costs. This trend suggests that operative management of fracture dislocations may be preferable. However, additional experience, careful scrutiny, and some well-designed controlled studies, including cost analysis, will be necessary to make a definitive conclusion.

The Role of Traction and Manipulation

Chahal reported on a group of seven patients that he had treated. He reported excellent results in those patients treated with the use of a pelvic girdle,

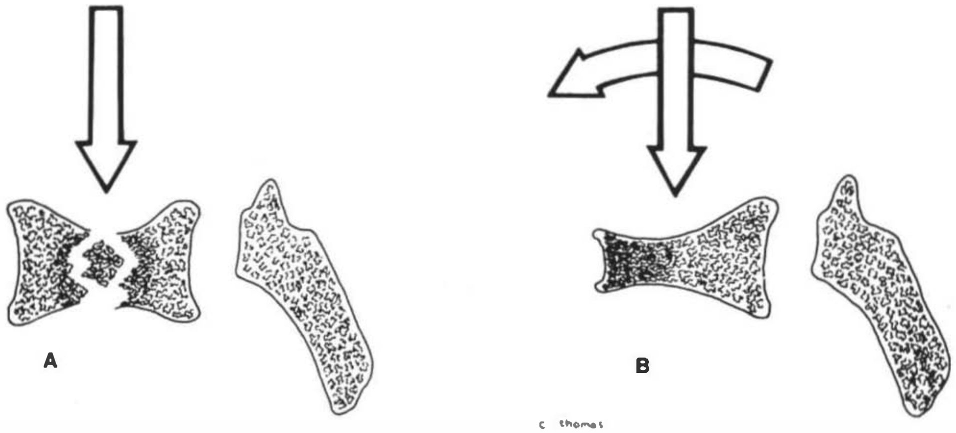

FIGURE 5-50 Probable correlation between the mechanism of injury and configuration of vertebral body compression fracture. (A) Pure compression loading gives biconcave and comminuted type pattern. (B) When there is a significant bending moment, there is more of a wedgelike configuration. This type is more prone to subsequent angulation than that shown in (A). In reality and in the experimental setting it is rare that a pure vertical compressive force is applied. There is generally at least some small element of a flexion or extension bending moment.

through which he applied 40–45 lb of traction.[29] With regard to manipulations, Böhler reports a technique in which patients are suspended prone, with arms and legs on two separate chains, in order to achieve reduction through gravity, and a plaster cast is applied in the corrected position.[16]

The Role of Laminectomies and Anterior Decompressions

Laminectomies

This highly controversial procedure was first introduced by Paul of Aegina between 625 and 690 A.D. The use of this operation has, unfortunately, preceded and superseded the critical evaluation of its advantages and disadvantages. Laminectomies may contribute to clinical instability through the removal of supporting structures of the spinal column.[17,57,63,149] The basic biomechanics of the role of the posterior elements in the support of the spine are demonstrated in Figure 5-51. The procedure itself may be associated with additional neurologic deterioration.[25] In addition to its well-documented attendant risks, the procedure itself is often not helpful. The posterior decompression does little or nothing to relieve anterior pressure (see Fig. 8-1).

Benassy and colleagues reviewed 600 cases and showed that laminectomy was not helpful in most thoracic and lumbar spine injuries.[14] Morgan and associates[100] compared a total of 198 injured spinal cords treated with and without laminectomy. They found that delayed laminectomy (more than 48 hours postinjury) produces an increased incidence of poor results compared with treatment by early laminectomy. Their basic conclusion was that the operation is indicated only when required for debridement of a compound wound or reduction of an otherwise irreducible fracture or dislocation. It is distressing and disappointing to observe how frequently this virtually useless and damaging procedure is inappropriately employed.

Anterior Decompression

Tencer and associates in 1985 reported some cogent biomechanical studies of the thoracolumbar spine in human specimens. The excellent though not definitive biomechanical studies support the idea, at least in the thoracolumbar spine, that laminectomy is not an effective technique for removing anterior pressure on the spinal cord, at least up to a 35% occlusion of the canal.[152] This work, which must be viewed and evaluated in context, is discussed further in Chapter 8.

With the development of anterior spinal surgery, there has been an increasing interest in anterior decompression through partial or total vertebral body resection. The basic biomechanics of the role of the anterior elements in the support of the spine are demonstrated in Figure 5-51. Proponents argue that when the block is anterior, posterior decompression

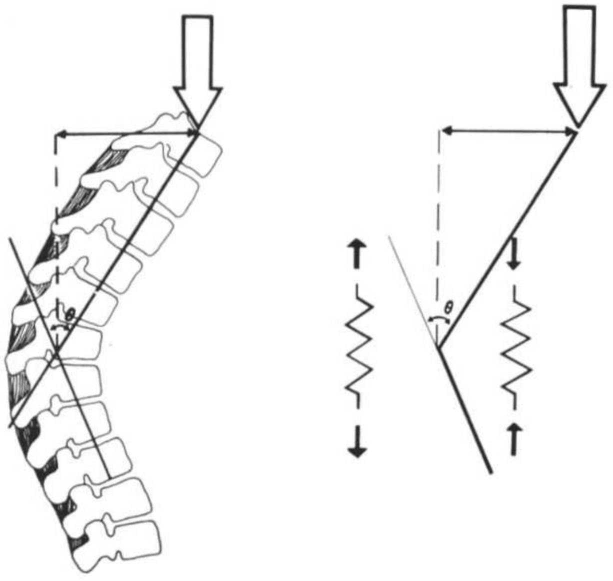

FIGURE 5-51 This diagram is designed to illustrate several points about clinical instability and kyphotic deformity: The posterior elements contribute to stability primarily by resisting tensile loads; the anterior elements function primarily by resisting compressive loads. The greater the load applied, the greater is Cobb's angle (θ); as Cobb's angle increases, there is a greater moment arm, tending to cause kyphotic deformity.

is of little value. In addition, significant vascular compromise of the cord can readily occur as a result of anterior encroachment into the spinal canal. Anterior decompression of thoracic and thoracolumbar fractures has been recommended by Whitesides and Shah. They advocate posterior Harrington instrumentation for reduction, followed by anterior vertebral body resection, with iliac or rib bone graft replacement. They reported experience with five cases.[166] This treatment rationale appears to have some advantages.

In recent years there has been a trend toward more surgical decompression followed by the use of internal fixation. These complex topics are discussed in Chapter 8.

Evaluation of Popular Arguments for Surgical Treatment

Arthrodesis, Pain, and Deformity

Evidence of functional, painless spines after trauma without surgical arthrodesis has been presented. Some of these spines showed spontaneous fusions; others did not. The study of Burke and Murry reported a higher incidence of spine pain in patients who had undergone surgery than in those who had not.[23] Other studies have shown little or no progression of deformity without fusion, while Roberts and Curtiss noted that an increase did occur in some patients in the absence of surgical stabilization.[132]

Effects on Nursing Care and Hospitalization Time

There have been few studies designed to evaluate either of these considerations in the thoracic or the thoracolumbar spine. It appears *a priori* that surgical stabilization might be expected to reduce nursing care and hospitalization time. However, an accurate evaluation involves a number of complex considerations. For example, the intensive nursing care involved in the preoperative, operative, and postoperative periods is to be balanced against the less intense and possibly more prolonged nonoperative management. Moreover, nursing care of the complications associated with treatment methods should be studied. Comparative requirements of care by other members of the team, especially the physical therapist, also seem to merit consideration. There remains the important consideration of hospitalization. We believe that it is very worthwhile to have specific, active, nonoperative programs in conjunction with surgical management.

The question of hospitalization time has become of paramount importance in the United States in the mid-1980s. Reimbursement is actually related to length of hospital stay. This is becoming a part of the consideration in the decision to operate or treat conservatively.

Effects on Overall Prognosis

This is an even more complicated issue. It brings into consideration cosmesis, psychologic well-being, attitudes toward rehabilitation, overall health, productivity, and activity levels achieved by the therapeutic programs under evaluation.

Obviously, all this information would be extremely expensive to obtain. Nevertheless, these considerations are presented to point out the need for more objective information and to show that it is not meaningful to assume certain advantages of operative intervention as if they were proven facts. We also wish to emphasize that a lack of proof does not necessarily invalidate indications for operative treatment in thoracic and thoracolumbar spine trauma.

RECOMMENDED EVALUATION SYSTEM

The Checklist

We have presented here an objective system that allows proper identification and treatment of the clinically stable and unstable conditions in the thoracic and thoracolumbar spine (see Table 5-9, p. 335).

We suggest that injuries in this region be evaluated using CT scans in addition to the regular anteroposterior and lateral radiographs. It is appropriate to emphasize here the importance of and difficulty in obtaining a good lateral radiograph of the C6–T1 area of the spine when evaluating a patient with spinal trauma. This issue has been well presented by Lauritzen, who recommends the use of lateral tomograms.[88] The swimmers view can also provide a satisfactory lateral radiograph of this region. Figure 5-49 stresses the importance of radiographs that satisfactorily show the vertebrae in this region.

Anterior Elements Destroyed or Unable to Function

The results of our laboratory studies[110] suggest that with all the anterior elements cut, the loaded thoracic and thoracolumbar spine in extension is either unstable or on the brink of instability. In the rotatory fracture dislocation, well known from clinical experience to be unstable, there is disruption of the anterior elements through the vertebral body, the annulus, or some combination of the two. With total or partial vertebral body resection for decompression, these structures may also be lost or compromised.

There are some other important characteristics of the anterior elements worthy of consideration that relate to the status of the vertebral body. With regard to configuration, the sharply wedged compression fracture shown in Figure 5-50 is more likely to be associated with a clinically unstable situation than is the less severely wedged fracture with a more central, vertically directed major injuring vector (see Chapter 4). There are several negative factors that are likely to be associated with the wedged configuration of the vertebral body fracture. Aseptic necrosis, which has been observed following vertebral fracture,[11] is more likely to occur in a highly *impacted*, severely wedged fracture, resulting in the loss of vertebral blood supply. When an injured vertebra undergoes aseptic necrosis, it can lose its ability to withstand compressive loads, especially during the stage of creeping substitution. The angulated compression fracture is more likely to be associated with progression of deformity. This is due not only to more efficient loading through a greater moment arm, but also to a possible collapse from aseptic necrosis. Although Leidholdt's clinical findings are reassuring,[89] from basic mechanics it is apparent that with increasing angulation, greater moments are applied per unit force (Fig. 5-47). Finally, the sharply angled compression fracture is more likely to be associated with significant posterior element disruption. The practical significance of this discussion involves biomechanical theory, which suggests that in the evaluation of the anterior elements, excessive wedging (approximately 50% or more of the estimated original anterior height) is suggestive of clinical instability.[G]

In the evaluation of the status of the anterior elements, special attention should be paid to the articulations of the ribs and vertebral bodies in laminographic studies. These articulations, through the linkages of vertebra, ligaments, and rib, provide considerable stability to the FSUs.

Posterior Elements Destroyed or Unable to Function

Key factors in making this determination are the physical examination and the evaluation of the position of the spinous processes on the anteroposterior radiograph. When there is extensive destruction of the posterior elements, there may be localized swelling, tenderness, edema, and a palpable defect under the skin. Wide separation of the spinous processes may be discernible. The anteroposterior radiograph shows wide separation of the spinous processes at the level involved. If there is a Holdsworth rotatory dislocation, there will be an offset of the spinous processes, showing axial rotation at the level of the injury. More subtle fractures, subluxations, and dislocations of the posterior elements are seen on the usual lateral films or laminagrams.

The results of our experimental studies suggest that with all the posterior elements destroyed or unable to function, there is likely to be enough abnormal displacement to cause additional neural damage. Kinematic studies reported on page 105 show that there can be abnormal movement in flexion, extension, lateral bending, and especially axial rotation when these structures are rendered functionless (see Fig 2-22). All of these abnormal motions are potentially injurious to the cord. With the normal kyphosis of the thoracic spine, the posterior elements play a significant mechanical role by withstanding tensile loading in the erect position, as well as during forward flexion. The presence of a wedged compression fracture may be a clue suggesting that some posterior ligaments are torn. Thus, the thoracic spine is probably more unstable in flexion when these structures are disrupted. The role of the posterior elements in the balance of the intact spine is shown in the theoretical model of Figure 5-51. The importance of the functional integrity of these elements in the thoracic spine is augmented by the normal kyphosis in this area.

Relative Sagittal Plane Translation

A relative sagittal plane translation of >2.5 mm is highly suggestive of thoracic spine clinical instability. This present criterion is based on experimental biomechanical studies completely analogous to those upon which the criteria suggested for the cervical spine were based.[110]

Relative Sagittal Plane Rotation

A relative sagittal plane rotation of >5° is strongly indicative of clinical instability in the thoracic spine. This criterion is also based on the previously mentioned biomechanical study.[110]

Spinal Cord or Cauda Equina Damage

One might consider giving this entity a full 5 points on the checklist. However, there are situations in which there is no recognizable structural damage and yet there is neurologic deficit. Some of these may be overlooked structural lesions, but it is also possible to have cord damage with a truly intact column. This, of course, would not be a clinically unstable situation. This entity is given a high value, but in such a manner that some other evidence of instability must also be present in order to make the diagnosis. Neural damage is also assigned a high value, because in the thoracic region, the space occupied by the cord is such that there is minimal opportunity for any abnormal displacement to occur without damaging the neural structures. Therefore, once the structural integrity has been altered enough to cause damage, the risk of subsequent damage is very high, and the patient should be protected against this.

Dangerous Loading Anticipated

This is the same as that recommended in the evaluation of the cervical spine. However, it is even more crucial in this region. Therefore, a 2 rather than a 1 is assigned to this factor in the thoracic spine. There may be relatively less space and less blood supply available to the spinal cord in this region (see Fig. 6-25). The forces applied to this region are likely to be greater, because the superincumbent weight is greater, and the operative moment arms are greater (Fig. 5-47).

RECOMMENDED MANAGEMENT

Flow Chart

The flow chart given in Figure 5-52 may be helpful as an outline of the major considerations in managing patients with disruption of the thoracic and thoracolumbar spine.

The patients may be placed in bed or on a turning frame. The latter should be used if there is significant neurologic deficit. The patient is given the necessary supportive therapy and is thoroughly evaluated clinically. Within the first 24 hours, a decision about the indications for operative decompression is made and carried out as soon as the patient's condition permits. Decompression is indicated when there is an incomplete neurologic deficit with radiographic evidence of spinal cord compression. The patient, with or without decompression, is then evaluated according to the clinical stability checklist.

Treatment of the Clinically Stable Condition

Patients who have been evaluated and determined to be clinically stable are treated with 1 to 2 weeks of bed rest[106] and an appropriate orthosis. They are then followed by radiographic and clinical evaluation according to the schedule on page 327. Once the patient can ambulate, physical therapy may be instituted. The patient is given exercises to strengthen the spinal and abdominal muscles.

Treatment of the Clinically Unstable Condition

There is no definitive study that convincingly shows the superiority of either operative or nonoperative therapy. Therefore, based on currently available information, we suggest that both approaches are justified. There are, however, certain situations in which we strongly favor operative stabilization of an un-

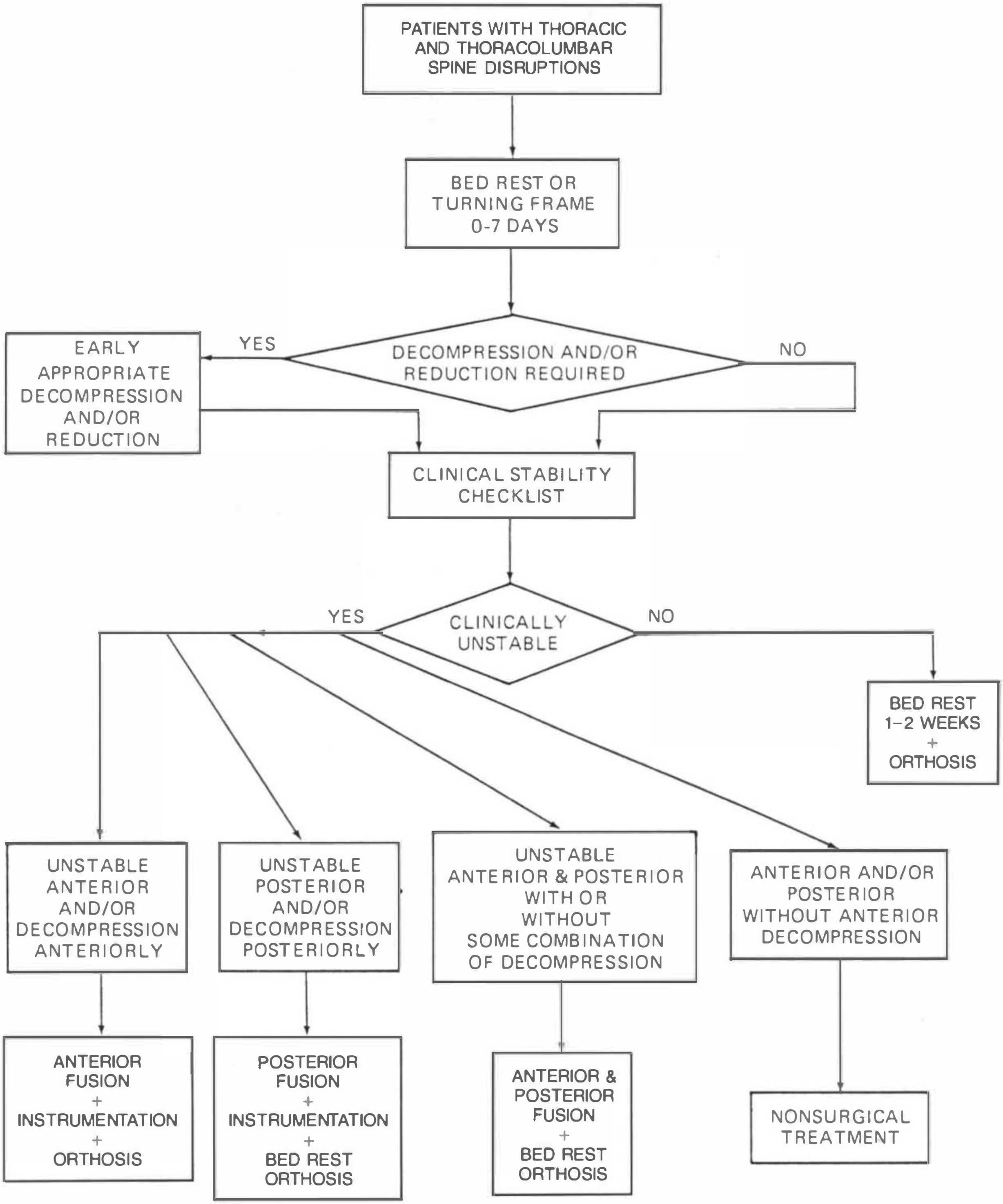

FIGURE 5-52 A recommended flow diagram for the management of patients with disruptions of the thoracic and thoracolumbar spine. The extent of bed rest, the type of orthosis, and the treatment time for each depends upon the surgeon's judgment about the immediate postoperative stability of the surgical construction. All treatment regimens include occupational and physical therapy, as tolerated, and follow-up schedule.

stable spine. Surgical stabilization is required following anterior decompression with removal of a complete vertebral body. Other indications for stabilizing operations include situations in which the patient's mental or physiologic status is such that the required bed rest would obviously be detrimental or impossible.

Several types of clinically unstable situations are included in Figure 5-52. As shown by the grouping, clinical instability may be present in several situations. It may be the result of anterior and/or posterior structural and pathological damage, as well as the result of anterior and/or posterior surgical decompression. Surgical considerations are also presented in some detail in Chapter 8.

Clinically Unstable Anteriorly, With or Without Anterior Decompression

We suggest anterior fusion followed by 1–2 weeks of bed rest and rehabilitation as tolerated. A plaster jacket with chin and occipital support or a Milwaukee brace is recommended. If there is instability at T6 or above, chin support is important. The individual is ambulatory in the orthosis for 12–24 weeks. Use of some of the more rigid implants may obviate the need for an orthosis. The patient is then managed by the guidelines suggested in the follow-up schedule.

Clinically Unstable Posteriorly With or Without Posterior Decompression

When instability is primarily posterior due to disruption of the posterior elements from natural or iatrogenic disease, posterior fusion with instrumentation is indicated. Appropriate wiring or one of the newer instrumentations is a feasible approach. We suggest that the patient be kept in bed no longer than is needed for pain control and to gain strength and balance for ambulation. Following this, an orthosis is prescribed as needed for comfort and/or stability. The rehabilitation and follow-up schedule is the same as that described on page 327.

Clinically Unstable Both Anteriorly and Posteriorly With or Without Anterior and/or Posterior Decompression

In some situations, posterior fusion alone is satisfactory, even when there is anterior and posterior instability. However, when there is a need for vertebral body removal, it is generally better to operate posteriorly first. Instrumentation or wiring can be used to provide some *immediate* postoperative stability, and posterior fusion can be applied with no additional loss of stability. Then the spine may be approached anteriorly, allowing decompression and fusion to be carried out with the assurance of some stability from posterior instrumentation. The regimen is carried out as follows. Posterior instrumentation and fusion is followed immediately or in 3–7 days by anterior decompression and/or fusion, as indicated. The postoperative regimen is the same as that for patients with posterior instability.

Pedicle fixation, spinous process wiring, lamina wiring, and other systems of rodding and fixation may provide enough stability to avoid the necessity of the anterior approach, particularly when no anterior decompression is necessary.

Clinically Unstable Anteriorly and/or Posteriorly Without Anterior Decompression

Except for clinical instability following anterior decompression by vertebral body removal, our interpretation of the present knowledge suggests that other types of instability may also be treated by nonsurgical methods. These nonsurgical methods offer several justifiable alternatives. One may elect to proceed with postural reduction followed by additional bed rest and then some orthotic device.[63] Another approach is to simply have the patient rest in bed, or, if there is significant sensory deficit, the patient may use a turning frame. After 6 weeks, the patient may begin to walk, using an orthosis for additional support.[12] With these and other nonsurgical regimens, physical and occupational therapy and appropriate exercises in bed are recommended. The patient is then given rehabilitation and activity as tolerated and is followed according to the schedule suggested on page 327.

Discussion

Several nonsurgical options have been offered here. There are a number of other regimens and combinations to be considered. Bedbrook and Edibaum, for example, suggested that possibly there should be trials of mobilization of some patients with neurologically involved spines after 3 weeks.[12] If such a series were shown to be successful, a number of the surgical "advantages" would be eliminated. The nonoperative regimens offered by us are all based on well-documented experience and sound clinical biomechanics and rehabilitation. They all seem to be appropriate, and the nonsurgical regimens selected might well be determined by some equilibrium between the individual patient's and physician's needs and preferences, as well as the local practices and available facilities.

PART 4:
THE LUMBAR AND LUMBOSACRAL SPINE (L1–S1)

The problem of clinical stability in the lumbar spine has some unique considerations related to both aspects of the definition of clinical instability. The associated neurologic deficits are relatively rare, less disabling, and patients are more likely to recover. A large epidemiologic series of all spine injuries reported that only 3% of patients with lumbar spine dislocations and fracture dislocations had neurologic deficits.[130] The second consideration is related to the phenomena of subsequent pain, deformity, disability, and the very high loads that must be borne by this region of the spine. The clinical biomechanical problem of pain and its management is discussed in Chapter 6.

ANATOMIC CONSIDERATIONS

Anterior Elements

The anterior longitudinal ligament is a well-developed structure in this region. The annulus fibrosus, which has received an enormous amount of attention in the literature, constitutes 50–70% of the total area of the intervertebral disc in the lumbar spine. As in other regions of the spine, it contributes in a major way to the clinical stability of the FSU.

Posterior Elements

The posterior longitudinal ligament is less developed than its anterior counterpart in the lumbar region. All the ligaments and the facet orientation are shown in Figure 5-53.

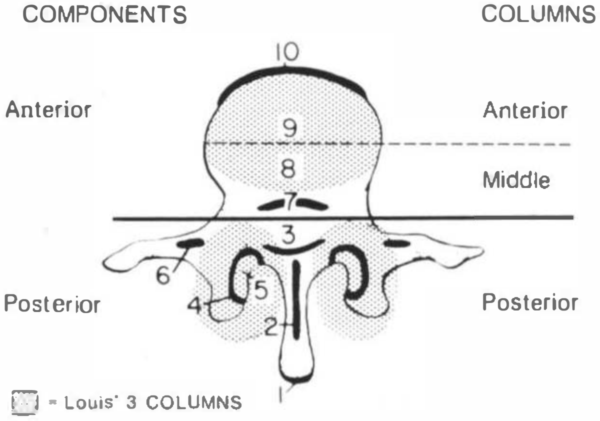

FIGURE 5-53 This schematic representation of the anatomy of a lumbar vertebra illustrates three categorizations of the columns. First, the anatomy: 1, supraspinous ligaments; 2, interspinous ligaments; 3, yellow ligament; 4, capsular ligaments; 5, the articular facet; 6, intertransverse ligament; 7, posterior longitudinal ligament; 8, annulus fibrosus, posterior; 9, annulus fibrosus, anterior; 10, anterior longitudinal ligament. This is also the sequence in which the ligaments were cut 1 to 10 or 10 to 1 in the Posner experiment.[125] The elements and columns are indicated as follows. On the left, the anterior elements and posterior elements are divided by the uninterrupted horizontal line just posterior to the posterior longitudinal ligament (7). The three columns of Denis[38] are shown to the right of the vertebra. The *anterior* column is anterior to the interrupted horizontal line, the *middle* column is between it, and the uninterrupted horizontal line and the *posterior* column are behind the uninterrupted line. Finally, Louis's[93] three-column concept is described under the stippled area of the figure (i.e., the two facet joints and the vertebral body).

The facet joints play a crucial role in the stability of the lumbar spine. Usually, a flexion rotation injury and displacement are required for dislocation to occur. The well-developed capsules of these joints play a major part in stabilizing the FSU against this type of displacement. When there is a fracture dislocation of the facet joints, there may be abnormal displacement, primarily axial rotation, and possibly lateral bending. When these displacements are observed, fracture or fracture dislocation of the facet articulations must be suspected. Figure 5-54 shows how the spatial orientation of the articular facets prevents excessive axial rotation. It is clear that these structures must fracture and/or dislocate in order to permit abnormal axial rotation. Moreover, Sullivan and Farfan showed in the laboratory that axial rotation of the lumbar spine of 30° or more caused failure of the neural arch, progressing from facet joint dis-

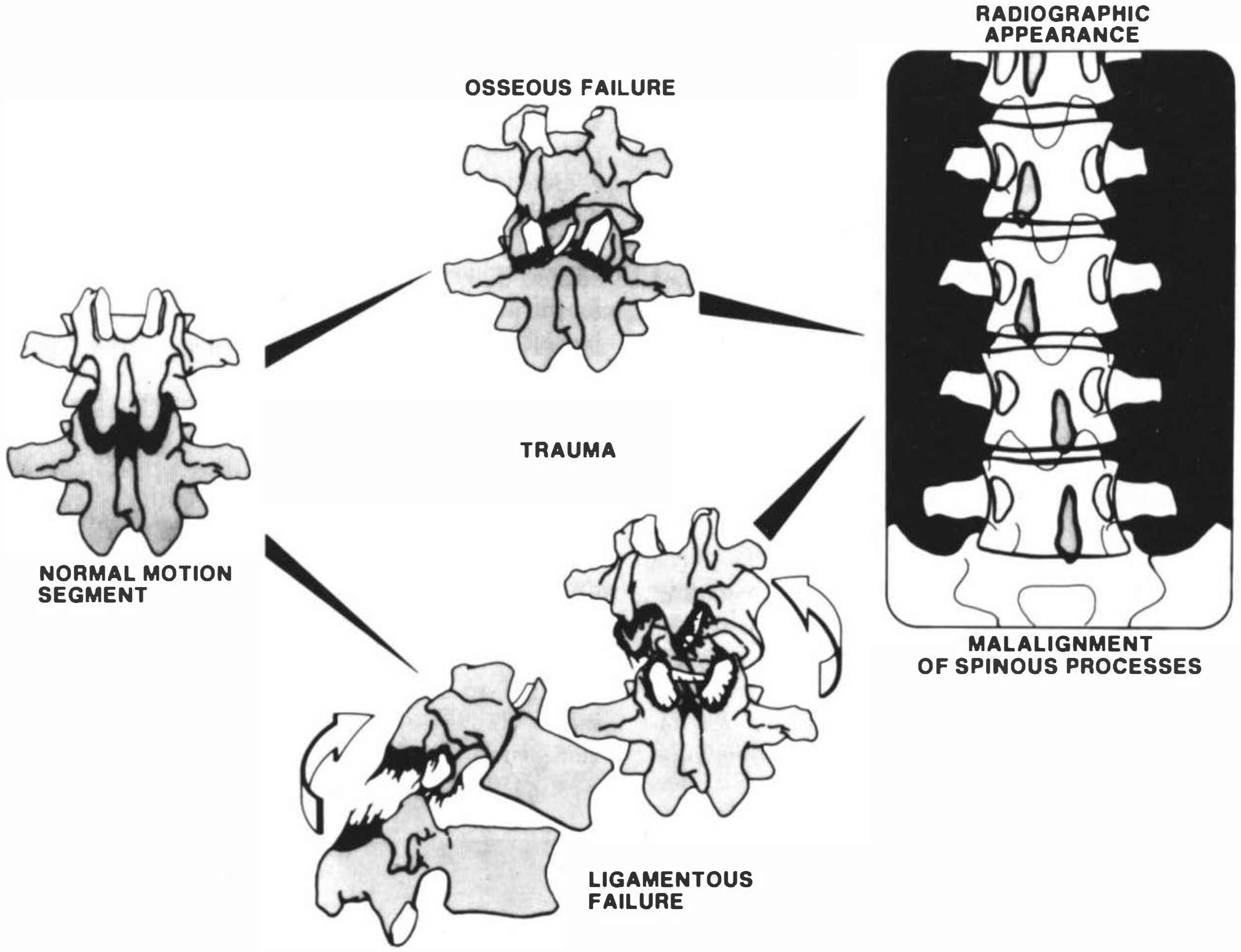

FIGURE 5-54 This diagram emphasizes that in order for abnormal displacement of the spinous processes in the lumbar spine to occur, it is necessary to have a dislocation of or a fracture in one or more facet articulations. The final radiographic view of the malaligned spinous processes may also show vertical separation of spinous processes and asymmetrical projection of the pedicles. These findings post-trauma are all suggestive of significant posterior element injury. Always look for vertical and horizontal displacement of the spinous processes when examining a post-traumatic anteroposterior x-ray of the spine.

location to fracture of the articular process.[148] It seems obvious that with the forces involved in these kinds of disruptions and dislocations, there would also be injury involving the yellow ligament. With the relatively modest posterior longitudinal ligament and the degenerating and ruptured interspinous ligaments, it is readily apparent that the dislocations and fracture dislocations of the facet articulations tend to be associated with a loss of clinical stability.

Rissanen carried out an anatomic study of 306 cadavers, which resulted in some pertinent and interesting observations. The age distribution was from fetal life to 90 years, with significant representation throughout the range. He found that 21% of those subjects over 20 years of age had rupture of one or more of their interspinous ligaments, occurring mostly at L4–L5 and L5–S1 FSUs. He also found that 75% of the subjects between ages 31 and 40 showed cavitation where the ligaments are normally located.[131] These findings suggest strongly that the interspinous ligaments can be expected to make very little or no contribution to the clinical stability of the lower lumbar spine in the adult.

In contrast, however, the supraspinous ligaments appear to play a major role in the lumbar spine. Myklebust and comrades studied ligaments individually by removing all but the ligament to be tested.[103a] The interspinous ligaments failed in the range of 95–185 N, and the supraspinous ligaments yielded in the range of 293–750 N. The strength and the mechanical advantage (distance from instantaneous axes) indicate that there are many major fractures in lumbar spine biomechanics.

The exact role of the yellow ligaments has not been determined in regard to instability. It is interesting to note that the ligament is considerably thicker (4–6 mm) in the lumbar region than at the L5–S1 region (1.5 mm).[70] This anatomic finding may in fact be related to the greater flexion/extension and axial rotation of L5–S1 in comparison with some of the more cerebrally located lumbar FSUs.

The status of the facet articulations is crucial in other ways. There is a consideration here that is similar to the observations of Beatson on the cervical spine.[10] Most probably, damage of the posterior longitudinal ligament and annulus fibrosus is associated with disruptions of the facet articulations. Such a correlation has not been demonstrated experimentally, but a study of the anatomic relationships with an evaluation of the amount of displacement involved in dislocation of the facets will support this hypothesis (Fig. 5-54).

The work of Posner and colleagues,[125] and subsequently Adams and colleagues,[2] through biomechanical component ablation studies, demonstrated that the major players in the stabilizing role for the posterior elements in the lumbar spine are the facet joints. This is an important point in all types of clinical stability evaluation.

The final consideration with regard to the posterior elements relates to the informative experimental setup that nature provides in the form of spondylolisthesis. The radiographs of a healthy 17-year-old male are shown in Figure 5-55. This well-known and well-recognized entity demonstrates a situation in which the posterior elements are rendered unable to function because of a defect in the pars interarticularis. When this situation occurs, the remaining anterior structures may allow progressive displacement. This is due to plastic deformation of the anterior structures, the two longitudinal ligaments and

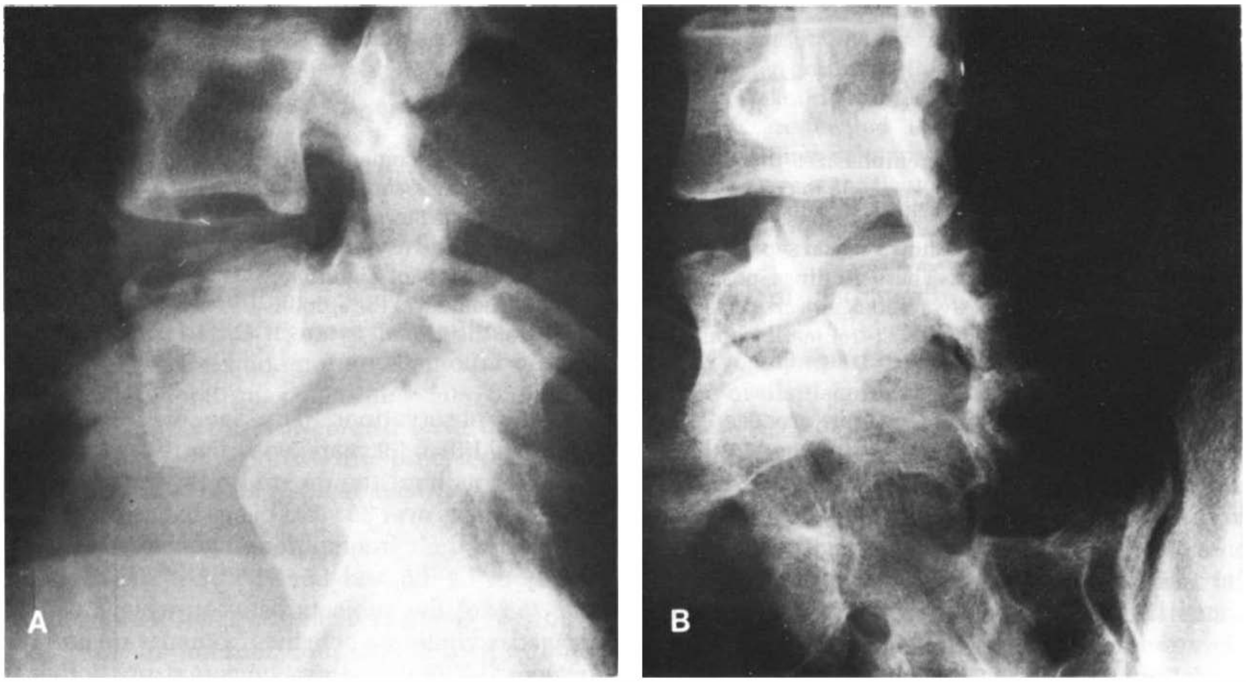

FIGURE 5-55 Radiographs of a patient with spondylolisthesis show that when all the posterior elements are unable to function, there may be abnormal displacement. **(A)** Lateral view and **(B)** oblique view, showing the "Scotty dog" sign.

the annulus fibrosus. The lateral radiograph graphically shows anterior displacement in the sagittal plane of approximately 50% of the sagittal diameter. The oblique views clearly demonstrate the "Scotty dog sign," with the defect in the pars interarticularis that allows the forward displacement. The "broken neck" of the "Scotty dog" represents the actual defect. This spondylolisthesis shows that there can be clinical instability of the lumbar spine when the posterior elements are compromised.

BIOMECHANICAL FACTORS

Mechanism of Injury in Spondylolysis

Although the exact mechanism for the development of spondylolysis is not known, there is some suggestion that it may be a fatigue fracture. This fits with clinical observations of the development of pain associated with a positive bone scan following repeated hyperextension in the gymnast. Cryon and Hutton[34] have described specific forces that could result in failure at the pars interarticularis region. Their analysis is presented in Figure 5-56.

Muscle Forces

The role of the muscles in clinical stability, as previously discussed, is difficult to evaluate. Certainly the lumbar region is well endowed with active muscles. The erector spinae, abdominal, and psoas muscles are all actively involved in maintaining the functional upright and sitting stability of the lumbar spine.[9,101,104] They also contribute to the very high loads to which the lumbar spine is subjected. In this region, these well-developed muscles and their characteristic loading patterns may render the lumbar spine less vulnerable to clinical instability. However, it must also be considered that the large and variable loads due to muscle and gravitational forces increase the likelihood that disruptions of the lumbar vertebral column will be associated with severe pain.

The Mechanics of Component Degeneration and Motion Patterns

Most of the biomechanical studies have simulated physiologic spinal motion and loads, monitored the normal, then altered various structures and remeasured the motion characteristics. A review of the more cogent studies provides useful background information for the understanding and evaluation of lumbar spine stability. This was done by Nachemson in 1985,[105] and we are presenting a review of several additional studies here.

Nagel's experiments showed that at the L1–L2 level, 20° of flexion or 10° of lateral bend occurred in specimens in which all the posterior elements and part of the annulus were removed. This is thought to represent instability. Adams and colleagues, using

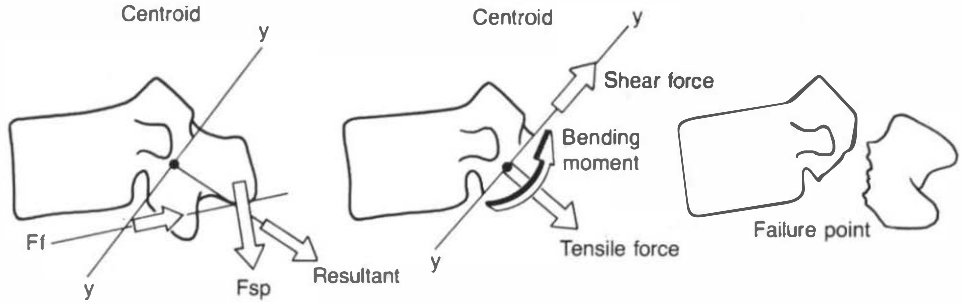

FIGURE 5-56 Analysis of forces thought to cause spondylolysis by fatigue failure in the lumbar spine. It is thought that the defects are produced by fatigue of the neural arch. *Ff* is the force acting on the inferior articular facets, and Fsp is the force acting on the spinous process. The combined forces produce a resultant force that can be resolved into a shear force, a bending moment, and a tensile force at the centroid of the cross-sectional area of the pars interarticularis. The line yy represents a plane through the narrowest region of the pars interarticularis. *(Hypothesis based on the work of Cyron, B.M., and Hutton, W.L.: Variation in the amount and distribution of cortical bone across the pars interarticularis of L5: a predisposing factor in spondylolysis? Spine, 4:163, 1979.)*

simulated physiologic loads and motions, ablated components and showed that the facet capsule is the major stabilizing structure and is capable of resisting about one-half of the full flexion force. Thus, the facet joint capsule is probably the major structure among the posterior elements in resisting flexion.[2] This role is even more prominent in view of the frequent absence of the interspinous ligaments in the lumbar spine.[131] These investigators also studied torsional loading. In this modality, it was not the facet joint capsule but the bony articular process that offered the major resistance. Only 1–2° of axial torsion were allowed. Much greater angles of torsion would be required to damage the intervertebral disc.[1]

Van Akkerveeken and associates focused on the effects of the removal of the nucleus pulposus and section of the posterior annulus fibers and the posterior longitudinal ligament on the mobility of the spine. The authors reported abnormal posterior sagittal plane translation (3 mm) and hypermobility in horizontal plane rotation (y-axis rotation).[154] Although a broad perspective is considered and a number of studies are reviewed, our recommendations are largely influenced by the work of Posner and co-workers.[125] This was a study of the lumbar and lumbosacral spine in which physiologically preloaded FSUs were studied in flexion and extension following individual component ablation. From these studies it has been possible to determine the upper limits of normal motion and the tolerance limits of the units with various components ablated. Once all the posterior elements were ablated in flexion, displacements that were significantly greater than those observed as the upper limits of normal motion were noted. Moreover, it was noted that any additional component ablation would result in failure of the FSU. This work pointed out significant differences in the biomechanics of the lumbar and lumbosacral FSUs and indicated different criteria for the evaluation of clinical stability in the two regions.

The preceding studies focused on quantity of motion. The work of Gertzbein and associates observes in a precise manner the quality of motion. Using a Moiré Fringe technique, the loci of instantaneous centers of motion were determined for several points in flexion and extension. The clusters of centers were called centrodes, and their characteristic patterns were observed.[59] Additional studies characterized centrode patterns in spines with varying degrees of disc degeneration. In those spines with degenerated discs, the centrode pattern increased significantly in length and was in a different location. The authors indicated that this test was highly sensitive in that it was positive in 94% of abnormal spines, whereas flexion/extension excessive range of motion studies were positive in 25% of the abnormal spines.[59,140] The authors show pattern changes with disc degeneration and thus appear to be defining disc degeneration as either a precursor of or a state of instability.

We must also mention here the work of Penning and associates, who studied flexion and extension films in two groups of back pain patients. One group did their bending while standing, and the other group did their bending while in the lateral decubitus position. The position of bending was not the important point of this study. The importance of this work is that the expected "abnormal" patterns of centrode distribution in those FSUs which had radiographic suggestions of disc degeneration were not present.[120]

We have employed a different definition of instability and an alternative approach to the clinical problem.

Clinical Biomechanical Studies

An important and frequently asked question is, What is the importance of discectomy as a cause of lumbar spine instability? Tibrewal and colleagues, in a prospective study with biplanar radiography, observed 15 patients. At the level above the abnormal disc in these patients there was up to a 50% decrease in flexion/extension and an increase in the coupled motions of lateral bending and axial rotation. Surgery that relieved the symptoms did not alter the range of flexion/extension but did reduce the increased coupled motions. Discectomy by fenestration and minimal resection of the lamina did not produce instability.[153] Pearcy and colleagues,[114] with three-dimensional x-ray analysis, studied patients in an attempt to document any association of back pain with abnormal spinal motion. They noted less flexion/extension than in normal controls. There have been a number of attempts to correlate abnormal quantities and qualities of motion with spine pain. For the most part, the data and information are inconsistent, and little substantive evidence has emerged to support distinct correlations.

Another important question is, What is the propensity, if any, to develop spinal instability adjacent to a fusion? Scoliosis fusions that go below L3 have a high prevalence of low back pain.[33] Single-level lumbar fusions resulted in increased motion to the FSU below and resultant disc narrowing.[127] Spinal stenosis has been reported adjacent to spinal arthrodeses.[88a]

Olerud and associates[108a] have completed an important clinical study in which they combined an old concept (immobilization test) with some new technology (percutaneous transpedicular fixation) to develop a promising clinical methodology for the evaluation and management of low back pain associated with instability. There were 18 patients with low back pain. Schanz screws 5 mm in diameter were placed through the skin and into the pedicles (Fig. 5-57). An external fixation was used to immobilize the spine. Some of the 18 patients had chronic severe back pain with sciatica. With external fixation, all but one patient experienced relief of back pain and often the sciatica. However, eight patients had severe pain following fusion. The authors indicated that they viewed this test as useful for selecting patients for surgery and for determining the levels to be fused. It was also thought to have potential for determining the success of a given fusion in eliminating motion or pain.

The Olerud study has been confirmed by another study in which predictive value of temporary external fixator is compared to that of plain radiographs, discography and facet blocks.[47b] Of the 35 patients who underwent external spinal fixation, all 35 had plain radiographs, 32 had discography and 14 had facet injection blocks. Twenty-seven of the 35 patients underwent surgery. Of the 25 who had relief of pain following temporary external fixator applica-

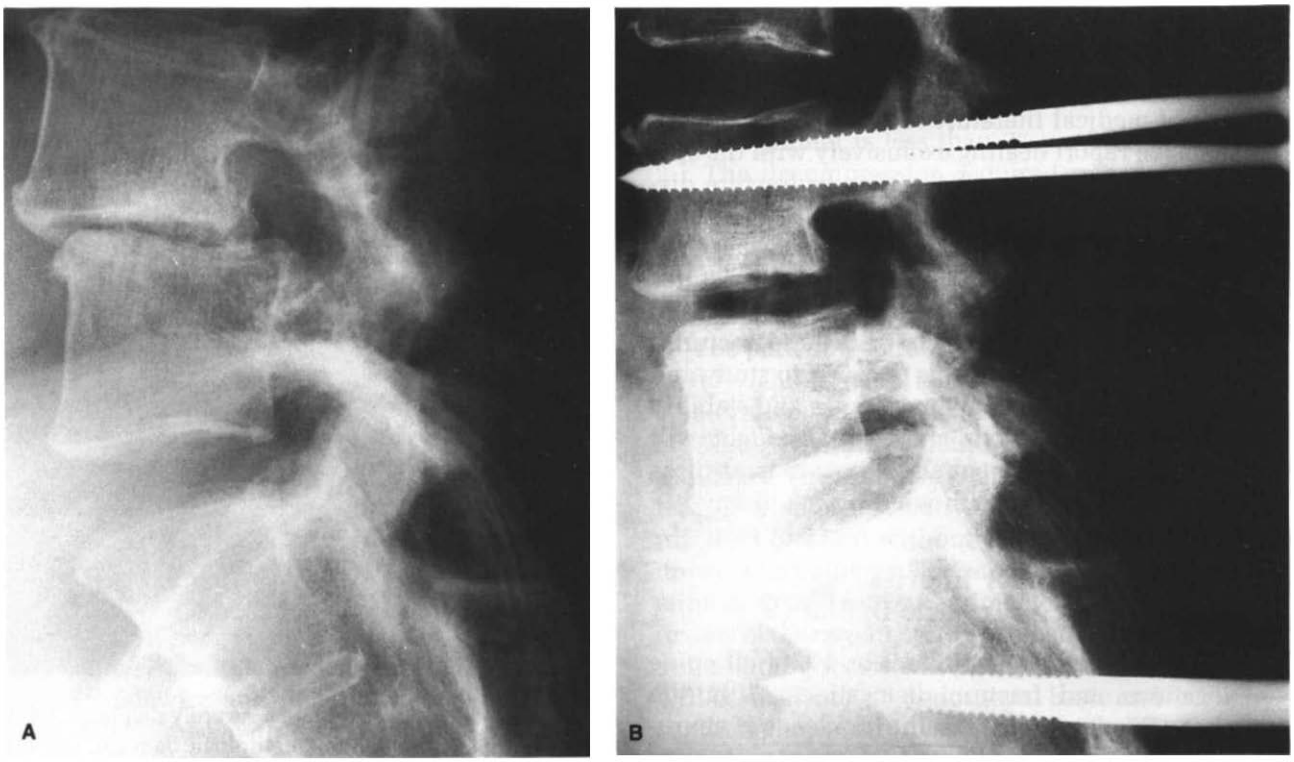

FIGURE 5-57 (A) Lateral lumbar spine radiograph of a patient with degenerative spondylolisthesis and severe low back pain. (B) Schanz screws have been placed percutaneously. The vertebrae are distracted and immobilized with an external fixator. Repeated correlation of pain with loosening of fixation and relief with immobilization suggest that the pain is due to clinical instability and can be helped by arthrodesis of the functional unit. *(Reproduced with permission from Sven Olerud, M.D.)*

tion, 17 had significant or complete relief of pain after surgery. Of the 21 patients who showed degenerated discs, only 9 had complete or significant relief of pain after surgery. Of the 15 patients who had reproduction of pain by discography, only 6 had complete or significant relief of pain following surgery. The facet block data, because of the small number of patients, was not analyzed. The authors conclude that temporary external spinal fixation is an invaluable additional test in patients considered for surgical arthrodesis.

Holmes, Brown, and associates have begun *in vivo* studies of the stiffness of FSUs at the time of surgery. The investigators observed a reduction in FSU stiffness following decompression. The reduction was in the range of 5–40%, when comparing the pre- and postoperative stiffnesses. This study fits the surgical maxim of measuring forces whenever feasible. It also provides an interesting potential for better future understanding of clinical instability.

In 1966, Kaufer and Hayes wrote the following: "Since lumbar dislocation is not uncommon and possesses special characteristics of therapeutic and prognostic significance, we were surprised that a search of medical literature failed to reveal a comprehensive report dealing exclusively with the special problem of lumbar dislocation or fracture dislocation."[84] More than 20 years later, this statement is far from correct. During the 1980s, a large number of articles on the topic of lumbar spine fractures appeared. In addition, there were numerous internal fixation devices with which to treat these fractures. The reader is admired for his decision to study and understand the clinical biomechanics and stability of those fractures. Your study and thought will greatly enhance judgment about how to treat these various fractures and with what device, if any.

Incidence of Lumbar Injuries and Associated Neurologic Deficit

"Lumbar segments constitute 3 or 4% of all spine dislocations and fracture dislocations."[84] In the study by Riggins and Kraus, the incidence of neurologic damage associated with lumbar spine injuries was just 3%.[130] Kaufer and Hayes, however, reported an incidence of 53% in their series of 21 cases.[84] Others have reported 60–70% with neurologic deficit. These figures vary widely, depending on whether or not T12–L1 is included separately or with the lumbar spine.

Structural Damage and Neurologic Deficit

Riggins and Kraus noted that there was a greater incidence of neurologic deficit in fracture dislocations, but here, too, there was no consistent correlation.[130] The radiograph in Figure 5-58 shows quite dramatically the discrepancy between structural damage and neurologic deficit. This extreme fracture dislocation had no neurologic damage associated with it.

Neurologic Deficit and Clinical Instability

A so-called burst fracture may have a neurologic deficit because a fragment of bone is displaced into the neural canal. However, this does not in itself render the FSU unstable. This is an example of a

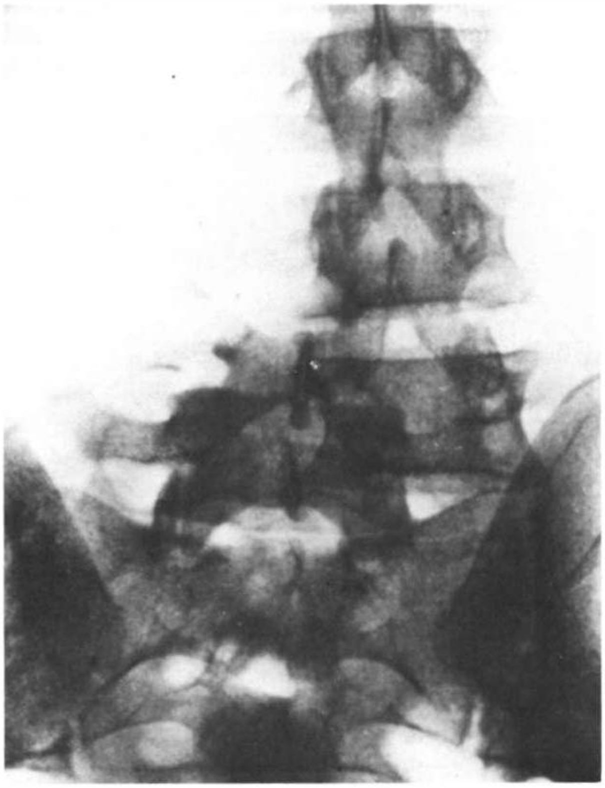

FIGURE 5-58 Radiograph of a severely displaced L4–L5 fracture dislocation without neurologic damage. This is an astonishing example of the occasional gross lack of correlation between structural damage and neurologic deficit. This disassociation is more likely to occur in the lumbar region than in other regions of the spine. (Steinger, K. K.: *Fracture-dislocation of the thoracolumbar spine with special reference to reduction by open and closed operations. J. Bone Joint Surg.*, 29:107, 1949.)

situation in which the anterior, middle, and posterior three-column concept may confuse the need for fragment reduction and/or neural decompression with a definition of clinical instability. Neural decompression and clinical instability are separate issues that may on occasion be related in that neural decompression in some circumstances may cause or contribute to clinical instability.

Classification System

Kaufer and Hayes classified fracture dislocations of the lumbar spine and appropriately recognized the disruption of the facet joints as an indication of more severe injury.[84]

SPINAL STENOSIS AND INSTABILITY

This brings us to another critical issue that involves the relationship between spinal stenosis, instability, and arthrodesis. Here the question is, In spinal stenosis surgery, which anatomic components can be ablated without rendering the spine unstable and in need of an arthrodesis?

The previously described biomechanics experiments by Posner and colleagues are helpful, and, in addition, some clinical studies are presented. Hazlett and Kinnard[67] completed a 2–5-year follow-up of 33 patients and noted no instability following discectomy and facet joint removal. They noted that multiple levels (presumably more than two) may have a different outcome. Johnsson and coworkers,[80] in their study of 45 patients, determined that postoperative instability as determined by slip was correlated with *degenerative spondylolisthesis but did not affect outcome when it occurred.* However, the patients who had acquired spinal stenosis and postoperative slip (instability) did tend to have a poor result. This study also showed that surgical outcome for these patients was not different if they had facetectomy in addition to laminectomy. However, Johnsson and co-workers showed a *tendency* toward less slipping after a more limited laminectomy. Sienkiewicz and Flatley,[142a] in a study of a group of 8 women with an average age of 62, observed a distinct propensity to develop a L4 or L5 spondylolisthesis following laminectomy associated with partial facetectomy or transection of the pars interarticularis. This observation should stimulate further study and alert us to the possibility of a propensity for the L4–L5 level to be clinically unstable in women. With such a paucity of studies in the literature, perhaps one should not be unduly influenced by this well-done study, particularly when it goes contrary to current teaching and practice.

Here are some suggestions for the management of spinal stenosis patients, with particular emphasis on the issue of instability. First, complete an adequate decompression of the spinal canal and nerve roots. This should be done conservatively. Take only that portion of the annulus that is herniated into the canal with a small additional margin. *In vitro* studies by Goel and colleagues showed that in the presence of laminectomy, a subtotal discectomy caused significantly less motion than did a total discectomy.[60] Next, remove only that portion of the facet articulation that is necessary for nerve root decompression. We suggest this conservative approach; moreover, it is reassuring to report that a clinical study of 33 patients with a 2–5-year follow-up showed that facet joint ablation in conjunction with discectomy at one level did not result in instability.[67]

An operating table-side decision to arthrodese the altered FSUs is suggested in the following situation.

1. The patient is less than 75 years of age
2. The decompression required removal of more than 50% of the total facet joints at one level
3. The decompression required the removal of a significant portion (30–40%) of the annulus fibrosus

The patient is followed and monitored postoperatively. The issue of instability can then be evaluated through the clinical stability checklist (see Table 5-10). If there are 5 points or more, spine fusion is indicated. The rationale is that the few patients who may develop postoperative instability can be recognized and treated without problems and can avoid unnecessary surgery. These guidelines are based on clinical experience, experimental studies,[125] and a review of the cogent clinical and biomechanics literature. More precise and totally accurate guidelines will be available after additional basic and carefully conducted clinical investigations.

SPONDYLOLYSIS, SPONDYLOLISTHESIS, AND INSTABILITY

When there is a defect in the pars interarticularis, the role of the posterior elements in stabilizing the FSU is significantly reduced. The soft tissue in the defect itself and in the interconnecting ligaments of the FSU may undergo plastic deformation. By physiologically stressing the spine (standing, hanging, and posterior loading with gravity in the sitting position), Kessen and colleagues[86] were able to show a statistically significant difference in the anteroposterior translation between a group of 15 normals and a group of 16 patients with spondylolysis. When there is a more readily perceptible displacement, there is spondylolisthesis and instability. The checklist (see Table 5-10) is useful in evaluating the clinical instability.

When the criteria for the diagnosis of instability are met and nonoperative treatment relieves neither pain nor neurologic deficit, arthrodesis is indicated. If the slip is extensive, reduction prior to fusion may be indicated. If there is a nonprogressive displacement without neurologic deficit in which pain is absent or not severe, the patient can be followed without surgery. There are several radiologic measurements that are used to characterize, monitor, and predict spondylolisthesis. The most cogent are reviewed in the following section.

Penning and Blickman studied *in vivo* sagittal plane motion using lateral x-rays of the lumbar spine in 24 patients with spondylolisthesis. These findings, though interesting, do not simplify the issue. The cogent findings were as follows. The vertebra involved in the spondylolisthesis showed a wide distribution of axes of movement and hypermobility, but no abnormal translatory movement. There was increased motion of the vertebra *above* the one involved in spondylolisthesis.[119]

It is of theoretical interest to note the variety of biomechanical observations that have been made in association with spondylolisthesis. Both increased and decreased motion have been observed for the vertebra that involves the listhesis. We note also that the IARs described by the involved vertebra as it moves in the sagittal plane have demonstrated a greater scatter. The important clinically relevant theoretical question is, Are these mechanical changes related to nociceptive stimulation and pain behavior?

In contrast to the work of Kessen and colleagues, who found increased motion in patients with spon*dylolysis*, Pearcy and Shepherd, with biplanar radiography studies, observed decreased motion in all parameters (flexion, extension, lateral bending, axial rotation, and even coupled motions) in symptomatic patients with spond*ylolisthesis*.[115] Perhaps this was due to muscle spasm associated with pain.

There is some basic and clinical information that supports the hypothesis that spondylolysis and spondylolisthesis may be related to mechanical loading. The process may be instantaneous or through negative loading. Studies by Detrich and Kurowski[41] employed mechanical models of muscle and bone and photoelastic materials to determine the sites of stress concentration in the lower lumbar spine. The analyses show that the highest stress concentration is in the region of the pars interarticularis. This investigation offers a neat, though not well-substantiated, hypothesis that addresses the combined role of the material and structural characteristics of the pars interarticularis as major factors determining the ability of the region to withstand potentially damaging loads. In other words, the cross-sectional geometry and the mass of cortical and cancellous bone may determine whether or not the patient acquires spondylolysis.[34]

Some clinical support of the mechanical failure advocates is offered by a study comparing 43 nonambulatory patients with a group of normals. The prevalence of spondylolysis was 5.8% in the normals and zero in the nonambulatory patients. This is significant at the $p < 0.001$ level.[135] This information supports a mechanical failure etiology. The racial differences (whites, 5%; Eskimos, 9%; blacks, 3%) may be explained on the basis of different thresholds for failure due to the structural and material properties of the pars interarticularis in the various racial groups. Differences in certain athletes (gymnasts, sumo wrestlers, football interior linemen) may be explained by the frequency, magnitude, and vectors of loads involved in the respective sports activities.

There are a number of geometric measurements that constitute a detailed quantitative description of the condition. These have been reviewed by Wiltse and Winter.[172] In addition to the utility of these measurements in monitoring changes, some of them may prove to be useful as prognostic indicators. On a theoretical basis, our opinion is that for following patients, the percentage of anterior translation (anterior displacement) and the sagittal rotation are the most important parameters. Measurements that may portend a progression of deformity are: (1) percent rounding of the top of the sacrum, (2) wedging of the displaced L5 vertebra, (3) lumbar lordosis, and (4) a large sacrohorizontal angle. These parameters tend to indicate geometric relationships that, in the standing position, would maximize gravitational loading vectors and increase the deformity.

TREATMENT CONSIDERATIONS

Effects of Reduction on Prognosis

Reduction of lumbar spine fractures and dislocations is thought to help reestablish stability and may decompress the neural canal.[84] As yet, attempts at closed reduction have been associated with increases in neurologic deficits.[81,133] Kaufer and Hayes recommended open reduction, posterior fusion, and wire fixation of the involved FSUs.[84]

The Role of Laminectomy

Laminectomy is rarely a useful procedure in fracture dislocations of the lumbar spine. Many of the injuries include decompression associated with the injury itself. Surgical exploration may reveal a subcutaneous hematoma that, when removed, extends all the way to and exposes the dura or even the cauda equina (Fig. 5-59).[84] Clearly, there is no need for a laminectomy in this situation. Because of the anatomy and the available space for the cauda equina, mechanical pressure is rare, but when present it can usually be corrected by realignment of the vertebra. Certainly, if there is clear radiologic or other convincing evidence that some localized particulate matter is encroaching on the cord posteriorly, laminectomy is indicated.

RECOMMENDED EVALUATION SYSTEM

The Checklist

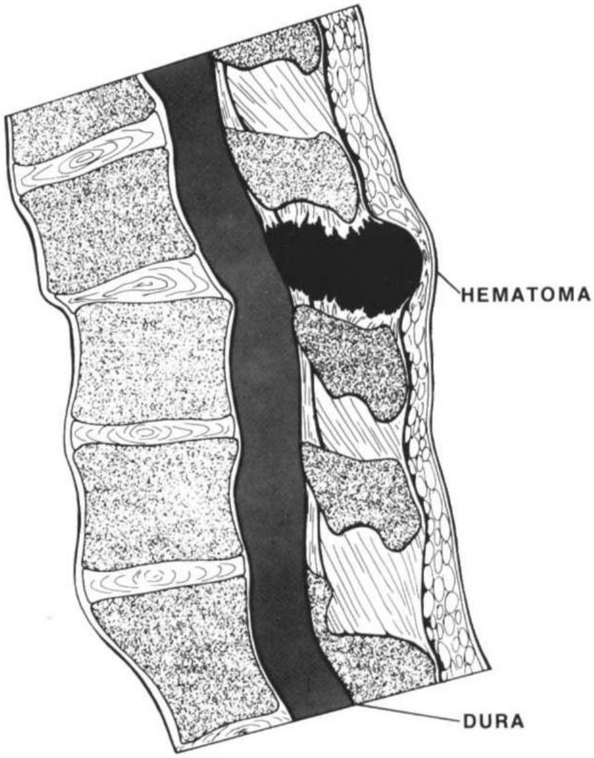

FIGURE 5-59 Depiction of the lumbar spine flexion injury in which all the posterior elements are disrupted and the dura is separated from the outside by only hematoma and skin. This is hardly an indication for decompressive laminectomy. The hematoma is readily palpable between the spinous processes on physical examination.

At the time of publication of the first edition of this text there was a distinct paucity of biomechanical or clinical studies to provide a solid basis for the systematic approach to the problem of clinical instability in the lumbar spine. Apologies were made, and the guidelines suggested were not as vigorously proposed as are those for other regions in the spine. That situation has changed drastically, and although there is not yet a definitive solution, considerable progress has been made! The checklist based on biomechanical experiments and clinical analysis is provided in Table 5-10. Some of the criteria in the checklist are based on the experimental work of Posner and colleagues. These investigators completed studies of FSUs in which flexion and extension were simulated under physiological loads (Fig. 5-60). The anatomic components were sequentially ablated going anterior to posterior in some specimens and posterior to anterior in others. This showed that when all anterior or posterior elements were out, the FSU was either unstable or on the brink of instability.

The problem of clinical instability in the lumbar spine has some unique considerations related to both aspects of the definition of clinical instability. The associated neurologic deficits in the lumbar spine are less frequent, less disabling, and more likely to recover than they are in the cervical and thoracic regions. The second consideration is related

TABLE 5-10 Checklist for the Diagnosis of Clinical Instability in the Lumbar Spine

Element	Point Value
Anterior elements destroyed or unable to function	2
Posterior elements destroyed or unable to function	2
Radiographic criteria*	4
A. Flexion/extension x-rays	
1. Sagittal plane translation > 4.5 mm or 15% (2 pts)	
2. Sagittal plane rotation	
> 5° at L1–L2, L2–L3 & L3–L4 (2 pts)	
> 20° at L4–L5 (2 pts)	
> 25° at L5–S1 (2 pts)	
OR	
B. Resting x-rays	
1. Sagittal plane displacement > 4.5 mm or 15% (2 pts)	
2. Relative sagittal plane angulation > 22° (2 pts)	
Cauda equina damage	3
Dangerous loading anticipated	1

Total of 5 or more = unstable

*See Figures 5-61, 5-62, 5-63 for measurement techniques.

to the phenomena of subsequent pain, deformity, disability, and the very high loads that must be borne by this region of the spine.

Basically, we want to take full advantage of the recuperative power of the cauda equina and minimize the possibility of prolonged disability associated with low back pain. The use of the checklist will not indicate fusion in the lumbar spine purely as a treatment for pain.

In order to provide some perspective, the rationale and experimental basis of the checklist (Table 5-10) are reviewed briefly. The goal was to establish reproducible guidelines based on anatomy, biomechanics, and clinical observations. It was considered desirable to establish criteria that would be applicable to all types of instability analyses. The list is designed with some internal checks and balances, some partially overlapping criteria, and some latitude for "gray zone" weighting of a given criterion (i.e., if the criterion is weighted a 2 and it is not possible to arrive at a definitive yes or no, then it is assigned a 1). Finally, the checklist is based largely on an experiment designed explicitly to address the issue of lumbar spine instability.[125] Eighteen FSUs from three levels of the lumbar spine (L1–L2, L3–L4, and L5–S1) were preloaded so as to simulate the load calculated to be present for a person lying supine and standing, each with maximum physiologic flexion and extension (Fig. 5-60). The sagittal plane translations were then measured using linear variable differential transformers and a minicomputer. Sequential transection of components in the posterior to anterior direction until failure occurred constituted the essence of the experiment. From this it was possible to determine the upper limits of physiologic motion of the intact FSU and the effect of component ablation on normal motion. In addition, it was possible to determine the point in the sequence of component ablation at which the FSU either failed or was on the brink of failure.

Now that the background and rationale of the checklist have been established, let us work through the various considerations involved in using it.

Anatomic Considerations

A schematic representation of the major ligaments and the facet orientation in the lumbar spine is presented in Figure 5-53. The anterior longitudinal ligament is a well-developed structure in this region. The annulus fibrosus, which has received an enormous amount of attention in the literature, constitutes 50–70% of the total area of the intervertebral disc. As in other regions of the spine, it contributes in a major way to the clinical stability of the FSU. The posterior longitudinal ligament is less developed than its anterior counterpart. The facets play a crucial role in the stability of the lumbar spine. The well-developed capsules of these joints play a major role in stabilizing the FSU against axial rotation and lateral bending. When these displacements are observed, fracture or fracture dislocation of the facet articulations must be suspected. Sullivan and Farfan showed in the laboratory that axial rotation of the lumbar spine of 30° or more caused failure of the neural arch, progressing from facet joint dislocation to fracture of the articular process.[148]

The annulus fibrosus and/or vertebral body may be destroyed or rendered unable to function by surgery, trauma, tumor, or infection. The annulus can also be compromised by chemonucleolysis. Extensive plastic deformation of the annulus may occur in long-standing spondylolisthesis and may contribute to instability. Excessive vertebral body wedging can contribute to instability, particularly if the anterior ligaments are not intact. Surgery, trauma, tumor, and infection may also destroy the posterior elements or render them unable to function. The accompanying display lists situations in which the contribution of

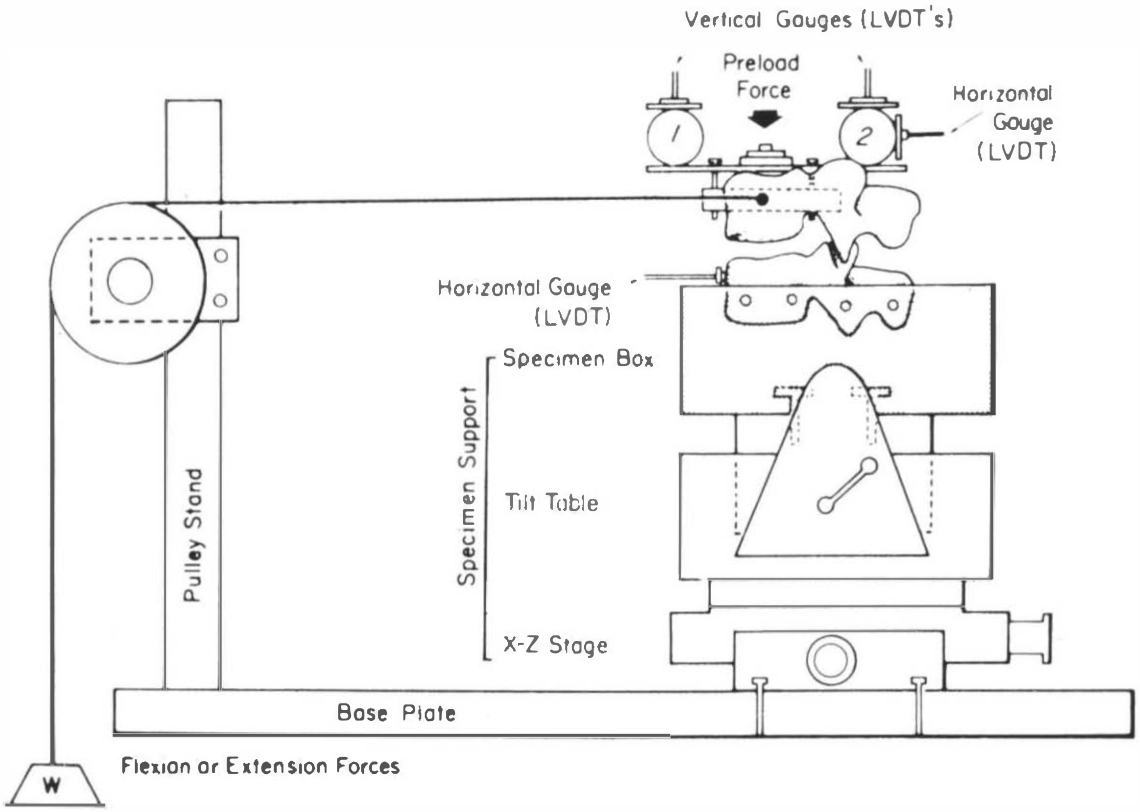

FIGURE 5-60 The experimental setup (1) and (2) consists of measuring balls attached to a top plate, which is attached via a cylindrical aluminum piece to the upper vertebra. The figure shows a functional spinal unit being subjected to a preload and flexion producing horizontal force, with transection of components from posterior to anterior. The specimen is reversed 180° for testing in extension force. *(From Posner, I., et al.: A biomechanical analysis of clinical stability of the lumbar and lumbo-sacral spine. Spine, 7:374, 1982.)*

the anterior elements to stability may be compromised.

SITUATIONS IN WHICH THE ANTERIOR ELEMENTS ARE DESTROYED OR UNABLE TO FUNCTION

- Failure of anterior ligamentous structures
  - Infection, tumor, disease
  - Surgery
  - Chemonucleolysis
- Failure of vertebral body
  - Slice fracture
  - Severe wedge compression fracture
  - Excessively comminuted fracture
  - Aseptic necrosis
  - Infection, tumor, disease
  - Surgery

A classic example of loss of the contribution of the posterior elements to stability is the condition of spondylolisthesis due to lysis of the pars interarticularis. Two points in the checklist are given for the loss of the contribution of the anterior or posterior elements to stability.

Radiographic Criteria

The method of measurement of translation and displacement in the lumbar spine is the same as in the cervical spine. Figure 5-61 depicts the measurement of translation and displacement additionally as a percentage of the width of the vertebral body.

When readily apparent residual displacement remains after the recoil and rebound of injury, the structural damage is obvious. However, where there is little or no residual displacement of the position of

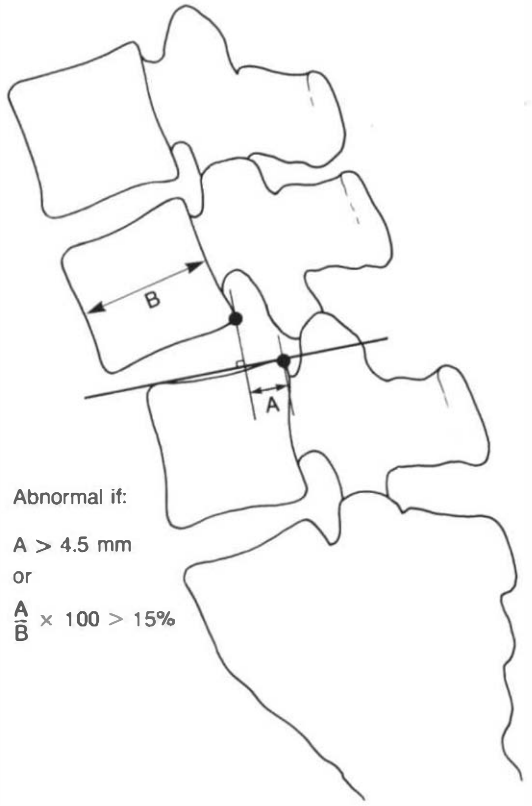

FIGURE 5-61 Measurement to determine vertebral translation or displacement in the lumbar spine. A method for measuring sagittal plane translation or displacement. If the translation or displacement is as much as 4.5 mm or 15% of the sagittal diameter of the adjacent vertebra, it is considered to be abnormal. These measurements are to be used in conjunction with the checklist in Table 5-10.

the vertebrae following injury to the cauda equina, clinical instability must be suspected. The physician should look for other evidence of clinical instability in order to make a diagnosis when there is no residual displacement.

Measurements can be made directly from resting or flexion/extension radiographs. In the acute traumatic setting, resting radiographs are usually performed. *Sagittal plane displacement greater than 4.5 mm or 15% of the anteroposterior diameter of the vertebral body on a static (resting) lateral radiograph should be considered potentially unstable.* These values were obtained from the aforementioned biomechanical experiment.[125] *Relative sagittal plane angulation greater than 22° is abnormal and potentially unstable* at any level in the lumbar spine. Note that 22° of relative angulation means 22° *greater* than the amount of angulation at the FSU above or below the FSU in question (Fig. 5-62). These norms were obtained from a review of the literature of the normal resting sagittal posture of the lumbar spine.[14a,145a] This value was tested on the data of 102 normal subjects. This standard of comparison takes into account the normal angulation between FSUs.

After evaluation with resting radiographs in the acute traumatic setting or in the nonacute setting of evaluating for lumbar spine clinical instability, additional information may be gained by obtaining flexion/extension radiographs. *Sagittal plane translation greater than 4.5 mm or 15% of the anteroposterior diameter of the vertebral body on dynamic (flexion/extension) radiographs should be considered potentially unstable.* These values were obtained from the aforementioned experimental study and several other kinematic studies.[44a,68a,114a,125,173a] *Sagittal plane rotation on dynamic radiographs greater than 15° at L1–L2, L2–L3, and L3–L4, greater than 20° at L4–L5, or greater than 25° at L5–S1 is abnormal and potentially unstable* (Fig. 5-63). These values were based on a review of the literature of *in vitro* and *in vivo* lumbar spine ranges of motion. Clinical techniques designed to accentuate or bring out lumbar instability are shown in Figures 5-64 and 5-65. There is a third set of techniques for bringing out instability that involves hanging from a bar to create distraction forces, and placing a weighted backpack on the torso to cause vertical compression forces. In the checklist, these three techniques are treated as flexion/extension observations.

Two points in the checklist are given for abnormal sagittal plane translation *or* displacement. These are considered as *resting* observations. Two points in the checklist are also given for abnormal sagittal plane rotation or abnormal relative sagittal plane angulation. These are considered as acting or *dynamic* observations.

Clinical Considerations

There is a relatively large margin of safety in the lumbar spine because the space available for the neural elements amply exceeds the space occupied by them. Therefore, the presence of neurologic deficit is very likely to be the harbinger of clinical instability. In other words, if there is enough displace-

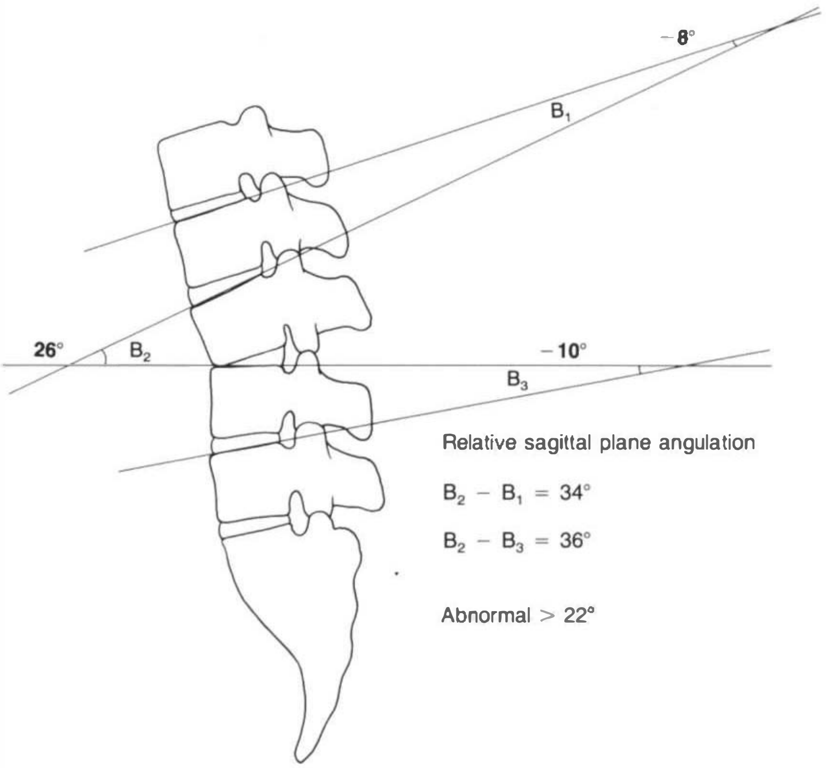

FIGURE 5-62 Measurement of relative sagittal plane angulation in the lumbar spine. A method of measuring relative sagittal plane angulation of the L4–L5 functional spinal unit on a static (resting) lateral radiograph. Relative sagittal plane angulation greater than 22° is abnormal and potentially unstable in the lumbar spine. Note that this means 22° greater than the amount of angulation at the FSU above or below the FSU in question. By convention, negative values denote lordosis and positive values kyphosis. These measurements are to be used in conjunction with the checklist in Table 5-10.

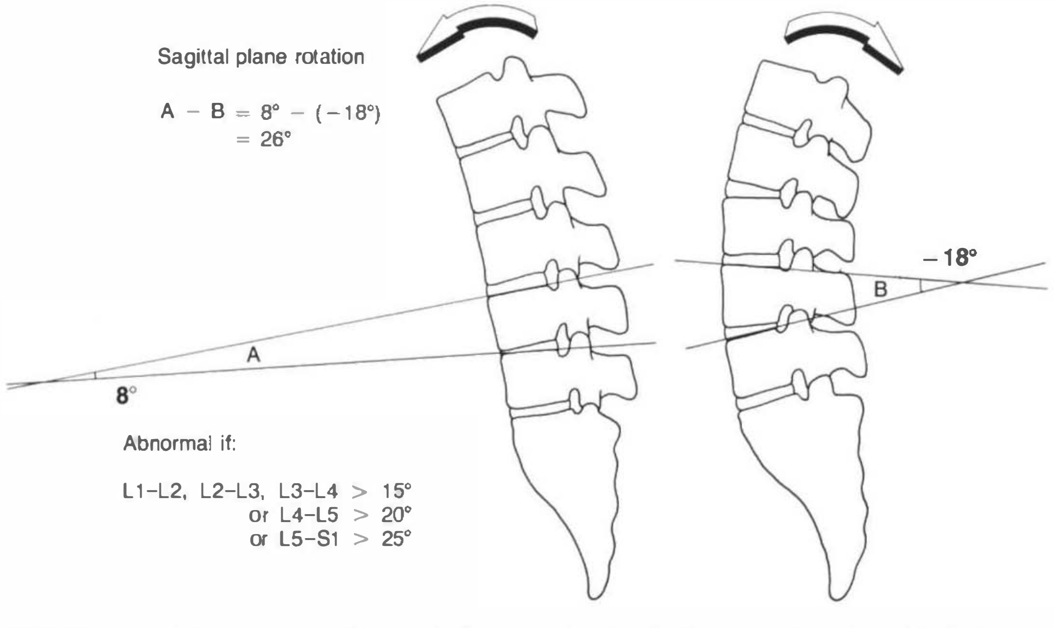

FIGURE 5-63 Measurement of sagittal plane rotation in the lumbar spine. A method of measuring sagittal plane rotation of the L4–L5 functional spinal unit on dynamic (flexion/extension) lateral radiographs. The sagittal plane rotation is the difference between the Cobb measurements taken in flexion (A) and extension (B). Sagittal plane rotation greater than 15° at L1–L2, L2–L3, and L3–L4, greater than 20° at L4–L5, or greater than 25° at L5–S1 is abnormal and potentially unstable. Note that negative values denote lordosis and positive values kyphosis. These measurements are to be used in conjunction with the checklist in Table 5-10.

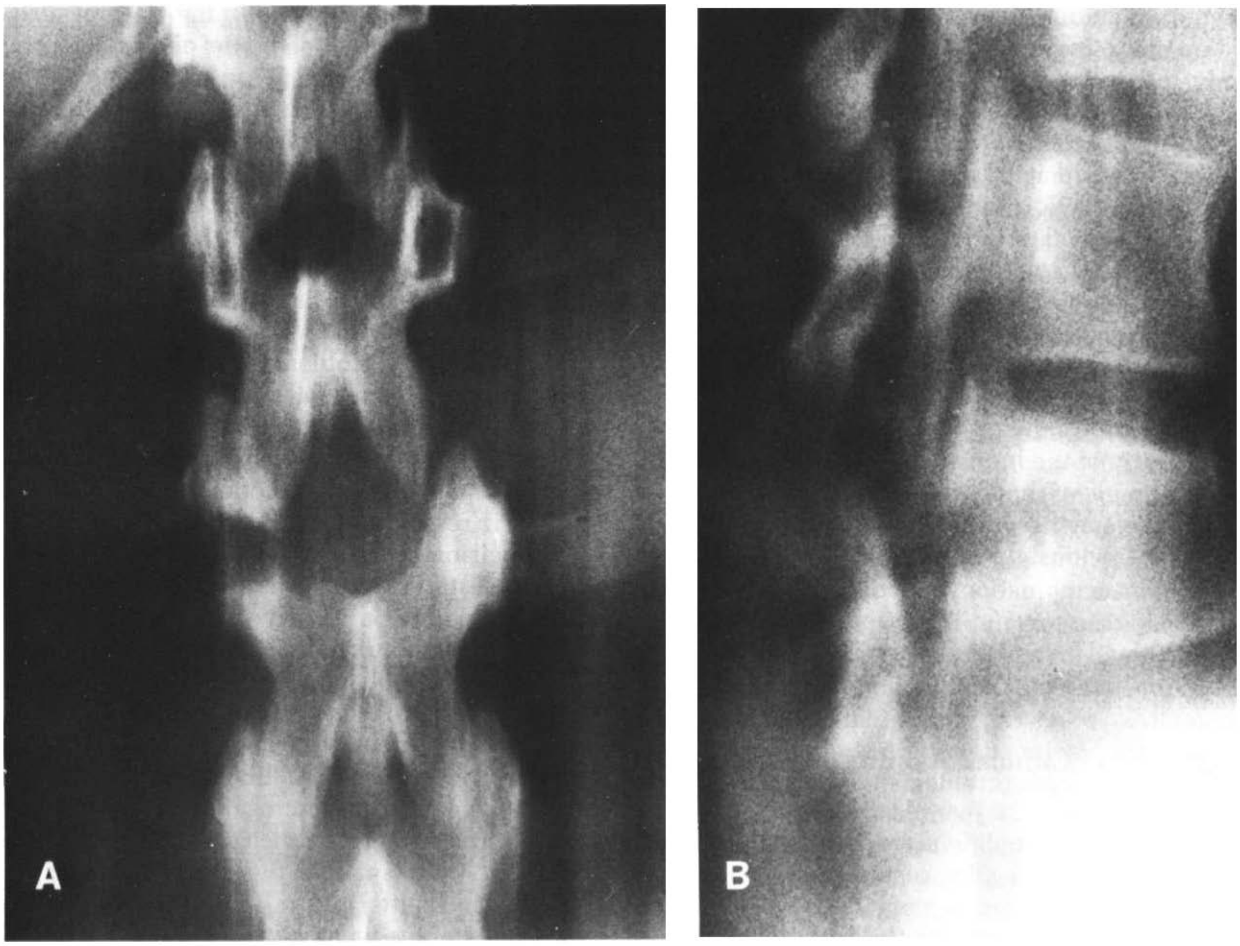

FIGURE 5-66 Radiographs of flexion injury with slight axial rotation. **(A)** A laminagram clearly illustrates that significant abnormal separation of spinous processes occurs only with fracture and/or dislocation of the facet joint complex. On the left side there is a fracture and on the right side there is a dislocation. **(B)** A lateral laminagram shows that there is a fracture of the posterior portion of the vertebral body. Since a major portion of this fracture is through the cancellous bone of the vertebral body, unlike the disc tissue it can be expected to heal and provide ample stability.

oped by Denis and Armstrong[38,40] is as follows. The *anterior column* is composed of the anterior longitudinal ligament, the anterior annulus fibrosus, and the anterior part of the vertebral body. The *middle column* includes the posterior annulus fibrosus, the posterior longitudinal ligament, and the posterior wall of the vertebral body. The posterior *column* is formed by the posterior ligaments and the posterior bony arch. Instability is specifically not defined or classified by Denis. He describes four groups of fractures.[39] The first is described as basic stress failure of the anterior column with the middle column intact. The second is a burst fracture described as failure under compression of the anterior and middle column. The third is the seat belt type of spinal fracture and is the result of failure of the posterior and middle columns under tension (see Fig. 5-67). The fourth is a fracture dislocation of all three columns that can result in neurologic deficit and instability. We have not used the classification to evaluate instability because it has not been helpful in the context of our definition of instability. In addition, there has not, to our knowledge, been any experimental or clinical investigation designed to show the relation-

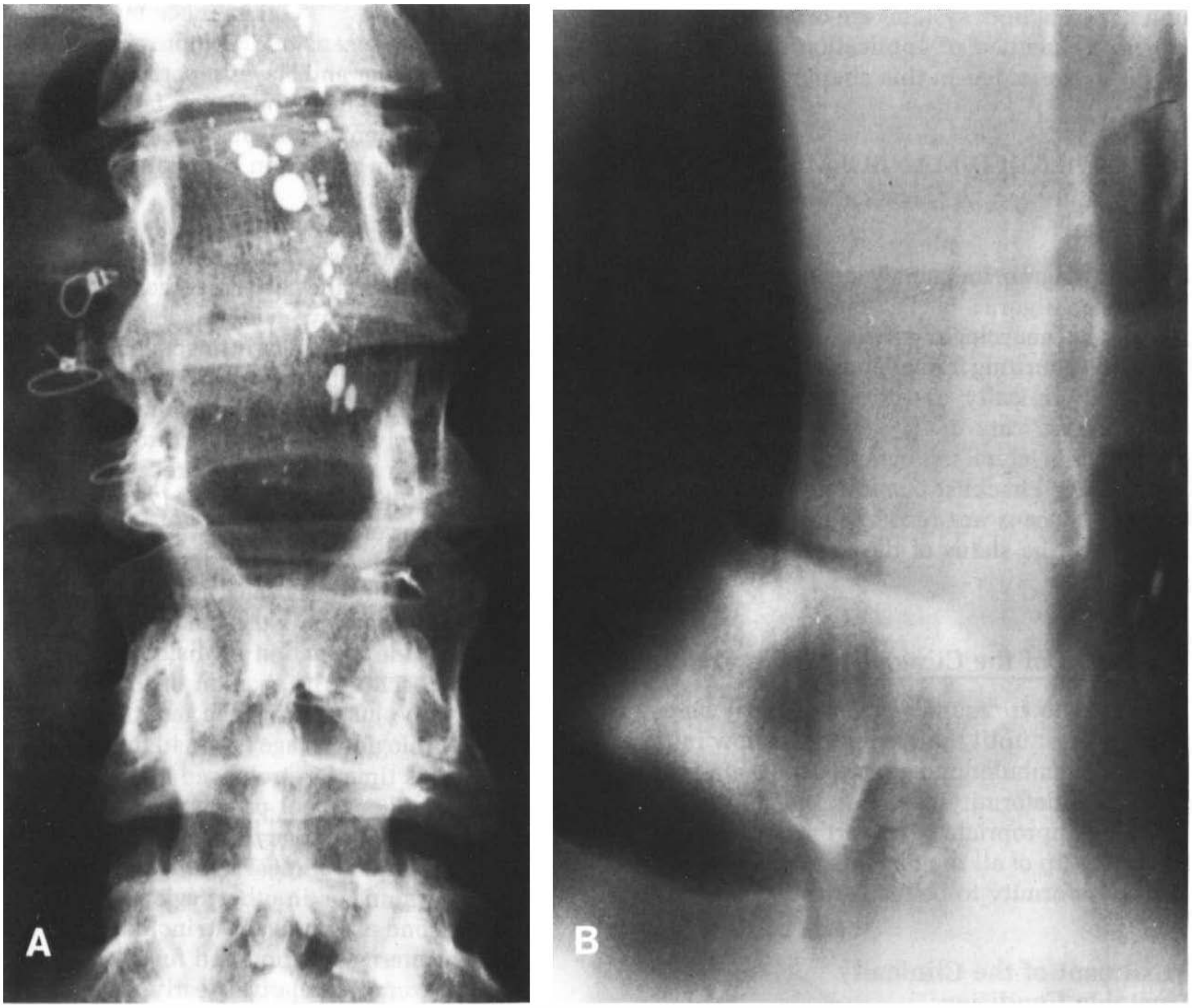

FIGURE 5-67 Radiographs of a patient with clinical instability of the lumbar spine. **(A)** A wide laminectomy at L2, L3, and L4. **(B)** Abnormal posterior displacement of L2 on L3. In addition to surgery of the posterior elements, this patient also shows removal of the disc, which disrupted the stability anteriorly.

ship of the classification to a specific definition of instability. As can be seen in Figure 5-53, by dividing the anterior components between 8 and 9 (anterior and posterior one-half of the annulus), we have the three columns. However, we do not have a systematic method for clinical stability evaluation using these three columns. The concept is nonetheless neat and is a useful framework for classification of fractures in this region.

There is another three-column concept of spine stability proposed by Louis.[93] This will be reviewed briefly to minimize confusion and to compare and contrast the two approaches. Anatomically, the difference is simple. Denis columns are based on an anterior to posterior sequential division into three columns. The Louis columns are based on a tripod anatomic division. The anterior leg or column consists of the vertebral bodies and disc. The posterior two legs or columns consist of the paired joints (see Fig. 5-53). In this system, points are given to various components, and if there is a total of 2 points or greater, the spine is considered to be unstable. This system is significantly different from the Denis system, although they are both called the "three-column" system. Both systems are different in concept and in the method of application to the checklist approach presented in this chapter.

RECOMMENDED MANAGEMENT

Flow Chart

The flow diagram for management of these problems is shown in Figure 5-68. Patients are treated in bed if there is no neurologic deficit; otherwise, they are treated on a turning frame. The patient is thoroughly evaluated clinically, and the necessary supportive and specific care is provided. Regular anteroposterior and lateral radiographs are taken. The clinical stability checklist is applied. In difficult judgments, CT scans are recommended for a detailed analysis of the status of the various anatomic elements.

Treatment of the Clinically Stable Condition

Patients who are found to be stable may be treated with bed rest until their symptoms allow initiation of gradual ambulation and exercises. Pain or fear of increasing deformity may lead the surgeon to prescribe an appropriate spinal orthosis. The schedule for follow-up of all the patients allows any progression of deformity to be recognized (see p. 327).

Treatment of the Clinically Unstable Condition

Conditions determined to be clinically unstable are separated into two groups, based on whether or not there is clinical evidence of cauda equina damage and evidence of a neural impingement on appropriate imaging studies.

Cauda Equina Damage

If there is evidence of impingement upon the cauda equina or nerve roots, there should be early exploration, appropriate relief of the impingement, reduction, internal fixation, and arthrodesis. In view of the excellent recuperative potential of the cauda equina, we do not think that nonoperative treatment is justifiable in the presence of documented cauda equina impingement. Closed reduction of injuries in this group is not recommended, because there have been reports of additional neurologic damage with such attempts.[15,81,141,147] Laminectomy is generally a less effective means of decompression than open reduction. Kaufer and Hayes reported a situation in which laminectomy of four levels failed to relieve a block in a patient who subsequently recovered 1 week after open reduction and stabilization.[84]

No Cauda Equina Damage

If there is a diagnosis of clinical instability *without* neurologic deficit, the need for surgery is less urgent. The available objective evidence does not lead to the conclusion that all clinically unstable lumbar spines must be treated with surgery. We suggest that there are at least three currently justifiable alternatives. The first involves performing arthrodesis, with internal fixation as an elective procedure at a later time. This is done relatively early (7–21 days) or at a later time (several months to years), based on the patient's symptoms and the judgment of the surgeon. Both approaches seem justified by available objective information. Note that these two options do not include reduction of dislocations. The delayed approach to the fixation of these injuries is thought to be justified by the fact that the risk of initial neurologic damage is less in this region. This factor allows time for prolonged observation to determine whether or not pain will be a problem. In other words, the urgency for the establishment of early or immediate clinical stability is not as great in the lumbar spine as in other regions.

The second alternative is to include *open reduction* with internal fixation and fusion. This option should be exercised relatively early, at 7–14 days. In the opinion of some physicians, spondylolisthesis falls into this category. The necessity or desirability of reduction in spondylolisthesis is controversial. However, this can be surgically reduced months or years after its occurrence.

The third option (a nonsurgical alternative) is justified in this group of clinically unstable lumbar spine injuries. Patients are treated with bed rest and active exercise in bed for 6 weeks, followed by gradual ambulation, physical therapy, and protected activity for another 6–12 weeks. A lumbar orthosis of intermediate control may be useful if symptoms of pain demand it.

The key issue is careful follow-up according to the schedule outlined on page 327. Should clinical instability develop, it can be managed effectively. The risks of a major problem in the lumbar spine are small due to the following factors:

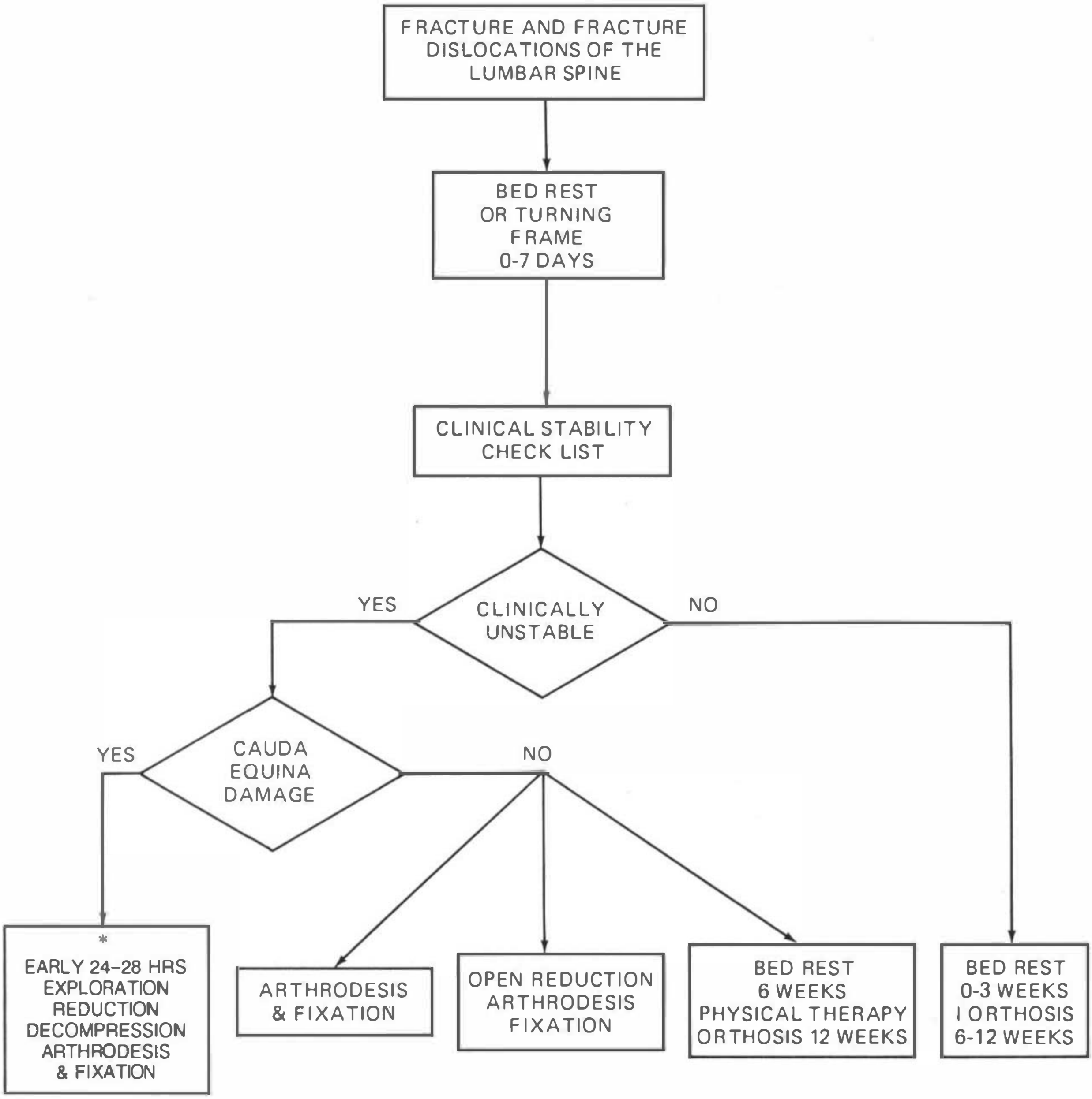

FIGURE 5-68 A recommended flow diagram for the management of patients with disruptions of the lumbar spine. All treatment regimens include occupational and physical therapy, as tolerated, and follow-up schedule, page 327. The decompression or reduction is determined by need. If there is imaging evidence of a fragment in the canal that is not relieved by reduction, then an appropriate decompression is indicated. When there is vertebral malalignment with cauda equina damage, then reduction is crucial.

1. There is a generous amount of free space in the lumbar spinal canal.
2. The large and powerful erector spinae, psoas, and abdominal muscles lessen the risk of catastrophic displacement in the lumbar spine.
3. The recuperative ability of the cauda equina is superior to that of the spinal cord.

For these reasons, clinical instability of the lumbar spine is less dangerous than in the cervical or thoracic spine.

PART 5:

THE SACROILIAC JOINT AND PUBIS

Even a cursory review of the anatomy and kinematics of the three-joint complex in Chapter 2 will make it obvious that establishing, defining, and recognizing abnormal quantity and quality of motion here constitute a great challenge. In other words, considerable additional knowledge is required before sacroiliac instability can be diagnosed and treated.

The clinical stability of the sacrum and pelvis poses a problem slightly different from that in the previously described regions. The main concern here is the ability of these structures to perform their mechanical function after disruption from trauma, disease, or surgery.

ANATOMIC CONSIDERATIONS

The sacrum is stabilized in the pelvic ring by a somewhat unique, ear-shaped articulation that is ingeniously reinforced by several structural characteristics. The joint is narrow and is provided with elevations and depressions. These characteristics limit motion and provide stability. It is fixed posteriorly and superiorly by a strong, stiff articular capsule. This capsule is further reinforced posteriorly and inferiorly by the sacroiliac ligaments, which are the strongest ligaments in the body.[143]

The major load-bearing portion of the pelvic girdle has been described as analogous to an arch with lateral pillars and a keystone (Fig. 5-69).[62] This construct is designed by nature for the support of very high loads. The vertical loads are resisted by the irregular surface of the joint and the wedge-shaped configuration of the sacrum. The separation of the pillars (the femora) is prevented mainly through the tension created by the tensile resistance of the large sacroiliac ligaments posteriorly and the interosseous ligament. The effectiveness of this coapting mechanism is due to the fact that it becomes increasingly stable with increasing loads.[H] There is a similar, secondary role played by the pubic symphysis anteriorly. The pubic symphysis generally consists of a thick cartilaginous disc. There is an inferior pubic ligament that is thought to provide the major stability to the joints.[58]

BIOMECHANICAL FACTORS

It is sometimes necessary that certain portions of the sacrum and ilium be removed to treat a tumor. Gunterberg carried out some tests on fresh autopsy speci-

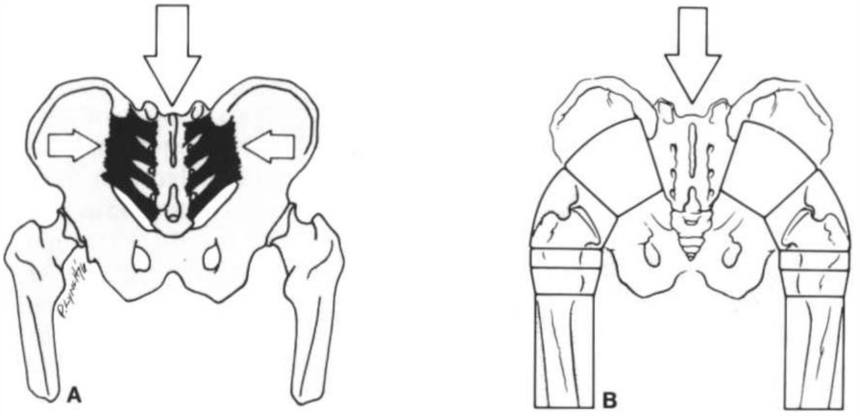

FIGURE 5-69 **(A)** The posterior sacroiliac ligaments are oriented so as to provide additional stability to the sacroiliac articulations with increased loading. **(B)** The sacrum is analogous to the keystone in the sense that it wedges in under compression loading and causes tensile loading in the ligaments. Thus the articulating surfaces and the sacroiliac ligaments work together to provide a shock-absorbing mechanism for the base of the spine.

mens of the pelvis and sacrum to compare the load-bearing capacity of the structures under different conditions. The available specimens were divided into three groups (control group, group Resection A, and group Resection B), with comparable age representation in each group. The control group was left intact. Group A and Group B were resected as described in Figure 5-70. Each specimen was loaded vertically at the top of the first sacral vertebra one to three times with a force up to twice the estimated normal physiologic load. The test specimen was then loaded to failure. As might be anticipated from the anatomic descriptions, these studies showed that the sacroiliac joint articulations remained in-

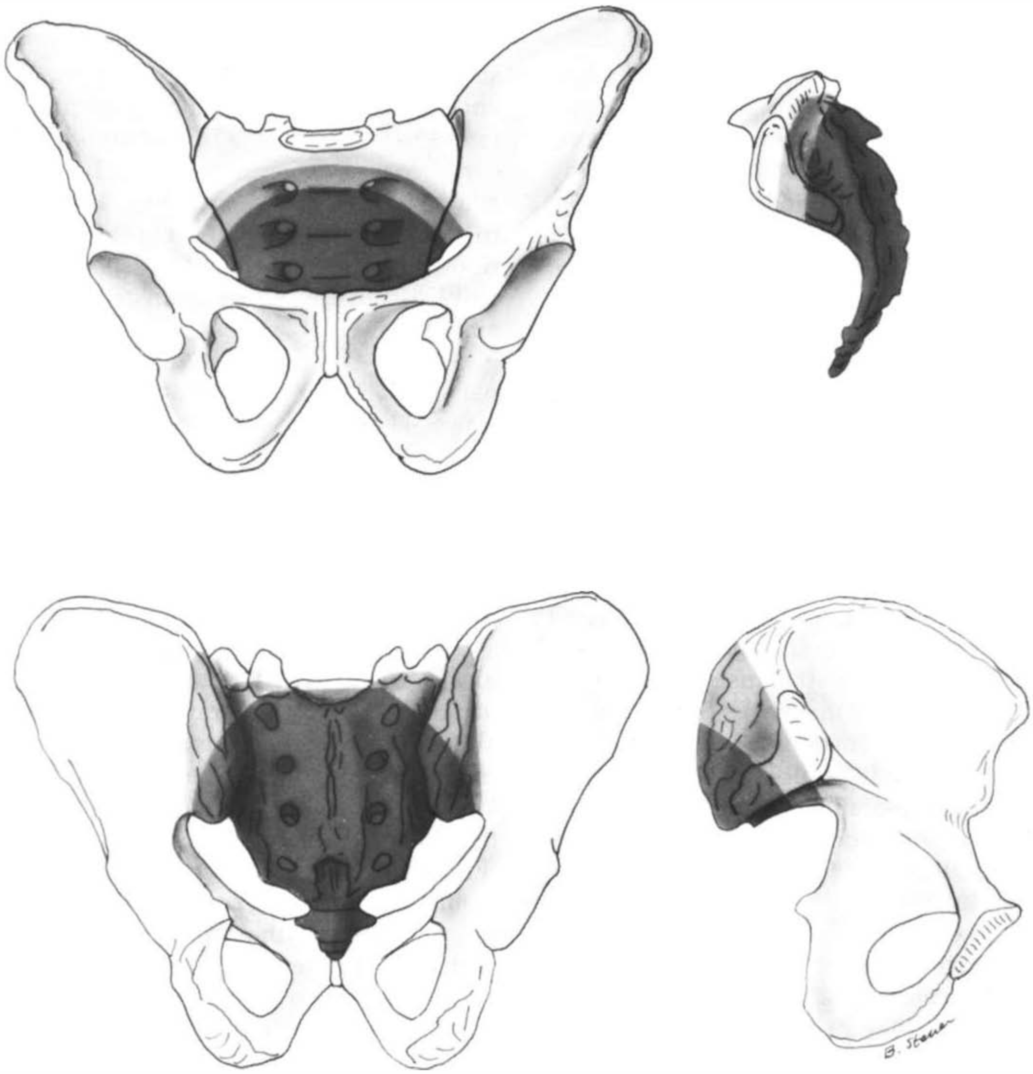

FIGURE 5-70 Resections of two portions of the sacroiliac joints in an experimental study of the load-bearing capacity of the partially resected sacrum. Resection A: resection between the bodies of S1 and S2. In addition, about one-third of the sacroiliac joint and corresponding ligaments were removed on both sides. This resection is indicated by the dark shading. Resection B: resection through the first body of S1. In addition, about one-half of the sacroiliac joint and corresponding ligaments on both sides were removed. This resection includes the dark-shaded structures plus the light-shaded structures. *(Gunterberg, B., Romanus, B., and Stener, B.: Pelvic strength after major amputation of the sacrum. An experimental study. Acta Orthop. Scand., 47:635, 1976.)*

tact. The failure occurred in the lateral parts of the sacrum, in both the resected and the unresected specimens. The data showed that with resection of the sacrum between S1 and S2 (group Resection A), approximately 30% of the ultimate load-bearing capacity is lost, and with resection through S1 (group Resection B), approximately 50% is lost. The failure load was four to eight times the calculated upright standing load in the intact specimen, 1.5 to five times in the resection between S1 and S2, and about two times in the resection through S1. The investigators concluded that the residual strength of the pelvic ring is adequate following resection of the sacrum through S1, leaving some associated iliac bone to allow early ambulation with full weight-bearing in the early postoperative period.[62]

The preceding biomechanical studies are useful from a component ablation perspective. There are also some relevant biomechanical data from studies of motion analyses. One approach has been to evaluate normal and abnormal motion at the pubic symphysis. Radiologic studies have suggested that pubic separations of 10 mm or more are abnormal.[82,83,95] Chamberlain[29a] measured vertical movement of the pubis radiographically with alternate-leg standing and concluded that more than 2 mm of displacement was abnormal. Steiner and colleagues[146] suggested 4 mm, and Hagen[64] reported 5 mm as the upper limits of normal. Some of the investigations interpreted vertical displacement of the pubis greater than normal to be an indication of pelvic instability. Subsequent work by Walheim involving patients diagnosed as having pelvic instability was thought to refute the assertion that pubic symphysis hypermobility is pathognomonic of pelvic instability. Walheim, in his *in vitro* and *in vivo* studies, including some patients, concluded that there is motion of the pubic symphysis in all three planes and that the largest displacement is in the vertical direction, with 3 mm being the upper limits of normal.[157] This investigator also reported a small amount of rotation (0.5°) in both the x,y and the y,z planes (*i.e.*, the frontal and sagittal planes, respectively). The *in vivo* studies included data collected from electronic displacement gauges attached to pins implanted in the bone on both sides of the pubic symphysis. We suggest that, in addition to translating in all three planes, the bones across the symphysis probably rotate in relation to each other about three mutually perpendicular axes. Thus, there is probably a small amount of motion in all six degrees of freedom.

CLINICAL CONSIDERATIONS

Instability in the sacral area may result from trauma, destruction by tumor, infection, or surgical resection or debridement. The question of management involves essentially bed rest versus ambulation. Evaluation of the ligamentous structures is difficult. The main evaluation is with radiographic examination. The basic guideline is the previously described biomechanical study. This study suggests that as long as the destruction leaves intact a fair portion of the first sacral body and its corresponding lateral structures and articulations, the patient may gradually ambulate. There is evidence that additional load-bearing capacity may develop through biomechanical adaptation. Fractures or dislocations and other disruptions of the architecture of the pelvic ring anterior to the hip joint may be associated with severe pain, but they are rarely unstable with regard to ambulation.

Pain in the sacroiliac and/or pubic region of the pelvis has been alluded to as pelvic instability. This is not a well-documented pathologic clinical entity. Nevertheless, a number of clinicians consider it important, and thus it is useful to study it. The association with major trauma and the hormonal changes in pregnancy is well recognized. The validity of pelvic instability as a cause of low back pain and leg pain is the challenging issue.

The clinical diagnosis generally is based on the localization of pain in the region, the demonstration of provocative tests, x-ray evaluation for abnormal alignment, and a pelvic external stabilization test to relieve pain. Using a trapezoidal Hoffman compression frame in 12 patients thought to have pelvic instability, Walheim[157] was able to show some effect (Fig. 5-71). Subjective symptoms decreased considerably in 11 of the 12 patients, and there was also improvement in their analyzed gait and in their Trendelenburg test.

There was no change in symphyseal mobility, and the sacroiliac joint motion was not measured. These preliminary studies suggest by the relief of pain and improvement of function that stabilization of the pelvis may serve as a diagnostic aid in the evaluation of patients suspected of having pelvic instability. It is our view that the available evidence and the risk/benefit ratio are such that the use of Hoffman pelvic fixation for this diagnostic purpose would be best considered in unusual circumstances and/or as experimental clinical development. Wal-

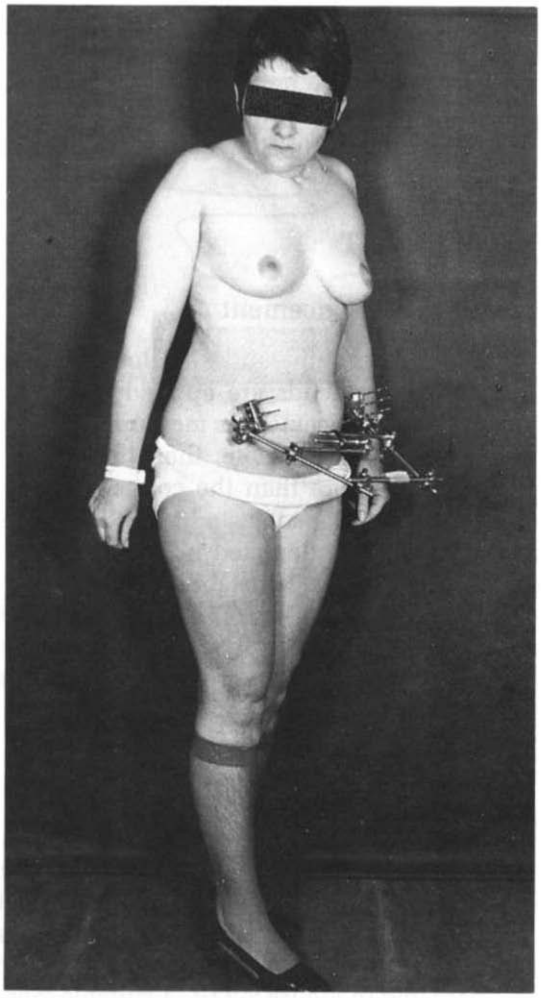

FIGURE 5-71 This external fixture is thought to be capable of immobilizing the pelvis effectively enough to determine whether or not low back pain is coming from the sacroiliac joints. (From *Walheim, G.: Pelvic instability aspects of diagnosis and treatment [Thesis]. Karolinska Hospital, Stockholm, Sweden, 1983.*)

heim, in collaboration with Olerud,[157] describes two groups of patients with pelvic instability who had fusion of the pubic symphysis, the sacroiliac joints, or all three. The patients enjoyed some degree of relief. This work constitutes an innovative approach and will be followed with interest. Certainly, a test that can provide quantifiable mechanical changes that can be correlated with pain relief and quantifiable clinical improvement is a sound approach to the problem of clinical stability in any region of the spine.

Trauma and Pelvic Instability

There are some urgent considerations that relate to pelvic instability following trauma. Gamble and colleagues[58] described several types of injury of the pubic symphysis. They include: (1) diastasis, (2) straddle fracture, (3) intra-articular fracture and overlapping dislocation, and (4) combinations of the above. The diastases were subclassified into three types: (a) open book, (b) vertical displacement, and (c) posterior displacement. Peltier and associates[117] considered a displacement of >1 cm evidence of pubic instability and a diastasis of >2.5 cm indicative of sacroiliac joint damage.[121]

Pennal and co-workers[121] have classified fractures of the pelvic ring into three distinct types based on specified radiologic analysis presumed mechanisms of injury. All three types have an anterior disruption occurring either through or closely adjacent to the pubic symphysis and a posterior disruption. The posterior disruption involves (1) sacroiliac dislocation, (2) sacral fracture, or (3) a fracture of the ilium near the sacroiliac joint. These authors suggest that *instability* will always be present with vertical shear fractures. These fractures generally involve disruption of the pubic symphysis and the sacroiliac regions. Although the authors do not define instability in the article, the radiologic classification and hypotheses constitute a useful contribution.

RECOMMENDED EVALUATION SYSTEM

A checklist is proposed for this region (Table 5-11). The list is designed to evaluate chronic pain instability as well as post-traumatic instability.

After an astute review of the literature, Schmidek and colleagues[138] noted that neurologic deficits of the cauda equina and peripheral nerves involved with the pelvic ring were at least grossly associated with more severe trauma and vertical shear fractures, although this was not universally so. We know that Pennal and associates noted a correlation between vertical shear injuries of the pelvis and instability. We put this together with our hypothesis that there is an association between neurologic deficit and instability. This is based on the logic that if the anatomic structures are destroyed and displaced enough to cause neurologic damage, there is a significant possibility that the structure will remain vulnerable to that same displacement or more. There-

TABLE 5–11 Checklist for the Diagnosis of Clinical Instability of the Sacroiliac Joint and Pubis

Element	Point Value
Pain relief with pelvic fixation	3
Abnormal displacement	
>1 cm sacroiliac	2
>2 cm pubic diastasis (horizontal)	1
>1 cm pubic displacement (vertical)	1
Audible click with associated pain	1
Pain with maneuvers to stress pelvic ring	1
Dangerous loads anticipated	1

Total of 5 or more = unstable

This checklist is *preliminary* and is offered as a guideline because it does not have the experimental support and clinical experience upon which the middle and lower cervical spine checklist and the lumbar and lumbosacral checklist are based.

Though not included in the checklist, remember the work of Gunterberg, which showed that removal of the sacrum at and below the sacroiliac level results in 50% loss of ultimate load-bearing capacity.

fore, the structure may be clinically unstable. There are, of course, exceptions, and this is the reason for the checks and balances of the checklist approach. Pain relief with pelvic fixation is included because of the work of Walheim reported on page 364. The figures for abnormal pubic displacement are based on the literature.[58,117,121,157] They have been weighted for extent of displacement. The presence of a click, the pain with maneuvers to stress the pelvis, and the dangerous loads are some additional clinical varieties that seemed important in our review of the literature.

When instability persists or can be expected to persist, we recommend stabilization with internal or external fixation and arthrodesis for sacroiliac joint dislocation. This decision for the extent and type of fixation is best determined by the clinical conditions and the judgment of the surgeon.

PART 6: SOME THEORETICAL CONSIDERATIONS ON THE BIOMECHANICS OF INSTABILITY

The basic mechanical phenomenon in instability is the abnormal displacement of portions of the spine under physiologic loads. The displacement may take the form of translation, rotation, or some combination of the two. Similarly, the physiologic load may be a force or a moment but in reality is usually some combination of the two. However, for the purpose of analysis, they may be thought of as separate entities.

DISPLACEMENT

Translatory Displacement

Study a functional spinal unit. The lower vertebra is fixed. Physiologic loads are applied to the upper vertebra, and displacement is measured. If the FSU is biomechanically unstable, then the upper vertebra translates more than the corresponding vertebra of a stable FSU subjected to the same physiologic force. This is depicted in Figure 5-72A. An example of translatory instability is anterior displacement of C5 on C6 after a bilateral facet fracture dislocation. An anteriorly directed physiologic force would be expected to produce greater anterior translation of C5 in the unstable spine than it would in a corresponding stable spine.

Rotatory Displacement

The situation is similar for rotatory displacements. Here an unstable spine will show greater rotatory motion than a stable spine when the two are subjected to the same physiologic moments. This concept is depicted in Figure 5-72B. A suitable example of rotatory instability is a spine with unilateral facet fracture dislocation and partial rupture of the disc. When this spine is subjected to an axial torque, the upper vertebra may be expected to rotate about an axis near the intact facet joint.

LIGAMENTS AND STABILITY

For a basic understanding of the stability of the spine, it is helpful to visualize the roles played by different ligaments. The intrinsic translatory and rotatory stability of the spine is provided by the ligaments. The contribution of a given ligament depends not only upon its particular strength but also upon its location. Moreover, a given ligament may contribute relatively more to either translatory or rotatory stability, depending upon the loading circumstances. For example, the interspinous ligaments may contribute significantly toward the rota-

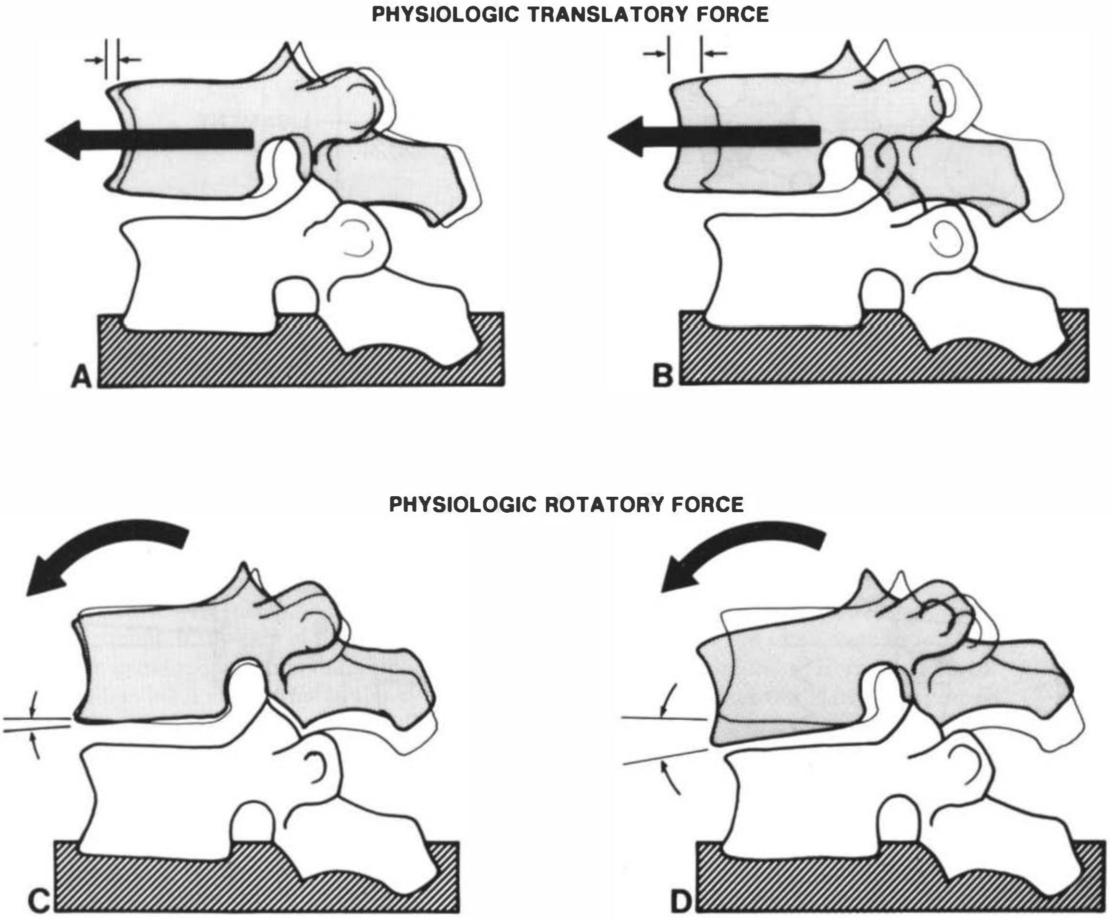

FIGURE 5-72 (A) A stable spine FSU that translates very little when subjected to an anteriorly directed force. (B) An unstable FSU, which characteristically translates more under the same load. (C, D) Here, the FSUs are being subjected to a physiologic bending moment. A greater angulation occurs in an unstable spine (D) as compared with a stable spine (C).

tory stability in flexion $(+\theta x)$ but little toward translatory stability in the anteroposterior direction.

Assuming that all ligaments are made of the same material, the strength of a ligament will be proportional to its cross-sectional area. A ligament with a larger cross-sectional area will provide greater stability and less displacement when the FSU is subjected to physiologic loads. An example of this is the annulus fibrosus, which has much greater area as compared with the interspinous ligament and therefore provides much greater stability.

Another factor that contributes to stability is the distance of a ligament from the center of rotation. An analysis of a single ligament may be done with the help of a simple mechanical model of an FSU. The concept is depicted in Figure 5-73. The model, consisting of a block (upper vertebra) and a spring (the ligament under analysis), is shown in Figure 5-73A. The ligament in Figure 5-73B is closer to the center of rotation than is that in Figure 5-73C. As the moment M is applied to the two constructs, the resistance to motion is provided by the forces in the ligaments multiplied by their corresponding lever arms, L_1 and L_2, respectively. As L_2 is the larger of the two, the design of Figure 5-73C provides greater rotatory stability than does that of Figure 5-73B. The example mentioned earlier may again be used to illustrate this point. The centers of rotation for flexion are in the posterior region of the vertebral body. This gives a greater lever arm for the interspinous ligaments as compared with the annulus fibrosus. Therefore, the contribution toward the rotatory stability due to the

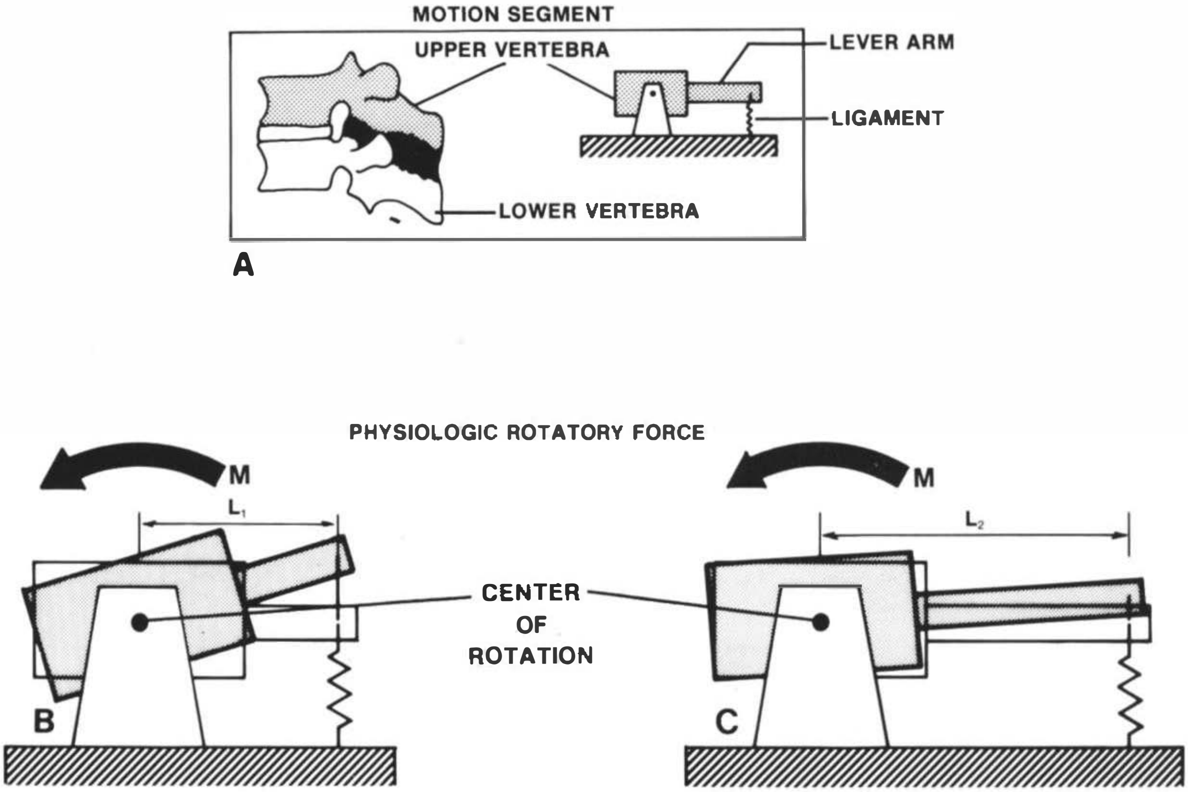

FIGURE 5-73 The location of a ligament with respect to the center of rotation determines its contribution to the stability of the spine. (**A**) A model of an FSU and a single ligament. The lower vertebra is represented by the white trapezoid. (**B**) The ligament that is located nearer to the center of rotation provides much less stability against bending than (**C**) a ligament that is further away from the rotation center. Additionally, orientation of a ligament also plays a part in providing stability (see Chap. 1).

location only would be greater for the interspinous ligament. But, of course, the strength of the annulus outweighs this advantage many times over.

In discussing the stability of the spine and the various factors that contribute to it, the real situation has been considerably simplified for the sake of analysis. However, this helps one to make certain judgments about the stability of the injured spine in an objective manner. The analysis requires two types of information—the extent of structural damage to the spine and the physiologic loads. The former consists of identifying the ligaments that are nonfunctional, their cross-sectional areas, and their locations. The latter depends upon the anticipated physical activities of the patient. This type of analysis, together with the relevant clinical information, permits assessment of the clinical stability of a given spine.

DISPLACEMENT AND CORD ENCROACHMENT

What is the relationship between the displacement and the actual decrease in the vertebral canal space at the level of the displacement? It is important to determine the degree of this decrease because it is closely related to the potential for cord damage.

The unstable FSU is represented by two identical rectangular blocks (vertebrae) with circular holes (spinal canal). The canal space is maximum with perfect vertebral alignment. Any relative displacement of the blocks results in a decrease in the space available for the spinal cord.

Figure 5-74A shows the situation in which, for a given angular displacement, there is the least possible decrease in canal space. This occurs when the center of rotation coincides with the center of the

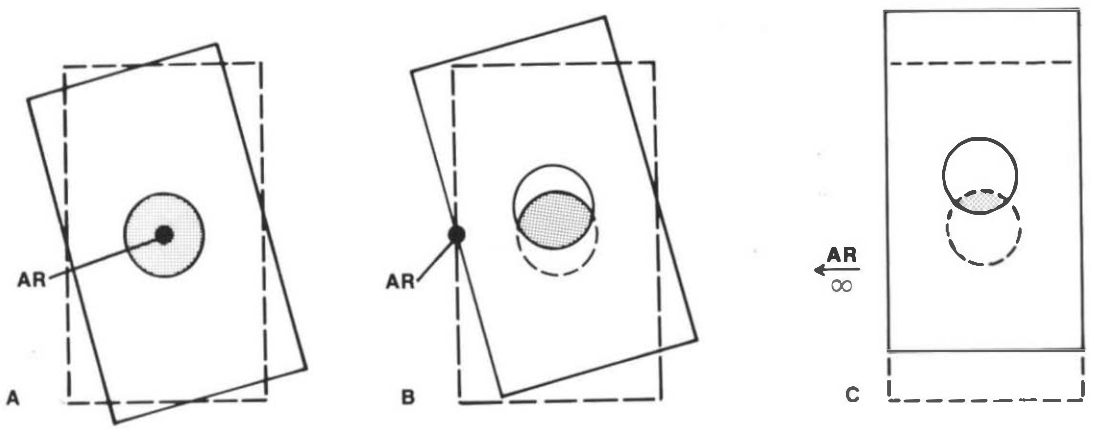

FIGURE 5-74 Spinal cord encroachment is not just a function of how much angulation is produced when the spine is subjected to an axial torque. It is also dependent upon the location of the axis of rotation (*AR*) in the horizontal (axial plane). This concept is exemplified here by a pair of blocks (vertebrae) with holes (canal space). **(A)** Minimum encroachment. One block is rotated with respect to the other around the center of the holes. For the circular hole (canal space), there is no decrease in the available canal space. **(B)** Intermediate encroachment. If the axis of rotation is shifted to the side, then for the same angulation, the available canal space is markedly decreased. **(C)** Maximum encroachment. Here, the axis of rotation has been shifted far to the side (at infinity), producing near translatory displacement.

cord. In Figure 5-74B, a greater encroachment of the space is observed, although the same amount of rotatory displacement is present. This is due to the fact that the center of rotation is located away from the canal. An example is a fracture dislocation in which one of the facet joints is destroyed along with enough ligaments to allow rotation to take place about the relatively intact facet joint. Finally, there is the case in which the decrease in space around the cord is maximal (Fig. 5-74C). This occurs in association with a pure translation. An example is a fracture dislocation of both facet joints, allowing a large translatory displacement in the anteroposterior direction. In such a situation, the displacement of the upper vertebral canal is equal to the displacement of the vertebral body translation. The three situations are well exemplified by the various types of dislocations and fracture dislocations at the level of C1–C2 (see Figs. 5-26, 5-23, and 5-19, respectively).

CONCLUSION

Our basic approach has been to take what is known of anatomy, biomechanics, and documented clinical experience and to analyze it in a manner that is clinically useful. A major anatomic consideration is the clinical significance of regional variations of several structural characteristics. Examples include the anterior longitudinal ligament, the yellow ligament, and the spatial orientation of the facet articulations and the disc in the standing posture. The relative size of the neural elements and the space in which they are enclosed is a cogent consideration. Regional variations also exist in mechanical properties such as kinematics, stiffness, and physiologic loads. We have emphasized the importance of a proper interpretation of the significance of neurologic deficit in the determination of clinical stability. Generally, when a deficit is associated with significant structural damage, clinical instability should be suspected. The importance of standardization of radiographic techniques for more precise interpretation cannot be overemphasized.

The bony architecture and the ligamentous elements constitute the structural components of the spine. With all components intact, the biomechanical function is normal. When sufficient anatomic disruption causes or threatens to produce an inability to function normally, we recommend that the spine be considered clinically unstable.

The goal of good patient management is to gain maximal recovery as rapidly as possible and avoid unnecessary treatment (surgical or nonsurgical), and at the same time prevent the unhappy tragedy of initial or subsequent neurologic damage. There is no convincing evidence that a diagnosis of clinical instability demands that the treatment be surgical reduction, fusion, or fixation. However, the management of a patient with such a diagnosis should definitely differ from that of clinically stable patients.

The role of laminectomy in the management of spine trauma should be diminished and is indicated in only a few special situations. The indications for decompression need considerable elucidation.

Checklists, flow charts, and follow-up schedules have been presented to conveniently organize and summarize the information, to stimulate others to criticize and improve upon them, and to provide clinical protocols for systematic evaluation, management, and study. The concluding principle is that it is only through clear prospective clinical protocols that we can ever really improve our knowledge and base our decisions more on solid scientific evidence and less on well-meaning speculation.

> Without theory, practice is but a routine bore of habit. Theory alone can bring forth and develop the spirit of invention.
>
> PASTEUR

■ CLINICAL BIOMECHANICS

■ Standardization of radiographic techniques is important, and specifications (source, subject, and film distances) should be known when interpreting the significance of measurements.

■ The expression of linear measurements as a percentage of some constant anatomic reference can control for magnification.

■ The normal spinal cord has an ample range of elasticity when deformed in the axial direction. However, it is more prone to damage if displaced in the horizontal plane. Therefore, axial separation between vertebrae is safer than flexion/extension.

■ Even though there is mainly passive elongation of the spinal cord within the canal, there is some cephalocaudal displacement within the canal.

■ The space available in the cervical canal (anteroposterior diameter and cross-sectional area) is a critical variable in the clinical biomechanics of instability and myelopathy secondary to trauma and degenerative processes.

Occipital-Atlanto-Axial Complex

■ Dislocations at the C0–C1 level are usually fatal. Any patient who happens to survive is clinically unstable and should be fused, C0 to C2.

■ More subtle translatory and axial rotatory instabilities may exist at the C0–C1–C2 complex, following trauma and with other conditions.

■ With transection of the tectorial membrane and the alar ligaments, there is an increased flexion of the units of the C0–C1–C2 complex and a subluxation of the occiput.

■ Transection of the alar ligament on one side causes increased axial rotation on the opposite side. This may be analogous to traumatic rupture in the clinical setting.

■ The articular capsules between C1 and C2 are designed loosely, such that they allow a large amount of rotation but provide a small amount of stability.

■ The major stability is provided by the dens and the transverse ligament. The latter is the most important structure in preventing anterior translation.

■ The alar ligaments are complex and variable anatomically; however, they play important roles in the clinical biomechanics of the C0–C1–C2 complex.

■ The apical and alar ligaments may be expected to contribute significantly to C0–C1 stability and also to C1–C2 stability.

■ Although the C1–C2 complex is clinically unstable after failure of the transverse ligament, the tectorial membrane, the alar ligaments, and the apical ligaments probably provide some resistance to gross dislocation.

■ With 7 mm total lateral overhang following a burst fracture of the ring of C1 (Jefferson fracture), the transverse ligament is torn and the situation is clinically unstable.

■ The nuchal ligament hypothetically has potential as an important structure in the clinical biomechanics of postsurgical kyphosis and/or whiplash-type injuries.

The Middle and Lower Cervical Spine

■ The annulus fibrosus is the crucial stabilizing structure, largely because of the attachment of the peripheral annulus directly to the bone through Sharpey's fibers.

■ Clinical stability is lost or in danger when either all the anterior elements or all the posterior elements are destroyed or unable to function.

■ Although there are exceptions, there is a rough correlation between magnitude of structural damage of the spine and extent of neurologic deficit.

- Although there are exceptions, the evidence suggests that when there are spinal cord or root symptoms or signs associated with spine trauma, a clinically unstable situation is to be suspected.
- Unilateral facet dislocations, when associated with fracture and/or neurologic deficits, are probably unstable. Those which have no neurologic signs, symptoms, or fractures and are difficult to reduce are probably stable.
- Removal of more than 50% of the normal facet structures in the presence of a laminectomy significantly compromises clinical stability.
- Bilateral facet dislocations and fracture dislocations are unstable. Abnormal anteroposterior displacements occur with relative ease following facet fractures.
- The major factor in determining the overall prognosis for recovery from neurologic deficit is the nature and magnitude of the initial trauma to the neurologic structures.
- Based on the available knowledge and current biomechanical analysis, a checklist for diagnosis and a flow diagram for management of clinical instability have been provided.
- Controlled, monitored axial traction (the "stretch test") can be helpful in the evaluation of the integrity of the ligamentous structures of the lower cervical spine.
- Patients with anterior elements destroyed are more clinically unstable in extension, while patients with posterior elements destroyed are more unstable in flexion.
- Given the diagnosis of clinical instability, the patient may be adequately treated by surgical fusion, a halo apparatus with trunk attachment, or prolonged traction. There is no convincing evidence in the literature that any particular method is superior. However, recent trends have been toward surgical stabilization for earlier discharge from the hospital and commencement of rehabilitation.
- We usually employ surgical fusion for clinically unstable spines because it has been successful in our experience.

The Thoracic and Thoracolumbar Spine

- The thoracic spine is mechanically stiffer and less mobile than the other regions of the spine.
- Thoracic spine stiffness, which is due to intrinsic qualities as well as the rib cage and its manner of articulations with the vertebrae, provides considerable stability.
- As a result of normal kyphosis, the thoracic spine is more likely to be clinically unstable in flexion.
- The anterior and posterior longitudinal ligaments, as well as the yellow ligaments, are well-developed structures in the thoracic spine.
- The ligaments of the facet joints are not well developed and offer little support following laminectomy, which results in a loss of yellow ligament support.
- Because of the anatomy and geometry of the lower thoracic and thoracolumbar spine, radiographic evidence of local malalignment of the spinous processes in axial rotation is suggestive of dislocation or fracture dislocation of the facet articulations.
- Removal of or loss of function of the posterior elements of the thoracic spine allows for significantly more motion in flexion, extension, and axial rotation. Removal of posterior elements tends to make the spine unstable in flexion.
- Removal of or loss of function of the anterior elements tends to make the thoracic spine unstable in extension.
- Extensive patient management experience and biomechanical evaluation both suggest that the rotational (axial rotation) fracture dislocation is one of the most clinically unstable injuries recognized.
- Bursting fractures may progress to spontaneous arthrodesis.
- Although it is not always clear-cut, it is more commonly the case that extensive structural damage tends to be correlated with neurologic damage.
- The isolated vertebral body fracture is least likely to be associated with neurologic damage, and the fracture dislocation with body and posterior element damage is most likely to be associated with neurologic damage.
- Frequently, with a fresh vertebral body fracture or a fragment of bone in the spinal canal, an intact posterior longitudinal ligament made taut by distraction will reduce the fragment.
- Most traumatic kyphotic deformities do not progress significantly. However, there are certain biomechanical considerations that indicate a likelihood of progression.
- Paraplegics with 10° or more of frontal plane angulation in the thoracic or thoracolumbar spine may be more prone to ischial ulcers from sitting.
- There is no correlation between severity of stable

compression fracture deformity and ultimate clinical outcome.

- The theoretical probability of progression of deformity depends upon the continuity of the posterior elements, the sagittal plane angulation, the amount of wedging of the vertebral body, and the presence or absence of aseptic necrosis.
- Laminectomy in the thoracic and thoracolumbar spine should be discouraged because it is detrimental to the biomechanical functions of this region of the spine. Nevertheless, there are situations when it is appropriate.
- Anterior decompression is more rational biomechanically and shows promising results. However, the indications and benefits of this procedure need careful scrutiny to thoroughly determine its value.
- The Milwaukee brace, designed to treat kyphosis, is thought to be a satisfactory orthosis for fixation of this region of the spine.

The Lumbar and Lumbosacral Spine

- Recent studies show that the facet joints are major players in the stability of the lumbar and the lumbosacral spine.
- The osseous structures of the facet joints limit motion considerably, particularly in regard to axial rotation. The capsule is a major player in the limitation of flexion.
- Once all the posterior elements are removed, or if all the anterior elements are removed, the FSU may be unstable or on the brink of instability.
- Facet dislocations are suggestive of clinical instability and are associated with a variety of other bony and ligamentous injuries, mainly posterior but also anterior.
- The interspinous ligaments in the adult lumbar spine are frequently absent, ruptured, or degenerated and contribute nothing to stability. However, the supraspinous ligaments do play a role.
- The quality as well as the quantity of motion may be important variables in instability of this region of the spine.
- The biomechanics of the lumbar spine and that of the lumbosacral joint are significantly different in regard to clinical instability.
- Although there are intriguing studies documenting changes in the normal *quantity* and *quality* of motion, there has not yet been a compelling correlation of either with pain behavior.
- The presence of neurologic deficit post-trauma is *strongly* suggestive of clinical instability.
- Laminectomy to treat trauma in these regions is not automatically the best choice. Other alternatives should also be considered.
- Two key radiographic observations are abnormal rotatory alignment and abnormal separation of the spinous processes. Either or both of these alert the observer to the possibility of clinical instability.
- On physical exam, an obvious, palpable, subcutaneous hematoma between two abnormally separated spinous processes suggests a clinically unstable lumbar spine until proven otherwise.
- The decision to fuse following spinal stenosis depends upon anticipated patient activity, the structural integrity of the facet joints, the annulus fibrosus and "factor-x," to be determined by further research.
- The mechanical changes that follow spondylolysis and spondylolisthesis are complex and variable and incompletely understood. Biomechanical principles are helpful in determining prognosis and management.
- There is adequate data for use of a systematic checklist approach to the evaluation of clinical stability in this region of the spine. There are two checklists—one for the lumbar spine and one for the L5–S1 FSU.
- Vertebral wedging probably increases the propensity for progressive deformity by shifting the center of gravity forward and increasing a deforming bending moment.
- The highest stresses in the lumbar vertebra with physiologic loading occur in the region of the pars interarticularis.
- Several of the geometric measurements used to characterize various aspects of spondylolisthesis are biomechanically useful and can be expected to predict progression of deformity.

The Sacroiliac Joint and Pubis

- The characteristics of pubic symphysis disruptions provide useful insight into both the traumatic and nontraumatic problems of this region.
- Clinical and biomechanical evidence suggests that following resection of a sizable portion of the first sacral body and its corresponding lateral structures and articulations, the patient may gradually begin to walk with clinical stability.

NOTES

[A] For exceptionally large patients, alternative values for the spine-to-film distance may be necessary. For these cases we suggest the alternative distance of 0.41 m (16 in), which will give a magnification of 29%.

[B] The first verified case of C1–C2 dislocation due to rupture of the transverse ligament was described by Bell in 1830.[13] The patient had a syphilitic ulceration of the pharynx. The postmortem examination showed rupture of the transverse ligament and compression of the spinal cord as the cause of death.

[C] The management of bilateral anterior displacement in rheumatoid arthritis is controversial. We do not recommend surgery for neurologically intact patients who have tolerable pain.

[D] A. L., a 39-year-old black male, fell down nine steps, striking the back of his head. On examination he complained of stiffness of the neck but had no neurologic deficit. Plain lateral radiographs of the neck revealed an anterior step-off of C4 on C5 of 2 mm, which increased to over 4 mm with flexion (Fig. 5-35). Observation over the course of the next 3 weeks clearly indicated that this spine was unstable as progressive anterior subluxation and widening of the interspinous space at C4–C5 occurred. Posterior interspinous fusion was required for stabilization. This patient demonstrates the fact that a spine can be unstable without immediately causing neurologic deficit. The presence of 4 mm of horizontal displacement with flexion indicated the instability. Under these physiologic conditions, a spine that permits such displacement does not have an adequate margin of safety.

[E] J. S., a 20-year-old white man, sustained a traumatic paraplegia with loss of spinal cord function at C7 and below. Plain lateral radiographs revealed angulation of C5 on C6 20° greater than the angulation of adjacent vertebrae (Fig. 5-36). Although this spine was clearly unstable, this patient was initially treated with only posterior decompressive laminectomy. Because of subsequent progression of the kyphosis at C5–C6, he required a posterolateral facet fusion to correct the instability and prevent progression of the flexion deformity.

[F] Angulation is measured as shown in Figure 5-36 at the interspace(s) under evaluation for ligament disruption.

[G] The original anterior vertebral height is estimated by determining the average height of the anterior portion of the vertebra above or below.

[H] The exceptional stability of the sacroiliac joints is due to two separate aspects of the anatomy—ligamentous and bony. The sacroiliac ligaments run from ilium to sacrum, directed medially and caudally. The weight borne by the sacrum produces tension in these ligaments. The tension may be divided into vertical and horizontal components. The vertical components support the downward weight. The horizontal components pull the sacrum equally in the left and right lateral directions, thus stabilizing it. A bending moment attempting to tilt the sacrum in the frontal plane will be effectively resisted by these stabilizing forces in the sacroiliac ligaments. The action of the ligaments is much like that of the oblique ropes tied to a flagpole for its stabilization. Seen from the bony aspects, the sacrum has two joint surfaces with the ilium. Although the surfaces are irregular, the main planes of the surfaces are slightly inclined to the sagittal plane, forming a wedge into the two iliac bones. Downward weight acting on the sacrum produces forces in the sacroiliac joint. These forces have two components, one along (frictional) and one perpendicular (normal) to the joint surfaces. Because of the small wedge angle, the frictional component is large. It is these large frictional forces that provide the bony stability to the sacrum. The wedging action of the sacrum is similar to the keystone in an archway.

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The Clinical Biomechanics of Spine Pain

I looked at the flaming clouds that hung like blood and a sword over the blue-black fjörd and city . . . and I felt a loud unending scream piercing nature.

—Edvard Munch[274]

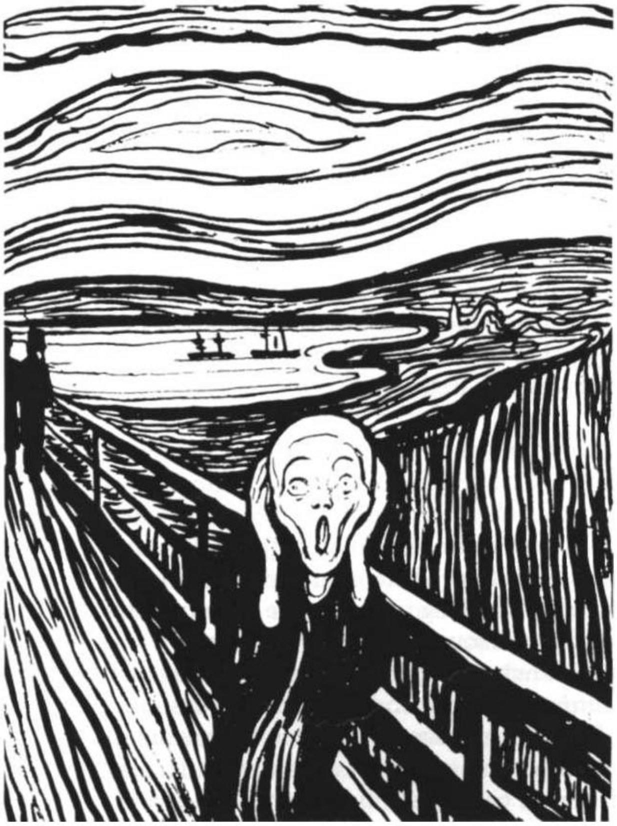


Figure 6-1. "The Scream," by Edvard Munch. 1895 Lithograph—OKK G/1 193. (Reproduced with permission from The Munch Museum, Oslo, Norway.)

6

Not many patients experiencing spine pain are able to express their pain as creatively as Munch did in "The Scream"—so physicians often must experience significantly less aesthetic aural manifestations of their patients' inner turmoil. Nevertheless, *all* patients have unique emotional and environmental factors affecting their particular response to spine disease.

The problem of pain has not only challenged the artist, poet, and physician, but it has always been an engaging topic for the psychologist, philosopher, and theologian. We may become a bit philosophical in our discussion of the many subjective factors affecting the patient in pain. Even though we are not philosophers and metaphysics is not the quest of our readers, some such non-scientific deliberation is required in order to discuss spine pain comprehensively.

In this chapter, we will present some of the major reliable clinical and biomechanical information that may be helpful to the clinician in understanding the etiology, diagnosis, treatment, and prevention of spine pain.

The term spine *pain* is used in this chapter to refer to cervical, thoracic, and lumbar pain that is not known to be related to infection, tumor, systemic disease, fractures, or fracture dislocation. The common neckaches with or without arm pain and backaches with or without leg pain that are so frequently encountered are discussed in this chapter.

It is well known that spine pain may be caused by tumor, trauma, infection, and a long list of systemic diseases. Psychologic, socioeconomic, biomechanical, biochemical, and immunologic factors also play a role. There may be any number of yet-to-be-discovered causes.

Although there are unique considerations associated with spine pain in different regions of the spine, the information in this chapter, unless otherwise stated, is meant to apply to all regions of the spine. There has been considerable similarity between the problems of neck–shoulder–arm pain and low back–hip–leg pain. They both occur in the more mobile and lordotic portions of the spine. They have similar characteristics with regard to age of onset (30–50 years of age),[A] the frequency with which they affect the population, and their typical pattern of exacerbations and remissions.[163,164] Spine pain occurs most frequently in the lumbar region, followed by the cervical region, with the lowest incidence in the thoracic region.[162,163] Horal showed that there is a significantly increased incidence of cervical and thoracic spine pain in individuals with low back disorders. Of patients who missed work because of low back pain, 50% had cervical spine pain, as compared with 38% of controls. For thoracic spine pain, the respective percentages were 23 and 17.[162]

A discussion of pain should not continue without some information about the pain-sensitive structures in the spine. The posterior annular fibers and the posterior longitudinal ligament are innervated by the sinu-vertebral nerve (Fig. 6-2).[358] The capsular structures have a sensory innervation, as do the osseous structures through the autonomic nervous system. The paraspinous muscles have a sensory innervation also. Direct spine pain can come from physical, chemical, or inflammatory irritation of any of the previously described nerves. Nerve root pain is thought to come from any of the three types of irritation to the nerve roots. Finally, there is indirect or referred pain, which is not fully explained.

ETIOLOGIC CONSIDERATIONS

The exact cause of most spine pain remains unproved. Most of the theories implicate the intervertebral disc through a variety of mechanisms, and consequently there are likely to be multiple causes. Information presented here may help to form some hypothesis about the causes of spine pain based on a synopsis of some of the most important current considerations.

The classical concepts of infectious disease etiology are based on the work of 1905 Nobel laureate Koch.[188] Three conditions known as "Koch's postulates" are necessary to establish a parasite as an etiologic agent[96,324]:

1. The parasite occurs in every case of the disease in question and under circumstances that can account for the pathologic changes in the clinical course of the disease.
2. The parasite occurs in no other disease as fortuitous and nonpathogenic.
3. After being isolated from the body and repeatedly grown in pure culture, the parasite can induce the disease anew.

Considering the concepts of Koch and others, we offer four postulates to be used to determine the etiology of what is now idiopathic spine pain:

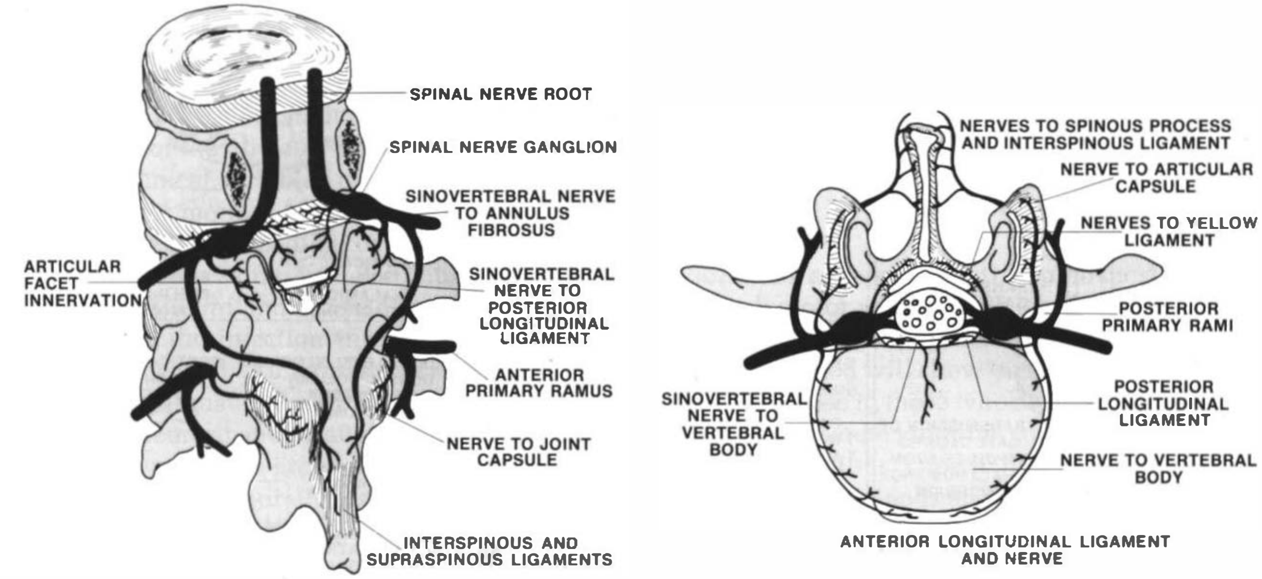


FIGURE 6-2 This drawing demonstrates clearly the sensory innervation of practically every anatomic structure in the spine. The annulus fibrosus, the major ligaments, the intervertebral joints and their capsules, the vertebral body, and all the posterior osseous structures are provided with sensory innervation. Thus, virtually any structure can be a potential source of spine pain.

1. The condition or agent occurs in every case or there is a statistically significant correlation.
2. The condition or agent occurs rarely or not at all in the absence of idiopathic spine pain.
3. Experimental induction of the condition or agent can produce spine pain.
4. Experimental correction or treatment of the condition or agent can eliminate spine pain.

Because of the experimental limitations involved in research on human pain behavior, it is extremely difficult, if not impossible, to fully satisfy all four postulates. Nevertheless, we view them as rigorous criteria worthy of consideration, certainly before one can take a dogmatic position as a proponent of some particular hypothesized cause of spine pain. The complexity of pain behavior is best appreciated through a review of the "gate" control theory presented in the following section.

"Gate" Control Theory of Pain

The "gate" control theory of pain is based on the work of Melzack and Wall.[218] Their historic article should be consulted for a full development, justification, and exposition of the concept. Although the theory is mostly unproven, it serves as a useful framework upon which to discuss pain. A freely interpreted and simplified synopsis of the theory as it may relate to spine pain follows. The essence of the theory is that within the substantia gelatinosa, several factors are able to block or facilitate the transmission of pain-producing impulses to the thalamus. The degree to which a theoretical gate is opened or closed to the transmission of these impulses depends upon blocking or facilitating influences from the cortex and/or midbrain as well as influences within the spinal cord. There are fibers of small diameter that tend to open the "gate" and facilitate pain transmission. The fibers of large diameter are thought to close the "gate." The nerves in the latter situation are thought to be involved in the mechanism of pain relief with electrical stimulations. Acupuncture may have the effect of stimulating the midbrain to send efferent impulses to close the "gate."

More of our liberal interpretation of the theory is depicted in Figure 6-3A. The known anatomic pathways are shown in Figure 6-3B, and a number of clinically recognized phenomena are listed in relation to their possible mechanism with respect to the gate theory in Figure 6-3C. The various psychosocial

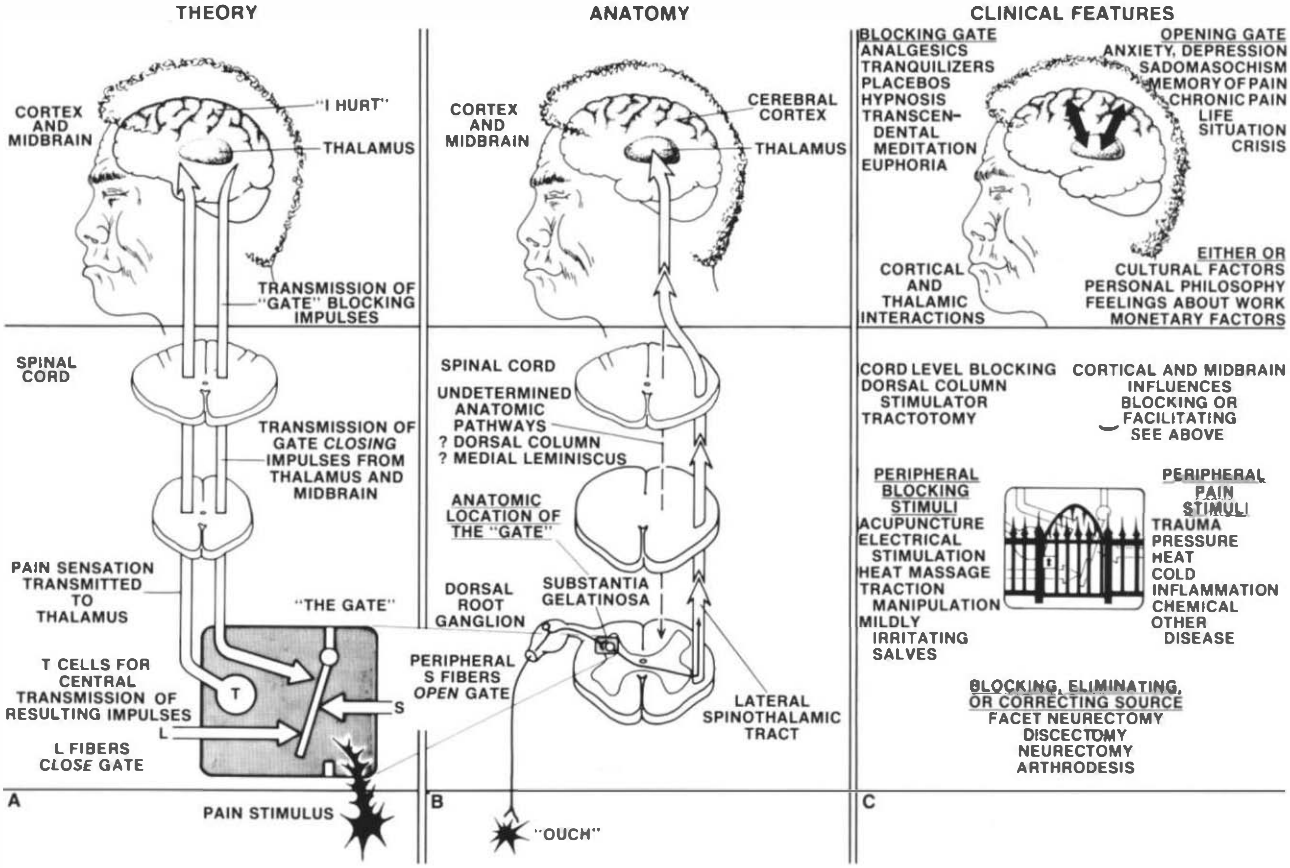


FIGURE 6-3 (A) The "gate" theory of pain. The intensity of the pain stimulus as experienced by the individual is dependent on the extent to which transmission of the stimulus is blocked. The S fibers tend to open the "gate" and facilitate the pain stimulus. The L fibers and fibers that transmit impulses from the thalamus and the midbrain tend to close the "gate" and inhibit or reduce the pain. (B) This diagram shows the *anatomy* of the key structures involved in the "gate" theory of pain. The location of the gate shown in A is in the substantia gelatinosa. The lateral spinothalamic tract is the structure that transmits the T-cell impulses to the thalamus (*descending arrow*). The specific location of the tracts through which the thalamus and midbrain exert their control on the gate is undetermined. (C) The clinical features may be viewed as an interaction between painful stimuli and facilitating and inhibiting factors that may operate through the "gate." Pharmacological, psychological, and socioeconomic factors probably have their origin in the cortex thalamus and midbrain but are to some extent mediated through the "gate," as shown in A and B. These factors also exert their influence to a considerable extent through interactions between the thalamus and cortex (*two-headed arrows*). The therapeutic value of a number of treatments, such as heat and massage, may be interpreted in the context of the theory as peripheral stimuli that work through the spinal cord and midbrain and tend to close the "gate" and protect the T cells from painful stimuli.

and cultural influences are likely to be mediated either between the cortex and the thalamus or between the midbrain and the substantia gelatinosa. In a similar manner, the influences of transcendental meditation, hypnosis, placebo reaction, and psychomimetic and analgesic drugs may be mediated here.

Biomechanical and Anatomic Factors in the Etiology of Spine Pain

If biomechanical mechanisms act as causative agents in back pain, they must provide some nociceptive stimulus to a specialized, pain-sensitive nerve ending or to the central nervous system through some other mechanism of nerve stimulation. The mechanical stimulus can be abnormal either quantitatively or qualitatively. Fig. 6-4 summarizes some of the obvious sequences that may connect mechanical variables to pain.

Abnormal motion, forces, and vibrations, high-quantity repetitive loading, or any combination of these may serve as the initial stimulus. This stimulus may cause inflammation. There is also the possibility that the infection threshold of the tissues might be lowered. Abnormal mechanics also may result in local material and structural changes and cause biochemical and/or nutritional changes in the region. Any of these groups of factors—inflammation/infection, structural/material changes, and biochemical/nutritional changes—may be causally related among themselves.

Inflammation or infection, the structural or material changes (e.g., a ruptured ligament), and the biochemical and nutritional changes may produce nociceptive sensation. There are two possible intermediate steps through which this last group may cause pain: direct chemical irritation or an immunologic mechanism.

This section is designed to review some of the most important biomechanical considerations that contribute to the understanding of spine pain.

Vibrations

Vibrations are applied to the spine most commonly through motor vehicles and less frequently through heavy vibrating equipment. Epidemiologic studies to be presented will show increased spine pain and/or disc disease in those who drive more than 3 hours per day or operate vibrating equipment such as jackhammers.[117] It is known that the resonating frequency of the spine is about 4–5 Hz both *in vitro*[361] and *in vivo*.[256] This is the vibration frequency of most automobiles. Consequently, it is hypothesized that road vibrations may be a causative agent in back pain and herniated discs. In this regard, it is also of interest that drivers of Japanese- and Swedish-made automobiles have a lower prevalence of herniated

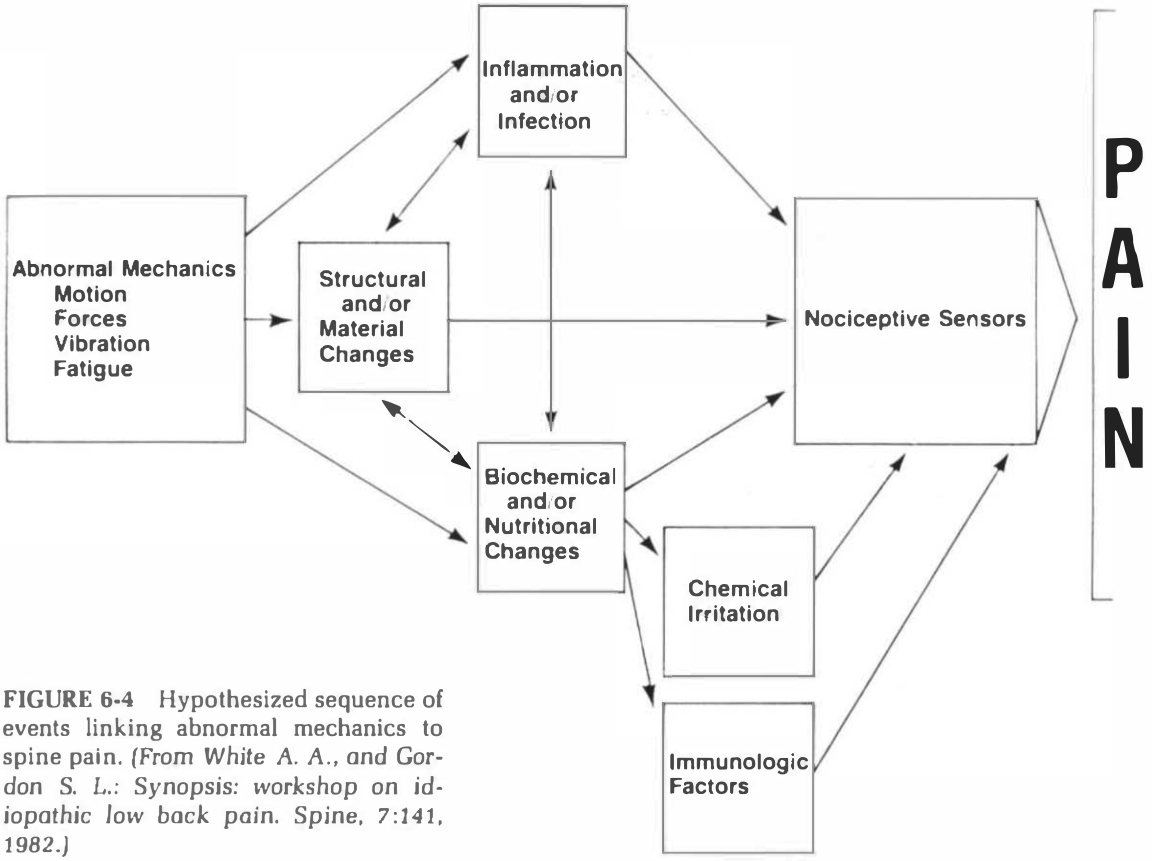


FIGURE 6-4 Hypothesized sequence of events linking abnormal mechanics to spine pain. *(From White A. A., and Gordon S. L.: Synopsis: workshop on idiopathic low back pain. Spine, 7:141, 1982.)*

lumbar discs, and the vibration frequency of these cars, unlike that of other cars, is outside the 4–5 Hz frequency range.[179] These are interesting observations with no powerful evidence as to any pathophysiologic mechanism. There is speculation that there may be a vascular mediation with vasospasm and nutritional changes resulting in disc failure and/or inflammation. This hypothesis fits with a similar mechanism to explain the association of disc disease with vascular compromise due to nicotine. Another possible mechanism for the vibration etiology could be a fatigue failure of the annulus fibers leading to disc failure and/or inflammation.

Torsion

Axial rotation (±y-axis) of the spine (twisting) is considered by some to be a possible mechanism of damage and pain to the lumbar spine. The assumption is that there may be shear loading of the annulus and/or damage to the osteocartilaginous or the ligamentous component of the facet articulations. There is agreement that the normal range of axial (y-axis) rotation in the lumbar region is 1–2° per functional spinal unit (FSU).[1,203] However, there is disagreement about the threshold for damage with axial torque. Adams and Hutton in 1981, based on *in vivo* human autopsy studies, expressed the view that the facets in torsion do not offer significant resistance. The investigators assert that the facet is the first structure to yield at the limit of torsion. They opined that because the facet joints limited torsion to 1–2°, this did not allow enough displacement to damage the disc. Therefore, they indicated that axial torque is not a great factor in lumbar intervertebral disc disease.

In contrast, Liu and co-workers in 1985[203] studied the lumbar facet joints in axial torsional loading and emphasized associated damage to the posterior elements. They described failure of facets, laminae, and capsular ligaments.

Our view is that the anterior or the posterior elements can probably be damaged or irritated by axial torque of the lumbar spine. It appears, however, that with torsional loading about the vertical axis, most of the forces, at least initially, will be borne by the posterior elements.

Lordosis

The issue of back pain and lordosis has been discussed with interest for some time. It has been stated that cultural groups that spend considerable time in the squat, lumbar *flexed* position suffer less with low back pain. Those who sit a great deal with a slightly extended (lordotic) lumbar spine have more low back problems according to this speculation. To our knowledge, there has been no satisfactory study to address this question. We have not seen even low back pain or herniated disc prevalence data in cultural groups that use the squat position extensively. Some indirect evidence of some preventive value of the squat position is found in the work of Fahrni and Gordon,[98] who conducted a radiographic study of degenerative changes in the lumbar disc while comparing a primitive squatting with a nonprimitive sitting society. There was a lower incidence of radiographic evidence of disc degeneration in the squatters compared with the sitters.

The biomechanics of lumbar lordosis and back pain are important for several reasons. There is concern that with hyperlordosis there could be excessive loading of the facet joints or posterior bulging of the disc with nerve root and/or posterior longitudinal ligament irritation. It has also been suggested that wearing high heels causes increased lumbar lordosis and back pain through one of the mechanisms described above. Some individuals have a naturally exaggerated lumbar lordosis, and there is a question of this being the cause of pain. There are also considerations on the therapeutic side, where some favor voluntary muscle stabilization of the spine to avoid lordosis and others advocate extension exercises to place the spine in lordosis in hopes of shifting the nucleus and decreasing posterior disc bulging. Although definitive answers are not available to all these questions, an updated review of the most cogent studies is presented.

Let us begin by looking at some of the studies that address the issue of lordosis as a source of or contribution to low back pain. Frymoyer and colleagues reported an extensive study of 292 males in 1985 in which the investigators correlated low back pain with various radiologic findings. There was no association of lumbar lordosis with low back pain.[119] Ferrand and Fox also found no association of lumbar hyperlordosis (defined as >63°) or hypolordosis with back pain. They also reported no differences between blacks and whites in lumbar lordosis but did observe significantly greater lordosis in females as compared with males.[103] In regard to high heels and lumbar lordosis in females, Bendix and colleagues have shown that increasing heel heights up to 4.5 cm actually decreases rather than increases lumbar lordosis.[21] There does not appear to be evidence to support the view that hyperlordosis is associated with back pain.

There are some cogent observations on the significance of the secondary changes in the facet joints and the outer vertebral discs associated with spine extension. Dunlop and colleagues studied disc space narrowing and the effects on lumbar facet joints.[83] They note that the female is more extended, and, according to these calculations, each degree of increased extension leads to a 4% increase in peak articular pressure in the facet joints as well as extra-articular impingement. This is offered by the investigators as a possible explanation of the existence of relatively more facet joint problems in women. Yang and King[374] reported that arthritic facet joints may bear up to 47% of the load transmitted to an FSU. There is an abnormal rearward rotation of the inferior facet articulation and damage to the facet capsule, which can cause low back pain. Adams and Hutton[3] studied the effects of posture on the lumbar spine and concluded that the flexed lumbar postures have better metabolic transport to the disc, reduce stress on the apophyseal joints, and give the spine a high compressive strength. Although there were also some disadvantages, they concluded that for standing, sitting, and heavy lifting, a *flexed* lumbar spine is preferred.

The available evidence does not support any strong conclusions. There do appear to be disadvantages to hyperextension if there is facet joint arthritis or disc narrowing. It seems that women should not be forbidden high heels because of fear of damaging or irritating their lumbar spine, and clearly there is no rationale in the mechanics presented here to suggest that extension exercises are therapeutic for low back pain. This does not imply that strengthening the erector spinae is not useful.

Cervical Spine

An important epidemiological study by Kelsey and associates[180] addresses the issue of acute cervical intervertebral disc prolapse. There were several statistically significant findings. There was a greater prevalence among smokers, materials handlers involved in frequent heavy lifting, those who dove frequently from diving boards, and those between 40 and 50 years of age. The 4 : 1 male : female predominance was also statistically significant.

There was a nonstatistically significant finding of a greater prevalence of disc prolapse in those who spent a great deal of time in motor vehicles as well as those who drove heavy equipment. Not related were: other sports, the number of pregnancies or live births, frequent twisting of the neck on the job, and time spent sitting on the job.

Epidemiologic Factors

It is generally helpful in searching for the cause of a disease to study certain characteristics of a large group of individuals with the disease. In other words, how is spine pain related to age, sex, occupation, socioeconomic status, weight, or any other observable characteristic that can be studied? We have updated this epidemiologic review by including the most cogent new studies and placing an emphasis on those fractures which may relate to biomechanical variables.

Carefully designed and executed studies of the epidemiology of lumbar intervertebral disc diseases were conducted in New Haven, Connecticut.[178, 177, 181–183] Some of the major observations are presented below.

Although the trends are not always consistent, it is interesting to discuss some of the mechanical

RELATIONSHIP OF CERTAIN EPIDEMIOLOGIC FACTORS TO LUMBAR DISC DISEASE OR LOW BACK PAIN*

Well-Established Risk Factors

Driving motor vehicles, especially trucks and heavy equipment
Male gender
Smoking, especially with chronic cough
Materials handling: standing, lifting, carrying, pulling, pushing, twisting, bending

Suggestive but Inconclusively Related Factors

Use of jackhammers
Cross-country skiing, jogging
Emotionally stressful occupations
More than two full-term pregnancies
Nursing (nurses and nurse's aids)
Sedentary occupations
Leg length discrepancy[113, 128]

Factors Not Related to Any Increase in Risk

Race
Baseball, golf, bowling, tennis
Swimming, bicycling
Height, weight

* Based on updated review of the literature cited in text.

considerations that may be operative in these findings. With regard to driving motor vehicles, the data suggest that men who spend 50% or more of their work time driving a motor vehicle are three times more likely to develop a herniated disc than someone who does not have such an occupation.[182] Men or women who drive at work or elsewhere for more than two hours per day are more likely to develop a herniated disc than those who do not. More recent studies confirm this observation in the work setting as well as outside it.[117,120,179] It is known that sitting puts more pressure on the intervertebral disc.[172,227,228] This, in addition to the schedule of vibratory forces that are transmitted to the spine, may be a possible mechanism.[137] In addition, the position of the legs and the limited variety of optional sitting positions available to the driver may result in a predisposition to disc herniation. The study also showed that individuals with sedentary occupations were at significant risk of developing disc problems.[177]

However, driving a motor vehicle, particularly a truck, because of the frequently associated materials handling, cannot be considered sedentary activity.

The suggestion of full-term pregnancy as a risk factor may be explained on the basis of the hormone relaxin and the increased load on the disc structures imposed by the increased weight of the uterus and its contents.[181] A study of 347 patients who had given birth to one or more children revealed that 39% of the women developed symptoms of back pain and possibly sciatica either during pregnancy or during the puerperium.[247] The data in Kelsey's study suggested that the causative factor was related to the pregnancy rather than to the care of the children after pregnancy. Consequently, the lifting of children may not be an important consideration.[177] Kelsey and associates, in a more recent study,[179] found that the number of pregnancies was not associated with acute prolapsed disc as a positive finding or risk factor. However, it is important to keep in mind that with altered, diseased, or irritated discs, the application of any incremental loads cannot be expected to be desirable or helpful.

It is interesting to find that acute herniated lumbar disc disease was found to be a risk factor associated with sedentary occupations. However, this has not been borne out by subsequent epidemiologic studies in the industrial setting. These are reviewed in a subsequent section on industrial low back pain. There are, however, significant risks associated with jobs involving materials handling. These are described below in the section on biomechanics of low back pain in industry.

The 1975 Kelsey study also showed a correlation between weekend sitting and disc herniation in males.[177] Goodsell studied 402 consecutive operations, and from this review he expressed the opinion that heavy labor predisposes to disc rupture.[131] Other investigators have expressed the view that heavy labor is not a significant factor.[114,164] A controlled epidemiologic study of 429 subjects, divided into eight preselected occupational categories, provides some relevant data. Interestingly, the incidence of spine pain, not necessarily disc herniation, was positively correlated with the subjects' subjective evaluation of the type of work they were performing. There was more low back pain in subjects who thought their work was physically demanding[207] as well as emotionally stressful, anxiety provoking, or demanding.[120,172,173] It was found that a patient complaining of acute low back pain often gives a history of making a sudden unexpected exertion while carrying a heavy object.[208] The study also showed a significant association of low back pain with sitting and lifting weights when the spine is flexed as opposed to proper lifting when the spine is straight with hips and knees flexed. This observation fits with the experimental and simple biomechanical modeling data of Nachemson and Troup.[232,335]

A study by Magora and Taustein showed that the subjects in the following occupations were most likely to experience low back pain at an early age: bank clerks, workers involved in heavy industry, farmers, and nurses.[209]

Comments

In some studies, data are available on specifically herniated disc disease; in others, the data are simply related to back pain with or without sciatica. Nevertheless, there appear to be some general trends. A person between the ages of 30 and 40 who is involved in heavy labor is likely to get a herniated disc. Individuals who spend a good deal of time driving trucks or automobiles are likely to have back pain with or without sciatica. Women in the later stages of pregnancy or in the early postpartum periods are prone to spine pain. The reader interested in a more detailed study of the epidemiology of spine pain is referred to comprehensive review articles by Kelsey[178] and Frymoyer and associates.[117,120]

Many of these problems may be related to a mechanical etiology. The intervertebral disc, between the ages of 30 and 50, is changing from one with a rather healthy, resilient, high water content to a relatively dry, scarred disc characteristic of individuals over 50. The well-hydrated, resilient disc under age 30 and the dry, scarred disc over age 50 may be mechanically less likely to fragment and displace. Obviously, other variables during these stages may also cause pain. Either the position of the spine or the patterns of the forces applied to it in driving a motor vehicle may well cause pain. A person whose job involves heavy labor, especially if he lifts improperly, can exert considerable forces on the structures of the spinal column, resulting in mechanical failure and/or pain. When the ligamentous structures of the pelvis and lower spinal segments alter their physical properties, mechanical disruption and pain may be the result.

There also may be a hereditary predisposition for a variety of different conditions that may lead to spine pain. This has been shown to be the case for spondylolisthesis and to a lesser extent for intervertebral disc disease.

The data on occupational and epidemiologic factors in back pain and disc disease are difficult to summarize. The more definitive relationships depend upon a clearer elucidation of the causes of low back pain and disc herniation. However, given present knowledge, it seems fair to suggest that driving a motor vehicle, smoking, and materials handling are related to spine problems. In addition, although it is not consistent, sedentary and emotionally stressful occupations may be associated, but the relationship is not distinct.

Biomechanics of Low Back Pain in Industry

This section reviews some of the most important studies involving the cause and prevention of back pain in industry.

Causes

The high-risk workers have been appropriately identified as truck drivers, materials handlers, and nurse's aides, as well as nurses. Truck drivers have been identified in several studies as being at risk to develop back pain.[47,371]

Kelsey and Hardy showed that male truck drivers are about five times more likely to develop an acute herniated lumbar disc than are non–truck driver males.[182] Prolonged sitting, which imposes significant (greater than when standing) loads on the lumbar discs,[231] and road vibrations[264,361] are thought to be the important variables.

Materials handlers are another industrial high-risk group likely to develop low back pain.[28,69] Several studies of materials handlers support the suggestion that, in decreasing order of frequency, the following tasks are associated with the onset of low back pain. They are: *lifting, pulling, pushing, carrying,* and *lowering.*[45,303,304] We note that in the analysis, pulling was more frequently associated (9–16%) with the onset of low back pain than was pushing (6–9%). This certainly fits with the free-body analysis presented in Figure 6-55 (p. 459). We note that *bending, twisting,* and *falling* or *slipping* are also significantly related to the onset of low back problems.[45,303,304]

Nurses and nurse's aides constitute another group reported to have a high prevalence of low back problems.[47,283,371] These workers are actually materials handlers but in the unique sense that the material they handle is the patient.[340] There is the added factor of having to lift frequently in an awkward and sometimes sudden fashion. Although we are not aware of particular studies, our clinical experience suggests that there is a high prevalence of back pain in emergency medical technicians, perhaps for the same reason of frequent and awkward lifting. We note that studies from Sweden[72] and Britain[88] do not support the observations of a higher incidence of back pain in nurses.

Prevention Measures for the Industrial Back Problem

Rowe[282] suggested that a prior history of back pain indicated a greater probability of developing low back disability. It has been suggested that lumbar spine films not be used as a routine screening procedure for back problems.[8] However, preplacement strength testing seems to be a promising screening technique. Research at Harvard University[185,357] and the University of Michigan[51] has shown that if the lifting requirements of a job exceed the strength capability (as determined by isometric job simulation tests), the likelihood of sustaining a musculoskeletal injury can be as much as three times greater. Because of both legal and methodologic reasons, the test of strength capability must be specific for the job to be done by the aspiring worker.

Another strategy for prevention involves worker training and education programs. There has been a long tradition of emphasizing "squat" or *correct* lifting.[147,240] The teaching has been to lift with the legs while keeping a straight back. Unfortunately, neither epidemiologic analyses[38,304] nor biomechanical studies[122] substantiate the efficacy of the squat lift in preventing back injuries. This lifting technique places the quadriceps muscle at a significant mechanical disadvantage and also requires more energy than lifting with the back.[38,122] Given current knowledge, probably the best advice for the worker is to keep the *object close to the* body and to lift slowly and smoothly while avoiding twisting. Finally, it is very important to maintain good physical fitness.[46]

It is also appropriate to discuss the ergonomic approach, which can be defined as designing the work situation to suit the worker. Snook and associates[304] concluded that for manual handling tasks, proper design can reduce low back injuries by 33%. They note that inexpensive changes such as raising a work surface or changing a container size may reduce back injuries. One final precautionary note is based on a study that showed that while materials handling is an important variable, unsafe floor surfaces at home as well as at work can be a contributing factor.[211]

Studies show that once injury has occurred, systematically monitoring the medical management results in better care, shorter disability, and significantly less expense.[198,360] Careful follow-up and monitoring has added import in view of the observation that there is a 50–60% recurrence rate of industrial low back pain within the first year.[9]

Socioeconomic and Psychologic Factors

These factors obviously overlap considerably with epidemiologic factors; however, we have chosen to discuss them separately. Studies have shown that the tendency to report sick from work with spine pain is correlated with lower intellectual capacity, educational level, socioeconomic status, and the patient's own idea of "self-importance" on the job. Patients who missed work because of low back pain or sciatica tended to have subordinate positions and to be less satisfied with their work.[326,327] Moreover, parameters of social insufficiency, such as divorce rate, alcoholism, and various psychiatric disturbances, all tended to correlate with missed work and spine pain.[30,352]

It is well known that the complaint of pain or onset of illness may be significantly influenced by psychologic factors.[36,93,94,157,368] The secondary gain may be relief from guilt, responsibility, or challenge, or the pain may be simply a manifestation of depression. Some of the recent psychiatric theories have generated evidence that a sadomasochistic patient may complain of spine pain in search of a surgical procedure.[29]

A recent study by Leavitt and colleagues[197] compared patients on compensation with a group not on compensation. When objective evidence of disease and psychologic stability were evaluated in the two groups, there was little difference. The only difference was that the compensation group used 43% more words to describe the quality of their pain. The study offers little or no justification for the suspicion that surrounds patients on compensation who have no evidence of organic disease. The overall issue is more complicated than this. Nevertheless, this study suggests the use of caution in these judgments.

Spine Pain and Vertebral Pressure Dynamics

Many clinicians accept the hypothesis that hemodynamic abnormalities, mainly an increase in blood pressure in bone, can result in pain. It has been suggested that the success of just about any geometric configuration of osteotomy of the hip in diminishing pain may be the result of the relief of internal osseous pressure.[16] Arnoldi compared the intraosseous venous pressures in the spinous processes of 43 vertebrae from ten subjects. The mean pressure was 8.3 mm Hg. Twenty-two vertebrae from ten subjects with low back pain had an average venous pressure of 28.0 mm Hg. The difference was statistically significant ($p < 0.01$).[14]

More recent work has shown that the lumbar vertebral intraosseous pressure is equal to and dependent upon the pressure in the inferior vena cava.[312] Another interesting study compared lumbar intravertebral pressure with lumbar cerebrospinal fluid (CSF) pressure.[140] These two measurements showed the same change patterns with different positions. The pressures were lowest in the prone position and highest in standing. In extension they were even higher than in standing, and in flexion they were as low as when lying prone. Patients studied had disc disease or spinal stenosis.

These findings have some neat implications in several clinical settings. With regard to clinical signs, the potential irritant effect of extension is obvious in the presence of disc disease and spinal stenosis. Vascular supply as well as pressure and inflammation may be further compromised in the spinal stenosis patient who experiences exacerbation of symptoms with extension. Space, more precisely cross-sectional area, is also further compromised by invagination of the yellow ligament and posterior disc bulge. In other words, to the extension test for spinal stenosis we may add an increased CSF pressure. In disc disease and sciatica, extension may cause CSF pressure increases, which may irritate the most proximal portion of the nerve root still within the subarachnoid space. Here, too, increased posterior disc bulge may be a contributor.

On the therapeutic side, both of these disease conditions would benefit from the significantly reduced CSF pressure that accompanies the slightly flexed position associated with bicycling. Clearly, extension exercises imposed in this situation could be a liability.

Soft-Tissue Structures and Spine Pain

Spine pain has been attributed to trigger points in the skin, and relief has been reported from injections of the same. Back pain has been reported to be cured following resection of portions of deep fascia over the paravertebral muscles thought to be responsible for pain.[225]

Spasm of the muscles themselves is generally thought to be either a primary or a secondary source of back pain. This has resulted in the sale of a great quantity of muscle-relaxing drugs and extensive use of massage administered by physical therapists and others.

Any of the numerous musculotendinous or ligamentous structures of the spine may suffer strain, sprain, or rupture. This may result in pain and inflammation and may be a stimulus for paraspinous muscle spasm, a cause of considerable spine pain. The list below, based on the work of Wyke, names the various structures of the spinal column that are known to have pain receptors.[373] There are nerve fibers capable of transmitting pain present in the lumbodorsal fascia, the supraspinous and infraspinous ligaments, the vertebral periosteum, and the anterior and posterior longitudinal ligaments, as well as the outer layers of the yellow ligament and posterior annulus fibers.[152]

SOURCES OF PAIN RECEPTORS IN THE VERTEBRAL COLUMN

Anterior and posterior longitudinal ligaments
Posterior annular fibers
Yellow ligaments
Interspinous ligaments
Intervertebral joint capsules
Periosteum of vertebrae
Fascia of vertebrae
Blood vessels of the vertebrae
Walls of epidural and paravertebral veins
Paravertebral musculature

(Wyke, B.: The neurological basis of thoracic spine pain. Rheumatol. Phys. Med., *10*:356, 1970.)

Posterior Elements

The intervertebral joints are cartilage-covered articulations with a synovial and a fibrous capsule. Any disease of the cartilage or the synovial tissue can affect this joint as well as any other in the body. Ankylosing spondylitis and degenerative arthritis tend to involve the joint more often than some other diseases. These joints may be the source of a significant amount of spine pain. Hirsch and colleagues reported that injections of hypertonic saline in either the posterior annular fibers or the intervertebral joint areas produced similar clinical presentations.[151] However, when the annular fibers were injected, the clinical presentation was more characteristic of the lumbago seen spontaneously in patients. The role of the posterior elements in low back pain is also presented under lordosis on page 384.

The Etiologic Relationship of Some Radiologic Findings to Spine Pain

Which of the many radiologic irregularities seen in the spine can be presumed to cause spine pain? In order to convincingly demonstrate this, it is necessary to show that patients with the particular radiographic irregularity have a significantly higher incidence of spine pain than individuals without the irregularity. This is no easy task, given all the subjectivity of complex variables involved in the complaint of pain and given the fact that about 80% of the population at one time or another will have a complaint of back pain.[232]

The valid correlation of plain x-ray findings with low back pain remains a major challenge to the clinician. There are several studies on this topic. Frymoyer and associates in a study of 292 males noted an increased incidence of low back pain in patients with L4–L5 so-called traction spurs and/or disc space narrowing.[119] A study compared the x-rays of 238 patients with back pain and sciatica with 66 patients without pain. No differences in x-ray evidence of spondylosis or disc degeneration were found.[372] Biering-Sorensen and associates analyzed x-rays of 666 subjects and reported that disc degeneration was significantly more common among those subjects with low back pain.[27] The severe back pain sometimes observed in idiopathic vertebral sclerosis[356] may be a special example of severe disc degeneration causing low back pain.

The possible role of transitional vertebrae in back pain remains controversial. Frymoyer and colleagues found no association. However, Castellvi and co-workers[49] reported that in their classification, the Type II transitional vertebrae were associated with an 11% incidence of disc herniation at the transitional level and an 83% incidence at the level above. These important observations deserve further study. Type II involves incomplete lumbarization/sacralization. There is a large transverse process that appears, on anteroposterior x-ray of the lumbosacral spine. One or both of the transverse processes form a diarthrodial joint with the alar of the sacrum.

The radiographic conditions are listed below in three groups according to the probability of an association with spine pain (very likely, questionable, very unlikely). The information comes largely from the review article by Nachemson,[232] with some updates substantiated from a variety of more recent sources. This discussion is based on the assumption that there is no other clinically obvious explanation for spine pain.

ASSOCIATION OF RADIOGRAPHIC IRREGULARITIES OF THE SPINE WITH SPINE PAIN

Very Likely

Spondylolisthesis (moderate or severe)
Multiple, markedly narrowed intervertebral disc spaces
Congenital kyphosis
Scoliosis (severe)
Osteoporosis
Ankylosing spondylitis
Lumbar osteochondrosis (Scheuermann's disease)

Questionable

Spondylolysis
Spondylolisthesis (mild)
Kyphosis (severe)
Scoliosis (mild to moderate)
Retrolisthesis of cervical, thoracic, or lumbar vertebrae
Lumbar scoliosis (>80°)
L4–L5 disc space narrowing and/or traction spurs
Idiopathic vertebral sclerosis
Type II transitional lumbosacral vertebra

Very Unlikely

Spina bifida occulta
Acute lumbosacral angle
Facet arthrosis, subluxation, and trophism
Disc calcification (except in thoracic spine)*
Extracervical, extralumbar, or extrathoracic vertebrae
Hyperlordosis
Intravertebral body disc herniation (Schmorl's nodes)
Accessory ossicles
Transitional lumbosacral vertebra (other than Type II)

(Data from Nachemson, A. L.: The lumbar spine, an orthopaedic challenge. Spine, 1:59, 1976.)

* Disc calcification in the thoracic spine should raise a high index of suspicion of a herniation.

A careful review of this list does not seem to reveal any patterns in the radiologic findings that might be related to some common factor, mechanical or otherwise. Perhaps its main value is to provide some helpful guidelines for the proper interpretation of the findings in the first two groups in the list. It is also important to keep in mind that there can be marked disc degeneration in the absence of *any* radiographic changes.[152] When there is radiographic evidence of decreased disc space, sclerosis, and osteophytes, the corresponding disc is severely damaged but may not be painful. The plain radiographic findings are the same for disc degeneration as for disc herniation.[114]

The Intervertebral Disc

Ever since the milestone investigation by Mixter and Barr, most of the clinical and research work on spine pain has focused on the intervertebral disc.[222] Actually, the disc has been proved to be the cause of pain in only a very small percentage of patients. Many physicians agree that it has probably been overstudied and overrated as a cause of spine pain. Even though there are still a number of questions to be answered, there is a fund of anatomic, biomechanical, biochemical, immunologic, and clinical information on this structure. These data are valuable and useful. Other possible causes of spine pain have not been so vigorously studied. It is possible that the nonherniated intervertebral disc is the cause of much of the severe, clinically significant low back pain that is observed, but this may be difficult to prove. In support of this hypothesis, it is known that there is neural innervation to posterior annulus fibers and that the clinical pattern of low back pain tends to precede the herniated intervertebral disc. In opposition to this hypothesis, it is known that the spine pain that precedes herniation often has a course of gradual exacerbations and remissions that somehow belies a purely mechanical explanation. This phenomenon is not satisfactorily explained by movement of the disc in and out of or into and away from the area of sensory innervation. This may be explained on a biochemical or immunologic basis, but current evidence is unconvincing and contradictory. The physician must accept that spine pain associated with nerve root irritation can be caused by disc disease. Whether or not the disc can account for a significant portion of other spine pain remains in question.

In addition, a large number of other diseases and conditions are known to be associated with radiculopathy, with or without spine pain. Most of these conditions are rare; however, they are important. Before focusing in detail on the disc, it is worthwhile to scan the list below.[5,178,193]

Normal Disc

This discussion refers to the normally functioning disc in a young individual, described in Chapter 1. The external loads create tensile forces in the peripheral annular fibers that are up to four to five times the superincumbent forces. There is a slight physiologic protrusion of the disc on the concave side of a physiologic curve, with a slight shift of the nucleus pulposus in that direction. Large tensile stresses are also applied to the peripheral annular fibers with torsional loading (y-axis rotation). This is thought to be especially true of the posterolateral annular fibers. As long as the nucleus is well hydrated, the annular fibers are well nourished, and no irritating or immunologically active chemical organic substances are present, the intervertebral disc is healthy and causes no pain.

CONDITIONS OTHER THAN INTERVERTEBRAL DISC DISEASE KNOWN TO BE ASSOCIATED WITH RADICULOPATHY

- Osteoarthrosis of the spine
- Intraspinal tumors
- Other tumors
- Epidural venous anomalies
- Spondylolisthesis
- Rheumatoid spondylitis
- Generalized toxemia
  - Alcoholism
  - Lead poisoning
- Radiation radiculopathy
- Diabetes
- Syphilis
- Sarcoidosis
- Behçet's disease
- Intraspinal synovial cysts
- Poliomyelitis
- Herpes zoster
- Diphtheria
- Meningitis
- Leprosy
- Tuberculosis
- Meylomeningocele
- Perineural cysts
- Extradural or subdural cysts
- Root avulsion
- Megacauda
- Widening of nerve root socket
- Abnormal anatomic location of nerve root
- Facet entrapment syndrome

(Agnoli, A. L., et al.: Differential diagnosis of sciatica. Analysis of 3000 disc operations. *In* Wöllenweber, R., et al. (eds.): Advances in Neurosurgery, vol. 4. New York, Springer-Verlag, 1977. Kelsey, J. L., and Ostfeld, A. M.: Demographic characteristics of persons with acute herniated lumbar intervertebral disc. J. Chronic Dis., 28:37, 1975.)

Pathologic Problems

As a normal result of the process of use and aging, several phenomena may take place in the disc. The water content of the disc diminishes, and the ability of its component structures to be nourished may be altered. There may be fatigue failure of some of the annular fibers, which may undergo a variety of different degenerative and chemical decompositions.[125] The products of decomposition may be protein substances that stimulate immunologic responses and inflammatory activity.[30]

Other more purely mechanical possibilities relate to the peripheral annular fibers, which may rupture from either fatigue failure or some particular traumatic episode. Radical bulge and tangential strain of the lumbar disc are maximal at the posterior lateral surface.[293] This is where most herniations occur. It is known that sometimes episodes of a sudden unexpected load cause the onset of acute spine pain. This may be due to the sudden rupture of some of the fibers of the annulus fibrosus. As these various phenomena occur, they may be associated with a variety of clinical findings.

In 1955, Charnley wrote a stimulating and provocative paper in which he sought to describe the mechanisms of intervertebral disc pathology and to correlate the hypothesized pathoanatomic factors with the clinical presentation and treatment of acute low back pain and sciatica.[52] Here we attempt to update his hypotheses.

Acute Back Sprain (Type I)

This is the acute back sprain that characteristically occurs when a laborer attempts to sustain a sudden additional load. There is immediate severe pain that may last for several weeks. The pain is primarily in the low back, without sciatica. This may be due to several factors. Charnley suggested the possibility of rupture of some of the deep layers of the annulus. We believe that while this is possible, the inner fibers are not innervated, and there is relatively less loading and deformation of the deeper fibers than of the periphery. There are several other possibilities. One is that peripheral annular fibers may be injured or ruptured along with any of the other posterior ligaments or musculotendinous structures. Also, there is the possibility that some of these injuries may involve rupture of muscle fibers or may be associated with nondisplaced or minimally displaced vertebral end-plate fractures (Fig. 6-5). The answer awaits further investigation. These conditions should respond to a period of rest, followed by a gradual resumption of normal activities.

Fluid Ingestion: Organic or Idiopathic? (Type II)

It was hypothesized that an attack of low back pain and muscle spasm may be produced by the sudden passage of fluid into the nucleus pulposus for some unknown reason (Fig. 6-6).[52,241] Charnley suggested that this irritated the peripheral annular fibers, causing the characteristic pain. There is little to discredit the hypothesis. Naylor suggests that increased fluid uptake in the nucleus is a precipitating factor in the biochemical chain of events that can lead to disc disease. Indirect evidence suggests that increases in fluid in the disc structure may not cause spine pain. This is based on the observation that astronauts returning from outer space have heightened disc space and spine pain. There are data, although inconsistent, that suggest that fluid injection into the normal disc causes low back pain.[148] This discrepancy may be partially explained by the differences in the *rate* of change in fluid pressure. The hypothesis of fluid ingestion fits with clinical data, because it is compatible with the characteristic clinical course of exacerbations and remissions, with or without progression to other clinical syndromes. In other words, movement of fluid in and out of the disc can explain the onset and resolution of the clinical symptoms. We suggest that this may be the explanation for spontaneous idiopathic organic spine pain (cervical, thoracic, or lumbar) unrelated to trauma, which accounts for a significant number of the many cases of spine pain.

Posterolateral Annulus Disruption (Type III)

If there is failure or disruption of some of the annular fibers, posterolateral irritation in this region may cause back pain with referral into the sacroiliac region, the buttock, or the back of the thigh (Fig. 6-7).

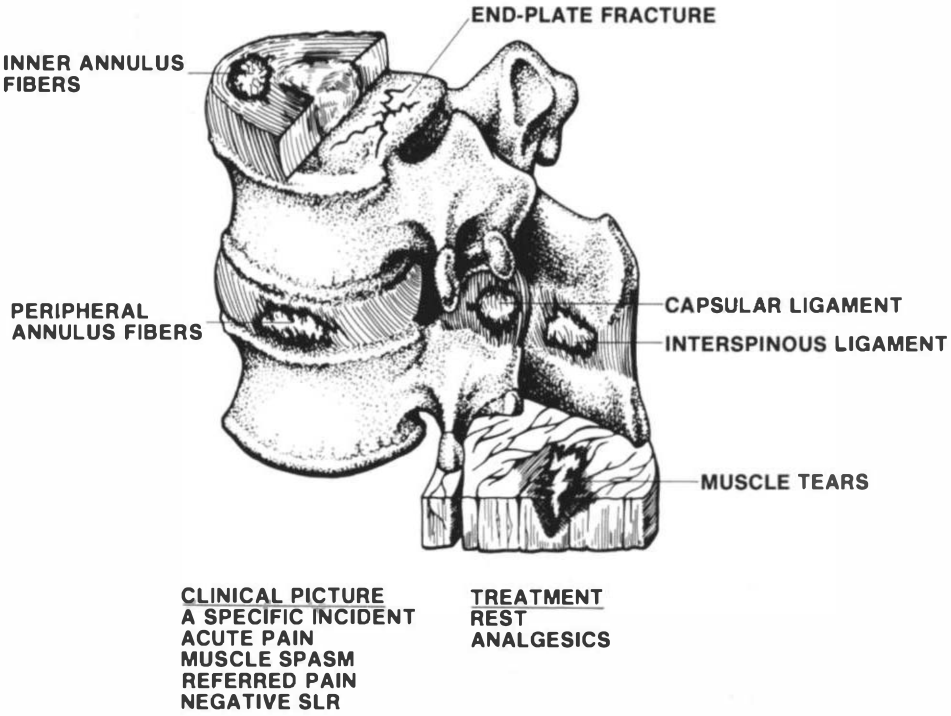
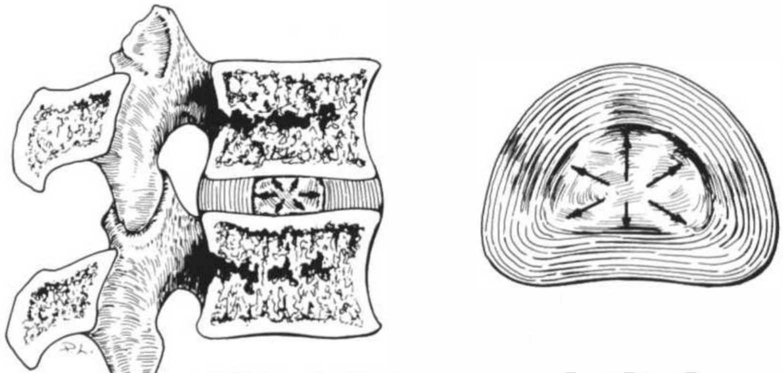


FIGURE 6-5 A clinical picture of acute back sprain (Type I) may involve damage to any number of ligamentous structures, the muscle, or even vertebral end-plate fracture. (*SLR*: straight leg raising test.)

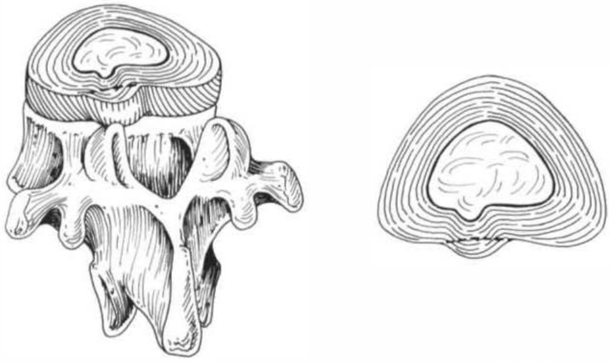


CLINICAL PICTURE
BACK PAIN
MUSCLE SPASM
NO REFERRED PAIN
NO SCIATICA
NEGATIVE SLR

TREATMENT
BED REST
ANALGESICS

FIGURE 6-6 Organic or idiopathic fluid ingestion (Type II). This mechanism may account for a large portion of back pain for which no distinct diagnosis or etiology has been determined. (*SLR*; straight leg raising test.)

This is referred pain and is due to stimulation of the sensory innervation by mechanical, chemical, or inflammatory irritants. Thus, "referred sciatica," as Charnley called it, is distinguished from true sciatica by a negative straight leg raising test and a lack of any neuromuscular deficit. As suggested previously, this referred pain may be explained by the "gate" control theory. This situation may resolve itself through reabsorption or neutralization of the irritants and/or phagocytosis and painless healing of the disrupted annular fibers.



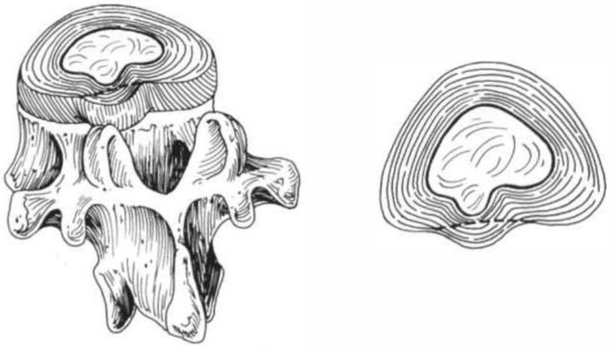
CLINICAL PICTURE
BACK PAIN
HIP, UPPER LEG PAIN
NEGATIVE SLR

TREATMENT
BED REST
ANALGESICS

FIGURE 6-7 Posterolateral annulus disruption (Type III). The dotted line represents theoriginal normal contour of the disc. Hip and thigh pain are referred pain rather than true sciatica. (*SLR*; straight leg raising test.)

Bulging Disc (Type IV)

Another proposed mechanism involves protrusion of the nucleus pulposus, which remains covered with some annular fibers and possibly the posterior longitudinal ligament (Fig. 6-8). There may be "true



CLINICAL PICTURE
BACK PAIN
INCREASED WITH COUGHING
AND SNEEZING
TRUE SCIATICA
POSITIVE SLR

TREATMENT
REST
ANALGESICS
TRACTION
MANIPULATION
PERCUTANEOUS
DISCECTOMY

FIGURE 6-8 Bulging disc (Type IV). In this situation, the annulus is bulging to such an extent that nerve root irritation has caused sciatica. The dotted line shows the normal position of the annulus rim. This type of disc herniation theoretically is amenable to chemonucleolysis or percutaneous discectomy because a part of the herniation is in continuity with the central nucleus pulposus. (*SLR*; straight leg raising test.)

acute sciatica" with mechanical and possibly chemical and/or inflammatory irritation of the nerve roots. The pain may include the back, buttock, thigh, lower leg, and even the foot. The pain may be increased with coughing and sneezing, and the straight leg raising test is positive. Radiographs in this situation usually do not indicate narrowing. It is feasible that traction or spinal manipulation may alter the mechanics in this situation and may possibly be therapeutic. With rest, the irritation may subside and remain stable or return spontaneously after mobilization.

When this kind of disc pathology has persisted and is causing pain and radiculopathy, there is an excellent chance for success with chemonucleolysis or percutaneous discectomy. A mechanism of injury that applies to this and other types of disc prolapse has been suggested by Adams and Hutton to be a hyperflexion injury.[2] The most vulnerable disc is thought to be a slightly degenerated one at the lower lumbar level. This is further discussed in Chapter 1, under Mechanisms of Disc Prolapse.

Sequestered Fragment (The Wandering Disc Material; Type V)

A theorized mechanical etiology is that of a sequestered nucleus pulposus and/or annulus fibrosus (Fig. 6-9). This may develop over a period of time associated with the normal degenerative processes of the disc and/or other presently unknown pathologic changes. This sequestrum may move about in a random fashion in response to the directions and magnitudes of forces produced at the FSU by the activity of the individual. This movement may permit the sequestrum to irritate (by physical presence and/or chemical breakdown products) the annular fibers and to produce low back pain with or without referred sciatica. It may also produce a bulge in an area in which it can cause true sciatica. The sequestrum may move about such that in some positions it is either asymptomatic or causes some combination of spine pain, referred pain, and true radiculopathy. Because of the movement of the sequestered fragment in response to forces at the FSU, it may be possible, through axial traction or spinal manipulation of the FSU, for the sequestrum to be moved temporarily or permanently from a location in which it stimulates a nerve to one in which it causes no irritation. Subsequent motion of the disc fragment into areas of pain insensitivity or subsequent scarring may result in no recurrence. On the

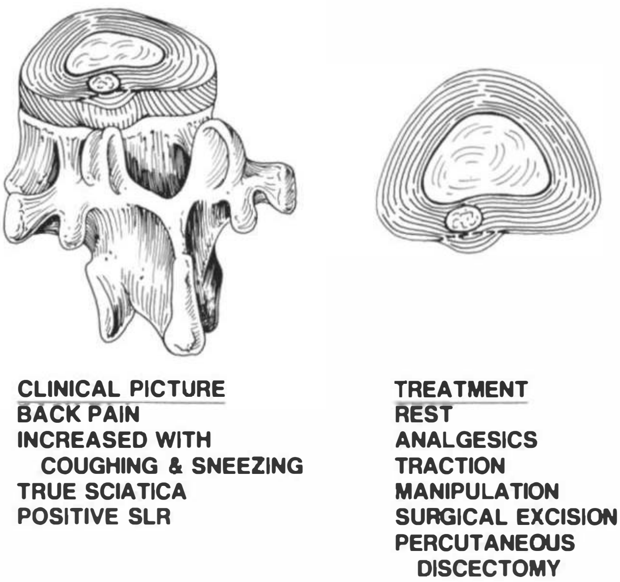


FIGURE 6-9 Sequestered fragment (the wandering disc; Type V). The results of treatment with surgery are better than those of Types I to IV but probably not as good as those of Types VI and VII. The wandering disc is a possible explanation for the clinical picture of exacerbations and remissions that is so frequently encountered. It may also be a partial explanation of why some patients show a good response to traction or manipulation. (*SLR*; straight leg raising test.)

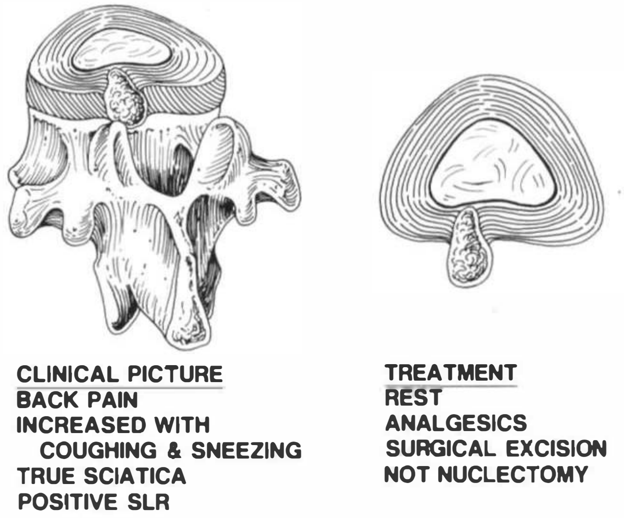


FIGURE 6-10 With Type VI there is sequestration and displacement but some anchoring of the ligament, so that it cannot move about. This is less likely to be helped by traction, manipulation, chemonucleolysis, or percutaneous discectomy.

contrary, if there is no scarring, the random movement of the sequestered portion of the disc may include positions of subsequent nerve root irritation.

Depending on the location of this hypothetical "wandering disc," it is possible that chymopapain, collagenase, or a nucleotome might successfully remove it.

Displaced Sequestered Fragment (Anchored; Type VI)

There is another clinical and mechanical situation that may develop. This is the displacement of a sequestrum of the annulus and/or nucleus into the spinal canal or intervertebral foramen (Fig. 6-10). The fragment is to some degree fixed in position. The nerve root irritation results from inflammation due to mechanical pressure, chemical irritation, an autoimmune response, or some combination of the three. There is true sciatica with positive straight leg raising tests. In association with a displaced portion of the intervertebral disc (sequestration), there may be narrowing of the interspace at the involved level. Axial traction, manipulation, or random movement is *unlikely* to help. Chymopapain injected into the disc space may never reach or affect the sequestrum, especially if there has been scarring or blockage of the hole in the disc structure. This is probably also the case should a percutaneous nuclectomy procedure be attempted. When this situation subsides spontaneously, we hypothesize that it is a result of phagocytosis and/or some physiologic adjustment of the neural structures to the irritation. These patients show the best results when treated with surgery, as suggested by Charnley and subsequently confirmed by Spangfort.[52,309]

Recent studies by Yasuma and associates,[375] who completed investigations of autopsy and surgical herniated disc specimens, are relevant here. The group identified a myxomatous degeneration and were able to categorize disc herniation into three types. One was primarily a nucleus pulposus extrusion through fissures; the second was mainly annulus fibers (a large sequestrated fragment); and the third was mixed and presented more as a bulge with nucleus pulposus behind it.

Degenerating Disc (Type VII)

Another stage may occur when the disc degenerates (Fig. 6-11). This involves a disruption of the normal annular fibers of the disc to such an extent that the disc is no longer able to serve an adequate mechanical function. This may be associated with degenerative arthritic processes of the vertebral bodies and/or the intervertebral joints. There may be chronic pain or intermittent pain, or such individuals may even be asymptomatic.

Organic Idiopathic Spine Pain

This is the type of pain present in patients who are diagnosed clinically as having organic spine pain without sciatica for which there is no known etiology. Pain may emanate from the disc; it may come from increased fluid uptake by the disc (Type II); it may come from any combination of the previously described etiologic factors; or it may come from some mechanism yet to be discovered.

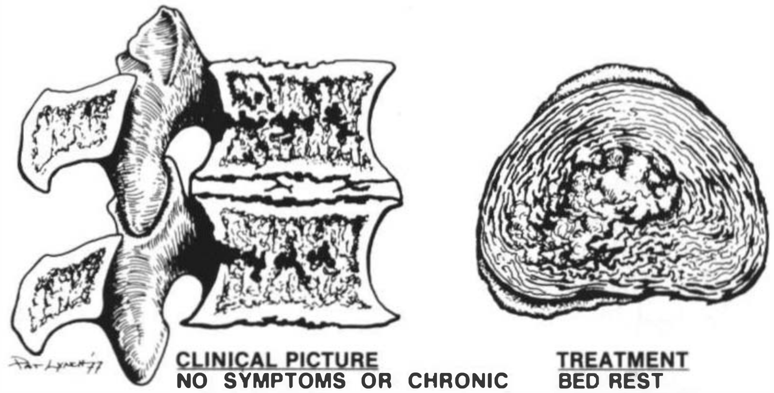


FIGURE 6-11 A degenerated disc (Type VII) may be the end process of the mechanical and biological effects of normal functioning or it may be associated with considerable pain and disability. There may also be arthritis in the intervertebral joints. It is important to emphasize that these various stages are a continuum. A given disc may move through several types or stages. The changes may accelerate or decelerate, stop, or in some instances they may even reverse.

CLINICAL PICTURE
NO SYMPTOMS OR CHRONIC SPINE PAIN
± SCIATICA
± SPINAL STENOSIS
OSTEOPHYTES AND NARROWING

TREATMENT
BED REST
ANALGESICS
SOMETIMES ARTHRODESIS

Immunologic Factors in Spine Pain

The basic hypothesis is that during the degenerative processes of the intervertebral disc, one or more of the degradation products stimulate the autoimmune response. The associated inflammatory response is the cause of the spine pain, with and without nerve root irritation. Autoantibodies to autogenous nucleus pulposus have been experimentally demonstrated in both animals and humans.[30,258] Investigators have identified in humans, through the leukocyte migration–inhibition test, the presence of a cellular immune response in patients in whom a sequestered disc was found at surgery.[127]

Elucidating the possible mode of an immunologic inflammatory response is helpful in the explanation of several characteristics of disc disease. This hypothesized mechanism aids in accounting for the chronic course of exacerbations and remissions and the success of anti-inflammatory drugs, such as aspirin, phenylbutazone, and steroids (administered locally and systemically). The predominant neurologic symptom is *pain*, which implies irritation rather than numbness, a symptom compatible with simple chronic neural pressure. Gross and microscopic observations of inflammation, granulation, and fibrosis in excised discs have been noted.[39] Subsequent research will further explain and confirm or invalidate this hypothesis.

Biomechanical Factors in Spine Pain

Naylor, in a review article, has presented an excellent summary of the biochemical and biomechanical factors that constitute a hypothetical explanation of the clinical phenomena of disc disease.[241] Although the disc is the largest avascular structure in the human body, there is considerable chemical interchange and activity there. The process of disc degeneration is summarized in Figure 6-12. The initial change is thought to be a disruption of the balance between the synthesis of normal proteinpolysaccharide and its depolymerization. The disequilibrium is such that there is an increased depolymerization. There is an associated increased fluid content in the nucleus pulposus, resulting in greater intradiscal tension. The increase in discal tension alone, as hypothesized by Charnley in 1955, can cause backaches. This may also be the cause of organic idiopathic back pain. From this point, according to the Naylor hypothesis, the situation may develop in at least three different manners (see Fig. 6-12). Processes 1 and 2 are set in motion when there is a cessation of the conditions that disturbed the equilibrium of synthesis, allowing it to be reestablished at a new but lower level. From this point, the process can progress in one of two disparate directions. Process 1 involves *repeated cycles* of abnormal proteinpolysaccharide synthesis, accompanied by increased collagen fibrillation. The repeated cycles may explain the clinically observed course of intermittent attacks of spine pain that tend to follow their own schedule of exacerbations and remissions. This may continue on to extreme nuclear degeneration and a fairly rigid, scarred disc that cannot develop tension or prolapse. Presumably, this stage may be reached with or without either spine or radicular pain. In addition to or because of the abnormal nuclear synthesis, Process 2 involves disruption of the disc mechanics and damaging stress and results in disruption and failure of the annular fibers. The end point of Process 2 is prolapse of the nucleus pulposus and/or some portion of the annulus. In this situation, prior to frank prolapse the patient would be expected to have a history of intermittent spine and radicular pain. Process 3 is a more direct progression to nuclear or annular prolapse following the initial biochemical and mechanical changes in the disc. This could be the pathophysiologic course followed by the patient with no spine pain who subsequently experiences rapid onset of radicular signs and symptoms, with or without spine pain.

As a simplified summary, we suggest that Process 1 is normal disc degeneration, 2 is the subacute or chronic symptomatic degeneration, and 3 is the acute prolapse of a disc with varying degrees of degeneration. There may well be some overlap among the three hypothesized courses that a given disc may follow. This probably depends upon genetic factors, mechanical factors, treatment, or some combination of these or other presently unknown considerations. This theoretic analysis offers an explanation for a good deal of what is observed clinically.

Biomechanical and Anatomic Factors in Nerve Root Irritation

We know that nerve root compression and the production of nociceptive stimuli involve several factors. The compression causes deformation, changes in microcirculation, ischemia, edema, demyelination (possibly inflammation), radiculopathy, and nociception.[286] The deformation may be due to di-

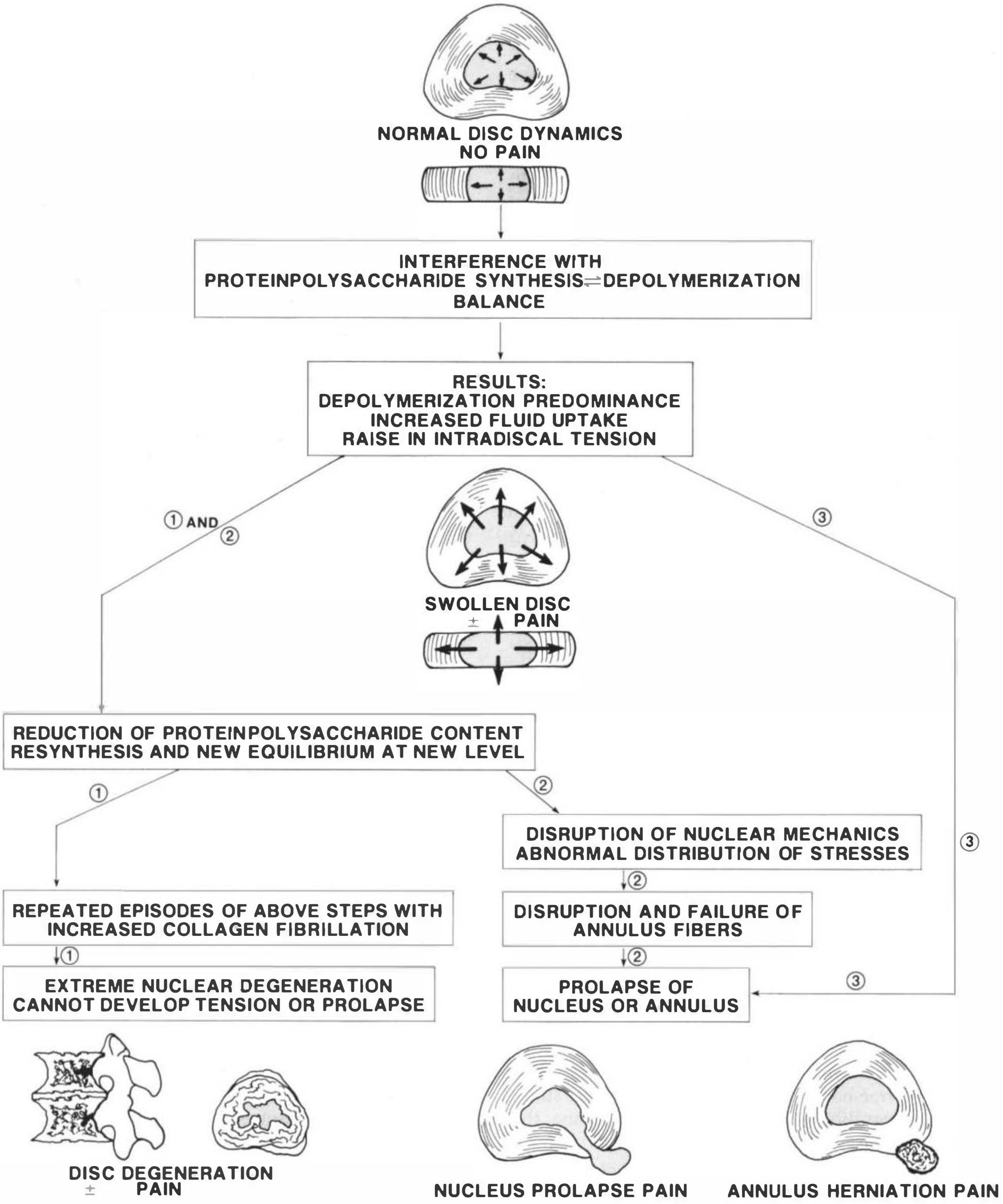


FIGURE 6-12 This flow diagram explains the biochemical hypothesis of the basic mechanisms of spine pain, disc prolapse, and degeneration. A number of mechanical factors mentioned in this chapter probably play a large role in the clinical presentation and outcome of these various biochemical phenomena. (*Based on Naylor, A.: Intervertebral disc prolapse and degeneration; the biochemical and biophysical approach. Spine, 1:108, 1976.*)

rect load application at the point of contact with the nerve root, and it may come from tensile loading at points above and below the point of contact at the level of the disc space. These tensile forces may be exerted by the attachment of the dural ligaments (ligaments of Hoffman) to the posterior longitudinal ligament and the vertebral body periosteum cephalad to the intervertebral disc. The dural attachment of the ligament is just at the level where the nerve root exits the dura. The caudad fixation occurs at the intervertebral foramen where the epineural sheath of the spinal nerve is attached to various structures within the intervertebral foramen (see Fig. 6-13).[310,311]

We can take these anatomic facts and put them together with some basic biomechanical modeling to provide an interesting hypothesis to explain some familiar clinical observations.[310,311] Although there are anatomic variations generally as to just how they are fixed, the lumbar nerve roots are in fact attached through the dural ligaments above to the posterior longitudinal ligament and periosteum of the vertebral body. Because of the variation, we chose to present them in the diagram conceptually rather than anatomically. The root is attached below the disc to the joint capsule, the pedicle, and the intervertebral foramen through multiple trabecular fibrous tissues.[310] We should also note here that work involving measurements of the sagittal diameter of the vertebral canal suggests that patients with more shallow canals are more likely to have surgery for radiculopathy. Presumably, a smaller disc protrusion may be enough to significantly impinge the nerve root because of the smaller canal. This study describes the measurement on lateral x-rays.[370]

Spencer and colleagues[311] studied 12 fresh cadavers using an instrumented force probe in contact with the L5 nerve root to measure forces at what would be the contact point between a bulging or protruding disc and the nerve roots. The effects of

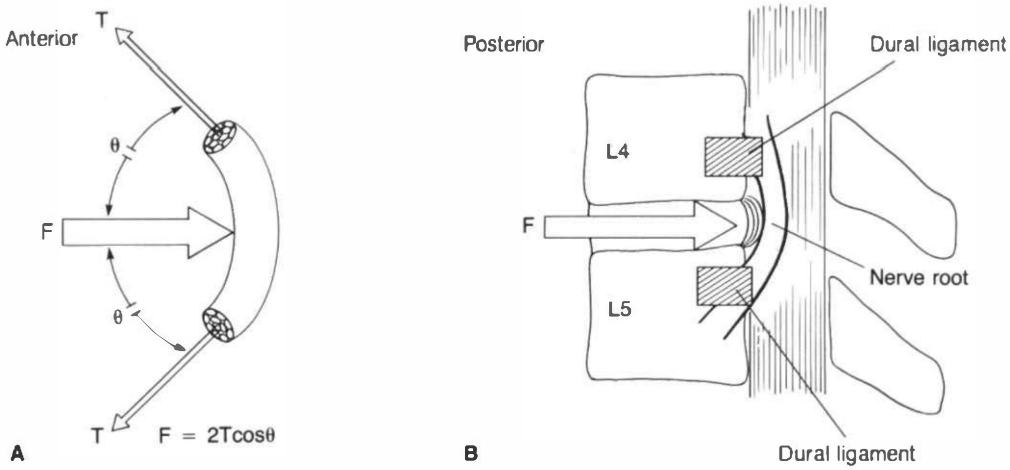


FIGURE 6-13 This figure illustrates schematically and anatomically various clinical and biomechanical factors that may be associated with the causation of nerve root irritation. **(A)** Pressure is exerted on the nerve root by a posterior bulging or herniation of the disc, provided there are counteracting tensile forces T exerted on the nerve root and dural sheath. A static free-body analysis shows that for the sum of the forces and moments to remain at zero (in equilibrium) in the sagittal (z,y) plane, the sum of the anteriorly directed forces ($2T\cos\theta$) must be equal to the contact force (F). Thus, $F = 2T\cos\theta$. **(B)** The anatomic portion of the figure in which clinical pathoanatomy has been depicted. The dural ligaments (ligaments of Hoffman) have been schematically presented to emphasize the concept of anchoring. The ligaments themselves come in a variety of numbers, shapes, and positions within this general region. Several clinical pathoanatomic situations are discussed in the text and presented in the displays on lumbar radiculitis and radiculopathy. These all focus largely on factors that increase the magnitude of force F, as with disc herniation shown here, and factors that increase the counteracting tensile force T exerted by the dural and Hoffman ligaments.

two rather important variables were observed. The effects of probe protrusion depth ("disc herniation") and disc space narrowing were investigated. Pressures went up with increasing probe depth (or "disc herniation"), as might be expected. However, pressures also increased with greater disc space height. This tends to confirm the hypothesized mechanism shown in Figure 6-13, in which it is asserted that tensile forces T on the dural ligaments play an important role in the trauma to the nerve root and the attendant pathophysiologic changes that lead to radiculitis, radiculopathy, and summarily sciatica. This hypothesis provides an excellent basis upon which to explain *increased* sciatica in the morning, presumably from increased disc height and *relief* of sciatica following chemonucleolysis.

Based on this information, here are some hypotheses that may explain certain observed clinical phenomena. The analysis can be set up as a kind of theoretic construct in which there is an equilibrium where both sides of the equation can be altered to produce mechanical damage and irritation of the nerve root. On one side there are mechanical variables having to do with the intervertebral disc, and on the other side there are factors related to the constraining ligaments above and below it. It may be that excessive bulging of a normal or slightly abnormal disc causes tensile loading irritation and inflammation of a nerve root because it is anchored in the manner shown in Figure 6-13. Depending on the anatomic location of the binding and the length and or elasticity of the ligaments, the threshold for irritation could vary. Obviously, another important variable would be the extent and frequency of associated disc bulging.

The trauma to the nerve root is expected to increase as the disc bulging becomes more abnormal, extensive, and prolonged. With a more frank herniation, the persistent protrusion would produce more mechanical irritation and damage to the nerve.

If the associated ligaments become more shortened, contracted, or scarred by the current irritation or cumulative irritation from previous episodes, one would expect more tension, irritation, and damage of the nerve root. Should there be additional scarring from inflammation, irritation by substances in the region, infection, or surgery, one would expect an increase in damage and irritation.

On the contrary, if, following herniation, inflammation or some other process causes lysis or *relaxation* of these ligaments, there would be improvement in the status of the nerve root. Surgical release of scar tissue and bulging or herniated disc tissue certainly can reduce irritation and inflammation.

Anatomic and pathoanatomic variation in the protruding disc and restraining ligaments may explain some well-documented clinical observations. The patient with a massive disc herniation who's doing well may have no Hoffman ligaments (dural ligaments in the region). The patient with a full clinical picture of disc herniation but with minimal or no imaging evidence of it may have very tight restraining dural ligaments and a modest disc herniation or just a physiologic bulge that is not brought out at the time of imaging studies. Patients with no restraining ligaments (they were absent two out of nine times at L4–L5 in the study by Spencer and colleagues[310]) may not show significant signs and symptoms of nerve root involvement until the nerve root is entrapped between a huge extruded disc and

ANATOMIC AND BIOMECHANICAL FACTORS INVOLVED IN THE PRODUCTION OF LUMBAR RADICULITIS AND RADICULOPATHY

Increasing disc bulge or herniation	F ↑ *	
Increasing frequency of impact of disc bulge	F ↑	Cumulative
Decreasing size of lumbar vertebral canal	F ↑	
Increasing size (swelling, inflammation) of nerve root	F ↑	
Increasing regional scarring	T ↑ *	
Increasing stiffness of dural ligaments	T ↑	
Decreasing length of dural ligaments	T ↑	
Decreasing size of intervertebral foramen	T ↑	
Increasing height of disc	T ↑	
Increasing proximity of anatomic attachment of dural ligaments to the disc	T ↑	

* F and T are forces diagrammed in Figure 6-13. As either F or T increases, the other increases. Decreases in F or T tend to decrease the nerve root trauma and the attendant radiculitis, radiculopathy.

the posterior wall of the lumbar canal. It is known that patients with decreased sagittal diameters of the lumbar vertebral canal are more likely to be operated on for lumbar radiculopathy than a control group with normal canal diameters.[370] The work of Porter and associates,[266] which compared 73 patients with disc symptoms with 200 normals, showed that at the L4–L5 level, subjects with an anteroposterior diameter of less than 14 mm were more at risk for needing disc surgery.

These various anatomic and biomechanical factors contributing to lumbar radiculitis and radiculopathy are summarized in the chart on page 399.

This anatomic theoretic analysis can also be used to explain certain therapeutic interventions. Disc space narrowing has been reported following chemonucleolysis. Hence, there are three possible benefits: (1) removal of disc pressure by the breakdown of proteoglycans in the nucleus pulposus, (2) reduction in disc volume and thus a significant overall reduction in disc pressure as explained by the bulk modulus (Fig. 6-14), and (3) a decrease in the tensile force (T in Fig. 6-13) by bringing the anchoring points of the dural ligaments closer together.

Percutaneous discectomy functions through the removal of nucleus pulposus and some annulus fibers, which may allow some displacement of a bulging disc back toward the interspace. In view of the fact that the normal repositioning of the disc is rarely observed following the procedure,[25] we hypothesize that the therapeutic benefit may be pressure release (decrease in force F) at the contact point between the bulging disc and the nerve root. This is theoretically feasible if there is a high bulk modulus.

A high bulk modulus is a situation in which the disc demonstrates a pressure volume relationship in which there is a steep slope (see Fig. 6-14). In other words, a relatively small change in volume is associated with a large change in pressure. If this is the case, then removal of a small volume of disc material, nucleus, and/or annulus could result in large pressure changes and relief of back pain and sciatica. There are several factors that relate to this hypotheses, both pro and con. Some data from Rydevik and co-workers[286] suggest that it is tension, not pressure, that causes nerve root pain. However, tension within the disc may be a cause of back pain. We know that in the normal situation, the pristine, nondegenerated disc tends to have a stiffer bulk modulus.[257] Should a young disc bulge and the pressure cause sciatica due to pressure, removal of a small volume of disc nucleus could then favorably alter the situation. This may be related to the tendency for younger patients to do well with percutaneous discectomy. Most herniated discs, however, have a significant element of degeneration. There is a tendency for changes in volume or displacement of the herniated disc out of the canal to be associated with

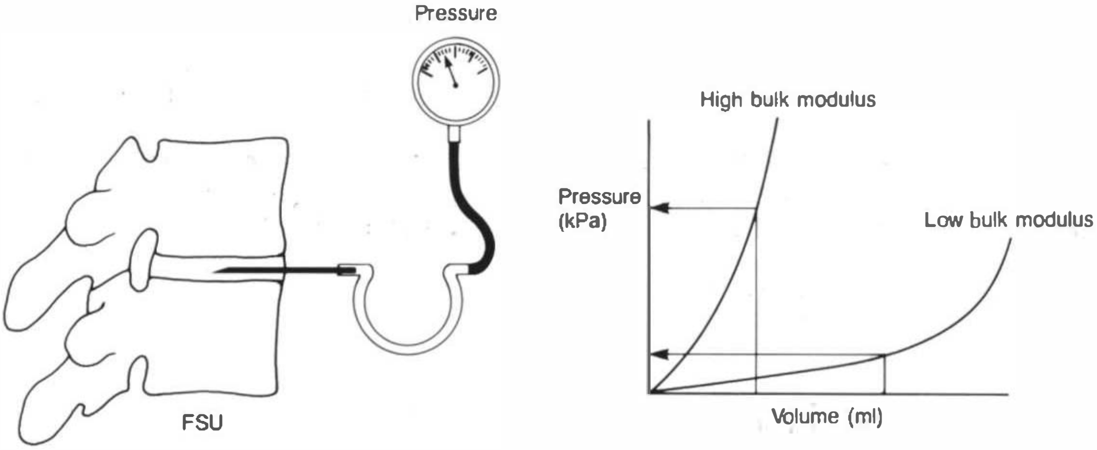


FIGURE 6-14 The concept of bulk modulus as applied to disc pressure measurement. If there is a high bulk modulus, then a relatively small change in volume will result in a large change in pressure. Conversely, if the bulk modulus is low, a larger volume change results in less pressure change.

a successful outcome with percutaneous discectomy.[25] These relationships require additional basic and clinical study in order to determine whether or not they are clinically significant.

Disc excision removes the protruded or extruded disc and relieves nerve root irritation by eliminating force F in Figure 6-13.

One situation in which repeat surgery may be successful is when residual or recurrent disc protrusion is associated with scarring of these dural ligaments. The binding from the T forces (Fig. 6-13) is a major factor because the offending disc material can readily irritate the unyielding nerve root. When there are anatomically correlated clinical signs and symptoms and imaging evidence of disc material in the canal, surgery is likely to be helpful. The scar generally must be released in order to expose and remove the disc. Appropriate laminectomy and foraminectomy with fat pad grafts help to keep the nerve mobile, free, unbashed, and clinically improved.

The various therapeutic factors that may contribute to the reduction of lumbar radiculopathy and radiculitis are summarized in the following chart.

Arthrodesis of the FSU can alter the equilibrium by decreasing the damaging impact of the bulging on the nerve. It is also possible that opening the posterior elements of the FSU, as with flexion, may increase the tension of a nerve root as its proximal and distal ends become slightly more separated.

Facet Joint Hypertrophy and Nerve Root Irritation

Burton and colleagues[44] and others have emphasized the potential role of the hypertrophied superior articular facet in the production of lumbar nerve root irritation (Figs. 6-15 and 6-16). The emerged facet articulation may project medially and engage the nerve root between it and a normal or a herniated disc (Fig. 6-15). An alternative mechanism is for it to project cephalad and trap the nerve root as it exits underneath the pedicle (Fig. 6-16). These specific mechanisms of nerve root involvement remind the spine surgeon that he or she should consider all pathoanatomic mechanisms for nerve and nerve root irritation when evaluating a patient both preoperatively and intraoperatively.

THERAPEUTIC AND BIOMECHANICAL FACTORS IN THE REDUCTION OF LUMBAR RADICULITIS AND RADICULOPATHY

Chemonucleolysis (disc narrowing, volume and pressure decreasing)	T↓	F↓ *
Percutaneous discectomy	F↓	
Disc excision	F↓	
Disc excision, release, ligaments, scarring, foramenectomy	T↓	F↓
Arthrodesis	T↓	F↓

* F and T are forces diagrammed in Figure 6-13. As either F or T decreases, the other decreases.

Lumbar Nerve Involvement in Spondylolysis and Spondylolisthesis

In an examination of 34 bony specimens of isthmic spondylolysis, Edelson and Nathan noted a 32% incidence of stenosis of the intervertebral foramen.[66] They also noted other sources of nerve root involvement, including degenerative changes in the lamina and at the lysis in the pars. The authors suggest that the stenotic foramen fixes the nerve root, and the "hooked" inferior portion of the lamina of the fifth lumbar vertebra catches the fifth lumbar nerve root and applies traction to it. This is combined with anterior pressure from the annulus or vertebral body below. Fibrocartilaginous tissues in the area of lysis contribute to the nerve root irritation. The best treatment when conservative management of spondylolisthesis with severe back pain and radiculopathy fails is laminectomy, foraminotomy, and posterolateral fusion, as recommended by Henderson.[143] This should relieve the nerve root irritation and compensate for the instability.

The "far-out" syndrome is another situation in which the L5 spinal nerve may be involved. The nerve is compressed between the transverse process of L5 and the ala of the sacrum. A Type I "far-out" syndrome is usually found in an elderly person with degenerative lumbar scoliosis. Type II is a spondylolisthesis with a 20% slip. A 25° caudocephalad (Ferguson) view is the best plain x-ray to show the condition. Computerized tomography (CT) scan with a "widely opened window" is the most definitive imaging study. Extensive lateral decompression of the nerve is required. Complete removal of the L5 transverse process with the lower half of the pedicle should be satisfactory. One should be cognizant of the possibility of this type of decompression creating or contributing further to the development of instability.

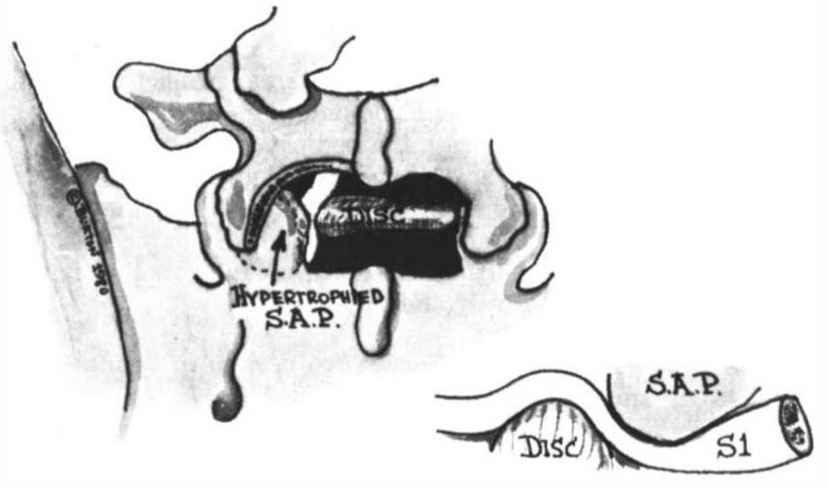


FIGURE 6-15 This hypertrophied degenerated superior articular facet of S1 has developed an osteophytic projection medially. This spur has impinged the nerve root against the herniated disc. This is shown from a posterior view and also from a left lateral perspective. *(Illustration by Charles Burton, M.D.)*

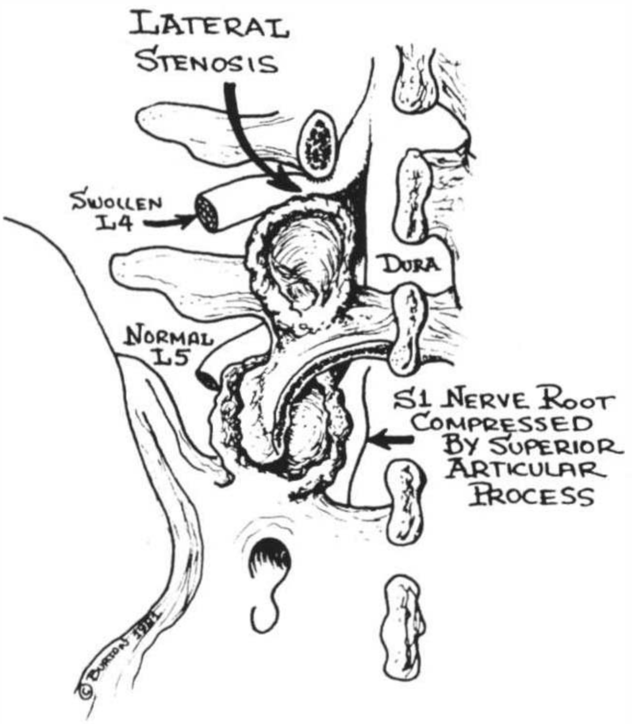


FIGURE 6-16 The hypertrophied degenerated superior articular facet of L5 has developed an osteophytic projection in the cephalad direction (+ y-axis). This spur has trapped the *4th* lumbar nerve root against the inferior portion of the pedicle of L4. *(Illustration by Charles Burton. M.D.)*

Cervical Spinal Stenosis

The biomechanics of cervical spinal stenosis are covered in Chapter 4, Practical Biomechanics of Spine Trauma. A developmentally small cervical canal as recognized by a Pavlov ratio of less than 0.8 puts one at greater risk of neurologic damage should there be trauma to the cervical spine. Cervical spondylotic myelopathy is more likely to occur in a patient with degenerative changes and a developmentally narrow canal. We also note that, analogous to the lumbar spine, the transverse or cross-sectional diameter of the dural contents is correlated with histopathology of the spinal cord, clinical signs and symptoms of myelopathy, and successful surgical treatment.

Except for long tract clinical signs and symptoms, the absence of claudication, the greater vulnerability of the spinal cord to trauma, and the greater recovery potential of the cauda equina and nerve roots, the cervical spine is quite analogous to the lumbar spine in regard to issues of pathophysiology and biomechanics of stenosis. Pain from the stenosis itself is usually not a major component of cervical spinal stenosis; thus, we have discussed this condition in more detail elsewhere.

Lumbar Spinal Stenosis

Spinal stenosis of the lumbar region is a condition that may not be recognized as frequently as it should be. Paine and Huang described 227 cases of the lumbar disc syndrome.[254] In this series, disc herniation alone was present in 31% of patients, developmental stenosis alone in 2%, degenerative stenosis alone in 27%, and combined lesions in 39% of cases. This shows that spinal stenosis is frequently present and probably plays a significant role in low back pain and sciatica, in conjunction with disc disease as well as independently.

Clinical Examination

The clinical presentation may consist of constant or intermittent, vague, atypical complaints. The pain distribution and radiculopathy may be multilevel or unilevel, either bilaterally or unilaterally. The highly characteristic feature is that of intermittent claudication. Physicians should be highly suspicious of patients with a history of intermittent claudication who have no clinical evidence of occlusive vascular disease. "Drop attacks" (sudden falling down because of leg weakness without loss of consciousness) are a rare finding but when present serve as a clue to the diagnosis. Some patients will note back and/or leg pain when walking down stairs or down an incline. Both of these activities tend to extend the lumbar spine. This may be associated with a relief of pain upon flexing the spine, either by sitting and bending forward or simply by walking in the slightly forward flexed position. The classic sign is the patient who walks and feels best when slightly bent and pushing a shopping cart. Presumably, the slightly flexed position reduces the invagination of the yellow ligaments, thereby providing more space for the neural and vascular elements within the lumbar canal. Moreover, we know from the work of Hanai and colleagues[140] that with extension, the CSF pressure in the lumbar spine is higher than when standing. With flexion, the CSF pressure is lower and is in the range that occurs when the patient is prone. The decreased pressure with *flexion* also fits the rationale for advising spinal stenosis patients to use bicycling as an exercise because it puts the spine in some flexion and provides good aerobic conditioning. This increased pressure on the cauda equina with extension could aggravate symptoms directly or indirectly through the production of ischemia to the regional neural elements. Parke and associated[259] have emphasized the possible role of vascular compromise in spinal stenosis and demonstrated a relative hypovascularity of the neural structures below the cavus in perinates.

The findings upon physical examination, like the clinical history, are likely to be mixed and vague. There may be a subtle suggestion of radiculopathy at several nerve root levels. The straight leg raising test may be weakly positive but not distinctive.

The diagnosis is made primarily on the basis of the nature of the history and physical findings, electromyogram (EMG), and the radiographic evaluation. On the plain films, one may see osteoarthritic involvement of the posterior joints in which the anteroposterior and lateral diameters of the bony canal have been reduced. Myelography and computerized axial tomography are helpful in a more accurate evaluation of the size of the lumbar spinal stenosis.[349] Some examples of spinal stenosis are diagrammed in Figure 6-17. The disease may pre-

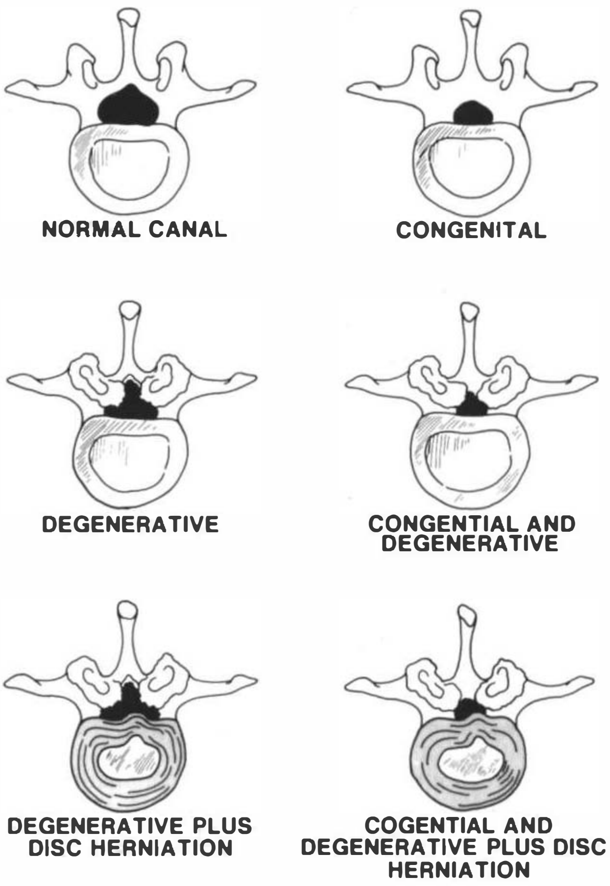


FIGURE 6-17 This diagram shows the normal canal and various combinations of conditions that may cause spinal stenosis. Congenital stenosis with disc herniation alone, not pictured here, is another possibility. *(Arnoldi, C. C., et al.: Lumbar spinal stenosis and nerve root entrapment syndromes: definition and classification. Clin.* Orthop., *115:4, 1976.)*

sent with the usual symptoms of disc disease or with low back pain and leg pain with no abnormal neurologic signs. Treatment includes nonoperative techniques, postural exercises, orthotic support, with lumbar spine in neutral or slight flexion, and epidural steroid injections. Although these have been recommended in the literature, there is little reason to be optimistic about their efficacy. Surgical treatment is through decompression by laminectomy and root canal decompression if needed.[187,367]

The key clinical biomechanical, anatomic, and pathophysiologic factors in lumbar spinal stenosis are described in the next few paragraphs. The space available for the neural and vascular elements is too small. This is due to a developmentally small canal and/or any number of several acquired conditions and anatomic changes. These anatomic and pathologic factors combine to cumulatively compromise the space available to the extent that the patient develops the previously described complex of symptoms known clinically as spinal stenosis.[242,349] The classification chart below lists various conditions that may encroach upon the canal. The following anatomic structures can encroach upon canal space: displaced disc, scar tissue, osteophytes of the vertebral body or facet joints, hypertrophic or deformed facet joints, joint swelling or cyst, and invagination of the yellow ligament.[65]

Once any combination of the above conditions and structures compromises the canal to some critical point, the signs and symptoms occur. The best quantification of this point is the cross-sectional area not of the lumbar canal but of the dural sac at its most constricted level or levels.[32,289,D] This can be measured by use of a CT scan with or without contrast. CT scan measurement of the transverse or cross-sectional area is the most accurate and reliable technique for clinical diagnosis. A cross-sectional (x,z plane) area of the dural sac of 100 mm^2 or less in an adult is considered diagnostic of central spinal stenosis. A dural sac cross-sectional area of 100–130 mm^2 is considered early stenosis, and the normal measurement is 180 ± 50 mm^2.

CLASSIFICATION OF SPINAL STENOSIS

- Congenital developmental stenosis
  - Idiopathic
  - Achondroplastic
- Acquired stenosis
  - Degenerative stenosis
    - Central canal
    - Peripheral canal and neural canal
    - Degenerative spondylolisthesis
  - Combined stenosis
    - Herniated disc combined with combinations of the above
  - Spondylolisthesis
  - Postoperative stenosis
    - Laminectomy, fusion, chemonucleolysis
  - Post-traumatic stenosis
  - Miscellaneous stenosis
    - Paget's disease, fluorosis

(Arnoldi, C. C., et al.: Lumbar spinal stenosis and nerve root entrapment syndromes: definition and classification. Clin Orthop., 115:4, 1976.)

Imaging Analysis

There are other imaging methods available for making the determination of the normality of the canal space. These are discussed here for the sake of completeness and because there are situations in which it may not be possible to calculate the cross-sectional area of the dural sac. The linear measurement of the anteroposterior diameter of the lumbar dural contents measured on a myelogram is a useful measurement. The lower limit of normal is 12–14 mm.[187] Measurements of the bony canal are not particularly helpful because they often do not correlate with the more crucial measurement of the space available in the dural sac.

Another specific imaging method to evaluate patients for spinal stenosis is magnetic resonance imaging (MRI), which can be especially helpful in identifying specific anatomic structures that are contributing to the cumulative stenosis at any particular level in the canal. We are not aware of any studies that correlate specific measurements of the dural sac by MRI with the diagnosis and treatment of central spinal stenosis; however, Schnebel and colleagues[288] reported a 96.6% agreement between MRI and contrast CT in the diagnosis of spinal stenosis.

Comments

There is a cogent theoretic construction for the production of symptoms in spinal stenosis (see Fig. 6-18). We have described previously the various factors that may constrict the space available in the spinal canal. As these accrue, the space available gradually diminishes. At some point that is not well defined, the threshold of space compromise that produces the symptoms is transcended. The various components may be removed either temporarily or

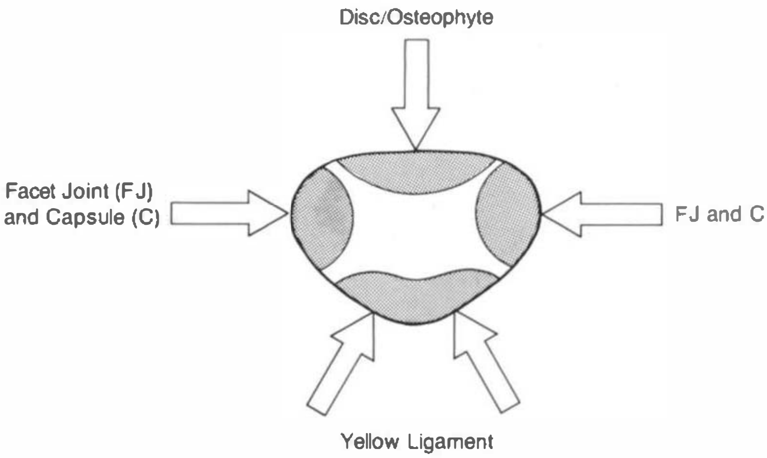


FIGURE 6-18 Spinal canal encroachment. Theoretical representation of the elements that cumulatively reach the threshold, causing symptoms of spinal stenosis. Any of the elements may be present and enlarge with progression. Various activities and therapeutic measures may temporarily or permanently reduce their effects.

permanently. Incremental increases in space available may reach the threshold for taking the patient back to an asymptomatic level. For example, anti-inflammatory agents may reduce facet capsule swelling and/or cauda equina edema. Sitting, bending and flexing the lumbar spine may temporarily reclaim the yellow ligament from the canal. Open or percutaneous discectomy may reduce the disc encroachment into the canal. Laminectomy and/or facetectomy will reduce the posterior and/or posterolateral encroaching elements.

We should not leave this topic before reviewing the work of Tsuji and associates.[336] These investigators have completed extensive studies of the role of redundant nerve roots in degenerative lumbar stenosis. They noted redundant nerve roots in 45% of cadavers studied and in 39% (22 of 56) of patients with degenerative lumbar spinal stenosis. It was hypothesized that the redundant roots may slide up and down in the caudal sac, become hypertrophied, and develop a kind of friction neuritis. The authors also suggest that the redundancy may develop as a result of shortening of the axial length of the canal due to the degenerative process in the lumbar canal. They showed that the canal becomes shorter with age. The importance of this observation was further magnified by the observation that there was a statistically significant correlation ($p<0.01$) between the claudication distance and the extent of caudal root redundancy. Figure 6-19 shows only moderate root redundancy but emphasizes the dramatic kinking or blocking with extension that can occur in patients with degenerative stenosis and root redundancy. (We suspect that to some extent this occurs with degenerative disc disease elicited by extension or simply standing or sitting.) The pathophysiologic and clinical consideration of redundant nerve roots in degenerative spinal stenosis merits careful attention and additional study. The author's (A.A.W.) experience with patients in the United States suggests a lower prevalence of nerve root redundancy in this condition.

Lumbar Scoliosis and Low Back Pain

Nachemson takes the position that severe low back pain is probably no more common in patients with lumbar scoliosis than it is in those with a straight spine.[233] He suggests conservative treatment with facet injections, traction, and transcutaneous electrical nerve stimulation (TENS) units. Kostuik and Bentivoglio reviewed intravenous pyelogram x-rays of 5,000 patients and found 2.9% who had lumbar or thoracolumbar curves.[189] In a review of 159 of 189 scoliosis patients available for following, they noted that the overall incidence of back pain was 59%, the lower end of the expected 60–80% range for the general population. It was observed that pain increased significantly with curve severity. There was an association of pain with radiographic evidence of facet scoliosis in 64% of the scoliosis patient group. Epstein and co-workers reported on a group of elderly scoliotic patients with degenerative changes, scoliosis, and back pain.[95] Simmons and Jackson discussed the problem of nerve root entrapment in patients with collapsing scoliosis.[298] These investigators considered Dwyer instrumentation and correction a satisfactory management technique of both the scoliosis and the nerve entrapment, which was generally on the concavity of the curve. Recent work

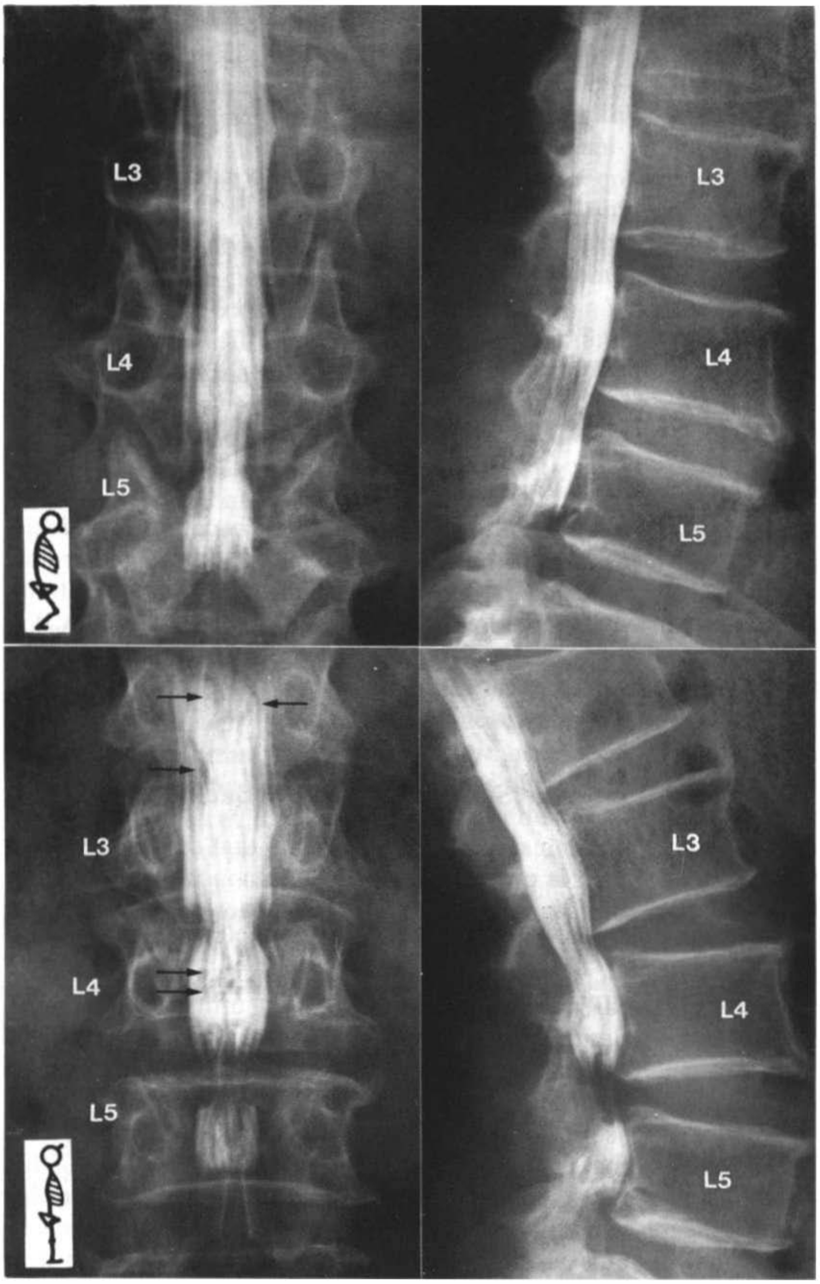


FIGURE 6-19 This composite is interesting from several perspectives. (**A** and **B**) show anteroposterior and lateral views with the spine in neutral or slightly flexed. Note absence of nerve root redundancy and minimum block on myelogram, except the large L5–S1 disc at bottom level. (**C** and **D**) show dramatic changes with spine extended. There is nerve root redundancy (arrows), evidence of major cauda construction at L4–L5, and a small impingement at L3–L4. (From Tsuji, H., et al.: *Redundant nerve roots in patients with degenerative lumbar spinal stenosis. Spine,* 10:72, 1985.)

by Sponseller and associates,[316] who followed 45 patients treated surgically as adult scoliotics, indicated that the improvement in pain, activity levels, and pulmonary function was not outstanding. There were only modest improvements in pain. They reported that 40% of these patients had minor complications and 20% of them experienced major complications.

The biomechanics and pathophysiology of the pain mechanisms in scoliosis are not well understood. Rigorous, conservative patient management followed by appropriate decompressions, possibly corrections, and arthrodesis is reasonable provided there is adequate explanation of the pain mechanisms on an individual basis. Care should be taken not to create a multivertebrae fusion mass that extends down to the transitional functional spinal unit (FSU). Ideally, the motion of the FSU above that should also be saved.

Loads and Motion

In general, the cervical spine has the most intersegmental motion if one considers all the parameters of motion (see Fig. 2-24). However, the loads are relatively low in this region. In the thoracic spine there is relatively little intersegmental motion, and the stiffness is high because of intrinsic mechanical properties, the rib attachments, and the thoracic cage. The thoracic spine loads are intermediate between those found in the cervical and the lumbar spine. In the lumbar spine, there is an intermediate degree of intersegmental motion, but the loads applied are of the highest magnitude. It is in this region that patients most often experience pain.

A biomechanical analysis of the incidence of pain in the different regions of the spine suggests a relationship between *loads* and *motion*. The degree of motion is highest in the cervical spine, but the loads are small. It ranks second in incidence of pain. There is relatively little motion in the thoracic spine, and the loads in this region are moderate. It ranks third in incidence of pain. The lumbar spine undergoes a moderate degree of motion and very high loads. It ranks first in incidence of pain. These relationships are summarized in Table 6-1.

Because of the plane of orientation of the facet articulations in the lumbar area, there are relatively more shear forces on the intervertebral discs during axial rotation.[135] In addition, discs L4–L5 and L5–S1 are subjected to high shear forces because of their high angles with the horizontal plane. It has been suggested that this may be a factor in the higher rate of development of disc herniation there. Farfan and colleagues have proposed that torsional loading causes failure of the annular fiber, which results in disc disease.[100] These factors may account at least in part for the high incidence of disc disease at L4–L5 and L5–S1.

TABLE 6–1 Relationship Between Motion, Loads, and Pain in Regions of the Spine

	Degree of Motion (Rank)	Magnitude of Loads (Rank)	Incidence of Pain (Rank)
Cervical	1st	3rd	2nd
Thoracic	3rd	2nd	3rd
Lumbar	2nd	1st	1st

Comments

> Having engaged in research in the field for nearly twenty-five years and having been clinically engaged in back problems for nearly the same period of time, and as a member and scientific advisor to several international back associations, I can only state that for the majority of our patients, *the true cause of low back pain is unknown.*
>
> (ALF NACHEMSON, 1975)

Several possible mechanisms that may be involved in spine pain have been presented. The preceding quotation is included as a poignant reminder that the current state of knowledge does not yet permit a full scientific understanding of spine pain. Although there are some good working hypotheses, a tremendous amount of research is needed to resolve this protean problem that frequently compromises the quality of life for so many people.

It is now almost 15 years after Professor Nachemson's comments. There has been significant progress in the management of back pain and the clinical imaging of various pathologic and anatomic components of the spine. However, if one uses Koch's postulates, modified to address low back pain or simply "the true cause," the challenge remains before us.

DIAGNOSTIC CONSIDERATIONS

This section focuses on factors that are of biomechanical significance or are crucial to the satisfactory evaluation of typical spine pain syndromes with associated neurologic problems. The discussion assumes that the physician has basic skills and knowledge in taking a general history, performing a physical examination, and maintaining the clinical management of an adult patient.

Clinical History

Cervical Spine Pain

A review of the salient clinical features of cervical spine pain follows.

Cervical spine pain is found in any combination of sites involving the neck, shoulder, and arm. The history of onset in the three regions may have any sequence. Usually, there is neck and interscapular pain, followed by pain in the other two areas. The location of arm pain and/or dysesthesia can be helpful in determining the level at which the pathology may exist. The neck pain or the brachial pain may be increased with coughing, sneezing, or Valsalva's maneuver (Fig. 6-20).

There may be a history of whiplash injury or a strain of the neck with some physical activity, or there may simply be a spontaneous onset, gradual or sudden. One of the common sites of referred pain from the cervical spine is the interscapular region. There may be associated problems, such as frozen shoulder (28%), epicondylitis, or carpal tunnel syndrome. The reasons for the associations are not clearly understood. Cubital tunnel syndrome and carpal tunnel syndrome may be related to the "double jeopardy" concept, in which it is suggested that when there are two sites of painful irritation of a nerve, they may reciprocally potentiate pain sensation associated with the two sites of irritation. In other words, the cervical nerve root disease lowers the threshold at which any irritation at the other sites may become symptomatic.

FIGURE 6-20 Valsalva's maneuver, forced expiration against a closed glottis with tight perineal sphincters, increases venous and cerebralspinal fluid pressure. When there is cervical spondylosis or soft cervical disc disease, this maneuver may cause neck or neck–shoulder–arm pain.

Thoracic Spine Pain

Thoracic spine pain syndromes, though relatively less frequent than pain in other regions of the spine, can be more serious if they are associated with a herniated disc. There may be a dramatic onset of symptoms such that the patient drops to the floor with paralysis. The shoulder blade is a recognized location for pain associated with thoracic spine disease.[164] The data from this study also suggested that a history of heavy work before age 15 may contribute to the development of kyphosis, which is associated with thoracic spine pain.

The clinical patterns of presentation of a thoracic disc are quite variable. Many of the cases, however, show a rapid onset of thoracic or low back pain, followed by sensory disturbances and motor weakness; about 50% of the patients have visceral dysfunction (bladder and bowel disturbances). The pain is increased by activities that involve Valsalva's maneuver. Trauma is the precipitating factor in the onset of the symptoms in roughly one-third of patients.[334]

Lumbar Spine Pain

Spine pain of acute onset associated with a particular mechanical incident may be related to a strain or rupture of some of the annular fibers of the disc or other muscular or ligamentous structures. If there is associated sciatica, the presumption is strengthened. Spine pain with or without sciatica, occurring without specific incident even in sleep, does not rule out disc disease. It is known that disc degeneration occurs as a gradual process, and the ultimate displacement to the point of irritation may be a subtle insignificant event. Because vascular, inflammatory, and biochemical factors may be operative in the production of pain, the onset may be gradual and progressive.

Lumbar disc pain is generally alleviated by rest, with the hips and knees flexed. The pain is accentuated by coughing, sneezing, and straining at the stool. These phenomena are thought to be mechanically related. In the erect position, the disc pressure is greater and the disc tends to bulge about its periphery. It is also known that the venous system is connected with the ventricles and the subarachnoid space, and a Valsalva's maneuver (coughing, sneezing, or straining at the stool) can increase pressure in the subarachnoid space. This space extends out along the nerve roots just into the intervertebral foramen in the lumbar region. If there is inflammation and engorgement of this already crowded space, the slightest change may constitute a pain stimulus. A slight increase in subarachnoid space pressure or a slight stretch of the nerve rootlet could easily trigger the pain-eliciting mechanism. It is for these biomechanical reasons that a Valsalva's maneuver tends to aggravate pain and the position of flexed hip and knee, which gives the lowest intradiscal pressure and causes the least stretch of the sciatic nerve, tends to relieve pain.

Physical Examination: Cervical Spine

The physical examination of the spine has been well presented in other publications.[73,107,161,173] It is presumed that an adequate general physical and musculoskeletal evaluation will also be carried out. Aspects of the physical examination that have some biomechanical relevance are discussed here.

Comments

There should always be a thorough motor and sensory examination, with care to rule out any physical evidence of myelopathy. Pain with firm palpation or percussion over the spinous processes of the involved FSU has been noted in the cervical spine, as in the thoracic and lumbar spine pain syndromes.[164]

A standard chart for localization of nerve lesions is provided here; however, it is important to indicate that there is often variation in the anatomic levels, one level cephalad or caudad to the classic textbook description.

Spurling's Test

This is a helpful test. The head is turned in maximal axial rotation facing, for example, first to the patient's right. Then it is laterally bent maximally to the right. With the head in this position, a vertical blow is delivered to the uppermost portion of the cranium. With the head and neck in this position, the vertebrae are vertical, approximately, and the disc on that side bulges maximally into the intervertebral foramen, which is also at its smallest size in this position (Fig. 6-21). The blow to the head is transmitted to the disc, which spreads a bit further and causes maximum encroachment on the intervertebral foramen. The left side is then tested by axially rotating and laterally bending the patient's head to the left and delivering a new blow. This should stimulate any nerve root or other pain-sensitive structures related to disc disease and cervical spondylosis. A positive Spurling's test, then, would be a complaint of any combination of neck, shoulder, and arm pain when the blow is delivered with the head and neck in the described position.

Range of Motion

These tests offer some crude, indirect evidence of disease. During extremes of motion, pain similar to that which the patient generally suffers is some indication of disrupted mechanics, as with cervical spondylosis. The other aspects of the physical exam are designed to localize dermatomal or myotomal dysfunction. The manual muscle tests, reflex changes, and sensory tests are carried out. When these findings correlate with the plain radiologic and imaging evidence of the level of cervical spondylosis, the prognosis for a good or excellent result with surgery is very much improved.

The Extension Test

The examiner gently assists the patient in putting the neck in maximum extension and then holds it there for 15–25 seconds. The test is positive if the

LOCALIZATION OF CERVICAL NERVE ROOT LESIONS BY PHYSICAL EXAMINATION

Neurologic Level	*Weakness*	*Reflex Depression*	*Sensation Decreased*
C5	Deltoid and biceps	Biceps	Lateral aspect of upper arm
C6	Biceps flexion and wrist extension	Brachioradialis	Thumb and index finger
C7	Finger and elbow extensions	Triceps	Middle finger
C8	Finger flexions	None	Ulnar aspects of forearm

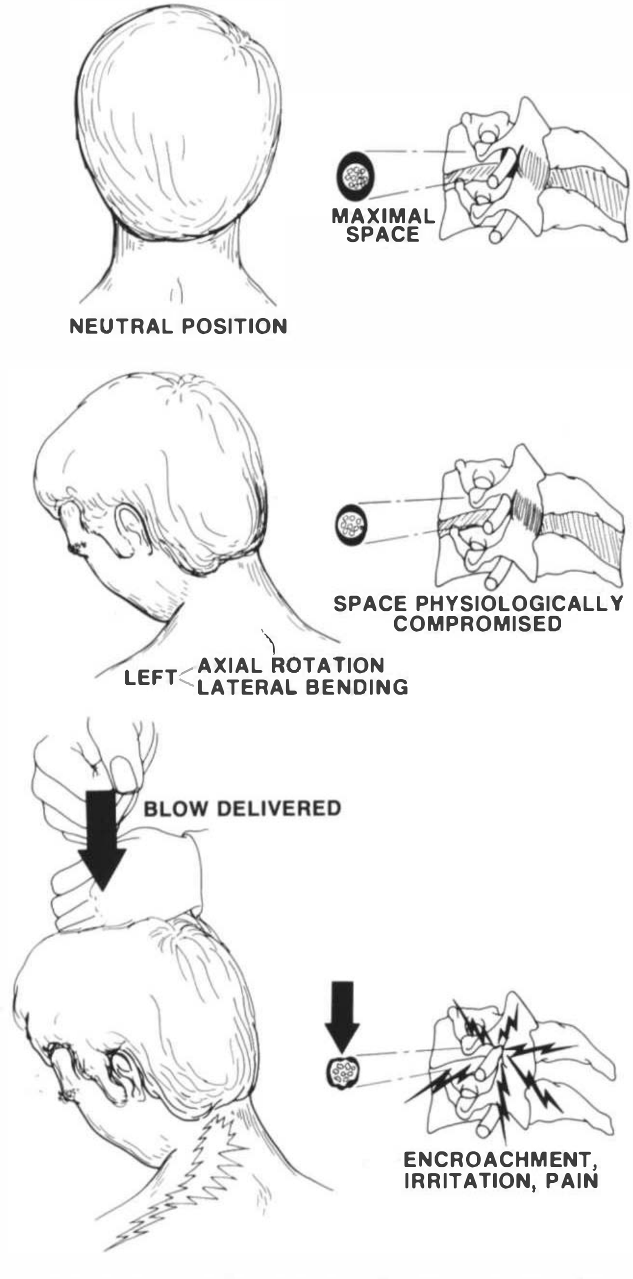


FIGURE 6-21 Spurling's test is based on several biomechanical factors. If there is some pathological compromise or irritation of the nerve root, when the root passes through the intervertebral foramen the irritation is aggravated. In order to demonstrate this, the head is positioned as shown, and the coupled motions of axial rotation and lateral bending will further compromise the space available in the foramen. When the test is positive, a vertically directed blow of moderate impact produces an additional lateral bending moment that reduces this space, irritates the nerve root, and causes some combination of neck, shoulder, or arm pain. This does not occur in a normal person.

patient complains of increased severity of neck, interscapula, shoulder, or arm pain (Fig. 6-22). This test was described to one of the authors (A.A.W.) by Professor Lee Riley of Johns Hopkins University. A positive test is indicative of cervical disc disease. We hypothesize that with extension, the herniated disc bulges posteriorly into the foramen, further irritating or aggravating the inflamed nerve root or the posterior longitudinal ligament. Biomechanical studies in Japan by Professor Hattori and associates showed that the maximum cervical disc pressure is developed with neck extension (Fig. 6-23).[298] Therefore, neck pain could come from direct pressure on an inflamed disc. There may also be some decrease in the intervertebral foramenal space, which could irritate an inflamed nerve root.

The Shoulder Abduction Test

A patient being evaluated for cervical radiculopathy and/or radiculitis may demonstrate this test. There may also be a history of pain relief in the arm when the hand is placed on the head, as shown in Figure 6-24. The test involves the examiner putting the patient's painful arm in the abducted position with

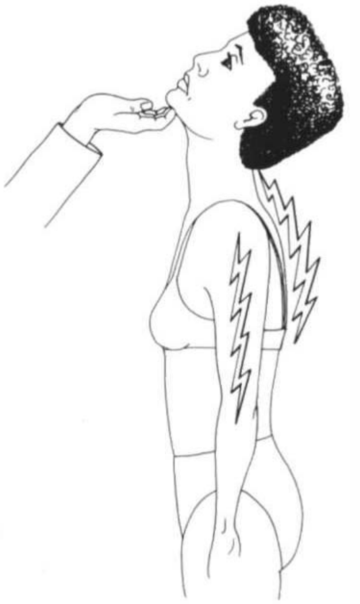


FIGURE 6-22 The extension test is performed with the examiner gently extending the neck and holding it there for 15–25 seconds. The test is positive if the patient complains of neck and/or shoulder/arm pain. This test, as far as we know, has not been reported in the literature. *(Personal communication from Professor Lee Riley, Johns Hopkins Medical School, Baltimore, Maryland, December 1985.)*

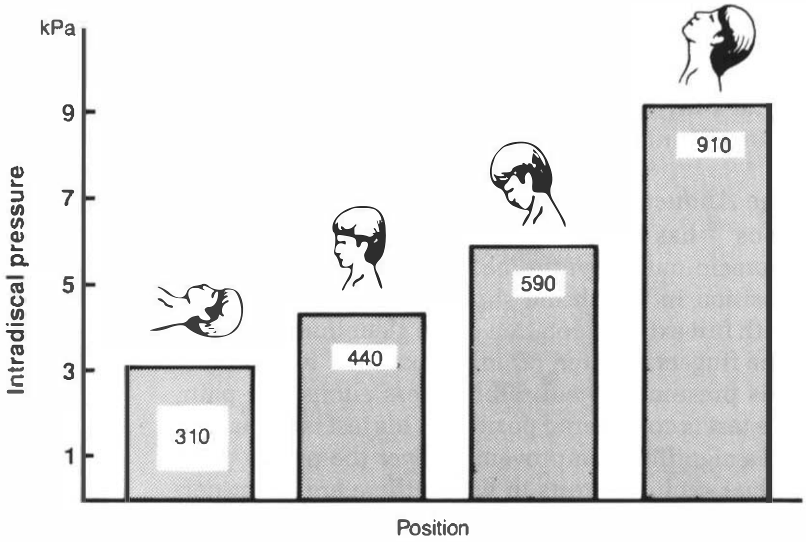


FIGURE 6-23 Normal intradiscal pressure in various positions of the cervical spine. (*Adapted from Hattori, S., Oda, H., and Kawai, S.: Cervical intradiscal pressure in movements and traction of the cervical spine. J. Orthop., 119:568, 1981.*)

the palm of the hand resting on the head as shown in Figure 6-24.

If the arm pain goes away or is significantly diminished, the test is considered positive for an extradural compressive monoradiculopathy. The most likely agent is a herniated cervical disc and/or osteophyte. The abducted position significantly relaxes the nerve by bringing the scapula several (3–4) centimeters closer to the spine.[276] The converse of this test can present as a clinical symptom. Patients will sometimes report exacerbation of arm pain after carrying a briefcase, suitcase, or shoulder bag.

FIGURE 6-24 The painful arm is placed in abduction and external rotation by putting the patient's hand on top of her head. If the arm pain is reduced or diminished, the test is considered positive for extradural compressive monoradiculopathy. (*From Robin, D. I., Dunn, E. J., and Metzmaker, J. N.: The shoulder abduction test in the diagnosis of radicular pain in cervical extradural compressive monoradiculopathies. Spine, 6:441, 1981.*)

Lhermitte's Sign

This is, in the obvious sense, the opposite of the extension test. This may be reported as a symptom by the patient or elicited by an examiner. The patient is asked to *flex* the neck maximally. The test is considered positive if the patient reports tingling in the back and the legs. This is due to spinal cord damage, and it may occur in several conditions, including cervical spondylotic myelopathy, syringomyelia, multiple sclerosis, herniated cervical disc, and spinal cord tumor. If plain lateral x-rays show a Pavlov's ratio in the middle or lower cervical spine of >1, then the clinician should evaluate the patient to rule out the preceding group of diseases.[8] The probable mechanism is the mechanical lengthening and the increase in the cervical spinal cord with flexion counteracted by some opposing or pressure-exerting force in the canal (disc, osteophytes, yellow ligament, bone tumor, meningioma) or within the spinal cord (intraspinal tumor, syrinx, plaque, or multiple sclerosis).

The Pancoast Tumor

Although well described in several other texts, this is presented here because the more often we think about it, the more likely we are to recognize it early enough to make a difference in the treatment. The findings are: *an apical mass palpable in the supra-*

clavicular fossa and neck/shoulder/arm pain, with or without Horner's syndrome and with or without wasting of the muscles of the hand.[255] Apical lordotic x-ray views and MRI are the major diagnostic determinants short of biopsy.

The Abduction External Rotation Test

Roos[279] has described this test as an indication of thoracic outlet syndrome. The patient is put in a position in which the shoulder is abducted to 90° with full external rotation of the glenohumeral joint. The fingers are then rapidly flexed and extended. If this presumed claudicatory stress elicits arm pain, the test is considered positive. This test is thought to be a significant improvement over the purely positional occlusive tests in which there are larger percentages of normals who will test positive. For example, 53% of normals may have an "abnormal" Adson's test,[126] and 68% and 54% of normals will have radical pulse alterations with the costoclavicular and hyperabduction tests, respectively.[331] With the Roos abduction external rotation test, only 7.5% of normals developed weakening of the pulse.[70]

Other findings on physical exam may include supraclavicular tenderness, a supraclavicular mass or swelling, a bruit, upper extremity weakness, hypesthesia, edema, or upper extremity blood pressure asymmetry. Several physical findings may be helpful in distinguishing upper plexus from lower plexus involvement. Findings in upper plexus involvement include: point tenderness over C5 and C6 nerve roots and upper trunks; pressure over the side of the neck reproducing symptoms; tilting or turning of the head to the opposite side causing pain; weakness of the biceps, triceps, and wrist; and hypesthesia in the radial nerve distribution. The physical findings of lower plexus involvement include the following: pressure above the clavicle producing pain and reproducing symptoms; tenderness along the ulnar nerve from the axilla to the inner arm; weakness of hand grip; weakness of the interosseous muscles; hypesthesia throughout the ulnar nerve distribution; and reproduction of symptoms by the 3-minute abduction/external rotation test. In general, about 15% of patients appear to have upper plexus involvement.[260]

Physical Examination: Thoracic Spine

There may be pain over the spinous processes of the involved FSU. There is sometimes neurological evidence of myelopathy. A sizable number of combinations of neurologic disturbances may be seen with thoracic disc disease.[334] No particular pattern appears to predominate. There may be abdominal-level sensory disturbances (numbness, paresthesias, and loss of vibratory strength), motor disturbances (paraparesis and paraplegia, muscle spasm, fasciculations, atrophy), and abnormal reflexes (hyperactive or hypoactive with or without symmetry). Some patients have had a positive Romberg's sign. All the varied combinations observed are presumably due to the distinct sensitivity of the spinal cord in the region.

The vulnerability can be attributed to the minimal amount of free space available to the cord when it is impinged by displaced disc material. The spinal cord has less freedom of movement, and therefore there is a greater possibility for a contrecoup disruption and production of neurologic problems. Thus, the dorsal column signs and Brown-Séquard neurologic signs are sometimes present.

Moreover, the blood supply of the cord in this region is precarious. Dommisse showed lucidly that the thoracic spine between T4 and T9 exhibits the least degree of vascularity and space for the thoracic spinal cord.[79] Consequently, this is the region where there is the lowest threshold for spinal cord damage. This very important relationship of blood supply, available space, and the possible pathologic effects of mechanical disruption is shown in Figure 6-25.

Physical Examination: Lumbar Spine

There are characteristics revealed by physical examination that are helpful in the diagnosis of organic pain and in the recognition of disc disease. Findings that have a relevant mechanical basis are discussed here, along with some of the tactics that aid in diagnosing nonorganic and functional disease.

Body Stances

The patient awaiting examination will consciously or subconsciously stand with the hip and knee both slightly flexed in the leg in which the pain resides (Fig. 6-26). This is a very reliable sign in our opinion, because the patient without realizing it has learned to stand in this position to relieve nerve root pressure. By slightly flexing the hip and knee, there is less tension on the sciatic nerve. Often, if the patient is asked "Why are you standing that way?" the response will be "Which way?"

There may be a *list* to either the ipsilateral or the contralateral side of the sciatica. Biomechanical

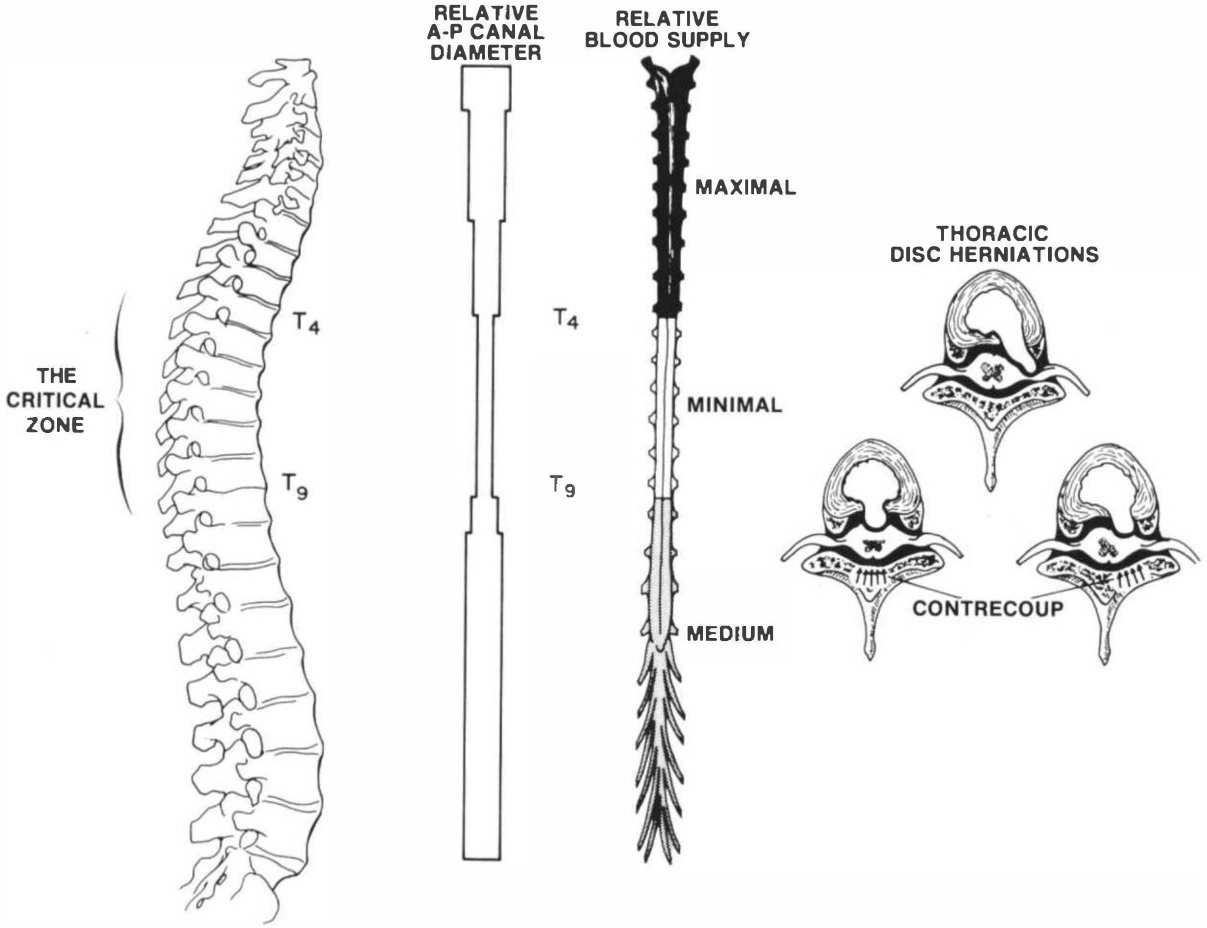


FIGURE 6-25 This diagram emphasizes several very important points in the understanding of thoracic spine pain, disc disease, and clinical stability in the thoracic spine. In the critical zone, the canal space and the free space between the spinal cord and the borders of the spinal canal are minimal. Although the thoracic cord is small, the relative free space is still minimal in the region of T4–T9. Moreover, the blood supply is less than elsewhere in the spinal cord. Therefore, this zone is doubly sensitive to any encroachment of the space available. A herniated disc not only causes spinal cord impingement, but there may be a contrecoup phenomenon in addition, so that both factors interfere with an already modest blood supply. It can be seen from the relative diameter of the anteroposterior canal that the situation is not quite as crucial in the highest and lowest regions of the thoracic spine. These considerations explain the catastrophic nature of thoracic disc disease and clinical instability in the thoracic spine.

considerations suggest the following: If the patient lists to the side of the sciatica, the disc herniation is in the axilla of the nerve root; if the patient lists away from the side of the sciatica, the herniation is lateral to the nerve root (Fig. 6-27).[107] This hypothesis seems reasonable based on theoretic pathoanatomic evaluation. To our knowledge it has not been documented with clinical investigations. On the contrary, recent work shows that the side of the list is unrelated to both the position of the disc herniation and the side of the sciatica. There was a much higher percentage of patients with a list who failed conservative treatment and required surgery. Forty percent of patients with a list required surgery, while only 5% of those without a list needed surgical treatment.[266]

Camptocormia

Occasionally, a modestly educated young male patient presents with a complaint of severe low back pain and an inability to straighten up. The patient is usually grotesquely bent forward and tilted to one

FIGURE 6-26 The disc stance is highly suggestive of the presence of a herniated disc or some other form of mechanical nerve root irritation in the lower lumbar spine. The patient sometimes unconsciously stands in this position, usually with a straight back (as emphasized by the ruler) and hips and knees flexed. The former reduces the posterior bulge of the disc, and the latter minimizes the stretch on the sciatic nerve roots.

side or the other. Attempts to have the patient actively or passively straighten up while standing are usually met with total failure. However, when he lies down on the examining table, the "deformity" is readily corrected. This is a unique and classic type of hysteria, the treatment of which is in the psychosocial realm rather than the biomechanical realm.[205,278]

An update on camptocormia shows that it does occur in the female, although rarely. A recent case report indicated that it was only the fourth case reported in the literature involving a female.[281] The differential diagnosis should include spinal cord neoplasm, vertebral infection, intradural or extradural hematomas, herniated disc, and spinal stenosis. It is of interest that this particular patient had a decreased ankle jerk that cleared after the camptocormia went away. The nerve root involvement presumably was secondary to the abnormal posture. MRI scan would help considerably in the diagnosis. The next-best tests are myelogram and bone scan.

Numerous treatments have been tried, and none has been consistently satisfactory. They include: persuasion, electrotherapy (placebo), back board (strap the patient to a vertical board), lumbar puncture (placebo), thiopental sodium (Pentothal) interview(s), fingers up the wall gradual daily progress, medical discharge from military service, and psychotherapy. Patients do not respond to direct confrontation on the issue of "inconsistency," as manifested by the relief of the deformity upon lying down.

Muscle Spasms

Unilateral or bilateral paraspinous muscle spasm is not diagnostic of a herniated disc, but if present and involuntary, it is suggestive of organic disease. Paraspinous muscle spasm associated with nonorganic disease or hysteria tends not to relax on the side of the stance phase during ambulation. This can be tested by walking behind the patient with both hands on the paraspinous muscle masses.

Naffziger's Test

When positive, this is a significant indicator of intervertebral disc disease with nerve root irritation (Fig. 6-28). The mechanism involves increased nerve root pain within 15 or 20 seconds of bilateral jugular compression. This is due to increased pressure of the subarachnoid space at the intervertebral foramen. We have employed a slight modification of the test, which we find to be useful. While compressing the jugular, the patient is asked, "Does this make your leg pain go away?" If the patient responds "No, it makes it worse," the test is positive and suggestive of disc disease. If the patient says, "Oh yes, Doctor, it feels better!" this is suggestive of nonorganic or functional disease.

Forward Bending

The patient is then asked to bend forward. A cooperative attempt to do this along with simultaneous flexing of the hips and knees is suggestive of organic disease with or without nerve root irritation. With the attempt to bend forward, the lumbar region remains relatively fixed, while the rest of the spine moves above it. This is suggestive of organic disease. Dramatic refusal or half-hearted attempt is suggestive of nonorganic problems.

Percussion of Spinous Processes

Percussion over the spinous processes with a neurologic hammer sometimes elicits severe pain, localized maximally around two or three adjacent spinous processes. When this finding is consistent, it is suggestive of organic disease. There may be tumor, osteomyelitis, disc space infection, or a herniated disc. If increased vertebral fluid pressure

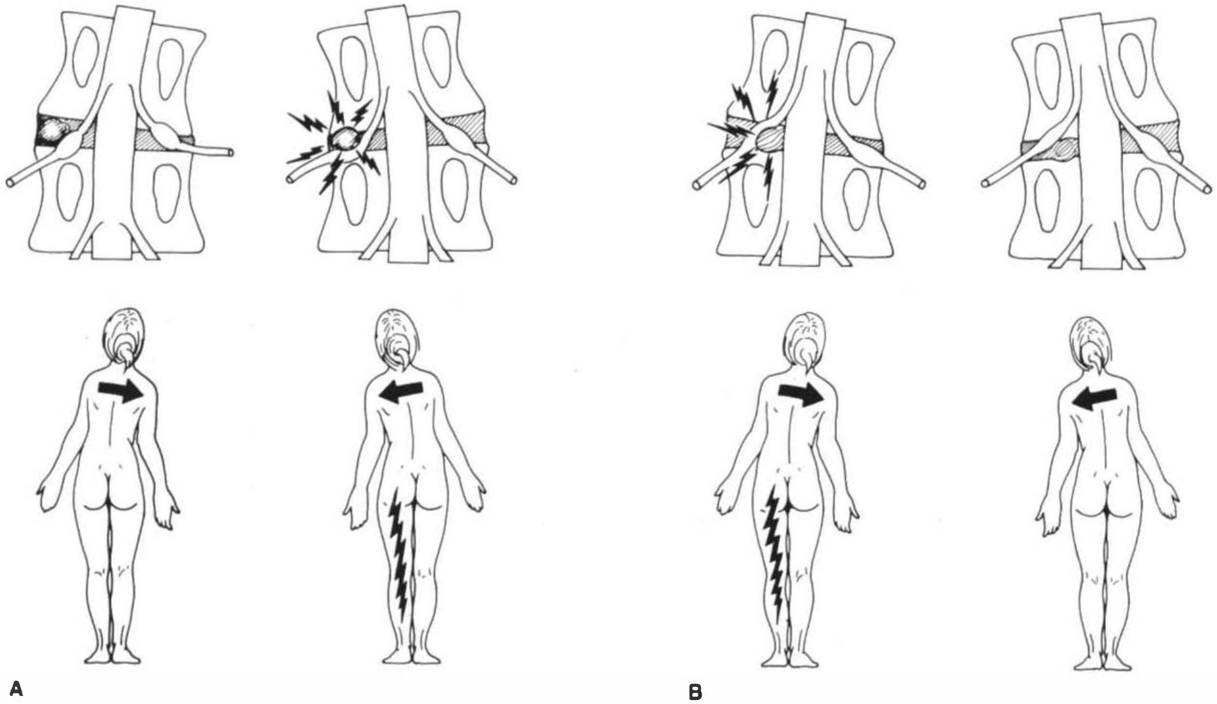


FIGURE 6-27 Patients with herniated disc disease may sometimes list to one side, usually the left. This is an involuntary mechanism. The list in some patients is toward the side of the sciatica; in others it is toward the opposite side. A reasonable hypothesis suggests that when herniation is lateral to the nerve root (A), the list is to side opposite the sciatica because a list to the same side would elicit pain. Conversely, when the herniation is medial to the nerve root (B), the list is toward the side of the sciatica because tilting away would irritate the root and cause pain. However, recent work has shown no correlation between direction of list and either position of disc or side of sciatica. There is a greater likelihood of a patient with a list having surgery.

does in fact cause spine pain, then this too would be stimulated by percussion.

Reflexes and Muscle Tests

Any gross muscle weakness from nerve root compression at L5 or S1 can be ruled out if the patient first is able to walk on the heels, fully extending the toes, and then subsequently is able to walk on the toes. When the patient walks on the heels, the absence of toe extension on one side may indicate weakness of the extensor hallucis longus and an L5 myotomal paresis. Decreased reflex of the Achilles tendon is suggestive of nerve root irritation at S1. A diminished knee jerk may be indicative of nerve root irritation at L3 or L4.

In addition to the previously described muscle power tests, we suggest several additional manual muscle tests. With the patient supine and both ankles held at 90° of extension, the examiner grasps a foot in each hand, holding them in the region of the metatarsal heads. The examiner then puts his feet together and with arms extended leans back so as to apply an equal force to the feet. This test will show weakness of the ankle dorsiflexors. The same test is done holding each of the patient's big toes. When these tests are positive, it is suggestive of nerve root irritation at L5. With both knees extended off the end of the table, the examiner can apply his weight equally to the dorsum and the anterior aspects of the distal tibias, and he can test the relative strength of the patient's quadriceps mechanism and the L3 and L4 root motor function.

A standard chart for localization of nerve root lesions is provided on page 416; however, it is important to point out that with muscle testing as well as sensory distribution, there is often anatomic varia-

FIGURE 6-28 Naffziger's test may be done while the patient is standing or lying down. The test is based on the hypothesis that bilateral jugular compression increases cerebral spinal fluid pressure. The pressure increase in the subarachnoid space in the root canal may cause back or leg pain by irritating a local mechanical or inflammatory condition.

tion on the dermatomal level or one myotomal level above or below the classic textbook description.

Leg Raising Tests

There is some confusion about the consistent nomenclature of some of the leg raising tests for examination of the lumbar spine. Some physicians consider the Lasègue's test to be a simple straight leg raising test.[138] Others believe it to be flexion of the hip followed by extension of the knee.[73, 107] Both interpretations are supported by reference to the original article by Lasègue.[195] This disagreement is of academic and historic interest. We suggest that the recent article by Dyck be allowed to put the issue to rest. His work shows convincingly that Lasègue never wrote about or described any such test.[88]

The straight leg raising test is done with the patient in the supine position (Fig. 6-29A). Most normal subjects can have the hip joint flexed 80–90° without back or leg pain. When there is back or ipsilateral leg pain, the test is considered positive. The sooner the pain occurs, the more definitive is the test. We do not think that the production of back pain without leg pain is as significant in this test as is the production of leg pain with or without back pain. If the test is consistent and is associated with voluntary extension of the lumbar spine to reduce the sciatic nerve stretch, it may be thought of as significantly positive. The examiner should take care to distinguish the discomfort associated with the stretching of a normal but tight hamstring muscle from leg pain similar to that for which the patient is being evaluated.

Recent research from the People's Republic of China[297] has shown that on the basis of the distribution of pain with the straight leg raising test, it is possible to localize the disc protrusion in the lower lumbar spine 88.5% of the time. If there is back pain with straight leg raising, the protrusion is likely to be central. Lateral protrusions cause leg pain, and those in between cause both leg and back pain. The investigators noted that a disc located in the axilla may be associated with the findings of back pain and ipsilateral leg pain when the straight leg raising test is performed. Also, contralateral straight leg raising may cause back and/or leg pain.

Straight leg raising with ankle dorsiflexion is a very useful test that is helpful in documenting the presence of nerve root irritation. It may be regarded as a type of check or confirmation of the straight leg raising test and a maneuver that distinguishes posterior leg pain from pain that may be due to stretching of the hamstring muscles. With the patient in a supine position, the straight leg raising test is done. The angle of hip flexion at which posterior leg pain is

LOCALIZATION OF LUMBAR NERVE ROOT LESIONS BY PHYSICAL EXAMINATION

Neurologic Level	*Weakness*	*Reflex Depression*	*Sensation Decreased*
L4	Anterior tibial, quadriceps	Knee jerk	Medial foot
L5	Extensor hallucis longus	—	Mid-dorsum of foot
S1	Peroneals, calf muscles, hamstrings	Ankle jerk	Lateral foot

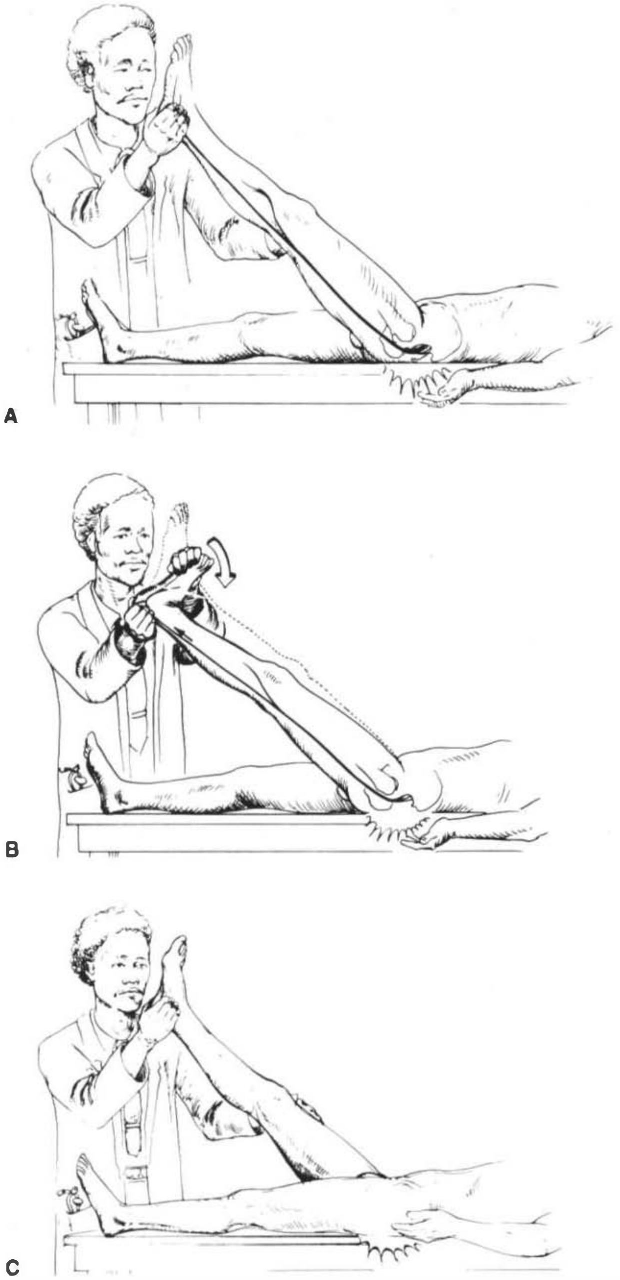


FIGURE 6-29 The various straight leg raising tests are useful clinical signs. They are based on the mechanical principle of stimulating an irritable situation at the lower lumbar region of the nerve roots and the intervertebral discs. The sciatic nerve is readily stretched by flexion of the hip and ankle (when the knee is straight) because of its anatomic location away from the motion centers of the two joints. **(A)** Straight leg raising test. **(B)** Straight leg raising with ankle dorsiflexion. **(C)** Crossed straight leg raising test. *(From Hoppenfeld, S.: Physical Examination of the Spine and Extremities. New York, Appleton-Century-Crofts, 1976.)*

elicited is found. The leg is then lowered to just below this level so that the pain subsides. While the leg is held at that level, the ankle is slowly but firmly dorsiflexed to the maximum (Fig. 6-29B). If this maneuver causes the patient's characteristic leg pain, the test is considered positive and strongly indicative of nerve root irritation.

Another reinforcing test for straight leg raising is the *internal hip rotation* test. In principle, it is analogous to the ankle dorsiflexion test. The straight leg raising test is taken to near the limit of the pain-free range. Then the hip is internally rotated. This puts tension on the sciatic nerve, and if it is physically irritated or inflamed, additional pain may be elicited. Concluding the test in this manner is thought to enhance the diagnostic value and repeatability of the test.[35]

The crossed, straight leg raising test (Bekhterev's test), although not often positive, is strongly indicative of a herniated disc or some other structure causing irritation of the nerve root.[309] This test is positive during flexion. The asymptomatic or relatively asymptomatic leg when flexed at the hip causes pain in the asymptomatic leg (Fig. 6-29C). The mechanism of this test is thought to be as follows: The inflammation or irritation of the nerve root of the symptomatic leg has sensitized it so much that the minute amount of motion produced by movement of the nerve root on the other side is enough to cause pain.

With conjoined nerve roots, a slight compromise of the intervertebral foramen can lead to symptoms of severe sciatica[262] (see Fig. 6-30). It is of interest here that in sciatica due to disc disease there is some possibility of relaxation of the nerve and roots with various positions. The nerve is relaxed with hips and knees flexed. When testing for straight leg raising, the knee is extended and the nerve is relatively relaxed; however, with a straight leg raising test, the irritated nerve is displaced in relation to the disc and it becomes more painful. In contrast, with radiculitis due to irritation and fixation of a conjoined nerve in the intervertebral foramen, relaxing the nerves by position has little effect, and thus the straight leg raising test is not operational. In this same view, bed rest is more likely to help disc disease symptoms than foramenal encroachment symptoms. Bed rest changes the mechanical variables to relax the nerve and take pressure off the disc. However, the bed rest makes relatively less difference in the mechanics of nerves pinched in the intervertebral foramen and unrelated to disc disease.

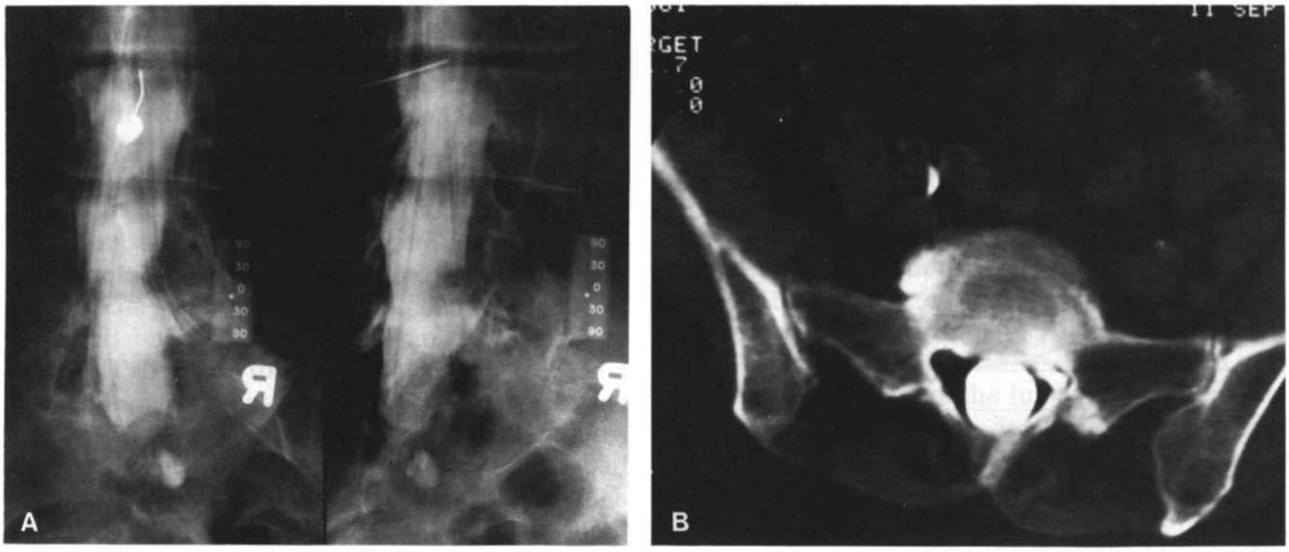


FIGURE 6-30 (A) These are the x-rays of an 80-year-old patient with severe sciatica unrelieved by several weeks of bed rest and not accentuated by the straight leg raising test (negative SLR test). There is an anteroposterior and an oblique view. Both show the two nerve roots entering the L5–S1 intervertebral foramen. Note relative radiolucency of contrast just at the entrance of the foramen, seen best on the oblique view. **(B)** It is interesting to note on this CT with contrast that with double roots entering the L5–S1 intervertebral foramen on the right, there is an associated absence of the S2 nerve root on the right in the upper portion of the sacral canal.

The Pedal Pulse Test

While the patient sits on the side of the examining table with hips and knees flexed, the examiner looks attentively at the dorsum of the foot, raises it by the heel with one hand, and carefully palpates for the dorsalis pedal pulse with the other hand (Fig. 6-31A). The heel is returned to its original position, and the same thing is done with the opposite side. This test provides considerable information. If both dorsalis pedal pulses are normal, it is highly unlikely that the patient's back or leg pain is due to occlusive vascular disease. (In order to be more certain of this, the examiner should determine that the femoral pulses are also present.) This maneuver also gives an excellent straight leg raising test, as the patient goes from a position of hips and knees flexed to one of a fully extended knee. This puts a large stretch on the L4, L5, and S1 roots, which contribute to the sciatic nerve. A patient with real sciatic nerve root irritation on the side that is being manipulated will *automatically* lean back on the examining table as the knee is extended, in addition to which the patient may share various exclamations with the examiner (Fig. 6-31B). By leaning back, the individ-

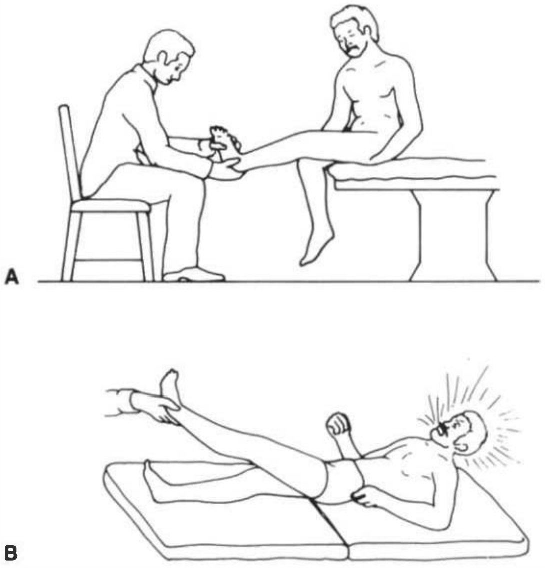


FIGURE 6-31 (A) The patient is relaxed on the table while pulses are examined. The straight leg raising here is greater than 90°. (B) Now that the patient is lying down, there is vigorous complaint with even less than 90° of straight leg raising. This is inconsistent and suggestive of other than organic disease.

ual is essentially *extending* the hip joint to compensate for and reduce the pain caused by excursion of the sciatic nerve. Another valuable aspect of this test is its potential to provide evidence supportive of nonorganic or functional disease. Consider the patient who sits quietly while the knees are extended to 180° and the hips are flexed at 90° as the pulses are being checked, but who complains vigorously of pain when the straight leg raising test is being done in the supine position. This may be thought of as a "positive pedal pulse test," suggestive of nonorganic disease.

This is actually one of Wadell's tests of nonorganic physical signs of low back pain. This entire group of physical tests is presented in the following section.

Waddell's Tests

This is a series of tests that have been developed, shown to be correlated with an abnormal Minnesota Multiphasic Personality Inventory (MMPI), and thought to be representative of nonorganic pain behavior. The findings are listed and the tests are described below.[342, 368]

Scoring—if three out of the five tests listed here are positive, then nonorganic psychologic pain behavior is likely.

1. *Tenderness*—This category should be scored positive if light touch or rolling of the skin on the back causes pain, or if deep tenderness is spread over large areas of the body.
2. *Simulation Test*—This is scored positive if gentle axial rotation of the pelvis and shoulders causes back pain or if *light* pressure *of 1–2 lb of force* applied to the head by the examiner's hand causes back pain (see Fig. 6-32).
3. *Distraction Test*—This is scored positive if the patient's pain, which is present in some activities, is not present when those same activities occur in a different context—for example, if the straight leg raising test is present in the supine position but *totally* absent when sitting. This is considered a positive distraction test.
4. *Regional Disturbances*—This is a positive score when the patient demonstrates "cogwheel" weakness* or non-neuroanatomic numbness, such as stocking and glove numbness in the absence of peripheral neuropathy.
5. *Overreaction*—A positive score is awarded here when there is excessive body language, grimacing, verbalization, groans, tremors, collapsing, excessive sweating, or stumbling.

Gluteal Skyline

Katznelson and associates have made a new observation in the physical examination that my be helpful in the evaluation of a patient with lower lumbar nerve root involvement.[171] The test is based on the recognition of the fact that L5, S1, and S2 myotomes

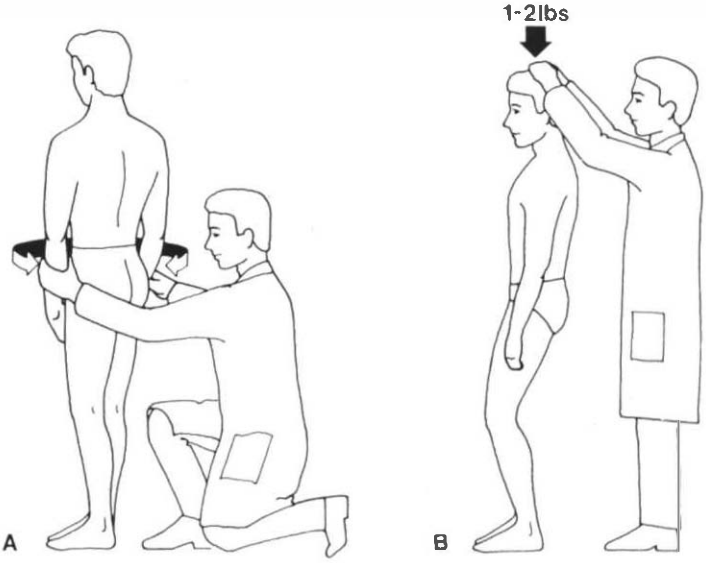


FIGURE 6-32 The simulation tests of Waddell involve: (A) simulations of rotation of the hips and shoulders by the examiner and (B) pressing gently on the head with 1–2 lbs of force. If either of these causes "back pain," the test is considered a positive nonorganic test and 1 point is assigned.

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* Neurologists sometimes use the term "cogwheeling" to indicate a type of spastic weakness that is associated with extrapyramidal systemic disease. Here we refer to a situation in which there is a "give-away weakness" followed by increased resistance.

innervate the gluteal muscles. Therefore, it is possible to recognize gluteal asymmetry from the "south" end of the examining table with the patient lying prone and contracting the buttocks. Asymmetry of the gluteal skylines is considered evidence of possible L4–L5 on L5–S1 disc herniation. The test was positive in just 60% of patients with myelographic evidence of disc disease; however, in 13% of these patients, this was the only physical finding. Moreover, it was second as a correlate to disc pathology only to positive straight leg raising with positive ankle dorsiflexion. We believe that the sign is useful and should be looked for especially where disc disease is suspected but unsupported by other objective findings on physical exam. We sometimes call it the "A/2 sign."

Kernig's Test

When there is enough mobility of the neck and significant encroachment and/or irritation of the meningeal structures, this test may be positive. Even though there is considerable accordionlike folding of the cord, there can be sufficient motion transmitted to the cervical or lumbar region to cause neck, back, or leg pain with flexion of the neck (Fig. 6-33). Biomechanically and pathoanatomically, this test may be regarded as analogous to a leg raising test performed from the opposite end.

Of the tests discussed here, we consider the following to be the most reliable: the crossed, straight leg raising test; the straight leg raising test with ankle dorsiflexion; protective flexion of the spine during the pedal pulse test and straight leg raising tests; the standing posture; deep tendon reflex changes; and distinct localized muscle weakness. Sensory examination is not discussed here. This, too, is important for a complete physical examination. However, it is often difficult to evaluate. Certainly, some of these other findings must be present in addition to sensory changes in order to convincingly indicate the presence of nerve root pathology.

FIGURE 6-33 Kernig's test is based upon the mechanism of increasing the tension in the meninges or the nerve roots by flexing the neck as shown. The test is positive when back, leg, or arm pain is elicited.

Dejerine's Test

This test is used to corroborate or elicit a history of sciatic pain due to the cough, strain, or sneeze effect. Patients with acute radiculitis due to nerve root irritation may have a positive Dejerine's test. The patient lying supine is asked to put his hands behind his head and raise his upper torso off the examining table or bed. If the abdominal pressure and secondarily the CSF pressure is increased when this is attempted, the patient may experience leg pain. If so, the test is considered positive, and it is assumed that there is nerve root inflammation secondary to disc herniation.

Prone Knee Flexion Test

This test, described by Herron and Pheasant,[144] is used to produce or enhance Achilles tendon reflex suppression or motor weakness. The patient is prone on the table with both knees maximally flexed. Motor strength and reflexes are tested with the lumbar spine in considerable extension. This can be expected to maximally irritate the nerve root through posterior disc bulging and possibly some foramenal constriction.

Imaging and Electrical and Psychologic Evaluation of Spine Pain

We agree with the Spitzer study recommendations that for patients in the age group 18 to 50 with no extenuating circumstances such as specific trauma, no diagnostic tests beyond the completion of a history and physical examination are needed within the first 7 weeks following the onset of symphysis.[315] Obviously, the history or physical exam may indicate other studies within this time frame.

Magnetic Resonance Imaging (MRI)

MRI or nuclear magnetic resonance (NMR) imaging technology is progressing and developing rapidly.[318] This technology, when more available, may become the *first* imaging study of choice in the evaluation of spine pain patients. This modality has proved useful in the diagnosis of lateral canal entrapment and in the recognition of a degenerated disc and spinal stenosis.[64] A more recent use has been to help differentiate a disc herniation that is still contiguous with the nucleus pulposus from one that is sequestered and extruded and no longer in continuity.[339] This differentiation may prove to have important therapeutic implications in the likelihood that chemonucleolysis or percutaneous lumbar discectomy will be successful for a given patient.

As may be expected, there is a significant (20–30%) incidence of disc herniation observed in asymptomatic or normal individuals. This is a point worthy of attention and reflection.

CT Scans

Computerized tomography of the spine has come to play a major role in the imaging of patients with a variety of spine problems. A symposium in *Spine* (1979)[328] provides an excellent overview of the basic information and the capabilities of the technology. There are also ample superb clinical examples for study.

Burton and colleagues emphasized the use of CT to diagnose central and lateral stenosis in the lumbar region.[43] This has been a technology that has grown and developed many uses in recent years.

Myelograms

Amidst considerable enthusiasm for CT scanning, a well-known and controversial study was completed by Bell and associates.[19] The investigators collected the x-rays of 122 patients with known disc disease confirmed at surgery and had them read by several neuroradiologists who did not know the diagnosis. The conclusion was that metrizamide myelography is more accurate than CT scan in the diagnosis of herniated lumbar discs. A rebuttal of Bell's study and of a study by Wiesel and associates was published.[273] Experienced neuroradiologists criticized these two studies and indicated that the superior technical quality of *their* scans and *their* interpretations would not allow as many false-positive readings.

Wiesel[359] found that 35% of asymptomatic patients had abnormal CT scans. Based on the reading of these films, a diagnosis of a herniated disc was made in 19% of the study group under 40. A 2% incidence of false-positive myelograms has been reported.[152] The false-negatives with this test have been reduced by the introduction of the water-soluble material metrizamide as the contrast medium. This material was not allowed in the United States before 1976.[232]

These are some interesting comparisons. With more experience, the most appropriate use of the CT scan and myelogram in diagnosing spine problems will clearly emerge. MRI may come to be the technology of choice as clinical experience and availability increase. We are gradually replacing the use of the myelogram with MRI.

Discograms

The term discogram has at least three different connotations. It sometimes refers to the amount of fluid that can be injected into a particular intervertebral disc. It also has to do with interpretations of the location and distribution of radiopaque fluids following their injection into an intervertebral disc. And finally, there is the use of injection as a clinical test of the extent to which this manner of irritation of the disc elicits the patient's characteristic pain.

The problem with discograms is that it is not possible to distinguish the distribution of the contrast media in the normal disc from that reported in the pathologic disc. Holt was able to do cervical and lumbar discograms on groups of volunteers from the Missouri State Prison. The study of cervical discograms was done with 50 subjects, ages 21 to 50, without histories of neck or arm pain or injuries to the cervical spine. In only 10 disc spaces out of 148 injected did the contrast medium remain within the confines of the annulus. In addition, the volume of the injectable material was not a useful indication of an abnormal disc. Severe pain was produced by the injection of the contrast medium in every subject. This eliminated pain production as an indication of an abnormal disc.[158] Holt conducted a similar study in the lumbar spine. There he found 37% false-positives in 30 normal volunteer prisoners from Illinois.[159]

Clearly, the ability of the intervertebral disc to retain a fluid within its confines is not related to its ability to cause spine pain. There does not seem to be adequate evidence that discograms provide any useful diagnostic information with regard to localization of spine pain.

However, there are studies that support the assertion that discography has been useful in identifying the cause of sciatica in unusual circumstances of relationships between herniated discs and nerve roots.[192,218] A discogram may be helpful when there is good clinical evidence of sciatica with poor or absent imaging evidence to explain it.

Currently, discograms are used with considerable enthusiasm by some clinicians. The procedure is also being used in combination with CT scans. Some investigators have considered this useful, particularly in the evaluation of difficult clinical low back pain problems.[18] However, the precise indications and documented sensitivity and specificity of the procedure and its variations await further investigations, and the discogram remains controversial. A superb editorial comment by Nachemson[234a] reviews the cogent points documented in the literature and closes with the statement that discographic studies should not be done except in the context of prospective clinical studies approved by human investigation committees.

EMG, F Response, and SEP

Electromyographic studies, including fibrillation potentials, H-reflexes, and ankle reflex latencies, have been shown to be more accurate than radiculography or clinical examination in the diagnosis of lumbar root compression.[201] These studies did not give false-positive results.

A study of the correlations of several variations in F response latencies showed a good correlation with imaging evidence of disc herniation.[66,144a]

Somatosensory evoked potentials (SEP) were shown to be useful in the localization and documentation of the site of laterality of nerve root entrapment in spinal stenosis patients.[186]

We have found these more objective electrical assays to be helpful in the evaluation of complex clinical problems in which the evidence for radiculopathy is equivocal, confusing, inconsistent, or atypical. The presence of one or more abnormal electrical assays can be crucial in the evaluation and management of a patient when compensation, litigation, and psychiatric problems are significant factors. Remember that the EMG will not be positive within 14 days of the onset of root pathology.

Psychologic Testing

Several psychologic tests may be used in the evaluation of patients with pain—particularly spine pain. Southwick and White[308] have reviewed the various tests in the context of their use in the evaluation of patients with low back pain. The authors stated that the Minnesota Multiphasic Personality Inventory (MMPI), the Middlesex Hospital questionnaire, and the Mooney pain drawing diagram were very helpful diagnostic tools.

It is appropriate, if not auspicious, that psychologic testing is the last evaluation to precede the section on the treatment of spine pain. We know that psychologic factors play a significant role in chronic back pain. Psychologic analysis and evaluation prior to treatment of chronic spine patients is important for several reasons. First, many of these patients may have any of the following personality problems: distrust, alienation, impulsiveness, poor frustration tolerance, or excessive extroversion, or they may be demanding, somatically preoccupied, overly dependent, or preoccupied with feelings of anxiety, inadequacy, and inferiority. Certainly, it would be helpful to know if any given patient had one or more of these personality traits. Second, psychologic factors are useful predictors of treatment outcome. The best predictions are obtained with the somatic scale (measuring somatic concern) of the Middlesex Hospital questionnaire and the hypochondriasis hysteria scale of the MMPI.

Pope and colleagues[263] noted that positive psychologic factors were associated with biomechanical factors of good mobility and spine flexion/extension muscle balance, in contrast to the association of negative psychologic factors with reduced spinal mobility and altered spinal flexion/extension muscle balance. Although the practical implications of these observations are not clear, they do add significantly to the impact of psychologic factors on the clinical manifestation of the disease.

Some investigations have shown that, unlike nonspinal orthopedic problems,[133] psychologic involvement in chronic low back pain is more predictive of surgical outcome than is an estimate of the organic component.[308] It appears that a battery of psychologic tests enhances the predictive strength. Most probably, the work test score is the best predictor. Psychologic evaluation of chronic back pain patients can be expected to improve our overall care and management of these patients.

TREATMENT OF SPINE PAIN

Well-controlled prospective clinical trials in orthopedics are rare.[74,272,284] There are several reasons for this, and most of them have been addressed. The active clinician, particularly the surgeon, lacks the time and sometimes the expertise to design and complete an excellent clinical study. The basic solution is to collaborate with a statistician and a research nurse or some other person who has the research project as a primary responsibility. This is a very expensive approach, and clinical research is difficult to fund. Even with the necessary resources, clinical research, particularly that involving surgical procedures, has some significantly challenging problems of both design and execution.[272,284] The surgical procedure cannot be double-blinded or even single-blinded. The reluctance on the part of patients and surgeons to participate in a randomization process is formidable. There is also the problem of different technical skills and experience among surgeons, which can make it difficult to control the procedural (treatment) aspect of the experimental design. Some useful suggestions have been put forward by Rudicel and Esdaile as a rational alternative to the double-blind and randomization dilemmas imposed by surgical therapy.[284] It is imperative that these problems be solved. There is a great need for more well-designed and well-executed clinical studies in the nonoperative and operative treatment of spine pain. The works of Deyo,[76] Rudicel,[284] and Raskob[272] and their respective associates are highly recommended as excellent sources of understanding and inspiration to those interested in this enormously important clinical research in orthopedic surgery and related disciplines.

There are few ideally designed prospective, controlled studies to guide the clinician in decisions about treatment of spine pain.*[74] This is unfortunate, and physicians are currently striving to improve the situation. At present, however, there are patients to be taken care of and treatment decisions to be made. Recognizing that the "gold standard" for decision making is the controlled, prospective, double-blind study or some appropriate modification of it for "surgical treatment,"[284] we submit some suggestions for today's treatment decisions.

Recommended Guidelines for Decisions in Treating Today's Patients

The diagnosis of the condition being treated is confirmed based on generally accepted clinical criteria.

The treatment in the common vernacular makes sense.

The treatment is *rational* and *logical* based on generally accepted knowledge about the pathoanatomy and/or pathophysiology of the condition being treated.

The risks/benefits for the particular patient are mutually discussed and clearly understood by both patient and physician, and both parties agree that the reasonably presumed benefits are *worth* the inherent risks.

New information from both ordinary and ideal studies is taken into consideration and factored into this framework. New etiologic, pathologic, diagnostic, risk, and innovative information can readily be included in these four components of decision analysis. Considerations of cost have not been included in the chart. The decision is easy when costs are accurately determined, and treatments with the same risks/benefits for a given patient can be selected on the basis of cost-effectiveness. Beyond this, however, we venture into complex and profound political, ethical, and philosophic considerations that are not the purview of this text.

The problem of the evaluation of various treatment programs for spine pain is an extremely difficult one and should be approached with deference, humility, patience, and determination. The numerous psychologic and socioeconomic factors that are involved have been discussed. The issue is further complicated by two additional factors. The first is that in 30–35% of patients there are placebo reactions to any form of treatment, including surgery.[78,320] The second is that the natural course of disease(s) that causes spine pain is such that 90% of the patients will be significantly improved within 2 months with either no treatment or some form of treatment.[138,164,235,309,347] It is therefore extremely difficult in many instances to know if a patient recovers because of a placebo response, because the treatment was effective, or as a result of the natural course of the disease. In addition, for those patients who do not show good results, it is sometimes difficult to determine if it is because of incorrect diagnosis, the particular character of the disease, insurmountable psychologic problems, or inappropriate, untimely, or ineffectively administered treatment. With this optimistic, inspiring background in mind, an attempt to review and evaluate some of the more widely used treatments for spine pain follows. Most of the information relates to low back pain and sciatica. However, the basic principles apply to both cer-

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* Personal communication, Gary Onik, 1988.

vical and thoracic spine pain. The unique considerations related to cervical and thoracic pain are discussed when pertinent.

It is understandable that some physicians do not like to treat patients for low back pain. However, there are few diseases in which one is assured improvement in 70% of patients in 3 weeks and 90% in 2 months, regardless of the type of treatment employed.[232] Given the present state of knowledge and the objective information about the various forms of treatment for spine pain, it is possible to build an argument for withholding treatment. However, this is not feasible. Patients expect and demand to be "treated" and relieved of their misery. The physician is compelled to do something. Therefore, the main goal is to make the patient rest comfortably, to maintain confidence, and to ensure against the occurrence of anything that is unnecessary and/or potentially harmful to the patient's physical and fiscal well-being.

This evaluation applies to patients who have spine pain that has been adequately evaluated and diagnosed as a herniated disc or organic spine pain of undetermined etiology. The patients do not have tumors, infections, specific arthritis (e.g., rheumatoid arthritis, ankylosing spondylitis, lupus erythematosus), significant trauma, or some other systemic disease that is the cause of spine pain. This evaluation does include osteoarthrosis of the cervical, thoracic, and lumbar spine. There are certainly an ample number of treatment options that are available to the patient with spine pain (Fig. 6-34). Since it is sometimes difficult for the specialist to choose among them, one can be sure that the patient probably has even greater difficulty deciding.

Rest, Analgesics, and Anti-inflammatory Drugs

In most instances of spine pain, the patient improves in 2 to 3 weeks. Rest minimizes mechanical irritation of what may well be a local inflammatory response. Anti-inflammatory drugs contribute to the alleviation of symptoms, as do analgesics.

Treatment

In the acute situation, the patient may be treated with bed rest, preferably on a firm bed[121] or surface, and given analgesics. The position suggested is lying either on the back or the side, with hips and knees flexed (Figs. 6-35, 6-36). This effectively reduces the

FIGURE 6-34 This cartoon includes most of the options currently employed by patients and therapists in the treatment of spine pain. There must be a best choice, and medical science should continue to search for it.

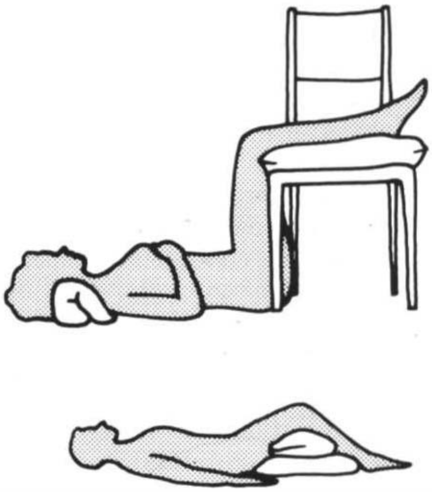

FIGURE 6-35 These positions have several important characteristics that should be beneficial to most patients with low back pain. The supine position reduces disc pressure. The straight back minimizes posterior disc bulging. With the hips and knees flexed, there is elimination of psoas muscle tension and thus of disc pressure, and there is minimal stretch on the sciatic nerve.

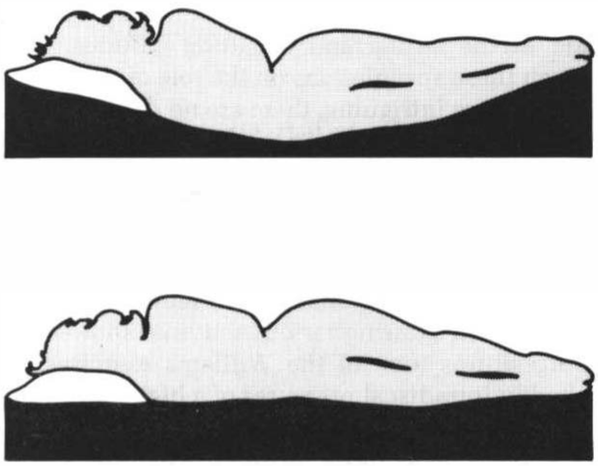

FIGURE 6-36 (*Bottom*) The firm mattress allows one to splint the spine better and reduce the tension associated with excessive curves. If the patient lies on the side with hips and knees flexed on the firm mattress, he can maintain the position and accomplish essentially the same goals achieved when the hips and knees are flexed in the supine position. (*Top*) Bending in any plane maintained for a prolonged period of time can cause excessive stress on the disc.

loads on the lumbar intervertebral discs.[232] The patient should assume the most comfortable position and level of activity. Some patients will be most comfortable on the side, lying in the fetal position with hips and knees flexed. This position obviously puts minimal stretch and irritation on the sciatic nerve. Some patients treated with bed rest will find that they are more comfortable if they can be up and about intermittently for minor activity. We actually advise against strict bed rest because this can be inconvenient to say the least. Moreover, on a theoretic basis, some motion is probably good for disc nutrition[155,191,337] and is of some preventive value in regard to venous thrombosis. The recommended time for bed rest must vary with the individual patient. However, a recent study suggests that for backache with sciatica, 2 days of bed rest seems to be as effective as 7 days.[76]

Results

The treatment is usually successful in 1 to 3 weeks. Occasionally, the patient's condition will worsen or proceed to a protracted course. In either instance, additional, similar therapy may be continued, or some other treatment program may be substituted or added.

Complications

There are the usual risks of bed rest in the middle-aged and aged. Prolonged bed rest has liabilities for any adult. These include depression, gastrointestinal disturbances, and loss of muscle mass. There are also the usual pharmacologic complications associated with the commonly used analgesics and anti-inflammatory drugs. The risk of addiction to narcotics is present, but this is not a high risk.

Comments

Bed rest at home is inexpensive, low-risk treatment that does no harm and generally is associated with relief of symptoms. It is a noninvasive treatment requiring no particular technical knowledge or experience. The use of medications in addition is a reasonable adjunct that makes the patient more comfortable and, in the case of the anti-inflammatory drugs, may accelerate the recovery rate. We recommend this regimen as the initial treatment of spine pain with or without sciatica.

The value of resting on a hard surface may simply be due to the more efficient, *immobilizing* capacity. By eliminating the bending of the spine that occurs because of a soft mattress, the patient is better able to attain and maintain a constant position of the spine without sagging and bending (Fig. 6-36).

Garfin and Pye have completed a useful and interesting study of chronic low back pain patients that compared the use of four different types of beds. The best was a 720-reinforced-coil "orthopedic" bed. This would be considered a *hard* bed. The next best was a standard 10-in-thick waterbed. A softer bed (500 coils) and a mixed foam and water bed were of no benefit.[121]

In addition to the above regimen, there are a number of nonsurgical treatment programs of varying degrees of complexity, intensity, and risks that may be employed. They are used independently or in a variety of combinations and sequences. Many of the conservative treatment modalities were evaluated in a study by Soderberg.[306] He found that 67% of patients with sciatica treated with a combination of nonsurgical techniques, including bed rest, physiotherapy, plaster jacket, manipulation, local injections, and systemic medications, were symptom-free in an 8-year follow-up. A review of some information about the various treatment programs and some comments follow. An excellent updated review of the conservative treatment of low back pain has been completed by Deyo.[74]

Medications

During the time interval when spontaneous remission of symptoms is being awaited, it is desirable to provide some medication to alleviate the pain. Most physicians tend to employ some combination of narcotic or non-narcotic analgesic, anti-inflammatory, muscle relaxant, and psychotropic drug. The choice of drugs is usually the result of some interaction between the knowledge and attitudes of the doctor and patient. We generally start with salicylates supplemented with codeine, if needed in the acute phase. Rest and reduction in pain with analgesics tend to relax the muscles. If the patient has psychiatric problems needing medication, psychiatric consultation is advised to help with the medication and the general management of the patient.

Williams Exercises

The therapeutic goal of the classic Williams exercises is to strengthen the lumbar spine flexors and stretch those muscular and ligamentous structures that tend to hold the spine in the extended position.[364] The premise is that the straight or slightly kyphotic lumbar spine is less painful. Supporters argue that people who live in Asian and African cultures, where a good deal of time is spent in a squatting position with the lumbar spine flexed, do not have as high an incidence of spine pain.[97] A radiologic study of cultural squatters and nonsquatters showed a greater incidence of lumbar spondylosis in the sitting and squatting cultures.[98] Although these speculations on the role of sitting and squatting are intriguing, there are no data about the association with back pain to substantiate them. We are not aware of an investigation in which cross-cultural comparison has ever been satisfactorily studied.

Another important point related to Williams exercises should be discussed. Intradiscal pressure measurements during various activities showed that doing sit-ups (one of the Williams exercises) resulted in intradiscal pressures of a high magnitude, as measured at the L3 disc. The observed pressures were the same as those recorded when the subject was lifting 20 kg (44 lb) by bending the back with the knees straight. This is hardly a task that the clinician should recommend as being therapeutic for a patient with acute low back pain. Of course, we are not sure that this organic, undiagnosed spine pain is due to disc disease, but assuming that it is, we consider sit-ups to be contraindicated in patients with acute or subacute lumbar spine pain, and they are probably not advisable as part of an exercise program for people over 40 years of age.[C]

Treatment

The exercises are shown in Figure 6-37. They are usually prescribed in conjunction with other forms of physical therapy, such as heat and massage. The

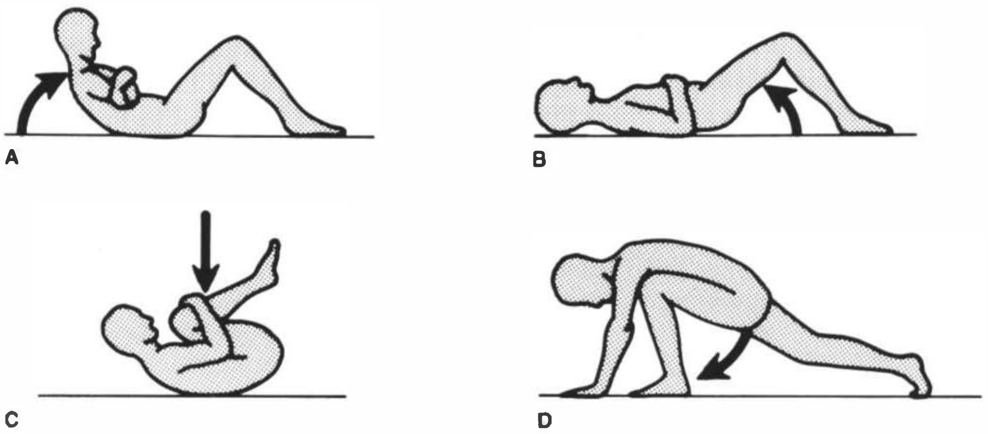

FIGURE 6-37 The goal of these exercises is to attain a less extended and a more flexed position for the lumbar spine. Exercise A strengthens the abdominal muscles, which can increase flexion. Exercise B, the pelvic tilt, strengthens the muscles that rotate the pelvis (about the x-axis) so as to reduce lumbar extension. Exercise C is designed to stretch the posterior structures, which allows more flexion. Exercise D is designed to stretch the hip flexors, which, when tight, also contribute to an extended or lordotic position of the spine.

exercise program is more commonly used in the subacute and chronic low back pain syndromes.

Results

We are not aware of studies that specifically evaluate Williams exercises.

Complications

There do not appear to be any documented examples of complications related to the use of Williams exercises.

Comments

These exercises are based upon the assumption that achieving and maintaining a flexed lumbar spine is preferred. This has some theoretic support in lumbar spine biomechanics. However, the sit-ups exercise, which is a part of the Williams exercise program, may aggravate symptoms, especially in the acute phase, because of the high loads (comparable to lifting 44 lb improperly) exerted when doing sit-ups. (See also the general comments at the end of this section on exercises.)

Trunk Muscle Exercises

The rationale for this form of treatment is based on several different observations. The load-bearing capacity of the lumbar vertebrae *in vitro* is significantly less than the load-bearing capacity in vivo[17]; a schematic analysis of the mechanics of the spine during lifting shows that the position and rigidity of an air- and fluid-filled column can efficiently and effectively reduce the weight on the lumbar spine.[335] It has been shown that the ability of a weight lifter to generate large thoracic and abdominal pressures is correlated with the amount of weight that can be lifted.[90,91]

A more recent study by Fairbank and associates concerns intra-abdominal pressure rises in males lifting in a variety of different ways. The study showed higher rises in intra-abdominal pressure in those who had pain.[99] The meaning of this is unclear, and it is tempting to speculate that this may represent some unconscious preventive mechanism.

Gracovetsky and colleagues[132] submit an interesting hypothesis based on biomechanics theory and the anatomic structure of and interplay between the abdominal muscles and the lumbodorsal fascia. The fascia attaches to the spinous processes, the hips, the pelvis, and the *transversus abdominis* and the *internal oblique* muscles. The Poisson's ratio of the fascia is 1 (see Chap. 9). The authors indicate that since there is *no* narrowing of the fascia with flexion, it must be assumed that the two abdominal muscles contract to prevent the narrowing. This represents a mechanical conversion of a lateral pull into a longitudinal tension. Suzuki and Endo,[325] using a quantitative isokinetic dynamometer, studied patients with low back pain syndrome. They were able to show that these patients had weaker, more fatigable trunk muscles. Soderberg and Barr[305] used EMG to study trunk flexors and extensors in healthy and chronic low back pain subjects. They found significantly different activities of the muscles in the two groups during Valsalva maneuvers and portions of sit-ups.

An important investigation was conducted by Halpern and Bleck[139] in which they used EMG to evaluate abdominal muscle in five different types of sit-ups. They opined that the best sit-up was one in which there was the *greatest* amount of abdominal muscle activity and the least amount of lumbar flexion. This was achieved by doing a sit-up in which the knees and hips are flexed and the trunk is flexed just enough to lift the scapula off the mat (Fig. 6-37A). We note that Grew[136] questions the mechanism of spinal support by the abdominal muscles. Nevertheless, the several studies presented here provide a rational basis, though not strong evidence, for using abdominal strengthening exercises for patients with low back pain.

We will also discuss some of the studies that address specifically the erector *spinae* muscles. Addison and Schultz noted as much as 50% extension weakness compared with flexion and extension in patients being hospitalized for low back disorders.[4] Poulsen demonstrated the importance of the strength of the erector spinae muscles through isometric studies of the back muscles.[268] McNiell and colleagues[217a] compared healthy subjects with patients with low back disorders and discovered that the latter group (especially those with sciatica) had low performance on tests of extension strength. Schultz and Andersson,[290] who analyzed loads in the lumbar spine using several methodologies, demonstrated the importance of the trunk muscles, the abdominal muscles, and the erector spinae muscles. Recent modeling and myoelectric trunk muscle work by Schultz and associates[290a] suggested that trunk muscles, through intra-abdominal pressure and tensing of ligamentous tissues, may play a role in resisting twisting-type injuries. Tesh and associates[331a] conducted cadaver studies using large latex balloons and a methodology to compare the role of increased abdominal pressure with thoracolumbar fascia in stabilizing the spine. The fascia was thought to play the major role. Here again is a rational though unproved basis for strengthening these muscles.

The evidence for assuming a relationship between trunk strength and back pain is of three general types:

- The trunk muscles are active in lifting.
- Subjects with back pain have different (or abnormal) characteristics of trunk muscle strength and fatigability when compared with normals.
- There are biomechanical and anatomic reasons for assuming that muscles are important in spine function.

Truncal (abdominal and thoracic) muscle tone is of some importance in protecting the spine from the loads that are applied to it, especially when lifting. They also help to maintain the lumbar spine in the less lordotic, more therapeutic flat back position (Fig. 6-38). Intra-abdominal and thoracic cage pressure may be factors in providing strength and mechanical stability to the spine. We have suggested that this situation is somewhat analogous to a football in the abdomen (Fig. 6-39). The abdominal and thoracic air and fluid contents are compressed, creating turgor in the soft tissues and providing support. Thus, it is suggested that these muscles be successfully toned and conditioned by exercises.

This rationale is based on a hypothesis. There is not substantial data to support it at present. Nevertheless, no harm is done by trunk muscle exercise, so many clinicians recommend it.

Treatment

The isometric truncal exercises are done as follows. The patient is told to inhale normally, to close the windpipe and the rectal and urinary sphincters tightly, and to push hard with the trunk and abdominal muscles. In other words, the patient should push against the windpipe as though blowing up a hard balloon and push against the closed rectal sphincter as though constipated. The idea is to have the patient maximally compress the thoracic and abdominal contents against a closed glottis and the perineal sphincters with all available truncal musculature.

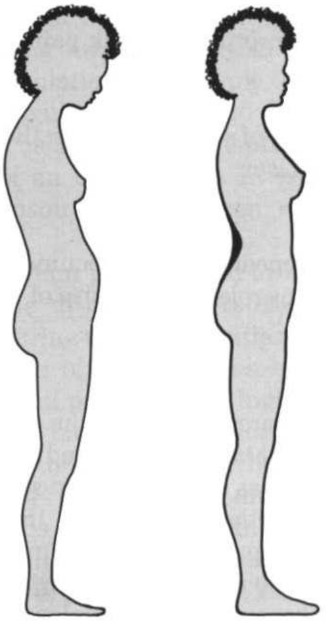

FIGURE 6-38 The posture of patients with a lordotic spine (*left*) is associated with posterior bulging of the disc and also with greater intradiscal pressure. Both factors are reduced considerably by correct posture (*right*), which is maintained by good abdominal musculature, also of therapeutic value.

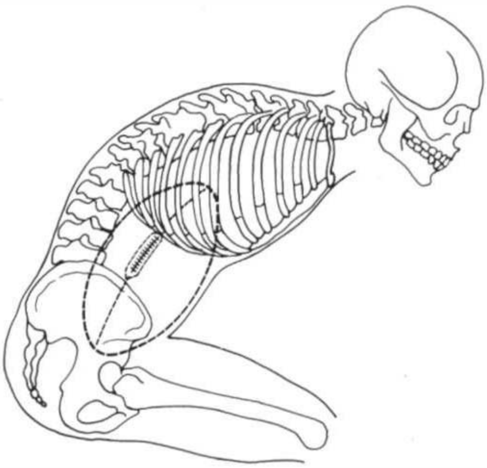

FIGURE 6-39 The football shown here within the body cavities is analogous to the turgor created in the abdominal, pelvic, and thoracic cavities through the compression of fluid viscera and air by contraction of the truncal muscles, primarily the abdominals. (*White, A. A., Southwick, W.O., Panjabi, M. M., and Johnson, R. M.: Practical biomechanics of the spine for orthopaedic surgeons (Chapter 4). In Instructional Course Lectures, American Academy of Orthopaedic Surgeons. St. Louis, C. V. Mosby, 1974.*)

This should be done at least 10 to 15 times, holding the contraction for 3 to 5 seconds, 3 to 4 times per day. Because some of the experimental studies have emphasized the role of the erector spinae muscles, we believe that they too should be exercised. We have used head and upper chest raises in the prone position and the contralateral arm and leg lift exercises shown in Figure 6-40.

Because of the ergonomic relevance of proper lifting, it has been suggested that quadriceps exercises be included in the exercise therapy for low back pain.[229]

Results

A double-blind study compared back extension exercises (strengthening paravertebral extensor muscles), mobilizing exercises (mainly involving flexion), and *isometric abdominal exercises.* There was a distinct and statistically significant superiority in the patients treated with the isometric abdominal exercise program.[164]

In another study, conventional physical therapy (*i.e.*, heat, massage, extension and flexion exercises) was compared with a program that involved isometric abdominal exercises and axial pelvic traction. There was also a control group that received only heat treatment. Patients who were treated with isometric abdominal exercises with traction did significantly better statistically than did the other two groups.[202]

Based on the preceding studies as well as others examining the biomechanical functions of the spine, it is to be expected that the isometric abdominal exercises offer considerable support to the spine with minimal negative risks.[229,232]

Complications

The potential problems with this treatment are related to Valsalva's maneuver. If the patient is still in the acute phase of symptoms because of nerve root irritation, the exercise may cause considerable pain. The more important complication may occur in an individual with heart disease. In this situation, the alteration in pulmonary and myocardial circulatory dynamics associated with Valsalva's maneuver may cause myocardial ischemia. Thus, the exercise should be avoided in patients with heart disease. Such patients may have to use a well-fitted thoracoabdominal corset or some other spinal orthosis with an abdominal support.

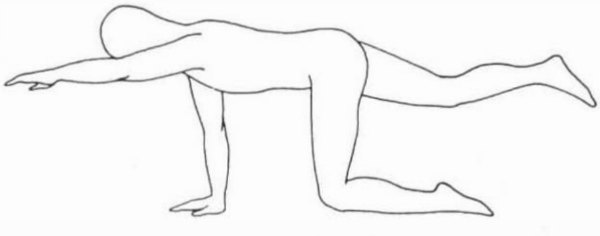

FIGURE 6-40 An excellent, though not easy, exercise to strengthen the erector spinae muscles.

Comments

Although several exercises have been discussed in some detail, we would like to submit our overview of the role and significance of exercise in the management of low back pain. We must first acknowledge that we don't know enough about the role of exercise in low back pain. Nevertheless, present knowledge permits the following reasonable assumptions on the topic.

First, regular aerobic exercises, if of adequate intensity, can cause endorphin secretions, which help control pain and elevate mood. Second, a patient actively participating in an exercise program is "working to help himself or herself" while time is passing and the natural course of the problem may be moving toward resolution. Third, there is good physiologic evidence that exercise, through its mechanical pumping mechanisms, improves nutrition of the intervertebral disc.[155,191,337] Exercises that build the strength and endurance of the trunk (*i.e.*, erector spinae and abdominal) muscles may have some therapeutic value in the short- and long-term protection of the back from becoming painful. Fourth, studies support the assertion that, other things being equal, the physically fit individual is less likely to have a backache problem than one who is not.

The next question is, How do we achieve fitness and exercise goals? The answer is, In whatever way we can. Our "philosophy" is to recommend whatever we think the patient is most likely to do that is likely to be helpful and harmless. Certainly if the patient has a hobby or a sport that is likely to be helpful, that is encouraged and carefully monitored. The basic options that we recommend follow:

A *swimming program* is probably the best exercise because it is aerobically excellent, gravitational forces are minimized, and it strengthens the trunk and abdominal muscles as well as those of the upper and lower limbs. The patient is encouraged to work up to a 30–45-minute period of relaxed swimming three to four times per week. Any stroke or combination of strokes may be used. Sometimes the breast stroke or the crawl aggravates back pain. If this occurs, avoid those stroke(s) and use the back stroke.

Stationary or *mobile bicycling* in a slightly flexed sitting position is also good exercise. Here, too, the recommendation is for 30–45-minute periods three to four times per week. This is good for aerobics, the lower limbs, and the stabilizers.

Walking is another exercise option. This should also be done for 30–45 minutes three to four times per week. Walking is good because it is readily available, patients will do it, and it is social. This exercise is aerobically beneficial and is good for strengthening lower limbs and trunk musculature.

In order to specifically address the trunk muscles in a patient who does not or cannot do some combination of walking, biking, or swimming, we have also employed isometric abdominals, sit-ups (Fig. 6-41), and the erector spinae strengthening exercises depicted in Figure 6-40.

There is evidence of the role of the truncal musculature in protecting and assisting the spine in heavy lifting.[71,90,99,132] We hypothesize that that same role is crucial in protecting and improving the diseased symptomatic spine. As can be seen in Figure 6-38, the exercise also has the effect of placing the spine in a less lordotic position. Although there is controversy about the lordotic position as a cause of back pain and disc degeneration, it is true that the flat position is generally more comfortable and is associated with about one-half as much intradiscal pressure as is found in the lordotic position.[227]

McKenzie Program

This popular form of exercise therapy for the painful back has numerous advocates and opponents. Thus, there is considerable controversy. We will not review the controversy, but the program and rationale are briefly described. Each patient is put through a series of lumbar spine movements. These include flexion, extension, lateral bending, and axial rotation. The movements that are associated with diminution of the patient's most peripheral symptoms are identified and converted into an exercise program specifically designed for that particular patient. The patient, in terms of the *theory* of the program, "centralizes" and eventually eliminates the pain through this individualized exercise program.

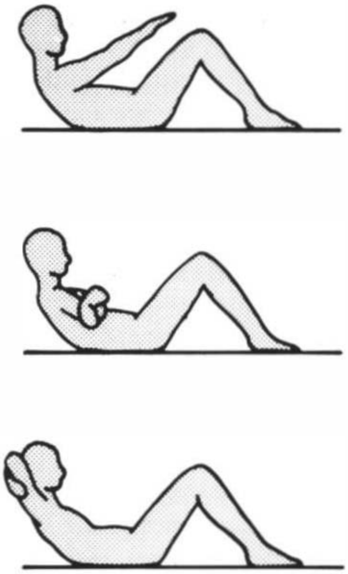

FIGURE 6-41 The exercises shown here have been suggested as a methodical of strengthening the abdominal muscles. Note that the feet are always kept flat on the floor. The three exercises increase in difficulty from top to bottom. Greater abdominal muscle forces are required in the bottom exercise as compared with the other two, because in this exercise the center of gravity of the upper body, due to the rearward position of the arms, is farthest away from the axis of motion. These exercises are done slowly and with the head raised initially, followed by a type of curling up of the upper trunk.

In a prospective randomized comparison trial, the McKenzie program was shown to be twice as effective as traction and back schools in alleviating back pain.[77] This is promising, and with further studies it may or may not prove to be as exciting as this trial investigation suggests.

Physical Therapy

This section includes an evaluation of massage and the various physical modalities of heat application.

Several rationales are used as the basis of these therapeutic modalities. Massage and heat application are soothing and relaxing, which makes the patient feel better and may have value in breaking up the cycle of muscle spasm, pain, and muscle spasm by alleviating the muscle spasm. Based on the "gate" control theory of pain (see p. 381), it may also be suggested that sensory input from the heat and massage somehow plays a blocking or inhibitory role in the transmission of pain sensations. There is also the hypothesis that the heat is transmitted to the deep tissues and provides a curative function. Finally, physical therapy may be used for whatever placebo effect it can deliver. Investigators[238] have suggested that endorphin secretion within the central nervous system may explain some of the benefit of massage. There are, however, no controlled studies to support the use of massage as effective therapy for low back pain.

Treatment

There is considerable work that has been done on the technology of massage and the various instrumentation that is employed to deliver heat. Such a discussion is not included here. The patients are generally given a few minutes of heat treatment and/or massage one to five times per week for 1 or more weeks.

Results

The results of these treatments do not differ from those of the various other forms of nonsurgical therapy.

Complications

These are limited to the risks of skin burns from heat therapy of too high intensity and duration and any social complications that might result from massage.

Comments

Massage and heat application constitute a low-risk form of therapy that satisfies the need on the part of physician and patient to treat and be treated. This is a fairly expensive emotional exercise that is worth avoiding if the patient can understand and tolerate a program of rest, isometric abdominal exercise, and education.

Patient Education and Group Therapy Programs

Based on the hypothesis that in most instances the natural history of the disease is one of a satisfactory recovery for the patient, a treatment of "wait and see" is justified. Moreover, the results of virtually all forms of conservative therapy are the same. However, there is an implied or expressed demand by the patient that something be done. There are a number of practical, reliable "tips" that the knowledgeable patient can take advantage of to reduce the pain and improve the quality of life while getting well. Finally, there is the positive reinforcement, understanding, and sympathy that one can receive from a group of fellow sufferers. Put all of these facts together, and they constitute the rationale for a spine pain school. These have been in operation in Sweden for many years now.[232] Some similar ideas have been instituted in the U.S., with relatively more emphasis on the group dynamics. These have been used for patients in whom surgery is not thought to be of value.[224]

Treatment

The program in Sweden is called the Low Back School. The goals are: to create self-confidence so that the patient may most effectively adjust to and manage the back condition; to avoid excess or potentially harmful treatment; and to decrease expenses. Numerous modifications have developed around the world from this concept.

The program consists of four 1-hour sessions that are essentially teaching demonstrations. The patients are also given an exercise program as shown in Figure 6-41.[C] The course material is outlined in the accompanying display.

The program in the U.S. involved teaching and encouragement of exercises, activities, and weight loss when necessary, with social facilitation to achieve these goals. There are a number of useful booklets and "handouts" that are available to give to patients. Listed on page 461 are some "tips" that, based on experience and current information, in our judgment are worthy of consideration.

Nachemson's advice to patients returning to

LOW BACK SCHOOL FOR PATIENTS (COURSE OUTLINE)

I Anatomy and function of spine back pain
  Cause, incidence, treatment effects
II Biomechanics of spine
  Effects of various activities on intradiscal pressure
  Importance of decreasing loads on back
III Ergonomics and practical application
  Individual advice about working, resting, and other activities;
  Teaching isometric abdominal and back exercises
IV Repetition, synopsis and test
  Instilling self-confidence; encouraging sports and other activities

work, based on disc pressure measurements,[234] is as indicated below.

We consider these guidelines to be sound advice. Some, but not all, studies have demonstrated the usefulness of this advice.[22,53,321] There is an important theoretic biomechanical consideration to present here. This has to do with the fact that disc pressure measurements were done on normal discs. We know from the work of Merriam[220] that while discographically normal discs behave predictably with regard to pressure measurements with different activities, degenerated discs behave differently and rather unpredictably with various postural changes.

This does not invalidate the guidelines offered, but it does indicate some potential limitations as they apply to the patient with degenerated discs.

Results

The Low Back School program has proved, through a well-designed and controlled investigation, to be superior to a placebo and moderately better than physical therapy. The patients treated in Low Back School reported less frequent absence from work than those treated with physical therapy. The physical therapy program, which consists mainly of manual therapy, was also found to be more effective than a placebo.[23]

A recent prospective study of the short- and long-term follow-up of the effectiveness of a back school showed it to be effective in both the short and long term.[223]

Complications

None were reported.

Comments

Because of the large number of patients that are afflicted with spine pain, the continuously increasing medical costs, the psychologic overlay, and other considerations, this seems to be an approach worthy of consideration.

ADVICE TO THE RETURNING WORKER

- Do not lift heavy objects
- Be as close as possible to object being lifted
- Avoid bending
- Avoid axial torsion
- Change positions frequently
- Avoid sitting in a low chair
- Use chairs with armrests and lumbar support

Axial Traction

Continuous and intermittent traction has been used for treatment of spine pain. There are a number of hypothesized mechanisms through which axial (y-axis) traction is thought to offer some therapeutic benefit. Most of these mechanisms are listed below. It is probably true that none of these theories have been proved or have substantial evidence to support them.

Treatment

Some of the various techniques and schedules for the application of cervical and lumbar traction are discussed, with information about their effectiveness.

There is considerable disagreement in the literature about the technique, degree, and duration of traction that should be applied to the spinal column. The duration of traction recommended varies from 4 minutes to 1 hour. In the cervical spine, the range of suggested forces is from 25 to 300 lb. Straight, axial traction is applied with the patient either supine or sitting. Varying degrees (0–30°) of neck flexion have also been employed.[58] In the lumbar spine the recommended forces range from 40 to 730 lb![81]

A traction weight of 30 lb for 7 seconds produces posterior separation of the cervical vertebrae. This appears to be the least weight and duration that effectively separate the vertebrae. Greater time causes additional discomfort without any signifi-

THEORETICAL AND OBSERVED CHANGES WITH AXIAL SPINE TRACTION

Diminution of disc protrusion[270]
Reduction of cervical disc space pressure[298]
Enlargement of intervertebral foramen
Opening up of the intervertebral disc space
Separation of intervertebral joints
Stretching a tight or painful capsule
Release of entrapped synovial membrane
Freeing of adherent nerve roots
Production of central vacuum to reduce herniated disc
Production of posterior longitudinal ligament tension to reduce herniated disc
Relaxation of muscle spasm

cant increment of mechanical change. An increase in force to 50 lb increases the separation between vertebrae.[60] The greater the angle of neck flexion, the greater is the posterior elongation, and therefore the greater is the opening of the intervertebral foramen.[59] The greatest separation of the cervical vertebrae occurs at a flexion angle of 24°. The amount of separation achieved with this degree of flexion is essentially as good with only 30 lb of traction as the separation obtained with 50 lb of traction without any flexion. The mechanical effects of traction are short-lived. It has been shown that with more than adequate cervical traction techniques (25 minutes, 30 lb) there is no significant residual intervertebral foramen separation 20 minutes after completion of traction.[60]

Recent work by Zibergold and Piper[376] demonstrates a statistically significant superiority of cervical traction as compared with no traction for cervical spine disorders. Intermittent traction (25 lb, 25° flexion, 10 seconds on/10 seconds off) was used and was found to be superior to static traction.

Studies of the effects of traction on normal cervical spines showed that the axial stiffness of the spine *in vivo* was such that separation of the vertebrae was possible with an axial load of one-third of body weight. The range of separation observed was 1–2 mm.[287] This much displacement could separate the joint space, open the neural foramen, and conceivably result in several other theoretical and observed changes listed on page 432.

In vivo studies of 80 discs in 48 normal subjects showed a 50% drop in intradiscal pressure with 10 kg of traction. There was an even greater reduction in pressure in degenerated discs. This important *in vivo* study also showed a 40% reduction in cervical disc pressure going from a sitting to a supine position.[298]

The problem of body–bed frictional resistance in the dissipation of traction forces is well elucidated in the lumbar spine work of Judovich.[167] His work is based on the study of one cadaver and three normal subjects. By measurements before and after cutting through the cadaver at the L3–L4 interspace, he was able to establish that the frictional resistance of the lower half of the body was about 26% of body weight. Therefore, in order to apply traction to the lumbar spine, one must first overcome the *frictional* resistance of the lower body segment. Any traction force above this is then applied to the spine for whatever therapeutic benefit it can deliver. This problem of body friction may be overcome by adding the extra 26% body weight or by a split bed–mattress technique, with wheels on the lower segment to reduce friction, or by vertical application of the traction. With vertical traction, the lower body segment would, of course, add its own weight in traction force rather than dissipate the traction force through frictional resistance. Lehmann and Bruner described a hydraulically powered device that delivered sufficient traction to the lumbar spine to achieve separation of vertebrae along the y-axis. The force required (about 300 lb) was associated with "uncomfortable stretch." Moreover, the mechanical effects were found to last only a short time after treatment.[199]

Lawson and Godfrey studied spinal traction and concluded that as much as 100 lb of cervical traction or 150 lb of lumbar traction resulted in no significant y-axis separation of the vertebrae. There was, according to these investigations, a slight temporary increase in height, presumably due to a loss of cervical lordosis.[196]

However, other investigators showed that with the use of a split-traction table and the hips flexed 70° with the lower legs parallel to the floor, significant separation of the posterior elements of the lumbar vertebrae could be achieved with 50 lb. A weight of 100 lb significantly increased the vertebral separation more than that achieved by the 50-lb weight. The greatest separation occurred at the L4–L5 and L5–S1 interspaces.

Larsson and colleagues,[194] in a multicenter controlled study, demonstrated the superiority of autotraction over the use of a corset. The difference was noted at 1 week post-treatment but not at 3 months. The technique is one in which the patient, through a mechanism of ropes and pulleys, is able to exert axial traction in the spine and pelvis by pulling with the arms.

Results

In a study of 212 patients with a variety of symptoms thought to be related to the cervical spine, it was found that patients with the symptoms most likely related to nerve root irritation benefited most from cervical traction.[338] In this group, 68% of the patients were improved. The treatment consisted of heat and massage plus 3–13 kg of axial traction in slight flexion (sometimes intermittent motorized traction) 3 times per week for 4 weeks. Because the "nerve root" symptoms were relieved better than were the other symptoms, it was thought that the study supported the hypothesis that the intervertebral foramina are enlarged by traction. It is extremely difficult to evaluate this series. We note that the cure rate is in the familiar range of 60–70%, which makes it difficult to distinguish from placebo administration, the natural history of the disease, and most other forms of therapy.

Christie reported on the preliminary results of a study that compared the effects of traction with an oral placebo medication in acute and chronic low back pain with and without sciatica. There was no meaningful difference in the results of the two treatments.[56] Hood and Chrisman reported that 52.5% showed good and excellent results with intermittent pelvic traction for lumbar disc disease.[160]

Complications

There have been complications associated with the use of axial traction. Eie and Kristiansen reported that about 33% of a group of patients they were treating for lumbar spine pain had distinct aggravation of symptoms.[90] The major risk is neurologic damage secondary to overloading the spine beyond its tolerance limits.[115]

Comments

It does not appear that axial traction is a superior modality for the treatment of spine pain. There are no convincing data to support a contention that the benefits are related to anything other than the improvement expected in time with the natural course of the disease or many other treatments.

Axial traction has been shown, at least temporarily, to separate either cervical or lumbar vertebrae. It can also reduce lumbar spine intradiscal pressure. There is no evidence that these changes are therapeutic or that this technique of treating spine pain with or without nerve root irritation is superior to any other. We note that it has certain associated risks. However, in patients who are not responding to other nonsurgical treatments, axial traction may be given a trial. The justification for this suggestion is based on the fact that axial traction is capable of at least temporarily altering the mechanics of the spine, with the Type IV or V disc pathology discussed on pages 393 and 394. Should the pain happen to be emanating from a structure that can be altered by manipulation, there may be some benefit. As a therapeutic trial, we recommend two or three treatments per week for 2 to 3 weeks using the appropriate technical factors necessary to separate the vertebrae in the painful region of the spine.

Some investigators have expressed the opinion that traction should not be used if the patient is diagnosed as having a fully sequestered disc with sciatica.[52] However, the report of Hood and Chrisman showed that some patients with true sciatica improved with axial traction.[160] They have suggested that the position of neural encroachment upon the axilla of the nerve root may be a factor. If the impingement is in the axilla of the root, the patient tends to list to the side of the sciatica.[107] Such a patient may become worse with axial traction. The patient who leans away from the painful leg may show encroachment laterally and may achieve a beneficial result with axial traction (Fig. 6-42).

Spinal Manipulation

As part of the U.S. Senate Report on the Fiscal Year 1974, Appropriation for the National Institute of Neurological Diseases and Stroke (NINDS) of the National Institute of Health, there was a specification that " . . . this would be an opportune time for an 'independent, unbiased' study of the fundamentals of the chiropractic profession. Such studies should be high among the priorities of the NINDS . . . " In February 1975, NINDS sponsored a workshop on "The Research Status of Spinal Manipulative Therapy."[130]

The following salient points are taken from the editor's summary of the workshop. There was no quantitative or qualitative reproducible description of "subluxation" either mechanically or anatomically. The concept of chiropractic subluxation remains a hypothesis yet to be evaluated experimentally. We believe that this has been one of the most frustrating aspects of certain views of the pathology that is purported to be altered with spinal manipulative therapy. When one is correcting a "subluxation" that cannot be perceived by independent scientific observers, it is difficult to convince those observers that the treatment is effective.

In regard to advancement in scientific knowledge, we are not aware of any significant investigations that address the scientific validity of the theory. There are several cogent clinical papers to be presented as an update on some comparisons of spinal manipulation with other therapeutic modalities. However, there are two key publications that have appeared in the literature since 1975 that are

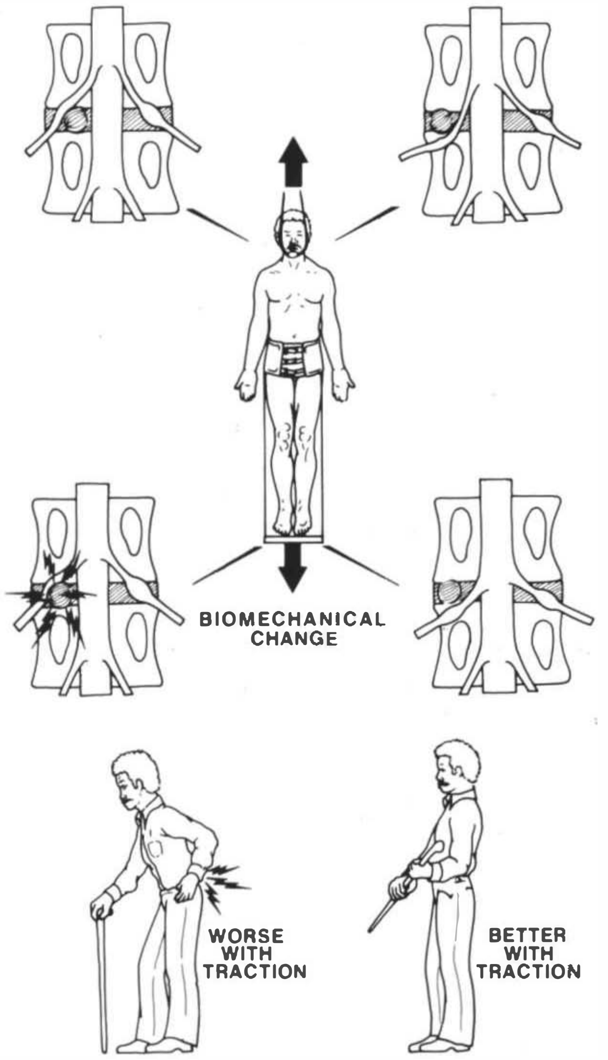

FIGURE 6-42 This is a hypothetical explanation of why some patients may respond well to traction, while others may respond with more pain. When the disc is in the axilla of the nerve root, axial traction may irritate the problem. A correlation of this with Figure 6-27 shows that the patient who lists to the side of the sciatica is likely to be made worse by traction. Conversely, one who lists to the side away from the sciatica is likely to be made better by traction. A biomechanical explanation for this is offered. The traction produces bending, which converts the lumbar spine from its physiologic lordotic posture to a more straight or flexed position. When this occurs, there is a relative displacement of the neural elements in the caudal direction. This results in either impingement by the herniated disc (*left*) or release of disc impingement (right).

presented here briefly as an update to the NINDS report.

One study is essentially a resume and a compilation of data on the status of manual medicine in the United States and Europe in 1982.[84] The work points out that there are about 19,000 osteopathic physicians and 23,000 chiropractors currently practicing this type of therapy in the U.S. Chiropractors alone collect $1.3 billion annually from the 130 million patients receiving their services. There are 15 osteopathic and 16 chiropractic schools, and together they graduate 3,000 students each year. The chiropractic schools are entered after 2 years of college. The chiropractic schools, some of which must be attended for 5 years in order to graduate, are including basic science pathology and clinical medicine courses similar to those taught in medical schools. A chiropractor works approximately 43–49 hours per week and earns $65,400 per year. Manual therapy is alive, well, and developing.

The second document ("Position Paper on Chiropractic") is a milestone contribution to the knowledge and understanding of those concerned about societal and community health care issues as they relate to chiropractic.[239] This document, produced by the National Council Against Health Fraud, Inc., provides an updated review and an objective analysis of manipulative therapy, various schools of thought among chiropractors, and, most important, some national well-based recommendations to help solve problems related to patients, preventative medicine, and societal improvements in regard to these issues. Specific recommendations for are made for consumers (patients); insurance carriers and third-party payers; legislators; basic scientists; academicians and educators; attorneys and law enforcement agencies; medical doctors, dentists, and other scientific health care providers; and reformist chiropractors.

Spinal manipulation is one of the most controversial but frequently used methods of treating spine pain. The controversy is charged with no small degree of emotion or bias. We claim no unique monopoly on objectivity. Nevertheless, the attempt here is to present a synopsis of the current status of spinal manipulative therapy as related to spine pain and to offer some comments on the subject. Because of its controversial nature, popularity among lay people, and obvious biomechanical relevance, it is discussed in some detail.

A large number of maneuvers may be employed in the manipulation of the spine.[210,248] However, regardless of what the external forces or manipulation may be, the movement of a vertebra is limited to the combinations possible with six degrees of freedom. Several of the best-known and most-used manipulations follow. The simplest one is the manual application of forces directly to the spinous processes and posterior elements of a given vertebra. These structures may be loaded and displaced along the +z-axis, the ±x-axis, the ±y-axis, and along various combinations of these (Fig. 6-43). In addition to this, through indirect means (manipulation of the head), the cervical spine may undergo various motions, including flexion/extension, lateral bending, axial rotation, and a variety of combinations. It does not matter whether the displacement results from the direct forces applied to the spinous processes (Fig. 6-44) or from forces applied by transmission through the ligaments and articulations of adjacent vertebrae (Fig. 6-45). The other variables available to the therapist are the rate of application of the forces and the magnitude of the forces. Of all the possible manipulations, those which appear to receive the most attention, at least in the medical profession, involve axial rotation.

What does manipulation do to the spine? From a study of the kinematics and physical properties of the spine, the basic constraints and patterns of motion of the spine are apparent. In Chapter 4, disruption of these patterns is shown to result in injuries when the normal tolerances of force or motion are surpassed. In order for spinal manipulation not to be harmful, it must not exceed certain tolerances, in either the normal or the diseased spine. In order for manipulations to be successful, they must somehow produce improvement using mechanical alteration, either directly or as a therapeutic stimulus to the diseased spine.

A more precise look at some structures that may be moved, stretched, stimulated, or relaxed by manipulation is necessary. Axial rotation is an effective means of applying tensile forces to the fibers of the annulus fibrosus of the disc. Various bending modalities are capable of applying tension to the annulus fibrosus and other ligamentous structures, as well as altering a bulge in the disc. The anterior and posterior longitudinal ligaments, both of which are innervated with sensory fibers, can be effectively moved by rotation about all three traditional planes of movement. This is also true of the yellow ligaments and the various interspinous and transverse ligaments. The importance of the fact that the inter-

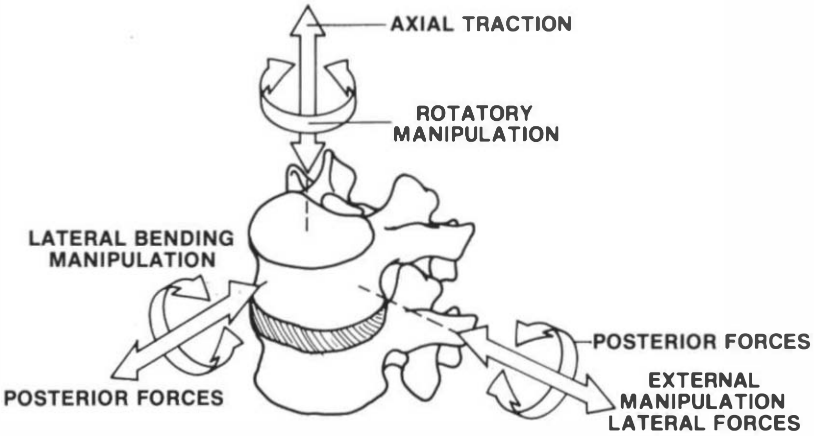

FIGURE 6-43 It is possible to achieve motion along all six degrees of freedom in the clinical situation through transmission of forces to the vertebra by indirect methods of manipulation. All of the various motions may be achieved clinically by applying the appropriate forces. The rate and magnitude of the different forces may be varied. It should be kept in mind that, no matter how complex or varied the external manipulation techniques, ultimately the vertebra is loaded and displaced according to some combination of these six degrees of freedom. The mechanism of their therapeutic benefit remains unknown. *(Rotatory manipulation and* posterior *forces are illustrated in Figures 6-45 and 6-46.)*

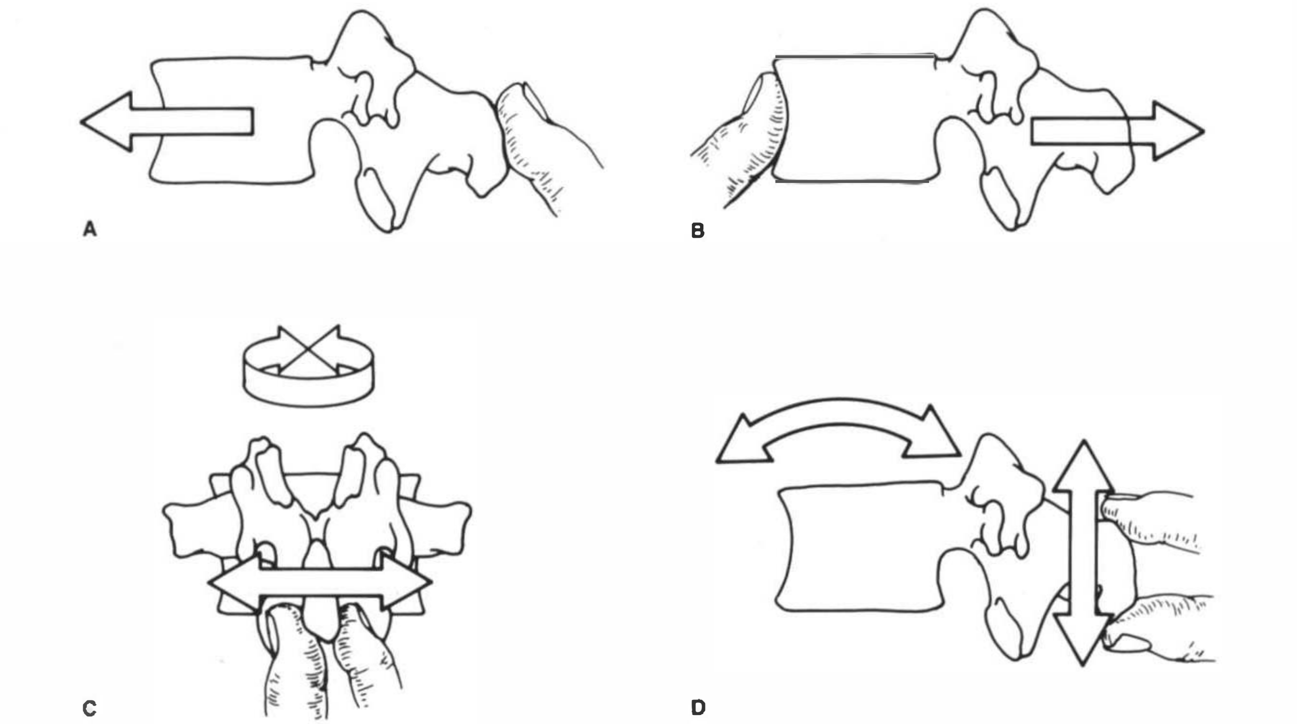

FIGURE 6-44 These are all of the possible direct forces (except for skin interposition) that may be applied for the purpose of manipulating the spine. The various motions and some of the major associated coupled motions are shown. In the clinical situation, the forces are not applied as shown here. However, it is possible to produce them with manipulation. **(A)** Direct +z-axis motion on the spinous process gives a forward translation (+z-axis). Since the spinous processes are just below the skin, this manipulation is possible in all regions of the spine. In the thoracic spine there may be x-axis rotation in addition to the z-axis translation. **(B)** Direct −z-axis motion can be achieved in the lower cervical spine by direct force application. When the neck is relaxed it is possible to palpate the anterior cervical spine just anteromedial and deep to the carotid sheath. **(C)** The spinous processes may be directly manipulated in all regions of the spine, resulting in ±y-axis rotation and, in the appropriate regions (cervical and upper thoracic), some amount of coupled ±z-axis rotation. **(D)** Here, the spinous processes are manipulated in the sagittal plane in the ±y-axis direction. This results in ±x-axis rotation. These are the major motions that are possible with direct manipulation. It has not yet been determined if these motions and forces are therapeutic.

vertebral articulations are true synovial joints has been emphasized. Axial rotation effects considerable movement and displacement of these joints in the cervical and lumbar spine and a fair amount of impingement and force application in the lumbar spine (see Figs. 1-19, 2-22). The impingement forces are taken by deformation of the cartilage, the facets, and also displacement between the vertebral bodies anteriorly. There is a possibility that a synovial fold or some intra-articular material may be altered. Certainly, anatomic studies have identified true menisci, fibrous invaginations, and fat-filled synovial reflections. However, these investigators considered the hypothesis that spinal manipulation exerted its therapeutic benefit through the repositioning of offending elements within the facet joints.[31] Another hypothesis is that changes in the mechanical status may alleviate or eliminate any associated synovial inflammation.

The theory that relief of nerve pressure may result from such limited possibilities of displacement does not fit well with present knowledge. There is no reason to assume that manipulation and displacement of an FSU through a normal range of motion can significantly move structures in or out of the intervertebral foramen.[8] This hypothesis has

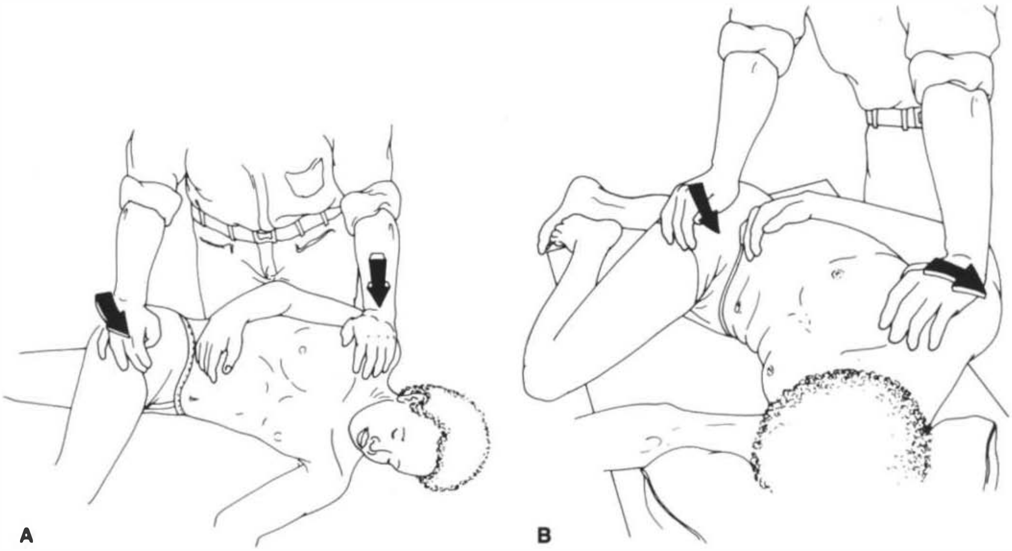

FIGURE 6-45 This particular manipulation designed for the thoracic and lumbar spine appears to be the one most frequently employed. **(A)** This shows the technique from the frontal view. The major thrust comes from the therapist's right hand, rotating the pelvis forward and indirectly applying an axial torque (about the −y-axis) to the thoracic and lumbar spine. The left hand is used to steady the thorax by taking up the reaction forces. **(B)** This shows the same manipulation from a different vantage point to emphasize the axial torque that is applied. Nothing is known about the magnitude of the torques exerted on the spine, the motion imparted, or the changes (therapeutic or otherwise) that may be associated. The axial manipulation of the cervical spine is done through the application of y-axis rotations of the head.

been studied in an experiment employing newborn cadaver material designed specifically to answer this question. The evidence from this study showed that this mechanism of nerve decompression is unlikely.[65] The work of Sunderland on adult anatomy tends to collaborate Crelin's findings in the newborn. Sunderland also warned against concentrating exclusively on the intervertebral foramen as the site of the pathologic mechanism.[323] The importance of the posterior vertebral joints and ligaments as potential sources of neurovascular bundle entrapment must also be considered.

A relatively more recent proposed mechanism suggests that the therapeutic benefit accrues from stimulation of certain large nerve fibers that then inhibit the transmission of nociceptive impulses through small-diameter nerve fibers. According to the theory, spinal manipulation achieves this by stretching the various ligamentous structures into a certain nonfailure range.[166] This highly speculative hypothesis is interesting.

Treatment

Treatments involve a number of manipulations, the details of which are available in the literature.[210,246] The process may be repeated one to five times per week. Sometimes manipulations are carried out under anesthesia. Probably the most frequently employed manipulation is the axial torque of the cervical or lumbar spine, shown for the lumbar spine in Figure 6-45. It is reasonable to assume that these manipulations can at least temporarily affect axial rotatory displacement of the cervical or thoracic spine. Direct manipulation is shown in Figure 6-46.

It has been pointed out in controlled experiments that sustained mechanical forces on axons or nerve trunks block rather than excite. This is the situation for transversely applied compressive forces as well as for longitudinally applied tensile forces. Transient, rapidly applied forces cause excitation.[261] If, indeed, the pathologic conditions exist at the nerve root level, then the relief of the compression or stretch by manipulation should not be expected to

FIGURE 6-46 This shows a frequently employed direct manipulation of the spine, which may be applied in either the thoracic or the lumbar region. As shown in Figure 6-43, it may result in ±z-axis translation and ±axis rotation.

relieve the pain. This information does not rule out the possibility that some other mechanism may be possible for the relief of back pain with spinal manipulation. It may be that there is associated inflammation, vascular engorgement, or chemical irritation of dural or ligamentous receptors that produces the pain. It is probably fair and accurate to indicate that at present there is no appealing substantial theory to explain the mechanism through which spinal manipulative therapy relieves pain, if in fact it does.

Results

Do manipulations help? As is the case with just about any form of treatment of spine pain, there is no well-documented clinical evidence that manipulation alters the natural course of the disease. Also similar to other therapeutic modalities, there is enthusiastic anecdotal evidence from patients and highly optimistic reports from therapists. There are *absolutely* no data that substantiate a direct relationship between manipulative therapy and clinical improvement of *visceral disease* through the improvement of segmental neural interactions. There is considerable controversy over the question of the efficacy of spinal manipulative therapy in the treatment of pain, particularly low back pain and neck pain. The following is a review of some of the major studies related to the question of whether or not spinal manipulations help patients with spine pain.

There are relatively fewer studies of cervical spine manipulation. However, the study by Sloop and associates[300] is of interest. By using diazepam to create amnesia in the manipulated group, it was possible to successfully execute a double-blind controlled study of patients undergoing cervical spine manipulation for chronic neck pain. The study showed a single manipulation of the cervical spine to be of no value.

Chrisman and colleagues investigated 39 patients with herniated intervertebral discs who had myelograms before and after rotatory (axial) manipulation of the lumbar spine under general anesthesia. The patients had been previously treated with conservative therapy. About half of the patients (51%) had good or excellent results after the manipulation. The myelograms, whether initially positive or negative, were unchanged after manipulation. Although some of the patients with positive myelograms were good or excellent 3 years after manipulation, in general, the patients without positive myelograms did better with manipulations than did those with positive studies.[55] These findings are the same as those of Mensor, who had a 51.2% success rate with rotatory manipulation under anesthesia.[219] Chrisman recommended a premanipulation myelogram to avoid the dangers of manipulating a patient with a large herniated disc.

There are other studies in the literature that report that manipulative therapy is used for spine pain. One of these shows statistically significant evidence that this treatment is better than other nonoperative therapy. Coyer and Curwen compared 152 patients with acute low back pain randomly selected for treatment with either manipulation or bed rest and analgesics. In the group that was manipulated, 50% were symptom-free in 1 week and 87% in 3 weeks. In the group that received analgesics, respective figures were 27% and 60%. However, the investigators stated that their figures were inadequate for statistical analysis.[62] In a randomized study, rotational manipulation was compared with de-tuned (simulated) short-wave diathermy as a placebo. The patients were evaluated 15 minutes, 3 days, and 7 days after treatment. Both groups of patients were markedly improved. Except for one factor, there was no demonstrable difference between the two groups;

the manipulated patients were better 15 minutes after treatment. It is also of interest that by subjective self-assessment, the patients in both the *placebo* and manipulated groups rated themselves as 70–93% improved.[129]

Doran and Newell carried out a prospective "blind" study to compare the efficacy of four different types of treatment. In this study, 456 patients were randomly subdivided into four groups according to the following treatments: manipulation, physiotherapy, a corset, or analgesic tablets. The patients were evaluated after 3 weeks, 3 months, and 1 year. There were never any important differences among the four treatment groups. However, as noted in other studies, some of the patients responded well and quickly to spinal manipulation.[80]

These more recent studies showed some immediate short-term benefits from manipulation but no long-term improvements.[101, 154] Farrell and Twomey[101] compared the response of low back pain patients to manipulation and to patient education. Flexion exercise and diathermy showed that those having manipulation returned to a symptom-free status sooner. Hoehler and colleagues[154] compared and studied low back pain patients and found that those having spinal manipulation experienced greater improvement than those receiving massage. The third investigation was by Jayson and associates.[166] They compared mobilization and manipulation with placebo and demonstrated a short-term benefit.

The work of Kane and colleagues is focal to the question, "Does 'manipulation' help?" This study reviews 122 patients treated by chiropractors for back or spine problems and 110 patients treated by physicians for the same type of disease.[170] With regard to patient perception of improvement and patient satisfaction, the chiropractors were at least as effective as the physicians. The two groups of patients were not significantly different with respect to demographic and socioeconomic factors, hypochondriasis, or their attitude toward the medical profession. It was found that the patients treated by chiropractors responded more favorably to the personality of their practitioners. This was thought to be due to the fact that chiropractors seemed *unhurried, sympathetic, communicated well* with the patients, and *treated them in an egalitarian manner*. These humanitarian considerations are important in all aspects of patient care, but they are especially crucial in the area of spine pain because the objective information is so meager and the subjective factors are so important.

The last three sentences in the preceding paragraph may be the most important ones in this section, chapter, or book. We attempt to share the Kane study every time an Introduction to Clinical Medicine lecture is presented to our medical students. We view it as crucial to *their* education and as an important reminder to ourselves. It is of interest that 14 years after the Kane study, Deyo and Diehl[75] have come forward with what we think of as a preeminently important sequel. This study also happens to be related to the study of low back pain, measuring the extent of *dissatisfaction* patients felt with the medical care received. The dissatisfied patients, unlike the satisfied patients, were characterized as wanting more diagnostic tests but not wanting to see the same doctor again. Why? The study gives the answer—because of the *failure to receive an adequate explanation of the problem*. The study also showed that those doctors who had satisfied patients and gave an adequate explanation did not order more tests and did not spend more time with the patients. One final point in this digression. In 1983, nine million patients were reported to have made 135 million chiropractic office visits.[239] The patients are in pain and are often anxious and/or depressed. We have some sense of the value and limitations of spinal manipulation. These patients, in order to make 135 million office visits, must be getting something more than just spinal manipulation. We believe that a significant part of what they are getting can be found in the studies by Kane and associates[170] and Deyo and Diehl.[75]

Nachemson, pursuant to a thorough review of the literature, indicated that there is no clinically significant proof that manipulation for acute or subacute low back pain is superior to bed rest and salicylates.[228]

There is an important study by Edwards[89] in which spinal manipulative therapy involved 184 patients who were divided into four groups (Table 6-2). The 46 patients in each of the groups were further divided into those to be treated with heat massage and exercise and those to be treated with spinal manipulation. The salient results are reproduced in Table 6-3.

In Group I, the results were acceptable in 82.5% for both treatments. However, they were achieved

TABLE 6–2 Subject Groupings of Patients in Edwards' Study of Spinal Manipulative Therapy

Group	Type of Pain	Treatment
I (46 pts.)	Central low back pain only	23 HME* 23 Manipulation
II (46 pts.)	Radiation to one buttock	23 HME 23 Manipulation
III (46 pts.)	Radiation down back of thigh to knee	23 HME 23 Manipulation
IV (46 pts.)	Radiation down posterior leg to foot	23 HME 23 Manipulation

* *HME* = Heat, massage, and exercise.

with spinal manipulation using about one-half the number of treatments that were needed for heat, massage, and exercise. In Group II, the results were slightly better with manipulation, and again they were achieved with about half as many treatments. In Groups III and IV, the manipulation therapy was statistically significantly better, and in Group IV, the results with manipulation were achieved with half as many treatments. If all groups are combined, the manipulative therapy is significantly better. This study certainly supports the efficacy of spinal manipulative therapy in comparison with heat, massage, and exercise. The results (80–95% satisfactory) are impressive in comparison with any form of therapy.

We suggest that the question of the effectiveness of spinal manipulation is still unanswered. There is evidence that the use of spinal manipulation is beneficial at least in the short term for treating spine pain with or without sciatica. Its use in the treatment of other diseases, in our opinion, is *totally* fallacious.

Complications

Do manipulations harm the patient? The answer is yes, sometimes. There are some reported cases in the literature in which neurologic damage was associated with spinal manipulation.[265, 332] Poppen described four cases in which patients experienced paraplegia or cauda equina syndrome following manipulation. Two of the patients were manipulated by osteopaths and the other two by orthopedic surgeons.[265]

Fisher reported a case of the precipitation of a large, midline L5 disc herniation that occurred following manipulation of the low back by a chiropractor. The patient developed the clinical neurologic signs and symptoms of cauda equina tumor. The disc herniation was noted at the time of surgery and removed. The neurologic recovery was complete.[109]

Pratt-Thomas and Berger reported two fatal cases of cerebellar hemorrhage and one fatal case of spinal cord injury following chiropractic manipulation.[269] Two additional patients suffered nonfatal vascular accidents following chiropractic manipulation of the cervical spine. One possible mechanism of vascular damage associated with axial torsion of the

TABLE 6–3 Results of Edwards' Study of Spinal Manipulative Therapy*

Group	Treatment	Total No. of Patients	Average No. of Treatments	Acceptable Results	
				No. of Patients	Percentage
I	HME†	23	9.7	19	82.5
	MM‡	23	4.8	19	82.5
II	HME	23	10.2	16	69.5
	MM	23	4.3	18	78.1
III	HME	23	8.5	15	65.2
	MM	23	6.2	22	95.7
IV	HME	23	13.3	12	51.7
	MM	23	6.4	18	78.5

* Edwards, B. C.: Low back pain and pain resulting from lumbar spine conditions: a comparison of treatment results. Aust. J. Physiother., 15:104, 1969.

† *HME* = Heat, massage, and exercise.

‡ *MM* = Mobilization and manipulation.

cervical spine is shown on page 93 (Fig. 2-8). It appears that these manipulations involve the application of high magnitudes of torsional loads to the cervical spine.[134]

More recent examples of complications of cervical spine manipulation are as follows. Myelopathy, stroke, acute brain stem stroke, vertebral artery occlusion, and brain stem lesion with coma are all serious complications that have been reported.[239] This listing is by no means complete. In the way of balance, there is the report of closed manipulation of the cervical spine *dislocation* causing immediate recovery from paraplegia.[81] There have also been ample complications associated with manipulation in the lumbar spine.[239]

Comments

As with other treatment modalities, there is the basic problem of demonstrating the effectiveness of spinal manipulative therapy over the improvement with a placebo that occurs in 33% of patients and the improvement that occurs in the natural course of disease. We suggest that additional pathoanatomic and biomechanical studies of the spine, including mathematical modeling, may provide a sound hypothetical basis for the mechanism of some types of spine pain. In addition, the development of theories explaining the therapeutic effects of manipulation is needed. The value of the therapy will have to be demonstrated by well-designed prospective clinical studies. Spinal manipulation seems to be about as effective as most other treatments. There is a recurrent observation that on a short-term basis, a number of patients, usually by their own subjective reporting and sometimes after a physician's evaluation, do better with spinal manipulative therapy than with other forms of nonsurgical treatment. We must, however, keep in mind the risks of complications. Patients with signs and symptoms of acute cervical thoracic or lumbar disc herniations should not be manipulated in our opinion. Likewise, those with possible cervical vascular disease, or those with spinal tumor or severe vertebral osteomyelitis, should not undergo manipulations.

There remains the practical question of whether or not spinal manipulative therapy is to be recommended as a nonsurgical treatment for low back pain and sciatica. Present knowledge indicates that the risk/benefit factors are such that spinal manipulation that is not too vigorous is a justified, alternative, nonsurgical treatment. Prolonged, expensive, repeated manipulations that offer only brief, transient improvement do not make the best use of the patient's resources.

Orthotic Devices

The practical expectations and biomechanical aspects of orthotic devices are presented in Chapter 7. The rationale for their use in the treatment of spine pain is based on immobilization, abdominal support, and the maintenance of a straight or slightly flexed lumbar spine. Immobilization is for purposes of splinting, supporting, and resting the spine to reduce mechanical irritation and muscle spasm. The abdominal support assists in the development of adequate intratruncal pressures to support the spine. This may be done with certain braces, a corset, or a plaster cast. The mechanism of this support is described on page 482.

Although there is not yet an ideal study to document the superiority of a lumbar orthosis over a placebo, there are several cogent studies that address and support a rationale for their use. Million and associates[221] conducted a well-designed study comparing lumbar corsets with and without an inserted molded support. The investigators reported significant subjective improvement of symptoms in the patient group that had the molded supports inserted in their corsets. Presumably, the added immobilization caused the improvement. There is evidence that a rigid brace can limit lumbar mobility.[365] Nachemson[236,237] reported a decrease in disc pressure associated with the slight flexion that is achieved with the use of a lumbar orthosis. It has also been recognized that the slight flexion provided by a rigid lumbar orthosis may enlarge the canal and improve a patient with spinal stenosis. Some of the common lumbar orthoses are shown in Figure 7-23.

Complications of lumbar orthoses are rare. There are two issues to discuss. It has been suggested that the use of a lumbar spinal orthosis may cause muscle atrophy. This has not been documented, and on the contrary, Waters and Morris have shown with an EMG study that abdominal muscle activity actually increases.[344] Thus, in our clinical practice, when a corset makes a patient "feel better" we simply use it. A potentially serious complication that may occur with a plaster jacket on a rigid lumbar lordosis is the superior mesenteric artery or "cast" syndrome.

When there is severe abdominal pain or vomiting following the application of a body cast or a body jacket or corset, the apparatus must be removed.

Local and Regional Use of Epidural Steroids

The epidural injection of locally acting corticosteroids and local anesthesia has been employed in the treatment of spine pain with or without associated radiculopathy. Novocain is administered to control the immediate pain due to the procedure and also to break up any cycle in which a regional irritant is causing extensive muscle spasm.

There have been studies in which the use of epidural steroids has been considered effective. Berman and associates reported a 70% success rate in patients with lumbosacral pain with negative EMG and myelograms.[24] In contrast, a prospective, randomized, double-blind study by Cuckler and associates involving 73 patients with radiographic evidence of a herniated disc or lumbar stenosis presents a rather different picture. Neither the patients with herniated discs nor those with spinal stenosis showed any statistically significant difference when compared with untreated controls. Moreover, the following complications were reported: tuberculous meningitis, adhesive arachnoiditis, aseptic meningitis, sclerosing spinal pachymeningitis, and hypercortisolism.[67] Transient hypercortisolism following epidural steroid injection was reported in a different study.[317]

The efficacy of this treatment method is at best equivocal. It appears that the likelihood of success is less when there is a distinct pathology such as disc herniation or spinal stenosis. The complications are not limited to "just a transient headache." A list of reported complications is provided. If this procedure is to be used, certainly adequate precautions should be taken.[317]

REPORTED COMPLICATIONS AFTER EPIDURAL ADMINISTRATION OF STEROIDS[317]

Headache
Dizziness
Transient hypotension
Increased back or limb pain
Elevation of cerebrospinal fluid protein levels
Septic or aseptic meningitis
Worsening of symptoms of multiple sclerosis
Sclerosing spinal pachymeningitis
Exacerbation of latent infection
Near-fatal septic meningitis (intrathecal injection)
Hypercorticism

Facet Joint Steroids

The combination of local anesthesia with cortisone injections of the posterior elements (synovial joints) has been thought to be successful in tendons and joint compartments in other areas of the body; therefore, this combination is also injected in the posterior intervertebral joints. These structures are well endowed with nerve supply and if inflamed may be expected to cause considerable pain.

Chemonucleolysis

This form of treatment, first proposed by Hirsch,[149] is based on the assumption that lumbar spine pain and sciatica are caused by abnormalities of the nucleus pulposus. Chymopapain is a proteolytic enzyme that effects a rapid hydrolysis of the protein–mucopolysaccharide ground substance of disc material, mainly the nucleus pulposus.[124] The disc space is commonly narrowed following injection of the material into the disc.[215, 369] The intradiscal injection is given because the dissolution of the nucleus pulposus results in the relief of pain and nerve root irritation. The actual mechanism of pain relief is unknown.

Treatment

Indications are the same as those for surgical laminectomy for excision of a herniated disc. There should be at least 6–8 weeks of conservative treatment, sciatica, positive tension sign(s), some evidence of radiculopathy, and imaging evidence of disc herniation. If there is a profound neurologic defect and/or imaging evidence that there is a herniated portion of disc material that is no longer in continuity with the central disc or nucleus pulposus, then chemonucleolysis should not be recommended. Also, the work of Mulawka[226a] suggests that if there is a defect that is greater than 50% of the anteroposterior diameter of the canal as shown on myelogram, this is a relative contraindication. This is based on the observation that under this condition a poor result is likely. Another study that compared chemonucleolysis with disc surgery noted that chemonucleolysis appeared preferable for patients with small disc herniations and those with medium-sized herniations, provided the preoperative clinical pattern indicated slight or moderate nerve compression. However, in larger-sized herniations, or the small and medium ones with clinical evidence of severe nerve root impingement, surgery is the treatment of choice.[287]

The technique has been well described by Brown and McCulloch.[40,215] The patient rests on the side with hips and knees flexed. A needle is introduced into the diseased disc. A discogram may be done to affirm the proper placement of the needle, and chymopapain is injected.

Results

This discussion is limited to the treatment of lumbar discs. Macnab and colleagues reported on the use of chymopapain in 100 patients. The best results were in patients with positive myelograms whose major symptoms were pure sciatica of short duration. The overall success rate was 67%.[206] A long-term follow-up series of 500 patients treated with chymopapain was reported by Wiltse and colleagues, and a similar study of 480 patients was reported by McCulloch. In both groups, 75% had effective relief of symptoms.[215,369] In the review article by Watts and associates, the range of satisfactory results using chymopapain in seven independent series was 49–75%.[346] Higher incidences of satisfactory results were reported by Brown (82.5%), Onofrio (83%), and Smith (90.7%).[40,250,301]

There have been studies in which chemonucleolysis was compared with surgical treatment of lumbar disc disease. In one study, 100 patients surgically managed prior to the availability of chymopapain were compared with an equal number of patients who had chemonucleolysis therapy. Chemonucleolysis resulted in a success rate of 74%, while surgical treatment yielded a success rate of only 48%.[245] Watts and colleagues compared the results in a group of 100 patients treated with chemonucleolysis with 174 patients who had undergone surgical disc removal. In the group of patients with a clinical picture compatible with a distinct posterolateral disc herniation, including myelographic confirmation, chemonucleolysis gave better results than surgery. Satisfactory results were achieved in 89% and 60%, respectively. In the patients who had previous surgery and back pain with or without nonradicular leg pain, satisfactory results were achieved in 55–60% with both types of treatment.[345]

Some recent clinically relevant research studies involving chemonucleolysis are reviewed. Several studies involving structural and biomechanical changes of the intervertebral discs in dogs showed similar changes. Initially following injection there was disc space narrowing accompanied by changes in the mechanical properties. The mechanical changes included increased stiffness in some studies and decreased stiffness in others,[33,34,168] accelerated creep rates, and an increase in torsional stiffness.[168] Over the ensuing 3 months postinjection, the disc height and biomechanical properties reached or approached their previous conditions. A possibly justifiable clinical extrapolation from the increased torsional stiffness is that patients should not return to work for 3 months following chemonucleolysis.[168] One study compared chemonucleolysis with discectomy. There were two differences at 5 months post-treatment: the surgical group was three times stiffer in medial and lateral bending than the chemonucleolysis group, and also there were more osteophytes.[188]

The utility of chemonucleolysis has been demonstrated in a number of studies.[37,217] Those by Fraser[111] and Javid[165] are two of the better designed and executed investigations. Surgery can be expected to provide a superior result,[63,92] although one report[350] did not support the generally assumed superiority of surgery over chemonucleolysis.

The reports on the efficacy of chemonucleolysis in the treatment of spine pain may be summarized by saying that the results are generally as good as or better than most of the other nonsurgical treatments of low back pain and sciatica. The patient who is most likely to show a good result with chemonucleolysis will probably also show a good result if treated surgically.

Complications

The major complications associated with chemonucleolysis are sensitivity reactions (some of which cause death), discitis, and arachnoiditis.[215,346] Experimental studies have shown that nerves exposed to chymopapain in clinically recommended concentrations for 2 hours may develop intraneural edema as an immediate result. With regard to long-term effects, investigators reported degeneration of nerve fibers, intraneural fibrosis, and impaired impulse transmission. Rydevik and associates and

Shealy also reported on the neurotoxic characteristics of the drug.[285,295] Although there is other evidence attesting to the safety of the enzyme,[110,123,369] it is important to be aware of these observations.

More recent studies show a low incidence of severe, very serious complications and a fairly high incidence of not serious but annoying complications of back spasms, stiffness, and soreness.[217] In 29,075 patients there were 11 deaths, 3 of which were related to the drug and 1 that was attributed to the procedure. Two were due to anaphylactic shock and one to a bacterial discitis that developed into a fatal meningitis. There was one paraplegia due to needle trauma or injection into the subarachnoid space.[6] There are two additional reports of paraplegia following the procedure.[87,351]

Comments

Chemonucleolysis in purposely selected patients and with appropriate precautions to minimize the risks of anaphylaxis and neurotoxicity can be a useful treatment modality for patients with herniated discs. Because of the neurologic problems (probably due to subarachnoid injection of the enzyme), both physician and patient interest in using the technique has diminished considerably in the U.S. In some other countries, such as the Netherlands, it remains popular. Early results with percutaneous discectomy suggests that this methodology may offer virtually all the benefits of chemonucleolysis without the attendant complications.

Percutaneous Discectomy

Percutaneous discectomy was first performed in 1975 by Hijikata in Tokyo.[146] The technique involved removing the disc with long pituitary forceps through a cannula put into the central portion of the disc. Several other surgeons have used a variation of the methodology, including Kambin and Gellman in 1983[169] and Suezawa and Jacob in Switzerland in 1986.[322] The methodology was refined and automated in 1985 by Onik and associates.[248,249] The investigators invented a reciprocating suction cutter for dividing the disc material into small pieces and aspirating it from the disc space.

Treatment

The indications and the technique are virtually the same as for chemonucleolysis. This includes the presumed absence of a free fragment of disc based on appropriate imaging studies. A CT scan is taken at the level to be operated on to be certain that there are no endangered structures in the intended path of the needle and probe. Under sedation and local anesthesia using x-ray control, the aspirator, following a series of specific steps, is placed at the center of the disc. The machine and its tubing are appropriately attached, and disc material is aspirated for 20–40 minutes; 1–2 cc of disc material is removed. The technique is presented in detail in the article by Maroon and Onik.[212]

Results

We are not aware of any prospective controlled studies. Hijikata[146] reported 68% excellent results in 80 patients. Kambin[169] states that in the treatment of 50 patients there was an 85% success rate. Onik and associates[246,249] reported successful results in 31 of 36 patients (81%). In a multicenter study of more than 200 procedures, the success rate has been reported at the 74% rate.

Complications

In over 2,000 cases there have been only two complications, and they were both disc space infections.* There are certain risks, such as puncture of viscera or major vessels, or damage to neural elements using the lateral approach. Up to the present (1988), these complications have not been reported in this procedure with a posterolateral approach.

Comments

This procedure appears promising in appropriately selected patients. Because of its apparent low risks, it is reasonable to expect that in the absence of an extruded sequestrated fragment there may be enough decompression to change the mechanical environment enough to cure the patient. Usually the initial enthusiasm for a new treatment becomes adjusted somewhat downward as experience accrues. On the contrary, the opposite can occur with modification of technology, better patient selection, and reduction of complications. We believe that this technique deserves careful attention and consideration for patients with nonsequestrated extruded lumbar disc herniation before they undergo microdiscectomy or formal surgical laminectomy and discectomy. Prospective controlled studies are needed to adequately evaluate this treatment.

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* Personal communication, F. D. Wagner, August 1972.

There is a relevant biomechanical concept that is worthy of mention because it may provide a basis for the analysis and some of the therapeutic benefit of percutaneous discectomy. The bulk modulus is the slope of a curve that describes the relationship between volume and pressure of a system in which a gas or liquid is in a closed space. The volume is measured within the space, and the pressure is measured on the same surface of the enclosed space. If the curve defines a steep slope, then relatively small changes in volume can result in rather significant pressure changes.

If this concept is applied to the intervertebral disc and some assumptions are made, interesting speculation can follow. The hypothesis is discussed on page 400 and in the following section.

Laser Discectomy

The relief of sciatic pain with removal or vaporization of small volumes of disc material may be due to the presence of a high bulk modulus.[54] The bulk modulus describes the relationship between pressure and volume in a given space. A high bulk modulus indicates a steep curve in which relatively small changes in volume result in relatively large changes in pressure. The hypothesis is that removal of a small amount of disc material by the various percutaneous needle techniques (percutaneous nucleotomy, removal with small rongeurs, and laser ablation) results in a significant decrease in disc pressure and relief of symptoms. Corroborative clinical evidence includes: the increase in sciatica and/or back pain with injection of small amounts of fluid in the disc; the increase in symptoms upon rising from bed (due to increase of fluid volume in the disc during recumbency); and the increase in symptoms during stress when an increase in fluid uptake of the disc is thought to occur. The dramatic relief of cardiac tamponade associated with the removal of a small portion (1–5%) of the effusion may be thought of as due to the presence of a high bulk modulus in the pericardium. We assume that the disc behaves like a closed homogeneous fluid-filled container. Pressure within the herniated disc (pressure on the posterior annulus and/or posterior longitudinal ligament) may be the cause of back pain and/or sciatica. The removal of 1–2 cc of nucleus and annulus material from the central portion of the disc, provided there is a high bulk modulus, will result in a relatively large decrease in intradiscal pressure, a decrease in nociceptive input, and a decrease in the back and leg pain. There may also be some structural change in which some of the herniated disc moves back toward the interspace. Casual observation of pre- and postnuclectomy procedures has not confirmed this latter assumption, however.

Disc Excision

Excision of all or part of the intervertebral disc is based on traditional surgical rationale. It is a diseased organ that is causing pain and disability to the patient, and it can be removed.

Treatment

There is some controversy about whether the disc or only the displaced fragment should be removed. In addition, there are questions about whether the disc removal should be done from the front or the back of the spine. And finally, there is the issue of whether or not the spine should be fused after removal of the disc (see Chap. 8). There are arguments to support all of these options, and there is no definitive clinical or experimental evidence that supports one particular method.

The issue of how much of the disc to remove has received considerable attention. We recommend removing the herniated disc and a conservative margin beyond, but not the entire disc. The rationale is as follows. The results are not improved by attempts to remove all of the disc. Spengler reported a 2% recurrence with limited discectomy in 44 patients.[313] The literature suggests 5–6% overall. Studies show that even with rigorous attempts to remove the entire disc, only about 23% is excised.[48] Certainly, the risks of perforation and damage of bowel or major vascular structures are likely to be enhanced by attempts at full disc excision.[294] Finally, the normal biomechanics of the FSU remain closer to normal if less of the disc is removed. There we suggest removal of the free fragment, the herniated portion of the disc, and a small margin of safety, including fragments in the posterior portion of the disc. To further protect structures anterior to the disc, we avoid taking the pituitary rongeurs to a depth beyond the mechanical joint at the base of the mobile portion at the end of the instrument. The distance from the tip to the joint on the rongeurs in our operating room is about 1.3 cm, or roughly + in. This guideline should allow a satisfactory disc removal. Just after the tip of this useful but potentially dangerous instrument goes into the interspace, open it fully, advance it gently 1–1.5 cm, then close and withdraw. Following these guidelines on *each* pass should suffice. Some like to touch the end-plate above and below before closing the instrument. These guidelines are likely to help prevent the potentially fatal complications of this otherwise benign procedure.

The surgical approach is a matter of preference and practical considerations. In the cervical region, the anterior approach is readily accessible and convenient. The posterior approach in the lumbar spine is generally easier and is associated with fewer complications than is the anterior approach.

The thoracic disc herniation is best removed from an anterior approach,[7,104,253] because the results with posterior removal are poor (Fig. 6-47).[334] The reason for this is related to the space available in the thoracic canal and the relatively lower blood supply to the midportion (T4–T9) of the thoracic cord (see p. 413, Fig. 6-25). The laminectomy and retraction of the cord to remove a disc herniation could result in cord change. Thus, a lateral[104] or anterolateral[7,253] approach is safer.

Patterson and Arbit[260] have described a technique in which a transpedicular approach is used to remove the thoracic disc. This requires no retraction of the spinal cord and is a reasonable alternative. Figure 8-11 shows this approach for decompression in the presence of a burst fracture.

A review and update of relevant information on thoracic disc herniation is provided. The prevalence of the condition is put at 1 in 1 million,[104] although some investigations indicate that it is not quite that rare.[7] The herniation usually occurs below T6. A review of the literature shows that the herniation is central in 57% of patients and lateral in 34%. Trauma was thought to be a factor in 35% of patients, namely, paratroopers, weight lifters, and tumblers. There were the expected neurologic signs of spinal cord involvement, visceral symptoms, and usually a poorly defined neurologic level. Cystometrograms and EMG tests are helpful. Plain x-rays may show disc space narrowing, sclerotic changes in the disc, and calcification of its margin. One reporter indicated that the myelogram is the most important study; however, in the future we anticipate that MRI will emerge as the definitive test.

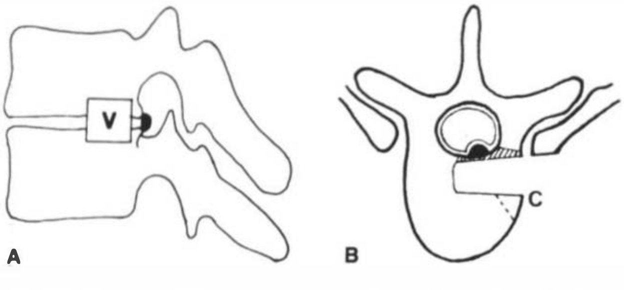

FIGURE 6-47 (A) Lateral view of two thoracic vertebrae showing starting area (V) for the tunnel of the vertebrotomy. (B) Tranverse section at the level of the vertebrotomy. Removal of the posterior wall of the tunnel (*shaded*) exposes the prolapse. A better angle of access can be gained by removing the anterior corner (*C*). This approach is advised because it avoids a manipulation of the thoracic spinal cord in its tight canal. (*Fidler, M. W., and Goedhart, Z. D.: Excision of prolapse of thoracic intervertebral disc. J. Bone Joint Surg.. 66B:518, 1984.*)

Spangfort[309] found that in lumbar disc surgery for all age groups, the best result, success in approximately 90%, was achieved in patients with complete herniation, followed by approximately 80% in patients with incomplete herniation and approximately 60% in those with a bulging disc. This 60% success rate is about the same as with most nonsurgical treatments. It is interesting to note that in those patients in whom there was no herniation at surgery, the pain relief was about 35%, the same as the placebo effect (Fig. 6-48).

In the cervical spine, we generally fuse at the time of disc excision (see Chap. 8). In the lumbar region, arthrodesis is not required after single disc excision. Techniques and rationale for various decompressions and fusions are discussed in Chapter 8.

Results

In the cervical spine, anterior disc removal with arthrodesis yields good or excellent results in 63–73% of patients.[275,277,354,363] However, by selecting the patients in whom the myelographic defect corresponded to the level at which surgery was performed, the percentage of patients with good or excellent results increased to 77% in one series[363] and to 91% in another.[354] Here, as in the case of lumbar spine disease, the percentage of good results increased with more accurate selection of the distinctly herniated disc. Results comparable to the preceding groups (83% improved) have been reported with cervical disc disease treated with disc removal without fusion.[153]

The result of surgery for thoracic disc disease is generally not good.[309] This is due to the severity of the pathoanatomic disease process, the characteris-

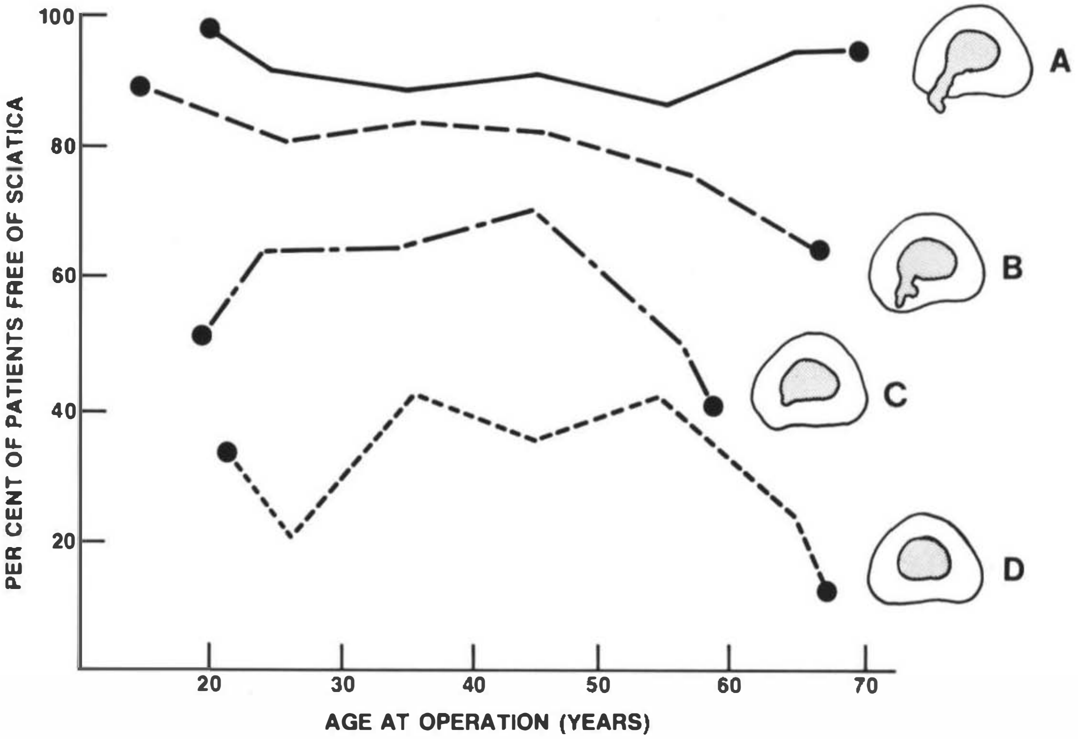

FIGURE 6-48 This graph shows the effectiveness of disc excision in the relief of sciatica in various age groups as a function of the actual pathology found at surgery. (**A**) Complete herniation; (**B**) incomplete herniation; (**C**) bulging disc; and (**D**) no herniation. *(Modified from Spangfort, E. V.: The lumbar disc herniation. A computer-aided analysis of 2,504 operations. Acta Orthop. Scand., 142 [Suppl.], 1972.)*

tic delay in diagnosis, and the relative lack of surgical experience in dealing with the problem. Patients without weakness, with a monoparesis or with absent or minimal sensory changes, generally show good to excellent results with surgery. Those with more severe neurologic deficit have a poor postoperative result. We are optimistic that with the newer emphasis on anterior approaches,[7,104,253] results for this rare form of surgery will improve.

The results with lumbar disc surgery have been carefully studied and have been shown to vary with the guidelines with which the patients are selected for the procedure. The trend noted in the cervical spine is distinctively demonstrated in the lumbar spine. If, through a careful preoperative clinical evaluation, the surgeon can select the patient with a complete herniation, the results with surgery are best.

Dunkerley operated on patients with a history and physical findings of disc disease, but without routine myelograms. The success rate among patients in that group was 75%.[82]

Nachemson, in a comprehensive review of the literature, demonstrated convincingly the importance of a thorough preoperative evaluation in order to accurately diagnose a disc herniation.[232] Using history and neurologic examination alone, a physician may expect to be correct 60% of the time. If a positive straight leg raising test is observed preoperatively, a herniation is present 70% of the time. Add a positive electromyogram to the evaluation, and the accuracy increases to 80%. If, in addition to the preceding, there is also a positive *water-soluble* contrast myelogram, the surgeon can expect to find a disc herniation in 90% of patients (Fig. 6-49). It may be that MRI will improve the accuracy beyond this 90% level.

Complications

There are always the standard risks of medications, anesthesia, and blood transfusions that are a part of surgical therapy. Complications associated with cervical spine disc surgery may be divided into major and minor. Minor complications subside without

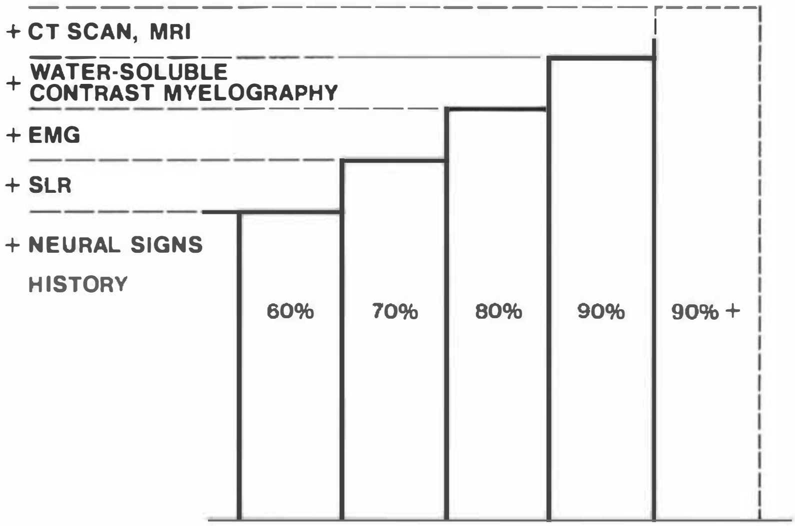

FIGURE 6-49 This histogram shows how the percentage of accurate diagnoses of disc herniation can be increased as more aspects of the clinical evaluation are found to be positive. It is reasonable to assume that the percentage of accurate diagnoses can be improved upon by using either the CT scan, MRI, or both. (*Nachemson, A. L.: The lumbar spine, an orthopaedic challenge. Spine, 1:59, 1976.*)

prolonged or vigorous treatment. They are hoarseness, dysphagia, and hematoma at the iliac donor site. Major complications include graft slippage, donor site problems, infections, pneumothorax, puncture of the esophagus or large vessels, and spinal cord damage.[354]

The lumbar complications include cauda equina damage; penetration of the aorta, vena cava, or iliac vessels; and wound infections.

Comments

Cervical and lumbar spine surgery is justified and indicated for carefully selected patients for whom the risk/benefit factor is appropriate. Surgery is mandatory for thoracic disc herniation once the diagnosis is made. Acute cervical myelopathy and cauda equina symptoms with bladder paralysis are situations in which immediate surgery is desirable.

Patients with disc disease, usually a large, centrally herniated L5–S1, who have saddle anesthesia bladder and/or rectal paralysis, and motor deficit are diagnosed as having cauda equina syndrome. Surgery is not elective but urgent. The surgery should be done as soon as possible, ideally within 2 weeks of onset of symptoms. While it is logical to assume that the longer the nerves are compressed, the less the neurologic recovery, published studies to date have not demonstrated this. Nevertheless, it is suggested that surgery not be delayed once the diagnosis is made.[190]

It is good to keep in mind that even though 90% of patients will enjoy the relief of leg pain after elective, carefully selected lumbar disc surgery, as many as 60–70% may continue to have some low back pain.[232] Subjective and objective information that should be of value to the clinician in discussions and deliberations about the surgical treatment of spine pain is provided below.

Who should do disc surgery? Selecki and colleagues studied a group of patients with low back pain and sciatica, some of whom were treated by neurosurgeons and others by orthopedic surgeons. The percentages of patients in the two groups who were successfully treated were nearly identical—59% in the orthopedic group and 63% in the neurosurgical group. The neurosurgeons operated on a larger percentage of their patients than did the orthopedic surgeons (75% vs. 56%). However, patients with radicular signs and symptoms tended to be in the neurosurgical group, whereas patients with only low back pain were more likely to be in the orthopedic group.[292]

Table 6-4 includes a partial summary of the treatment discussed. We have attempted to rank the various treatment programs. There are some oversimplifications. McKenzie exercise, for example, are rated very high. This is based on just one study, and additional studies are required to confirm and establish the superiority. The same situation exists for percutaneous discectomy. The reported percentages successfully treated have been informally averaged. However, they reliably represent the trends found in the literature. The time span of treatment has not been standardized. It is known that with waiting and doing nothing, 70% of patients will be well in 3 weeks and 90% in 2 months.[232] However, the table is

TABLE 6–4 Comparison of Treatment Regimens for Spine Pain

Relative Therapeutic Effectiveness	Risk Factors
Outstanding (>80%)	Serious
Disc excision (conservatively selected) McKenzie program*	Disc excision Chemonucleolysis
Good (70–80%)	Moderate
Isometric truncal exercises Basic patient education and group therapy Spinal manipulation† Disc excision (not conservatively selected) Chemonucleolysis Percutaneous discectomy*	Rest, analgesics, and anti-inflammatory drugs Williams exercises Spinal manipulation Local and regional use of drugs Percutaneous discectomy*
Average (60–70%)	Minor or Insignificant
Rest, analgesics, anti-inflammatory drugs Williams exercises Miscellaneous exercises Physical therapy (heat and massage) Axial traction Orthotic devices Local and regional use of drugs Disc excision (liberally selected)	Isometric truncal exericises Physical treatment (heat and massage) Basic patient education and group therapy programs Axial traction McKenzie Program

* Needs more clinical study.
†Short-lived benefit.

based on reported results in the literature that evaluate and/or compare some of the different treatment regimens.

Table 6-4 also summarizes the relative risks of the different regimens, based on reported complications and the authors' opinions. Risk represents a combination of frequency and severity of complications associated with the various treatments. With percutaneous discectomy the complications are very rare; however, when one occurs (disc space infection) it could be moderate or severe. Because infection is so rare with the procedure, we categorized the risk factor as moderate.

Although expected results and relative risks are not the only factors involved in the selection of a given form of treatment, they are important considerations. Availability of facilities and techniques, and individual patient and physician preferences are also major factors. For nonoperative treatment, isometric exercises, patient education, and especially the McKenzie program rate highly with respect to effectiveness and low risk. Spinal manipulative therapy is effective, and there is only moderate risk, provided the patient is properly diagnosed and does not have tumor infection or impending disc herniation. Other forms of conservative treatment may be employed. Obviously, drugs are frequently used effectively. Guidelines suggest that if the patient is not improved in 2 to 6 months with nonoperative treatment, surgery may be indicated, provided that the history, physical findings, and myelogram are indicative of a herniated, cervical, or lumbar disc or cervical spondylosis. When a lumbar disc problem is diagnosed and presumed to be present for 6–8 weeks, it is thought to be better to operate at about this time for the best results.[346] As mentioned previously, the diagnosis of a herniated thoracic disc is an indication for anterior discectomy.

Well, Doctor, what will happen if I don't have surgery? The study by Hakelius is of considerable help with respect to discussion with patients, decision making, and recommendations about the advisability of surgery in the presence of a herniated disc. The 583 patients in the study fit the subjective and objective clinical criteria for the diagnosis of disc herniation at the L4–L5 or L5–S1 level. The patients had symptoms for no more than 6 months. Disc exploration and removal of the herniated fragment was accomplished through no more than a partial unilateral laminotomy in 166 (28.5%) of the patients. The remaining patients were treated with varying degrees of relative immobilization. Both the conservatively managed and surgically treated patients were followed for an average of 7 years, 4 months.

The salient findings were as follows. When there is a distinct prolapse, the surgically treated patient experiences a speedier relief from sciatica and loses less time from work than the conservatively managed patient. The time away from work following surgery is likely to be less if the conservative treatment prior to surgery has taken less than 2 months. *However,* the study also showed that acute sciatica with neurologic symptoms is a transient condition and with few exceptions will subside with time. *Moreover,* in this study the results of the surgically and conservatively treated patients in this series were almost identical 6 months after the start of treatment. Table 6-5 presents a comparison of the two groups in more detail. The investigation reported that there were several points that indicated a slightly better prognosis for the surgically treated group of patients. These results are the patients' subjective evaluation of the improvement of their low back pain and sciatica. Also, follow-up treatment showed that the surgically treated patients took less sick leave for low back pain and sciatica than did the conservatively treated patients.[138]

The more recent work of Weber presents a somewhat better outlook for surgery. At the end of 1 year, the patients who had surgery were statistically significantly better. However, with the long-range follow-up at 4 and 10 years, the operated and nonoperated groups were the same.[348]

But what about my leg, Doctor? Will it be weaker if I don't have surgery? A prospective study by Weber showed that 1 year following surgery, the surgically treated patients were no better than the conservatively treated patients with regard to objective measurements of motor function.[347]

All right, I need surgery, but when should we operate, Doctor? How long can I wait without increasing the time lost from work? The amount of sick leave required after surgery is less if the preoperative conservative management lasts for less than 60 days.[138] Ninety percent of patients will be better by that time.[232] This suggests that physicians should wait at least 2 months before recommending surgery. Patients who undergo surgery after 1 year of incapacity have roughly a 50% less chance of gaining relief of symptoms than those patients treated by surgery sooner.[292] The poorer surgical results with long-standing preoperative disease may be contributed to by two factors—nerve root fibrosis secondary to long-standing compression and psychologic factors resulting from prolonged pain and disability.

Although this objective information is most helpful, there remains the problem of trying to weigh the disadvantages of exposing the patient to the risks of anesthesia and surgery against the pain, disability, and time away from work that may be involved in waiting. We have been especially attentive to the following hypothesis posed by psychologists: When a patient has had pain for as much as 1 year, the removal or correction of the organic source of that pain alone is not enough to free the patient of the pain. In addition to the surgical or other medical treatment, some psychologic treatment may also be required.

Spine Arthrodesis

The biomechanical rationale for arthrodesis to treat spine pain is based on the hypothesis that immobilization of the FSU should reduce or eliminate any pain associated with that particular FSU. The rationale, biomechanics, and indications for spine fusion are presented in detail in Chapter 8. The immobilization or, more realistically, increased stiffness of the FSU is thought to eliminate or decrease irritation at the intervertebral disc, the intervertebral joints, or other pain-sensitive structures.

Biomechanical studies creating an experimental

TABLE 6–5 Comparison of Results of 166 Surgically Treated and 417 Conservatively Treated Patients with Herniated Discs

Post-Treatment Results	Surgical Removal of Disc (%)	Conservative Treatment with Relative Immobilization (%)
Reduced working capacity	12	15
Complete loss of working ability	0.75	1.5
Restrictions in leisure activity	15	15
Regular sleep disturbances	20	13
Sick leave for back pain/sciatica (90 days)	13	14
Pronounced residual sciatica	12	20
Pronounced residual paresis	20	16
Pronounced subjective motor symptoms	7	6
Pronounced subjective sensory symptoms	8	5
Objective sensory loss	34	33
Surgery for recurrences	5	6

(Hakelius, A.: Prognosis in sciatica. A clinical follow-up of surgical and non-surgical treatment. Acta Orthop. Scand., 129[Suppl.], 1970.)

floating fusion in the lumbar spine showed that there was *increased* motion at an adjacent FSU, usually the one below the fusion.[271] Theoretically, this could be a liability leading to strain instability or spinal stenosis. However, we are not aware of any definitive clinical documentation that this actually occurs.

In the cervical spine, arthrodesis in addition to disc excision has not been proved to be superior to simple disc excision. However, with cervical spondylosis that involves osteophytes, joints of Luschka, and degenerative arthritis, arthrodesis is likely to give a better result. This observation has not been statistically documented, however.

In the lumbar spine, the evidence seems to indicate that there is no particular advantage to routine arthrodesis at the time of disc excision.[118,244,292,348] A recent prospective controlled study by White and associates[355] showed that excellent results were more likely to occur in laminectomy/discectomy patients *without* fusion than in those having the same procedure *with* fusion.

The biomechanical studies of Rolander (see Chap. 8) demonstrated that even with solid union of the posterior elements there was sufficient motion between the vertebral bodies with physiologic loading to cause irritation of the disc. For lumbar spine pain in the absence of spondylolisthesis, arthrodesis is very unlikely to succeed in the eradication or even satisfactory alleviation of spine pain. This is based primarily on the overall lack of clinical success with this procedure.

There are some intriguing problems raised by observations of patients following spinal fusions. Pseudarthroses are difficult to diagnose and when present are not always painful. In cervical spine fusions for spondylosis, patients with failure of fusion are sometimes relieved of pain.[354]

Newman observed that some patients treated surgically for lumbar spondylolisthesis had satisfactory relief of pain even though they developed pseudarthrosis.[329] Shaw reported that 77% of his patients with failed fusion were satisfied with their results.[328] Other surgeons have made similar observations.[282,328] The relationship between successful arthrodesis of the spine and the relief of pain remains obscure. Spinal fusion for the treatment of spondylolisthesis and some of the other radiographic entities that are very likely to be associated with spine pain appears to produce the most consistently good results with regard to pain relief. As shown in Chapter 8, the results of fusion alone for the treatment of other spine pain are variable and unpredictable.

Salvage Procedures and Failed Backs

This is a discussion of the advisability of performing surgery for a painful spine on a patient who has had two or more such operations. It is generally accepted that the probability of a satisfactory result is one in ten, or less. Yet in a neurosurgical, orthopedic, or pain referral center, one may encounter a patient who has had as many as 13 operations for low back pain.* When should a so-called salvage procedure be done, what should the procedure be, and what is the prognosis? These questions are not definitively answered at present. However, there are some rational guidelines that are moderately supported by clinical studies.

The third operation is generally one of extensive decompression of the cauda equina and nerve roots in the anatomic areas where they are found to be compromised. This is usually done in conjunction with fusion of all involved FSUs.

A group of 54 patients who had had two unsuccessful disc operations were studied at the Mayo Clinic. One-half of the patients were treated conservatively and the other half had a third operation. The patients were not assigned to the two groups on either a matched or a random basis. Thus, there was no statistical evaluation. The conservative therapy consisted of various spine orthoses, rest on a hard mattress, limitation of activity, heat massage, and special exercises. The surgery involved disc removal and/or spinal fusion. The patients treated by salvage surgery tended to show a better result. However, this group had a higher incidence of complaints of sciatica prior to the salvage procedure than did the conservatively treated group. This may account for the better results observed in the surgically treated group.[175]

Another report on salvage surgery for low back pain emphasized certain factors in the selection of patients. Patients with psychiatric problems and patients involved with compensation and litigation were not as likely to achieve an acceptable result with yet another operation. However, the history of a 1-year, pain-free interval following the last surgery was highly correlated with a successful salvage procedure.[105]

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* Personal communication, F. D. Wagner, August 1972.

Mechanical compression by disc, or spinal stenosis and the presence of lumbar instability or a pseudarthrosis, also increased the likelihood of a successful outcome. At a 2-year and 4-year follow-up, 80% of patients thought that the procedure was worthwhile.[106]

The fact that *recurrent* disc protrusion was found to be a problem in this series may be considered an argument for removal of the entire disc as well as the displaced fragment at the initial operation. Because the recurrence rate is so low (2%) in limited discectomy, we agree with the investigators who recommend excision of the displaced fragment only.[138,232,313]

Each patient should be evaluated and managed as an individual. A thorough psychologic evaluation should precede any salvage procedure. If the history indicates that the patient returned to work after previous procedures, there may be a more optimistic prognosis. When a specific pseudarthrosis is to be treated, there may be a greater chance for success. The basic surgical procedure adequately frees the nerve roots from scarring, impingement, or compression and reduces motion with arthrodesis. Thus, once all of the involved discs have been removed, the nerve roots have been freed through their course in the neural foramina, and the involved segments have been fused, there should as a general rule be no more spine surgery in the region. It is certainly desirable to avoid contributing one or more episodes to the saga of the painful, unhappy, totally disabled patient who has undergone multiple spine operations.

Salvage spine surgery probably has the best prognosis when done with a patient who has had a 6-month to 1-year pain-free interval, is not dependent on drugs, and is psychologically normal with objective evidence of radiculopathy and imaging evidence of disc or bone material on the nerve.[106,200] Strong clinical evidence of instability (see Chap. 5) or a pseudarthrosis would forbid a good outcome.

The surgery should include exploration and decompression of the nerve roots (*i.e.*, removal of offending disc tissue and osseous tissue and release of scar tissue that binds the nerve to other structures). When there is evidence of instability, or if the surgery at decompression is thought to create instability, an arthrodesis of the appropriate level(s) is completed. Wadell and associates, however, recommended fusion routinely at the time of the second operation in compression patients. This is because the second operation in their view was the *last chance* to make the patient better, and in our experience a third operation is likely to make the patient worse.[341] We believe that the risks/benefits are such that fusion should be done only when there is a specific indication.

A Regimen to Consider

Following a thorough diagnostic evaluation, any specific diagnosis that is made is treated appropriately either directly or by referral to the appropriate specialist. The remaining patients have a diagnosis of nonspecific organic low back pain and/or a herniated intervertebral disc. Both groups are treated essentially the same way, by waiting 6–8 weeks and employing one or more forms of conservative therapy. During that time, rest, activity, and anti-inflammatory narcotic and non-narcotic drugs are prescribed as indicated. We believe that time is the most important factor. However, any combination of physical traction and/or exercise may be employed, depending on the physician's clinical judgment and evaluation of the particular patient. If 2 to 3 weeks of one form of conservative therapy is not effective, it is advisable to switch to some other program. This is true even with bed rest. Sometimes a patient who does not improve with bed rest will do better with moderate activity and/or some other form of treatment. The time suggested for bed rest is decreased. A recent investigation showed that 1 or 2 days of rest may be better than resting for 1 week. For the reasons previously discussed, it is very difficult to determine why a particular patient improves.

When both the patient and the doctor decide that it is time to give up (the suggestion may come from either person), the patient is then scheduled for a CT scan or an MRI or is hospitalized for a myelogram. If the study is positive, a limited discectomy is planned. If the study is not positive, other diagnostic tests may be indicated, and the patient is followed and managed as indicated by diagnostic tests and the future course of the symptoms.

Guidelines for the management of patients with acute and subacute back pain have been presented.[108,314]

For cervical spine pain, the management is essentially the same. The nonsurgical treatment may include a cervical orthosis of *intermediate control* (see Chap. 7). Elective surgery requires an abnormal imaging and involves disc excision and Smith-Robinson fusion at one level or the two adjacent levels that best fit the clinical data.[354] This depends upon pain distribution, evidence of sensory and neuromuscular deficit, and location of cervical osteophytes in the intervertebral foramen.

PROPHYLAXIS AND ERGONOMICS

How can spine pain be avoided once a patient has been so afflicted, and how can the severity and probability of recurrence be minimized?

Certainly, it is reasonable to assume that spine pain resulting from most causes can be aggravated by mechanical factors. Prophylaxis and ergonomics are obviously intimately related to treatment, and there is some overlap with physical therapy, exercise, and low back pain school.

Prophylactic measures for avoiding spine pain are numerous. Those related to trauma involve all of the well-documented practices of street, road, and highway safety, which are generally well known. The value of seat belts and headrests is discussed in Chapter 4. The epidemiologic data show that truck driving and automobile commuting predispose certain people to the development of disc disease. A patient with spine pain that is related to disc disease should be advised to limit automobile riding and especially driving.

Sports injuries are another significant source of spine trauma. The basics of good conditioning, coaching, and proper supervision are of obvious benefit. Proper equipment and changes in rules may be helpful in preventing some of the sports injuries. Education of the public concerning recreational dangers, proper conditioning, and the use of "common sense" is valuable in the reduction of spine pain caused by trauma.

Perhaps changes in some of the compensation laws and practices may be a form of prophylaxis against spine pain.

In general, individuals in good physical condition are less likely to have back pain.[46] However, there is some evidence that supports the rationale that working specifically to strengthen trunk muscles may have protective or preventive value. A large study of flexibility and trunk muscle strength in 449 men and 479 women suggested that good *isometric* endurance of back muscles may prevent first-time back trouble.[26] Mayer and associates demonstrated that an impressive 87% of a selected group of chronic low back pain patients returned to work and remained there after 2 years.[214] These patients were treated with functional restoration, which included trunk-strengthening exercises in conjunction with a comprehensive psychologic support program. This record superseded the success rate in a control group by a factor of five.

Postural Biomechanics

The issue of ergonomics and prophylaxis calls attention to the question concerning the comfortable and desirable position of the lumbar spine. Some physicians suggest that the reason bars have bar rails is that while the customers enjoy conviviality and ethanol, they may be more comfortable with the back in the slightly flexed, relatively straight or nonlordotic position (Fig. 6-50). We have reviewed the cultural hypothesis that states that this position is used more frequently in populations that complain less often about low back pain. However, this statement is not documented. Experimental studies show that there is more of a posterior bulge of the disc with extension than with flexion. In contrast, there is evidence that shows that the reclined, slightly extended, and relatively unloaded spine has less pressure in the intervertebral disc and less activity in the paraspinous muscles. These studies are reviewed in the next section. Although these biomechanical data support the use of a reclined, slightly extended position at least for sitting, the definitive resolution of the question requires additional information. Meanwhile, we suggest that the most comfortable posture for each patient be determined and that the patient be advised and taught to maintain it.

Biomechanics and Sitting

There has been some interesting biomechanical information related to the question of sitting. These data are important to both prophylaxis and ergonomics because workers often perform their job in the sitting position. The investigations have analyzed the effects of backrest inclination and lumbar support on L3 intradiscal pressure and quantitated electromyographic recordings.[13] The goal was to determine the seat type and reclining angle that was associated with the lowest disc pressure and the least paraspinal muscle activity. The hypothesis is that these characteristic seat types and angles might be the least stressful and the most therapeutic for the

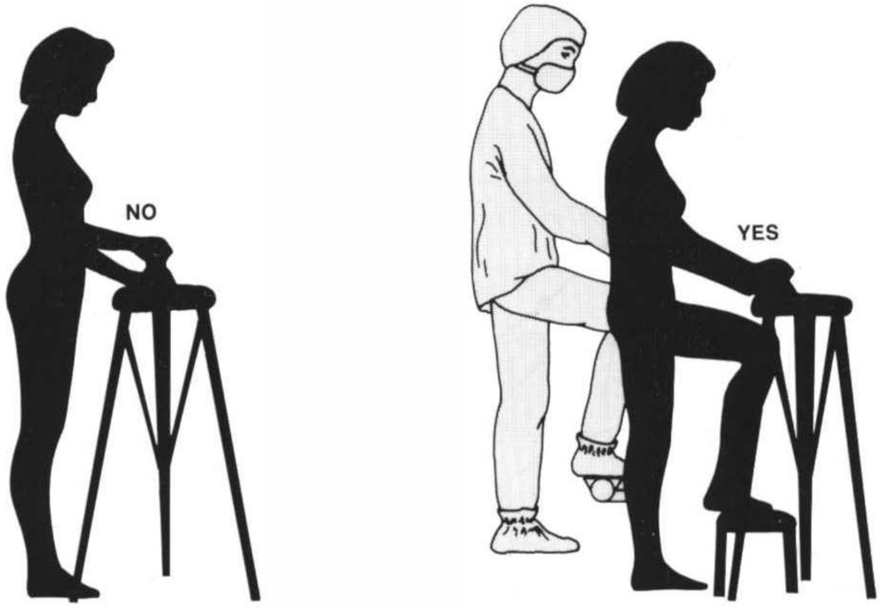

FIGURE 6-50 Flexion of the hip reduces the tension of the psoas muscle and the lordosis of the lumbar spine, resulting in reduced loads on the lumbar spine. This bit of ergonomic advice is particularly important to the surgeon who may stand for several hours on occasion at the operating table. For housework, the work position shown is advised. This is a valid ergonomic principle.

spine. The subjects also reported their subjective feelings about the comfort of the seats following several alterations of the two variables. A number of important points come from this study. It is assumed that the lowest disc pressure and the least electromyographic activity of the paraspinal muscles are the most desirable situations. The lowest electromyographic and intradiscal pressure recordings were found with a backrest inclination of 120° and a 5-cm lumbar support. The highest intradiscal pressure was found in the situation in which there was no lumbar support and a 90° inclination (i.e., a straight back; Fig. 6-51).

More recent work by Andersson and associates in 38 healthy subjects showed that the lumbar support had the greatest influence on lumbar lordosis and the inclination of the backrest had the most influence in reducing loads on the lumbar disc.[10] Increasing the incline simply shifted more weight onto the backrest and therefore reduced the load on the disc as well as the force extended by the erector spinae muscles necessary to maintain erect equilibrium.

In view of the study by Kelsey and Hardy in which the vehicle driver was found to be at risk to develop sciatica,[182] we recommend that a lumbar support be used by the frequent or the symptomatic driver. The study showed that the use of armrests as well as lumbar supports reduced intradiscal pressure. The use of arm supports in addition to lumbar supports is recommended for the symptomatic and frequent driver.

Thigh support and adequate space for alteration of position are also positive factors in the ideal seat design. An example of good seat design is shown in Figure 6-52. Obviously, for the worker there are practical considerations about where the person must look and what he or she does with the arms and legs. These and other factors may require design adjustments that might interfere somewhat with the use of the ideal seat.

There has been considerable interest recently in a new type of chair called a Balans chair. There is no back support, and the weight is borne on the knees, shins, and buttocks. A study of this chair showed that it was not more comfortable, nor was it associated with less trunk muscle activity as reflected by EMG.

Biomechanics of Work Activity

Ergonomics is important in many ways to prophylaxis in different work situations.[299] Crucial to ergonomics is the question concerning the proper way to lift an object. The loading mechanics of the lumbar spine, for example, are such that any increased load that is anterior to the vertebral bodies greatly increases the forces that are exerted on the lumbar spine. This is due to the forces that must be

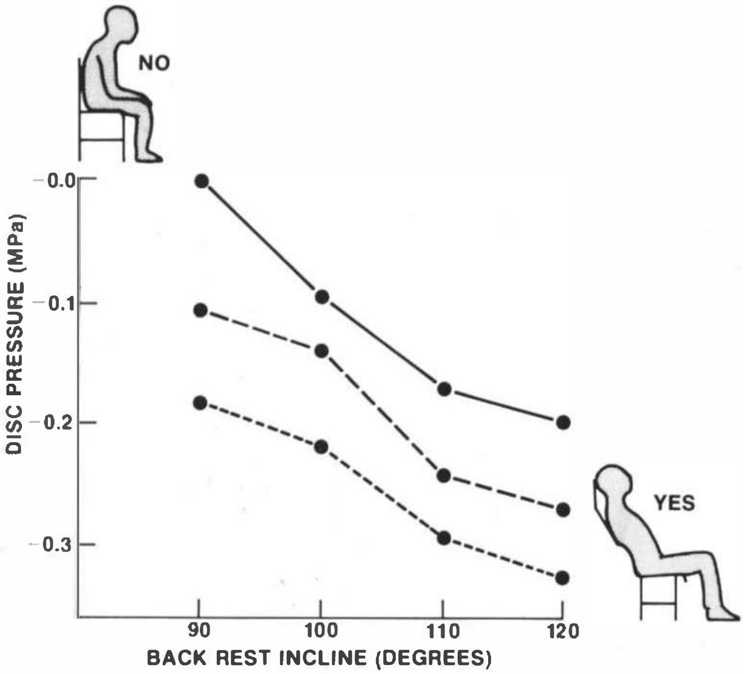

FIGURE 6-51 This graph depicts the effects of the variables of back rest inclination and size of lumbar support on intradiscal pressure. The solid line represents no lumbar support; the large dashed line represents a 3-cm support; and the small dotted line represents a 5-cm lumbar support. *(Nachemson, A. L.: The lumbar spine, an orthopaedic challenge. Spine, 1:59, 1976.)*

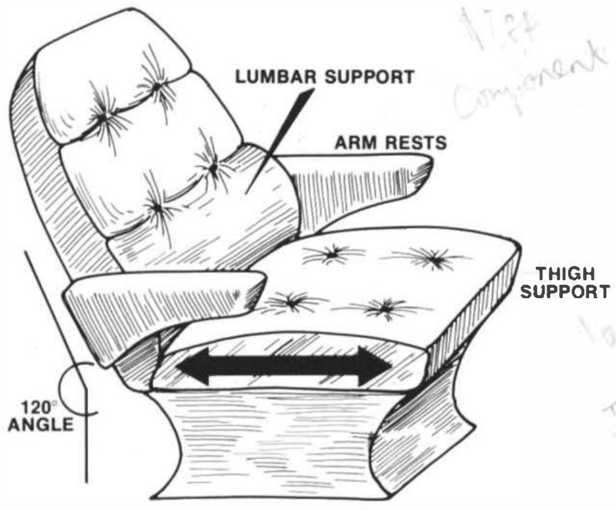

FIGURE 6-52 The current available knowledge suggests that this is the ideal chair biomechanically for comfort and relaxation (not desk work). This takes into consideration the proper inclination, lumbar support, arm rests, thigh support, and space to move around and change position.

lift
Component 4
3
4
load

extended by the paraspinous muscles in order to maintain equilibrium. The resultant forces at the fulcrum, which is the lower lumbar FSU, are very high. This is shown in Figure 6-53.

Although there has been suitable emphasis of the importance of leg lifting as opposed to back lifting,[228,232] the distance of the object from the body at the time of lifting has also been shown to be a very important ergonomic consideration.[11-13,226,251] Simultaneous electromyogram and truncal and intradiscal pressure measurements were made while normal subjects went through different types of lifting procedures. These studies showed that with all three procedures, the distance of the weight from the body was directly related to high measurements. This is due to the high forces necessary to maintain equilibrium because of an increased lever arm (see Fig. 6-53). There is a larger joint reaction force (high intradiscal pressure), a greater force required by the erector spinae muscles (high electromyographic activity), and a need for greater truncal support to protect the spine (high truncal pressure).[11-13] This shows the significance of lifting with the object close to the body. It is of considerable importance in both industrial and domestic ergonomics. Recent CT studies by Nemeth and Ohlsen[243] showed that the average erector spinae moment arm is 68 mm, a greater distance than has been used in most biomechanical modeling of this situation. The authors also demonstrated a significant difference in the erector spinae moment between males and females. This information, when taken into consideration, will improve the accuracy of future biomechanical

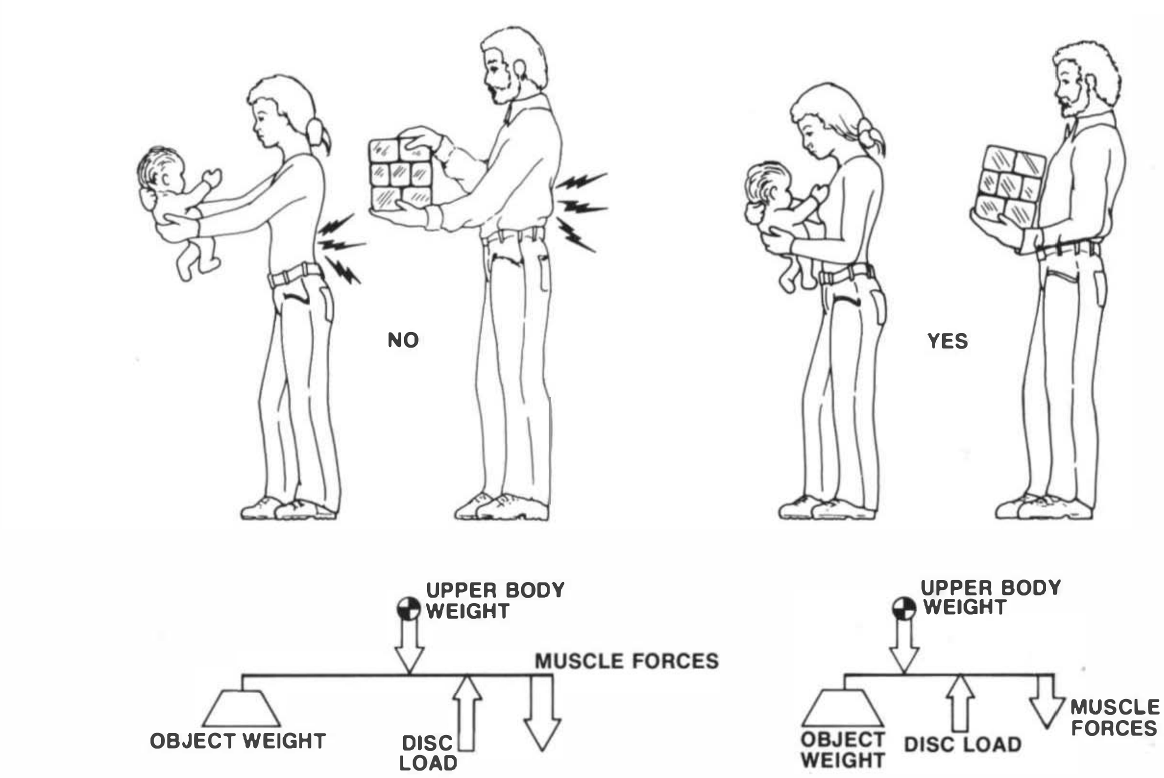

FIGURE 6-53 The ergonomics of proper lifting is shown in this diagram. The load on the discs is a combined result of the object weight, the upper body weight, the back muscle forces, and their respective lever arms to the disc center. On the left, the object is farther away from the disc center compared with the situation on the right. The lever-balances at the bottom show that smaller muscle forces and disc loads are obtained when the object is carried nearer to the disc.

modeling of the role of the erector spinae muscles in exerting forces on the lumbar spine. The reader interested in a thorough and detailed explanation and a free-body diagram is referred to the work of Hayes and associates.[142]

Several experiments have been carried out in order to correlate abdominal and/or thoracic cage pressure with quantity of weight lifted by different subjects.[71,91,174] These studies showed a correlation between the ability to increase the fluid pressure in the two cavities and the amount of weight lifted (Fig. 6-54). This attests to the value of abdominal and thoracic cage muscles in supporting the spine when it is carrying heavy loads. In view of this work, we suggest good muscle tone, especially for the abdominal muscles. However, the strength is not to be developed by sit-ups. Isometric abdominal exercises achieve the same goals without excessive intervertebral disc loading. In addition, this work is partial justification for the use of a spinal corset or a brace with an abdominal corset in situations where development of the abdominal and thoracic muscles (truncal pressure) is not feasible.

It has been shown that intratruncal pressures increase when heavy weights are lifted. The pressure increase is greater when heavier weights are lifted, and it is also increased when the speed of weight lifting is faster.[71,174] Studies of simultaneous intrathoracic and intra-abdominal pressures comparing pulling, pushing, and lifting were carried out. The results show that the largest pressures were recorded when subjects were pushing, and the smallest ones occurred during pulling.[71] It was also observed that during pulling, the back muscles were tense (Fig. 6-55A), while during pushing, the rectus abdominis muscle was tense (Fig. 6-55C). Thus, the intratruncal pressures probably reflect the tension in the abdominal muscles. The biomechanical explanation

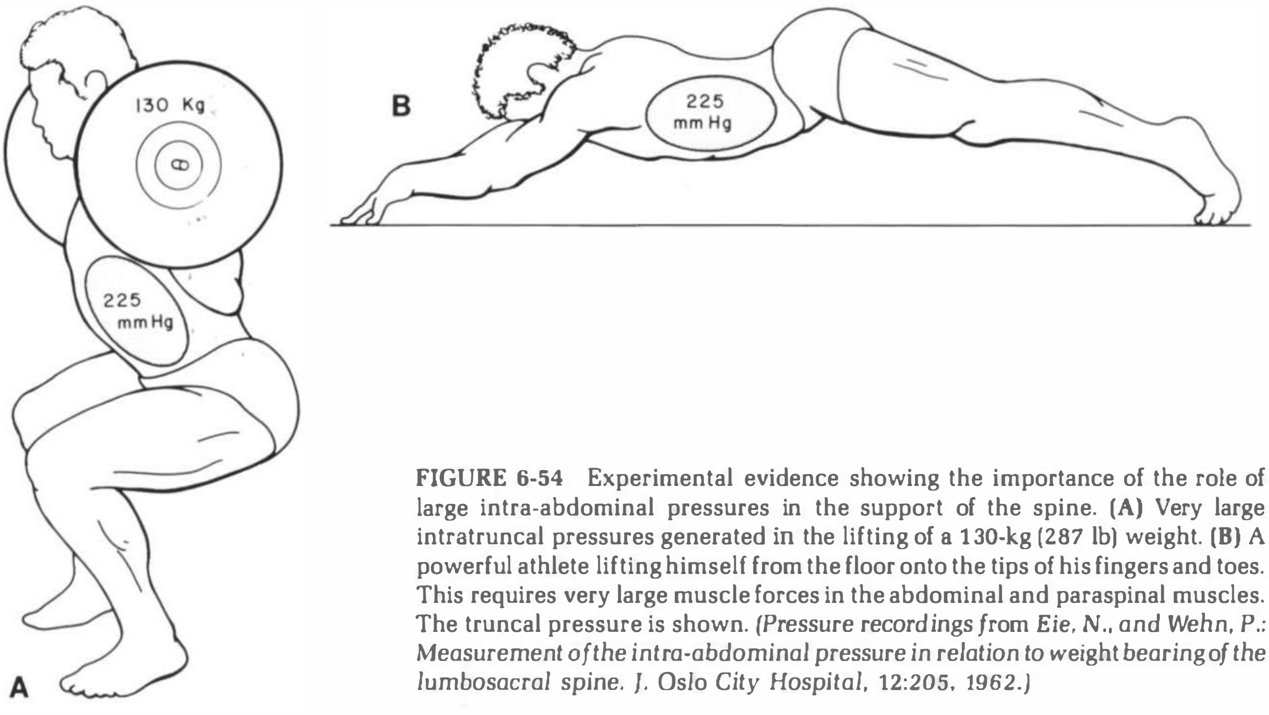

FIGURE 6-54 Experimental evidence showing the importance of the role of large intra-abdominal pressures in the support of the spine. **(A)** Very large intratruncal pressures generated in the lifting of a 130-kg (287 lb) weight. **(B)** A powerful athlete lifting himself from the floor onto the tips of his fingers and toes. This requires very large muscle forces in the abdominal and paraspinal muscles. The truncal pressure is shown. *(Pressure recordings from Eie, N., and Wehn, P.: Measurement of the intra-abdominal pressure in relation to weight bearing of the lumbosacral spine. J. Oslo City Hospital, 12:205, 1962.)*

for the reduced disc load in pushing versus that in pulling is depicted in Figure 6-55. The probable loads acting on a lumbar vertebra during pulling are seen in Figure 6-55B. The pull force, directed anteriorly, increases the bending moment and the erector spinae force considerably because of the short lever arm this muscle group has with respect to the axis of rotation. Thus, the disc load is also increased. The situation is different in pushing (Fig. 6-55D). The horizontal push force is now directed posteriorly. Its bending moment is counterbalanced by the rectus abdominis force. Because this muscle has a larger lever arm as compared to the erector spinae muscle, its force is relatively smaller. Thus, there is a smaller increase in disc load with pushing than with pulling.

Studies have been carried out to determine whether or not an isometric strength test of a person's ability to lift weights is correlated with the incidence of low back pain.[50, 51] The results showed that workers who were doing jobs in which their isometric test strength did not equal the strength required by their jobs had a much higher incidence of job-related low back signs and symptoms. The proper use of this type of testing and information can be most useful in the ergonomics and the prevention of spine pain. For the present, it is suggested that a patient returning to work involving lifting should have a thorough and well-executed program of isometric exercises beforehand.

Biomechanics, Sexual Ergonomics, and Low Back Pain

Although sexual disability often accompanies back pain, not very much has been written about it. The incidence of sexual problems in medical practice is higher among patients seen by physicians who routinely ask about sexual problems than among those whose physicians do not ask routinely. A study of married patients involved in a chronic pain center revealed a considerable number of sexual problems, such as sexual impairment, deterioration of quality of sex life, and decreased quantity of sexual activity. It is of interest that 86% of these patients had back pain or limb pain. Of two studies addressing back pain specifically, one reported impotence in 63% of 43 men with industrial back pain. The other involved 50 men with low back pain, 30% of whom reported loss of libido.

Detailed reviews of help for back pain patients with sexual disability are available for the advisor[116,252] and the advisee.[353] This section will present the biomechanical rationale and some very fun-

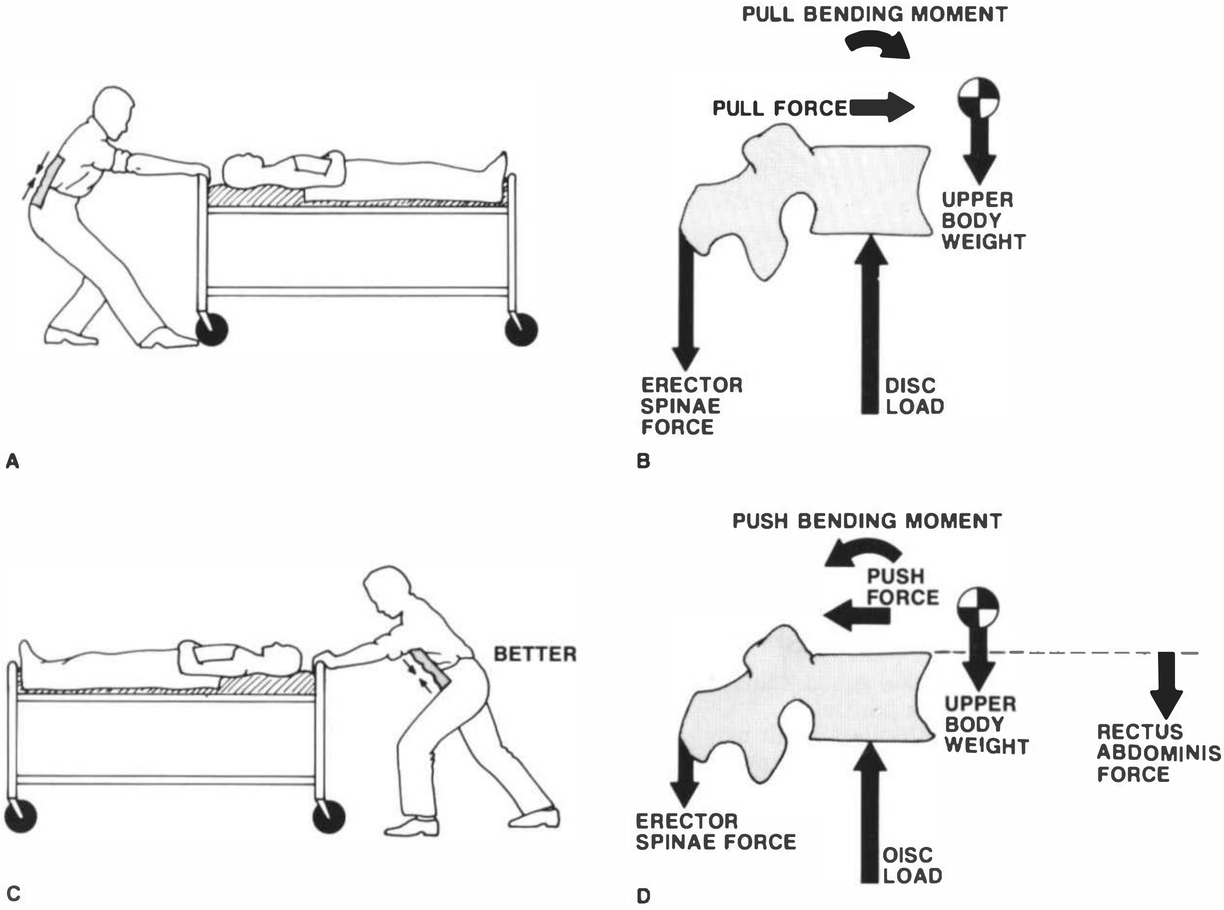

FIGURE 6-55 The ergonomics of pulling and pushing and the probable forces and bending moments involved. (**A**, **B**) During pulling, the erector spinae muscles resist the bending moment created by the horizontal pull force. (**C**, **D**) On the other hand, during pushing, the rectus abdominis resists the bending moment produced by the pushing force. Because this muscle has a much larger lever arm than the erector spinae muscles, its force requirement is relatively smaller. Therefore, the load on the disc is smaller during pushing than during pulling.

damental recommendations from the ergonomics point of view. See the chart on sexual ergonomics and Figure 6-56.

There is a great deal more to be presented in a full counseling program for patients; however, this basic advice is well founded in the best current knowledge about the biomechanics and general pathophysiology of low back pain.

Activities to Avoid

We believe that the following recommendations are well supported by the data depicted in Figures 6-57 and 6-58 and in Table 6-6. Patients with spine pain may be expected to aggravate their condition by coughing, straining, and laughing. Also, activities such as bending forward and lifting are associated with large increases in intradiscal pressure. A variety of exercises are prescribed in physical therapy for patients with back pain. Sit-ups with or without the hips flexed cause large loads to be exerted on the lumbar spine. The intradiscal pressure generated by sit-ups with or without the hips flexed is comparable to pressure generated by bending forward 20° holding 20 kg, hardly an exercise that a physician would suggest for a patient with acute low back pain.[C] An examination of Figure 6-58 shows that the patient with acute spine pain should also avoid straight leg

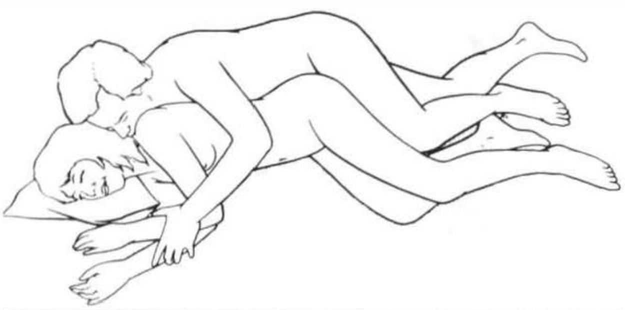

FIGURE 6-56 The side-lying position is the best all-around basic position for making love when one has a back problem. *(Reproduced with permission from White, A. A.: Your Aching Back. A Doctor's Guide to Relief. 2nd ed. New York, Simon and Schuster, 1990.)*

SEXUAL ERGONOMICS

- Don't bend forward with knees straight: this applies to both standing and lying positions.
- Avoid excessive swayback (lordosis). This may irritate the disc, nerve root, or facet joints.
- Avoid lying prone or supine if flat with hips and knees extended. This loads the lumbar spine through the psoas and also produces lordosis. Thus, the simple "missionary position" is potentially aggravating to the back-pain sufferer, top or bottom.
- Flexed hips and knees tend to relax the psoas and the sciatic nerve, straighten the lumbar spine, reduce disc bulge, and decrease loads on the facet joints. Thus, the side-lying position is fundamentally the best basic position for either partner with low back pain (see Figure 6-56).

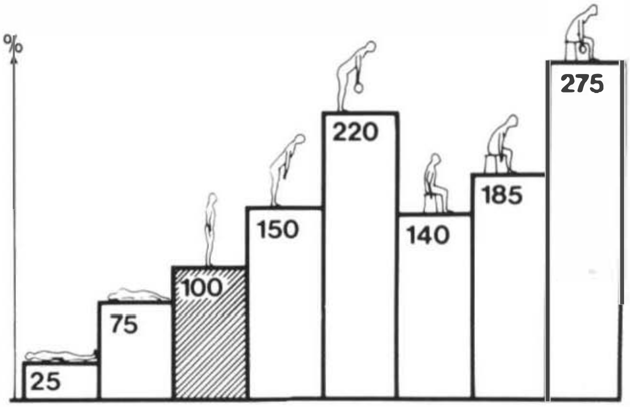

FIGURE 6-57 A diagrammatic comparison of in vivo loads (disc pressures) in the third lumbar disc during various activities. Note that sitting pressures are greater than standing pressures. *(Nachemson, A. L.: The lumbar spine, an orthopaedic challenge. Spine, 1:59, 1976.)*

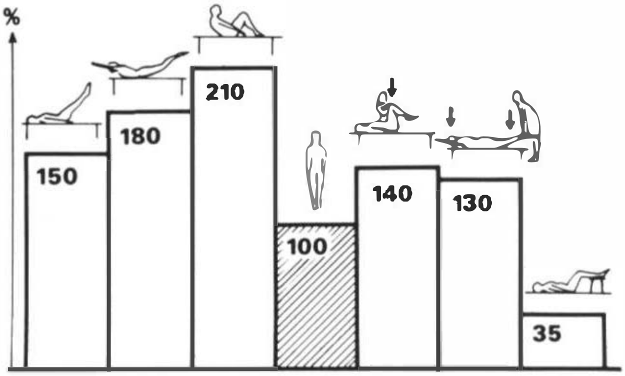

FIGURE 6-58 This figure compares disc pressures in vivo at L3 during various exercises and positions. Note pressures during sit-ups with legs bent, hyperextension exercises, and back lying with hips and knees flexed. *(Nachemson, A. L.: The lumbar spine, an orthopaedic challenge. Spine, 1:59, 1976.)*

raising exercises and lumbar hyperextension exercises. The least loads are exerted on the lumbar spine in the supine position with 30 kg of traction or in the semi-Fowler's position (see Fig. 6-35). The spine is also flexed, an excellent resting position for the patient with low back pain.

In one study, pressure-sensitive needles were injected into the third lumbar intervertebral disc of subjects who subsequently performed a number of tasks. Some of the most important activities are depicted in Figure 6-57. The chart is presented so that the various activities may be compared on the basis of percentage of standard. The standard selected was that of the force on the third lumbar disc recorded with the subject involved in normal standing. A large amount of valuable information has come from these data. The actual measurements are given in Table 6-6. A good deal of our prophylactic and ergonomic recommendations are based on and supported by these data.

Obesity greatly increases both the direct vertical compressive load on the spine and the anteriorly acting loads, which, through the action of the muscles, create very large joint reaction forces. The pan-

TABLE 6–6 **Nachemson's Data on Loads in the Third Lumbar Discs During Various Positions and Activities**[227,228,230]

Activity	Load (N*)
Supine in traction	100
Supine	300
Standing	700
Walking	850
Twisting	900
Bending sideways	950
Upright sitting, no support	1000
Coughing	1100
Isometric abdominal muscle exercise	1100
Jumping	1100
Straining	1200
Laughing	1200
Bending forward 20°	1200
Bilateral straight leg raising, supine	1200
Active back hyperextension, prone	1500
Sit-up exercise with knees extended	1750
Sit-up exercise with knees bent	1800
Bending forward 20° with 10 kg (22 lbs) in each hand	1850
Lifting of 20 kg (44 lbs), back straight, knees bent	2100
Lifting of 20 kg (44 lbs), back bent, knees straight	3400

* To obtain load in lbf, multiply by 0.225.

X1.5

niculus in Figure 6-59 is presented to emphasize this point. Consequently, obesity should be avoided.

A list of tips is shown below for patients who have or have had spine pain. Of course, none of them apply to all patients.

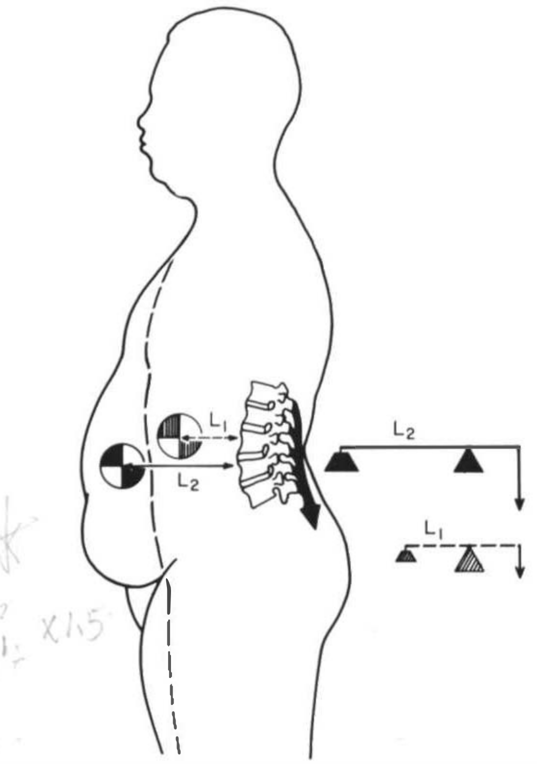

FIGURE 6-59 The mechanics here are the same as those in Figure 6-53, except that the weight here is adipose tissue rather than an external object. In the latter case, it is much easier to correct the lever arm. However, this diagram does emphasize yet another prophylactic and therapeutic value in avoiding or eliminating obesity.

PROPHYLACTIC AND ERGONOMIC TIPS FOR PATIENTS WITH SPINE PAIN

- Exercise to maintain painless range of motion and muscle tone.
- Avoid improper sit-ups and back extension exercises.
- When sitting, use a lumbar support and recline the seat when possible.
- Use armrests when possible.
- Move around within the seat; also, get out of the seat occasionally.
- Determine whether the flexed or extended lumbar position is better for you. Usually, straight or slightly flexed is better.
- Use this position in walking, standing, sitting, and lying.
- When lying in bed during severe pain, flex hips 90°.
- Try sleeping on the floor on three blankets during severe pain.
- Use a flat, firm bed otherwise.
- Lift with legs and with object close to you while doing a Valsalva's maneuver.
- Develop truncal muscles with isometric abdominal and erector spinae exercises.
- Be careful about opening and closing windows (especially postmenopausal women).
- Avoid sudden jerks or incremental loads when lifting or carrying.
- Avoid heavy lifting and strenuous activity when the back is symptomatic.
- Swimming is generally an excellent exercise in both acute and chronic phases.
- Avoid obesity.
- Sit on the bed or use a high table for changing diapers.
- Avoid pain-causing activities.
- When standing for a long period of time, elevate one foot on a footrest.
- Make love in the side-lying position.

■ CLINICAL BIOMECHANICS

Etiologic Considerations

■ From the biomechanical vantage point, in attempts to determine etiology, most of the attention is directed to the intervertebral disc and the facet articulations. However, other anatomic factors are important, and there are numerous psychologic and socioeconomic factors that are crucial to the clinical evaluation and management of spine pain.

■ Epidemiologic studies suggest somewhat paradoxically that individuals involved in sedentary occupations as well as those who are involved in heavy labor are prone to spine pain and/or radiculopathy. Those who spend a good deal of time driving motor vehicles, those who smoke cigarettes, and those who experience full-term pregnancy are also at risk. Vibration may be a factor for drivers, and increased loads associated with alterations in the mechanical properties of the ligaments may be accountable for the problem associated with pregnancy.

■ The following radiographically demonstrable conditions are thought to be causes of spine pain: spondylolisthesis; multiple, narrowed intervertebral discs; congenital kyphosis; scoliosis; osteoporosis; ankylosing spondylitis; and lumbar osteochondrosis. There does not appear to be any biomechanical thread that is common to these conditions.

■ Fibers of the annulus fibrosus during the process of maturation and aging may undergo fatigue failure or rupture during degeneration.

■ The acute back sprain is probably due to sudden loading that causes rupture of peripheral annulus fibrosus fibers or some of the other ligamentous or musculotendinous structures associated with the spine. A nondisplaced or minimally displaced vertebral end-plate fracture may also be responsible. However, this is speculation, and there has been almost nothing in recent research to help in the understanding of idiopathic organic spine pain.

■ Low back pain or cervical spine pain may be caused by irritation of the peripheral annular fibers associated with the passage of fluid into the nucleus pulposus.

■ Referred hip and thigh pain associated with back pain may result from irritation of the posterolateral annular fibers. This may occur in the absence of true radiculopathy, which results from irritation of the nerve root.

■ When there is significant bulge of the disc posterolaterally, there can be true sciatica or radiculopathy in addition to the low back pain.

■ A biomechanical analysis of the relationships between the nerve root anchored by ligaments above and below in conjunction with the force exerted by a herniated disc provided an interesting theoretical framework upon which to explain several important clinical phenomena.

■ The actual sequestered portion of the disc that can move randomly about, depending on the direction and magnitude of forces involved, may either be asymptomatic or cause any combination of spine pain, referred pain, or true radiculopathy. This pathologic condition may respond to axial traction or manipulation.

■ When there is a displaced sequestration, partially or completely fixed, the clinical picture is less changeable. If there is radiculopathy, it is distinct and more persistent. Axial traction or manipulation seems unlikely to be helpful in this situation, although both have been used with reported success.

■ A degenerating disc is associated with mechanical disruption, and there may be evidence of degenerative changes of the intervertebral joints.

■ There is evidence that the increased disc pressures associated with and combined with the biochemical changes in the nucleus pulposus may lead to pathologic disruption and failure of the annular fibers and ultimately disc herniation.

■ Of considerable importance in the problem of spine pain are all the various factors and combinations of factors that can compromise the space available for the neural elements and result in spine and radicular pain. The cross-sectional or transverse area of the thecal sac is shown to be the most significant measure for evaluating spinal stenosis.

■ There is some biomechanical correlation among loads, motion, and pain. The lumbar spine carries the heaviest loads, exhibits the second largest range of motion, and has the highest incidence of pain.

Diagnostic Considerations

■ Thoracic kyphosis, which is associated with heavy work before age 15, is also associated with thoracic spine pain.

■ Irritation of any painful structure adjacent to the subarachnoid space (nerve root, peripheral annular fibers) may be caused by any alteration of venous pressure. Thus, coughing, sneezing, or straining at the stool may cause severe spinal or radicular pain.

■ In Spurling's test, one takes advantage of the knowledge of cervical spine kinematics to position the vertebra so as to lower the threshold of sensitivity in order to irritate the nerve root in the intervertebral foramen. For the nerve root at the right intervertebral foramen, the patient should *look* and *bend* to the right. The opposite should be done for the left side. In each case, additional impact further compromises the foramen by approximating the vertebra and causing the disc to bulge. This causes radicular pain from the nerve that is compromised in a crowded foramen. The extension in the cervical spine is clinically useful, and its explanation and rationale are based on the biomechanics of the disc. Another useful test is the hand-on-head test, in which arm pain due to compressive monoradiculopathy will go away.

■ The sensitivity and complexity of neurologic problems in the thoracic spine due to disc disease are largely explained by the paucity of spinal cord blood supply and free space in the canal.

■ The patient with lumbar disc disease and sciatic nerve root irritation may stand with hip and knee slightly flexed, because this puts the least stretch on the nerve root at the site of irritation.

■ There is no longer thought to be a correlation between the direction of a patient's list with respect to the side of the sciatica and the position of the disc herniation in the canal. The list is important, however, as a harbinger of a poor prognosis when accompanied by sciatica.

■ The patient with camptocormia is able to straighten his spine in the reclining but not in the standing position because psychologic, not biomechanical, factors are involved.

■ Naffziger's test is based on the secondary pressure changes in the subarachnoid space around the irritated nerve root due to manual compression of the patient's jugular veins.

■ The leg raising tests, in which pain is elicited by sciatic nerve movement, are important and reliable indicators of nerve root irritation.

■ The patient with a "positive pedal pulse test" is able to flex the hip to 90° with the knee extended without pain when sitting but not when supine because of psychologic and/or socioeconomic, not biomechanical, factors.

■ The identification, development, and clinical correlations of Waddell's tests have provided a significant incremental usefulness in the evaluation of patients with low back pain.

■ Based on sound clinical studies and pathoanatomic and biomechanical analysis, there seems to be considerable legitimate controversy regarding the clinical usefulness of discograms.

Treatment of Spine Pain

■ Rest reduces mechanical irritation and is especially useful during the acute phase of the symptoms. There is a distinct trend toward using rest in the acute phase for several days rather than several weeks.

■ The goal of Williams exercises is to maintain the spine in the straight or flexed position. This seems to be reasonable, but tradition appears to have inflated the value of these exercises far beyond that which is justified by solid supportive evidence.

■ Sit-ups and active hyperextension exercises exert forces on the lumbar spine that are comparable to *improperly* lifting 20 kg (44 lb), hardly a therapeutic exercise for a patient with acute low back pain.

■ Erector spinae exercises may be helpful to the athlete or laborer who is over the acute phase of spine pain and needs to strengthen the posterior spinous muscles.

■ There is theoretic, clinical, and experimental biomechanical evidence to support the advisability of isometric truncal exercises for patients with lumbar spine pain.

■ A supervised walking, bicycling, or swimming program can provide trunk strength, general muscle tone, and aerobic fitness.

■ Physical therapy (heat and massage) has not been shown to have any particular biomechanical effect on the spine. The therapeutic response is satisfactory for about 60–70% of patients.

■ Spine pain school relies on patient education directed toward an understanding of the problem and practical advice about how to best care for oneself. Ergonomic advice based on sound biomechanical knowledge is provided.

■ It has been shown that it is possible, with appropriately applied axial traction, to increase the separation between vertebrae and consequently enlarge the neural foramen in the cervical and lumbar region of the spine. However, the separation does not persist when the traction is released. This form of therapy gives satisfactory results in 60–70% of patients. Certain specific pathologic conditions may be expected to respond better than others to axial traction.

■ Regardless of how the forces are applied and transmitted to a vertebra, it can only move within some combination of six degrees of freedom.

- Only the magnitude of the forces and their rate of application may be altered by the therapist.
- It has been shown that axial rotatory manipulation does not alter the size of the intervertebral foramen.
- Based on current traditional anatomic, pathologic, and physiologic scientific knowledge, there is no basis for any assumption that spinal manipulation is therapeutic for visceral disease.
- Spinal manipulation is as effective as a number of other forms of therapy that offer satisfactory results in 60–70% of patients. There is a suggestion that there is an immediate but transient post-treatment benefit.
- Cervical spine manipulation can be fatal.
- Orthotic devices reduce spine motion to some extent. This has the value of reducing irritation. If the orthosis has an abdominal component, it can further reduce irritation by providing support to the spine.
- The enzyme chymopapain is capable of hydrolyzing and at least partially dissolving the intervertebral disc. Several studies report results that are significantly better than the good-to-excellent results commonly observed in 60–70% of patients with other treatment.
- The risk of using chymopapain, in addition to possible death from anaphylaxis, includes the possibility of neural damage, discitis, and arachnoiditis.
- Chymopapain injected into the nucleus pulposus may be ineffective in the treatment of a sequestered portion of disc material. The enzyme may not pass from the nucleus to the displaced, sequestered material.
- The early experience with percutaneous discectomy is promising—70–80% success and few complications. Some of its mechanism may be related to the biomechanical concept of bulk modulus. More clinical studies are needed to determine the ultimate usefulness of the modality.
- When patients are carefully selected by extensive preoperative evaluation to ascertain the presence of a herniated disc, the results are satisfactory in 90%. The results are best in patients with lumbar spine disc herniations. In the cervical spine, the results of treatment are not quite as good, and they are generally not good in the thoracic spine.
- In the surgical excision of a herniated disc, the radiculopathy tends to improve more than the spine pain.
- Once the diagnosis of herniated lumbar disc is made, the patient is generally better served by surgery at 6–8 weeks after onset of severe sciatica. The advantages and disadvantages of surgery at any given point in the course of the disease vary with the individual patient.
- Fusions may be employed on the assumption that decreasing mobility of the FSU reduces irritation of all related structures and alleviates or eliminates pain.
- It has been shown that posterior fusion reduces but does not eliminate motion.
- A *third* spine operation in the same region should thoroughly free all neural elements, fuse the involved segments when indicated, and should be the last operation.

Prophylaxis, Ergonomics

- There is indirect pathoanatomic and biomechanical evidence that the slightly flexed or straight back position of the spine is less painful and more comfortable. This is controversial, and the desirable position may be best determined for each individual patient, as is suggested by the McKenzie program.
- There is solid biomechanical evidence that shows that a backrest inclination of 120° and a 5-cm lumbar support constitute the ideal seat. Armrests, adequate thigh support, and space within which to comfortably alter position are some additional biomechanically important factors in design.
- When an object is being lifted, the distance that it is held away from the body has been shown to be the factor of major importance in the ergonomics of lifting. To greatly reduce forces on the lumbar vertebrae, objects should be held close to the body when lifting.
- The ability to generate large intratruncal pressures through the use of powerful abdominal muscles may protect the spine as well as increase the capacity to lift heavy loads. It is recommended that patients with spine pain have a program of trunk-strengthening exercises before returning to work that involves significant lifting.
- Flexing the knees and hips and lifting with the legs while maintaining a straight back are also important in the ergonomics of lifting. However, these

factors may not be as important as the two preceding considerations.

■ It is suggested that patients with acute lumbar spine pain avoid sit-ups and hyperextension exercises because these activities exert such large forces on the lumbar spine. The exception is the worker or athlete who must perform isotonic flexion or extension muscle activity.

NOTES

[A] The onset of cervical disc disease is later than that of lumbar disease.[162] Patients in the age range of 30 to 39 years are most likely to have lumbar disc surgery, as opposed to age 40 to 49 for cervical disc surgery.[178,183]

[B] It is possible that Type V disc pathology may respond to spinal manipulation. This constitutes an exception to the statement that manipulation and displacement of an FSU through a normal range of motion cannot significantly move structures in or out of the intervertebral foramen.

[C] This statement apparently conflicts with the recommendations of the Low Back School (see Fig. 6-41). In an attempt to resolve this, M. Zachrisson-Forssell was consulted. The discrepancy was acknowledged, but it was suggested that the exercises are not performed in the acute phase of spine pain and that they are very carefully taught and supervised. The technique is one of a slow curving upward, starting with the head, followed by the shoulders, upper thoracic spine, and the lumbar spine. Although there are no measurements, this is thought to be associated with much lower loads than the ones recorded in the studies conducted by Nachemson. Despite the findings of Nachemson, some physical therapists state that partial sit-ups, with hips and knees flexed, do not cause any detrimental loads or irritations. It is more generally agreed, however, that sit-ups with legs straight are to be avoided in patients with back pain. With this type of sit-up, the lumbar spine is loaded by the erector spinae, psoas, and abdominal muscles.

[D] Although canal size does not correlate with clinical symptomatology as well as cross-sectional area of the dural sac, this should not be construed to mean that the canal size in the lumbar spine is not important. Studies have shown that disc disease is more common in those with small canals. Moreover, the margin of safety can be presumed greater than all the various potential encroaching factors in lumbar spinal stenosis when there is a normal or larger than average canal.

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Spinal Braces: Functional Analysis and Clinical Applications

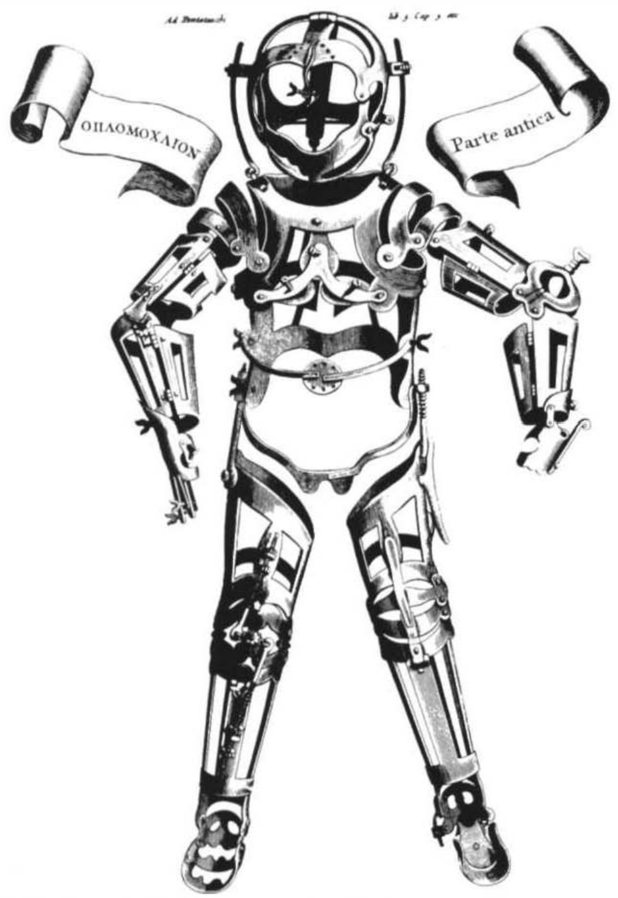

Figure 7-1. This shows how easily the skill of the armorer can be transferred to the fabrication of orthoses. With modest imagination, one can extrapolate the components of almost any brace directly from this picture. (From Orthopaedic Appliances Atlas, vol. 1. American Academy of Orthopaedic Surgeons, Ann Arbor, J. W. Edwards, 1952. Original from Heironmus Fabricus ab Aquapendente. Opera Chirurgica. Patavii, Bolzetti, 1641. Courtesy of the Armed Forces Institute of Pathology.)

7

Although progress has been made, there is still a relative paucity of research on spinal orthotics, particularly controlled clinical studies. Thus, a significant portion of this chapter is based on inferences from objective information about clinical biomechanics and physical properties of the spine. The end result is not hard science but a rational approach based on the available biomechanical knowledge and clinical studies.

The names of the various spinal orthoses are complex, confusing, and frustrating. Hopefully, this chapter will not add to the appellative turmoil but will instead alleviate it through an analytical approach that emphasizes components and functions rather than cities, states, and eponyms.

HISTORICAL BACKGROUND

This section is based largely on information from the *Orthopaedic Appliances Atlas*.[47] Corpus *Hippocraticum*, a sifting of all contemporary medical knowledge as well as a humanistic approach to medical ethics, was written by Hippocrates in the first century A.D. Two of his books, *On Fractures* and *On Articulation*, dealt with methods of treating orthopedic problems.

Galen (131–201 A.D.) was the first to employ the terms scoliosis, kyphosis, and lordosis. Because of his experiments with animal dissections, he is given credit for the first attempt at active correction of spinal deformities based on his broad concept of physiology and morbid anatomy. He advocated breathing exercises, singing, and chest strapping for the correction of scoliosis. In the Middle Ages during peacetime, when the armorers were not involved in military endeavors, their talents were displayed in brace making. Figure 7-1 readily shows the similarity between orthotic appliances and armor. It is difficult to resist the suggestion that the spirit of these armorers may have been the precursor of the contemporary spirit of bioengineers, motivated to employ their knowledge and expertise for humane rather than martial purposes. There are certain observable patterns, even in these crude, old, armorlike braces that have molded pelvic supports and spinal uprights.

Ambrose Paré (1509–1590) is considered a pioneer in the modern art of brace making. Among his inventions were metal corsets, leather walking splints, and different types of shoes for sufferers of clubfeet. In the 17th century, though orthopedics was far from a recognized specialty, supportive appliances, slings, and extension devices were being improved far beyond the advancements of previous centuries. Nicholas Andry (1704–1756), professor of medicine at the University of Paris, led many of these advances and combined the words that gave this field of medicine its name: *orthos*, meaning straight, and *paidios*, meaning child. Two of his colleagues, Lorenz Heister and Levache, made substantial contributions to the development of brace making. Heister is credited with the development of the first crude spinal brace. This apparatus was quite aptly known as the "iron cross" (Fig. 7-2). Note the "halolike" head piece and the support for the sling, both principles that are still currently employed. Levache devised a suspension brace by elongating the posterior bar of a spinal brace over the head ("jury mast") and attaching it firmly to a snug-fitting cap. This same principle, although often revised, is now used in a different manner with the halo traction apparatus.

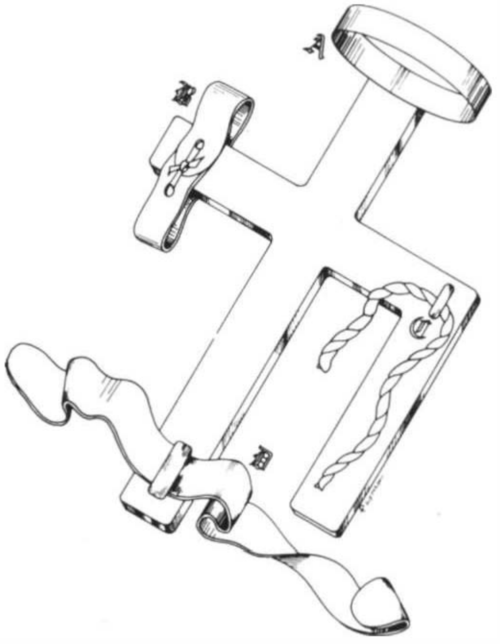

FIGURE 7-2 An interpretation of the "iron cross." Lorenz Heister, a student of Professor Andry, is credited with having developed this *first spinal brace* in the 18th century. In the art of bracing, variations are prolific and originality is precious. In this construction one can see some basic components. *(A)* A halolike structure. *(B)* An axillary sling. *(C)* Shoulder straps with splint for the upper arm. *(D)* A waist or pelvic band.

Dr. Lovett wrote the following:[47]

> . . . although the period at the middle of the 18th century . . . was one of considerable activity and progress, in the development of scoliosis, at about this time, there began and there lasted for over 100 years, the dreariest and most confusing period in the history of the affectation. The theorist and the apparatus inventor went mad, and every form of device appeared. Braces and corsets, infinitely complicated, worse than useless, appeared by the dozen. Beds, especially constructed chairs, slings, swathes, belts, levers and the like all found their advocates and theories as to causation also ran riot, but on the whole, the invention and the elaboration of apparatus held the center of the stage.

Notwithstanding Lovett's views, this period of epidemic creativity yielded several valuable principles or devices in the nonsurgical treatment of scoliosis, such as the plaster jacket, the thigh attachment, pressure pads, pelvic pads, pelvic bands, axillary supports, and the important concept of dynamic bracing.

An analysis of contemporary orthotics reveals that there have not been very many significant changes in the basic concepts and mechanisms of bracing. There have been some worthwhile refinements, however. The 19th century witnessed the emergence of modern medical science. Surgery became more established on a scientific basis rather than as a craft. Along with this change there were the corresponding refinements in the design of mechanical appliances. Since the orthopedist of the 19th century was not primarily a surgeon, his reputation grew in proportion to his ability to effectively utilize *mechanical principles.* Probably the most famous orthopedist of this period was Hugh Owen Thomas, a prolific inventor of appliances and one whose influence and simplicity of design are still present in modern orthotics. This dedicated, energetic chain-smoker is credited with the development of the frequently used and generously modified Thomas cervical collar. He established his own workshop where braces and splints were fashioned, and his work represents the beginning of the fusion between the mechanical and surgical phases of orthopedic treatment.

Another advance is credited to Anthonius Mathijsen, who used a plaster of Paris bandage in 1852 as a substitute for the cumbersome splints that had been used to immobilize limbs. Lewis Sayer (1820–1901), considered the father of orthopedic surgery in the U.S., was the first to apply a plaster of Paris jacket. It was not long before the principle of uninterrupted rest was appreciated, and plaster supplemented many techniques.

The science and technique of brace making have been largely influenced by other American surgeons who devoted much of their time and study to the improvement of existing designs. The last 60 years have seen tremendous strides in the facilities made available to orthopedic patients, and ongoing research programs continue to discover new materials and techniques that will contribute to more effective spinal bracing. During this period, the highlight has been the development of the Milwaukee brace by Schmidt and Blount.

FUNCTIONS OF SPINAL ORTHOSES

The clinical science of spinal orthotics is essentially that of the application of forces to the spine in order to control it. The goals may be any combination of the following: support, rest, immobilization, protection, and correction. The application of forces alters the existing patterns of deformation and kinematics of the spine. To rest the spine, the orthosis must substitute for or assist the actions of muscles. The rationale may be to limit the range of motion when certain positions or movements are painful to the patient. It may be desirable to protect the vital cord and nerve roots immediately following surgery or after injury. In this instance, the brace carries out a function that either the intrinsic structure of the spine or the muscles normally achieve. If inappropriate judgment is employed and erroneous assumptions are made, there is potential danger to the patient. The orthosis can sometimes function purely as a "comforter" or a psychologic reminder; when the patient moves, his brace touches or irritates him in some way, serving as a stimulus to limit that particular activity. An orthosis for the neck or back pain sufferer often has a subjectively recognizable "supportive" function that may be due to some mix of biomechanical change, heat, massage, and placebo.[39,46] Finally, there are the correctional uses of spinal orthoses, as in the treatment of scoliosis and kyphosis.

These functions have been summarized by Nachemson,[39] as presented in Table 7-1. This listing of orthotic functions is important in the interpretation of some very useful tables that follow.

TABLE 7–1 Spinal Orthosis Function

1 = Correct deformity
2 = Limit motion
3 = Stabilize
4 = Unload
5 = Miscellaneous effects (massage, heat, psychological placebo)

(From Nachemson, A. L.: Orthotic treatment for injuries and disease of the spinal column. *In* Physical Medicine and Rehabilitation: State of the Art Reviews, vol. 1, pp. 11–24. Philadelphia, Hanley and Belfus, 1987.)

BIOMECHANICAL FACTORS

Physical Characteristics of the Spine

The spine may be viewed mechanically as a series of semirigid bodies (vertebrae) separated by viscoelastic linkages (discs and ligaments). Elasticity is exemplified by a spring. If a load is applied, there is an immediate deformation. When the load is released, the spring goes back to its original position. A syringe may be used to describe viscosity. When a load is applied to the plunger, it does not return to its original position upon removal of the load. The rate of application of the load on the plunger is related directly to resistance. Viscoelasticity is a combination of viscosity and elasticity.

The clinician is working with a series of linkages suspended in the body, with viscoelastic structures of various stiffness attached to them. In the cervical spine, the vertebrae and the linkages are surrounded mostly by muscle. In the thoracic spine, they are encased by muscles, ribs, air, and the lungs. In the lumbar spine, there are the muscles, air, water, and the viscera. Skin and subcutaneous fat are involved in all three regions. The materials encasing the structures have different moduli of elasticity and viscous properties. Materials adjacent to the spinal column are all of fairly low stiffness, except for the ribs, which are significantly stiffer. The ribs stiffen the thoracic spine by forming a boxlike construction. It has been calculated by Andriacchi and colleagues that ribs increase the stiffness of the thoracic spine in bending by 200%.[2]

All of these elements may be viewed as sitting in a cylinder, the body. The goal of the orthotist is to transmit force through the cylinder to the spinal column in order to exert some control on it.

The Transmitter Problem

In the science of orthotics, the force is not applied directly to the spine but must be transmitted. Whether the goal is support, immobilization, or correction, the mechanism will depend on the transmission of forces. The major mechanical factor that limits the transmission of forces to the spine is the *stiffness* of the structures through which the forces must be transmitted. If a feather is used to push a deformity, very little force is transmitted. This is true regardless of the amount of force that is available. The feather has a low stiffness and will deform. Thus, essentially no force is transmitted. If, on the other hand, a steel rod is employed to push the deformity, it will transmit the force almost completely. The same principle holds when there is an attempt to apply forces to the spine with an orthotic appliance.

The basic biomechanical problem of spinal orthoses is one of transmitting sufficient forces to a series of vertebrae through low-stiffness, viscoelastic transmitters. The stiffness of these transmitters varies considerably; the ribs (though not especially stiff) represent the stiffest available transmitter. Fat, which has a much lower stiffness, is at the other end of the continuum. The overall biomechanical problem of spinal bracing is summarized schematically in Figure 7-3. These factors are of considerable importance to the clinician in his evaluation of the forces that can be expected to be transmitted to any particular region of the spine in order to achieve a desired therapeutic goal. It is possible to apply forces more effectively to a thoracic scoliosis than to one in the lumbar region because the ribs are better (stiffer) transmitters than the muscles and viscera of the lumbar region. It is known that the Milwaukee brace is less effective for holding or correcting lumbar curves than it is for holding or correcting thoracic curves.

Other Limiting Factors

The pain sensitivity of the skin and the deeper tissues must be considered. Also, there are biologic functions of the skin that have a limiting influence. The skin must be freed of dirt, debris, and its own excretions. It must also be ventilated. These factors limit the magnitude and the duration of pressures that may be applied. As a result, the clinician is able

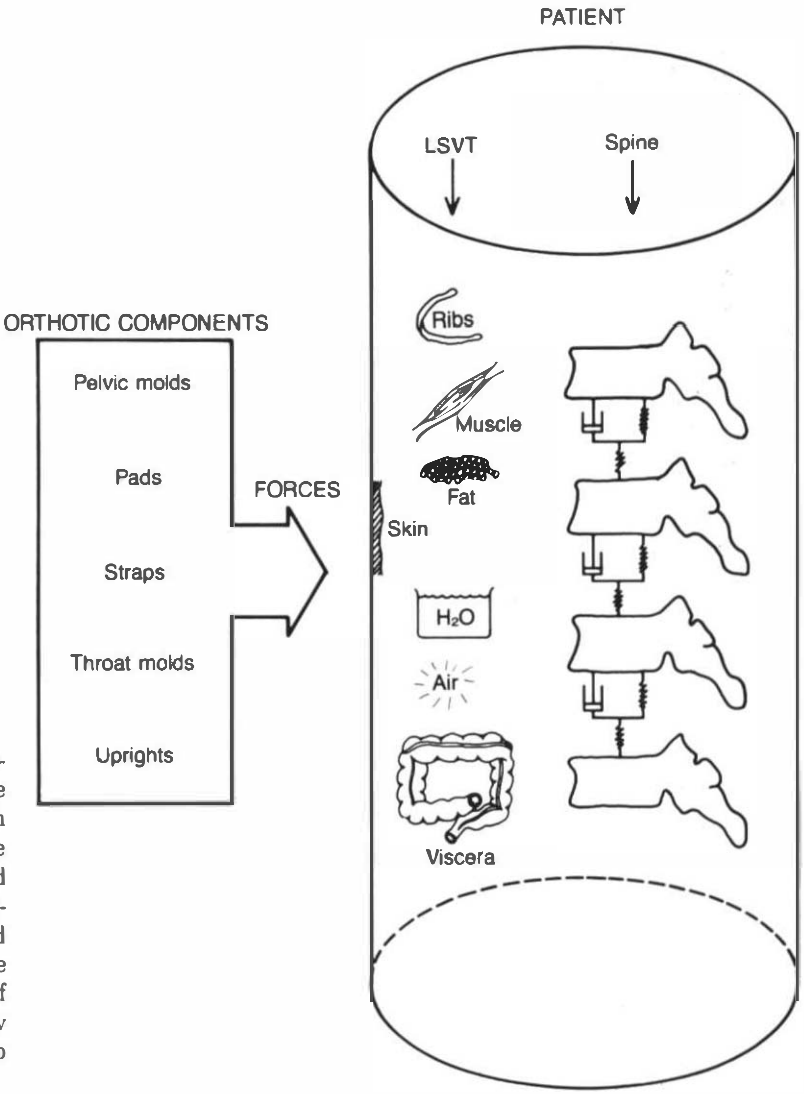

FIGURE 7-3 This diagram depicts the biomechanics of spinal orthotics. On the left are the various orthotic components through which forces are applied to the patient on the right. The anatomic components are divided into various low-stiffness viscoelastic transmitters (*LSVT*) and the spine, a series of rigid bodies separated by viscoelastic linkages. The mechanics of bracing involve the process of transmitting forces to the spine through low stiffness viscoelastic transmitters in order to exert various controls on the vertebrae.

to apply forces to the spine with most orthoses but is not actually able to control it completely.

The characteristic elasticity of bone is a limiting factor. Rolander carried out experimental cement fusions of the laminae and spinous and transverse processes of functional spinal units (FSUs) and observed motion between the vertebral bodies under loads that did not exceed the physiologic range.[50] This was due to the elasticity of the unfixed portion of the vertebra, the pedicles, and the body. Although this experiment has limited application to the spine *in vivo*, it points out the improbability of completely immobilizing an FSU with an orthosis when a direct fusion may not do so (see Chap. 8).

In a similar sense, it is important to be aware of the limitations of a brace in protecting an osteoporotic vertebra. (The weakened vertebral body in osteoporosis is discussed in Chapter 1 on p. 40). It should be recognized that a brace will be significantly limited in its ability to compensate for the very large loss of supporting elements in the osteoporotic vertebra.

The Normal Kinematics

An analytical approach to the biomechanics of orthotics involves some additional considerations. It is worthwhile for the clinician to keep in mind the normal kinematics of the spine as bracing problems are approached. The cervical spine is the most mobile. There is generally more flexion than extension. Most of the motion is in the sagittal plane, and that usually occurs at C5–C6. The C1–C2 joint, however, has the greatest axial rotation in this region. There is strong coupling of axial rotation and lateral bending in the lower cervical spine, and there is a generous amount of axial rotation in this region. In the thoracic spine, there is significant motion, but it is certainly less mobile than either adjacent region. Here, too, flexion is greater than extension. The amount of rotation in the sagittal plane progressively increases cephalocaudally. The coupling has the same characteristics as that in the cervical spine, but it is not as strong or as consistent. There can be a transition to the kinematic pattern of the lumbar spine anywhere between T9 and T12. There is also an axial rotation in the thoracic region that decreases cephalocaudally. In the lumbar spine, there is little axial rotation. Most of the motion is in flexion/extension (x-axis rotation). At the lumbosacral joint, there is relatively more rotation. Also, any pelvic motion will move the lumbosacral spine.

For the most complete evaluation of the controls that a clinician can apply to the spine with a brace, two factors should be considered. First, there should be a consideration of the characteristic regional kinematics involved. Then an analysis with respect to the six degrees of freedom should be carried out (see Chap. 2). This includes evaluation of probable translation along each of the three coordinates and rotation about each of the three axes. The clinician decides which movements he wants to restrict and then selects the appropriate orthosis to achieve that control.

The clinician is usually most concerned with only two or three of these six degrees of freedom. However, an awareness and an analysis of all six are desirable. Consider a patient with low back pain. If an orthosis that effectively discourages flexion and extension (rotation about the x-axis) but allows considerable axial rotation (rotation about the y-axis) is prescribed, this may cause difficulty. If the patient has a significant synovitis of the facet joints, the axial rotation may irritate the joints and elicit significant pain, despite the fact that they remain reasonably well protected in flexion and extension.

When orthotics are employed to compensate for instability, a basic understanding of clinical instability is of value. It is necessary to consider which structures have been rendered nonfunctional so that appropriate support may be instituted. Spines that are unstable as a result of the loss of the functional integrity of the anterior elements are more unstable in extension. Spines unstable because of disruption of the posterior elements are more unstable in flexion (see Chap. 5). Certain orthoses protect better against anterior displacement, and others protect better against posterior displacement. Here again, all six degrees of freedom should be considered, and decisions concerning the type of motion to be controlled are necessary. Attention is then given to the question of how rigid the fixation should be. The clinician must be certain that it is possible to compensate for the instability with an orthotic device and that the device selected is most appropriate for the particular instability under treatment.

Creep and Biomechanical Adaptation

The creep phenomenon is based on the characteristic of viscoelasticity. It manifests itself in the form of additional deformation over a period of time that may vary from several seconds to several minutes. After several weeks, biomechanical adaptation comes into play. *Biomechanical adaptation* may be defined as biologically mediated changes in mechanical properties of tissues (material properties and/or structural changes) in association with the application of mechanical variables to the tissues. For example, if the hardness of skin under the pelvic band of a Milwaukee brace was measured after the first and the 99th day, the values would be different. The change would be due to biomechanical adaptation.

In long-range responses to forces, there are differences in the configuration of ligaments and bone. The so-called giraffe-necked women of the Padang tribe of Indonesia demonstrate biomechanical adaptation (Fig. 7-4). A radiograph of such a person shows that the shoulders are pushed caudad; however, there is an intrinsic loss of physiologic cervical lordosis and some exaggerated elongation and separation of vertebral bodies. Another example of adaptation to long-range forces is the change seen in Scheuermann's disease with Milwaukee brace treatment (see Fig. 7-21). Here, the actual configuration of the spine changes. The alterations are more readily mediated in the growing skeleton, where Heuter Volkmann's laws can operate through the epiphysis. In early gradual correction and also in long-range

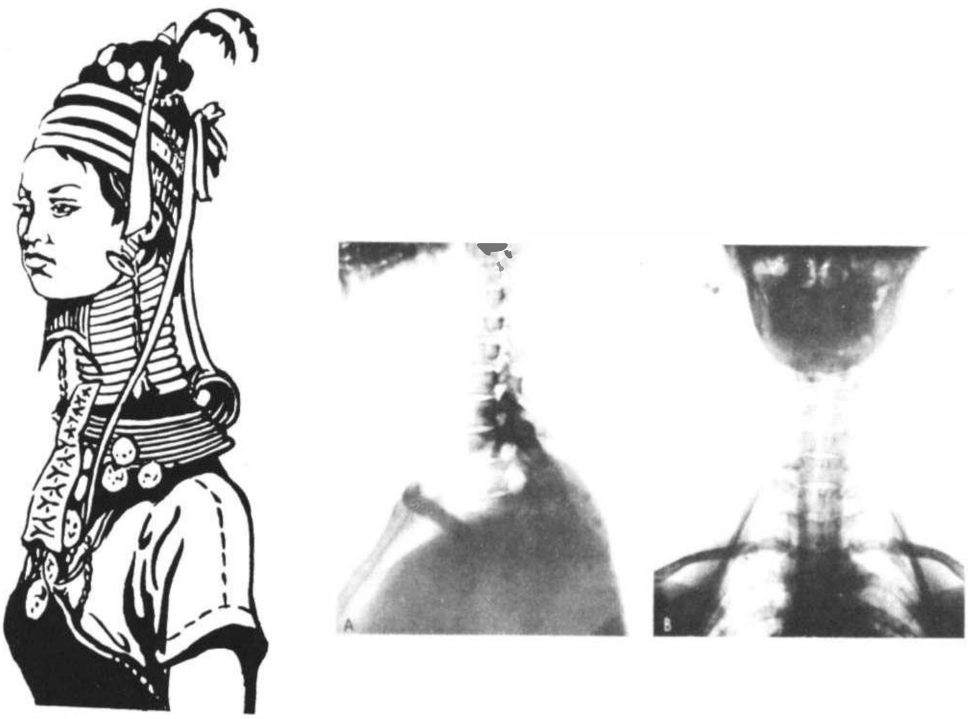

FIGURE 7-4 This is a demonstration of *biochemical adaptation*. In the Padang tribes of Indonesia, rings are placed above the neck, added in gradual increments, and left over a long period of time. The shoulders are pushed down, which adds to the appearance of length. Biomechanical adaptation is also shown by the loss of normal cervical lordosis, elongation of vertebral bodies, and more than the usual separation between vertebrae. It has been reported that an old tradition involved removal of the rings if the bride was unfaithful. This resulted in paralysis and death due to the loss of intrinsic clinical stability. *(Roaf, R.: Scoliosis. Edinburgh, Churchill Livingstone, Ltd., 1966.)*

biologic adaptation, time is an important consideration in the use of spinal orthotics.

Five Mechanical Principles of Spinal Orthoses

Balanced Horizontal Forces

Horizontal forces are eminently suitable for providing efficient bending moments for the correction of lateral curvature, derotation of vertebrae, and immobilization of the spine. Most of the loading situations in braces can be shown to be mediated through a three-point loading system; this is analyzed in some detail below.

Three horizontal forces are applied at points along the length of the spine. Two are in one direction, and one is in the opposite direction (Fig. 7-5A). There are some fundamental characteristics of this force system. Since the system is in equilibrium, the sum of the forces and the sum of the bending moments they create must be equal to zero. Therefore, the site of application of the forces and their magnitudes are interrelated.

In a general case shown in Figure 7-5, the forces at points B and C have to be in inverse proportion to their perpendicular distances, D_B and D_C, from point A. Furthermore, the sum of the forces at points B and C must always be equal to the force at point A. Thus, with the perpendicular distance D_C being twice that of D_B, the magnitudes of forces at points A, B, and C must be in the ratio of 3 : 2 : 1.[A]

This information has clinical relevance. In order to adjust the skin pressure at the three force points, the pad sizes must be proportioned according to the force magnitudes. In the case of the Jewett brace, shown in Figure 7-6, with two anterior pads placed at an equal distance from the posterior pad, the force at the posterior pad is twice that of the anterior pads

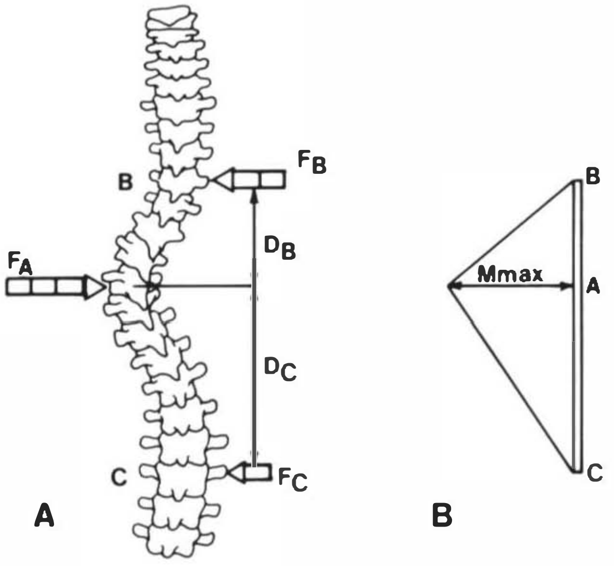

FIGURE 7-5 (**A**) A three-point force system is formed by forces F_A, F_B, and F_C. Relative vertical (y-axis) distances (i.e., D_B and D_C) dictate the relative magnitudes of the three forces. (The lengths of the arrows correlate with the relative forces.) (**B**) The bending moment diagram for the three-point force system is a triangle with its apex at the level of the middle force. This implies that the maximum corrective potential or control is thus at a point on the spine at the level of the application of force F_A.

and therefore should have twice the pad area in order to have the same skin pressure.

Another important characteristic of this three-point force system is the bending moment applied to the spine. Actually, it is the bending moment and not the forces that produces the angular correction. The bending moment at the various intervertebral spaces varies. It is maximum at the level of the middle force F_A and linearly decreases to zero at the level of the two end forces, F_B and F_C (Fig. 7-5B). By placing the middle force at the apex of the curve, the clinician maximizes the correctional efficiency of this force. It can be further shown that forces F_B and F_C should be located as far away from F_A as possible.[A]

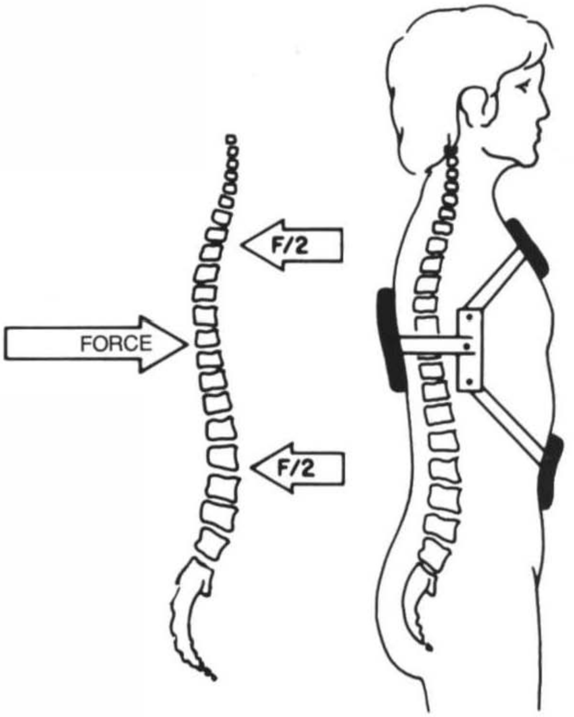

FIGURE 7-6 The Jewett brace functions on a three-point force system, but on a different plane than that depicted in Figure 7-5. As is shown, the posterior pad has been placed midway between the two anterior pads. Therefore, each of the anterior forces is one-half of the posterior force. The greatest potential for correction is at the level of the posterior force.

Fluid Compression

It is possible to use soft tissues (muscles, fascia, and tendons) to support a compressive load. Nature has used the diaphragm and abdominal muscles to compress the contents of the trunk cavity. Thus, the turgor of fluid under pressure is employed to support or splint the spine. The orthotist makes valuable use of this concept by applying compression externally, through the use of a corset or an abdominal support, either attached to an appliance or worn alone. A Williams brace (thoracolumbosacral orthosis [TLSO]) is one that uses this corset effect anteriorly on the abdomen to achieve additional support for the spine. (See Table 7-2 for orthotic nomenclature.) This technique is especially effective in resting and unloading the lumbar spine. This mechanism was no doubt operative when fashions were different and the woman would note that her back felt better when she "had on her foundation."

Distraction

By the application of tension through distraction, it is possible to achieve a certain amount of immobilization and stability of the spine. The value of distraction as a form of fixation can be readily appreciated with an ordinary sheet of paper held vertically and stretched between the two hands. The paper becomes rigid because of this tension and can resist a lateral force. Without tension applied to the two ends, the paper has negligible lateral stability.

Sleeve Principle

This essentially involves the construction of a cage around the patient. There are basically two semicircular fixation points, one above the other. Then, between the two, there are various uprights. The uprights may be at the sides of the patient, or they may be posterior and paraspinous. These uprights serve as a sleeve, a splint, a distractor, and as a point for attachments of various accessory devices, such as localizer pads, axillary slings, or abdominal pads.

Skeletal Fixation

This is another useful orthotic technique. The prime and perhaps only examples are the halo fixation and the halo pelvic fixation devices. These appliances provide the most effective methods of applying reliable controls to the spine.

CLINICAL REVIEW OF SPINE REGIONS AND THEIR SPECIFIC ORTHOSES

After the clinician makes a diagnosis, he determines the specific mechanical goals that are to be achieved—whether to support (rest, assist), immobilize (protect), or correct the spine. An analysis of the six degrees of freedom in which the involved vertebra or vertebrae can move is carried out. The clinician then determines which degrees of freedom are to be controlled as well as the manner and extent to which they are to be altered.

When these determinations are made, the orthosis that is best able to achieve these goals can be selected. The authors have not chosen to present this section of material as a catalog of diseases and braces recommended for treatment. The major types of orthoses are reviewed. In an attempt to systematize, these arbitrary groupings have been chosen, based on the effectiveness of control applied by the orthosis: minimum *control* (least effective); *intermediate control* (a broad range with some effectiveness); most *effective control* (the best in thegroup). A more precise classification is desirable; however, present knowledge dictates this somewhat arbitrary classification. The list above outlines a systematic clinical analysis of the biomechanics involved in the selection of a spinal orthosis.

SYSTEMATIC ANALYSIS FOR THE SELECTION OF ORTHOSES

- *Determine the goal of orthosis:*
  - Support (rest, assist)
  - Immobilization (protection)
  - Correction
  - Reminder
- *Determine how many degrees of freedom are to be constrained:*
  - Flexion
  - Extension
  - Lateral bending
  - Axial rotation
  - Axial distraction
  - Anterior translation (z-axis)
  - Lateral translation (x-axis)
- *Determine the magnitude of control:*
  - Minimum
  - Intermediate
  - Most effective

Cervical Region

Experimental Studies

An *in vivo* study by Hartmann and colleagues has provided some relevant guidelines about the effectiveness of immobilization by cervical spine or-

TABLE 7–2 Common Abbreviations for Spinal Orthoses

Abbreviation	Area Included	Mainly Used for Disorder in
CO	Cervical orthosis	Neck
CTO	Cervicothoracic orthosis	Neck
CTLSO	Cervicothoracolumbosacral orthosis	Thoracic & lumbar spine
TLSO	Thoracolumbosacral orthosis	Lower thoracic & lumbar spine
LSO	Lumbosacral orthosis	Lower lumbar spine
SIO	Sacroiliacal orthosis	Lumbosacral areas

(From Nachemson, A. L.: Orthotic treatment for injuries and disease of the spinal column. In Physical Medicine and Rehabilitation: State of the Art Reviews, vol. 1, pp. 11–24, 1987.)

thoses.[19] These investigations evaluate normal motion and motion in five different orthoses by the use of moving pictures and cineradiography. Their findings are shown in Table 7-3. The most difficult motion to restrain was rotation that occurred between C0 and C2. Therefore, an effective cervical orthosis must fix the head directly or hold the occiput and mandible through effective molding.

It is useful to keep in mind that these studies were conducted with *normals*. In view of other observations of patients with rheumatoid disease or cervical fractures, we must recognize that the immobilizing efficiency of these orthoses may be less than the studies on normals would suggest.[1,24,57]

An evaluation of cervical braces by Johnson gives some additional information about their immobilizing efficiency. He placed normal subjects in four different orthoses and took photographs and radiographs of their cervical spines in full flexion and extension. The total motion between C1 and C7 was studied. Angles were drawn as shown in Figure 7-7. The findings are presented in Table 7-4.

In a quantitative study of cervical orthoses by Johnson and colleagues, there were several cogent findings.[21] Increasing the length and rigidity of a cervical orthosis generally improved the effectiveness of its control of motion. There was not much effective control of lateral bending or axial rotation of the cervical spine by the conventional orthosis. The most effective conventional braces were able to restrict C1–C2 flexion/extension by only 45% of normal. The halo apparatus restricted the same motion by 75%. The major quantitative findings from this study are shown in Table 7-5.

TABLE 7–4 Efficiency of Cervical Braces in Immobilization

Orthoses	Type	Total Movement from Full Flexion to Full Extension (degrees)
Soft cervical collar	CO	101
Hard plastic collar (Thomas)	CO	58
Four-poster cervical	CTO	25
Duke (occipital, chin, and chest piece)	CTO	2

The median normal is approximately 90°.[29]
CO = cervical orthosis; *CTO* = cervicothoracic orthosis.
(Johnson, R. M. et al.: Cervical orthoses. A study comparing their effectiveness in restricting cervical motion in normal subjects. J. Bone Joint Surg., 59*A*:332, 1977.)

Several generalizations may be made from these studies. The soft collar does little in the way of immobilization. The efficiency of fixation at the chin and occiput are major elements in the design of a cervical orthosis. For the most satisfactory immobilization in this group, the use of some type of shoulder and thoracic fixation and support should be added to the cervical and chin occipital components of the brace. When the chest support not only rests upon but is fixed to the thorax, the immobilizing efficiency is even greater.

Minimum Control

Since the time of Sir Thomas, collars have been popular for the treatment of a variety of problems in the cervical spine. They vary in height and in rigidity. They may be altered or worn so as to limit flexion or extension, either one relatively more than the other. If the high portion of the collar is worn anteriorly, there is relatively less flexion. If the high portion is worn posteriorly, there is relatively less

TABLE 7–3 Effectiveness of Cervical Spine Orthoses in Immobilization

	Approximate % Restriction of Range of Motion C1–C7					
	Motion Picture			Cineradiograph		
Orthoses	FE	LB	AR	FE	LB	AR
Soft cervical collar	5–10	5–10	0	0	0	0
Hard plastic collar (Thomas)	75	75	50	75	75	50
Four-poster cervical	80–85	80–85	60	85	85	60
Long two-poster	95	90	90	90	90	90
Guilford two-poster	90–95	90–95	90–95	90	90–95	90
Halo device		Essentially no motion				

(Data from Hartmann, J. T., Palumbo, F., and Hill, B. J.: Cineradiography of the braced normal cervical spine. Clin. Orthop., 109:97, 1975.)
FE = Flexion/extension (x-axis rotation); *LB* = Lateral bending (z-axis rotation); *AR* = Axial rotation (y-axis rotation).

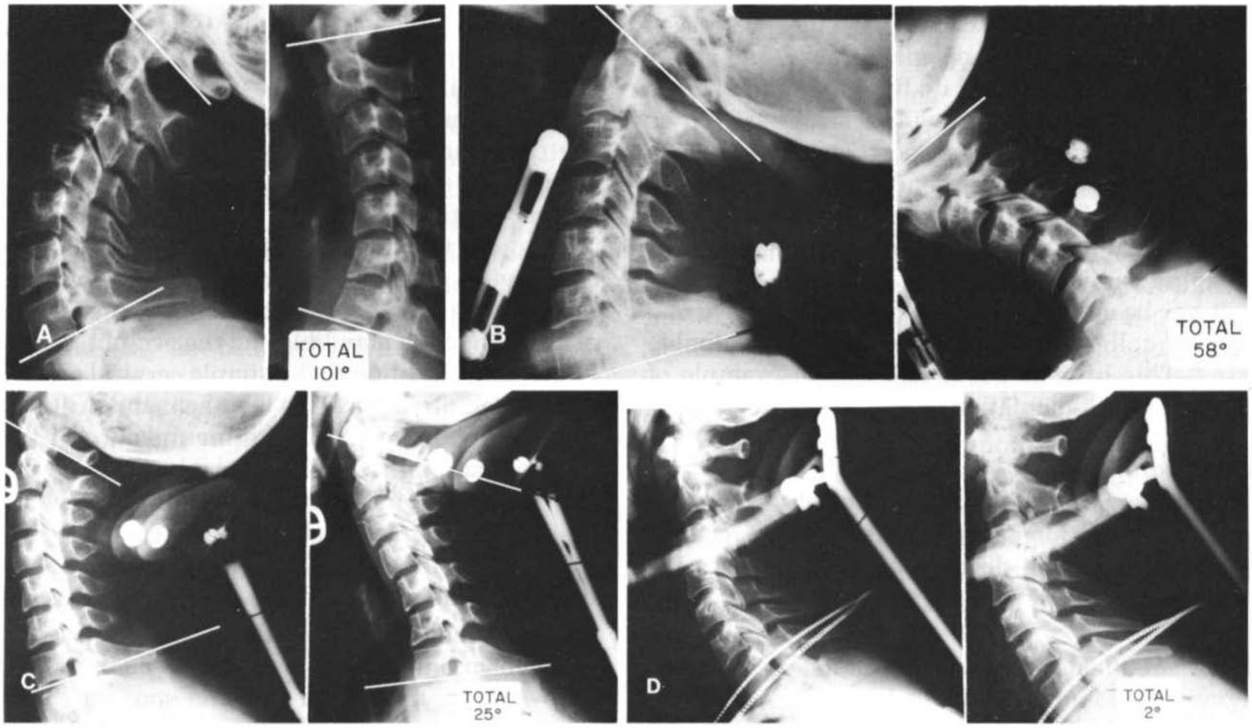

FIGURE 7-7 The actual radiographs of the extension/flexion ranges of motion achieved by normal subjects. The range of motion at C1–C7 achieved by these subjects is shown for the different cervical orthoses. **(A)** Soft cervical collar. **(B)** Hard plastic collar. **(C)** Four-poster cervical orthosis. **(D)** Duke orthosis with occipital chin and chest piece. *(Courtesy of R. M. Johnson, M.D.)*

TABLE 7–5 Rigid Conventional Braces that Provide the Best Control of Flexion and Extension at Different Levels of the Cervical Spine

	Flexion/Extension		**Flexion**		**Extension**	
Segmental Levels	Brace	Mean Motion Allowed (degrees)	Brace	Mean Motion Allowed (degrees)	Brace	Mean Motion Allowed (degrees)
C1–C2	(Halo)	3.4	SOMI	2.7	Cervicothoracic	2.5
C2–C3	(Halo)	2.4	SOMI	0.9	Four-poster	2.0
	Four-poster	3.7	Four-poster	1.6	Cervicothoracic	2.1
	Cervicothoracic	3.8	Cervicothoracic	1.8		
Middle	Cervicothoracic	4.6	SOMI	1.7	Cervicothoracic	1.8
(C3–C5)			Four-poster	2.0		
			Cervicothoracic	2.8		
Lower	Cervicothoracic	4.0	Cervicothoracic	1.5	Cervicothoracic	2.5
(C5–T1)			SOMI	2.9	Four-poster	2.5

(Johnson, R. M., et al.: Cervical orthoses. A study comparing their effectiveness in restricting cervical motion in normal subjects. J. Bone Joint Surg., 59A:332, 1977.)

extension. The cervical collars have the advantages of being inexpensive, convenient to use, and easily fabricated. Although they do little to immobilize or unload the spine, they provide warmth as well as psychologic comfort and support. These devices are useful in a broad variety of conditions, such as minor sprains and strains, some whiplash cases, cervical spondylosis, and postoperative management when the spine is clinically stable.

It has been observed that the long-term use of a cervical collar can prevent or reduce a "double chin." This little cosmetic aside is an example of biomechanical adaptation.

Intermediate Control

There are a variety of modifications of the cervical collar. Within this range of intermediate control there are different degrees of effectiveness. For slightly more restriction, a beefed-up cervical collar, such as the Philadelphia collar CTO (cervicothoracic orthosis), may be employed (Fig. 7-8). Through its rigidity, its sternal and base of the neck support, and its anterior and posterior reinforcement under the chin and the occiput, it is able to offer better restriction, especially in flexion and extension (x-axis rotation).

In order to achieve greater degrees of control of the cervical spine, it is necessary to have some purchase on the shoulders and the thoracic cage, as well as fixation of the mandible and the occiput. This, in effect, is an addition to the simple cervical collar in the caudad direction. This lengthens the sleeve and provides more effective anchoring and purchase. For example, suppose a patient has had an elective anterior cervical spine fusion at two levels. There is essentially adequate stability, but there is a need in the early postoperative period to prevent excessive cervical spine motion. In this situation, a well-fitted

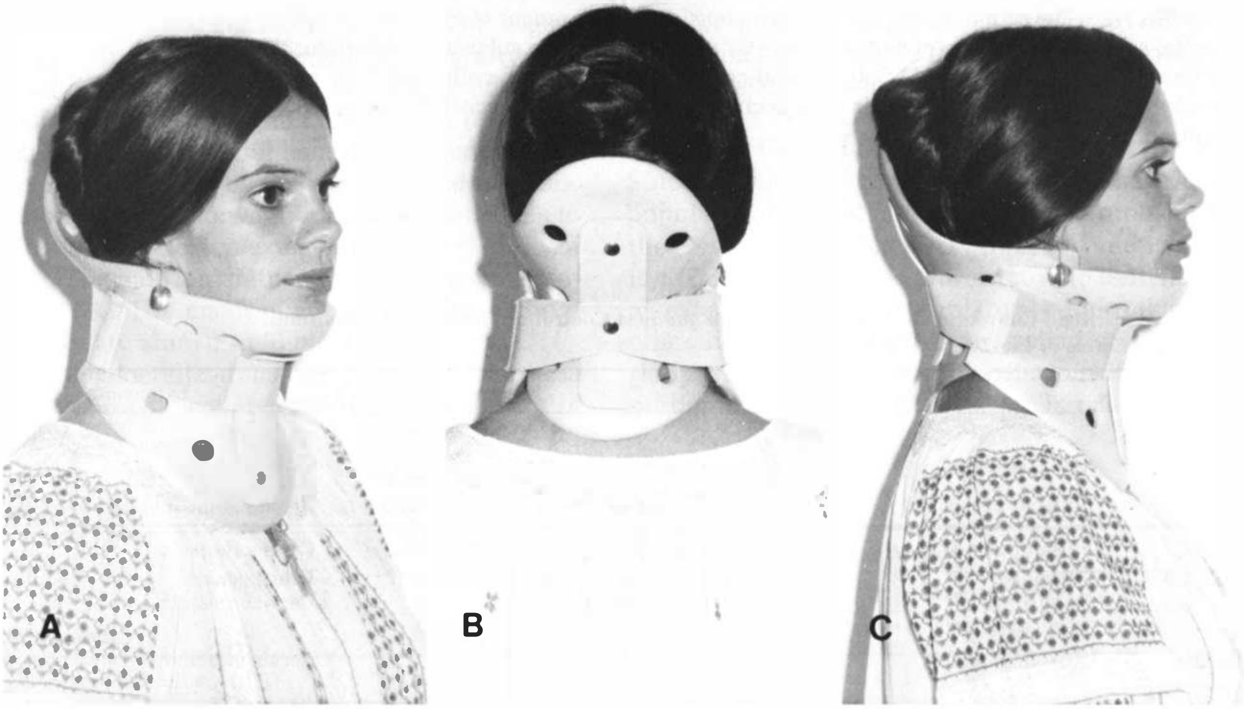

FIGURE 7-8 The Philadelphia collar in three views: **(A)** anterolateral; **(B)** posterior; **(C)** lateral. The device has some purchase on the upper chest and back but no fixation there. It is still enough to allow some distraction. Note the anterior and posterior splints. There is some degree of support and fixation of the mandible and occiput, which should reduce axial rotation. This orthosis has been classified as exerting controls in the intermediate range.

cervical brace with shoulder or chest and shoulder fixation would be satisfactory. There are several braces that are in this category: the four-poster brace (Fig. 7-9A), the Duke brace, the Guilford brace, and others (Fig. 7-9B). It should be kept in mind that shoulder-supported fixation and unloading are most valuable and necessary in the erect position. However, in the supine position or any situation in which there is considerable rotation of the shoulders and spine, the directions of the forces are altered and the appliance may not be as effective.

Most Effective Control

If the clinician determines that there has been a significant loss of stability through destruction or removal of supporting structures in the cervical spine, then the maximum amount of immobilization and unloading is desirable. Major control is needed in all six degrees of freedom.

For more effective fixation, the Thomas collar may be extended in both directions and made more rigid. Thus, one employs the Minerva cast, which includes the forehead, goes high upon the occiput, and extends all the way to the pelvis.

This device is appealing, if for no reason other than its glorious and powerful appellation from the highest echelons of Roman mythology. Minerva was born by popping from the head of Jupiter fully armored. (Knowing how she was born, it is challenging to speculate about how she may have been conceived.) This cast, which constitutes a sizable portion of armor, encases the head, shoulders, thorax, abdomen, and pelvis. This device does indeed offer considerable control and is especially useful for protection of an irresponsible patient. It should be kept in mind, however, that a few degrees of cervical spine motion are present even in the carefully applied Minerva jacket.[51] The limitations of force that may be applied depend somewhat on the ease with which talking and eating is to be permitted. Opening the mouth requires space for either the mandible to move caudad or the head to extend. Either of these motions allows a displacement of the occiput and thus some motion at C1–C2.

Although not as effective as the halo, the SOMI (Sternal Occipital Mandibular Immobilizer) cervical orthosis has been shown to be an excellent immobilizer of the upper middle and lower cervical

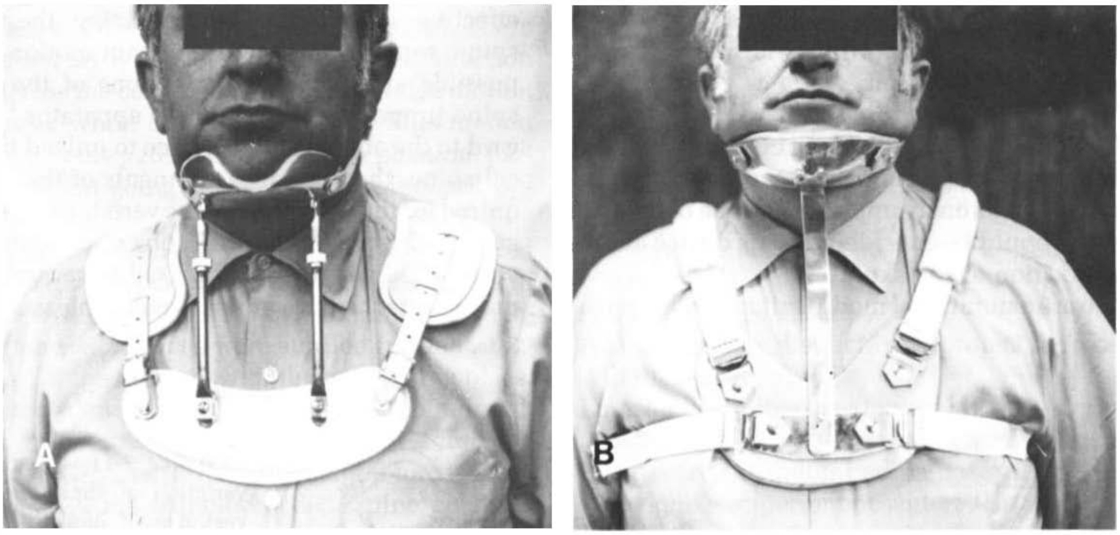

FIGURE 7-9 **(A)** The four-poster cervical brace. This provides adequate mandibular and occiput fixation. There is purchase on the thorax anteriorly and posteriorly. The four posts may be adjusted to alter distraction. The anterior and posterior pairs may be adjusted to determine the amount of flexion and extension. **(B)** The Guilford, Duke, or long cervical brace prototype. There is fixation of mandible and occiput as well as straps to anchor the purchase on the chest. These devices have both been classified as exerting intermediate control. The long cervical braces with thoracic anchoring have been shown to be most effective in the intermediate range. *(Courtesy of J. T. Hartmann, M.D.)*

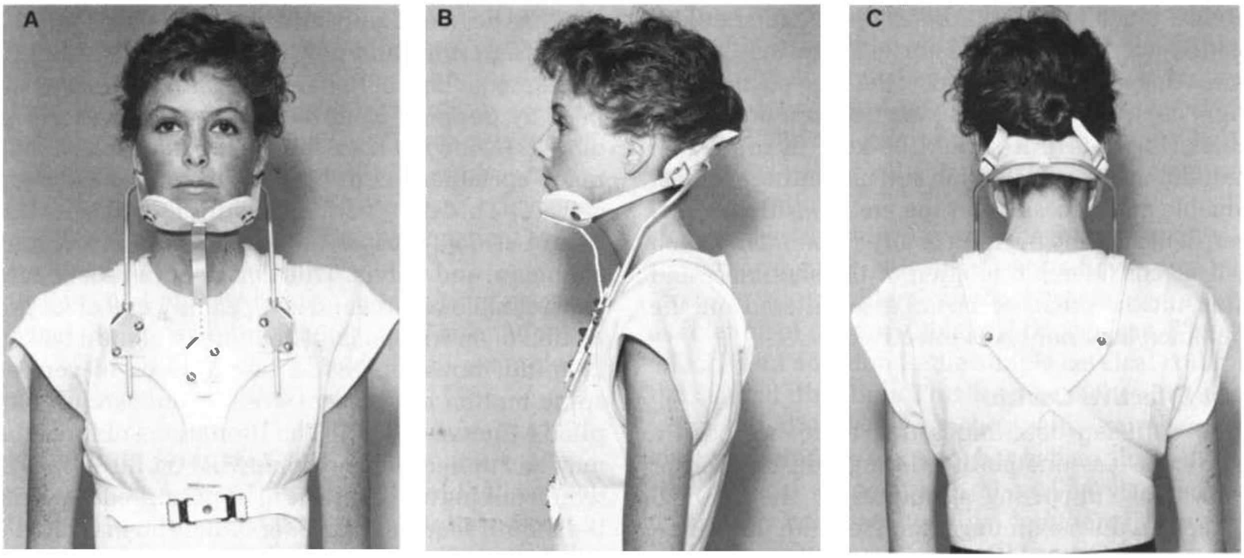

FIGURE 7-10 The SOMI is a moderately effective immobilizer of the cervical spine. It has the advantage that it can be applied and used with the patient in the supine position.

spine.[21] This orthosis, shown in Figure 7-10, has no posterior uprights and a flat back, which makes it well tolerated in supine bed rest. It also has alternate forms of fixation of the head. A chin cup can be interchanged with a headband to rest the chin or facilitate chewing. This orthosis is most effective with a cooperative patient.

The Halo Device In precarious clinical situations where extensive disease or surgery renders the cervical spine dangerously unstable, the use of the halo apparatus should be considered. This device allows skeletal fixation at the skull.

There are a number of modifications of this apparatus. Two of them are illustrated in Figure 7-11. The Vermont modification is a relatively new development with some experimental data to support its rationale.[25a] Even though the halo device is the most effective orthosis for immobilizing the cervical spine, some potentially significant motion remains possible at the lowermost regions of the cervical spine properly fixed in a halo apparatus.[1,57] In regard to the ability of this device to unload the cervical spine, there are measurements of the forces required for the detraction of the vertebrae.[7] Detraction can be observed on radiographs of patients in halo vests.[46] This device may be used in several ways. It may be attached to a molded removable waist-length

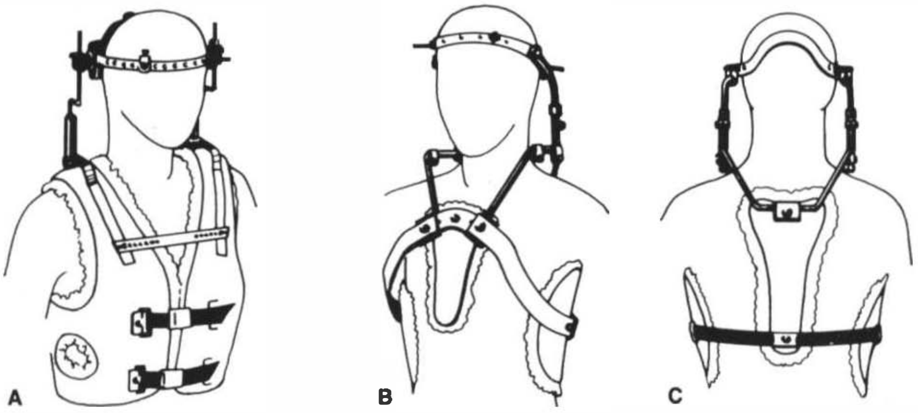

FIGURE 7-11 A shows one variation of the standard halo vest. B and C depict two views of the modified halo jacket (Vermont halo device). This device is lighter and is proportioned to provide better anchoring on the thoracic cage. *(From Krag, M. H., and Beynnon, B. D.: A new halo-vest: rationale, design and biomechanical comparison to standard halo-vest designs. Spine, 13:228, 1988.)*

jacket, a plaster waist-length jacket, a plaster jacket molded about the pelvis, or a pelvic hoop. All of these offer good fixation and are listed in order of increasing effectiveness. The combination to be employed depends on the problem, the involved region of the spine, and the clinician's preference and judgment.

Reports in the surgical literature show that the halo device is used effectively in a variety of cervical spine injuries. Chan and colleagues successfully treated unstable cervical fractures with 12 weeks of halo fixation after 10 days of cervical traction.[6] The group reported an 89% success rate with a few insignificant complications in these unstable fractures. Cooper and associates reported a success rate of 85% in a similar group of patients treated for 12 weeks with the halo.[8]

The halo may also be useful in other circumstances, such as the treatment of an athetoid patient or to stabilize a patient with cervical or upper thoracic spine instability[38] or impending instability due to metastatic disease. The halo may satisfactorily preserve neurologic function while radiation or chemotherapy is instituted.[9]

It is also important to be aware of the research documentation of some variation in the immobilizing capacity of the halo vest and the halo cast. There is variation in both compression and distraction *forces* across the cervical spine. There is also *motion* in the neck while in the halo device. This motion occurs when the patient changes body position, particularly when going between the supine and upright positions.[24]

Lind and associates[26a] prospectively studied forces and motion in a group of 31 patients treated in the halo vest for unstable cervical spines. Distraction/compression forces were studied in the last 20 patients. On the average, with the halo vest in place, the sagittal plane motion was about 70% of normal. Motion was restricted the most below C2 and the least above C2. There was a distraction force across the neck with the patients in the supine position. With various exercises, types of rehabilitation, and activities of daily living, there was considerable variation in the distraction/compression forces in the neck. It seems evident that there is not a great deal of unlocking or immobilization of the cervical spine with the halo apparatus, even though it suffices in the management of several clinical problems.

There are complications associated with the use of the halo device. These include penetration of the skull, brain abscess,[55] and abducens, glossopharyngeal, and facial nerve palsies.

Numerous other complications have been reported with the use of halo and halo pelvic treatments. The halo pelvic device may have up to 53% significant complications in a 5-year follow-up.[11] The complications are listed here and are documented in the references.[10,11,16,20,37,54]

Complications Associated with Use of Halo Apparatus

Perforation of the skull
Abscess of brain (with or without perforation)
Degenerative changes of facet joints (from distraction immobilization)
Avascular necrosis of dens
Loss of alignment of fractured cervical spine
Depressed adherent pin tract scars
Decubitus in elderly patients with sensory deficits
Abducens, glossopharyngeal, and facial nerve palsies
Problems with personal hygiene
Limitation of social life
Vascular compression of duodenum
Pain with spontaneous fusion
Dysphagia
Pain and discomfort

Reduction of Complications Associated with Use of Halo Apparatus

Evaluate carefully post–halo application headaches.
CT scan the skull for post–halo application headaches.[20]
Ascertain brain abscess.
Avoid excessive distraction to protect the facet articulations.
Look for decubitus, particularly in the elderly and in patients with impaired sensations.
Be attentive to abdominal pain or vomiting so as to recognize superior mesenteric artery syndrome.
Follow the placement of pins as recommended by Garfin and colleagues, based on studies of the osteology of the skull.[15,16] The ideal placement is anterolaterally above the orbital rim and posterolaterally below the greatest diameter of the skull. This avoids the thinner frontal sinuses and the temporal fossa and prevents piercing of the temporalis muscle (Figs. 7-12 and 7-13).
For halo pelvic device application, the open technique for insertion into the ilia is recommended.[25]

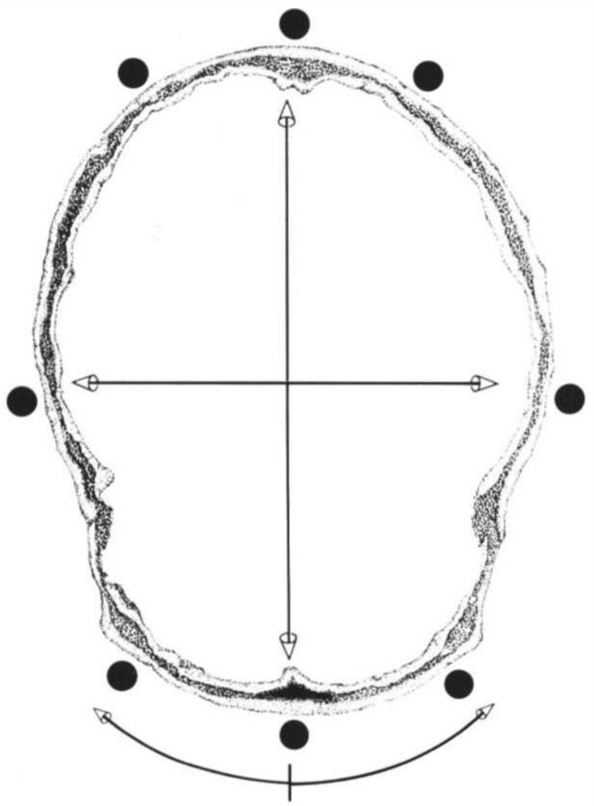

FIGURE 7-12 The study of the osteology of the cranium provides useful information about the thickness of the skull bones in various regions. The bone is thickest in the anterior lateral and posterior lateral positions. At these points below, the largest circumference of the horizontal plane of the cranium (x,z plane) is the ideal location of the fixation pins. *(Reproduced with permission from Garfin, et al: Osteology of the skull as it affects halo place. Spine, 10:696, 1985)*

A significant practical clinical contribution has come from the work of Botte and co-workers.[4] A well-designed clinical study showed that if the halo fixation pins were put at 0.90 Nm (8 in.lbf) of torque instead of the customary 0.68 Nm (6 in.lbf), the following advantages were noted. Pin loosening went from 36% down to 7%, and pin tract infections went from 20% down to 2%.

There are numerous modifications of the halo apparatus. Figure 7-11A shows the more standard type, which involves many variations on the basic theme. Figures 7-11B and 7-11C depict a more recent development of the halo apparatus. The major biomechanical advantage appears to be the fact that the apparatus, by resting on the thoracic cage and not the shoulders, is less mobile and more stable.[25a]

When a patient is in large magnitudes of skeletal traction, applying axial loads to the spine on daily rounds should include requests for the patient to smile, to roll the eyes, and to stick out his tongue at the doctor. If the patient is not able to do any of these three activities, then careful neurologic evaluation is indicated (see Chap. 8).

Nachemson[39] reviewed the literature and provided an excellent summary of cervical orthoses. Table 7-6 gives clinical indications, suggested cervical orthoses, and the functional rationale and level of significance of documentation of the treatment.

Thoracic Region

Minimum Control

In this grouping there are the long thoracic corsets. Some of these orthoses when well fitted can offer significant immobilization. However, they do not serve as effectively as some of the other appliances. Probably their main indication would be for chronic, benign, thoracic pain, in which the orthosis gives good "symptomatic" relief. In addition to the increased warmth and massage provided by a corset, there may be a distinct placebo effect.[46] This is fine if it helps the patient.[31]

Intermediate Control

The hyperextension brace is sometimes referred to as the Jewett or Griswold brace (Fig. 7-6). It is designed for resistance of motion primarily in flexion. This brace has fixation points at the manubriosternal area and at the pubis, and it employs counterpressure between these two pads from a posterior to an anterior direction, thus achieving a three-point fixation. The advantage of this brace is that there is the possibility of adjusting the levels at which maximum fixation can occur. Thus, it is possible to better immobilize a particular region of the thoracolumbar spine. This brace is less effective in restricting rotation in the coronal plane (about the z-axis), and there is virtually no resistance to axial rotation (y-axis rotation). When this brace can be adjusted to obtain some degree of hyperextension, it is reasonable to assume that it is capable of shifting the weight-bearing axis more toward the posterior elements of the vertebra. Thus, it may be helpful in diminishing stresses on the vertebral body and the anterior elements. Most probably, this orthosis cannot be relied

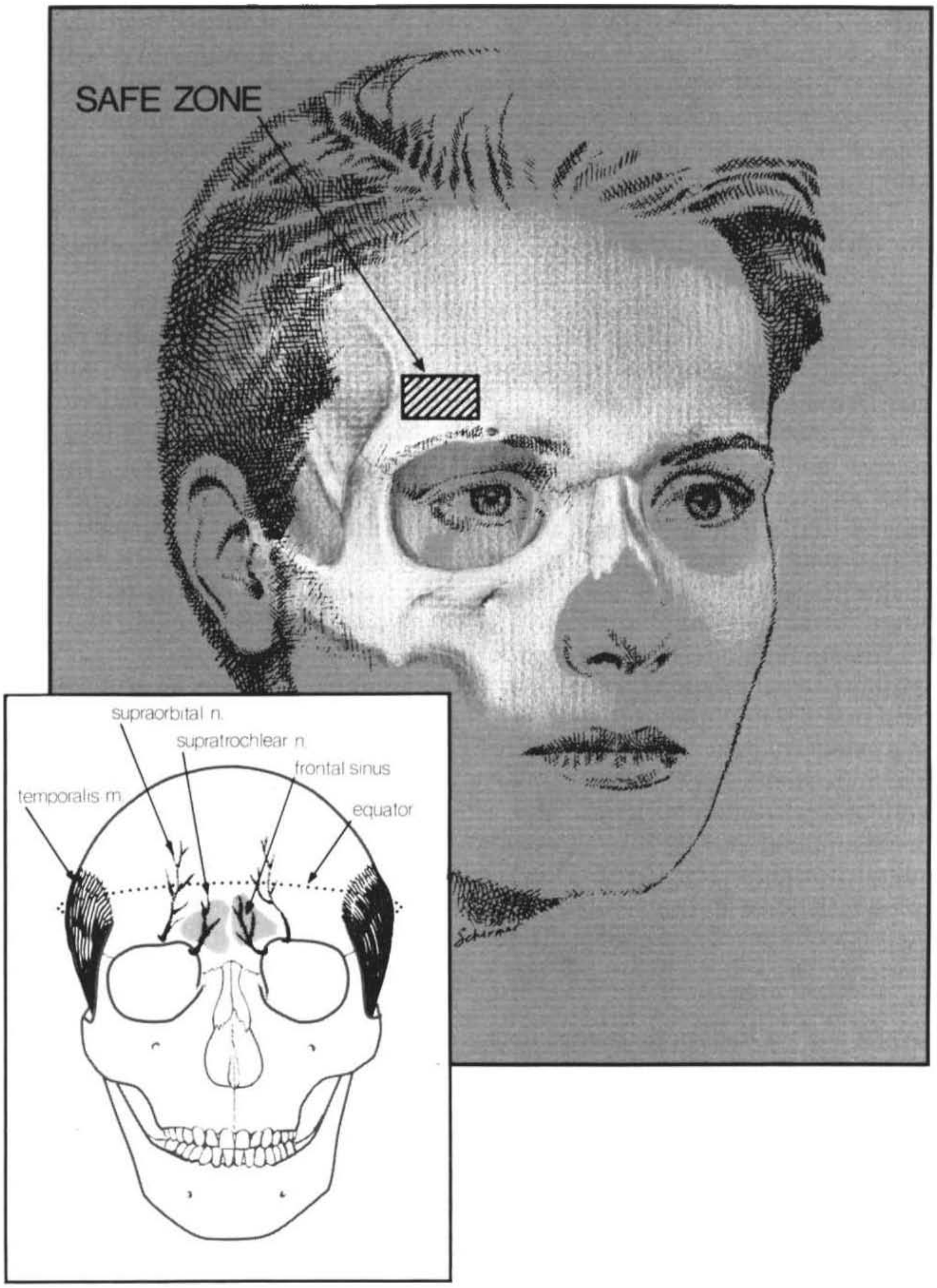

FIGURE 7-13 Illustration of safe zone for placement of anterior halo pin. Placement is best anterior to the temporalis muscle. Insertion lateral to the midportion of the supraorbital rim protects the supraorbital and supratrochlea nerves, as well as the frontal sinus. The pin should be below the equator (largest horizontal circumference) of the skull to prevent pin migration. The pin should be placed above the supraorbital ridge to protect the orbital contents. (From *Garfin, S. R., Botte M. J. Waters, R. L., and Nickel, V. L.: Complications in the use of halo fixation device. J. Bone Joint* Surg., 68A:320, *1986.)*

upon to prevent additional collapse of a severely comminuted thoracic spine fracture. If this is a major concern, a full body cast should be applied in hyperextension. It is also important to bear in mind that the decrease in the strength of the vertebral bodies in osteoporosis is such that the orthosis is not likely to be completely effective in protecting them from collapse.

The Taylor (TLSO) brace is one of the standards for the thoracic and the thoracolumbar spine (Fig. 7-14). This brace consists of a pelvic band with two long, posteriorly applied bars extending to the shoulders and joined with a transverse bar. There are straps that pass from these uprights around the shoulders and under the axilla. There is also a full-length abdominal pad that is attached to the uprights. Thus, it consists of a pelvic band and an axillary band attached by two rigid posterior uprights. The sleeve principle is employed for a splinting effect. The points of attachment at the axilla and the pelvis, with the abdominal pad anteriorly, constitute a three-point fixation.

The immobilizing efficiency of this brace, as shown by Nagel and colleagues,[41] is good in the lower thoracic area. The resistance against lateral bending is less effective because there are no lateral bars to prevent that motion. Although there are axillary shoulder attachments, resistance to axial rotation is not very satisfactory. This type of orthotic design functions largely as a reminder to resist excessive motion in flexion and extension. There are other models of this brace that enhance its usefulness in limiting other types of movement. To limit lateral bending, lateral uprights are added. They anchor the pelvic and thoracic bands and restrain lateral trunk bending (Fig. 7-15).

Sometimes there is a therapeutic indication to minimize axial rotation (y-axis rotation). This could be of primary importance in a patient in whom there was extensive pain elicited explicitly by that particular movement. For this, the clinician adds to the lumbosacral, anteroposterior, and lateral control brace bilateral subclavicular pads anchored to the lateral uprights (Fig. 7-15). This device provides some resistance and serves as an irritant and therapeutic reminder to the patient. There is no definite evidence that this device is superior to the axillary supports in resisting axial rotation.

Nagel and associates, in their cadaver studies of internal and external fixation of unstable thoracic spines, showed that in order to stabilize the upper thoracic spine, a CTLSO (cervicothoracolumbosacral orthosis) (some fixation to also include the neck) is required.[41]

Most Effective Control

A tightly worn Milwaukee brace and a well-molded Risser plaster jacket should be included in this category. Both of these appliances exert control against axial rotation as well as effective control of flexion/extension and lateral bending. The Risser cast controls axial rotation through its molding about the pelvis, the thoracic cage, the chin, and the occiput. The Milwaukee brace exerts its control through the pelvic mold, the localizer pads, and the axillary sling.

The most effective immobilization of the thoracic spine, as with all regions of the spine, is with the halo pelvic apparatus. With the skeletal fixation, this apparatus offers maximum control in all six degrees of freedom. Because of the viscoelastic properties of the spine and the strength of the bone to which the device must be fixed, it is not possible to apply enough tension to the spine to completely immobilize it, even with this orthosis (see Chap. 8).

TABLE 7–6 Indications for Orthosis in Disorders of the Cervical Spine

	CO (Cervical Orthosis)		CTO (Cervicothoracic Orthosis)	
	Soft	Rigid	Reinforced	Rigid
Congenital torticollis	1D	1D	1C	1D
Congenital malformations	1D, 3C	1D, 3C	1B	1A, B
Unstable fractures (includes postop. instability)	3, 4D	3, 4C	3, 4B	3, 4A, B
Stable fractures (includes postop. situations)	5C	5C	4C	4C
Rheumatoid arthritis and subluxation	3D	3D	3B	3B
Neurological disorders with paralysis	5D	5C	3B	3B
Acute torticollis (wry neck)	5D	5D	4, 5C	4, 5D
Cervico-brachialgia (e.g., painful conditions, soft and hard discs)	5C	5C	2B, 3, 4C	2–4B

(From Nachemson, A. L.: Orthotic treatment for injuries and disease of the spinal column. *In* Physical Medicine and Rehabilitation: State of the Art Reviews. Vol. 1, pp. 11–24, 1987.)

Key

CO: Soft felt collar; Rigid = plastic collar with occipital support.
CTO: Reinforced = Philadelphia collar; SOMI brace
Rigid = halo vest.

1 = correct deformity;
2 = limit motion;
3 = stabilize;
4 = unload;
5 = miscellaneous.

A = good clinical studies exist;
B = good biomechanical studies exist;
C = nonconclusive studies;
D = no support or negative studies.

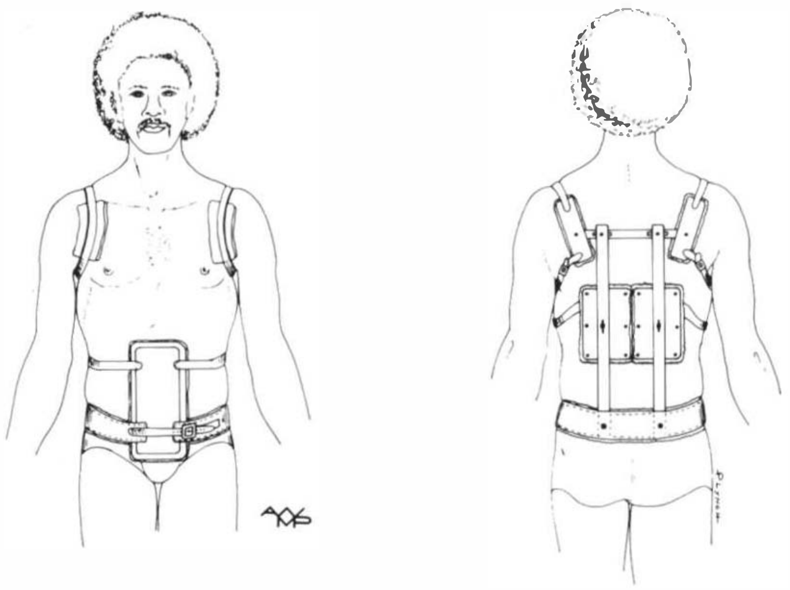

FIGURE 7-14 The Taylor (TLSO) thoracolumbar brace is one of the standards. Its components are described in the text. This orthosis offers *intermediate* control to the thoracic and the thoracolumbar spine.

The Milwaukee Brace in the Treatment of Scoliosis

The usual halo-hoop device has a turnbuckle mechanism in each of the uprights for displacing the halo on the y-axis with respect to the hoop. Distraction is produced by adjusting the turnbuckles. However, there is no way to determine the magnitude of the forces shared by the four uprights and the total force that is applied to the spine. A halo-hoop apparatus developed in Hong Kong and reported by Clark and Kesterton solves this problem in a neat and simple manner. They have incorporated a spring at the base of each of the four uprights so that the force in a given upright is transmitted to the hoop by way of the spring. By measuring the length of the previously calibrated spring, an accurate measure of the force is obtained. The total force applied to the spine is the sum of the four upright forces.[7] In the halo-hoop without measuring springs, an unknown amount of force is applied to the spine. In order to avoid the possibility of injuring the spine, the clinician may apply suboptimum levels of force. With the halo-hoop designed by Clark and Kesterton, a near-optimum level of force can be applied and maintained. Use of this device results in a highly efficient controlled form of distraction.

The Milwaukee brace represents a high degree of sophistication and elegance in the art of bracing.

FIGURE 7-15 In any thoracic or lumbar brace where exertion of an intermediate control of lateral bending (z-axis rotation) is desired, laterally placed uprights may be added to the basic Taylor orthosis shown in Figure 7-14. If the requirement is to control axial rotation (y-axis), the clavicular pads and straps shown here are recommended.

This complex apparatus embodies a number of mechanisms and principles involved in the clinical science of orthotics.

The principles involved in the brace and the technical considerations involved in fulfilling these principles are discussed here. The correctional effect is on a long-term basis and involves not only immediate mechanical effects but also biologic adaptation. Any mechanical analysis and optimization of such a system is extremely complex.

To gain some understanding of the mechanics involved, the Milwaukee brace and spine system may be studied in the simplified manner shown in Figure 7-16. The real situation is modeled as a plane-curved bar subjected to a set of forces. Forces F_1 and F_2 are the mandibular-occipital pads and the pelvic support forces that seem to correct the spine deformity by stretching. (Actually, the angular correction is obtained by producing bending moments in the scoliotic spine.) Forces F_3 (thoracic pad), F_4 (axillary sling), and F_5 (pelvic support) form a neat three-point force system. The purpose of these three forces is to bend the spine into a curvature opposite that of the scoliotic curve and thus correct it. The two force systems may be applied separately or together. In the combined situation, they are interdependent. The results of this interplay are studied in Chapter 3.

The brace is built to fit what would be expected to be the normal body of the patient. The normal mold is made with the body in the position of maximum attainable correction. This is achieved by the following procedures. In order to compensate for any functional or real leg length discrepancy, the pelvis is balanced by employing lifts on the short side. Any lordosis is then minimized by having the patient stand with the knees slightly bent in order to rotate the pelvis in the sagittal plane and minimize the amount of lumbar lordosis. Also, in order to gain maximum correction of the supple, growing scoliotic curve, the patient is suspended in a head halter traction apparatus. These important techniques are shown in Figure 7-17.

In order to obtain good fixation on the pelvis with additional support to the lower spine, the cast for the mold is carefully fitted to the pelvis with special attention to the iliac crests. When the final mold is made, the abdominal portion of the mold is carved out considerably before the actual pelvic girdle is fitted to it. This is done in order to assure significant compression and better fixation.

Erect uprights perpendicular to a level pelvis are then constructed. These uprights support an occipital headpiece and throat mold. A plastic throat mold has been employed in recent years as a substitute for

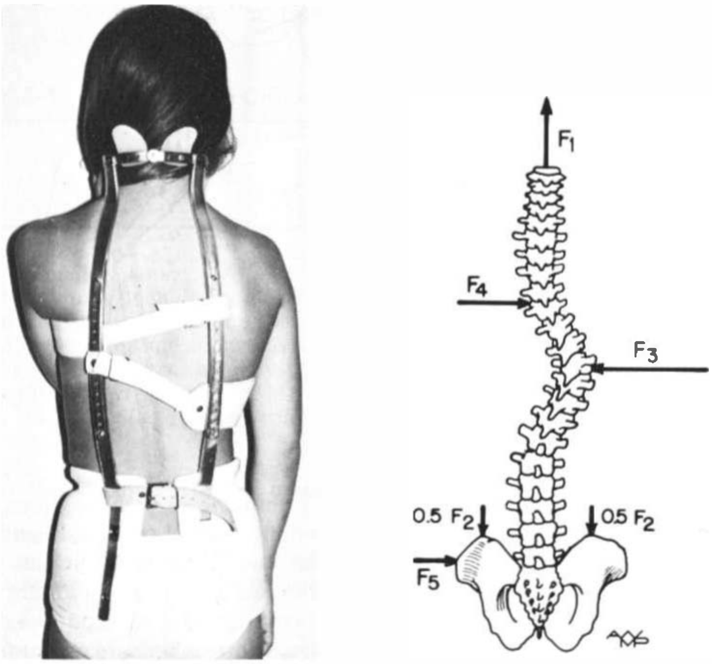

FIGURE 7-16 Here, the forces exerted by the Milwaukee brace for correction of a scoliotic deformity are analyzed. The spine is subjected to stretch by equal forces, F_1 and F_2, applied by the way of the occipital mandibular pads and the pelvic girdle, respectively. Forces F_3 (thoracic pad), F_4 (axillary sling) and F_5 (pelvic girdle) are all basically horizontal forces. They form a three-point system. The maximum bending moment for correction to the spine is applied at the level of the force F_3.

FIGURE 7-17 The Milwaukee brace is built to fit the expected normal body mold for the particular patient. Moderate axial traction is applied. The foot is elevated to attain a level pelvis, upon which perpendicular uprights of the proper length may be measured and constructed. In addition, both knees are flexed to rotate the hips and pelvis in order to minimize lumbar lordosis.

the chin piece used in the past. The throat mold is shown in Figure 7-18. This has been quite effective in reducing the amount of dental changes associated with previous methods. The throat mold and the occipital piece, in conjunction with the uprights, work together to provide a distracting force on the spine. Thus, they resist settling into the more deformed position. They resist gravity and the deforming forces intrinsic to a scoliotic spine. In addition, these components of the brace serve as reminders and reference points away from which the patient may actively move, employing his or her own muscles and actively correcting the deformity.

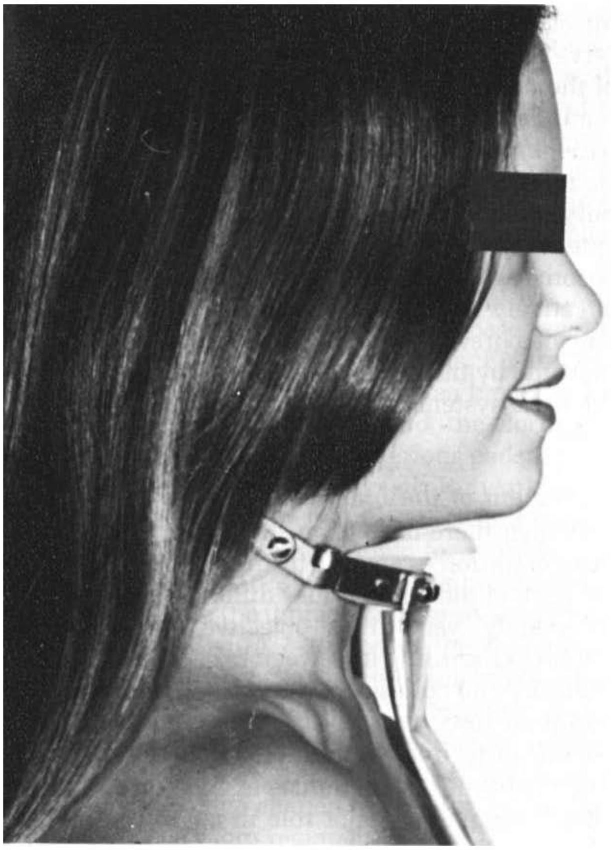

FIGURE 7-18 This type of throat mold now commonly used appears to be therapeutically just as effective as its predecessor, the chin piece, yet it is much less likely to cause disruptions of the teeth.

Localizer pads are used on this brace to provide the valuable function of both active and passive correction. The brace may employ any combination of three basic types of localizing pads. One is the typical thoracic pad, which is applied posterolaterally over the ribs. The pad is applied slightly posteriorly in order to have its force also serve to correct the rotational aspect of the scoliotic deformity by providing an axial torque. There is also a lumbar pad, which is generally smaller and more heavily padded. This is attached to the posterior uprights and presses posterolaterally on the erector spinae muscles at the apex of the lumbar curve. On occasion in scoliosis, a sternal pad may be employed. This pad fits on the anterior upright and is also generally quite well cushioned. These localizer pads have a function similar to the distracting function of the uprights. In effect, they apply correctional loads and serve as a check mechanism to prevent progression of the deformity. At the same time, they serve as a reminder or a reference point away from which the patient may voluntarily move, using his own intrinsic muscles, and thereby actively correct the deformity. The axillary sling is employed on the opposite side of the convexity of the curve. This offers a counterforce against the thoracic pad, which contributes effectively to a three-point fixation loading system and also prevents the patients from being pushed to that side by the thoracic pad. A detailed analysis of the force system and placement of lateral pads follows.

Placement of the Lateral Pads

Although there have been studies on the measurement of the forces applied by the occipital-mandibular pads to the spine under different activities, unfortunately very little objective information is available regarding force vectors of the lateral pads. Mulcahy and colleagues report that the average longitudinal force in the Milwaukee brace increases significantly upon removal of the thoracic pad.[36] They believe that the forces extended through the chest cage play a major role in passive correction. There are three lateral pads—axillary, thoracic, and lumbar. For a particular patient, how many of them should be used, and in what combination? Where should they be located, and how much pressure should be applied through each pad for optimum results? Which way should the forces be directed from the pads so that there is some improvement of axial (y-axis) rotation of the vertebrae? These are all crucial questions that merit consideration.

At present, all the answers are not available. However, the problem may be analyzed biomechanically, and some practical recommendations can be offered. One of the most popular concepts regarding the lateral pads is to assume that the axillary pad, the thoracic pad, and the pelvic support form a three-point force system. In discussing the concept of balanced horizontal forces (see p. 481), one of the conclusions reached was that the maximum bending moment occurs just under the middle force (*i.e.*, the thoracic pad). The question arises concerning the location of the thoracic pad with respect to the apex of the scoliotic curve. Orthopedic opinion seems to be divided on this question. However, a majority of physicians prefer that the thoracic pad be placed against the ribs *attached* to the apex of the curve and not *at the level* of the apex of the curve. However, simple biomechanical analysis, based upon the three-point force system, reveals that the optimum placement of the thoracic pad is midway between the pelvic support and the axillary pad.[8] Additional investigation is required to clarify these considerations more definitely.

The concept of dynamic bracing is employed with the Milwaukee brace in two basic manners. One has already been discussed—the active movement away from the localizer pads. In addition, an integral part of the basic Milwaukee brace prescription includes a series of well-conceived specific exercises to be followed under the supervision of a physical therapist. These consist of breathing exercises and activities to counteract a tendency for the development of an excessive lumbar lordosis. The routines include breathing, pelvic tilt, and abdominal, back, hip, shoulder, and arm strengthening exercises.

Experimental studies have shown that the Milwaukee brace is an effective appliance for applying corrective and immobilizing forces to the spine. The brace has been shown to function in these two ways in both the supine and the prone positions. It has been shown that removal of the thoracic localizer pad, the occipital piece, or the headpiece significantly impairs the effectiveness of the brace in applying correctional forces to the spine.[14,40]

Investigators have recently examined the efficacy of brace treatment for the correction of scoliosis. A prospective controlled multicenter study is being completed by the Scoliosis Research Society under the direction of Professor Alf Nachemson. Current available studies show different results for correction of middle and upper thoracic curves.[12,42,59] The success with lower thoracic and upper lumbar curves is more evident.[5,59]

Indications for orthotic treatment of scoliosis are given in Table 7-7.

Orthoses for Scoliosis with Pelvic Obliquity

There is an implementation of the halo apparatus that makes it the most effective orthosis for applying distractive correctional loads to the deformed spine. This instrument can effectively apply forces resulting in + y-axis translation and immobilization of the spine. The halo-hoop should be considered in the treatment of scoliosis with severe pelvic obliquities. Skeletal fixation is obtained in the outer table of the skull with the halo, and in addition, fixation to the pelvis is achieved with the use of large pins applied through holes in a circular hoop apparatus. The pins are applied through the upper and outer portion of the wings of the iliac crests. With a series of turnbuckle screw mechanisms, the desired amount of distraction is applied to the spine. This device is also useful in situations in which there is clinical instability in a portion of the spine that is yet to be stabilized surgically.

This device applies forces and controls to any area of the spine better than other currently available spine orthoses. Strain gauges can and probably should be incorporated in the apparatus so that the forces involved can be precisely controlled and monitored.

Milwaukee Brace in the Treatment of Kyphosis

This brace is used in essentially the same fashion, embodying in the treatment of this disease principles that are identical to those in scoliosis. The basic difference is that the deformity and the curvature are in the sagittal plane and are without a significant element of axial rotation. The scoliotic deformity is largely in the frontal plane and embodies a significant element of axial rotation. Both diseases involve deformities within particular vertebrae. This analogy is used for descriptive reasons alone and does not imply similarities in etiologic or other aspects of the two diseases. The corrective forces are applied through the use of distraction between the well-molded pelvic band and the occipital throat or chin piece. Two localizer pads are used. One is a sternal pad and the other is a dorsal pad that is applied to the apex of the kyphotic deformity. An analysis of this orthosis in the sagittal plane demonstrates three-point fixation systems (Fig. 7-19).

The correctional effects produced by the Milwaukee brace in treating kyphosis are based on the biomechanical principles for scoliosis. Figure 7-19 shows a patient using the Milwaukee brace for kyphotic correction. The spine and the correcting forces applied to it through the various pads are also shown. The five forces, F_1 through F_5, work in the sagittal plane in a manner similar to that of the five forces in the frontal plane, shown for scoliotic correction in Figure 7-16. The sternal pad replaces the axillary sling, and the thoracic pad is replaced by the posterior pad. Figure 7-20 shows the therapeutic effectiveness of this orthosis. *Biomechanical adaptation* as a long-term therapeutic response to the appropriate use of an orthosis is further demonstrated in Figure 7-21. In Figure 7-21A, the thoracic vertebrae are wedge-shaped. In Figure 7-21B, they are noted to be more rectangular.

Documented clinical experience shows that a TLSO is effective in correcting kyphosis in the growing child with Scheuermann's disease.[27,33,42,60–62]

TABLE 7–7 Indications for Orthosis in Scoliosis and Kyphosis (Growing Subjects Only)

	CTLSO (Cervicothoracolumbosacral Orthosis) Reinforced	TLSO (Thoracolumbosacral Orthosis) Reinforced
Scoliosis		
T1–T8	1C	1C
T9–L1	1A, B	1A, B
Lumbar		
L2–L4	1D	1A, B
Kyphosis		
T1–T8	1A	1C
T9–L1	1B, C	1A, B

(From Nachemson, A. L.: Orthotic treatment for injuries and disease of the spinal column. In Physical Medicine and Rehabilitation: State of the Art Reviews, Vol. 1, pp. 11–24, 1987.)

Key

1 = correct deformity;
2 = limit motion;
3 = stabilize;
4 = unload;
5 = miscellaneous.

A = good clinical studies exist;
B = good biomechanical studies exist;
C = nonconclusive studies;
D = no support or negative studies.

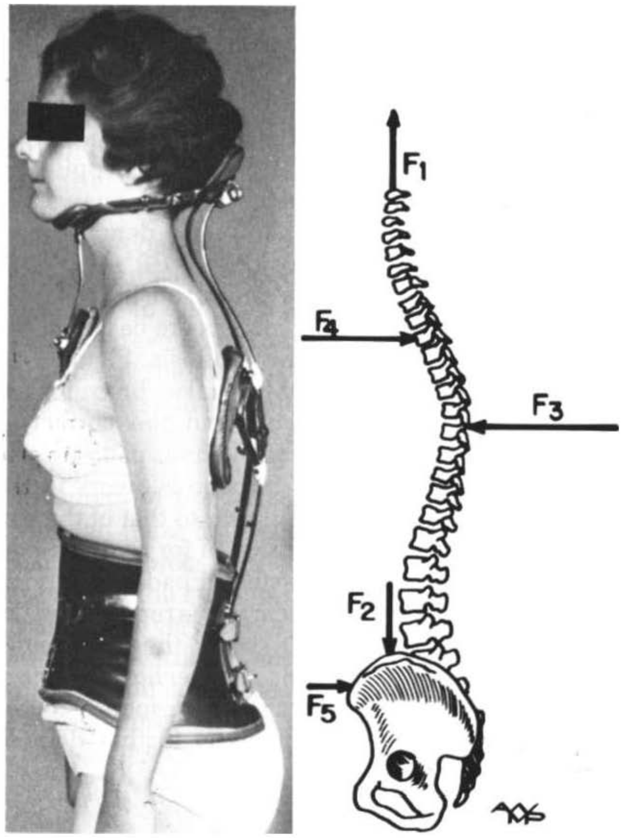

FIGURE 7-19 Here, the forces exerted by the Milwaukee brace for correction of a kyphotic deformity are analyzed. The spine is stretched by two equal axial forces, F_1 and F_2, applied to the spine by way of the occipital-mandibular pads and the pelvic girdle, respectively. The other three forces, F_3, F_4, and F_5, are basically transverse to the spine axis and are applied by way of the posterior pad, the sternal pad, and the pelvic girdle, respectively. These three forces form a three-point force system producing maximum bending moment and hence the correction potential at the posterior pad level.

Orthotic Treatment of Thoracic Spine Osteoporosis

This is a difficult and important problem that has not been completely studied. The elderly female presents with advanced osteoporosis deformity and severe pain. Presumably, the pain is from repeated microfractures, which also add to the deformity. It is difficult to provide a tolerable orthosis, not to mention one that can relieve the pain. There is also the dilemma that effective pain-relieving immobilization theoretically contributes to advancement of the disease process. We suggest treatment with a customized, well-padded Jewett brace for 6–8 weeks, or less if symptoms subside. When the Jewett brace cannot be tolerated, another thoracolumbar spine orthotic, such as the Taylor, is employed.

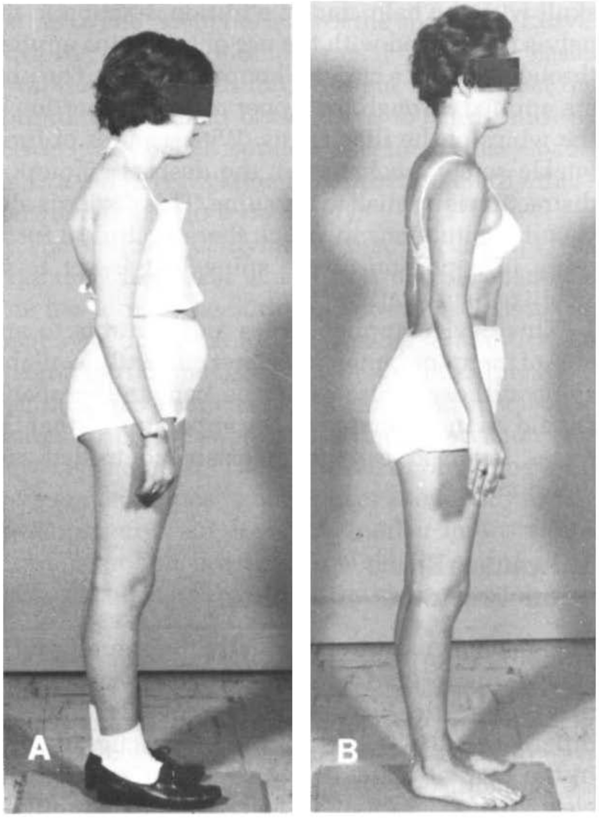

FIGURE 7-20 These photographs show the clinical effectiveness of the Milwaukee brace in the treatment of adolescent kyphosis. In this patient, the brace was worn for 3.5 years. **(A)** Before treatment. **(B)** After treatment. This change may be thought of as *biomechanical adaptation*.

Synopsis of Thoracic and Thoracolumbar Spine Orthotics

An updated review of the literature by Nachemson[39] shows that, based on various motion-restraining capabilities of orthotic devices,[13,17,18,28,30,45,63] the selection for prescriptions can be determined from Table 7-8. These studies also support the assertion that a Jewett brace is particularly effective in resist-

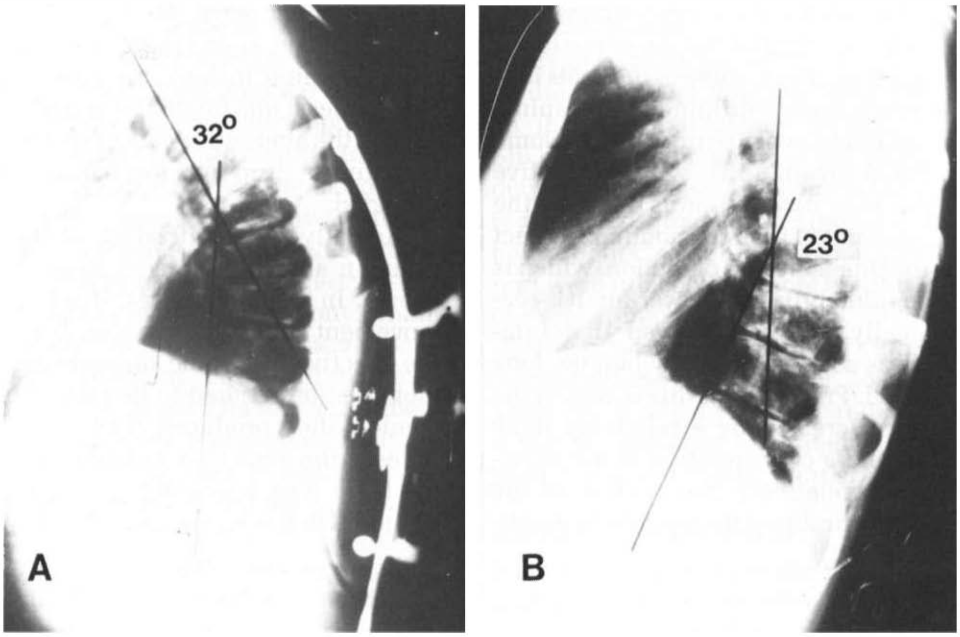

FIGURE 7-21 This shows changes in the shape of thoracic vertebrae associated with Milwaukee brace correction of adolescent kyphosis in the patient shown in Figures 7-19 and 7-20. **(A)** The deformity measures approximately 32°, and the vertebrae are wedge-shaped. **(B)** The spine after wearing the brace for 3.5 years. Note the new vertebral configuration as a result of *biomechanical adaptation*.

ing flexion, and the Norton–Brown brace (possibly because of the irritating effects of trochanteric pads) is useful in avoiding lateral bending.

Lumbar Region

Most of the long and sometimes confusing list of eponyms for spine orthoses are associated with this region of the spine. The number of appliances and their proliferation mirrors the confusion and complexity inherent in treatment of "low back pain." The "brace" represents a valiant effort among the many attempts to treat a formidable clinical problem.

Lumbar spine braces are most often used to reduce pain. They may be employed for giving support and/or immobilization following spine fusion or trauma to the spine. Mechanically, the braces generally seek to achieve increased abdominal support, to reduce the forces on the spine, and to achieve a straighter lumbar spine. Forces are applied to the normal or accentuated lumbar lordosis in order to hold it in a straighter position. It is commonly observed clinically that the less lordotic spine is more comfortable.

Experimental Studies

Although there are few investigative studies on the mechanics of braces, there is at least one important study, and it is probably the most objective scientific study of the effects of bracing on the lumbar spine. Norton and Brown investigated movement of the spine in braces using radiographs and the insertion of K-wires in the spinous processes for measurement.

Standing, sitting, and bending in flexion and extension were then studied with a number of braces. The braces include an experimental brace created by the investigators, a chairback brace, a Goldwaith brace, a Williams brace, the Arnold Albert brace, a flexion Taylor brace, a rigid Taylor brace, a reinforced Taylor brace, a Jewett brace, and a plaster jacket.[45]

Although the investigators considered their extensive work preliminary, it provided interesting and worthwhile information. The pertinent findings are reviewed. Sitting with a brace, even when erect, was associated with substantial flexion of the lower two lumbar interspaces. Thus, if one of the goals is to immobilize the lower lumbar or lumbosacral spine, the patient should either avoid sitting or wear some apparatus for immobilization that will be effective when sitting. The long back supports, such as the Taylor brace, concentrated their immobilizing effect in the region of the thoracolumbar junction, which is much too high to immobilize the lower lumbar segments. Paradoxically, it was observed that lumbosacral flexion was actually greater when the long brace was employed. Presumably, this is due to the increased lever arm created by a relatively more rigid upper spine and a concentration of the movement in that lower, relatively free portion of the spine. Thus, if it is desirable to thoroughly immobilize the lumbosacral joint (as, for example, with spondylolysis, spondylolisthesis, or following lumbosacral spine fusion), the cast or brace must include at least one thigh. If it is desirable to allow the patient the alternative of occasional hip joint flexion for sitting, a drop-lock mechanism at the hip can be included.

Actually, immobilization of the spine did not occur in any of the braces studied by Norton and Brown. In some instances, it was possible to limit movement in the interspaces. It is interesting that the effectiveness of the supports with respect to immobilization seemed to be related more to the discomfort they produced than to the actual magnitudes of the force transmitted from the apparatus to the body. The desirability of a paraspinal brace to immobilize the lumbosacral spine was questioned.

TABLE 7–8 Indications for Orthosis in Thoracic and Thoracolumbar Disorders*

	CTLSO (Cervicothoracolumbosacral Orthosis)	TLSO (Thoracolumbosacral Orthosis)	
	Reinforced + Rigid	Reinforced	Rigid
Congenital Malformations			
T1–T8			
Stable	1, 5C	1, 5D	1, 5D
Unstable	3B	3D	3D
T9–L1			
Stable	2C	1, 2C	1, 2C
Unstable	3B	3C	3C
Spine Fractures			
T1–T8			
Stable	2C	2D	5D
Unstable	3B	3D	3C
T9–L1			
Stable	2C	2C	2C
Unstable	3B	3C	3C
Neurol. Diseases with Muscle Paralysis			
T1–T8 OBS:			
incl. sitting orthosis	3A, B	3D	3C
T9–L1	3A, B	3A, B	3B, C

(From Nachemson, A. L.: Orthotic treatment for injuries and disease of the spinal column. *In* Physical Medicine and Rehabilitation: State of the Art Reviews. Vol. 1, pp. 11–24, 1987.)

* In this table, a body jacket is classified as a *rigid* TLSO, and the Milwaukee brace is classified as a *reinforced* TLSO.

Key

1 = correct deformity;
2 = limit motion;
3 = stabilize;
4 = unload;
5 = miscellaneous.

A = good clinical studies exist;
B = good biomechanical studies exist;
C = nonconclusive studies;
D = no support or negative studies.

They felt that these had limitations in two realms: the force was not localized low enough in the lumbar area, and these uprights did not produce the necessary discomfort. The investigators developed an experimental brace designed to utilize this information. The paraspinal uprights were replaced with lateral uprights, which extended downward to the greater trochanters. The brace also applied a force to the lumbosacral region by means of a single crossbar. This component exerts a force over the bony prominence so that pressure and discomfort accompany the early ranges of movement. This is all mediated through an abdominal pad with its low attachment straps. Good counterpressure is offered without impeding sitting in the erect posture. In addition, side bending is effectively blocked by the lateral uprights.[45]

Increases in intra-abdominal pressure may provide additional support to the lumbar spine. This is important in the use of abdominal corsets and also in the use of the lumbosacral corsets and the chairback brace. Walters and Morris carried out studies of the electromyographic (EMG) response of the paraspinous and abdominal muscles with and without either a lumbosacral corset or a chairback brace. These investigators found a decrease in the activity of the abdominal muscles with both the lumbosacral corset and the chairback brace.[56] This implies that these braces take over some of the function of the abdominal muscles by compressing and supporting the spine. Subjects wearing braces in the resting position showed either no effect or some decrease in the abdominal muscle activity. With ambulation, however, the wearing of the chairback brace was associated with an *increase* in muscle activity. This is presumably due to an attempt on the part of the muscles to overcome the immobilizing effect of the brace. If the use of the brace is based on resting paraspinous muscles, presumed to be causing pain through their spasmodic contractions, the desirability of using such a brace should be examined critically. This is because the study showed that there is *greater* paraspinous muscle activity with these two braces. In other words, the brace could worsen the patient's condition.

Morris and associates also studied the impact of an inflatable rubber balloon in the front of a corset. The investigators observed some mechanical unloading in symmetric forward-bent positions with the inflated balloon.[35]

Clinical Studies

There are numerous, not very rigorously controlled, studies of lumbar orthoses. They nevertheless show an interesting tendency to yield similar results. Slightly more than 50% of the patients feel that the device helps them, while the rest see no effect. These randomized prospective studies did not show any helpful effects.[39] Lumbar corsets with a hand support were subjectively more helpful than those without any support.[32]

The most cogent results of studies of the immobilizing efficacy of lumbar orthoses are presented in Table 7-9. The salient clinical biomechanical point is that one hip must be fixated in order to effectively immobilize the lumbar spine.

Decisions for Treatment

Because of the nonspecific nature of low back pain and the paucity of excellent clinical studies, this is not an area in which treatment determinations can be firmly based on hard evidence. There are, however, some basic principles and biomechanical data that provide a reasonable base. They are presented in the next several pages.

Willner[58a] has developed a test instrument for predicting the effect of rigid braces in patients with low back pain. The instrument shown in Figure 7-22 consists of an aluminum frame with an adjustable

TABLE 7–9 Reduction in Percent of Normal Mobility by Different LSOs and TLSOs from Different Lumbar Measurements

Orthosis	Flexion	Extension	Lateral Flexion	Rotation
TLSO				
Cloth, reinforced	50	50	20	0
TLSO				
Rigid	60	60	50	30
With one hip included	90	90	70	90
LSO				
Cloth	30	30	10	0
LSO				
Rigid (plastic)	60	60	40	20

(From Nachemson, A. L.: Orthotic treatment for injuries and disease of the spinal column. *In* Physical Medicine and Rehabilitation: State of the Art Reviews. Vol. 1, pp. 11–24, 1987.)

LSO = lumbosacral orthosis; *TLSO* = thoracolumbosacral orthosis.

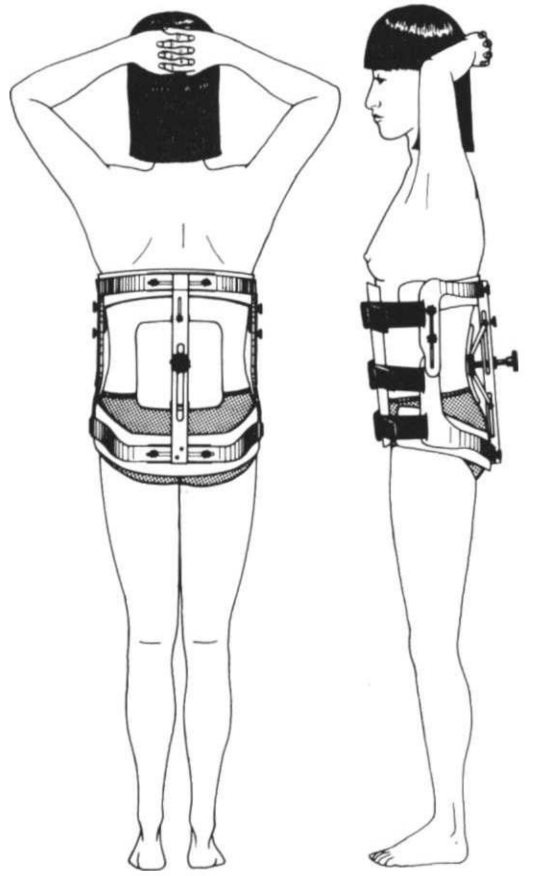

FIGURE 7-22 This is the W.I.S.S. (Willner Instrument for Spinal Stabilization). The height and protrusion of the lumbar support is adjustable and measurable, as is the abdominal support. The individually adjusted test instrument is worn as a trial for pain relief. When the pain relief is achieved, the measurements are used to facilitate the fabrication of the final orthosis.

width and length. This is applied to the patient's back. There is a back support adjustable for both height and lumbar support. Abdominal support is also adjustable. The device is designed to transfer measurement information to the brace module, which simplifies construction of the brace. Dr. Willner recognizes that it is difficult to predict both the efficacy and the acceptance of an orthosis prior to the expenditure of resources to create one. The ideal extent of lordosis, the level and extent of lumbar support, and the abdominal support are probably unique for each patient. Altering these variables to suit the individual patient can provide maximum pain relief. It was found that if the test orthosis did not relieve pain, neither would the rigid brace. The test brace measurements also simplified the maintenance of the brace. This apparatus appears to be a promising new development in the use of spinal orthoses for pain relief.

The functions of a lumbar orthosis are as follows: to serve as a reminder and an irritant to the patient for restriction of movements and activity in the lumbar spine; to act as a support and a vehicle for application of abdominal pressure (which should somewhat alleviate the loads imposed on the lumbar spine); to provide some immobilizing efficiency of the upper portion of the lumbar spine and the thoracolumbar area; and to maintain a straighter and more comfortable back by employing the principle of three-point fixation.

Instead of reviewing the extensive list of conditions of the lumbar spine that may be treated, the authors submit the following type of stepwise analysis for consideration. First, the clinician decides what goals he is attempting to achieve with the orthosis: what are the mechanical factors involved, the motions that are to be restricted, or the structures that are to be corrected or supported? When maximum immobilization is needed, a more rigid structure, such as a cast or brace, is required. For less rigid immobilization, a corset or a pelvic belt may be considered. If the goal is primarily to limit anterior or posterior movement (flexion and extension), then pelvic and thoracic bands connected to posterior uprights are probably the most effective. If lateral motion is also to be limited, then lateral uprights are desirable, along with the consideration of the trochanteric pads of Norton and Brown's experimental brace. When axial rotation is to be diminished, a well-molded body plaster is applied. If rotation is to be controlled, perhaps a longer brace with good pelvic fixation and fixation on the upper portion of the thorax may be necessary. It is useful to note here that the cadaver experiments of Nagel and co-workers[41] showed the plaster body jacket to be an excellent immobilizer in all planes. It was superior to other TLSOs and was even superior to Harrington distraction rods. The immobilizing superiority of the plaster body jacket over other TLSOs is also supported by the work of Fidler and Plasmans.[13]

Minimum Control

This group is composed of the various corsets that are available for the lumbar spine. They differ in the controls they apply, depending on the quality of the fit and the quality, quantity, and distribution of the staves.

Intermediate Control

Braces in this category include the low or short lumbar spine brace (Williams type); the slightly longer Knight or MacAusland (chairback) brace; and the long lumbar spine brace, which is actually a Taylor brace. These orthoses are used most commonly as a "crutch" for the patient with the chronically disabled back. Since they make the patient feel better and it is not certain that they cause disabling loss of intrinsic muscle function, it is reasonable to use them. Figure 7-23 shows back and side views of the basic designs of lumbar orthoses. The Norton–Brown experimental brace should be included in this group and is probably its most effective member. However, as a group, these braces provide little or no control of rotation; flexion/extension is not well controlled in the lower lumbar and lumbosacral area; and lateral bending is controlled to some degree (see Table 7-10.).

Most Effective Control

Based on our analysis, the appliances that most effectively control the lumbar spine are listed in their ascending order of control: Taylor brace with thigh attachment; molded plaster body jacket (lower lumbosacral area not immobilized); molded plaster body jacket with thigh included; and the halo pelvic skeletal apparatus.

Braces from this category are used when control must be maximal. However, a broad range of control is represented in this group. The halo pelvic apparatus is the most effective external device now available for controlling all six degrees of freedom. The plaster casts are effective in reducing axial rotation due to pelvic and thoracic molding. Because of their compression of the abdomen and their rigidity, they are also effective against flexion/extension and lateral bending. This does not apply to the L4–S1 area, however, and the thigh is best included when maximum control of this area is important. An example of an orthosis that includes the hip and thigh on one side is depicted in Figure 7-24. Axial rotation, as well as flexion/extension, is better controlled by this maneuver. The Taylor and Norton–Brown braces are more effective controls than those of the intermediate group, but they are not nearly as effective as the halo pelvic apparatus or the plaster jacket with the thigh included.

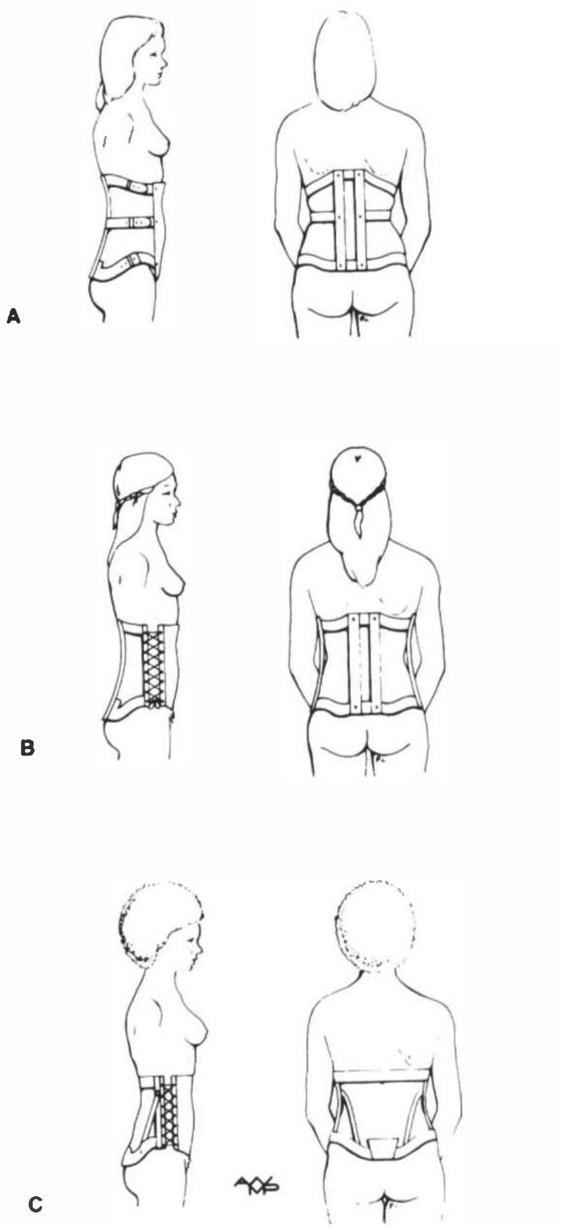

FIGURE 7-23 (A) MacAusland (chairback) brace. This orthosis offers intermediate control for the upper lumbar spine in flexion and extension. It is less effective in lateral bending (no lateral uprights) and least effective in controlling axial rotation. (B) Knight brace. This orthosis offers intermediate control for the upper lumbar spine in flexion/extension and lateral bending (note lateral uprights). It does not provide effective control of axial rotation. (C) Williams brace. This brace exerts intermediate control against flexion/extension and lateral bending but not against axial rotation. All three braces provide general support and stability through compression via the abdominal supports.

There are two additional tables that are of considerable value in prescribing lumbar orthoses. The first (Table 7-11) is used when there is a specific diagnosis that can be confirmed by objective observations. The second (Table 7-12) is designed to be used in situations where there is pain without a specific diagnosis but with a categorization based upon clinically meaningful characteristics.

CONCLUSION

Mechanical and psychologic factors can sometimes interact and become a major consideration in the prescription for an orthosis. For example, patients with severe neurologic deficits who do not have control of the trunk and pelvic musculature may not make use of a rigid spinal brace and attached lower extremity braces. In the hospital setting, the apparatus may seem to be helpful to some extent, but there is considerable psychologic support from physicians, nurses, and therapists. The same type of situation rarely works out well at home; frequently such orthoses are discarded, and the patient simply uses a wheelchair.[23] Thus, it is important to be as realistic as possible about what can be expected of an orthosis from a practical as well as a mechanical point of view.

With all the foregoing considerations, a clinical biomechanical approach to the use of orthotics may

TABLE 7–10 **Functional Analysis of Spinal Braces**

Orthosis	Standardized Nomenclature*	Spine Region	Degrees of Freedom Controlled	Effectiveness of Control
Soft cervical collar	CO	Cervical	FE, LB	Min.
Hard plastic collar (Thomas)	CO	Cervical	FE, LB, (AR)	Int.
Philadelphia	CTO	Cervical	FE, LB, (AR)	Int.
Four-poster	CTO	Cervical	FE, LB, AR	Int.
Long two-poster (Guilford, Duke)	CTO	Cervical	FE, LB, AR	Int. (high)
Minerva cast	CTLSO	Cervical	FE, LB, AR	Most
Halo device	CTO	Cervical	FE, LB, AR	Most (high)
Long thoracic corsets	TLSO	Thoracic	FE, LB	Min.
Three-point (Jewett, Griswold)	TLSO	Thoracic	FE	Int.
Taylor	TLSO	Thoracic, thoracolumbar	FE	Int.
Taylor (with lateral uprights)	TLSO	Thoracic	FE, LB, AR	Int.
Use of clavicle pads	TLSO	Thoracic	AR	Int.
Milwaukee brace (tightly worn)	TLSO	Thoracic, thoracolumbar	FE, LB, (AR) Kyphosis correction	Most
Risser plaster jacket	TLSO	Thoracic, thoracolumbar	FE, LB, AR	Most
Milwaukee brace (loosely worn)	TLSO	Thoracic, thoracolumbar	Scoliosis correction	Most
Halo pelvic device	CTLSO	Thoracic, thoracolumbar	FE, LB, AR	Most (high)
Corsets	TLSO	Lumbar	FE, LB	Min.
Williams	TLSO	Lumbar (except L4 to L5)	FE, LB	Int.
Knight	TLSO	Lumbar (except L4 to L5)	FE (LB)	Int.
MacAusland	TLSO	Lumbar (except L4 to L5)	FE (LB)	Int.
Taylor	TLSO	Lumbar (except L4 to L5)	FE, LB, AR	Int.
Norton–Brown (exp. brace)	TLSO	Lumbar (except L4 to L5)	FE, LB	Int, (high)
Taylor (with thigh attachment)	TLSO+	Lumbar	FE, LB, AR	Most
Molded plaster jacket	TLSO	Lumbar (except L4 to L5)	FE, LB, AR	Most
Molded plaster jacket (thigh included)	TLSO+	Lumbar	FE, LB, AR	Most
Halo pelvic device	CTLSO	Lumbar	FE, LB, AR	Most (high)

* See Table 7-2.

Key

FE: Flexion/extension (x-axis rotation)
LB: Lateral bending (z-axis rotation)
AR: Axial rotation (y-axis rotation)
( ) Slightly less controlled
Min: Minimal
Int.: Intermediate
Most: Most effective
+: Includes hip control

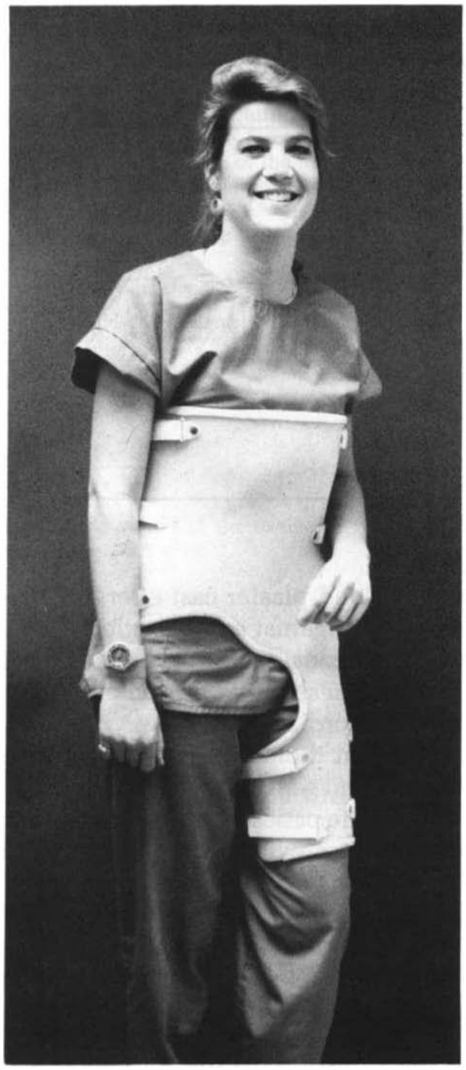

FIGURE 7-24 This TLSOT includes the thigh, in order to stabilize the lumbosacral joint and better stabilize the L4-5 segment. This can be used as a removable shell as shown here, or as an orthosis, fabricated either with plaster or polypropylene.

be taken through answers to the following questions. What are the pathologic conditions that are involved in the spine? What are the therapeutic mechanical goals? In what way should the mechanics of the spine be changed? Is the goal to protect the spine, to rest it, or to correct it? What kinds of forces are necessary in order to achieve the therapeutic aims?

The type of forces necessary can be determined by a review of the basic kinematics of the spine in the region where the forces are to be applied. Then it is possible to decide which orthotic devices are able to apply the needed loads so as to best achieve the desired mechanical result. There are limitations in the extent to which forces may be applied to the spine; these involve psychologic, physiologic, and mechanical factors.

In general, the corsets and collars and relatively flexible supports apply the least amount of control to the spine. Short models of the braces that employ rigid uprights and molds provide better control, followed by the longer braces. They have greater leverage, which offers additional mechanical advantage in the application of forces. Mechanical devices may be added to these braces to deliver additional therapeutic support. Examples include devices to limit axial rotation and a number of pads and supports that apply more discrete, localized loads for explicit purposes. This is effectively employed in the treatment of scoliosis or kyphosis. Spinal orthoses that incorporate the thigh or add some special extension, as in the Norton–Brown experimental brace to augment immobilizing efficiency, may also be used. Increased fixation where desired can sometimes be achieved through the use of more rigid material, such as plaster.

Finally, the level of maximum immobilization is

TABLE 7–11 Indications for Orthosis in Lumbosacral Disorders with Verifiable Diagnosis

	LSO (Lumbosacral Orthosis)		
	Soft	Reinforced	Rigid
Spondylolisthesis	2, 3D	2, 3B, C	1, 2, 3B, C
Disc Hernia	2D	2, 4, 5B, C	2, 4B, C
Degenerative Instability	3D	3D	3B, C
Spinal Stenosis	1D	1, 3C	1, 3B, C
Osteoporosis	2, 4, 5C	2, 4, 5C	2, 3B
Pelvic Instability	3D	3C	3B
Rheumatoid Diseases	1–5D	1–5C	1–5B, C

(From Nachemson, A. L.: Orthotic treatment for injuries and disease of the spinal column. *In* Physical Medicine and Rehabilitation: State of the Art Reviews. Vol. 1, pp. 11–24, 1987.)

Key

1 = correct deformity;
2 = limit motion;
3 = stabilize;
4 = unload;
5 = miscellaneous.

A = good clinical studies exist;
B = good biomechanical studies exist;
C = nonconclusive studies;
D = no support or negative studies.

TABLE 7–12 Indications for Orthosis in Lumbosacral Disorders with Nonspecific Diagnosis

	LSO (Lumbosacral Orthosis)			SIO (Sacroiliacal Orthosis)
	Soft	Reinforced	Rigid	Soft, Reinforced
Acute Low Back Pain	2, 4, 5D	2, 4, 5D	2, 4, 5D	2, 4, 5C
Chronic or Recurring Low Back Pain	2, 4, 5D	2, 4B, 5C	2, 4B, 5C	2, 4, 5D
Acute Low Back and Leg Pain	2–5D	2–5C	2–5C	2–5C
Chronic Low Back and Leg Pain	2–5D	2–4B, 5C	2–4B, 5C	2–5C
"Sacroiliac Disorders"	3D	3D	3C	3C

(From Nachemson, A. L.: Orthotic treatment for injuries and disease of the spinal column. In Physical Medicine and Rehabilitation: State of the Art Reviews. vol. 1, pp. 11–24, Philadelphia, Hanley and Belfus, 1987.)

Key

1 = correct deformity;	A = good clinical studies exist;
2 = limit motion;	B = good biomechanical studies exist;
3 = stabilize;	C = nonconclusive studies;
4 = unload;	D = no support or negative studies.
5 = miscellaneous.	

achieved through the use of external skeletal fixation with the halo apparatus in conjunction with a short or a long well-molded body cast. The long body cast offers a greater efficiency of immobilization because of greater purchase and leverage. The most effective device at present for immobilizing the entire spine is the halo pelvic apparatus.

■ CLINICAL BIOMECHANICS

■ The clinical problem is to apply forces to the spine in a manner that will somehow be therapeutic to the patient. The magnitude and resultant effects of these forces depend upon the biomechanics of the complex system.

■ Forces applicable to the spine from the outside are limited by skin discomfort and the stiffness of the structures through which they must be transmitted.

■ In a three-point fixation system, the middle force should be applied where the clinician wishes to obtain maximum correction or immobilization.

■ Selective discomfort imposed by the orthosis is one of the mechanisms through which motion is controlled.

■ For the most effective control of the degrees of freedom, the halo pelvic skeletal fixation, halo cast, halo vest, or a Minerva plaster jacket should be employed.

■ A well-molded plaster cast offers the best nonskeletal fixation against axial rotation of the spine.

■ In each case, consider the movement (i.e., flexion or extension) that must be prevented; then choose the orthosis accordingly.

■ Shoulder and thoracic support fixation adds to the unloading and general effectiveness of a cervical orthosis when the patient is standing but loses some of its effectiveness when the patient is recumbent. This difference in position as well as other trunk and shoulder movement can cause cervical spine motion when the halo vest is used.

■ In attempts to immobilize the upper cervical spine with the use of a cast or a brace, it should be remembered that complete immobilization is not possible. The patient's ability to talk and chew is inversely related to the effectiveness of the immobilization.

■ The Jewett hyperextension brace probably has some ability to shift weight from the anterior to the posterior elements of the thoracic vertebrae.

■ The use of an abdominal corset with the chairback brace may be useful in diminishing the loads applied to the lumbar spine.

■ A cast or orthosis that seeks to immobilize the lower lumbar spine, particularly the lumbosacral joint, should include at least one thigh as part of the fixation.

■ There may be a useful role for an adjustable trial brace to determine therapeutic position of the lumbar spine and as a trial of whether or not pain is relieved.

NOTES

[A] The forces. The law of equilibrium states that (a) the sum of the forces be equal to zero and (b) the sum of the moments be equal to zero. Therefore:

$$\text{(a)}\quad F_A - F_B - F_C = 0 \qquad (1)$$

$$\text{(b)}\quad F_B D_B - F_C D_C = 0 \qquad (2)$$

Solving these, we obtain:

$$F_B = \frac{F_A \cdot D_C}{D_B + D_C} \quad \text{and} \qquad (3)$$

$$F_C = \frac{F_A \cdot D_B}{D_B + D_C} \qquad (4)$$

Putting in the values of the example shown in Figure 7-5A gives

$$F_B = \frac{2F_A}{3} \qquad (5)$$

$$F_C = \frac{F_A}{3} \qquad (6)$$

Bending Moment Diagram. The bending moment diagram for the three-point force system is shown in Figure 7-5B. It is triangular in shape, and the maximum bending moment is equal to:

$$\text{Mmax} = F_A \cdot \frac{D_B \cdot D_C}{D_B + D_C} \qquad (7)$$

$$= F_A \frac{1}{\frac{1}{D_B} + \frac{1}{D_C}} \qquad (8)$$

Equation (8) clearly shows that the maximum bending moment is maximized by the largest values of D_B and D_C.

[B] Location of the thoracic pad. The goal of applying horizontal forces to the spine is to obtain maximum overall correction. Thus, for the three-point force system, the criterion for maximum angular correction is the angle change for the two vertebrae at the level of the two end forces. The engineering principles applicable here have to do with the deflection of a beam subjected to bending moments from applied forces. The principle states that the resulting angulation between the two points on a beam is proportional to the area of the bending moment diagram between those two points. By applying this principle and maximizing the area as a function of the location of the middle force, it is shown below that the optimum place for this force is midway between the two end forces.

Referring to Figure 7-5B, the three-point principle and its bending moment diagram, an equation for the area of the bending moment diagram can be written:

$$\text{Area} = (D_B + D_C)\,\text{Mmax}/2 \qquad (9)$$

Inserting the value for Mmax from equation (7) results in:

$$\text{Area} = \frac{F_A \cdot D_B \cdot D_C}{2} \qquad (10)$$

Furthermore, assuming that points B and C are given and that point A is varied to obtain the most efficient loading, it is possible to substitute for D_C:

$$D_C = D - D_B \qquad (11)$$

Where D equals $D_B + D_C$. Putting equation (11) in (10), differentiating Area with respect to D_B, and equating the expression to zero, the value of D_B for which Area is maximum is obtained:

$$D - 2D_B = 0 \qquad (12)$$

$$D_B = D/2 \qquad (13)$$

Therefore, for maximum angular change between the two end vertebrae, the point A should be located midway between B and C.

In the above analysis, the Milwaukee brace and the patient are modeled as a pure three-point force system. If the pelvic support is allowed to take up bending moments (which it always does to a degree), the above assumption and the conclusion are no longer true. For example, if the axillary sling is absent, the three-point force system degenerates into a cantilever system where the bending moment created by the thoracic pad is balanced solely by the pelvic support. It all depends upon the brace configuration (presence of the various pads), fit of the brace, and the patient's activity. In order to optimize these factors and thoroughly understand the mechanism that makes the Milwaukee brace so effective, additional biomechanical measurements on patients using the brace are required.

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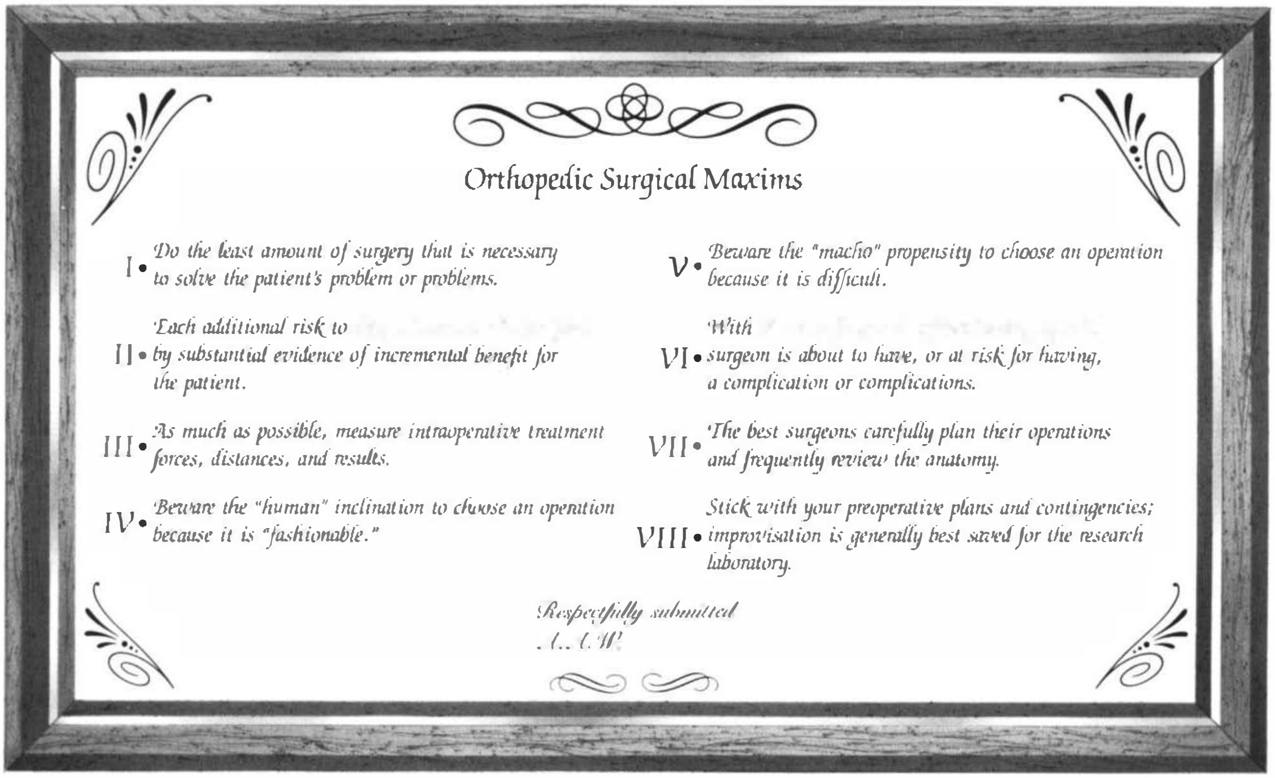
Biomechanical Considerations in the Surgical Management of the Spine



Orthopedic Surgical Maxims

I • Do the least amount of surgery that is necessary to solve the patient's problem or problems.

II • Each additional risk to [illegible] by substantial evidence of incremental benefit for the patient.

III • As much as possible, measure intraoperative treatment forces, distances, and results.

IV • Beware the "human" inclination to choose an operation because it is "fashionable."

V • Beware the "macho" propensity to choose an operation because it is difficult.

VI • With [illegible] surgeon is about to have, or at risk for having, a complication or complications.

VII • The best surgeons carefully plan their operations and frequently review the anatomy.

VIII • Stick with your preoperative plans and contingencies; improvisation is generally best saved for the research laboratory.

Respectfully submitted
A. A. W.

Presented by Augustus A. White, III, as part of the Presidential Address to the 16th Annual Meeting of the Cervical Spine Research Society, Key Biscayne, Florida, 1988.

8

Since the first edition of this text, the explosive interest in spine surgery has been associated with a vast increase in the volume of related information. This chapter will focus on the biomechanically relevant research and clinically applicable studies that impact significantly on the various facets of the surgical management of patients with spine and spine-related problems. Certainly a plethora of new implants for use in spine surgery has developed. Some cogent information and guidelines will be offered to help the reader evaluate and select among the numerous options.

The major goal of the surgeon is to utilize available scientific and clinical information to ensure that the risk-benefit scales weigh distinctively in the patient's favor.

PART 1: SURGICAL DECOMPRESSIONS

Decompression is indicated in the presence of an incomplete or progressive neurologic deficit in which there is clinical evidence of pressure or encroachment on the spinal cord associated with tumor, trauma, infection, degenerative changes, or a variety of miscellaneous developmentally related situations.

The goal of decompression is to effectively remove abnormal spinal cord or nerve root impingement with the least possible surgical risk and the least disruption of the structural integrity of the spinal column. The first consideration is to localize the site of the abnormal impingement. The compression may be anterior, posterior, lateral, or some combination. The offending structure(s) may be located in the midline, laterally, or both; at the interspace, behind the vertebral body, or both. Obviously, there are a number of possible combinations. The sources of compression may also be mixed or poorly localized. There are several accepted surgical procedures that may be employed to decompress the spinal cord and/or nerve roots. Each case should be carefully evaluated and the proper surgical procedure chosen.

Clinical evaluation is carried out to locate the source of abnormal pressure as accurately as possible. The history and neurologic examination are helpful; however, the determination of location is based mainly on imaging studies. All of the following studies need not and cannot always be carried out; however, using several in each case aids in localizing the source of impingement: laminagrams; radiopaque myelography; air contrast myelography, computerized axial tomography, and nuclear magnetic resonance imaging. The importance and availability of CT scanning and MRI imaging in diagnosis and preoperative planning compels us to recommend the anatomic work of Wolfgang Rauschning for review by the spine surgeon.[239] Clinical judgment and the equipment available dictate the combinations of studies that are used for a particular patient.

Generally, when the pressure is anterior, the decompression of choice should be anterior. Similarly, posterior decompression is generally best for relieving posterior pressure. If the surgeon decompresses the spinal cord anteriorly when the pressure is posterior, or vice versa, the procedure may be ineffective in relieving the pathomechanical problem, and the patient's neurologic status may not improve.

When the offending structure is between the vertebral body and the spinal cord, it may be necessary to remove all or part of the vertebral body to decompress the lesion. Posterior decompression of the spinal canal, even with dentate ligament transection, may not relieve anterior impingement. This observation was made by Verbiest in post-traumatic situations,[302] and it applies to most anterior encroachments.[328] There is controversy about the advisability of transecting the dentate ligaments. Since the work of Kahn[149] in 1947, there has been considerable attention directed to the assumption that transecting the dentate ligaments reduces spinal cord pressure because the dentate ligaments hold the cord anteriorly in the canal. Tunturi[297] showed that the dentate ligaments in dogs are elastic and prestressed. Stolzman and Blackworth[285] indicated that the dentate ligament restrained the cephalocaudal movement of the cord, but not the anteroposterior movement. Cusick and co-workers[59] monitored somatosensory responses in dogs with experimentally produced posterior cord elevations. They found that the dentate ligaments were the most significant elements affecting tension in these spinal cords, as evidenced by both mechanical and somatosensory response studies. These findings suggest that sectioning the dentate ligaments distributed the tension over a greater segment of the cord, with a reduction in tension and an improvement in axonal conduction at the level at which the experimental force was applied. As an important complement to the animal investigations, human cadaver studies showed a 50% reduction of force after dentatotomy. More recently, the clinical work of Miyazaki and Kirita[199] shows that in ossification of the posterior longitudinal ligament, the dentate ligaments do in fact hold the cord forward in apposition to the ossified posterior longitudinal ligament.

The issue of anterior versus posterior decompression remains controversial and unresolved. Nevertheless, logic dictates that if there is pressure anteriorly, removing structures posteriorly is unlikely to relieve the anterior pressure. Conversely, removing structures anteriorly is unlikely to decompress pressure that is being applied posteriorly (Fig. 8-1). Although the sagittal plane diameter may be the most important factor, hypertrophic or inflamed facet joints can also compromise the frontal plane diameter and thus diminish the cross-sectional area of the spinal cord. It is appropriate nevertheless to concentrate on the anterior and posterior structures. Spurs and herniated discs are compelling reasons for anterior resection, while invaginating yellow ligaments and developmentally narrow canals demand consideration of posterior decompression. Moreover,

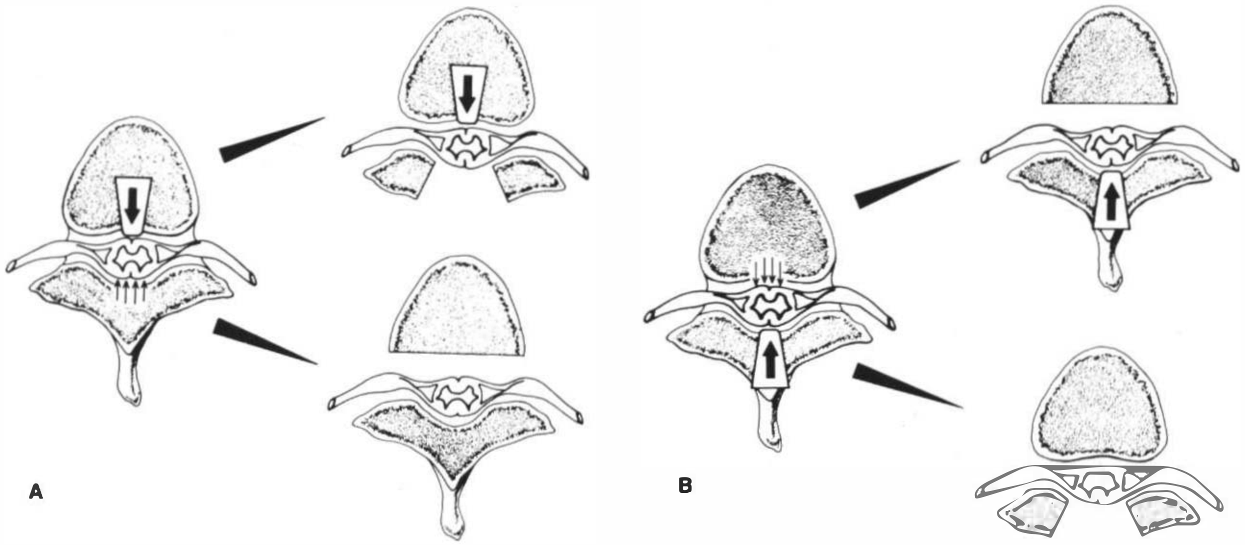


FIGURE 8-1 An illustration of the concept of contrecoup compression of the spinal cord and the possible methods of decompression. **(A)** Here, the primary compression is anterior, and the contrecoup compression is posterior. Posterior decompression solves only the secondary source of pathology. Anterior decompression resolves both sources of compression on the spinal cord. **(B)** This is the converse situation. Removal of the anterior secondary compression does not solve the problem, but removal of the primary lesion posterior to the spinal cord permits full decompression. These concepts point out the importance of preoperative localization of the primary lesion causing compression whenever possible. This is greatly facilitated with the use of computerized axial tomography.

when both anterior and posterior structures are impinging on the spinal cord, anterior and posterior decompressions should be considered.

A review of some cogent studies that relate to this critical issue follows. Tencer and colleagues,[288] using an ingenious *in vitro* experimental design, were able to provide compelling mechanical evidence. The experimental setup involved a microload cell and displacement transducer inserted through fresh human cadaver vertebral bodies and placed in apposition to the anterior spinal cord. This study showed clearly that the dura is tethered to the anterior canal. Laminectomy did not decrease the anterior pressure in the presence of 35% occlusion of the canal. The investigators also showed that in order to decrease anterior pressure, there must be some decompression anteriorly. It is important to point out, however, that this study was in the thoracic spine, where normal kyphosis is present. One cannot be certain that these findings apply to the cervical spine, where there is a normal lordosis.

There are some interesting experimental studies using benign tumors that better simulate the slow development of pressure that usually occurs in cervical spondylotic myelopathy. Ushio and colleagues[301] put spinal cord tumors anterior to the thoracic spinal cord in rats and observed tumor growth and associated neurologic changes. The neurologic symptoms were relieved by chemotherapeutic agents and by radiation, but they were not relieved and were not helped by laminectomy. In another study by Bennett and McCallum,[13] experimental tumors were implanted anteriorly in one group of cats and posteriorly in another group. The tumors were allowed to grow enough to produce histologic, neurologic, and electrophysiologic changes. They found that laminectomy helped the function of the animals with the posteriorly placed tumors but did not help those with the anteriorly placed tumors.

The evidence continues to accrue to help make a valid decision about this important issue, but it is not yet definitive. Two recent clinical studies add to the controversy. Nicholls and Jarecky[213] published a 20-year experience with decompression laminectomy for tumors of the cervical, thoracic, and lumbar spine in 38 patients. Seventy percent of these patients received no benefit from surgery. Hukuda and co-workers[136] reported an extensive clinical study of 191 patients operated on for cervical spondylotic myelopathy (CSM). They compared in a substantial number of patients the results of anterior and posterior procedures. Posterior operations gave better results in the more advanced myelopathies, such as transverse lesions and the Brown-Séquard syndrome. The brachialgias and central cord syndrome were satisfactorily treated with anterior techniques. There was no overall superiority of either the anterior or the posterior approach. These two studies are not comparable, but they are cogent to the issue of decompressive laminectomy.

DECOMPRESSION IN THE CERVICAL REGION

Some of the most dire neurologic consequences occur from spinal cord injuries in this region. The cervical spine is accessible surgically, either anteriorly or posteriorly. An interesting and general scholarly view and historical perspective of cervical spine surgery has been presented by Fielding and Hensinger.[81]

Anteriorly Located Compression

There are a number of sites in which the spinal cord may be compressed anteriorly. Generally, some appropriate *anterior* decompression should be employed.

Anterior Midline Compression

The transoral approach to the upper cervical spine[20, 192] is sometimes necessary for decompressions that cannot be successfully completed with the generally less complicated prone posterior approach. When the offending structure is between the dens and the spinal cord/brain stem, or when the structure is a part of the dens or the dens itself, the transoral exposure and excision may be the procedure of choice. Callahan and colleagues have emphasized the importance of pannus that may be present behind the dens on the anteriorly subluxed atlas in the rheumatoid patient.[40] We suggest careful imaging with MRI or CT myelography to look for pannus at and behind the dens. If a subluxated rheumatoid atlas is reduced posteriorly, there is a risk that a portion of callus could displace and impinge upon the spinal cord. Figure 8-2 shows a situation in which a large lump of pannus behind the dens is causing effacement of the spinal cord.

Sakou and co-workers[254] indicate that removal of the dens through the transoral approach can be

FIGURE 8-2 (A) This sagittal reconstruction of a CT scan at the level of the dens shows pannus (arrow) at and behind the dens pushing the spinal cord posteriorly. (B) This CT scan at approximately the same level shows effacement of the spinal cord by the pannus, which cannot be seen but is indicated by the arrow. *(Courtesy of Charles Clark, M.D., and Joseph M. Mirra, M.D.)*

made easier with the use of modern, high-speed drills. They also suggest that the surgeon leave a thin portion of the body of C2 so as to prevent the anterior migration of the posterior longitudinal ligament into the operative field. This provides additional protection to the spinal cord. If needed, an anterior C1–C2 fusion can be done at the same time.

Pressure may be exerted on the anterior midline portion of the cord at the level of the interspace. This may be caused by a "hard disc" (primarily an osteophyte), a "soft disc" (primarily the annulus fibrosus), tumor, trauma, or infection. We believe that the Smith-Robinson procedure is the operation of choice for anterior midline pathology located at the level of the interspace because it gives adequate exposure and provides a sound surgical construct for postoperative stabilization (Fig. 8-3, Part 2A).[268] The surgical constructs for anterior fusion are analyzed on page 547. The Bailey-Badgley[9] and the Cloward[50] procedures are effective in these cases, although we believe that they are somewhat more extensive than is necessary for anterior midline pathology limited to the region of the interspace. The Cloward procedure may not be such a stable construct (Fig. 8-3, Part 2C).[153] A modification of the three procedures may be required for anterior decompression of the spinal cord. The entire vertebral body is resected all the way back to the posterior longitudinal ligament and dura mater for full visualization and decompression.

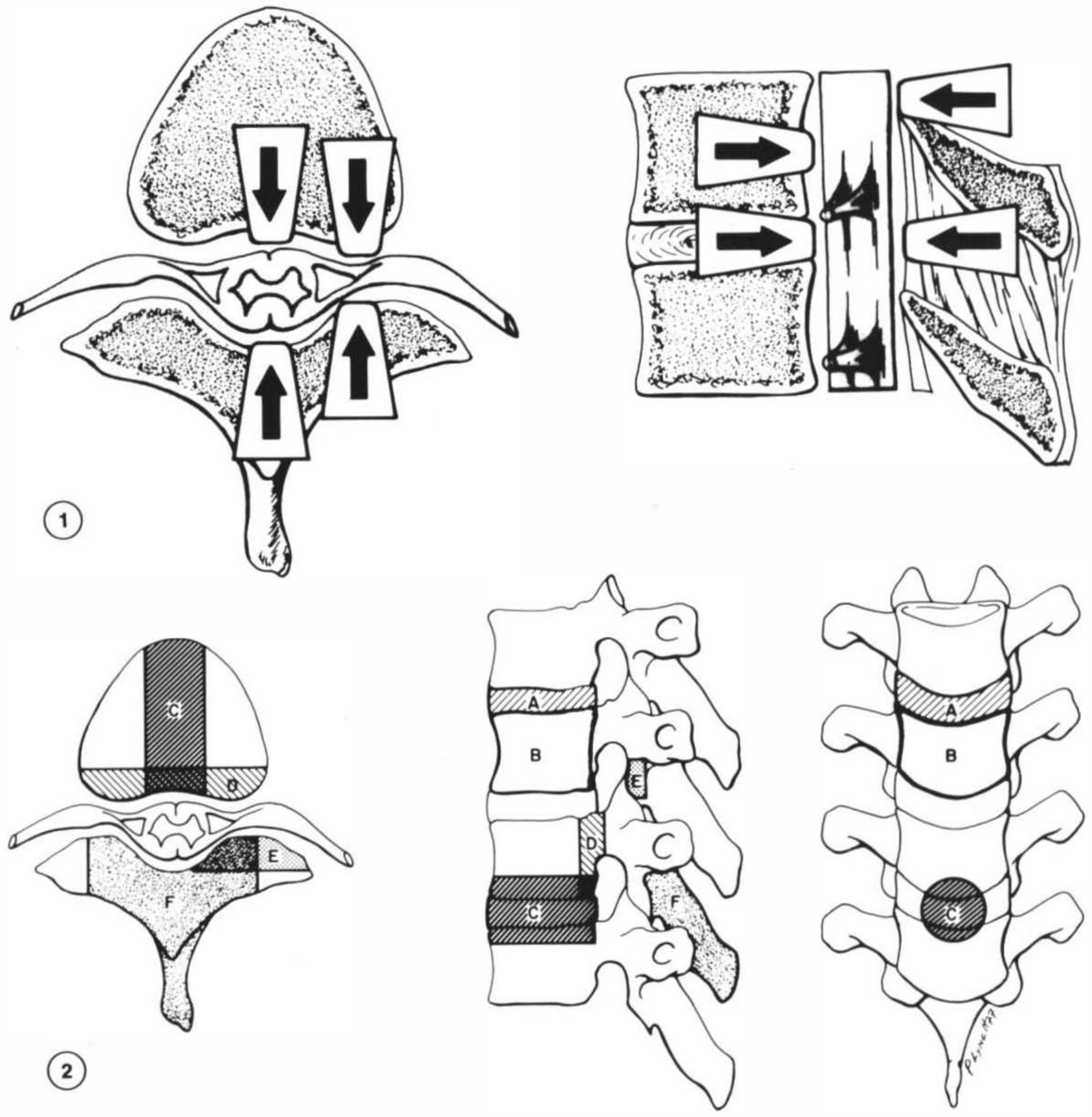


FIGURE 8-3 Part 1. Compression sites. This illustrates the basic sites of spinal cord and nerve root compressions. The anterior compressions may be at the interspace level or up behind the vertebral body. They may also be midline or lateral. The same is the case for the posterior compressions. In selecting the surgical technique for decompression, it is useful to first determine as accurately as possible the site of the compression. **Part 2.** Multiple decompressions. *(A)* The Smith-Robinson approach (removal of the intervertebral disc) decompresses the anterior cord and nerve root at the interspace level (see Fig. 8-31). *(B)* Vertebral body resection decompresses the anterior cord and nerve roots (see Figs. 8-38). The discs above and below are included, which provides even wider exposure and decompression. *(C)* The Cloward or dowel resection, when carried back through the posterior longitudinal ligament by careful dissection, exposes the central portion of the cord behind the vertebral body. The lateral areas are exposed at the interspace by removal of remaining disc material. *(D)* An anterior decompression that begins at the neural foramen and removes only the most posterior portion of the vertebral body. This leaves as much of the anterior structures as possible to maintain clinical stability (see Fig. 8-41). *(E)* This is the keyhole laminotomy and/or the facetectomy or posterior nerve root decompression (see Fig. 8-5). Keyhole laminotomy is suggested only for a soft, anterolateral cervical disc that can be removed by the cephalad or caudad retraction of the nerve root. *(F)* Bilateral laminectomy (total laminectomy) for posterior decompression of the cord.

When the lesion causing the pressure is cephalad or caudad to the level of the interspace, additional considerations become important. Such a lesion may result from the same pathologic conditions mentioned above or from ossification of the posterior longitudinal ligament. Surgery for this problem is discussed in the section on the management of cervical spondylotic myelopathy. The Cloward and the Keystone procedures are also useful when only limited access to the space behind the vertebral bodies is needed (Fig. 8-3, Part 2C). Another procedure that can be useful in this situation is vertebral body resection, especially when extensive exposure and good visualization of the cord are important.[146] This is also shown in Figure 8-3, Part 2B.

Another choice should be mentioned. If for some reason an anterior approach is not possible, then laminectomies at two or more contiguous levels may be helpful. This may alleviate some posterior contrecoup compression.

Anterolateral Compression

Anterolateral pressure limited to the interspace can be caused by protruding joints of Luschka, a hard or soft disc, tumor, trauma, or infection. The Smith-Robinson procedure is the one we recommend in this situation. Snyder and Bernhardt[270] recently described results of limited discectomy for the treatment of cervical radiculopathy for either "hard" or "soft" disc disease (Fig. 8-4). The procedure called *anterior fractional interspace decompression* by the authors is reported to give good or excellent results in 64% of the entire group and 70% of patients not on workers' compensation. These are not outstanding results. However, the procedure is interesting because of the application of the concept of limited discectomy to preserve biomechanical function and avoid fusion and also as an alternative to the treatment of disc herniation next to a fused segment. In our view, this procedure merits additional study. We believe that for a well-documented "soft disc," the keyhole laminotomy is also a useful operation (see Fig. 8-3, Part 2E, and Fig. 8-5).

Sometimes, an anterolateral lesion may extend or be entirely located behind the vertebral body. This may be caused by the same diseases described above. In these situations, we suggest a vertebral body resection. If the clinical problem is such that a somewhat limited exposure would suffice, then the previously described modification of the Bailey-Badgley, the Cloward, or the keystone procedure is a good choice.

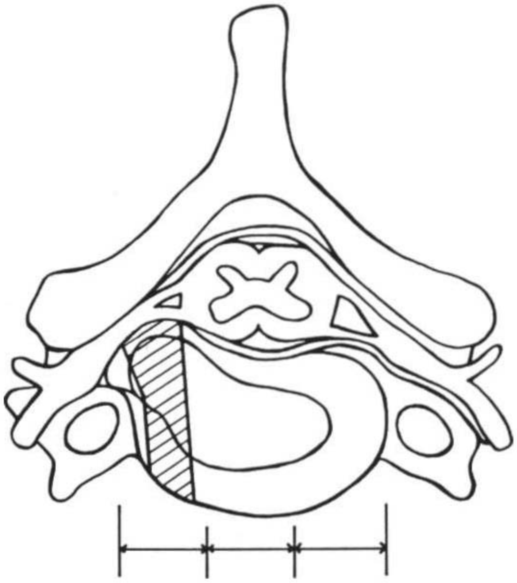


FIGURE 8-4 With magnification and the use of a high-speed burr, the tunnel to the herniated disc is made as indicated. The disc is removed with angled curettes and 2-mm rongeurs. The region is carefully explored, and all fragments are removed. *(Snyder, G. M., and Bernhardt, M.: Anterior cervical fractional interspace decompression for treatment of cervical radiculopathy. A review of the first sixty-six cases. Clin. Orthop., 246:92, 1989.)*

Anterior Midline and Lateral Compression

When the lesion is at the interspace only, the Smith-Robinson procedure is the treatment of choice. An isolated anterior and lateral disc compression limited to the level of the disc can be treated by the united discectomy of Synder and Bernhardt.[270] This is shown in Figure 8-4. Theoretically, this technique preserves some of the biomechanical integrity of the functional spinal unit (FSU). If there is extensive disease behind the vertebral body, then vertebral body resection is the treatment of choice. Laminectomy at multiple levels is a secondary choice when the anterior approach is not possible.

Posteriorly Located Compression

Posterior pressure on the cervical spinal cord may come from a variety of conditions. There are the standard causes—tumor, trauma, a developmentally small spinal canal, and infection. Additional conditions include yellow ligament encroachment, spinal stenosis, and laminectomy membranes. When the

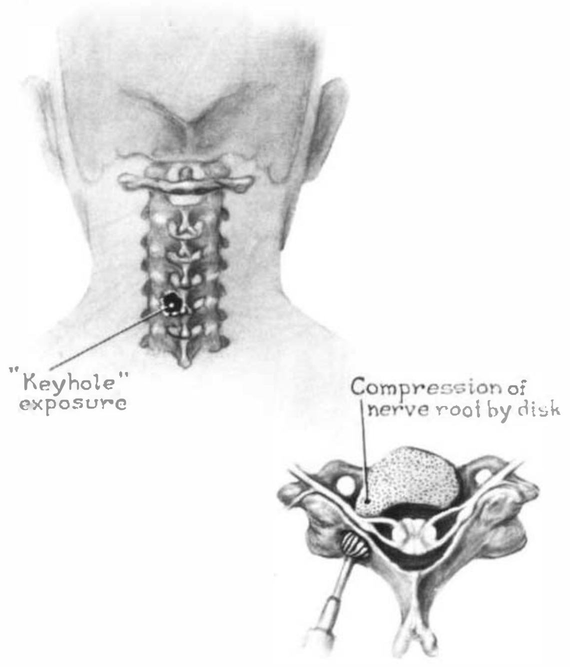


FIGURE 8-5 This limited laminotomy is adequate for removal of a herniated annulus fibrosus. In the cervical spine the roots pass through the intervertebral foramen approximately at right angles to the spinal cord. Thus, the roots should be carefully retracted cephalad or caudad, rather than medially as in the lumbar spine. *(Robinson, R. A., and Southwick, W. O.: Surgical approaches to the cervical spine. In American Academy of Orthopaedic Surgeons: Instructional Course Lectures. vol. 17. St. Louis, C. V. Mosby, 1960.)*

offending disease is posterior, the decompression should be posterior. When the source of compression is posterior, its relationship to the interspace (along the y-axis) is not surgically important. This is because of the relative ease of adjustment of the level of laminectomy and its extension cephalocaudally. However, with the anterior approach there is a significant difference between cutting through the disc space and cutting through the vertebral body. It is also important to remember that multiple laminectomies[A] in the cervical spine, even in the presence of intact facet articulations, may jeopardize clinical stability in children.[43] In other words, the surgeon should not decompress any more extensively than is necessary. Decompression of more than 50% of the structural integrity of the facet articulations puts the segment at risk, even in adults (see Chap. 5).

Posterior Midline Compression

This may be adequately decompressed with a single- or multiple-level bilateral laminectomy, as needed (Fig. 8-3, Part 2F).

Posterolateral Compression

Compression here may be relieved by a keyhole laminotomy, a unilateral laminectomy, or a laminectomy with facetectomy or nerve root decompression.

Posterior Midline and Lateral Compression

For this condition, we recommend the bilateral laminectomy or multiple bilateral laminectomies.

Anteriorly and Posteriorly Located Compression

There are situations in which the cord is compressed at or near the same level, both anteriorly and posteriorly. In addition to tumor, trauma, and infection, there are the pincer mechanisms[233] and combinations of hard and soft discs associated with yellow ligament encroachment.

Anterior and Posterior Compression, Limited to the Interspace

This may be caused by either the pincer mechanism or combined yellow ligament and disc encroachment. For osteophytes impinging on the anterior or anterolateral portion of the cord and nerve roots associated with a yellow ligament impinging posteriorly, the Smith-Robinson procedure is an effective construct. The technique permits removal of anterior spinal cord impingement, and by spreading the interspace, it reduces the yellow ligament encroachment and increases the longitudinal (y-axis) diameter of the neural foramen.

The pincer phenomenon occurs as a primarily translatory displacement in the sagittal plane.[233] This is shown in Figure 8-6. In some instances there has been extensive displacement, and the cord damage is due to the initial impact at the time of injury rather than the residual canal encroachment. In this type of situation, correction of the encroachment is unlikely to be helpful with regard to spinal cord recovery. However, with reduction and/or decompression, there may be some nerve root recovery. If the trauma is not acute, or if there is reason to assume that the neurologic problem results from the residual encroachment rather than the initial im-

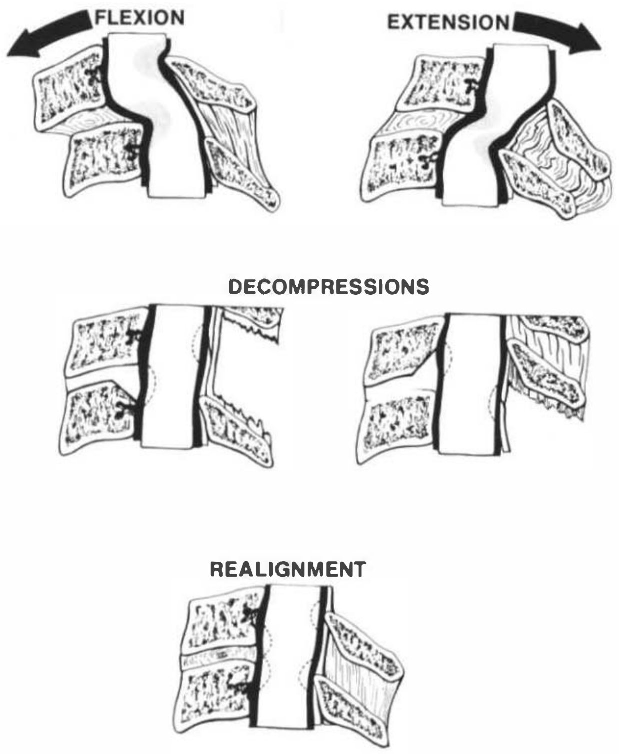


FIGURE 8-6 The pincer phenomenon may be associated with different patterns of cord encroachment, depending upon whether the displacement at the level of disruption is in a flexion or an extension mode. Although the indications for decompression are controversial, this figure is intended to point out some of the mechanical factors that are important. When indicated, adequate decompression of a pincer problem may require a posterior approach, possibly in conjunction with an anterior procedure. Realignment may also be as satisfactory in some situations.

pact, then one solution to the problem is to reduce the displacement and thus regain the original spinal canal diameter. It is very important in acute unilateral and bilateral facet dislocations to obtain an MRI prior to operation or closed reduction to be certain that there is no cervical disc herniation that could cause neurologic deficit at the time of reduction.[6a] This may be done through axial traction or open reduction in the case of fixed facet dislocation. If there is evidence that a displaced disc or bony fragment causes compression anteriorly or a bony fragment causes compression posteriorly, an appropriate decompression should be carried out. If the situation is such that the pincer mechanism is thought to be the only pressure-exerting pathology in an irreducible injury, we suggest a posterior decompression, which provides the opportunity for open reduction, if necessary. This could then be followed by anterior decompression should it be necessary.

Eyring and colleagues reported a case in which anterior and posterior decompressions and fusions were done under the same anesthesia with the patient in a sitting position.[73] Reasons for the decision to decompress as well as the surgical technique are presented in this chapter. A study of the evaluation of clinical stability in Chapter 5 shows that, in all probability, a pincer phenomenon requiring decompression is more likely than not to fit the indications of a clinically unstable situation, which should be managed accordingly.

Compression Behind a Vertebral Body, Which May or May Not Be at the Level of the Interspace

The possible causes include tumor, trauma, infection, and spinal stenosis.[210] The reasonable options are vertebral body resection, multiple bilateral laminectomies, or combined anterior and posterior decompressions.

Mixed or Poorly Localized Compression

This condition may occur as a result of any combination of the entities mentioned previously. If the indications for decompression are present but cannot be well localized, we suggest vertebral body resection or combined anterior and posterior decompression. If the situation is associated with fracture, reduction and realignment may be useful.

Biomechanical Considerations in the Management of Chronic Cervical Spondolytic Myelopathy (CSM)

Because of the increase in the number of elderly patients and the high prevalence of both CSM and myelopathy secondary to ossification of the posterior longitudinal ligament (OPLL), we considered it useful to have a section of this chapter specifically address this topic.

Anatomic and Pathophysiologic Considerations

Canal size is a very important consideration. Wolf and associates[330] measured over 200 adults and suggested that an anteroposterior diameter of less than 10 mm in the presence of posterior spurs is likely to be associated with spinal cord compression. A more recent publication by Kallen and co-workers[150] indicated the importance of the sagittal diameter in CSM. This work emphasized the following measurements: the DAD and the SAD. The DAD is the developmental anteroposterior diameter, and the SAD is the spondylotic anteroposterior diameter (see Fig. 5-11). This study suggests that by subtracting the SAD from the DAD, one gets the SSI, which is the segmental stenotic index. Individuals with SADs of 14.8 mm and an SSI of 1.5 mm or more may be at risk for developing CSM. The investigators found that patients with CSM have narrower cervical canals than normals. It was also noted that normal individuals and CSM patients have the greatest sagittal plane narrowing at the C5–C6 level. Statistical significance was not demonstrated. The work of Edwards and LaRocca[68] has supported the observation that a narrow canal in association with a posterior osteophyte in the canal is a mechanical factor of major clinical significance in CSM. Table 8-1 provides a synopsis of their analysis. Ogino and colleagues, in a study of nine autopsies, concluded that the presence of a developmentally narrow (small compression ratio) spinal canal was the most significant factor in CSM.[218] There are several structures in addition to a posterior osteophyte that may compromise available space and impact the spinal cord. These include a bulging or herniated annulus fibrosus or nucleus pulposus. Yellow ligaments that have lost elasticity may invaginate into the canal, take space, and irritate the spinal cord. Degenerated hypertrophic facet joints as well as uncovertebral joints can compromise canal space and contribute to CSM.

The disc and the yellow ligament in the human spine are at about the same vertical (y-axis) level. Therefore, if both are protruding into the canal maximally, as with extension of the neck, there is a likelihood of significant spinal cord irritation. Also important in the clinical pathology and surgical biomechanical considerations in CSM is the factor of motion, both normal and abnormal. Kallen and colleagues[150] pointed out the importance of the fixed subluxation. In addition, there is a dynamic situation known as the pincer phenomenon in which there is a guillotine effect on the cord (Fig. 8-6). There is also the theoretic possibility that cumulative microtrauma, from loads applied to the spine, may contribute to the pathologic processes.

TABLE 8–1 Cervical Developmental Segmental Sagittal Diameter

10 mm or less—with cervical spondylosis, likely to have myelopathy
10–13 mm—may be considered premyelopathy group
13–17 mm—tendency for symptomatic spondylosis
Greater than 17 mm—group less prone to develop disease

Gooding and co-workers[101, 102] simulated CSM in dogs and showed that spinal cord blood flow was reduced. The conclusion was that vascular compromise is a major factor in CSM. However, Korhine and colleagues,[160] from their spinal evoked response studies in monkeys, expressed the view that mechanical pressure was more important. It seems logical that both factors are significant contributors to the pathophysiologic processes. Fujiwara and co-workers[95] were able to show a distinct correlation between clinical severity and the reduced cross-sectional area of the cervical spinal cord with microscopic pathologic changes. Moreover, the study showed a statistically significant correlation between increased cross-sectional spinal cord area and improvement following surgery. The compression ratio as defined by these investigators was also associated with these variables, but to a lesser extent (Fig. 8-7).

In summary, it may be useful to think of the major mechanical factors in the pathophysiology of CSM

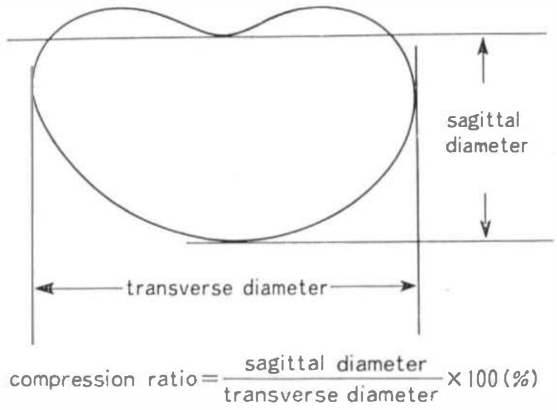


FIGURE 8-7 The compression ratio measurement provides a good index of spinal cord pathology. A compression ratio of 40% is considered abnormal. The cross-sectional area is a more reliable indicator of both spinal cord damage and associated clinical deficits. A cross-sectional area of 40 mm^2 or less is considered abnormal. *(Fujiwara, K. et al.: An analysis of the factors prognosticating therapeutic results of cervical myelopathy. Cervical Spine Research Society. Palm Beach, FL, 1986.)*

Static Culprits

Small cervical canal
Osteophytes
Disc herniation
Ossification of posterior longitudinal ligament
Deformed uncovertebral processes
Apophyseal joint deformation or inflammation
Hypotonic yellow ligament

Dynamic Culprits

Motion—normal and abnormal
Loads—normal and abnormal
Mechanical properties of spinal cord
Mechanical properties of spinal column
Hypotonic yellow ligament

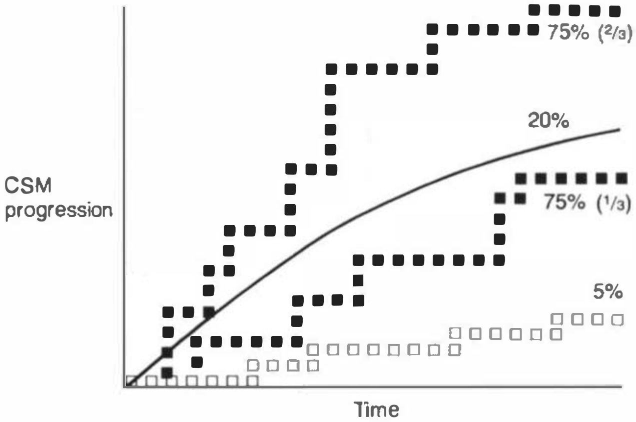


FIGURE 8-8 This graph shows the progression of severity of cervical spondylotic myelopathy with time. About 5% of patients experience a slow progression with superimposed periods of a slight increase in progression. This is indicated by the open squares. Approximately 20% present a moderate rate of progression that is smooth and gradual and more or less constant. This is represented by the uninterrupted line. The remaining 75% are represented by the black squares. One third of this group progresses moderately, and the other two thirds progress more rapidly. Both of these groups have superimposed rates of rapid acceleration.

as being separated into two groups—the static culprits and the dynamic culprits (see display). The major factor in CSM is a narrow canal with osteophytes. In addition, there are other possible causes of spinal cord compromise, namely, the intervertebral disc, the yellow ligament, the uncovertebral processes, the intervertebral joints, normal and abnormal motion, and the physiologic and abnormal forces. Any combination of these factors may cause the initiation and progression of CSM. However, the salient static measurement that has been shown to be correlated with the important clinical and pathophysiologic variables is the cross-sectional area of the spinal cord.[95]

An important consideration in the clinical management decision making about the CSM is the natural course of the disease. Unfortunately, there is very little well-documented information on this topic. LaRocca[169] reviewed the literature and summarized the major instrumentation. The pattern appears to be one of progression, with several different schedules of acceleration (Fig. 8-8).

Surgical Procedures for Cervical Myelopathy

In 1980, a presentation was made at the annual meeting of the Cervical Spine Research Society in Palm Beach, Florida. The title was "A Comparative Study of Spinal Canal Enlargement and Laminectomy in the Cervical Spine."[123] The laminoplasty operation with several modifications has become the procedure of choice for myelopathy secondary to ossification of the posterior longitudinal ligament in Japan. Most of these procedures are shown in Figure 8-9, from the work of Tsuyama.[296] A recent modification developed by Itoh and Tsuji[140] has demonstrated the efficacy of this procedure using the laminoplasty and a small bone graft to stabilize the spine and ensure that the canal enlargement is maintained. The enlargement of the canal is important; however, a more critical factor is the increase in the cross-sectional area of the spinal cord.[95] An enlargement of the cross-sectional area of this canal does in fact result in a corresponding enlargement of the cross-sectional area of the cord, which is more important and is the critical variable. Despite previous discussion about anterior versus posterior decompression, there is no doubt that, in some situations, decompression can be achieved with laminoplasty or laminectomy (Fig. 8-10).

Recent interesting work by Hukuda and colleagues is relevant to this discussion.[137] These researchers carefully compared, in a controlled prospective five-year follow-up, the results of laminoplasty and laminectomy. Ten patients with laminectomy and 18 patients with French door–type[136] laminoplasties were studied. There was no superiority of laminoplasty over laminectomy in CSM in regard to functional recovery and enlargement of the epidural space. Also, there was no difference in the occurrence of kyphosis or instability

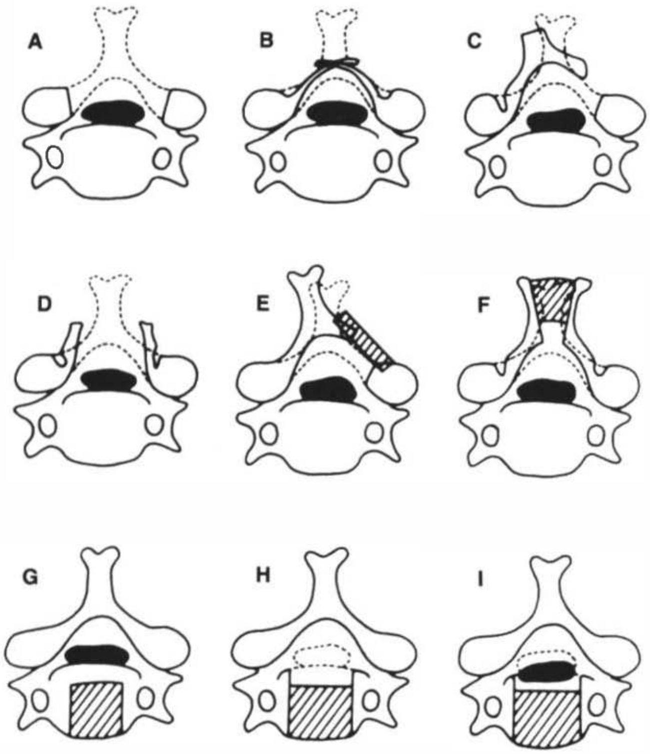


FIGURE 8-9 This shows the basic surgical procedures for treatment of ossification of the posterior longitudinal ligament. In addition to laminectomy, there are several laminoplasty procedures as well as anterior fusions with and without decompression. *(From Tsuyama, N.: Ossification of the posterior longitudinal ligament of the spine. Clin. Orthop., 184:71, 1984.)*

postoperatively. One difference was noted in the two groups—those patients who had laminoplasties had a reduction in neck extension. The investigators noted that there was a correlation of positive results with postoperative size of the spinal cord but not necessarily the spinal canal.

If laminectomy is chosen, it can be done in the traditional manner or in association with foraminotomy and removal of osteophytes from a posterior approach, as described by Epstein and associates.[72] Stability can be assured with one of the facet fusion procedures described by Robinson and Southwick[242] and Johnson and Southwick.[146]

The various anterior procedures provide access for removal of the anterior pressure and stabilization at the same time. In regard to the Smith-Robinson procedure, there is a theoretic consideration that deserves attention. It relates to the theory of Brieg and associates,[26,29] which states that tension is an important issue in the pathophysiology of pressure phenomena in the spinal cord. Based on this consideration, it may be advisable, when using the Smith-Robinson procedure, to use a bone graft 4–5 mm high. A larger graft carries the theoretic risk of adding tension to the region where the spinal cord has already reached its physiologic elastic limit. Other more extensive anterior decompression procedures to be considered involve full vertebral spondylectomy followed by iliac crest or fibular grafting procedures. Here, too, overdistraction should be avoided.

Yonenobu and colleagues[333] compiled a large clinical study of 95 patients with multisegmental CSM. These patients were treated as follows: 24 with extensive laminectomy, 50 with anterior interbody fusion (Cloward or Smith-Robinson), and 21 with subtotal spondylectomy. At a $p < 0.01$ level of significance, the subtotal spondylectomy provided the best clinical outcome for one-, two-, and three-level disease. When more levels were involved, laminectomy was better. The partial spondylectomy was thought to be better than the discectomy and interbody fusion because of the greater extent of canal enlargement it provides. Although other investigations have not noted problems with laminoplasty, these authors attributed late (12 months postoperation) neurologic deterioration to postlaminectomy instability. Oiwa and associates[220] completed experimental studies on dogs that implied that postlaminectomy cervical myelopathy deterioration *in patients* could be due to adhesion of scar to the spinal cord. The study showed that epidural fat served to prevent subarachnoid adhesions.

Discussion

This presentation does not address the indications for surgery in CSM. However, it is appropriate to mention that the best results occur in patients with moderate disease who have had clinical evidence of neurologic involvement for 1 year or less.

This selective review of the literature provides some useful guidelines based on substantial, albeit not definite, knowledge. Our interpretation suggests the following.

Laminectomy and laminoplasty probably do not adequately decompress anterior pressure contact points. However, in some patients there is an increase in the cross-sectional area of the spinal cord following laminectomy or laminoplasty.[95] Moreover,

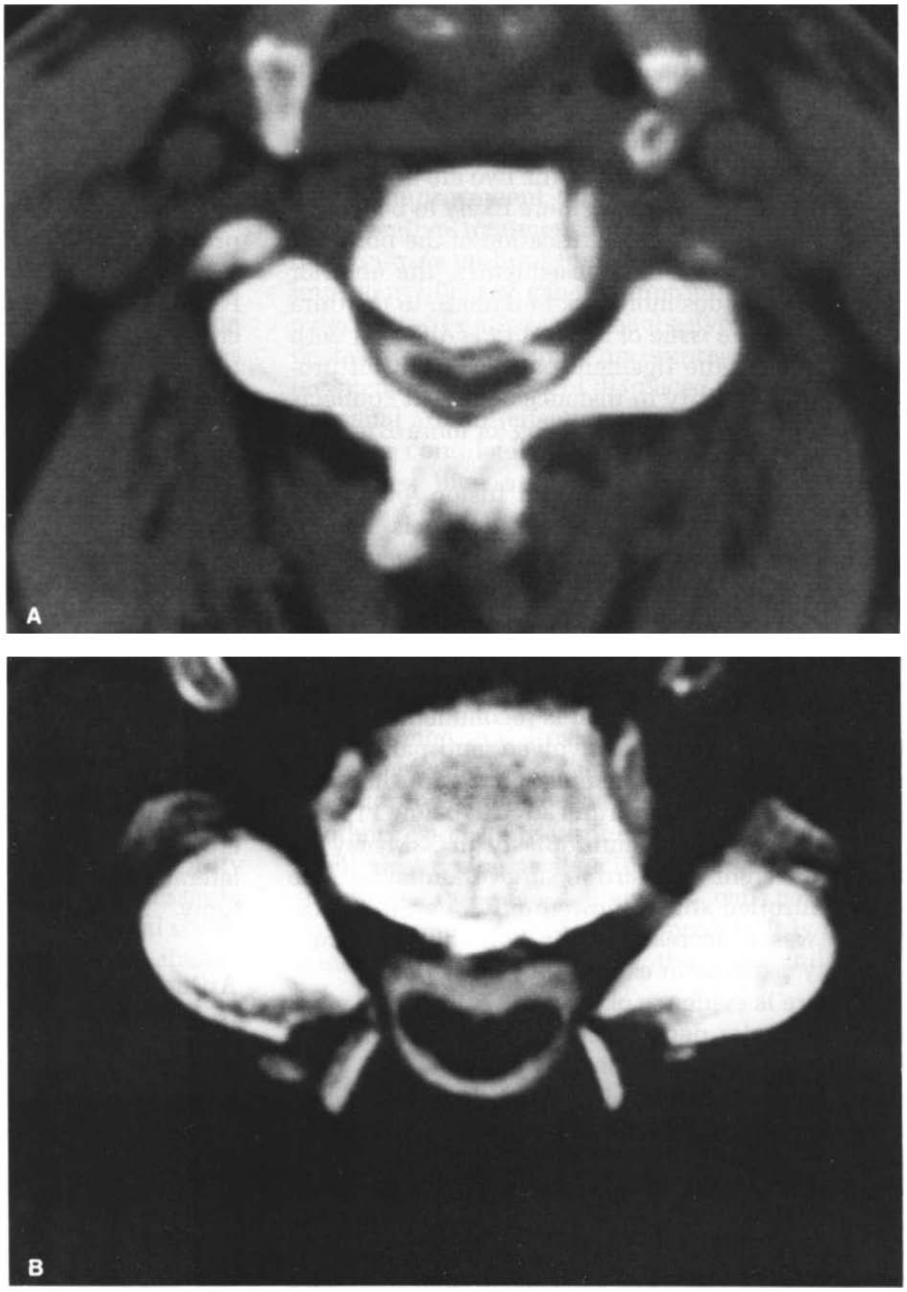


FIGURE 8-10 The compression ratio and the cross-sectional area of the cord can be changed with laminectomy. **(A)** Severely compressed cord with anterior osteophyte and developmental stenosis. **(B)** The same patient and level following French Door laminoplasty. This decompression could also have been achieved with a simple laminectomy. (From Hukuda, S., Mochizuki, T., Ogata, M., et al.: *Operations for cervical spondylotic myelopathy. A comparison of the results of anterior and posterior procedures. J. Bone Joint Surg.* 67B:609, 1985.)

when there is a developmentally narrow canal and/or more than three levels to decompress, laminectomy or laminoplasty should be considered. Laminoplasty has more assurance against instability but causes the loss of some extension. These procedures should be accompanied by a fat pad graft.

In a developmentally small canal associated with spondylotic myelopathy, there should be consideration of some procedure to enlarge the canal. A laminectomy or laminoplasty is probably advisable in these circumstances. In the face of changes like those shown in Figure 8-10, it is difficult to assert that posterior decompression is of no use in CSM associated with a stenotic cervical canal. In our opinion, the studies reported here suggest the following guideline: when CSM is associated with a

DAD of 13 mm and an SAD of 11 mm, laminectomies or laminoplasties should be considered if there are clinical indications for surgery.

Although there are similarities between ossification of the posterior longitudinal ligament and CSM, the surgical managements of the two are not exactly the same. In OPLL there is more likely to be attachment of the dura to the ossification of the posterior longitudinal ligament. Consequently, the anterior approach provides more risk of damage to the dura mater. Also, the issue of instability is different with OPLL because the ligament ossification itself provides some stability in instances where it connects adjacent vertebrae, thus allowing for more extensive posterior decompressions.

Anterior decompression and fusion, preferably with the Smith-Robinson technique, is recommended for patients with anterior impingement of the spinal cord at one or two levels in the absence of a narrow spinal canal. This procedure is also advantageous when there is significant radiculopathy associated with the level(s) of pathology.

Posterior decompression is recommended when three or more levels are involved, and particularly when there is developmental stenosis of the canal (*i.e.*, a DAD below 13 mm and an SAD below 11 mm).

Laminectomy and laminoplasty for CSM may not be any different in regard to surgical outcome. One well-controlled study showed only one difference, which was a decrease in the ability of the laminoplasty patients to extend their necks.[137]

If there is evidence of instability or the potential for it, posterior decompression procedures should be accompanied by a Robinson-Southwick or a Johnson-Southwick facet fusion, or, in the case of laminoplasty, some fusion modification such as that described by Itoh and Tsuji.[140]

There may also be circumstances in which significant multilevel anterior spur formation and compression in association with a stenotic canal should be treated with anterior and posterior surgery with appropriate attention to maintaining adequate stability.

DECOMPRESSION IN THE THORACIC REGION

Anterior Compression

The logical considerations involved are basically the same as those for the cervical region. However, the thoracic spine is stiffer, the potential for clinical instability is greater, the effectiveness of laminectomy for decompression and exploration is lessened, and the likelihood of worsening the neurologic condition is greater. These factors are mainly due to the relative lack of free space for the spinal cord and its precarious blood supply (see Fig. 6-25). For the treatment of the possible combinations of anterior compression, we suggest either *partial* (see Fig. 8-3, Part 2D) or total vertebral body resection (see Fig. 8-3, Part 2B). Wherever possible, partial resection is preferable, because this leaves more structural integrity to maintain clinical stability and simplifies the necessary reconstruction.

The recommended anterior approach for decompression and removal of a herniated thoracic disc is described in Chapter 6, Figure 6-48. There is no space to maneuver in the tight canal, so an anterior approach, which completely avoids putting anything into the canal, is suggested.

Posterior Compression

For midline as well as combined midline and lateral problems, we recommend bilateral laminectomy at a single or at multiple levels, depending upon the cephalocaudal extent of the lesion. For posterolateral compression, the recommendation is unilateral laminectomy or laminectomy and facetectomy.

Anterior and Posterior Compression

Here, we suggest vertebral body resection or a combined anterior and posterior decompression. In some fractures and/or dislocations, reduction and realignment may be helpful. If the pincer phenomenon is present, reduction and realignment should be considered, followed by bilateral laminectomy if the preceding is impossible or not effective.

Mixed or Poorly Localized Compression

In these cases, one may employ vertebral body resection, a combined anterior and posterior approach, and/or reduction and realignment.

Clinical Instability

Anterior or posterior procedures in some circumstances may render the thoracic spine unstable. Arthrodesis with or without instrumentation is required according to the analysis and indications discussed in Chapter 5 on the evaluation and management of instability. Considerations of the indications for and selection of internal fixation devices are discussed in a subsequent portion of this chapter.

Decompression for Thoracic and Thoracolumbar Burst Fractures

This fracture is specifically addressed in Chapter 4. With neurologic deficit and imaging evidence of bone or disc tissue impinging on the anterior portion of the canal, we suggest the following approach, as indicted in the algorithm in Chapter 4. Harrington distraction rods are employed with fusion one or two (usually two) levels above and below the fractured vertebrae. Cadaver biomechanical studies by Fredrickson and colleagues[89] showed that distraction is the *major* correcting force in burst fractures in regard to the reduction of intracanal fragments. In regard to reduction of intracanal fragments, the ability of a device to provide distraction force is more important than its ability to correct kyphosis. Any fragments unreduced by the distraction are exposed through a transpedicular approach and manipulated back into the vertebral body (Fig. 8-11). This technique of decompression is a slight modification of a technique described by Patterson and Arbit[231] for protruded thoracic discs. Additional stability may be obtained by wiring the spinous processes to the distraction rods. This provides additional stability without having to pass wires into the already traumatized spinal canal. See Chapter 4 for a more detailed discussion of the evaluation and management of this fracture. McAfee and co-workers have advocated a more aggressive approach in which anterior corpectomy and bone grafting are used.[190] These authors reported better results with more neurologic improvement than has been reported with nonoperative treatment and with Harrington distraction rods without fusion. They emphasized the important effect of canal compromise on microcirculation to the neural elements. We believe that if posterior stabilization, decompression, and attempted repositioning of displaced fragments out of the canal are not effective in correcting neurologic status or completely freeing the canal of extruded fragments, then anterior decompression is indicated.

DECOMPRESSION IN THE LUMBAR REGION

In the lumbar spine, the problem of compression is less common and less severe. The cauda equina starts below L2, and there is relatively more free space for the cauda equina than there is for the cord in other regions of the spine. Another unique factor about this region is that in the large majority of situations, the neural elements can be decompressed through a posterior approach. It is usually possible to carefully retract the dura and cauda

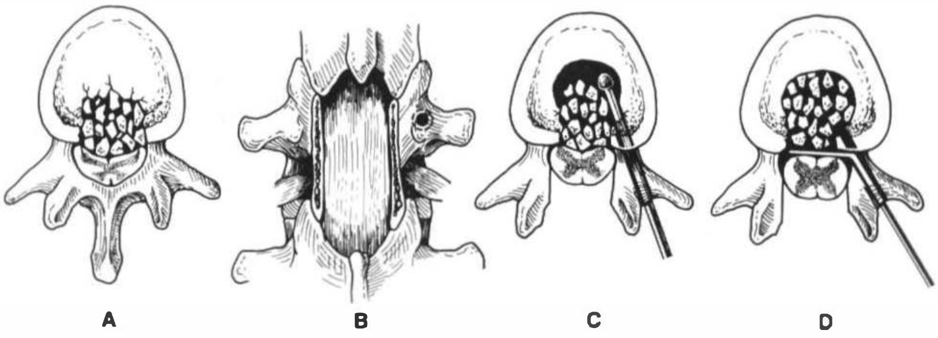


FIGURE 8-11 (A) Horizontal view of burst fracture with osseous fragments in the spinal canal. (B) Laminectomy is completed with care to avoid trauma to neural elements. If the canal is very tight, the laminectomy should not be employed. The pedicle is identified and confirmed. (C) A diamond burr is used to core out the pedicle and enter into the vertebral body. This tract is enlarged as necessary to essentially remove the pedicle from inside out and provide space in the vertebral body to facilitate repositioning of the bone fragments. (D) The fragments are readily and gently removed from the canal with a dental instrument and replaced in the vertebral body region.

equina far enough laterally to expose and remove anterior compressing structures. Anterior decompressions are sometimes necessary and can usually be achieved through a posterior approach. Although an anterior or anterolateral approach makes the vertebral body readily accessible, in this region it has some serious complications.[134,264] Nevertheless, anterior lumbar spine decompressions are sometimes indicated, most often, perhaps, with neurologic deficit associated with persistent fragments in the canal following burst fractures. See page 604 for a more detailed discussion of burst fractures and their management.

Anterior Compression

Anterior Midline Compression

Anterior midline lesions at the intervertebral disc only, as well as those behind the vertebral body, may be decompressed with bilateral laminectomy or multiple posterior laminectomies. Usually, the anterior structures may be exposed by side-to-side retraction of the dura and the cauda equina. In some instances it is necessary to include a partial or total vertebral body resection for the purpose of decompression or excision.

Anterolateral Compression

This condition may require one of several different procedures, depending upon the nature, size, and location of the lesions. When they are at the interspace, a simple partial laminotomy will suffice. In most cases, those compression sources at the interspace or at the level of the vertebral body require unilateral laminectomy, laminectomy with facetectomy, or in some instances partial vertebral body resection.

Anterior Midline and Lateral Compression

These types of compression lesions may be decompressed by several different procedures. They are bilateral laminectomy, multiple bilateral laminectomy, transpedicular partial vertebrectomy, or, when necessary, total or partial vertebral body resection through an anterior approach.

Posterior Compression

Posterior Midline Compression

In addition to the common causes (tumor, trauma, and infection), spinal stenosis, laminectomy membrane, and yellow ligament encroachment may cause compression of the neural elements in this location. The problem can be solved by a bilateral laminectomy.

Posterolateral Compression

Tumor, trauma, and infection may be responsible. The procedure of choice is a unilateral laminectomy or a laminectomy and facetectomy (foraminotomy) if there is also compression in the root canal.

Posterior Midline and Lateral Compression

The causes are the same as those for posterior midline compression. The surgical procedure for decompression is bilateral laminectomy or multiple bilateral laminectomies, depending on the extent of the compressed area.

Anterior and Posterior Compression

When this occurs at the interspace only, the cause may be a combination of hard or soft disc disease with yellow ligament encroachment, spinal stenosis, or a pincer mechanism. Because of the space available in the lumbar spine, combined anterior and posterior encroachment at the interspace level is unusual. In most instances, bilateral laminectomy alone would be expected to be sufficient for decompression. In some instances, a partial vertebral body resection may also be necessary. If the lesion is also at the level of the vertebral body, multiple bilateral laminectomies may be required.

Mixed or Poorly Localized Compression Sites

In these unusual circumstances, discrete localization eludes a thorough clinical study. Bilateral laminectomy, multiple bilateral laminectomies, partial vertebral body resection, combined anterior and posterior decompression, and reduction and realignment are all the procedures that may be useful.

Lumbar Spinal Stenosis

It is important to recognize that this is a decompression in which the major constriction of the thecal sac occurs at the level of the disc and yellow ligament.[256] The laminectomy and removal of all stenosing portions of the facet articulations and their respective capsules are also important. Special attention to the portion of the yellow ligament in the lateral canals is required. The evaluation and management of lumbar spinal stenosis is discussed in more detail in Chapter 6, p. 402.

Biomechanically, it has been noted in *in vitro* studies of human lumbar spines that, contrary to expectations, removal of one facet *decreased* rather than increased the load on the remaining facet joint. There was, however, an increase in peak pressures in the remaining facet joint.[179] (One would also expect an increase in the load on the disc, assuming that the total load remains the same.) Since peak pressures may contribute to arthritis of a remaining facet in vivo, removal of an entire facet joint should probably be considered at *least* a partial indication for fusion.

GUIDELINES FOR SELECTING A SURGICAL PROCEDURE

The general guidelines for selecting a surgical procedure for decompression are summarized in Table 8-2. The basic approach is to accurately localize the

TABLE 8–2 **Recommended Guidelines for the Precise Selection of Surgical Procedures for Decompression of the Spine**

		Decompression Surgery		
Compression Sites	**Examples**	Lower Cervical	Thoracic and Thoracolumbar	Lumbar and Sacral
Anterior				
Midline At interspace only	Hard disc, soft disc, tumor, trauma, infection	SR, BB, C	VBR (P)	BL
Behind vertebral body ± interspace*	Hard disc, soft disc, tumor, trauma, infection Ossification of the posterior longitudinal ligament	BB, C, KS, VBR, MBL	VBR (P)	BL, MBL, VBR (P)
Lateral At interspace only	Hard disc, soft disc, tumor, trauma, infection	SR, KH	VBR (P)	KH, UL, LF
Behind vertebral body ± interspace*	Hard disc, soft disc, tumor, trauma, infection	BB, C, KS, VBR	VBR (P)	UL, LF, VBR (P)
Midline and lateral At interspace only	Hard disc, soft disc, tumor, trauma, infection	SR	VBR (P)	BL
Behind vertebral body ± interspace*	Hard disc, soft disc, tumor, trauma, infection Ossification of the posterior longitudinal ligament	VBR, MBL	VBR, VBR (P)	BL, MBL, VBR (P)
Posterior				
Midline	Tumor, trauma, infection, yellow ligament, laminectomy membrane, spinal stenosis	BL, MBL	BL, MBL	BL
Lateral		KH, UL, LF	UL, LF	UL, LF
Midline and lateral		BL, MBL	BL, MBL	BL, MBL
Anterior and Posterior				
At interspace only	Pincher mechanism with anterior or posterior displacement Hard or soft disc with yellow ligament encroachment, spinal stenosis Tumor, trauma, infection, spinal stenosis	SR, VBR, BL, CAPD, RR	VBR, BL, CAPD, RR	BL, VBR (P)
Behind vertebral body ± interspace		VBR, MBL, CAPD	VBR, MBL, CAPD	BL, MBL
Mixed or Poorly Localized				
	Tumor, trauma, infection, other	VBR, CAPD, RR	VBR, CAPD, RR	VBR, CAPD BL, MBL, RR

* The site of compression may or may not be at the level of the interspace.

SR: Smith-Robinson, disc removal, and anterior interbody fusion
BB: Bailey-Badgley trough decompression and fusion
C: Cloward anterior decompression and dowel graft fusion
KS: Keystone interbody resection and fusion
VBR (P): Vertebral body or bodies resected totally or (P) partially, replacement and graft
KH: Keyhole—small laminotomy
UL: Unilateral laminectomy or laminotomy
BL: Bilateral laminectomy or laminotomy
MBL: Multiple bilateral laminectomies
LF: Laminectomy and facetectomy (nerve root decompression)
CAPD: Combined anterior and posterior decompression
RR: Reduction and realignment

site of the compression and then choose the appropriate surgical procedure to relieve it. The constraints are the risks and limitations of the various exposures and the liabilities created by the structural damage to the spinal column that is required to achieve decompression. The different regions of the spine vary in the accessibility and the necessity of the different approaches and decompression procedures.

Table 8-2 is also a guideline for the surgeon in training. It should be useful in emphasizing the importance of careful thought and evaluation to localize as accurately as possible the site of compression. Then, an *appropriate* rather than a *routine* procedure may be selected to effectively decompress the spinal cord and/or nerve root.

PART 2: SPINE FUSIONS

Since the procedure was first introduced by Albee and Hibbs in 1911,[3, 132] arthrodesis has been one of the most important and frequently employed operations of the spine. This part of the chapter discusses the mechanical aspects of the various techniques of spine fusion. Theoretic and experimental background information is provided along with relevant clinical data. We do not comprehensively catalog all fusion operations. References for many of the procedures may be found in the work of Wu.[332]

Surgical fusions are recommended at various levels of sophistication. A physician may state, "The patient should have surgery," or "That patient needs a fusion," or "That patient needs a posterior C1–C2 Brooks fusion in order to provide for some immediate stability against anterior translation and restrict axial rotation while the union matures." Appropriate, detailed recommendations, such as the latter, require a sound mechanical understanding of the available surgical procedures.

CLINICAL BIOMECHANICS OF SPINE FUSIONS

Spine fusions are used for one or more of the purposes listed below. The ideal is to achieve the necessary therapeutic goals with the minimal effective decrease in motion and minimal disruption of normal structure and function of the spinal column—in other words, maximum therapy with minimum risk and liability to the patient.

REASONS FOR SPINE ARTHRODESIS

To support the spine when its structural integrity has been severely compromised (to reestablish clinical stability).

To maintain correction, following mechanical straightening of the spine in scoliosis or kyphosis or following osteotomy of the spine.

To prevent progression of deformity of the spine, as in scoliosis, kyphosis, and spondylolisthesis.

To alleviate or eliminate pain by stiffening a region of the spine (e.g., diminishing movement between various spine segments).

Biomechanical Factors Important in Spine Fusion Surgery

Surgical constructs (operations) should be chosen on the basis of suitability. In other words, for any given surgical problem, there are one or more constructs that effectively achieve the desired therapeutic goal. The various surgical procedures have their own unique structural and biomechanical characteristics. The surgeon's goal is to accurately understand the biologic and mechanical aspects of the problem and select the appropriate surgical construct to solve it.

Clinical descriptions of surgical procedures in the literature devote ample attention to the indications for surgery, the anatomic aspects of procedures, and postoperative care. A good deal less emphasis is placed on choosing the surgical construct. Factors that are not generally discussed are the mechanics of the surgical construct and the relationship of the mechanics to the clinical requirements and goals of the procedure.

Biomechanical Effects of Spine Fusions

In addition to the therapeutic value of spine fusions there are some theoretic and documented liabilities. Some experimental work shows increased motion below a "fusion."[237] The experimental work of Lee and Langrana showed increased stress on the adjacent unfused segments.[173] Some observations of excessive motion,[234] degenerative changes,[42, 139] spinal stenosis,[70,118] and even fracture dislocation[65] have been observed in adjacent segments in association with spinal fusions. All these changes adjacent to the fusion mass are most likely biologic, and in some instances pathologic, changes due to the stress concentration at the interface of the highly stiffened (fused) segment of the spine and the more flexible (unfused) segment of the spine.

Electrical Stimulation

A recent development has involved the use of electrical stimulation to enhance bone formation in spine fusions. A double-blind study of 30 pigs by Nerubay and associates showed a statistically significant increase in osteoblastic activity with bone formation.[209] The actual comparative union rate and mechanical properties of the fused spine were not reported. Nevertheless, this is cogent and interesting work.

Kane[151] has shown, in a randomized prospective controlled clinical study, that electrical stimulation improves the success rate in spinal fusion. The control group of 28 patients had a fusion rate of 54%, and the group of 31 that was electrically stimulated fused at an 81% rate ($p > 0.005$).

Graft Materials

There are several important considerations concerning the choice of graft material and its use in surgical constructs. The graft may be used in several different manners. For example, it may serve as a structure to contribute to immediate postoperative stability, as a scaffold, as a spacer, or as a bridge to span a particular spinal column defect. The basic biologic use of graft material is to induce, establish, or assist in osteogenesis. It has not been determined how this occurs or even if it does occur. The main effect may be limited to the provision of a latticework or some structure for the ingrowth of new bone.

Cortical bone, except for a small portion of its osteocytes, dies after transplantation. The bone is more rapidly revascularized if its periosteum is removed. During the process of remodeling and revascularization, there is a relative osteoporosis and weakening of the graft. As creeping substitution progresses, the new bone takes on the mechanical characteristics that are dictated by the regional biomechanical environment. Three phases have been well documented,[286] and they apply to cortical and cancellous grafts. Although the three phases have been given several names, the following seem to adequately identify the process: (1) creeping substitution, (2) osteogenic regeneration, and (3) functional adaptation. As would be expected, there is considerable overlap among the phases.

The fate of a cancellous bone graft is somewhat different from that of cortical bone. The cancellous bone, especially the red marrow of the ilium, will have a large number of surviving osteogenic cells in its deeper areas. In addition, revascularization is facilitated by the open spongy structure of cancellous bone. For a more detailed discussion and presentation of this material, the works of Burwell,[35,36] Enneking and colleagues,[71] and Stringa and Mignani[286] are recommended.

Other recent developments relate to the clinical science of bone grafting.

Biodegradable Bone Composites

Investigators are working to develop a biodegradable or bioabsorbable bone composite.[53] The goal is to have a readily available substance that can serve immediately as a load-bearing substance and also be gradually replaced by host bone. Progress has occurred in *in vitro* studies and in some animal investigations.[53]

Effects of Chemotherapeutic Agents

Burchardt and associates studied bone autografts in dogs medicated with Adriamycin and methotrexate. They observed that there was a suppression of new bone formation and suggested that with these drugs there would be a prolonged time for union and a greater incidence of nonunion.[34] The suggestion is that in the clinical setting where bone graft is used in the presence of these chemotherapeutic agents, maximum protection of the surgical construct for a long period of time is desirable.

Key issues related to bone grafts have been reviewed by Friedlaender.[92] This review and update suggests that while autografts may be slightly superior to allografts in terms of time required for incorporation, allografts are also successful. The practical advantages of allograft availability in regard to size, shape, and quantity, with no additional surgery for harvesting, may equal or outweigh the advantages of autogenous bone described above. Specific clinical circumstances can change this delicate equilibrium in favor of either of the choices. We have moved to the routine use of freeze-dried and irradiated allografts in initial spinal fusions.

Iliac Crests

Nature has thoughtfully and generously anticipated the needs of surgeons and provided the iliac bones. This ready source of bone has a number of advantages. It is expendable; it can be harvested with the patient in either the prone or supine position; there is ample cortical or cancellous bone; and the structure lends itself to the removal and carpentering of a variety of useful shapes and sizes. Moreover, the cancellous bone can be impacted and molded to fit the irregular contours of the irregular bony structures to which it must be apposed. A careful study of a freely dissected ilium will easily familiarize the surgeon with the numerous varieties of natural shapes and combinations of cortical and cancellous grafts that are available (Fig. 8-12). Clinical observations show that the use of iliac bone is preferable to the use of tibia in lumbar spine fusions.[125] The disadvantages of the ilium as a donor site include its not uncommon complications, including severe pain, hematoma, hypersensitivity, buttock anesthesia, herniation at the donor site, subluxation of the hip, numbness, myalgia paresthetica, infection, and bony overgrowth that sometimes has to be resected.[318] The complication of a postoperative fracture of the anterosuperior iliac spine (Fig. 8-13) can be avoided if the surgeon is careful not to take a deep cut into the ilium less than 3–5 cm behind the anterosuperior iliac spine.[164]

Experimental studies that provide useful information about the relative strengths of some of the different configurations of iliac bone grafts have been carried out on fresh cadaveric ilia.[314] Relative compression strengths of different grafts are related to the postoperative stability of the surgical constructs. The results of the study were as follows. The small horseshoe configuration of bone (Smith-Robinson procedure) was the strongest, followed by the large horseshoe configuration (modified Bailey-Badgley procedure), and the weakest of the three was the dowel configuration (Cloward procedure). It was found that *all* three configurations sustained high loads. All specimens withstood loads approximately 2.5 times the average body weight. The calculations based on the publications of Henzel and colleagues[128] and Ruff[252] indicate that about 10% of total body weight is above T1 and about 50% is above T12. The loads may be of greater magnitude if the joint reaction forces associated with functional *in vivo* loading from muscle forces are considered. In the lumbar spine, the lever mechanisms derived from the muscle forces impart loads to the spine comparable to as much as three to four times body weight (see Chap. 1). This information has significant implications. In the immediate postoperative period, with the muscles relaxed or the head supported, the load-bearing capacity of the grafts is adequate. However, with the introduction of such variables as dynamic loading, physiologic muscle forces, and creeping substitution of the bone graft, especially in the thoracic and lumbar region, the probability of the load-bearing capacity of the grafts being exceeded increases significantly. Also, their *relative* strengths become more crucial.

Fibula

A fibula graft is probably the strongest in resisting compression because of the relatively large amount of cortical bone. The weaker grafts from the ilium are probably strong enough in the immediate postoperative period. The fibula is reabsorbed slowly and therefore can be depended upon for a longer period of time for its structural support against compressive loading. The disadvantages are the small amount of cancellous bone and the potential for functional compromise of the mechanics of the donor site. It has been shown that the fibula bears approximately one-sixth of body weight.[166] This liability can be virtually eliminated by the use of hemicylindrical fibular grafts.

Ribs

Ribs are good bone graft material, especially for arthrodesis of the anterior thoracic spine, because they are so readily available.[74] Because of its structure, which consists of a modest cortex and porous cancellous bone, it has the advantage of reasonable strength without having a good deal of dense cortical bone that must be incorporated. Furthermore, its slight curvature gives it a certain resiliency and permits it to conform to a cervical or lumbar lordosis or a thoracic kyphosis

Tibia

Tibial grafts are occasionally used in spinal arthrodesis. Although this provides a strong graft, we do not think that the liability incurred justifies its use. When the structure of the tibia is changed from a closed section to an open one, it is considerably weakened. The structure is much less able to resist both torsional and bending loads.[88] Chrisman and Snook estimated a refracture rate in skiers at 3%.[45]

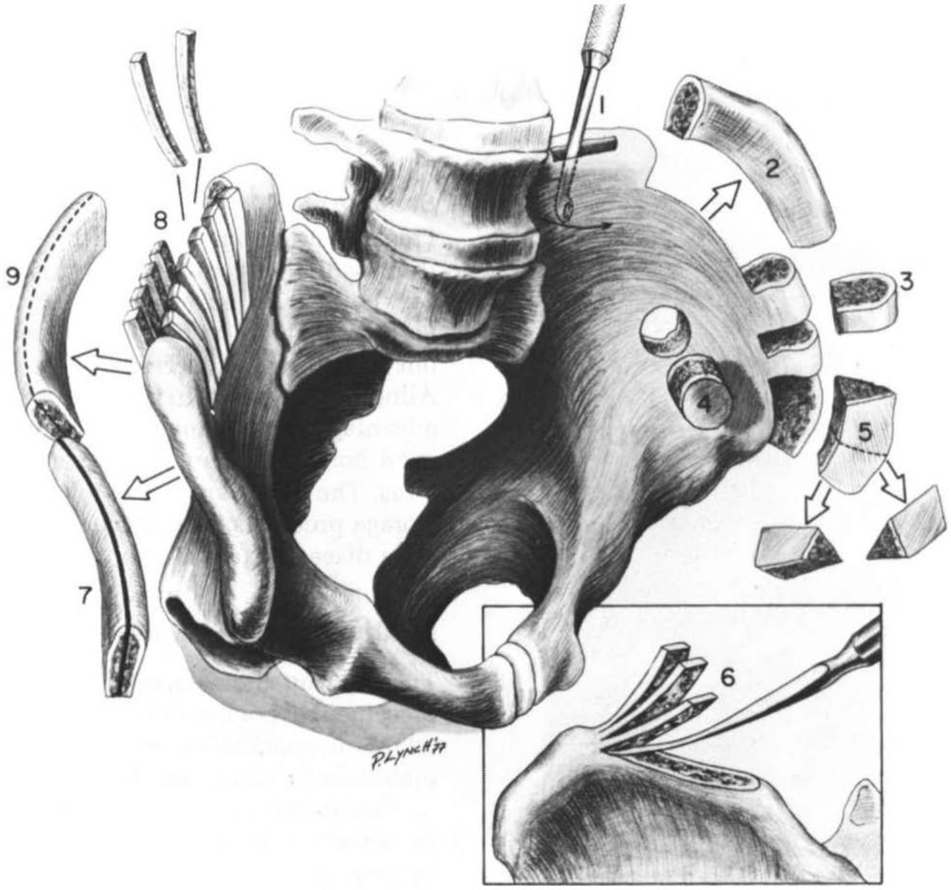


FIGURE 8-12 This is a diagrammatic representation of some of the various combinations and configurations of bone grafts available from the ilium. *(1)* Cancellous bone may be curetted from any portion of the iliac crest. *(2)* Various lengths of horseshoe-shaped cortical cancellous graft may be taken for trough grafts and vertebral body replacements. *(3)* The smaller horseshoe configuration used in the Smith-Robinson fusion. *(4)* The dowel configuration. *(5)* A part of the ilium of variable sizes may be taken and fashioned for the Brooks C1–C2 fusion. *(6)* A technique for obtaining multiple onlay grafts with generous portions of cancellous bone. *(7)* A large horseshoe may be split longitudinally and employed in a variety of constructs, to fit a kyphosis or a lordosis with cancellous bone facing up or down. These may also be cut from the ilium so that they are C-shaped, or a mirror image of the same. *(8)* A convenient source of multiple corticocancellous strips. *(9)* Instead of the portion labeled 8, this portion of the ilium, with a different natural shape, may be cut to provide additional uses, similar to those described for 7. When initial strength of bone graft is important, the anterior portion of the ilium may be preferable.[314]

This was due to a persistent cortical defect. Therefore, if at all possible, physicians should avoid weakening the tibia in this manner.

Allograft Bone

The physical properties of allografts may be presumed to be similar enough to those of autografts for the former to be used as a substitute, at least with respect to the immediate postoperative mechanics. If the immunologic factors could be eliminated or were shown to be insignificant, then the use of bank bone could be very advantageous. Besides eliminating all the donor site complications, the variety and availability of donor material could be enhanced. There are now several studies that suggest that the infection rate is not increased[293] and the fusion rate

FIGURE 8-13 This x-ray of a 65-year-old male who had a large bone graft removed from his left anteriosuperior iliac spine (ASIS) demonstrates one of the many complications of using the ilium as a donor site. When the patient was taken out of bed to ambulate on the second day following an anterior L3–L4 lumbar spine fusion, he had sudden onset of pain in the left hip region. The cutout of the ilium close to the ASIS (about 2.5 cm) left a thin spike of bone. Because of the depth of the graft taken, a long moment arm was created. This allowed the sartorius and tensor fasciae latae to easily cause a fracture at the base and pull a portion of the ASIS off. Removal of the graft at a point 3 to 5 cm from the tip of the ASIS would probably have prevented this. The patient was treated with crutches and progression to normal ambulation, which occurred in 4 to 6 weeks.

for spine surgery is about as good with freeze-dried allografts as it is with autografts.[51,208,237,255,335] We must note here an important study by Stabler and associates, who noted that in seven posterior spine fusions in children there were seven failures.[272] Also, Bosworth performed a study and found autograft bone to be three times as good as frozen allograft bone.[21] This issue is not yet settled. However, it appears that allograft is probably not quite as good as autograft. Controlled studies are needed to resolve this.

Summary

Present knowledge suggests that the patient's own ilium is the best source of graft material. With surgery in the thoracic region, the rib provides an excellent bone graft. The fibula is preferable to the tibia, but both of these sources have some disadvantages. Allografts are useful in many situations and have the advantage of no complications at the donor site; a good bone bank can offer a variety of shapes and sizes. The disadvantages of allografts are related to storage problems, immunologic reactions, and possible disease transmission.

Positioning of Bone Grafts

The relevant biomechanical considerations of the placement of bone grafts focus primarily upon sagittal plane mechanics, and frontal and horizontal plane mechanics in some instances.

The placement of a fusion mass at the maximum distance from the instantaneous axes of rotation will be more effective in preventing movement around those axes. In preventing sagittal plane rotation of the upper vertebra in relation to the lower one, a fusion mass located on the tips of the spinous process is more effective than one that is placed closer to the instantaneous axes of rotation. This concept, which relates to leverage and area moment of inertia, is exemplified in Figure 8-14. Thus, in terms of discouraging motion of an entire FSU, the further the graft is placed from the instantaneous axes of rotation, the more effective it will be. This principle also applies to axial rotation and lateral bending. Looking at this point alone, the posterior fusion established some distance from the instantaneous axis of rotation is better than one that is placed closer to it.

The concept of leverage is also important with respect to the instantaneous axes of rotation and placement of a fusion mass. During flexion, assuming that the axes of rotation are located in the middle or slightly anterior portion of the disc, the leverage situation is as shown in Figure 8-14, part 2. It is readily apparent that an anterior bone graft has relatively less leverage than a posterior one with regard to its efficacy in preventing rotation of the upper

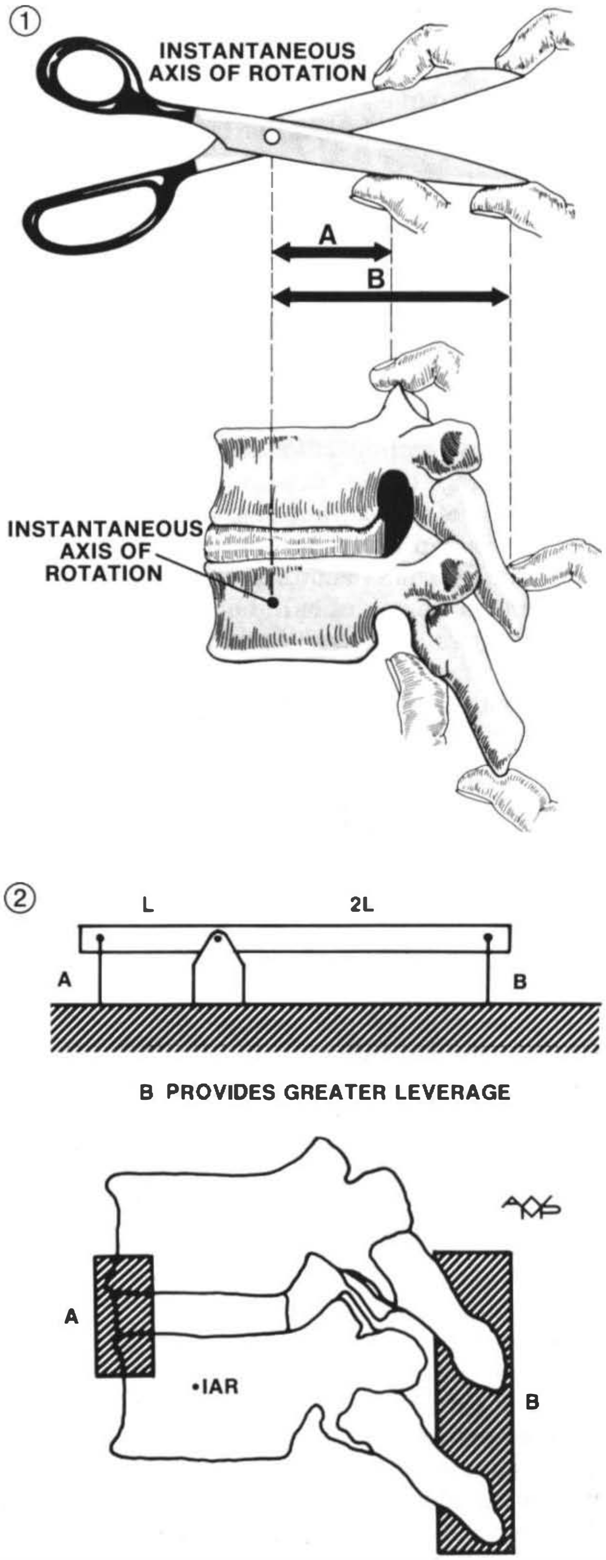


FIGURE 8-14 **(1)** To prevent the opening of the blades of the scissors by holding them together, it is distinctly easier to pinch the blades together at the tips (*distance B*) rather than at the midpoint of the blade (*distance A*). Because distance B is further away from the instantaneous axis of rotation, there is greater leverage. The same concepts apply to the vertebral FSU. Flexion, separation, or opening of the spinous processes, is more readily prevented by placing the fingers at the tips of the spinous processes (*distance B*) rather than at the facet joints (*distance A*). Thus, with regard to a flexion movement, a healed bone graft at distance *B*, at the tips of the spinous processes, is more effective than one closer to the instantaneous axis of rotation, other factors being constant. These concepts partially explain the efficacy of the rather delicate interspinous and supraspinous ligaments. **(2)** The concept of leverage is shown again here. The anterior bone graft *A* is a short distance (analogous to *L*) from the instantaneous axis of rotation and therefore provides less leverage than bone graft *B*, which is a greater distance (analogous to *2L*) from the instantaneous axis of rotation.

vertebra in flexion or extension. Go a step further and assume that the instantaneous axes for axial rotation bear the same relationship to grafts A and B in Figure 8-14, part 2. If this were the case, then graft B would also have more leverage in preventing axial rotation.

This is not the first introduction of the biomechanical concept of leverage to the literature on spine surgery. The following points were made in 1911 by Albee in a discussion of the importance of splitting the spinous processes and inserting a bone graft between the two parts during a spine arthrodesis: "This method is believed to be preferable to any where breaking or cutting of the spinous processes destroys entirely or for the time being . . . the desired leverage of the spinous processes . . . "[3]

There are other relevant mechanical considerations. The concept of rigidity is a crucial mechanical factor with regard to fusion. This concept is important from the viewpoint of the normal elasticity of the vertebral structure and the relative efficiency of the fusion mass in preventing deformation of the vertebra with various physiologic loads. The principle of rigidity and its application to the vertebral FSU are shown diagrammatically in Figure 8-15. The practical significance of this concept is the fact that a fusion mass that involves the spinous processes, lamina, and transverse processes is more rigid and immobilizes more effectively than one that involves only the spinous processes.

Rolander demonstrated experimentally that the normal elastic properties of the bone are such that motion may still take place with physiologic forces applied to the FSU after an adequate posterior fusion.[243] During in vitro experimental studies, he actually fixed *all* posterior elements except the pedicles with cement and found significant motion at the interspace with physiologic loading (Fig. 8-16). In the clinical situation, a bone graft is a more elastic structure than the cement used in the experiment; therefore, even more motion is permitted. Such a posterior fusion would be sufficient if its purpose were to substitute for the stabilizing role of destroyed ligaments. However, it would be sorely lacking if its goal were to totally eradicate motion at the disc interspace as a requirement for eliminating discogenic pain. Obviously, in the latter situation the principle of placing a fusion mass away from the instantaneous axes of rotation should be abandoned and an interbody fusion should be carried out. The interbody technique provides high rigidity by eliminating interbody motion (Fig. 8-17). This procedure, when feasible, not only eliminates movement between vertebrae to the maximum degree that is possible with bone, but also removes all or part of the intervertebral disc.

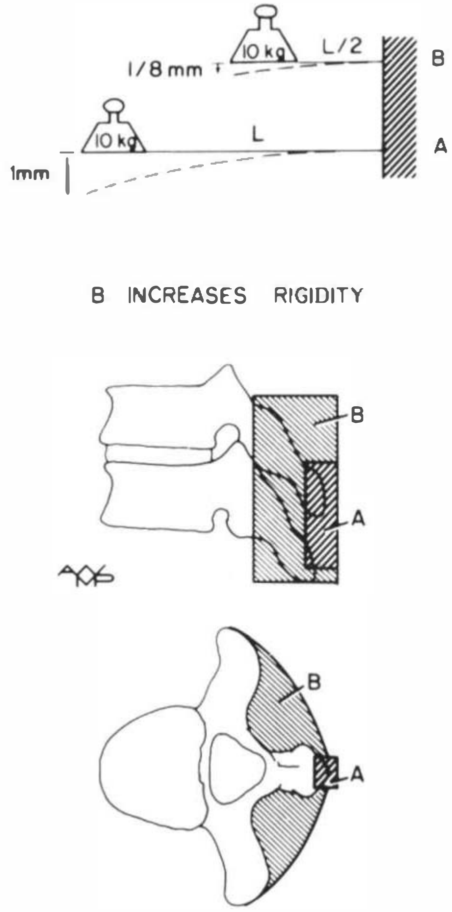


FIGURE 8-15 The concept of rigidity as applied to spine fusions is illustrated here. The top figure shows a movement of 1 mm with a 10-kg mass placed at a distance *L* meters away from the attachment of the beam. If that same 10-kg mass is only one-half the distance away, the motion is reduced to 1/8 mm. In the second example (with a shorter distance), there is more rigidity. The analogous placement of a fusion mass in relation to sources of motion in a vertebra shows that fusion mass *B* will provide much more rigidity than fusion mass A.

Experimental and Clinical Tests of Hypotheses

An excellent clinical biomechanical study demonstrates nicely some of the theoretic considerations espoused in this section on bone graft positioning.

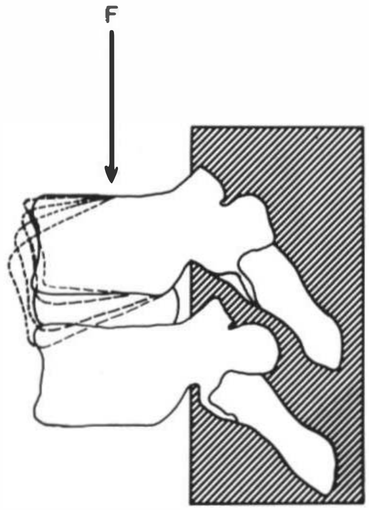


FIGURE 8-16 This experiment by Rolander shows the mechanism through which a force F can cause motion between vertebral bodies in the presence of a solid posterior arthrodesis. The motion is permitted by the elastic properties of the free osseous structures. Since it is not known how much motion causes pain or if motion is responsible for pain, it is readily understandable that spine fusion for pain in the lumbar region has not been a particularly good procedure. *(White, A.A., Southwick, W.O., Panjabi, M.M., et al.: The practical biomechanics of the spine for the orthopaedic surgeon. In American Academy of Orthopaedic Surgeons: Instructional Course Lectures. St. Louis, C. V. Mosby, 1974.)*

Lee and Langrana studied in 16 fresh cadavers the following types of spine fusions: posterior, bilateral lateral, and anterior under combined compression bending loads.[173] All three types increased the axial and bending stiffness, but the anterior fusions were the stiffest, followed by the bilateral lateral and then the posterior. All three increased stress at the adjacent unfused segments, especially in the facet joints. The bilateral lateral fusion was the best in terms of good stabilization with the least effect on the adjacent unfused segments. The posterior fusion permitted anterior motion and was associated with the highest stresses on the adjacent segments.

The clinical study of Percy and Burrough[234] evaluated lumbar spine interbody fusions with biplanar radiography. They noted very restricted motion of the fusion with paradoxical motion and also increased motion of the segment above the fusion when compared with the other unfused FSUs (see list on p. 564 for review of changes adjacent to a fusion).

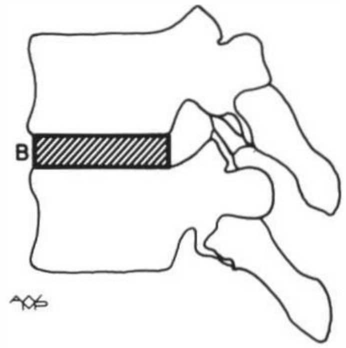


FIGURE 8-17 This illustration shows the position of a bone graft *B*, which can provide maximum rigidity by eliminating interbody motion.

This information is of particular interest in regard to the clinical decision to do a posterior interbody fusion, a posterior fusion in conjunction with an anterior interbody fusion, or just a bilateral lateral fusion. Other things being equal, this study would favor the choice of a bilateral lateral fusion construct.

Kyphotic Deformity and Bone Graft Positioning

Positioning of the graft is important in another context. This has to do with the use of a bone graft to prevent deformity, maintain correction, or substitute for damaged or absent structures in the presence of a curve. The mechanical principle involves the relationship of the bone graft material to the neutral axis. If the spine is thought of as being analogous to a beam that is bent and loaded as shown in Figure 8-18A, there are compression stresses (−) on the concave side and tensile stresses (+) on the convex side. Somewhere in the middle at the neutral axis there is neither compression nor tension. Furthermore, these stresses vary, with a maximum stress on the surface and no stress along the neutral axis (Fig. 8-18A). When an anterior bone graft is to be used as a spacer or to resist compressive forces, it should be placed at a more anterior location in the vertebral body. The closer the graft material is to the neutral axis, the less effective it will be. This applies to the role of the graft in resisting tensile as well as compressive forces. A graft placed at this anterior position offers more effective immediate postoperative stability against axial rotation and flexion or anterior collapse because it is placed further away from the respective axes of rotation. The more posterior of the

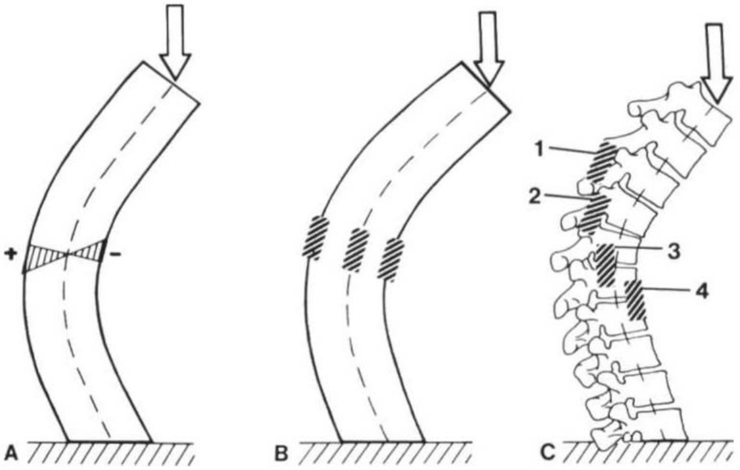


FIGURE 8-18 **(A)** The spinal column may be analyzed by regarding it to be somewhat like a beam. There is tension on the convex side of the curve and compression on the concave side. The dashed line is the neutral axis, and there is neither tension nor compression along this line. **(B)** Bone grafts inserted at the various points tend to behave as follows. The graft on the convex side of the curve is mainly under tension and cannot resist deforming forces until fully attached at both interfaces. The graft at the dashed line (neutral axis) provides little or no resistance to bending in the plane of the page. **(C)** In the spine, the graft at position 1 is well away from the neutral axis and when biologically fused at both interfaces can offer effective tensile resistance against progressive kyphosis. The graft at position 2 can do the same but is less effective because it is closer to the neutral axis. The graft at position 3 is not likely to be as effective as 1 or 2 in preventing progression of deformity because it is even closer to the neutral axis. Graft 4 is effective because it immediately begins to resist compressive forces, which tends to prevent additional deformity and angulation at that point. The graft is also some distance away from the neutral axis, giving it mechanical advantage. *(White, A. A., Panjabi, M. M., and Thomas, C. L.: The clinical biomechanics of kyphotic deformities. Clin. Orthop. 128:8, 1977.)*

two anterior interbody graft locations shown in Figure 8-18 is less effective in resisting the two motions than is the more anteriorly placed graft. These points are important in the treatment of kyphotic deformities.

Sometimes long, anterior strut graft fusions are required in the thoracic spine to maintain stability against progressive kyphotic deformity.[26,328] Two questions are posed in relation to this problem. How many struts should be placed, and where should they be placed in order to be most effective? Sometimes, major portions of the vertebral bodies have been destroyed or removed, and this factor largely determines where the grafts should be placed. These questions are addressed below through an analysis of the three anterior bone graft locations shown in Figure 8-19.

Graft A is some distance away from the neutral axis but offers support only at the FSU in which it is implanted. This can be effective if there is a sharply angulated kyphosis located at one FSU, which may occur in some cases of trauma. There is a surgical construct in which graft A is not employed but B and C are used together. When this is employed, we must assume that only one of them, either B or C, is bearing the major portion of the compressive load. This is due to the fact that in the immediate postoperative period, the surgical construct is unlikely to be designed and erected precisely enough to have both grafts participate equally in the load bearing. However, at later stages when biomechanical adaptation occurs, both columns of graft would be expected to bear an appropriate share of the loads.

The surgeon may choose one of three alternatives. The construct may consist of B, of C, or of B and C, shown in Figure 8-19. The further the graft is located from the neutral axis, the greater the lever arm with which the graft is working, and the more

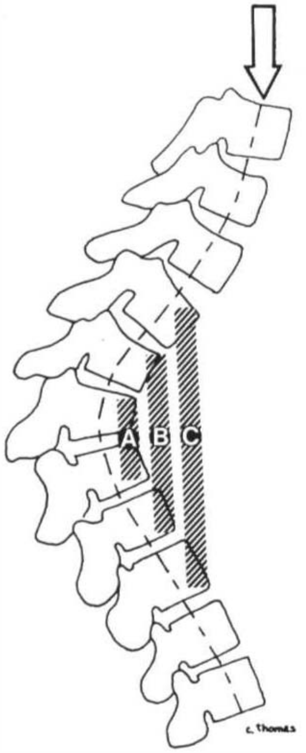


FIGURE 8-19 This is an illustration of the various locations of anterior bone grafts for kyphotic deformity. The biomechanical considerations involved in choosing graft A, B, or C are discussed in the text. *(White, A. A., Panjabi, M. M., and Thomas, C. L.: The clinical biomechanics of kyphotic deformities. Clin. Orthop., 128:8, 1977.)*

effective is the support. Thus, if only one is to be used, graft C may appear to be the most attractive alternative. However, there are other considerations. The longer, more anteriorly placed graft is more likely to fail from buckling. Its length is critical with regard to this situation. (A more detailed analysis of the concept of buckling is presented in Chap. 9.) It may also interfere with neighboring anatomic structures and is more difficult to revascularize because of its size and position. For these reasons, the more closely placed graft *B* is probably preferable, if only one is used. The best solution is to use both *B* and *C* and to place the most anterior graft as far from the neutral axis as is surgically feasible, recognizing that with biomechanical adaption they will *both* bear some of the loads in the long range.

Clinical experience suggests that for severe kyphosis, posterior fusion alone is likely to fail. Amed and colleagues, Bradford and colleagues, Hall, O'Brien, and Winter and colleagues have recommended in cases of severe kyphosis with tuberculosis that the patient have an anterior decompression followed by an anterior and a posterior fusion.[5,26,112,217,328]

Subsequent clinical follow-up studies by the Moe group[178,327] and by Herndon and co-workers[130] suggest that anterior and posterior fusions are the most likely to be successful. Also, the greater the deformity, the more difficult it is to correct and arrest. Thus, earlier surgery is more effective. Herndon, who reported on Scheuermann's disease, suggested that posterior fusion alone may suffice in curves 70° or less. However, Winter and colleagues,[327] from their studies of congenital kyphosis, suggest a cutoff of only a 55° deformity for posterior fusion alone. The trend is long anterior and posterior fusions early.

Wolff's Law and Spine Fusions

Physicians frequently hear the following orthopedic banality—"Don't put that bone graft under tension, it will be absorbed because of Wolff's law." Admittedly, most banalities are allowed to become such because they carry a certain element of truth or at least apparent truth. Wolff's law must be critically examined and put into some biomechanical perspective before we accept the above as true.

Wolff's law states that bone is laid down where stresses require its presence, and bone is absorbed where stresses do not require it. Somehow, the law has been misconstrued to mean that bone is laid down or built up where it is loaded in *compression* and absorbed where it is loaded in *tension*. This is wrong. Bone is also built up or laid down where it is undergoing tensile stress. There is plenty of good strong bone on the anterolateral aspect of the femur. This area is under considerable tension. Experimental studies have actually used tensile loading to effectively stimulate osteogenesis.[131,177] Thus, it should not be assumed that bone on the convex side of a scoliotic or a kyphotic deformity will be absorbed just because it is under tensile loading. However, this does not relate to the relative effectiveness of this bone in performing various biomechanical functions; it simply means that because bone is loaded in tension, one need not assume that it must be resorbed.

Extent of Fusion

Fusion of One FSU

When trauma or disease has disrupted the stability of one FSU, a simple posterior fusion fixing the vertebra of that FSU can be completely satisfactory, provided certain conditions are present. There must be adequate posterior osseous structures to which the vertebra can be fused, and the anterior and posterior bony structures that remain in each vertebra must be in continuity with the rest of that vertebra. Obviously, stability cannot be achieved if there is an ununited fracture of the pedicles of one vertebra and its posterior elements are fused to the adjacent lower vertebra.

Fusions Involving Two or More Vertebrae

When one or more vertebrae are structurally destroyed, partially or totally absent, or unable to provide clinical stability, then it is necessary to construct the fusion so as to attach it to one normal FSU above or below the pathology. Examples include fusions for vertebral body resection, spondylolisthesis, or ununited fracture of the ring of C1 (requiring C0–C2 fusion). The basic idea of this construct is to include in the fusion normal spine segments above and below the pathology. The abnormal segment(s) is included in the fusion when that is possible, and it is bridged over when this is not possible.

We disagree in most situations with the recommendation to include more than one normal vertebra above and below the pathology, unless there is a deformity. One normal FSU should be as good as its adjacent normal one in withstanding loads. Moreover, there is an unnecessary restriction of motion when additional normal FSUs are included. Finally, it is well documented that the FSU above a fusion may sometimes develop abnormal motion to the point of clinical instability, so that fusion is required. To fuse it before this is required eliminates an option unnecessarily and shifts the risk one FSU higher or lower. There are two exceptions to the principle of fusion to just the first normal adjacent vertebra on either side of the pathology. First, in patients with tumors, an adequate margin of resection is not always certain. There are cases in which a portion of destructive tumor must be left behind, and additional progression or recurrence is expected. Second, in some patients, maximum postoperative stability is required. In these situations, two adjacent normal vertebrae are incorporated into the fusion mass for additional purchase and stability and a margin of safety.

In the special situation of fusion for the arrest of progression or the preservation of correction in kyphosis, other biomechanical principles are operative. We believe that posterior fusions for kyphosis should include *all* the vertebrae in the deformity. A short fusion has to work against a large moment arm created by the weight of the trunk above. The larger fusion is probably superior because of its greater mass and the reduction of the effective moment arm acting on it.

By including all the vertebrae in the kyphotic curve in the fusion and reducing the effective moment arm operating at the end of the fusion mass, the probability that additional vertebrae will become part of the deformity is reduced. In addition, the forces that contribute to abnormal motion at the end of the curve (Fig. 8-20) are also reduced. Attention to this principle tends to decrease the incidence of the type of problems reported by Wagner and colleagues.[305] These investigators noted that when an inadequate number of vertebrae were included in the fusion, kyphotic deformities developed above and below the fusion mass.

This same principle applies to fusion of a scoliotic curve. We believe that it is adequate to include the transitional vertebrae at either end of the curve. The possible exception is fusion of a rapidly progressing curve in a young person. In fusions at multi-

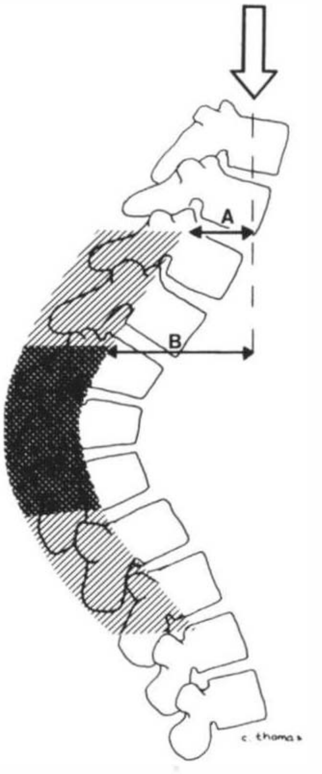


FIGURE 8-20 This shows the mechanical advantage of a longer posterior fusion (*light and dark shading*) over a shorter fusion (*dark shading*). With the short fusion alone, there is the possibility of an effectively longer moment arm B as opposed to the relatively shorter moment arm A. Also, the longer fusion, when mature, provides a more effective internal splint. (*White, A. A., Panjabi, M. M., and Thomas, C. L.: The clinical biomechanics of kyphotic deformities. Clin. Orthop., 128:8, 1977.*)

ple levels, the fusion should include the first adjacent vertebra above and below the pathology that is part of a normal FSU. To fuse beyond these limits is unnecessary and disadvantageous. Guidelines for scoliosis fusion are discussed in more detail in Chapter 3, page 153.

Spine Fusions in Children

Several questions are frequently raised about spine fusion in children. Will early fusion disrupt growth patterns and cause deformity or neural damage through an inability of the fused spine to accommodate the maturing neural elements? Another question relates to the feasibility of a therapeutic asymmetrical fusion in the correction of a deformity in the growing child. Will fusion on the convexity of a deformity (scoliosis or kyphosis) result in correction through subsequent symmetrical growth?

Children as young as 2 years of age may have cervical spine fusion without any of the problems of deformity or neurologic complications from relative hypoplasia of the fused section of the spine.[260] This is based on a report of 13 patients, aged 2 to 15 years. These observations are supported by the experience of Hallock and colleagues, who also noted that surgeons should not generally anticipate any correction of kyphotic deformity from any asymmetrical growth associated with posterior arthrodesis.[113] With posterior fusion, there was continued growth. However, the anterior elements grew 37% less than would have been expected without the fusion, and the posterior elements grew 47% less than would have been expected. The vertebral and disc space heights in the fused segments were both less than the expected normal.

Bridge Constructs, Spacers, and Prophylactic Fusions

Rules without exceptions are unique, yet boring. In some special situations and constructs, for maximum immediate or long-range stability it is necessary to include more than one normal FSU above and/or below the usual first vertebra in a normal FSU. With massive resection of all or part of one or more vertebrae, a bone graft is sometimes used as a spacer or a bridge to span a defect (Fig. 8-21). It may

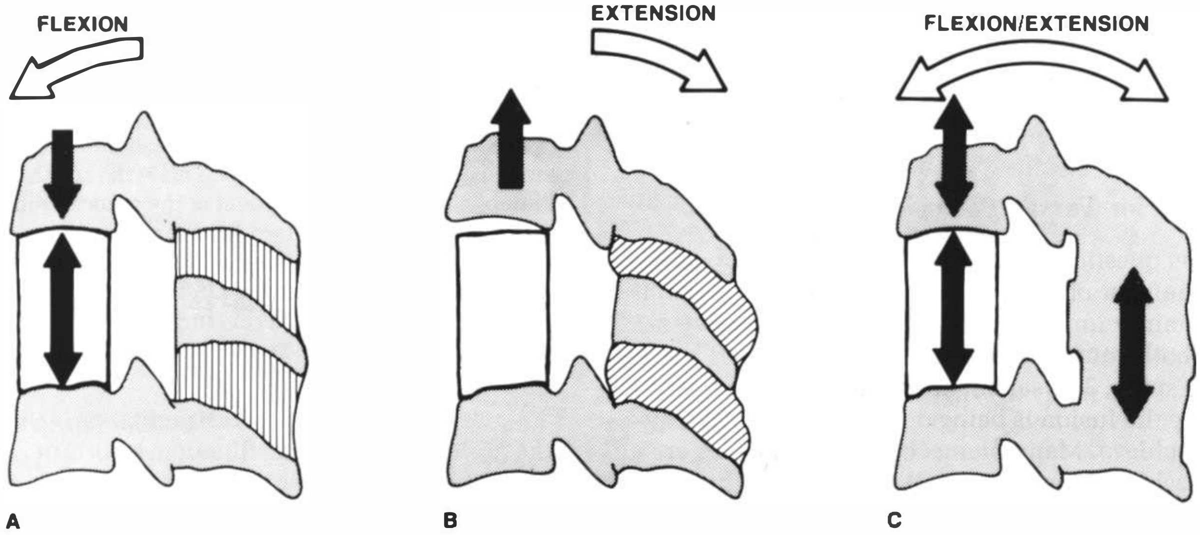


FIGURE 8-21 (A) In the *immediate* postoperative period, an anterior bone graft can serveas a spacer and can resist compression, provided the posterior elements are intact. Therefore, the construct is stable in *flexion*. (B) The construct cannot resist *tensile* loading and is therefore unstable in *extension*. (C) A construct that will resist compression and *tension* even if the posterior ligaments are destroyed. After incorporation of the graft, this construct offers stability in both flexion and extension. The posterior bone block fusion mass and the anterior cement or bone spacer together provide stability in flexion and extension. Some form of anterior and/or posterior instrumentation could also be used to resolve this challenge.

be necessary to include more than one normal vertebra at one or both ends of the fusion mass in order to obtain some immediate postoperative stability. This is the case when the surgical construct is relatively weak as a result of loss of structure and/or anticipated exposure to large loads.

The concept of a prophylactic fusion also involves a spacer or a bridge construct, but in addition the process must anticipate damage by an aggressive metastatic tumor that would otherwise cause spinal cord or nerve root damage or irritation secondary to structural failure. In such circumstances, fusion to normal vertebrae in addition to adjacent vertebrae is employed.

Immediate Postoperative Stability

Immediate postoperative stability is an important concept and involves the ability of the surgical construct to prevent subsequent neurologic deficit, deformity, or disruption of the spine construct under physiologic loads, prior to the contribution of any biologic processes of healing or bony union to resist potentially damaging loads. The biologic processes involved in the maturation of bone grafts may significantly alter the structure of the spinal column through changes in the mass and the distribution of the osseous material. The major indication for spinal instrumentation is the need for immediate postoperative stability.

Anterior Versus Posterior Fusion

This question is frequently discussed and debated. A number of complicated factors are involved. It is certainly important that the surgeon develop skills in both anterior and posterior fusions at all levels of the spine. The salient consideration is to determine why the fusion is being done and what one expects to achieve. Many biomechanical principles are applicable in this decision. When fusion is intended to establish clinical stability, generally the site of the major instability is considered. This is analogous to the perhaps facetious situation depicted in Figure 8-22. The fusion is done at the site where the structural damage has rendered the spine clinically unstable. This is usually the site of the major structural damage. When there is disruption of the anterior ligaments, excessive vertebral body destruction, or vertebral body resection, the fusion is best done anteriorly. When there is destruction or inability of the posterior elements to function, a posterior fusion is the procedure of choice. If a decompression is required and it becomes a source of clinical instability, here, too, fusion should be carried out at the site of the destruction necessitated by decompression. There are a number of instances in which there is a need for both an anterior and a posterior arthrodesis. The following provides an analysis of a number of surgical constructs for anterior or posterior spine fusions, along with comments about their biomechanical characteristics.

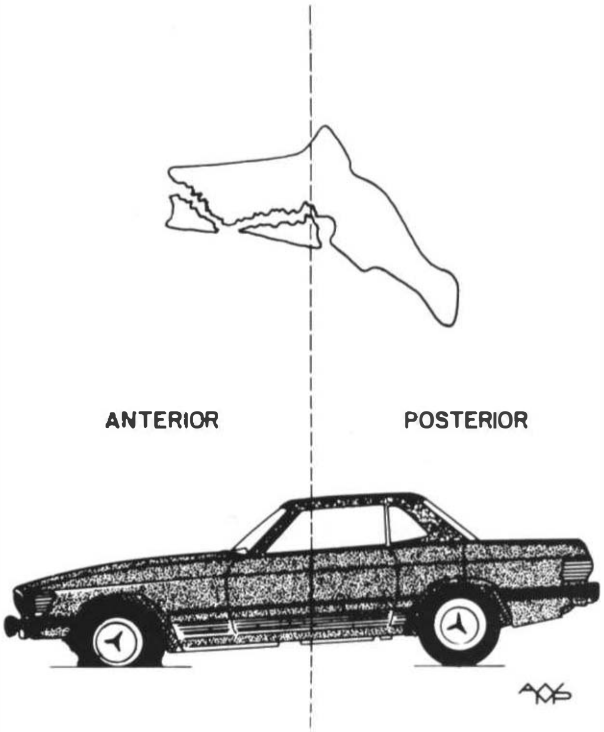


FIGURE 8-22 This analogy emphasizes the importance of evaluating the site of major clinical instability and selecting the proper surgical construct and approach to correct it (provided that the physician decides upon surgical treatment). If the front tire is damaged, the back tire should not be repaired, and vice versa. If the posterior structures of the spine are disrupted, surgery should not be performed on the anterior elements. This is the general rule of thumb in the surgical treatment of the clinically unstable spine: the surgeon should work at the site of the instability. However, there are exceptions. Also, there are situations in which anterior and posterior clinical instability can be solved by anterior or posterior surgery alone.

EVALUATION OF CONSTRUCTS IN THE OCCIPITOCERVICAL REGION

Anterior Constructs

The surgical approach and construct for this procedure has been described by DeAndrade and MacNab.[60] The exposure is essentially a cephalad extension of the Southwick-Robinson exposure of the lower cervical spine.[146] This procedure must be avoided in a person who sings high notes. There is little to discuss concerning the biomechanics of the surgical construct. The anterior surface of the occiput and the ring of C1 are roughened, and cancellous bone chips are applied in the hollow above the anterior portion of C1 and then covered over with longitudinal strips of cortical bone (Fig. 8-23). Postoperative immobilization is achieved and maintained with a halo apparatus.

The procedure is indicated when posterior stabilization is not feasible. It is important to be aware that fusion of C0 to C1 alone in the absence of an intact transverse ligament will fail to establish clinical stability. If the transverse ligament is not intact, the fusion should include C2. There is a significant biomechanical advantage in not including C2 in the fusion either anteriorly or posteriorly. Leaving the C1–C2 articulation unfused preserves a considerable amount of axial rotation (see Chap. 2).

Posterior Constructs

Simple Onlay Construct

This surgical construct for the posterior fusion of C0–C1 and C2 is uncomplicated.[211] The base of the occiput, the middle one-half to two-thirds of the posterior ring of C1, and the posterior elements of C2

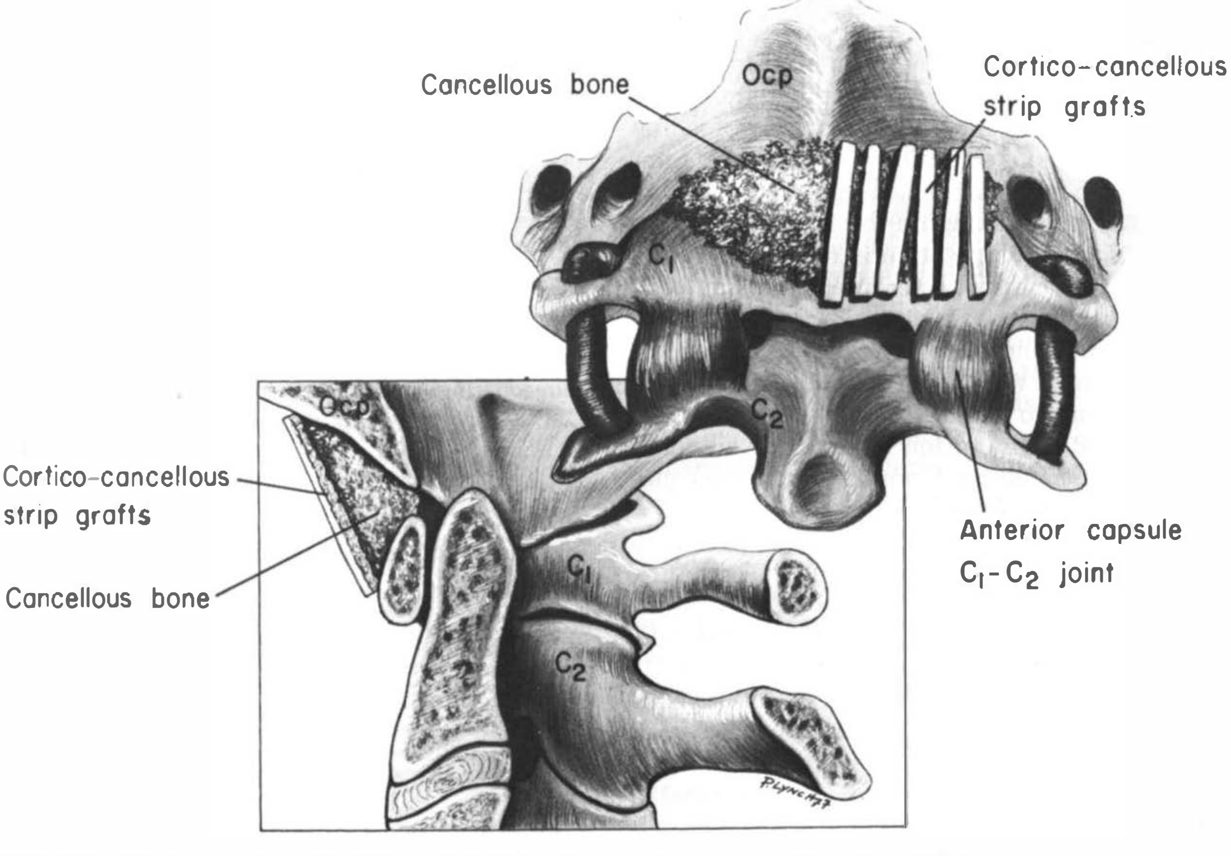


FIGURE 8-23 A useful construct for anterior C0–C1 fusion when a posterior approach or construct is not feasible. The construct has good osteogenic potential, but it offers no immediate postoperative stability. If there is clinical instability, some form of halo fixation is desirable.

are exposed. These structures are all decorticated, and cancellous chips are placed over the three decorticated structures. The patient is kept in a previously prepared plaster cast for 6 weeks. There is little to criticize about this surgical construct. It obviously provides no immediate postoperative stability; thus, it is not adequate alone when such stability is required, unless halo fixation is used. The value of this technique lies in its simplicity and accessibility. In the past we considered this to be the procedure of choice for a routine case and would advise the use of a halo device for postoperative immobilization. However, several techniques for internal fixation of C0–C1 fusions are now available. These are reviewed in the section on instrumentation. It is not yet clear which is the procedure of choice.

Construct With Wire Fixation

When fusion of the occiput to the cervical spine is required, the construct shown in Figure 8-24 provides an effective design.[114,242] The wiring and the bone graft provide some immediate postoperative stability. Both columns of the construct are able to effectively resist tensile and compressive loading during flexion and extension, respectively. Similarly, they can resist lateral bending in either direction by alternately taking up the tensile and compressive forces. Axial rotation is restrained by the anchoring effect of the posterior elements on each other as a result of their attachment to the graft. There is ample cancellous bone in the graft–recipient bed interface. We suggest that in most instances it is satisfactory to fuse distally only as far as C2, leaving C3 out of the fusion mass. This should be completely adequate and preserves precious motion, especially axial rotation.

Placing holes in the occiput involves the serious risk of bleeding from the sagittal sinus. Certainly, this procedure should not be used when a posterior C1–C2 fusion would suffice. This construct is indicated when the occipital-axial joint or the C1–C2 joint must be stabilized and it is not useful or possible to employ the posterior ring of C1 in the fusion mass. This construct or some modification of it is also indicated when a massive fusion is required to bridge a grossly unstable, structurally impoverished cervical spine.

Occipitocervical fusions posteriorly constructed are recommended for rheumatic patients with migration of the dens into the foramen magnum. Posterior fusion is suggested because the anterior elements are thought to be weaker in these rheumatoid patients in whom anterior fusion did not yield good results.[238]

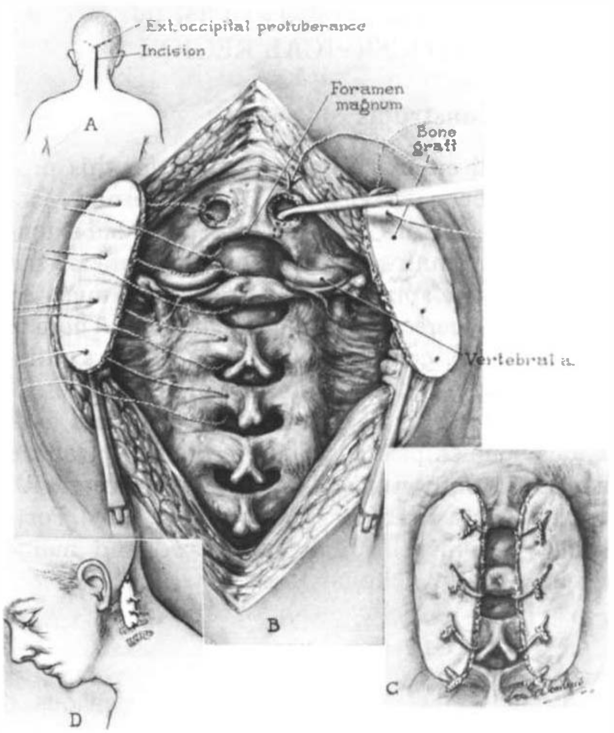


FIGURE 8-24 A biomechanically stable construct for arthrodesis of the occiput to the cervical spine. The advantages are its stability against all modes of motion and the option for stable facet fusion should the posterior vertebral elements be absent. The disadvantage is the necessity of placing two holes so close to the sagittal sinus. *(Robinson, R. A., and Southwick, W. O.: Surgical approaches to the cervical spine. In American Academy of Orthopaedic Surgeons: Instructional Course Lectures. vol. 17. St. Louis, C. V. Mosby, 1960.)*

EVALUATION OF SURGICAL CONSTRUCTS IN THE UPPER CERVICAL SPINE

Anterior Constructs

Bilateral Screw Fixation

This technique was designed as a method for internal fixation of a fractured odontoid process.[11] The surgeon, dissatisfied with other methods of managing this fracture, devised this technique of *bilateral* screw fixation of the lateral mass of C1 to the body of

C2 (Fig. 8-25). The proximal portion of the anterolateral approach described by Henry may be employed for exposure.[126] A 2.5-cm (1 in) screw is inserted through the lateral mass of C1 into the body of C2. The angle of the screw is determined by placing the drill at the anterior surface of the mastoid process and through the tip of the transverse process of C1, with the head in neutral rotation. The described landmarks keep the drilling and screw tract anterior and lateral to the vertebral artery. A guide to assure safe and proper orientation of the screw has been devised by E. H. Simmons. A neck splint is worn for 6 weeks as a precaution until the time that flexion/extension films show a stable, healed arthrodesis requiring no further treatment.

We believe that this technique offers a secure fixation of C1 to C2. Axial rotation and flexion/extension are solidly fixed. Although it may be a bit aggressive as treatment for a fractured odontoid, it has appeal as a fusion technique in situations where the posterior ring of C1 is not available for arthrodesis (e.g., the posterior ring has been removed, is congenitally absent, or is hypoplastic, necrotic, or detached from the anterior portion of C1). For atlanto-axial joint arthrodesis, the technique shown is supplemented by some bone graft in a trough along the anterolateral aspect of the trough between the C1–C2 articulations. This construct can be expected to offer good immediate postoperative stability.

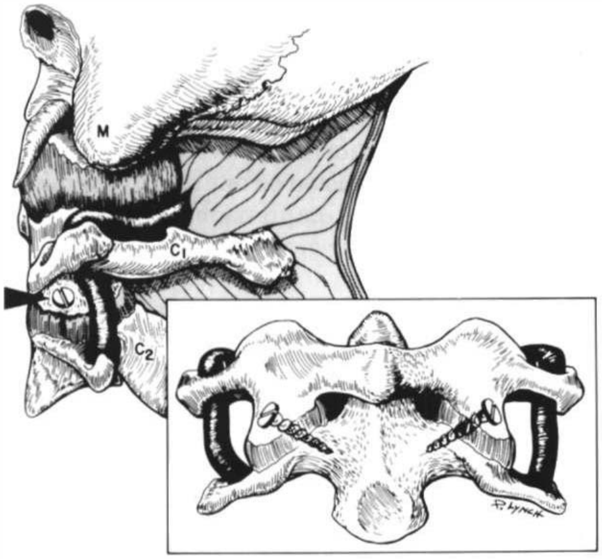


FIGURE 8-25 Barbour C1–C2 screw fixation—a stable construct for the anterior arthrodesis. It required two operations through surgically challenging anatomic regions. The two articulations are denuded of cartilage; the joint space is filled with cancellous bone chips. The advantage is that it provides immediate postoperative stability and can be used when there is an absent, diseased, or structurally useless posterior ring of either C1 or C2. Operating on only one side does not suffice.

The disadvantage of this procedure is that it requires two operations through a surgically rather difficult anatomic approach that is close to several important structures.

Anterior Screw Fixation of the Dens

The placement of one or preferably two screws across a fractured dens has been recommended as a primary treatment of the injury.[17] This is biomechanically sound in that clinical and anatomic evidence suggests that displacement of the fracture fragment may be a factor in the nonunion, and the problem can be solved without having to eliminate axial rotation should a C1–C2 fusion be required (see Trauma chapter). The liability of the procedure is that there is a risk of major neurologic complications, and other less risky methods are generally successful.[47]

Fang Construct

The anterior approach to C1–C2 through the mouth, described by Fang and colleagues, may be useful in some cases for drainage of abscess, excision, or biopsy of tumor or removal of the odontoid (Fig. 8-26).[74] However, for fusion, this is not one of the more biomechanically sound constructs. It is better to have more of an interface between the graft material and the vertebrae to be fused. There is not a secure fixation of the bone graft to the recipient site. Therefore, if this procedure is used, we recommend a halo fixation. Finally, there is an increased risk of osteomyelitis involved in the transoral approach.

Retropharyngeal Approach

McAfee and co-workers[189] have described and thoroughly demonstrated the efficacy of the retropharyngeal approach for the upper cervical spine. These surgeons have shown that it is possible to complete a variety of surgical procedures and reconstructions without the risks of the transmucosal approach. It is probably preferable to avoid operating on this region through the mouth when an implant is planned as part of the procedure.

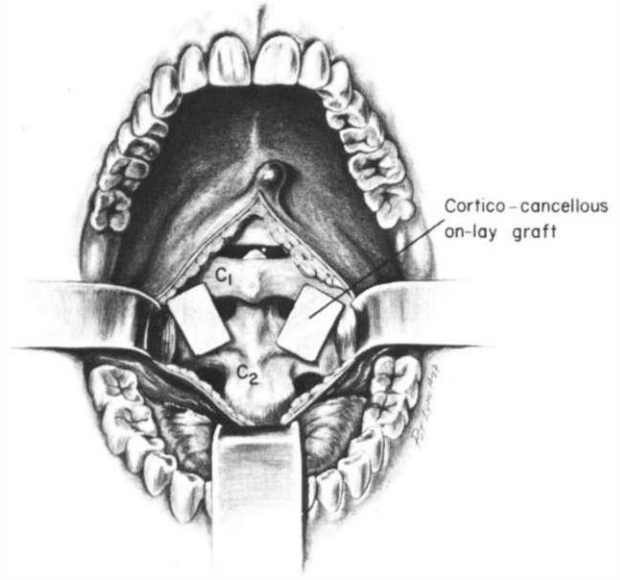


FIGURE 8-26 This construct is not stable. In addition, the surgeon must proceed through the oral cavity. We believe that the potential for infection must be greater than with other approaches. The procedure is useful for drainage of infection, biopsy of tumor, and excision of the odontoid.

Posterior Constructs

Comment

We would like to include two points about posterior cervical spine surgery. The first may be most useful in relation to posterior C1–C2 fusions. Longitudinal posterior midline cervical scars sometimes leave a great deal to be desired with regard to cosmesis. Thus, in situations where the need for extensive cephalocaudal access is not a factor, such as C1–C2 fusions, we use a high transverse incision that can readily be covered by the hair, fashion allowing. It is also noteworthy that Langer[187] described transverse anatomic skin lines in the posterior cervical region, particularly the more cephalad portion.

Another point to be considered in posterior cervical spine surgery is the following. Nolan and Sherk[215] dissected the posterior cervical muscles and noted that the semispinalis cervicis and capitis muscles appeared to be primarily responsible for extension of the cervical spine and head. They then established a biomechanical model and calculated that at least 14.38 kg are required to balance the cervical spine in the prone position. In view of the anatomic studies and biomechanical model, they suggested that the extensors play an important role in the dynamic stabilization of the cervical spine and head. The clinical biomechanics of these observations, along with some additional considerations, are discussed in the context of the effect on clinical stability (p. 311). We suggest that one avoid removal of the attachments of the semispinalis cervicis to the ring of C2 when possible and attempt to anatomically reapproximate the muscles at the time of closure.

Brooks Construct

The success rate of posterior fusions of C1–C2 in general is not considered to be especially good.[91] A variety of techniques are described.[31,60,96,168] We believe that the Brooks fusion is biomechanically sound, and it has been shown to be effective with clinical trials.[31,106] The surgical construct is shown in Figure 8-27.

Axial rotation is the major motion that occurs at the C1–C2 level. This rotation, along with flexion/extension that includes anteroposterior translation, is the movement that is clinically the most important one to control in order to achieve immediate postoperative stability. The bone grafts wedged and fixed circumferentially create a "friction block" effect and efficiently prevent axial rotation (Fig. 8-28). There is controversy about the amount of lateral bending at this level. If, however, there is lateral bending, the construct is effectively designed to prevent it. Tension is resisted on one side by the circumferential wires, and compression is resisted on the other side by the bone graft. There is one additional aspect of this design that is mechanically useful. The two wedge-shaped configurations of graft allow a snug approximation of recipient site and graft and a control of the amount of flexion/extension between C1 and C2 without bringing the posterior elements of C1 and C2 too close together. A direct approximation of the rings of the atlas and the axis could cause too much extension and could aggravate pathologic aspects of a lesion in the neural canal or the anterior elements. To assure proper separation between the posterior elements of C1 and C2, the vertical dimension of the graft when in place should be 1 cm.[31] The wedges of bone are removed from the ilium and fashioned so as to place cancellous bone at both interfaces of the fusion construct (Fig. 8-12:5). The wedged configuration also prevents graft migration toward the spinal cord. The relative mechanical advantages of the Brooks construct as compared with fusions involving simple midline wiring[96] and bone grafting are illustrated in Figure 8-28. Although there is the slight mechanical advantage of using two wires on each side, we suggest that the major practical biomechanical goals may be achieved with only one doubled or twisted wire around the middle of each of the interfaces of the bone graft–recipient site, as described by Brooks, who reported a success rate of 11 out of 12 fusions (Fig. 8-29).[31] Thus, only two rather than four wires need be passed under the laminae of C1 and C2. This reduces risk of neural damage and also shortens the operating time without any significant loss of mechanical advantage. The principle of the Brooks construct provides excellent immediate postoperative stability. The construct requires an orthosis of only minimal or intermediate postoperative control. We recommend a cervical brace with a thoracic support worn for 6 weeks.

A recent biomechanical study has compared stability provided by an anterior procedure (facet screw fixation according to Magerl) with stability provided by three posterior procedures. Grob and co-workers used fresh human cadaveric C0–C3 specimens for a comparative study of four stabilizing procedures of the C1–C2 joints.[107] To produce an unstable specimen, they transected most of the ligamentous connections between the atlas and axis (*i.e.*, tectorial membrane, both alar ligaments, transverse ligament, and left capsular ligament). Each injured specimen was stabilized in turn by Gallie, Brooks, and Magerl facet fixations as well as by Halifax clamps. With the use of pure moment applications and stereophotogrammetric techniques for motion measurements, multidirectional stabilities were determined for the intact, injured, and stabilized specimens. This provided a direct comparison of the stabilizing capabilities of the three surgical procedures.

The Gallie procedure proved to be the least stable of the four procedures under all physiologic loads.

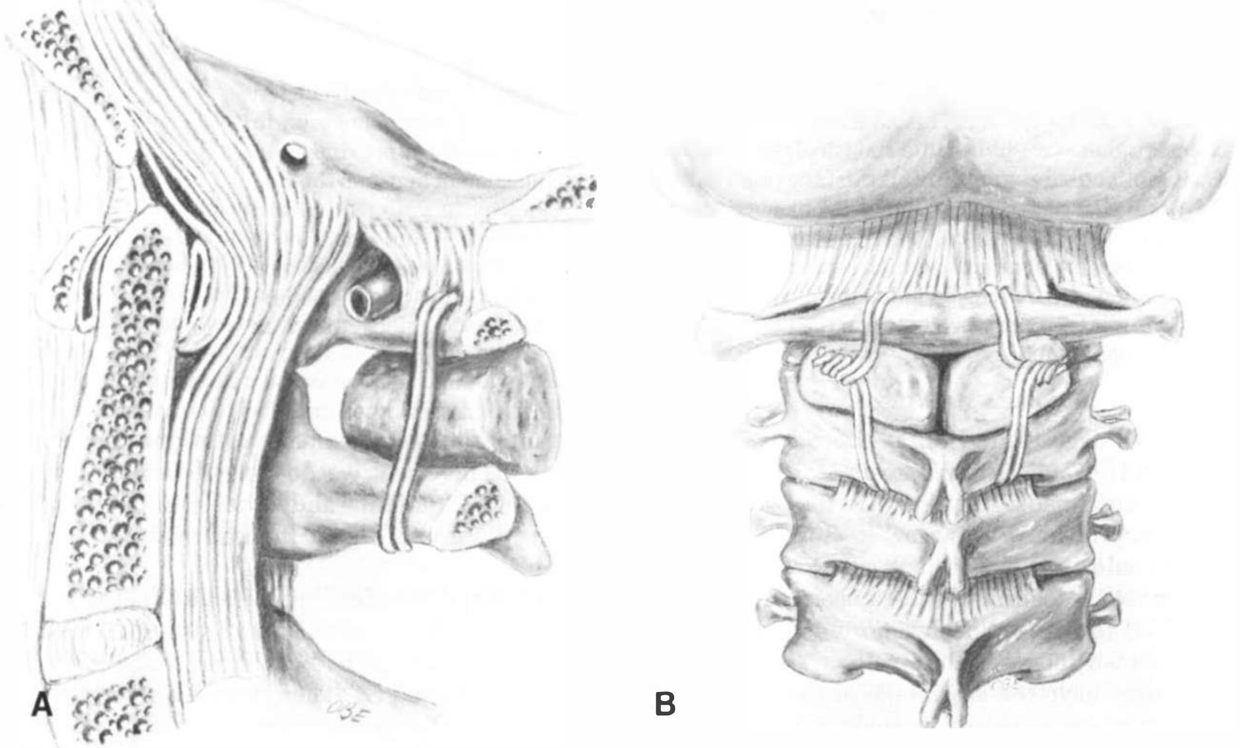


FIGURE 8-27 **(A)** A posterior, sagittal plane view of the Brooks fusion construct. The wedge of corticocancellous bone, in conjunction with the wire that incorporates it with the lamina of C1 and C2, is the essence of the construct. **(B)** A posterior view showing the two doubled wires, one on each side. (Brooks, A. L., and Jenkins, E. G.: *Atlanto-axial arthrodesis by the wedge compression method.* *J.* Bone *Joint* Surg., 60A:279, 1978.)

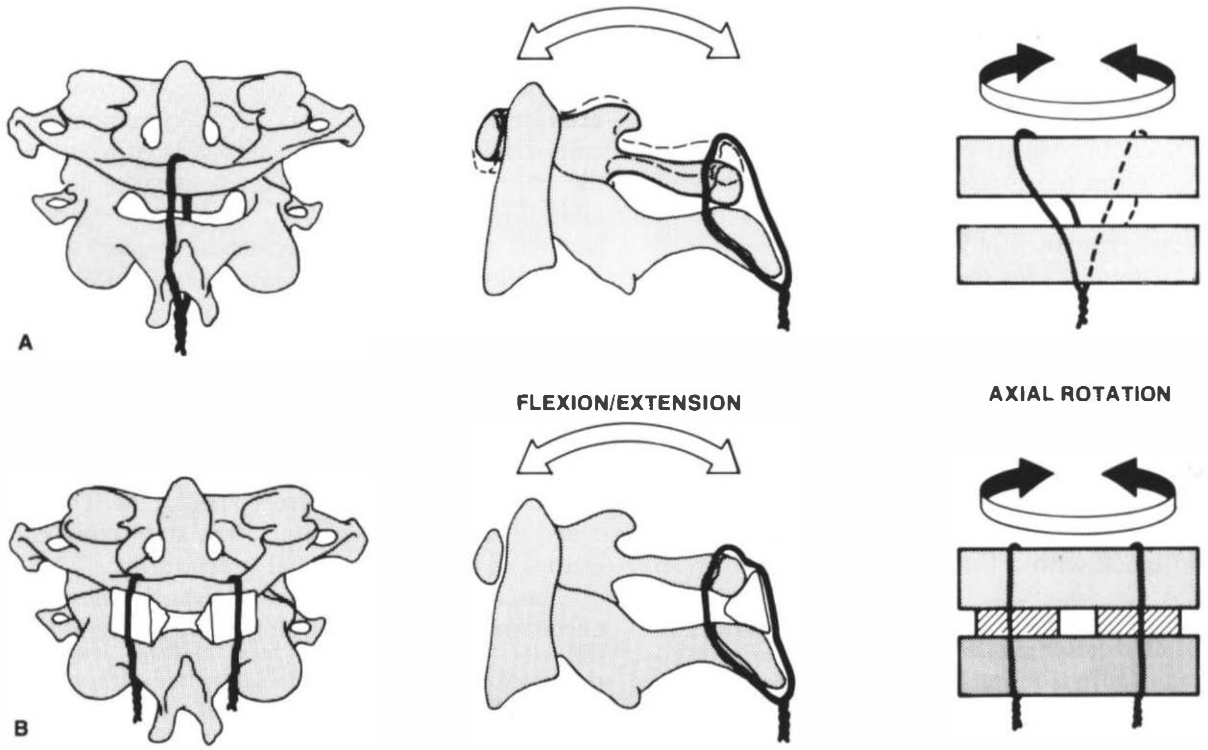


FIGURE 8-28 An illustration of the biomechanical advantages of the Brooks construct. **(A)** A single midline wiring. This construct would be relatively stable in flexion; however, in extension there would be little stability, since the two rings would readily approximate. In axial rotation there is nothing to resist the relative horizontal displacement between the ring of C1 and that of C2. **(B)** With the Brooks construct, there is stability in both flexion and extension. The flexion is restrained by tension in the circumferential wires, and extension is restrained by the bone graft, which serves as a buttressing block. Rotation is resisted by some combination of wire tension and bone block, but this time the mechanism is one of friction. The bone grafts compressed between the two posterior rings serve as friction blocks and offer stability against axial rotation.

The Brooks, Magerl, and Halifax procedures proved to be equally stable under flexion, extension, and lateral bending loads. However, in axial rotation the most stable procedure was Magerl, closely followed by Brooks and then the Halifax clamp.

Several other techniques for successful posterior fusion of C1 to C2 have been reported.[79,80,194] Some of these constructs are technically less dangerous because they have the advantage of having to pass only one doubled wire under one lamina. This is an important consideration, especially for the surgeon who has not had extensive experience operating in this region. One of the constructs described by Fielding is shown in Figure 8-30. Although we are not aware of any biomechanical studies, in our opinion this construct is not as stable as the one described by Brooks. Therefore, we do not advocate its use for rheumatoid patients and other clinical situations where there is a higher than normal risk for nonunion.

Nordt and Stauffer[216] reported two cases of quadriplegia in patients with Down's syndrome who had sublamina wires placed in the presence of an anteriorly subluxated C1. There was simply not enough space to pass the wires without damage to the spinal cord. Although this occurred in Down's syndrome patients, the risk is high in other patients with persistent subluxation for whom Steel's rule of thirds no longer holds. If reduction cannot be achieved with preoperative traction, some other technique,

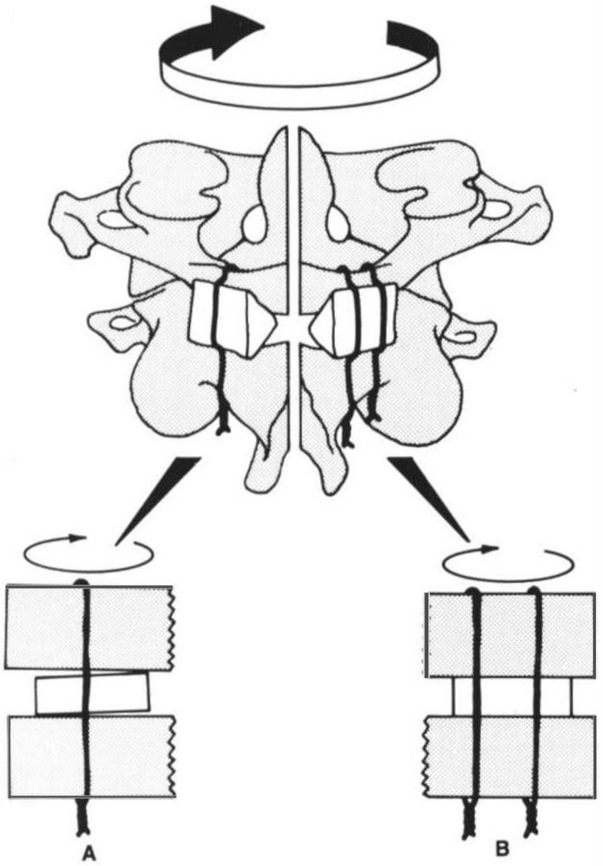


FIGURE 8-29 Biomechanical considerations in the use of a single or a double wire for each side of the Brooks fusion construct. **(A)** When there is axial rotation with a single wire, the friction may cause the graft to tilt. **(B)** This is less likely to happen with a double wire because the two wires at either end of the graft more effectively prevent tilting. This is more of a theoretic than a practical point, and its advantage must be balanced against that of passing two wires rather than four under the posterior elements of the two vertebrae. A construct with one wire on each side of the midline is thought to be strong enough to resist displacement with flexion/extension.

such as onlay graft or an anterior approach, should be considered.

There is a useful modification of the Brooks fusion for the surgeon faced with a central defect in the posterior arch of C1. Callahan and colleagues described the following alternative.[40] Since there is no posterior arch to pass the wire *around*, one wire is passed separately *through* the available posterior elements on each side. The rest of the procedure is as described, using only one wire on each side.

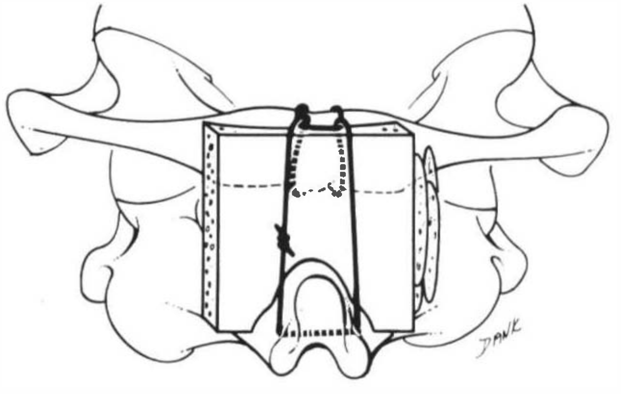


FIGURE 8-30 This is another sound construct for posterior C1–C2 fusion. It is a simple design that requires only the passage of one loop of wire around the ring of the atlas. The construct is most effective in preventing flexion. *(Fielding, J. W., Hawkins, R. J., and Ratzan, S. A.: Spine fusion for atlanto-axial instability. J. Bone Joint Surg., 58A:400, 1976.)*

EVALUATION OF SURGICAL CONSTRUCTS IN THE MIDDLE AND LOWER CERVICAL SPINE

There are several techniques for anterior cervical spine fusions. Only the most important and frequently used ones are discussed here. Careful attention to good technique is important in all these procedures.

Anterior Constructs

The Smith-Robinson Construct

This construct has several biomechanical advantages (Fig. 8-31).[268] The preparation of the graft bed removes the intervertebral disc and provides ample exposure for midline and lateral decompression of the anterior cord and nerve roots. The graft itself provides adequate support against vertical compression. The cancellous portion in contact with the vertebral end-plates readily permits revascularization and incorporation. The construct allows all or most of the vertebral end-plates to be left intact. The interspace is usually spread 7 mm, which opens the intervertebral foramen and reduces invagination of the yellow ligament. This procedure is best suited for the treatment of cervical spondylosis. Even though some surgeons think that only annulus re-

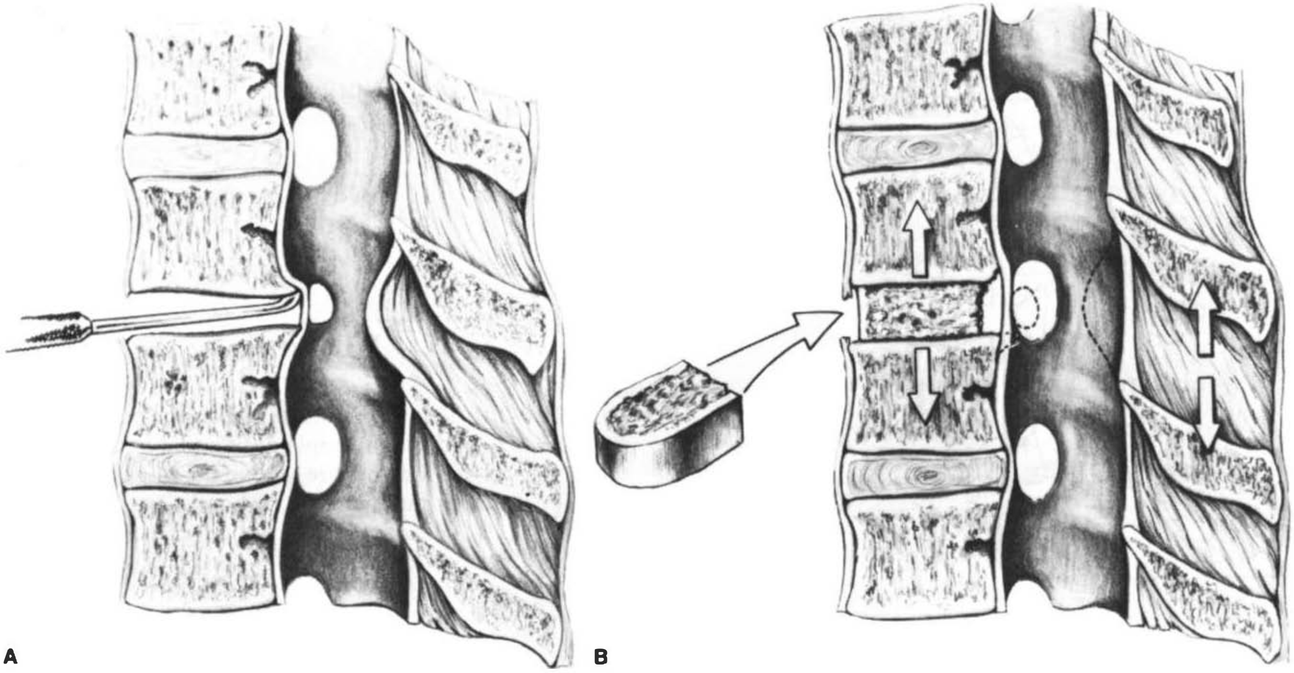


FIGURE 8-31 (A) A narrowed intervertebral foramen, invaginated yellow ligament, and a curette removing an osteophyte. This procedure can also be done with a neurosurgical burr. (B) The horseshoe graft is being inserted where the disc has been removed, exposing the posterior longitudinal ligament, or the spinal cord if the surgeon desires. The graft should be approximately 7 mm high and cut with a double-blade saw if possible, such that its top and bottom surfaces are parallel. If there is any tendency for it to be wedge-shaped, extrusion may occur. The graft immediately separates the interspace, opens the neural foramen, and reduces the yellow ligament invagination, and subsequently, with successful arthrodesis, the osteophyte resorbs.[318] This is the procedure we employ as the construct of choice when surgery is indicated for cervical spondylosis.

moval is necessary for cervical spondylosis with neck, shoulder, and arm pain, we believe that the bone graft has some advantages. It provides immediate interspace opening, relieves nerve root compression, and the subsequent arthrodesis is believed to be helpful in the relief of pain from any associated arthritis. Since the disc is cleaned out and the interspace opened, it seems reasonable to offer the patient the additional benefits of a bone graft.

The disadvantages of this procedure include relatively limited exposure to the spinal cord. When more than two interspaces are to be fused, a construct that uses a trough is preferable. Such a procedure is more convenient, and there are fewer interfaces between bone graft and recipient bed that must be incorporated. There is less probability of successful fusion when more than two interspaces are required.[315]

Among the several modifications of this procedure, we chose the following to include here. A simple 180° reversal of the orientation of the "horseshoe" bone graft has been suggested.[16] This procedure has the following advantages: the cortical end is posterior in the interspace where the greatest distraction is sought, and should the graft slip forward, the protruding end can readily be removed, leaving the stronger portion of the graft in the interspace.

A recent work by Geibel and colleagues[97a] reported on 55 patients in whom the bone graft was reversed. The surgeons reported results comparable with reported series with unreversed grafts. However, the procedure was thought to be technically less demanding with the reversed grafts. The question as to whether or not to reverse the graft is not yet resolved.

A recent work by Gore and Sepic[104] confirms most of the basic truths about this procedure as a treatment of cervical spondylosis. That is, the results are 80–90% satisfactory, there are no serious complications, posterior osteophytes are resorbed or not progressed, and patients with radicular pain have a better prognosis. We continue to suggest this as the procedure of choice. However, we must acknowledge that the issue of whether discectomy without fusion is preferable is not yet settled. Rosenorn and co-workers,[246] in a prospective randomized study, compared discectomy with fusion with discectomy alone and reported superior results in the latter group.[246] Of statistical significance is the fact that patients with discectomy alone returned to work within 9 weeks of surgery more often than patients with discectomy and fusion.

The Bailey-Badgley Construct and Modifications

This construct has some useful biomechanical advantages.[9] Figure 8-32 shows the Bailey-Badgley technique for fusing one and three FSUs. Also, a modification of the construct that can be used in a similar manner is shown. The modification is thought to supply additional mechanical support by providing more cortical bone and increasing the anteroposterior length of the bone graft. There is the possibility of conveniently fusing several interspaces. If exposure of the anterior midline portion of the spinal cord behind the vertebral body is necessary, it is readily achieved with this technique. The strength in the immediate postoperative period is adequate, and there is the biologic advantage of cancellous to cancellous bone contact across the fusion interfaces.

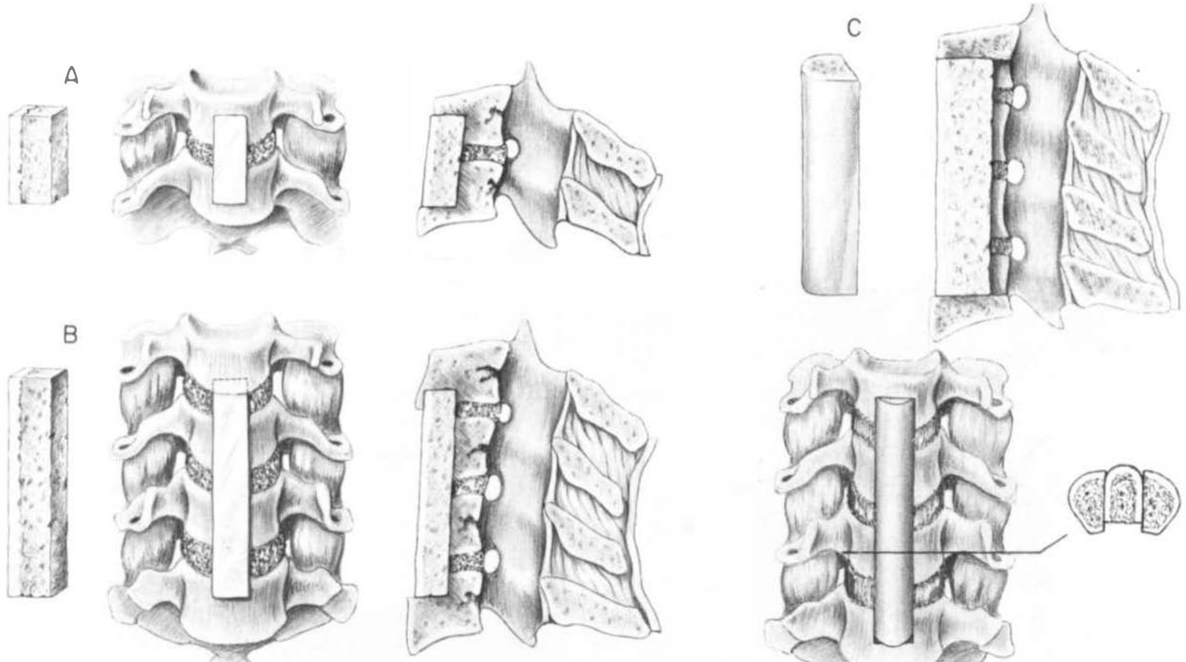
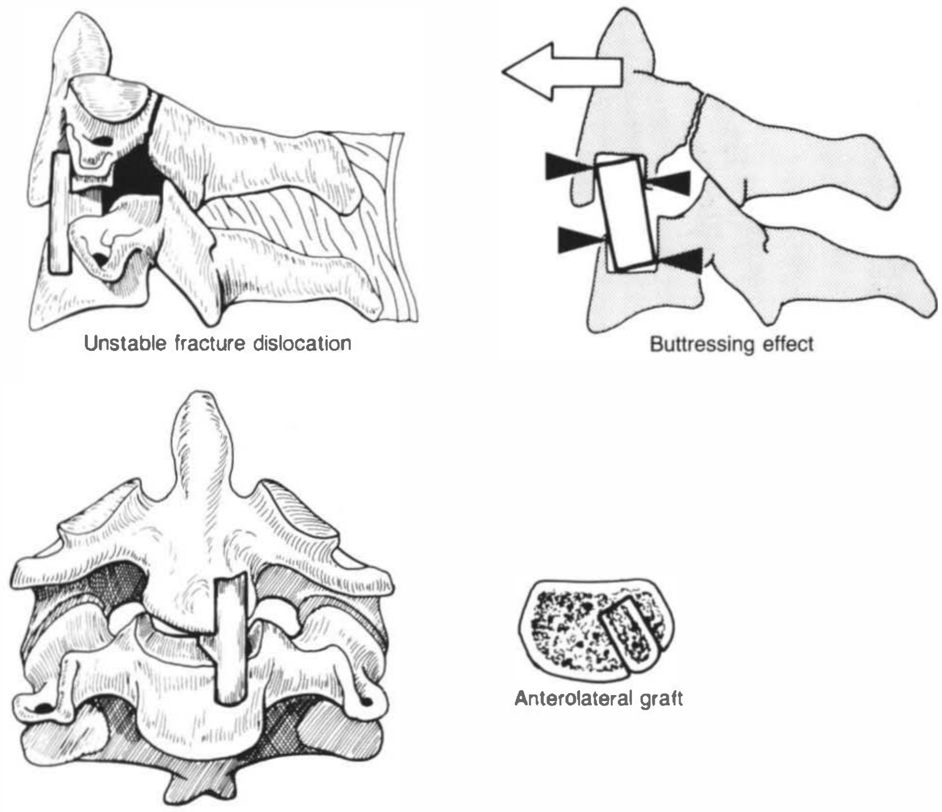


FIGURE 8-32 (A) This is a diagram of the construct described by Bailey and Badgley, as used for anterior fusion at one level. The interspace is packed with cancellous bone. (B) A modification of the procedure in which the graft is tunneled into the uppermost vertebral body and slotted into the lowermost one. That portion of the disc not removed for placement of the bone graft (as shown in C) may be left intact to provide some degree of clinical stability. (C) Another modification, which employs a horseshoe-shaped configuration of iliac bone that is stronger in resisting axial compression. This also shows how the procedure may be used to decompress certain anterior areas of the cord. The option of leaving a portion of the disc for purposes of clinical stability is seen.

One of the very important mechanical advantages of this technique is that the trough bed allows the surgeon the option of leaving some anterior stability through the annular fibers on both sides of the trough. This can be useful in a situation where there is instability posteriorly and there is a necessity to fuse anteriorly (e.g., following multilevel extensive posterior element removal). Cattell and Clark found the construct useful in this situation.[43] If the entire annulus has to be removed to insert a horseshoe graft, then in the immediate postoperative period the patient with posterior element injury is unstable not only posteriorly but also anteriorly. It is certainly possible for a graft to slip in such an unstable situation. If the peripheral annular fibers (which are the strongest because of Sharpey's fibers) can be left intact, there is some preservation of intrinsic stability. Stauffer has pointed out the liabilities of anterior interbody fusions in the presence of post-traumatic posterior instability.[273]

The Bailey-Badgley technique has been modified by using iliac bone with three surfaces of cortex, as shown in Figure 8-32C. In addition to this, we have found the trough construct to be useful in situations of clinical instability in which there is significant displacement between vertebrae that cannot or need not be reduced. With significant anterior displacement, there is no good location in which the graft can be placed. However, with an anterolateral or a lateral interbody placement of an iliac graft in a trough, a useful construct is developed (Fig. 8-33). The boneblock immediately locks the FSU so as to resist further displacement. The lateral positioning allows for the construction of a smooth trough of uniform depth in both vertebrae, without the step-off that is inevitable with anterior placement. Furthermore, by taking only a portion of the annulus, any residual stabilizing influence is preserved. We have used this construct in the cervical spine as well as in other regions under similar circumstances (Fig. 8-38).

We also recommend for your consideration a modification by Gore.[103] This procedure uses two bone grafts fashioned to be keyed and locked into partially excavated vertebral bodies. We consider


Unstable fracture dislocation

Buttressing effect

Anterolateral graft

FIGURE 8-33 An illustration of the principle involved and the construct employed for the placement of a trough graft anterolaterally across an interspace in order to provide immediate postoperative resistance against sagittal plane translation. The mechanics are such that when there is either anterior or posterior sagittal plane motion, a portion of the graft buttresses against the osseous structures of the two vertebrae.

that this is also a situation in which there is a relative indication of plate and screw fixation of C5–C7 bridging the grafted level. The role of the hardware would be to provide enough stability to hold the graft in place rather than to provide immediate postoperative stability, particularly if the posterior elements were not functional.

Cattell and Clark have employed another modification of the basic Bailey-Badgley anterior trough construct. They use a tibial graft that is tunneled into the body of the upper vertebra through its inferior end-plate and wired into the lowermost vertebral body in the fusion (Fig. 8-32B). We suspect that this modification offers little in the way of significant immediate postoperative stability. However, it is relatively more stable than the standard Bailey-Badgley construct with regard to the probability of displacement of the bone graft.

Boni and associates more recently demonstrated the use of this technique for the treatment of multilevel cervical myelopathy due to several midline osteophytes encroaching on the anterior portion of the cervical canal.[19]

The disadvantages of the Bailey-Badgley technique and its modifications are minimal. There is considerable bleeding when it is necessary to violate the central cancellous portion of a vertebral body.

The Cloward Construct

This construct is well instrumented, convenient, and provides good visualization of the midline anterior portion of the spinal cord at the interspace and for about 1 cm on either side of it.[50] The construct has been reported to collapse in a significant number of instances.[153] This is probably due to the fact that although the graft configuration itself is of adequate strength,[314] the total construct may be lacking after a period of time, possibly during the early phases of creeping substitution, when the graft is relatively more osteoporotic. Experimental vertical compressive loading of the construct suggests that failure may be caused when the two coinlike cortical edges cut into the adjacent cancellous bone (Fig. 8-34). The Cloward construct, like the Bailey-Badgley construct, has the advantage of preserving some degree of stability by leaving a portion of the intact annulus fibrosus attached to the vertebral bodies. However, we see no reason to use this instead

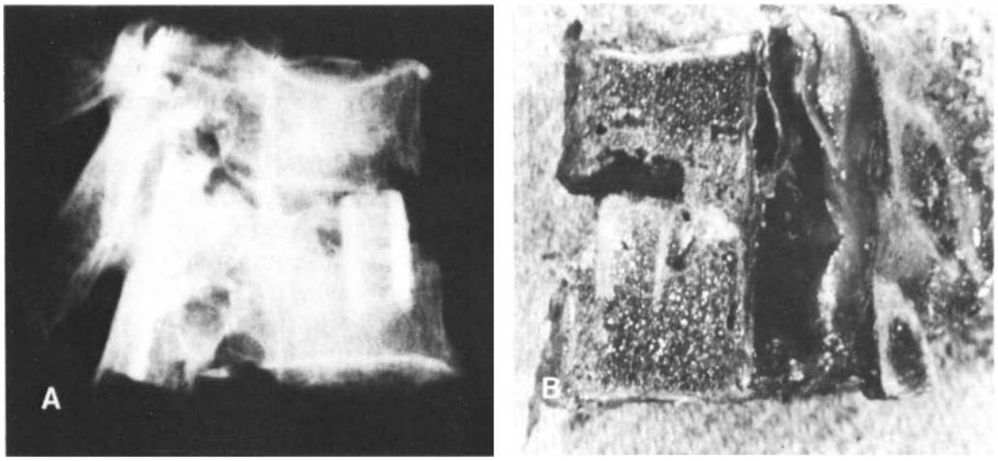


FIGURE 8-34 (A) A lateral radiograph of an experimental dowel construct, loaded to failure. (B) A sagittal section of the actual specimen oriented as a mirror image of the radiograph. In both pictures, it is apparent that the dowel graft migrates into the cancellous bone of the vertebral body as a result of the vertical compression loading of the vertebra. This is at least part of the mechanism of collapse that is so commonly seen clinically with this construct. We do not consider it to be sound biomechanically. *(White, A. A., Jupiter, J., Southwick, W. O., and Panjabi, M. M.: An experimental study of the immediate load bearing capacity of three surgical constructions for anterior spine fusions. Clin. Orthop., 91:21, 1973).*

of the Bailey-Badgley construct when clinical stability is important.

Biomechanical Comparison of Constructs

The immediate, vertical, compressive load-bearing capacity of three surgical constructs designated as Smith-Robinson, Cloward, and modified Bailey-Badgley procedures were tested experimentally.[315] The results are given in Figure 8-35. The immediate postoperative load-bearing capacities of the three constructs are listed in order of decreasing strength as follows: Type I, Smith-Robinson; Type II, Cloward; and Type III, Bailey-Badgley (modified). The failure loads were greater than the expected range of physiologic loads in these static tests. It would be more useful to know the relative load-bearing capacity of these constructs as they undergo creeping substitution. It is known that they will become mechanically weaker during the phase of creeping substitution. In dogs, experimental studies show that transplanted bone is greatly weakened between 6 months and 1 year after the transplant. It is reasonable to assume that the grafts continue to have the same relative strengths. Therefore, the relative load-bearing capacities of the three constructs, with regard to mechanical function, become quite important.

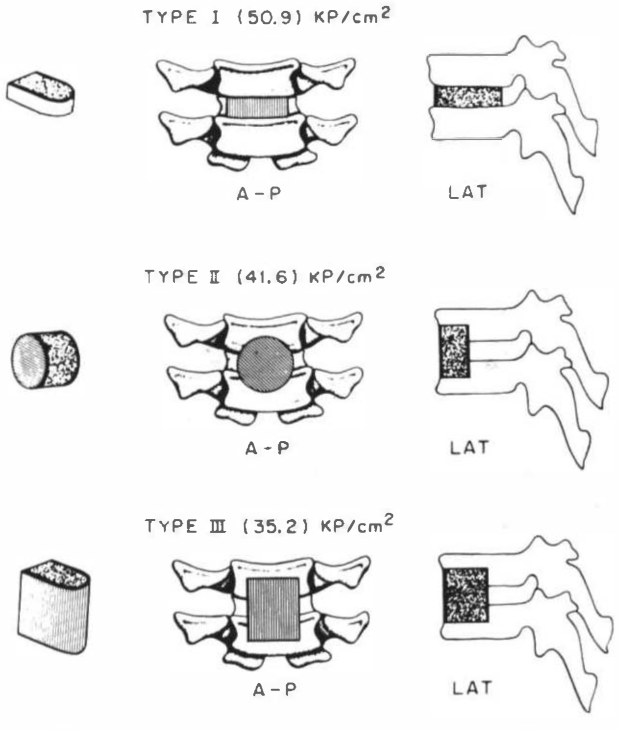


FIGURE 8-35 Graft configuration: how the graft fits into vertebrae, and how the vertebrae are altered to receive it. The numbers are mean values for the load-bearing capacity of each of the three surgical constructions. *(White, A. A., Jupiter, J., Southwick, W. O., and Panjabi, M. M.: An experimental study of the immediate load bearing capacity of three surgical constructions for anterior spine fusions. Clin. Orthop., 91:21, 1973).*

The Keystone Graft Construct

This construct, described by Simmons and Bhalla, has some biomechanical advantages.[266] Mechanical studies comparing the keystone graft with the dowel (Cloward) graft revealed some useful information. Studies of surface area, an important consideration in bone graft surgery with respect to fusion and incorporation, showed that for fusion of one interspace, the rectangular (keystone-shaped) graft had approximately 30% more surface area than the cylindrical graft of comparable size. For a fusion of two interspaces, the surface contact area was 70% greater in the rectangle than it would be with two separate cylindrical grafts.

The immediate postoperative stability of these grafts was also studied. In autopsy specimens in which the two surgical constructs were created, experiments were carried out to test flexion/extension and lateral bending. The flexion/extension studies were carried out with a constant ejection force applied to both types of grafts. The dowel grafts were extruded with 20–25° of extension, but the keystone grafts were not extruded before there was fracture and complete disruption of the spine. The two constructs were compared in the lateral bending mode by measuring the relative motion between the graft material and the recipient vertebral body. The keystone construct required four times more lateral bending force to produce such motion than did the dowel construct.[266] The investigators pointed out one additional mechanical point. If there is concern about extrusion, the graft should be placed as close as possible to the posterior portion of the vertebral body.

The investigators reported a clinical series of keystone and Cloward constructs in which there was a greater fusion rate and more relief of pain with the keystone construct.[266] This occurred despite the fact that the series of keystone constructs included a higher incidence of multilevel fusions. We believe that the keystone technique has some important advantages with respect to sound biologic and mechanical principles. The configuration and position are effective in providing excellent immediate stability against extrusion and motion in lateral bending. The large surface area of cancellous to cancellous bone contact should provide sound arthrodesis. If a trough construct is indicated and the surgeon is able to build a keystone construct, we recommend this as the procedure of choice.

Wedge-shaped configurations of bone grafts have been employed in anterior cervical constructs to compensate for wedging of vertebral bodies secondary to trauma.[302, 303] We advise against the use of any *unsecured* configuration of bone graft in which there is a wedge. If such a configuration is used with the base of the wedge facing anteriorly, extrusion is a risk; when the base is facing posteriorly, spinal cord impingement is a risk if there is nothing to block migration of the bone graft. The keystone graft, a trough graft, and a graft with carefully carpentered parallel surfaces are mechanically more sound, being less likely to become displaced anteriorly. The keystone construct, followed by the Bailey-Badgley construct, is the construct of choice for patients with multilevel anterior arthrodesis.

The Notched Fibula Construct

This technique, described by Whitecloud and LaRocca,[319] is similar in principle and indication to the Bailey-Badgley trough graft, except for the graft material. The construct is shown in Figure 8-36 and was designed to prevent collapse and extrusion, which would improve the success rate for arthrodesis. It is recommended for multilevel fusions. The initial experience shows more success with prevention of collapse than with prevention of extrusion. There have been problems with extrusion in three of 20 cases. The fibula has been described as an excellent source for grafting in the cervical spine.[104] We believe that the construct has the advantage of pre-

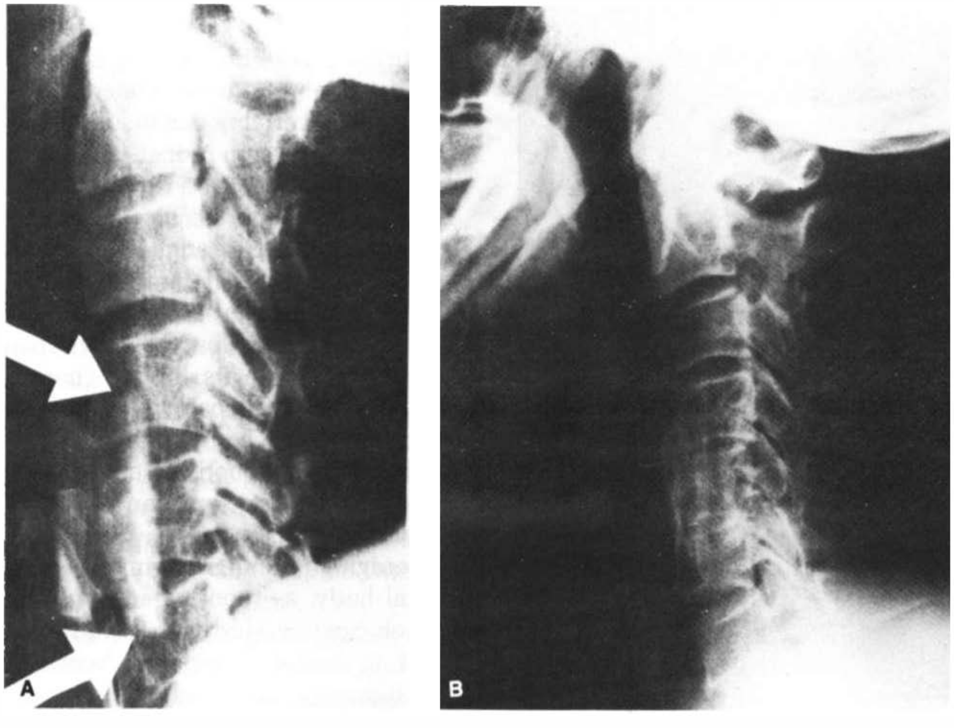


FIGURE 8-36 **(A)** The strength of the fibular graft and notching to straddle the anterior cortex (see arrows) of the vertebral body are the main features of this construct. **(B)** A graft solidly incorporated into the three vertebral bodies. Although it has advantages, we consider the disruption of a normal fibula and the long period of time required for incorporation of this type of graft to be disadvantages. (*Whitecloud, T. S., and LaRocca, H.: Fibula strut graft in reconstructive surgery of the spine. Spine,* 1:33, 1976.)

venting collapse. The notching technique is a useful way to lock in the graft.

Plate and Screw Fixation

The radiograph shown in Figure 8-37 demonstrates a form of fixation that is increasingly used. One surgeon has reported the successful use of this construct in 38 patients.[247] The objections about screw loosening, migration, and damage to vital structures may have been overemphasized. It is possible with proper radiographic monitoring to recognize this in time to prevent any catastrophe. We are not advocating this construct; however, it may be a reasonable option in a situation where immediate anterior postoperative stability is crucial.

FIGURE 8-37 This is included to suggest that surgical constructs in the spine employing plates and screws for anterior fixation of the spine need not be rejected without evaluation. There is not, to our knowledge, any documented basis to categorically consider such a construct wrong or undesirable.

Although progress has been made, as presented in Part 4 of this chapter, our current thoughts are that those things which can be achieved *without* internal fixation ought to be achieved that way. Nevertheless, there is a place for the appropriate use of anterior plates and screws. The precise indications have not yet been definitively determined.

Vertebral Body Replacement

Replacement of one or more vertebral bodies is sometimes indicated for wide decompression, visualization of the cord, or excision of an infected, tumorous, or grossly destroyed vertebral body.[146] The construct for the replacement of a vertebral body is shown in Figure 8-38, and it may be used in any region of the spine for vertebral body replacement. With some of the very large vertebrae, it is necessary to use several ribs or two or more pieces of ilium. This construct provides a spacer as a substitute support in the immediate postoperative period. Graft extrusion during extension is resisted by the spikes at either end (Fig. 8-38B, C). Its resistance against vertical compression comes from the strong, horseshoe-shaped configuration of iliac bone, with cortex on three sides (Fig. 8-38B, C, D). The resistance against axial rotation and lateral bending is modest. An excised vertebra can also be advantageously replaced by a keystone construct in this situation. The keystone construct has the advantage of greater stability, but because the dissection violates the intact cortical shell and extends into the central cancellous bone, there is more hemorrhage. Because the graft is seated in cancellous bone in the lower regions of the spine, the vertical load-bearing capacity of the construct may reach its limits. Therefore, we do not recommend its use below the cervical spine unless it is reinforced by some additional support.

This vertebral body resection and reconstruction with anterior bone graft and posterior stabilization when needed is a useful technique for the treatment of benign and malignant tumors of the cervical vertebral body, as reported in the Cervical Spine Research Society study by Fielding and associates.[82]

It is useful to present here the *in vivo* biomechanical studies of dogs by Whitehill and associates.[322] These studies support a fundamental biomechanical surgical principle. When there is instability anteriorly and posteriorly, some tension load-bearing construct is required both posteriorly and anteriorly. This point is depicted in Figure 8-21.

This anteroposterior instability may be an indica-

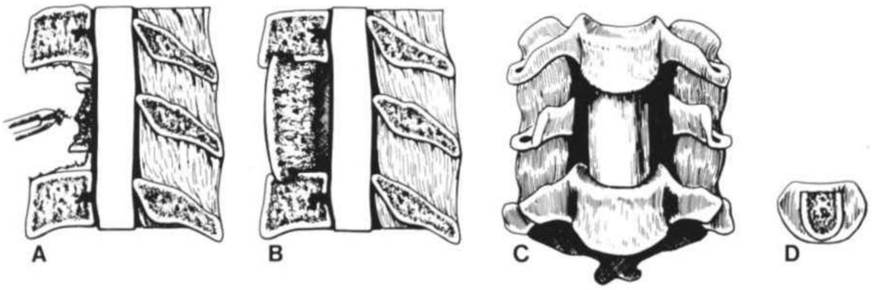


FIGURE 8-38 **(A)** Excision of infected, tumorous, or badly damaged vertebral body and anterior decompression of the spinal cord. **(B)** Sagittal section of the construct, showing the use of iliac bone graft as a spacer. This construct is clinically unstable in the immediate postoperative period. **(C)** An anterior view of the construct, with the bone graft notched in place to prevent extrusion or posterior displacement into the spinal cord. The construct is relatively stable postoperatively during flexion if the posterior elements are intact, but it is clinically unstable in both flexion and extension if they are not intact. **(D)** A cross-section to show how the cortical bone purchase on the intact portion of the end-plate offers resistance to vertical compressive loading.

tion for the use of an anterior plate in addition to a bone graft supplemented by posterior wiring or some other form of posterior fixation.

Posterior Constructs

Posterior Fusion and Wiring

The following technique is recommended for use in wiring around spinous processes. Rather than pass the wire through a hole in each spinous process, we suggest that the wire be passed around the caudal border of the spinous processes, which are generally at an angle that will not allow it to slip off. This is stronger because the wire is exerting forces on an intact cortex. For the cephalad process, we suggest a hole close to the anterior surface at the base of the spinous process. This will be the weaker point, but it provides the maximum margin of safety against pullout in the caudal direction (Fig. 8-39). (See also the section on wires, p. 580 of this chapter. One may select a figure-of-eight wiring technique.)

The surgical constructs for various extents of arthrodesis are demonstrated in Figure 8-39. One FSU can be wired as shown. This provides some restriction of motion. Strips of bone are added to achieve fusion. When it is necessary to fuse more than one FSU, each vertebra can be wired to the adjacent one and the entire group encircled by another wire (Fig. 8-39, middle). The posterior wiring supplies some immediate postoperative stability, provided the posterior elements are structurally intact or only minimally disrupted. If immediate postoperative stability is a major goal, plate and screw fixation or one of the stronger bone graft materials should be used, such as rib, fibula, or tibia, instead of ilium or bone strips. Increasing the number of wirings between the vertebrae in the fusion mass and wiring the graft to the vertebral elements both contribute to the immediate postoperative stability. Immediate stability can be further improved by including more intact FSUs in the fusion mass. This has the liability of decreasing motion and increasing loads upon the FSU above the fusion mass. However, it offers more structure upon which to securely anchor the surgical construct. The greater and more effective the purchase, the stronger is the construct.

Facet Fusion and Wiring

Experiments by Haas on dogs supported the idea that it is desirable to destroy the intervertebral articulations when a posterior spine fusion is performed.[111] The construct described here makes use of these facet articulations in a different manner. Instability in the presence of a unilevel laminectomy may be satisfactorily treated with posterior fusion and wiring. However, when the surgeon is faced with the problem of stabilizing a spine that lacks laminae at two or more levels, we recommend the construct shown in Figure 8-40.[39,242] The facet fusion construct offers considerable immediate post-

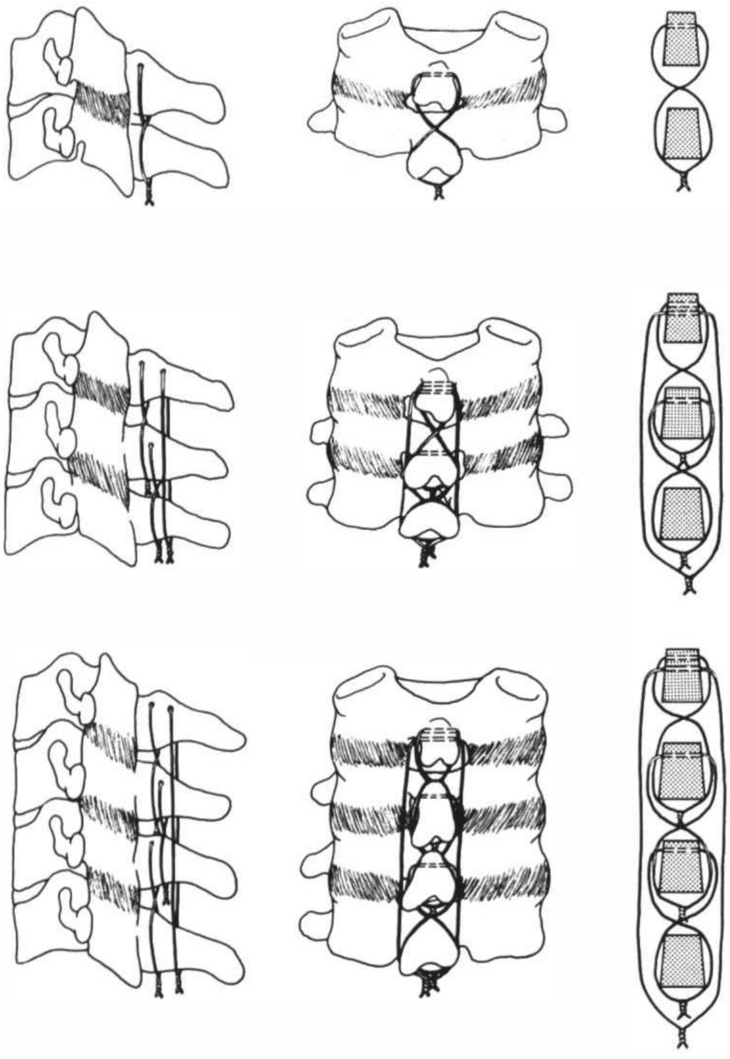


FIGURE 8-39 There are a variety of effective posterior wiring techniques. This figure shows the "figure-of-eight" wiring technique for fixation of 2, 3, and 4 vertebrae respectively. The "figure-of-eight" has been shown in laboratory and clinical reports to be satisfactory. We suggest that the cephalad purchase at the base of the spinous process be tunneled through the bone as shown here. For three- and four-segment wire constructions, three layers of overlapping wires are suggested. There are more complex constructions; however, these are thought to be the most effective, yet simple, ones.

operative stability with variations in effectiveness that are associated with choice of graft material. This construct provides stability against several patterns of motion, particularly flexion/extension and lateral bending. Motion in the sagittal and frontal planes specifically is effectively restrained. Most of the mechanical advantages result from the principle of bridging and the bilateral wiring of the lateral masses to a strong bone graft. We recommend the rib for this procedure. It has several mechanical advantages. Its natural curvature fits the cervical lordosis. It has relatively good strength compared with iliac bone configurations because of its higher moment of inertia, which is due to the closed section created by the tube of cortical bone. It has a flexibility that allows it to bend rather than break. There is also a biologic advantage: the loose cancellous bone in the medullary canal allows it to be relatively more readily absorbed. This procedure has an additional clinical advantage: in the event of a condition requiring reexploration of the spinal cord, the cord is not covered by bone graft.

Pelker and associates studied the multidirectional stability, strength, and failure mechanisms provided by several posterior surgical procedures.[232] Fresh human cadaveric three-vertebrae cervical spine segments were studied when intact, after injury. The injury involved complete ligamentous transection together with removal of an anterior wedge of vertebral body. Following this seven differ-

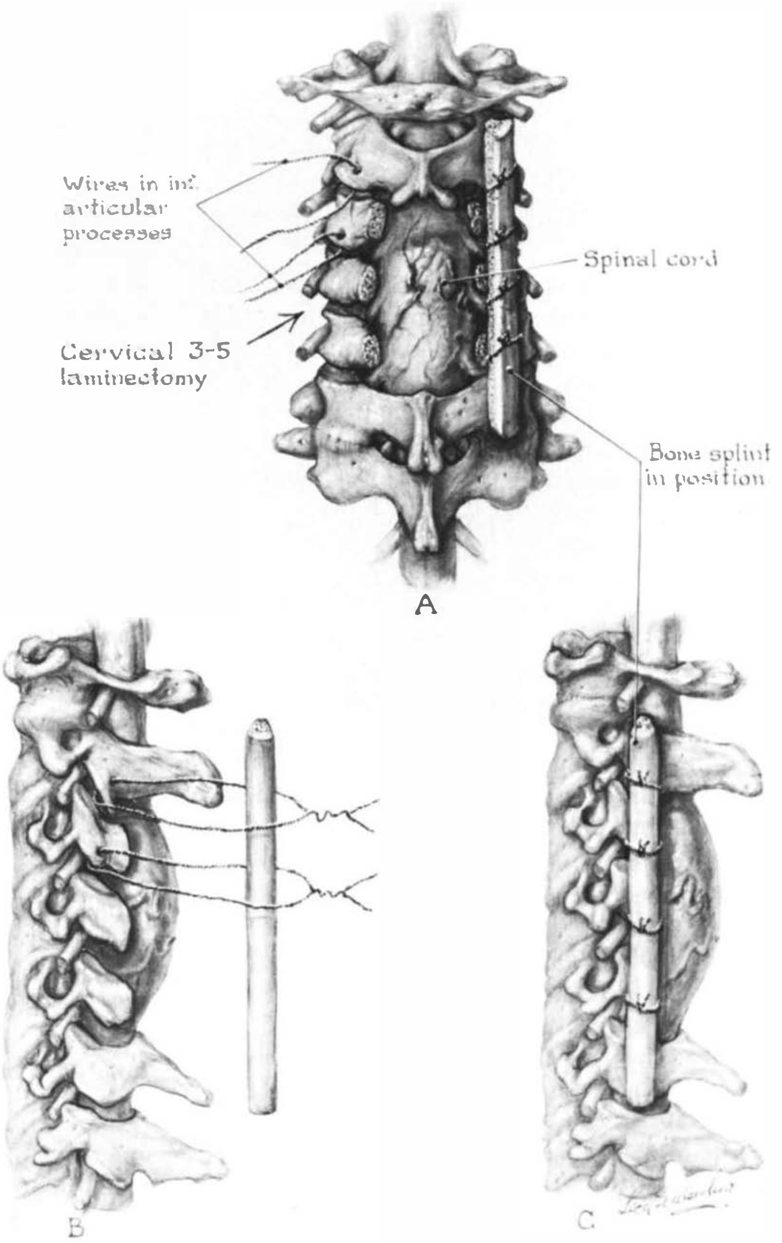


FIGURE 8-40 The facet bone graft wiring procedure. **(A)** The appearance of the exposed cord, with one strut of the tibia graft wired into place. **(B)** The wires are passed through the facet articulations and around the bone graft. **(C)** Here, the second tibial graft is fixed in place. A rib or a selectively removed segment of ilium that has a natural curvature that fits the cervical lordosis and also forms an oval around the exposed portion of spinal cord may be used. We expect that, in the future, surgeons may use with increasing frequency plates affixed to the lateral masses by screws. The facet bone graft wiring procedure, however, is probably safer and is capable of providing satisfactory immediate postoperative stability. *(Robinson, R. A., and Southwick, W. O.: Surgical approaches to the cervical spine. In American Academy of Orthopaedic Surgeons: Instructional Course Lectures. vol. 17. St. Louis, C. V. Mosby, 1960.)*

ent posterior repairs were completed. These repairs used combinations of wire, bone, and polymethylmethacrylate (PMMA).

The first biomechanical tests evaluated multidirectional (flexion, extension, lateral bending, and axial rotation) instabilities of the specimen in the form of ranges of motion. After these nondestructive biomechanical tests, the specimen with one of the seven repairs was loaded in flexion to failure, providing additional information about strength and failure mechanisms. From this comprehensive study, only a few results are presented here. The interested reader may seek the original article for further details.

All repairs provided adequate stability in flexion, while none of the repairs was stable in extension without PMMA supplementation. In lateral bending, all repairs proved stable except for the posterior wiring without PMMA. Axial rotation stability was adequate for all repairs except for the posterior wiring and facet fusion. The supplementation of all repairs with PMMA substantially improved their stability. The findings may be useful in clinical decision making and in the choice of postoperative bracing.

Johnson and co-workers completed biomechanical studies of bilateral facet wiring that did not include bone graft within the wire fixation. This construct provided stability and was the strongest wire construct tested in flexion. The strength was comparable to that of an intact spine.[145] This can be a useful construct for the provision of a high level of immediate postoperative stability. It does, however, carry the liability of placing and leaving wires in the adjacent unfused facet joints. It is not clear what the clinical manifestations of this unphysiologic situation may be. Nevertheless, this particular complication pales in comparison with another one. Whitehill and associates[322] have shown that the placement of *sublaminar* wires in the cervical canal in dogs resulted in a neurologic complication (partial or total quadriplegia) in about 20% (5/24) of the animals. When posterior wiring stability is required and the spinous process wiring technique is not feasible, the facet wiring is much preferred to laminar wiring. The absence of spinous processes and the need for immediate postoperative stability may constitute an indication for the use of the Halifax clamp (see p. 588).

Cervical Spine Fusion—Anterior Versus Posterior

We have generally asserted that if the clinical considerations afford the surgeon a more or less equal choice, posterior fusion is preferable. This has been based largely on the recognition of greater complication risks in anterior as compared with posterior cervical spine surgery. This has been recently confirmed by the work of Capen and associates,[42] who reported similar complication rates but more severe complications with anterior procedures. Tbe dire complications included esophageal problems, graft dislodgement, kyphotic deformity, and degenerative changes above and below the fusions.

EVALUATION OF SURGICAL CONSTRUCTS IN THE THORACIC SPINE

Most of the principles and techniques of surgical constructs in the cervical spine also apply to the thoracic spine. The thoracic spine is discussed to some extent in the treatment of kyphosis (see p. 160). Here, biomechanical considerations that are unique to the thoracic spine are discussed with respect to arthrodesis. It should be emphasized that the loads in this region are much higher than in the cervical spine (about fivefold) and that there is a normal kyphotic angulation. The relative advantages of different locations for placement of the grafts are discussed.

Anterior Constructs

Of the previously described anterior constructs, the trough construct, employing either a modified Bailey-Badgley construct or the keystone principle, is the most useful construct applied to the thoracic spine. Because of the great vessels and the relative ease of exposure, the anterolateral and lateral aspects of the vertebral bodies are more readily accessible than the midline, anterior aspect.

Decompression and Anterior Fusion

In some instances, it is desirable to decompress the thoracic spine anteriorly. When at least the anterior portion of the vertebral body or bodies involved is intact, we suggest the construct shown in Figure 8-41. This is essentially the same procedure described by Fang and associates.[74] It provides adequate decompression and exposure of the anterior portion of the thoracic cord and leaves the supporting structures of the anterior vertebral bodies intact. This may be done laterally and is very useful for removal of a herniated thoracic disc. The two segments of rib are embedded in the cancellous portions of previously intact vertebral bodies above and below. The ribs ultimately provide stability when incorporated. Initially, the postoperative stability depends upon the remaining portion of intact vertebral body anteriorly and the tensile supporting structures posteriorly. If the posterior elements are not intact, we recommend a Milwaukee brace if the surgery is above T6 and a Jewett brace if it is below that level.

The following is a unique construct. In the treat-

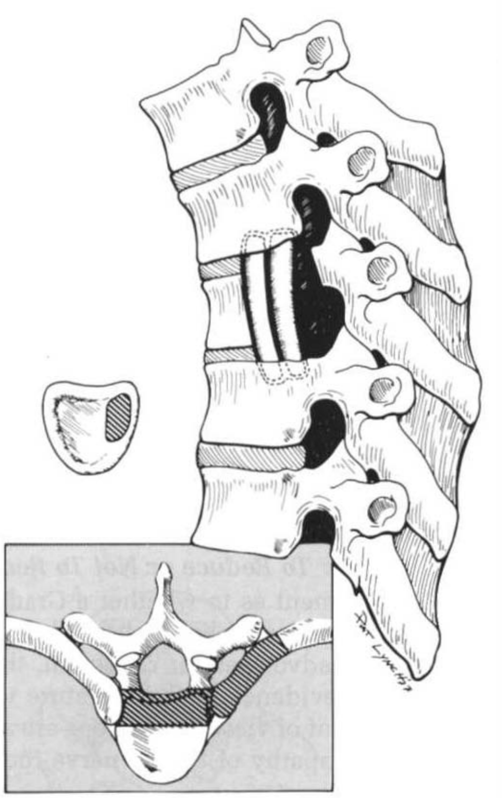


FIGURE 8-41 A construct for decompression and arthrodesis. The decompression is designed to effectively relieve anterior spinal cord pressure and leave some of the vertebra for structural support. A bone graft of fibula or iliac crest is implanted into the vertebral bodies above and below. In situations where the anterior portions of the remaining disc and vertebral body are intact, this construct has a moderate amount of immediate postoperative stability.

ment of kyphotic deformity, a rib with blood supply maintained is used. This functions as a living pedicle graft to the anterior portion of the thoracic spine.[245] Investigators report that although the graft is not effective in the correction of deformity, it does prevent progression. The graft is also reported to hypertrophy with growth.

Posterior Constructs

Location and Extent of Posterior Fusions

A posteriorly placed fusion is perfectly adequate for substitution of disrupted ligamentous structures (Fig. 8-39). More leverage and greater contact area for interface adhesion are provided by including all the posterior elements. The technique and principles described for the cervical spine apply equally well to the thoracic spine. This includes wiring techniques, the selection of graft material, and the proper instruments to enhance immediate postoperative stability. The experience with arthrodesis in the treatment of scoliosis suggests that decortication of the posterior elements and disruption of the facet articulations are essential.

Facet Fusion and Wiring

The principles, indications, and technique of facet fusions in the thoracic spine are the same as those of the cervical spine (see p. 555). The rib grafts may be placed with their convexity posteriorly to fit the normal lordosis of the thoracic spine.

When immediate postoperative stability is not important, simple fusion of the facets and transverse processes alone may be employed.

EVALUATION OF SURGICAL CONSTRUCTS IN THE LUMBAR AND SACRAL SPINE

Anterior Constructs

Interbody Fusions

There are a number of techniques for anterior interbody fusion of the lumbar spine. Most are performed through an anterior approach,[38, 119, 134, 138, 264, 277] but a posterior approach is possible.[326] Interbody fusions are advantageous when all the posterior elements are destroyed, in cases where repeated posterior endeavors have failed, or when the posterior approach is not accessible for clinical reasons. The technique was initially and is currently employed in the treatment of spondylolisthesis. Collapse in this region is not catastrophic, so constructs that have ample quantities of cancellous bone at the graft bed interspace are preferred. The major disadvantages of anterior lumbar approaches and fusions are the associated complications—death, venous thrombosis, retrograde ejaculation, and impotence.[134,264] It has been suggested, based on a survey of surgeons, that the risk of some of the complications has been overrated. The survey of 20 surgeons contributing to 4,500 cases of anterior lumbar spine surgery reported 19 problems of retrograde ejaculation, or 0.42% (one-fourth of these resolved), and 20 cases of impotence, or 0.44%. This was not an ideal experimental design, but it tends to support the assertion that this is not a high-risk complication.[85] In view of the risks involved and the adequacy of posterolateral constructs, we generally prefer the posterolateral fusion technique unless there is some reason to employ the anterior approach. Flynn and Hoque[86] reported a high incidence of nonunion (44%) with anterior lumbar interbody fusions.

Trough Graft Technique

This technique is valuable when the posterior elements are not available for fusion, or when the posterior approach is not possible. Also, it is useful as a spacer when all or part of a vertebral body must be removed. This type of construct offers considerable immediate postoperative stability, especially against sagittal plane translation, which can be the most devastating to the neural elements. A clinical example follows.

Patient I. L. is also discussed on page 357, in relation to clinical instability. Pain, neurologic deficit, and progressive posterior translation of L2 and L3 were present. By employing an anterolateral graft of a large piece of iliac bone wedged into the trough, immediate postoperative stability was attained (Fig. 8-42).

Peg Graft for the L5–S1 Joint

The relative inaccessibility of the anterolateral approach to the lumbosacral joint is dictated by the wings of the ilia and the common iliac veins and their associated branches. Therefore, a direct anterior approach to the L5–S1 joint with the use of a peg graft to gain immediate stability is used.[264] Fibula, rib, or ilium may be employed; the last is generally the most readily accessible. This construct has the added advantage of fusion and some moderate degree of fixation of a progressive or irreducible spondylolisthesis.

Spondylolisthesis: To Reduce or Not To Reduce

There is disagreement as to whether a Grade II or higher spondylolisthesis should be reduced. Although there are advocates for reduction, there is little convincing evidence in the literature to substantiate their point of view. Reductions are associated with radiculopathy of the L5 nerve root.[24, 185]

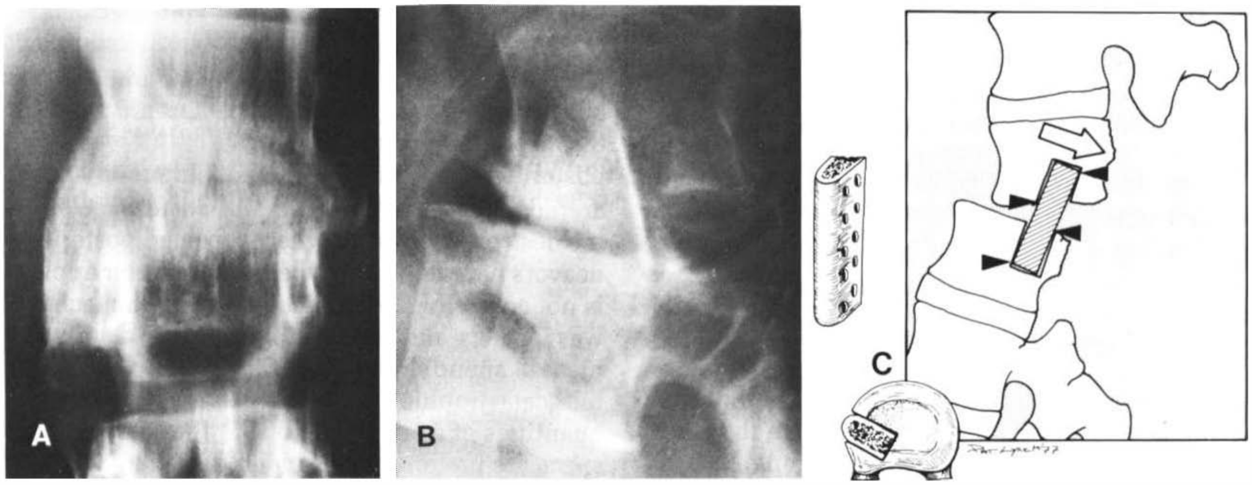


FIGURE 8-42 (**A**) An anteroposterior and (**B**) a lateral radiograph of patient I.L. following an anterolateral trough graft to prevent additional posterior translation of L2 on L3. (**C**) This diagrammatic representation shows the construct more clearly. The bone graft is placed in the region of the remaining overlap of the two vertebral bodies in the sagittal plane. Force arrows indicate the mechanism of locking, which provides immediate postoperative stability against posterior translation. The cortical sides of the graft have several drill holes to facilitate revascularization. The forces associated with the subluxation tend to lock the graft in place. This worked well in the past, and such a construction is completely satisfactory at present. However, for immediate postoperative stability, a Kostuik-Harrington device or a Syracuse I-plate could be considered.

This also fits with the understanding of the pathoanatomic relationships (see Chap. 5). Moreover, long-term follow-up of spondylolisthesis patients treated with fusions without reduction shows satisfactory results.[121,144]

The question of whether or not a posterior fusion alone is enough is open for discussion. Wiltse[326a] has been a proponent of posterior fusion alone. DeWald and colleagues[62] and Frennered and Nachemson[90] have advocated anterior and posterior fusion. This, of course, has the theoretic disadvantage, particularly in a younger person, of creating sequelae of fusion above the arthrodesis (see p. 564). We favor posterior fusion L5–S1 for Stage I slip, L4–S1 for Stage II, and for Stages III and IV, an anterior L5–S1 is added to the posterior fusion, as suggested by Frennered and Nachemson.[90]

Interbody Fusion, Posterior Approach

This technique, described by Wiltberger,[326] is a sound construct and is demonstrated in Figure 8-43. The specifics of the technique are important and of course should be reviewed in detail before using the procedure.

This construct may be useful when the surgeon is limited to a posterior exposure or when the posterior elements are inadequate for fusion. Such a situation might exist after failed attempts at fusion by more conventional procedures, either anteriorly or posteriorly. There may be a need for extensive removal of

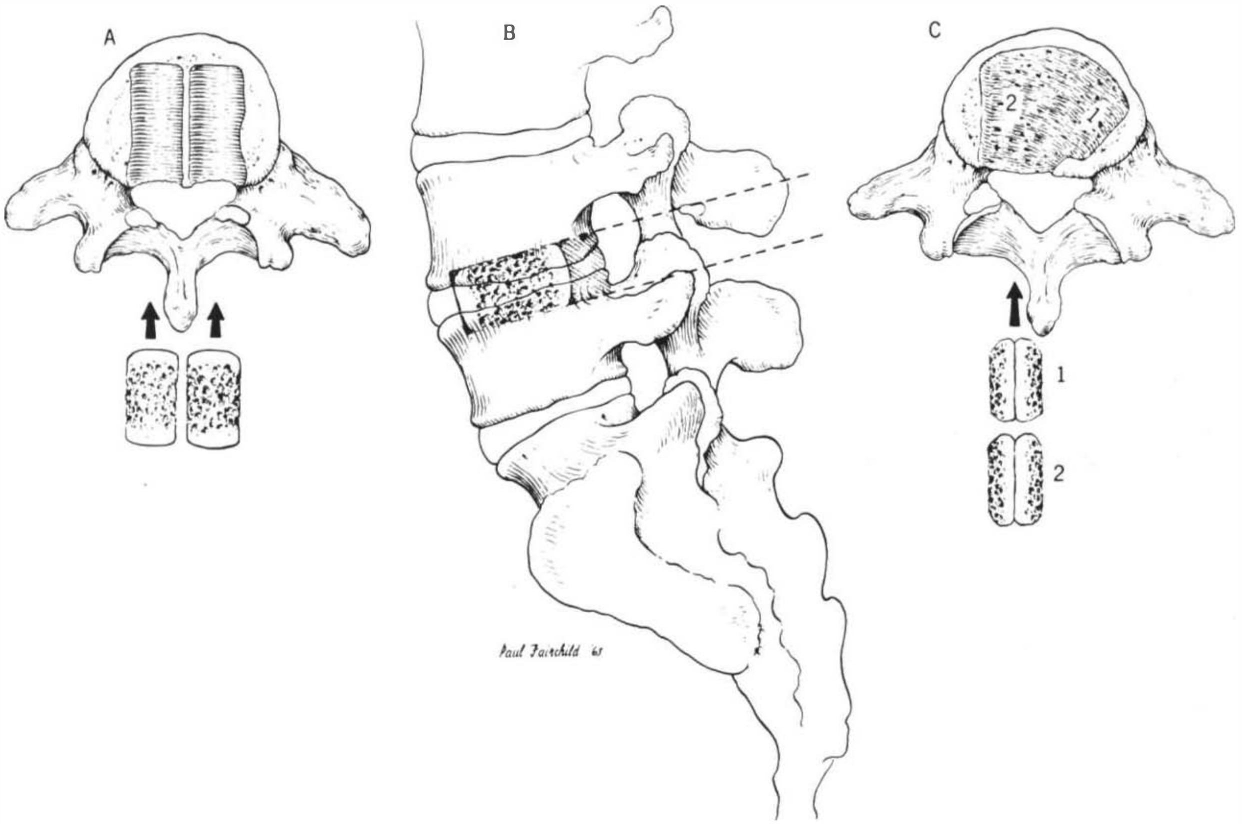


FIGURE 8-43 **(A)** Bilateral insertion of iliac bone dowels. **(B)** A lateral view of inserted dowels. **(C)** The insertion of two cortical bone dowels by a unlateral, partial laminectomy. This construct is useful if the physician wishes to do an interbody fusion at the time of posterior exposure or if an anterior approach is not feasible. The disadvantages are technical difficulties of adequate exposure and possible extrusion of bone graft. A tricortical iliac crest graft is also used with this technique. The tricortical graft is thought to be able to bear higher loads immediately. *(Wiltberger, B. R.: Intervertebral body fusion by the use of posterior bone dowel. Clin. Orthop., 35:69, 1964.)*

the posterior elements or good visualization of the cauda equina. This technique is also advantageous because it does not expose the sacral sympathetic fibers, and therefore there is no risk of impotence in the male. It also avoids the complex plexus of veins, which can sometimes obviate an easy anterior exposure of the lumbosacral joint. Biomechanically, the procedure is sound in that the bone graft is under some variable compressive force; it consists of ample cortical bone (although we do not encourage the use of the tibia) as well as some cancellous bone, and the construct is precisely carpentered for adequate immediate postoperative stability. The graft is in a position to provide both leverage and rigidity.

The major disadvantage of this procedure is the possibility of posterior protrusion of graft material and the necessity for ample nerve root retraction. The probability of posterior protrusion of graft material is reduced by good carpentry and adequate protection in the postoperative period (body jacket including one thigh, worn for 6 weeks); ample nerve root retraction is achieved with careful surgical technique. During surgery, the patient should be positioned with hips and knees flexed. This so-called tuck position has its own liabilities, which should be weighed against the advantages of allowing a more generous retraction of the cauda equina.[4]

Posterior Constructs

The various posterior spine fusions that include the spinous processes and laminae are satisfactory for arthrodesis designed to reestablish clinical stability.[21,278,291,294,309] Wires may be employed to provide some element of immediate postoperative stability and are especially needed when clinical stability has been lost. The basic constructs for posterior element fusions are shown in Figure 8-44. The variations on these basic surgical constructs are numerous.

The "H" Graft

The "H" or clothespin graft of Bosworth has received a good deal of attention.[21] This construct does not appear to have any particularly significant biomechanical advantage over the many other constructs for posterior lumbar spine fusion.

Posterolateral Arthrodesis

There is also the option of performing a posterolateral fusion. This involves the outer portion of the facet joints, the pedicles, the transverse processes,

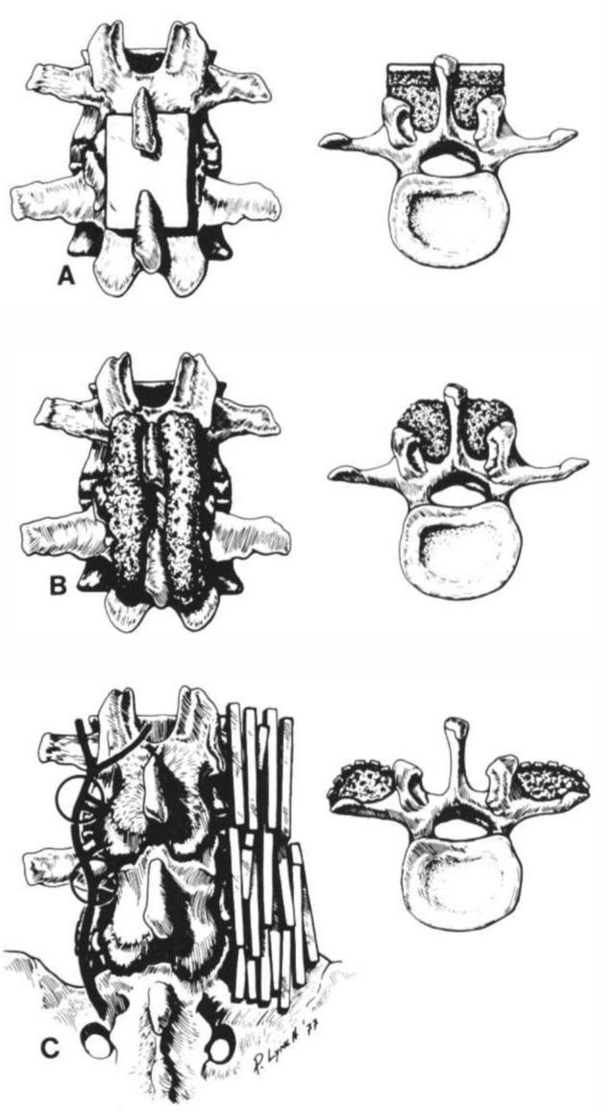


FIGURE 8-44 (A) The basic construct of the "H" graft. The added technique of constructing the "H" graft for distraction of the posterior elements and compression of the graft offers no real biomechanical advantage. (B) The midline (Hibbs) construct is a standard, acceptable one. (C) Fusion of the lateral elements, in our opinion, is the construct of choice, provided the facet joints are also involved in the construct. It has shown the best reported results with clinical experience; the fusion bed is in an area of good blood supply; there is less risk of lamina hypertropy and spinal stenosis; posterior decompression of the cord may be easier during initial surgery or subsequently; and the location of the fusion mass is biomechanically at the best site, without performing an interbody fusion, for maximum immobilization in all parameters. Note the two circles in C. These indicate points recommended by MacNab and Dall[182] for cauterization to control bleeding in this procedure.

and the gutters between them. This type of construct may be used in addition to, or, in the case of spondylolisthesis or multiple laminectomies, instead of, those which involve just the spinous processes and lamina. The bilateral posterolateral fusion technique described by Watkins[309] and modified by Truchly and Thompson[294] has been suggested for use instead of being just an adjunct to fusion involving only the spinous processes and laminae. The modifications suggested by Truchly and Thompson were very successful (92% union). They consist of the following: no screw fixation, use of slivers rather than large blocks of bone graft, two separate posterolateral incisions, and no attempt at facet articulation disruption. Truchly and Thompson report continued success in an additional 125 cases with essentially the same percentage of successful arthrodesis.[295]

Posterolateral fusion has the advantage of not obstructing the dura, which makes the area more readily accessible to subsequent surgery. The dura is protected from migrating bone fragments in the short-term postoperative period. It has also been suggested that the lateral gutters and recesses are more vascular.[309] Biomechanically, this construct (Fig. 8-44C) is superior to the posterior construct (Fig. 8-44B); it increases the area moment of inertia considerably so that greater stability is obtained in axial rotation and lateral bending. We recommend this construct as the procedure of choice for posterior lumbar spine fusion.

Clinical reports in the literature suggest that the inclusion of the posterolateral structures (facets, pedicles, and transverse processes) significantly increases the incidence of solid fusions (see Table 8-3).[278,294,309] Hensinger and colleagues reported 100% fusion using this technique to treat spondylolisthesis in 20 patients with an average age of 14.5 years.[127] These are thought to be more effective constructs than the simple, midline spinous process and lamina fusions and also preferable to the anterior interbody fusions.

MacNab and Dall reviewed a series of comparable reports in an attempt to compare different techniques of lumbar spine fusion. They found that for fusions of L4 to S1, the results were as follows. The pseudarthrosis rate for anterior fusions was 30%; for posterior midline fusion, 17%; and for posterolateral fusion, 7%. The advantages of posterolateral fusions were also pointed out: The graft bed is larger and uninterrupted (i.e., the yellow ligament in midline fusions); the zygapophyseal joints are included in the fusion mass; by including the transverse process and the pars interarticularis of the most cephalad vertebra, that vertebra is more firmly incorporated into the fusion mass; by avoiding decortication of the lamina, the syndrome of spinal stenosis from thickening of the lamina ventrally as well as dorsally is avoided. The work of these investigators includes some illuminating anatomy and describes the blood supply to this region. They suggest that hemorrhage can be greatly reduced by the use of a modified, flexed hip and knee position, relieving abdominal pressure. In addition, there is cauterization of vessels at the base of the transverse process on the caudal side and at the dorsal edge of the superior articular facet.[182]

Grafting for Spondylolysis

In cases of symptomatic spondylolysis and some spondylolisthesis in patients under 30 with minimal displacement, the defect can be grafted and fixed. The technique described by Bradford and Iza[25] employs two separate 18-gauge wire loops around each transverse process and the common spinous process. They report 22 patients with a

TABLE 8–3 Results of Lumbosacral Fusion (L4 to S1)

Author	No. of Operations	Technique	Incidence of Nonunion (%)
Cleveland, Bosworth, and Thompson (1948)	357	"H" graft	17.4
Thompson and Ralston (1949)	169	Hibbs	23.6
	49	Transfacet screws	55.1
Straub (1949)	80	Wilson's plate and cortical graft	14.
McBride and Shorbe (1958)	77	Facet block	36.
Shaw and Taylor (1956)	55	Onlay cortical	36.
Watkins (1953)	10	Posterolateral, block	20.
Truchly and Thompson (1961)	41	Posterolateral, slivers	7.3

(Truchly, G., and Thompson, W. A. L.: Posterior lateral fusion of the lumbosacral spine. J. Bone Joint Surg., 44A:505, 1962.)

90% fusion rate and 80% good or excellent results. Morscher and colleagues[201] described a procedure using bone grafting and a hook–screw combination for the treatment of spondylolysis. The hook is a modification of the Harrington hook and is designed to fit the contour of the inferior or caudad portion of the lumbar lamina. This is used in combination with a cancellous bone screw and two nuts. The investigators report good or excellent results in 10 of 12 patients.

Clinical Biomechanics of Lumbar and Lumbosacral Spine Fusions

Because of the relatively small fusion mass and the presence of the intervertebral disc, it is reasonable to assume that the posterior and posterolateral constructs are relatively less stiff than the interbody fusion constructs. They are certainly less stiff than the so-called 360° fusion construct, which includes an interbody fusion as well as a posterior and posterolateral arthrodesis. This "360° construct" and, to some extent, the anterior construct are relatively stiffer. Therefore, there is relatively more stress concentration than would exist with an isolated posterior construction.

Mechanical, Structural, and Clinical Anomalies Adjacent to Cervical or Lumbar Spine Fusions

Increased motion
Decreased motion
Paradoxical motion
Degenerative changes, discs, and facets
Spondylolysis aquisita
Spinal stenosis
Degenerative spondylolisthesis
Fracture dislocations

Theoretically then, one would expect above and/or below the high-stiffness fusion mass relatively more hypermobility, greater loads, more degenerative changes, more instability, and a higher incidence of spinal stenosis. There is some evidence to support these assumptions. Harabayashi and colleagues reported lumbar stenosis at the level above an anterior spinal fusion in three patients. This has also been reported in patients having posterior[118] spine fusions. Frymoyer and associates[94] noted traction spurs, disc narrowing, and hypermobility above the fusion mass. Although these were common findings, they were not thought by the authors to be correlated with any clinical problems. Failure of the spine (pars interarticularis defect) has also been reported above fusion masses. Brunet and Wiley[32] added 14 patients with this problem to the 23 previously reported in the literature. The problem presented within 5 years of an interlamina fusion. Lee[171] followed 18 patients an average of 8.5 years after lumbar spine fusion. The most common finding was hypertrophic degeneration arthritis of the facets. He also found spinal stenosis, severe disc degeneration, degenerative spondylolisthesis, and spondylolysis aquisita. These studies document the problem. The changes associated with lumbar fusion are listed above. Unfortunately, there is not enough information at this time to allow one to select a fusion construct that will be clinically satisfactory and yet prevent stress concentration of a magnitude that may expose the patient to the following problems. In our view, the current, albeit inconclusive, state of knowledge suggests that other things being equal, bilateral posterolateral fusions should be used in the lumbar spine.

Fusion should not be performed only to relieve pain from intervertebral disc pathology, because it is not reasonable to anticipate success. It is known from the biomechanical studies of Rolander that even with all the posterior elements fused, there may still be motion between vertebral bodies. This is due to the normal elasticity of the bone that comprises the pedicles (see Fig. 8-16). No particular technique of posterior lumbar fusion has been shown to be clinically superior in eliminating pain. The crucial consideration is to understand the reason for arthrodesis and then to design the construct so that the appropriate posterior elements of the vertebrae in question are incorporated into a fusion mass.

Lumbar spine fusions for low back pain in the absence of clinical instability or spondylolisthesis are not well justified by clinical experience. In June 1974, the International Society for the Study of the Lumbar Spine held its inaugural meeting in Montreal, Canada. The results and the techniques of several methods of spine fusion were discussed. We would like to share with the reader the comments of two distinguished surgeons who helped to place these numerous techniques in some perspective.

> We are all probably aware of the rather poor results that have recently been reported for these patients

regardless of whether the fusion was performed from the front, the back, or laterally. . . . In my own mind and also in the minds of many colleagues there is no doubt that for the majority of our patients suffering from low back pain, the treatment is not fusion, no matter what type of approach [construct] is used.[205]

—ALF NACHEMSON, M.D.

. . . Clearly the essential issue is not developing a better technique [construct] for obtaining spinal fusion but rather more clearly defining those instances in which spinal fusion is truly necessary and will yield a high degree of relief of symptoms. . . . For these reasons I feel a more restrictive role is indicated for the operation of spinal fusion in light of our present knowledge.[249]

—RICHARD ROTHMAN, M.D., PH.D.

SPINAL FUSIONS IN THE MANAGEMENT OF PRIMARY AND METASTATIC TUMORS

Basic Concepts and Principles

Because this is a rather complex problem, we shall discuss first some basic concepts and principles, which will be followed by some specific documentation from the literature.

The first decision is to determine whether the goal will be en bloc removal of the tumor with a margin of normal tissue followed by appropriate reconstruction, or removal of the bulk of the tumor to protect neural elements and stabilize the spine to keep the patient ambulatory and as comfortable as possible. The latter is the reasonable goal in most instances because most spine malignancies are metastatic and the patient has a limited life expectancy. Most primary malignant tumors of the spine are not resectable en bloc with a margin, usually because of two reasons. The first has to do with the fact that this is often not possible without including neural structures in the resection. There is some disagreement as to when the en bloc resection of a spinal column tumor with a margin of tumor-free tissue can be maintained while leaving the spinal cord intact. The other reason that this is sometimes not feasible is the limited training and experience of the available surgeons. Stener[262] has documented successful treatment with removal of entire vertebral segments with preservation of spinal cord function, reconstruction of the spinal column, and long-term follow-up without local or metastatic recurrence of the malignant tumor.

The interbody fusion has the biomechanical characteristics described on page 535. However, we do not advise that one extrapolate from those principles the decision that the posterior lumbar interbody fusion (PLIF) is the fusion technique of choice. There are a number of other important clinical considerations that affect the choice of a surgical procedure. We believe that, considering all the clinical and biomechanical risks and benefits, the first choice for lumbar spine fusion is the bilateral posterolateral construct depicted in Figure 8-44C.

We shall focus now on some of the fundamental considerations that relate to palliative spinal surgery for malignant disease. Then we will return to the definitive management of the primary malignancy of the spinal column.

We think the points suggested below are justified given currently available knowledge. The subsequent discussion attempts to explain, substantiate, and document these assertions.

1. Overall patient care goal: protect neural elements; maintain or attain pain-free ambulatory status.
2. Clinical biomechanical goal: decompress the neural elements; reconstruct the spinal column for short- and/or long-term stability. Obtain maximum immediate postoperative stability, possibly with internal fixation.
3. Surgical operative goals: attain the preceding goals with the safest, least extensive, lowest-risk surgical procedure(s).
4. The surgical operative goals can best be achieved by the following:
    A. Obtain full imaging evaluation of the tumor through some appropriate combination of plain films, CT scans (with various reconstitutions), MRI, myelogram, and selective arteriography (with possible embolization of the tumor).
    B. Organize team for preoperative planning session. Ideally, include all surgeons to be involved, neuroradiologist, oncologist, anesthesiologist, and operating room nurse.
        (1) Develop as clearly as possible a full three-dimensional concept of the relations of the spinal cord, the tumor, the spinal column, and the vital regional structures.

(2) Work first on a plan that will allow the resection and reconstruction to occur entirely through either an anterior or a posterior approach.

If this is not feasible, then a posterior approach with or without decompression but with extensive stabilization, followed by an anterior approach, is generally more likely to be the desirable strategy. (See discussion of anterior vs. posterior approaches, p. 540.)

(3) Plan surgical exposure and resection of the tumor from the region of the spinal cord. Note also what spinal column support structures have been or will be lost by the exposure and decompression. Diagram this if possible.

(4) Plan the surgical reconstruction. Is the goal long-term, intermediate, or short-term clinical stability? Is it some combination of the three?

Resources to achieve the goal(s) include bone graft, bone cement, various implants (plates, screws, wires, ceramics), and biodegradable bone composites.

The various implants can provide short-term and immediate postoperative stability. Bone graft is required for long-term stability.

Consider the patient's needs and prognosis and select among the options. Develop a specific plan for the surgical reconstruction. Draw diagrams of it if feasible. Also, discuss one or more alternative plans.

(5) Plan sequence in which the various steps of the procedure will be completed and be certain that all necessary implants and equipment will be available.

(6) The sequence planning includes decisions as to *which* surgeon will do *what*, and *when*.

(7) Discuss also immediate postoperative care for primary and alternative plans.

In the following display, we have offered some general guidelines that may help in decision making.

SPINE TUMOR SURGERY: SOME FACTORS TO CONSIDER

I. Define biomechanical goals for individual patient
  Immediate postoperative stability
  Short-term stability (<6 months)
  Long-term stability (>1 year)
II. Immediate postoperative stability
  Segmental fixation—posteriorly
    Pedicle fixation
    Facet wires to rods
    Lamina wires to rods
    Spinous process fixation to rods
    Wires to posterior elements
    Wires to posterior elements plus cement
  Anterior segmental fixation
    Plates and screws
    Rods and screws
    Screws and cables
  External fixation
    Screws in pedicles with external device
    Halo, jacket, cast
    Halo, pelvic fixation
III. Determine anterior versus posterior or combined approach
  For purposes of decompression, tumor removal, or resection
  For purposes of reconstruction and stabilization
IV. During reconstruction planning, take into consideration anticipated response of tumor (i.e., local recurrence, thus more extensive bone grafting).
V. Consider also implications of any adjunctive radiation and/or chemotherapy treatment.
VI. Open biopsy may be required for definitive diagnosis.

Clinical Studies of Primary and Secondary Malignancies

Prophylactic surgical stabilization should be considered for all patients with spinal metastases before deformity and paralysis occur. This rather forward-moving and active approach has been espoused by DeWald and colleagues.[61] This is a very reasonable proposition. Several key questions to be studied and answered are as follows. What is the real (measurable) benefit in the quality of life for the individual patient in regard to pain, ambulation, bladder and bowel control, and time at home versus time in the hospital? What is the relative cost to patient, family, and society of prophylactic treatment versus neurologic deficit followed by treatment? How can we accurately predict which metastases (tumor type, location in spine, extent of bone destruction) are associated with impending neurologic doom? The last is a solvable biomechanical problem, the answer to which we can reasonably anticipate in the next edition of this work.

Bohlman and associates[18] reported their experience with 23 primary neoplasms of the cervical spine in which they used many of the preceding principles, such as posterior stabilization prior to extensive anterior resection. Conversely, they noted that extensive posterior decompression may well be preceded by anterior fusion and stabilization. They did not attempt extensive total en bloc resection of these tumors. Fidler[76] repeated his experience with pathologic fractures of the cervical spine. This work recommended posterior instrumentation two levels above and below the tumor site using cement in one side and a bone graft in the other if life expectancy is greater than 6 months. For metastatic spinal fractures below T2, however, Fidler's basic construct is anterior fixation with distraction and screw fixation.[77]

Flatley and co-workers reported their experience with seven patients treated for spinal instability secondary to disc disease.[83] Their basic treatment was posterior wiring (16-gauge wire) to a rod 4.8 mm in diameter three levels above and three levels below the instability. They espoused the principle of including a fusion in the construct if life expectancy is greater than 12 months.

Siegal reported 47 patients treated with vertebral body resection for epidural compression by malignant tumors. He emphasized the importance of anterior decompression and observed that the results with laminectomy were less satisfactory.[263]

For descriptions and examples of extensive resection of entire vertebral bodies with *en bloc* resection of tumors, the reader is referred to the work of Crawford and to that of Stener and Gunterberg.[55, 262, 284] An example of such a construct is presented in the next section.

Weinstein and McLain have published the University of Iowa experience with 82 patients with primary neoplasms of the spine.[310] This large series provided some useful information. Plain x-rays showed 99% of the tumors. The malignant tumors occurred more frequently in the vertebral bodies than in the posterior elements. The ratio was 2 : 1. They were more often associated with neurologic deficit and, as may have been expected, were more often seen in the older patients. Five-year survival in patients with malignant tumors was associated with the type of surgery performed. The survival rates were as follows: those with curettage, 0%; those with incomplete resection, 18.7%; and those with complete excision, 75%.

Reconstruction Following Total Spondylectomy

Complete vertebral body removal and subsequent reconstruction and fusion in the patient discussed below exemplify sound biomechanical principles.

The patient, a 49-year-old farmer, was afflicted with a chondrosarcoma arising from the body of the seventh thoracic vertebra.[280,281,283] The tumor extended into the mediastinum and the spinal canal, where it displaced the spinal cord. All of the seventh thoracic vertebra and parts of the sixth and eighth were removed along with the tumor. The thoracic spine was reconstructed as shown in Figure 8-45. Fifteen months after surgery, the patient was well and walking. There were no signs of metastasis. Radiographs showed that the bone graft was completely incorporated and fused to the partially resected vertebrae above and below.

This construct uses two portions of iliac bone, with their cortices intact on three sides. This provides good resistance to vertical compressive loading, an excellent spacer, and good cancellous to cancellous bone contact at the graft bed interfaces. In order to resect the tumor with adequate margin, a modification of the keystone principle is used, and the graft is wedge-shaped, with the apex of the wedge pointing posteriorly instead of anteriorly, as in the keystone graft described by Simmons and Bhalla. The keystone is reversed here to prevent migration of graft material back into the spinal cord. The construct effectively compensates for the tendency for anterior displacement, caused by the passage of two cerclage wires in the horizontal plane around the grafts and through the two plates posteriorly. In addition, there are two silk threads in the frontal plane, at the caudal and cephalad ends of the graft, that go through the respective adjacent intact vertebral bodies, T5 and T9. These fixations also offer some resistance against anterior protrusion of the grafts. The patient in this case was kept in bed in a plaster shell for 3.5 months.

The primary, immediate postoperative stability

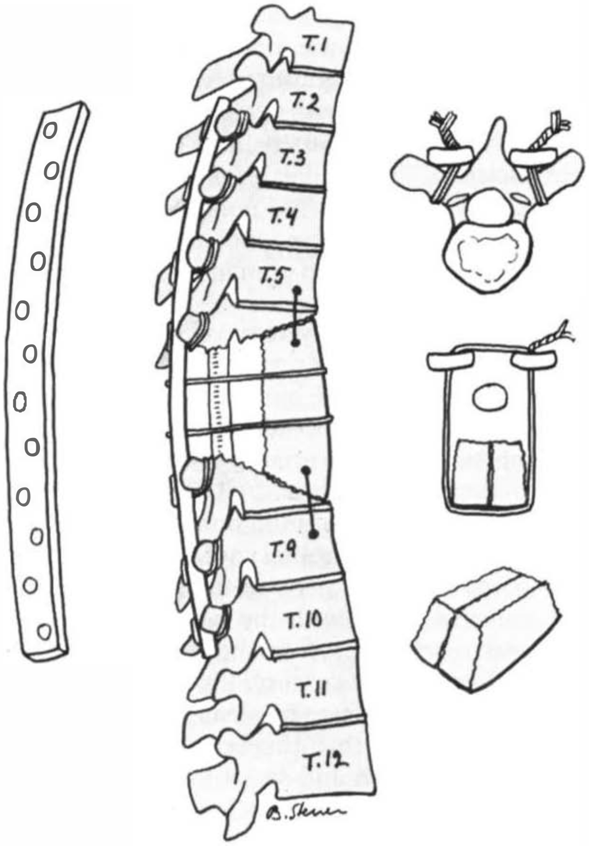


FIGURE 8-45 Reconstruction of the spine. The anteroinferior half of the body, the inferior articular processes, the spinous process of the sixth thoracic vertebra, all of the seventh, the anterosuperior half of the body, and the superior articular processes of the eighth have been removed along with the tumor. Two plates have been fastened with double steel wires to the transverse processes of the third to the sixth and the eighth to the tenth vertebrae (*middle and top right*). Two iliac bone-blocks with obliquely cut ends have been put together (*bottom right*) and inserted between the obliquely cut bodies of the sixth and eighth vertebrae (*middle*). The blocks have been fastened to the spine with silk threads passed through holes in the blocks and the vertebrae. Further fixation has been provided by two steel wires fastened to the plates and gripping the bone-blocks (*middle and middle right*). (From Stener, B.: *Total spondylectomy in chondrosarcoma arising from the seventh thoracic vertebra. J. Bone Joint Surg., 53B:288, 1971.)*

of the construct is provided by two posterior stainless steel plates that are bent to conform to the normal thoracic kyphosis and are wired to the transverse processes of the two partially resected vertebrae. They are also wired to *three* intact vertebrae above and *two* intact vertebrae below. This fixation and purchase on several normal vertebrae provides stability against both tensile and compressive loads. Thus, there is good stability against flexion and extension, and with the laterally placed attachments to the transverse processes, there is stability against lateral bending. Axial rotation may be resisted reasonably well through the fixation of the transverse process to the plate, which impinges against the lamina and the base of the spinous processes when axial rotation is attempted. Some degree of additional stability can be expected from the relative intrinsic stiffness and modest motion in the thoracic spine, as well as the stiffness and support supplied by the rib cage. Figure 8-46 shows a radiograph of the construct almost 5 years after surgery. The patient is free from evidence of disease after 19 years. He has a slight spastic paraparesis but is able to walk with a cane in each hand.

This construct demonstrates the sound judgment and ingenuity of the surgeon (Dr. Sterer) as well as good clinical biomechanics.

In principle, with difficult reconstruction problems of the spine following tumor resection, the surgeon is well advised to consider some of the implants that provide immediate postoperative rigidity. As described in the latter part of this chapter, the Cotrel–Dubousset system, the Steffee system, and others can impart high rigidity to the postoperative construction.

SOME GUIDELINES ON THE BIOMECHANICS OF POSTOPERATIVE MANAGEMENT OF PATIENTS UNDERGOING SPINAL FUSIONS

How long should the patient be kept in bed after undergoing a spinal fusion? Should some type of orthosis be used afterward, and if so, for how long? The literature recommends a broad variety of management programs. Of course, each patient must be evaluated individually, and the management program should be designed and adjusted to best serve the specific patient's needs and requirements. Some general basic guidelines follow.

The management of these patients is basically contingent upon the presence or absence of clinical stability. Management of clinically stable patients is simpler. There are four stages of maturation of the arthrodesis: I, fibrous healing; II, mixed fibrous and

FIGURE 8-46 Radiograph taken almost 5 years after surgery for chondrosarcoma in a 49-year-old man (see Fig. 8-45). The iliac bone grafts now form a block-vertebra in conjunction with the partially resected adjacent vertebral bodies of T6 and T8. The surgical construct is stable and completely successful from a biomechanics standpoint. Note the osteophyte between T9 and the bone grafts. There are no signs of recurrence or metastases. The patient is still well after 19 years.[288] He has a slight spastic paraparesis but is able to walk with a cane in each hand. *(From Stener, B.: Resección de columna en el tratamiento de los tumores vertebrales. Acta Orthop. Latinoam., 1:189, 1974.)*

osseous healing; III immature osseous healing; and IV, mature osseous healing. During Stage I, maximum protection should be provided. In addition to reduced activity, a protective orthosis may be worn. In Stage II, there is a need for relative protection either by some type of orthosis or through restrictions in activity. In Stage III, the patient is allowed normal, nonvigorous activity with progression of rehabilitative exercises. In Stage IV, the patient has reached maximum healing and may undergo a program of vigorous rehabilitative activities and exercises, within reasonable limits that are determined for each patient.

The first three stages involve roughly 6 weeks each, and Stage IV may be prolonged. The total time allowed for arthrodesis maturation may be increased or decreased for any given patient, depending on a variety of considerations. The cervical, thoracic, and lumbar spines, in that order, tend to require progressively more time for healing. The greater the extent of fusion (e.g., the number of FSUs), the longer it takes for healing. As a corollary to this, the larger the size of the bone grafts, the longer it takes for complete fusion. We must assume that wound infections, loosening of hardware, or one or more previous operations tend to slow the rate of arthrodesis. Patients in poor health also require relatively more time for completion of arthrodesis.

Clinically *unstable* patients require more time, especially in Stages I and II. The above considerations also apply to these patients. Precise schedules are difficult to generate because the number of variables and problems of evaluation of fusions in mechanical situations make it virtually impossible for an investigation to generate the necessary data. We occasionally offer specific schedules for the postoperative management of various conditions. However, our recommendations for basic general guidelines for postoperative management of patients with

FACTORS THAT TEND TO PROLONG HEALING TIME IN SPINAL ARTHRODESIS

- Extensive fusion
- Large bone grafts
- Wound infections
- Loosening hardware
- Previous surgery in the same area
- Aged patients
- Debilitated patients
- Clinically unstable spine

TABLE 8–4 Clinical Biomechanical Stages of Spine Arthrodesis and Their Management

Stage	Approximate Time Required (weeks)	Goal	Management
I Fibrous healing	6	Maximum protection	Restricted activity Possibly an orthosis
II Mixed fibrous and osseous healing	6	Relative protection	Less restricted activity No protection or protective orthosis with less control than above
III Immature osseous healing	6	Moderate but non-vigorous activity (swimming, biking, walking)	Minimal or no orthosis All but vigorous activity Regular or light duty if job permits
IV Mature osseous healing	6 or more	Maximum convalescence and activity	Maximum allowable activity

spine arthrodesis are summarized and presented in Table 8-4 and the following list. If factors in this list strongly favor the patient, or if the opposite is true, the time intervals for the different stages are adjusted accordingly.

With the advent of several relatively rigid surgical implants of the spine, much of the biology of the arthrodesis process and schedule may be altered. The critical factor may be the relationship between the progression of the stiffness of the arthrodesis to take over the loads and the fatigue life of the implant. This may be the major factor in healing, and issues such as bracing activity of the patient, orthotics, and so forth, may play a relatively insignificant role, depending upon the mechanical properties of the spine implant construct.

PART 3: SURGICAL CONSTRUCTS EMPLOYING METHYLMETHACRYLATE

The use of polymethylmethacrylate (PMMA) as an adjunct to or instead of spine arthrodesis is becoming more widespread.*[67,96,155,156,223,259,271] There does not appear at this time to be any definitive data that demonstrate an increased incidence of infection over comparable surgery without methylmethacrylate. Studies indicate that there is probably no direct toxic effect, no detrimental exothermic reaction, and no chemical reaction at nearby bone.[178,240] On the other hand, some more recent *in vivo* animal studies of surgical constructs have demonstrated some limitations of the cement and a superiority of bone graft[306,308,321] and fibrous tissue ingrowth between the cement and the bone.[320] This section presents the available biomechanical information that will be helpful to the surgeon in decisions about the use of methylmethacrylate for spine fixations.

BIOMECHANICAL FACTORS

Cement–Bone Interface

The material is a cement and not a glue. The bonding between methylmethacrylate and bone is not one of adhesion but is based on the interdigitation of cement particles and bone trabeculae, with the two being separated by a thin layer of fibrous tissue.[44,105] This fibrous layer has been examined closely following the application of PMMA to the spine of dogs. Within a month, a distinct fibrous band was present and was associated with loosening of the construct[320] (Fig. 8-47).

PMMA is likely to be most effective in resisting compressive loads. We should remember, however, that for most compressive loading situations, human bone itself can be expected to be strong enough for

* Personal communication, K. Ono, M.D., December 1984.

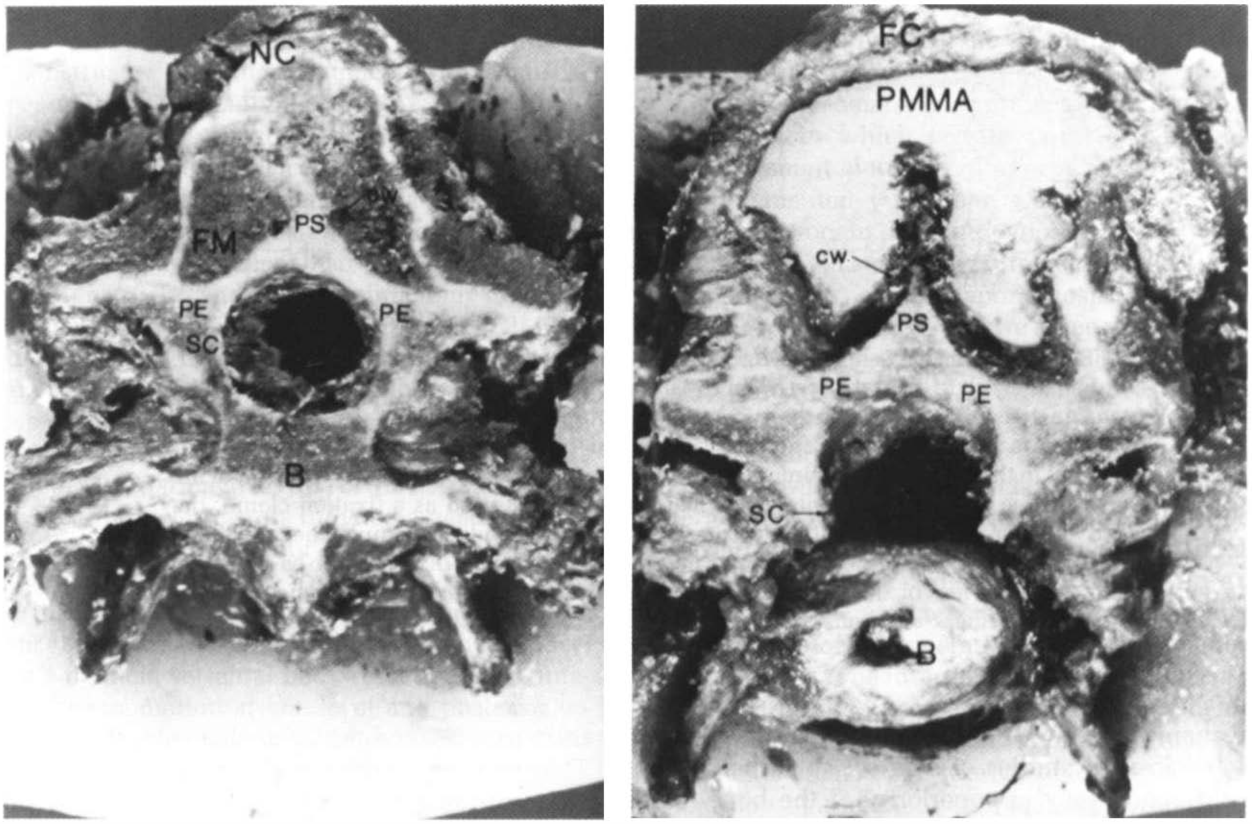


FIGURE 8-47 Photograph of horizontal section of the cervical spine through the center of a typical PMMA fusion. A space is noted between the methacrylate "fusion" mass and the underlying bone of the posterior elements. Although this section was obtained following mechanical failure, the tissue is similar in appearance to the overlying capsule that partially fills the interface between the total joint prosthesis and the host tissues. *FC* = fibrous capsule; *PMMA* = methylmethacrylate "fusion" mass; *CW* = cross-section of cerclage wire; *PS* = original posterior spinous process; *PE* = original posterior element bone; *SC* = spine canal; *V* = vertebral body. *(From Whitehill, R., Reger, S., Fox, E. A., Payne, R., Barry, J., Cole, C., Richman, J., and Bruce, J.: The use of methylmethacrylate cement as an instantaneous fusion mass in posterior cervical fusions: A canine in vivo experimental model. Spine, 9:246, 1984.)*

adequate support.[314] Methylmethacrylate is less effective in resisting tensile loading than it is in withstanding compression. Therefore, it will also be relatively weaker in bending. It has been shown, however, that, as is the case with concrete, its tensile load–bearing capabilities may be significantly enhanced through the incorporation of wire or wire mesh that has a high tensile resistance.[253] The additional advantage provided by the mesh is the great opportunity for interdigitation interface bonding between the host bone, surgical wire, mesh, and methylmethacrylate.

Experimental Studies

In vivo studies of PMMA surgical constructs by Wang and associates have provided evidence for several clinical biomechanical comments.[306,308] The strength of the constructs with cement decreased with time, while the strength of those involving bone grafts increased. Those constructs in which the cement was used as an anterior spacer (Fig. 8-48A) were the weakest constructs when tested in extension (see also Fig. 8-21). Constructs involving cement reinforced with wire or chain were found to be

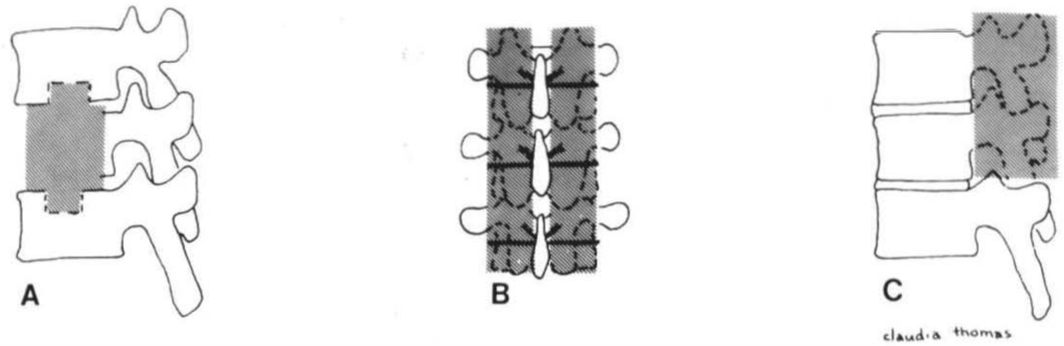


FIGURE 8-48 The uses of methylmethacrylate in surgical constructs. **(A)** As a *spacer*, methylmethacrylate is most effective in resisting compressive forces during flexion. During extension, tension is applied to the cement–bone interface and may cause failure. **(B)** When the cement is used as a *fixation splint*, it is subjected to bending loads. The key biomechanical factors in this situation are reinforcement of the cement and an effective attachment of the cement to the posterior elements. **(C)** When the cement is used as a *fixation clamp*, the effectiveness of the interface with the posterior elements depends solely upon their anatomic configuration and the placement of the methylmethacrylate. This may not be effective, particularly in view of the evidence that shows that a large fibrous membrane develops between the cement and bone.

mechanically better. Whitehill and colleagues completed *in vivo* studies of surgical constructs with PMMA and noted a superiority of the bone graft constructs over those involving the cement.[321]

In an *in vitro* study of spine models testing 14 types of instrumentation, Fidler[78] noted that PMMA placed on one side significantly increased rigidity and thus stability. The clinical implication is that a construct involving fixation that includes PMMA on one side and bone graft on the other may constitute an optimized trade-off, gaining immediate as well as long-term stability.

Postmortem Mechanical Tests of Surgical Constructs Involving Methylmethacrylate

M. A. is a 53-year-old male who was hit by a truck and sustained a fracture dislocation of C6 on C7 associated with paraplegia. Total laminectomies and foraminotomies were carried out at C6 and C7. The following surgical construct was established.

Wires were passed through the spinous processes of C4 and T2, and C5 and T1. These wires formed ellipses around the C6–C7 laminectomy. Stainless steel mesh was cut to size and placed over the laminectomy site.[C] Methylmethacrylate was then mixed and applied in one piece as an ellipse, the center of which was the laminectomy, and pressed down over the wires, C7, and posterior elements of T1 and T2. The central aperture was designed to allow flow of blood out of the spinal canal, in case of epidural bleeding and subsequent hematoma. Unfortunately, the patient died on the 18th postoperative day.

The surgical construct was set up in an Instron testing machine to be tested to failure in a manner that simulates flexion.[228] The first audible crack occurred when the upper wire pulled out of the spinous process of C4. The specimen reached its maximum energy absorption capacity (strength) when the wire pulled out of the spinous process of T2.

Although no firm conclusions can be drawn from the study of one specimen, there are some points that merit discussion. Methylmethacrylate significantly increased the stiffness of the spine between C4 and T2. Consequently, there is considerable stress concentration at C3–C4 and T2–T3. These are the obvious points at which failure is likely to occur and, in fact, did occur. The failure of the constructs due to the wires pulling out shows that, at least for this particular construct, the point of attachment of the wires was the weakest link in the chain. This is important in regard to the value of the construct in contributing to postoperative stability. The first crack (pullout of the wire from the spinous process of C4) occurred at 70 N (16 lbf). This is in the range of 10% of body weight, which is about the weight of the head. In other words, the weak link of the construct without the assistance of active muscle forces or braces is alarmingly close to expected physiologic loads. This constitutes a tolerance limit or margin of safety that is not particularly generous.

Panjabi and associates had an opportunity to test another authentic *clinical* surgical construct involving PMMA.[227] The patient had a posterior wiring and fusion C5–T2, and a C7 vertebrectomy followed by an anterior stabilization C6–T1 using PMMA and wire mesh. The specimen was tested after 7 years' maturation in the live patient. It was studied three-dimensionally and compared with a normal cervical spine. The finding was that this flexion was about the same strength as that of the normal control. However, the investigators indicted a need for bone graft in addition to cement for long-term stability and strength.

Fidler[76] studied postmortem specimens of two clinically operated human spines. One had undergone a thoracic vertebral body resection with anterior instrumentation at one level and posterior instrumentation at another level. Tests of this specimen showed that the posterior instrumentation was more effective in resisting tension. The second specimen had anterior fixation that had been strengthened by the addition of cement studs in the vertebral end-plate and a modified paravertebral Zielke screw/rod system. This second anterior system was an improvement over the previous case in regard to resisting axial torsion (y-axis rotation). The anterior interbody fixation was better than the posterior for resisting flexion (compression) forces. The investigation also observed a smooth fibrous capsule enveloping the anterior interbody device, which was comprised of metal and cement. This suggests that humans, like Whitehill's canine, probably form encapsulating fibrous tissue around cement spine implants.[321]

Clinical Studies

Bryan and colleagues[33] used PMMA with stainless steel wire and titanium mesh for posterior cervical arthrodesis in rheumatoid patients. The authors used one of two constructs in 11 patients. The two constructs involved long cervical fusions with and without inclusion of the occiput. The complications were: subluxation of an FSU at the end of a long fusion area and two wound dehiscences. This latter complication was due to an overabundance of cement material being put in the wound, a point worth remembering.

Clark and associates have prepared a comprehensive review in which certain points are emphasized.[46] Cement alone is not appropriate; it should be supplemented with bone graft, wire, or some instrumentation. In their indications for PMMA, the authors included stabilizing tumors, rheumatoid arthritis, fractures, cervical spondylosis, and others with osteopenic bone. While we agree with the first two indications based on the in vivo experimental studies and our analysis, we take a significantly more conservative position on the other indications.

Harrington[120] has reported his experience with the use of PMMA for vertebral body replacement in anterior stabilization of the spine in the treatment of metastatic disease. This report included 14 patients followed over 4.5 years. There was only one failure of fixation. He preferred cement to bone graft because of the option for postoperative irradiation of more than 1500 rads. The importance of good purchase of the cement in the intact vertebrae above and below was emphasized. Lee and co-workers, in a report of 20 patients treated for tumors of the spine, also noted the importance of approaching at the site of the lesion, being attentive to anterior and posterior stability, and using fixation devices, bone graft, and cement to develop an appropriate surgical construct.[174]

Mechanical Functions of Methylmethacrylate in Surgical Constructs

The material is used in spinal surgical constructs in several fashions. It may be employed as a spacer,[48, 115, 223] an internal splint,[155, 259, 271] and/or a fixation device (see Fig. 8-48). It is apparent that the spacer basically supports compression loading. This provides more than adequate immediate postoperative stability against flexion, with some clinical stability against lateral bending and axial rotation. Stability against extension is not provided in the spacer construct; therefore, the construct should be protected by an appropriate orthosis. The use of methylmethacrylate for splinting and fixation requires bonding of the cement to the bone, which does not occur.[321] When the cement is to withstand bending (tensile) loads, adequate reinforcement is required. The use of the splint here may be somewhat analogous to the plate described by Stener (see Fig.

8-45).[280] While the stainless steel plate is a well-known entity of proven worth with great tensile strength, methylmethacrylate does offer the advantage of a highly individualized configuration, because it can be poured, molded, and packed into various anatomic caverns and crevices.

PRINCIPLES AND INDICATIONS FOR THE USE OF METHYLMETHACRYLATE

A conservative approach to the use of this material is suggested. This is recommended because of the problems and risks associated with its use. Although it is not known whether the material itself potentiates infection, infection in a wound involving methylmethacrylate is very difficult to treat. If the various principles and indications provided are followed and if the physician is convinced that bone alone will not suffice, then the cement should be employed. If methylmethacrylate is used with a clear understanding of the precise role that it is to play in the construct, then clinical success is likely. Is it to be used as a spacer, a splint, or a fixation device? It is probably best as a spacer, then a splint to be affixed to the host bone, and it is most questionable as a solitary fixation device. The surgeon should also determine the absolute or relative importance of immediate postoperative and long-range clinical stability. This consideration is crucial in deciding whether or not to include bone graft arthrodesis in the surgical construct. If long-range stability is needed, then an arthrodesis should be included. The surgeon must also have a basic understanding of the biomechanics of spine stability, surgical constructs, and methylmethacrylate. Finally, it is necessary to have the technical knowledge and surgical ability to achieve the biomechanical goals. These and other principles are listed in the accompanying display and depicted in Figure 8-49.

Below we suggest indications for the use of methylmethacrylate in surgical constructs. At least one of the conditions should be present. One may anticipate that recommended indications will become less stringent in the future provided the clinical experience with the techniques is shown to have a low risk/benefit ratio.

The value of methylmethacrylate in the extremely ill patient should not be underestimated. Its use can readily reduce anesthesia time and blood loss by 50% or more. Pain and complications from a bone graft donor site are also eliminated.

ANALYSIS OF SOME SPECIFIC CONSTRUCTS

Construct for Maximum Immediate Clinical Stability

Given present knowledge and our interpretation of spine biomechanics, we suggest the construct shown in Figure 8-50 for achieving the *maximal* immediate postoperative clinical stability with cement. The important features are as follows: firm fixation of the cement to the spine; reinforcement of the cement by a stainless steel mesh; fixation of the spine by wires and mesh, which is effectively fixed with the cement, into which both are incorporated; the use of two normal vertebrae above and below the pathology for stable anchoring of the construct and for more purchase and interfaces between cement,

TEN PRINCIPLES IN THE USE OF METHYLMETHACRYLATE IN SPINE SURGERY

1. Consider the use of bone before cement.
2. Determine what is expected of the cement.
3. Determine immediate and long-range need of clinical stability. Fuse if necessary.
4. Analyze biomechanics of clinical and surgical constructs, both anteriorly and posteriorly.
5. Develop technical knowledge and ability to design and develop a construct.
6. Use a wire mesh to maintain a barrier between the cement and the dura.
7. Reinforce the cement with stainless steel mesh wire or pins if the construct must resist tension.
8. Provide vertebra with holes, pins, screws, and wires for as much interdigitation as possible between bone and cement.
9. Take care not to put in too much cement.
10. Bone graft if there is a chance for survival of 12 months.

Most of these principles are presented in the work of Clark and associates[46] (see Fig. 8-49) and Dunn[87] (see Fig. 8-51).

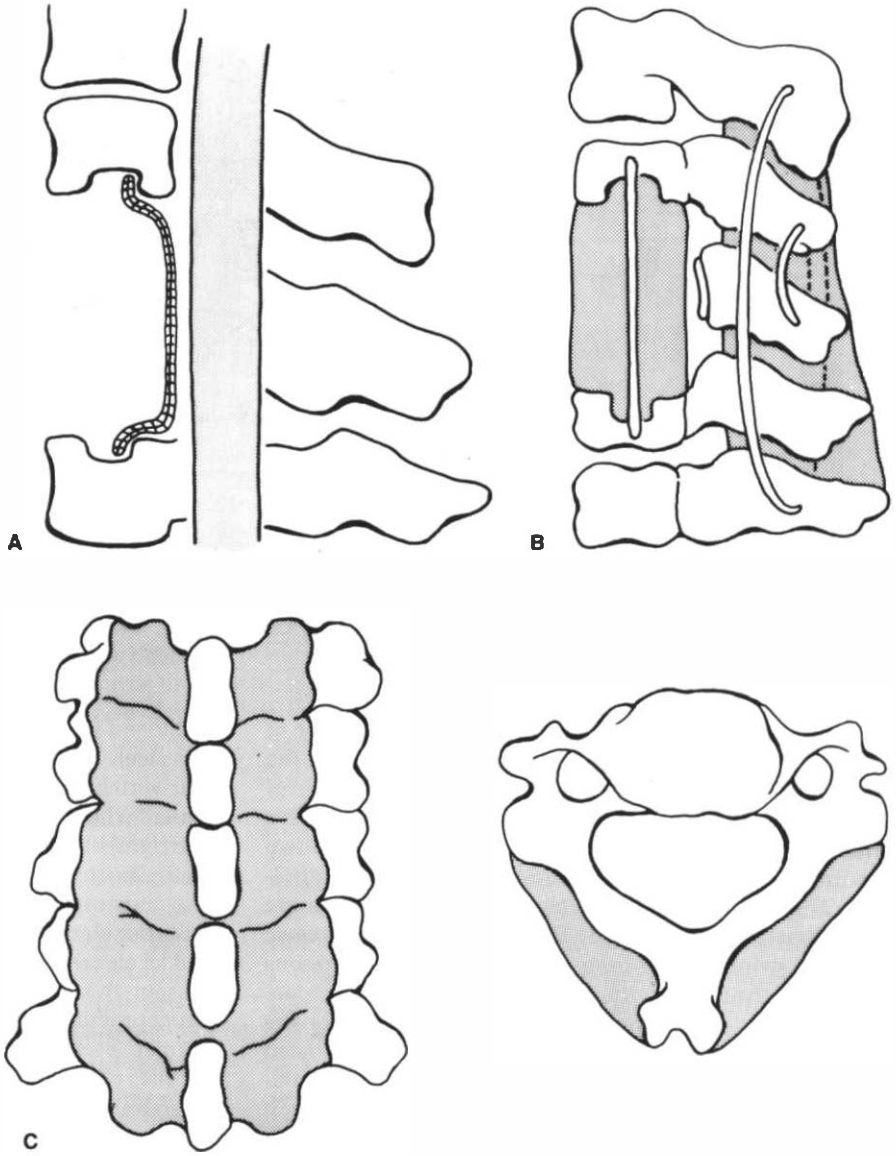


FIGURE 8-49 Illustration of the fundamental factors involved in the most appropriate utilization of cement in cervical spine surgery. **(A)** Provide some shield to protect the dura (wire mesh can be useful). **(B)** Reinforce cement with wire. Use anterior and posterior stabilization in grossly unstable situations. **(C)** Too much cement can obviate wound closure. If the tips of the spinous processes can be seen, the skin should close easily. *(From Clark, C., Keggi, K., and Panjabi, M.: Methylmethacrylate stabilization of the cervical spine. J. Bone Joint Surg., 66A:40, 1984.)*

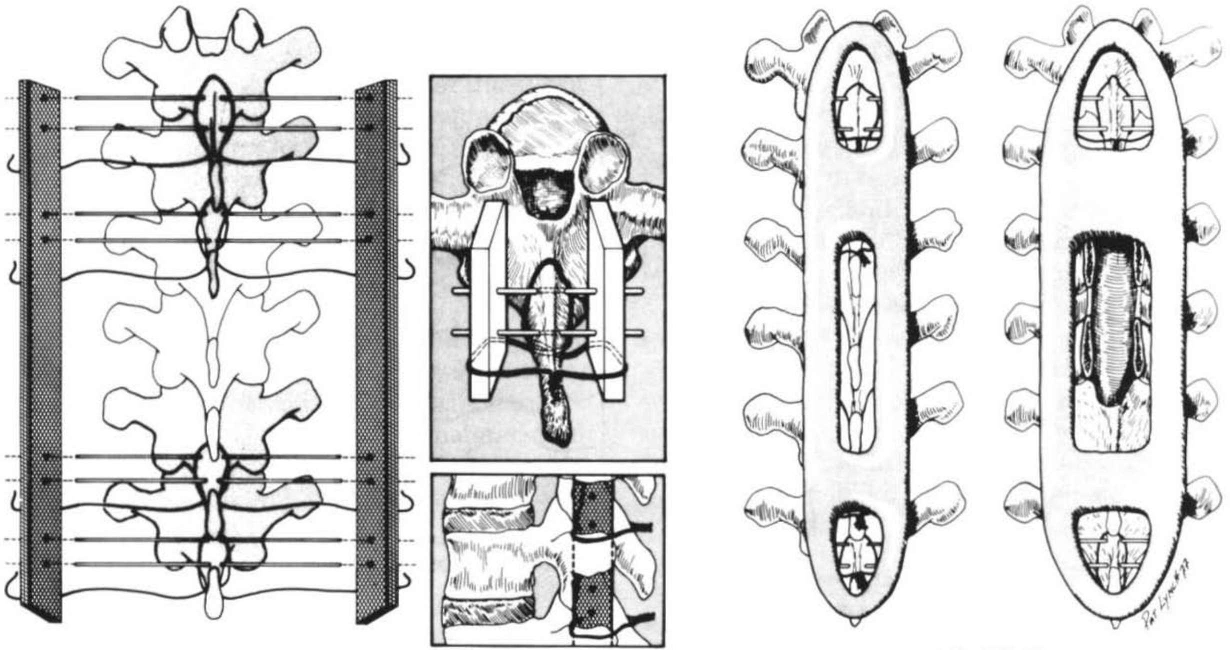
metal, and bone; curettage of the facet articulations in order to attain some biologic arthrodesis; and tapered and rounded structure of the cement ends.

SOME INDICATIONS FOR THE USE OF METHYLMETHACRYLATE IN SPINE SURGERY

Maximum immediate postoperative clinical stability is crucial to the survival of the patient.
There is no source for bone autograft or allograft.
The patient is extremely ill and requires stabilization.

The same basic construct may be employed in patients with multiple laminectomies. In such patients, there are two normal vertebrae above and below the uppermost total laminectomy, and the rolled wire mesh is wired to the facets, just as the bone graft is attached as shown in Figure 8-50. If there are facetectomies, the stainless steel mesh and cement filler are used to bridge the defect.

One possible advantage of the basic construction presented here is the relative ease of removal. If the facets are not wired, the entire wire–cement complex can be removed by resection of the spinous process near its base.



STAGES I AND II

STAGE III

FIGURE 8-50 Based on our analysis and appreciation of the clinical biomechanics involved, we recommend the surgical construct illustrated here as the procedure of choice to provide the maximum immediate postoperative stability with methylmethacrylate. ***Stage I*** exposes two *normal* vertebrae above and below the level of the pathology. This is to provide adequate purchase for the fixation splint (methylmethacrylate). Place two parallel transverse $^{5}/_{64}$-in K wires at the base of the four normal spinous processes. Anterior to the transverse K wires, wrap a no. 24 twisted stainless steel wire around the base of the spinous process and twist it five to ten times. (The twisted wire forms a more stable interface with the methylmethacrylate.) Let these wires extend out laterally in both directions. ***Stage II.*** Take two strips of stainless steel mesh of the appropriate length, 1 cm in width. This should be punctured with an awl, making holes large enough and in the proper position to admit the four transversely placed K wires. The ends of the twisted wires are then crossed and brought around the strips of wire mesh and twisted together so as to hold them in place. ***Stage III.*** Methylmethacrylate is then prepared, and while it is still soft, the oval shape with the tapered ends and transverse bars is fashioned in and around the previously implanted stainless steel. Stages I to III are also carried out on the abnormal vertebrae if the posterior elements are intact. In a patient who has undergone laminectomies, a modification (*far right*) should be used. The implantation of too much methylmethacrylate can make wound closure difficult. The construct may be used regardless of the status of the posterior elements of the involved vertebra. If both lamina and facets are absent, the region is bridged; if the facets are present, the facet wiring (see Fig. 8-40) is done. This construct provides secure splint fixation by attachment to two normal vertebrae above and below, secure methylmethacrylate–bone interface through stable wiring, wire mesh reinforcement of the methylmethacrylate, and minimization of stress concentration by decreasing the mass of material and thus the stiffness at each end of the oval. We consider that this construction appropriately demonstrates several of the principles of the use of cement. However, the use of rigid transpedicular fixation and bone graft is preferable when the bone is of good quality. If cement construction is used and is needed for more than 1 year, then bone graft should be included.

Cement as a Temporary or Permanent Spacer

In cases of extensive anterior element destruction by tumor or resections of one or more vertebral bodies, cement may be useful. Methylmethacrylate may be used in this situation as a spacer or a filler (Fig. 8-51). The cement provides a good purchase in the cancellous bone. If necessary, a notch, a trough, or some undermining procedure may be useful. Several examples of the use of methylmethacrylate as a spacer are given in Figure 8-51. Various techniques for pre-

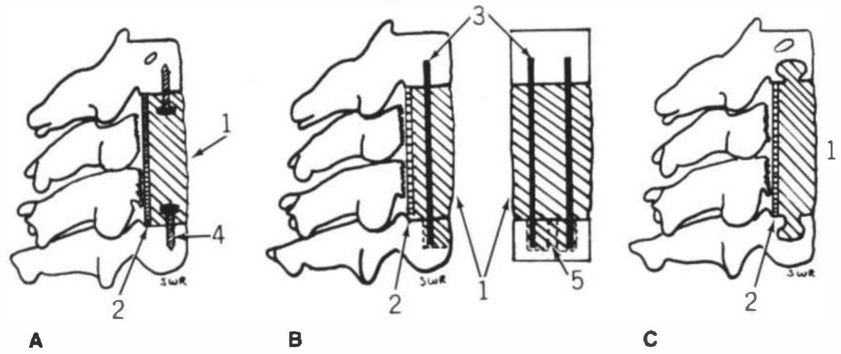


FIGURE 8-51 Some of the different constructs for securing the methylmethacrylate–bone interface when methylmethacrylate (*1*) is used as a spacer. **(A)** The use of screws (*4*) in cancellous bone. The construct can be expected to be more stable if a cancellous bone screw is used. This constitutes a sound construct against extrusion and against tensile loading. **(B)** Here, K wires (*3*) are employed. This construct is reasonably effective in resisting extrusion but not very effective against tensile loading. **(C)** This undermining technique, when well done, is effective against extrusion and tensile loading. (*2*) This represents a bone plate positioned to protect the dura and its contents. Experiments are required to determine assuredly the relative effectiveness of the three constructs. We favor the construct shown in **C**, followed by **A** and then **B**. *(Dunn, E. J.: The role of methylmethacrylate in the stabilization and replacement of tumors of the cervical spine. Spine, 2:15, 1977.)*

venting extrusion of the cement block and resisting tensile loading during bending (as in extension) are shown. It is difficult to compare the relative effectiveness of the three constructs without any experimental data. They all look as though they would be stable in flexion. The crucial factor is stability in extension. The construct with the cement packed into the mushroom-shaped space should be the strongest. We believe that screw-fixed cement is more secure than cement fixed by K wires.

Ono and Tada have employed a simple but ingenious device to be used as an anterior spacer to replace a vertebral body.[223] The device and the surgical construct are shown in Figure 8-52. The quadrilateral cylinder controls the placement of methylmethacrylate and prevents it from overheating the spinal cord. Ono has suggested that the construct be employed in those patients who have primary cancer localized in the spine or metastatic cancer in the spine without pulmonary metastasis. He warns that if the patient has a tracheostomy, this cement–metal construct is prone to serious infection.* We have suggested a modification of the Ono device, one that provides some immediate postoperative clinical stability against extension. The modification involves the use of a screw transfixing the prosthesis to the remaining anterior lip of the vertebra and to the methylmethacrylate inside the vertebra. This modification of the Ono device is shown in Figure 8-52E.

Another technique for ensuring clinical stability in these situations is to supplement the construct with posterior fusion immediately or later, when the patient's condition permits. The spacer resists compressive forces very effectively. However, there is no good resistance against tensile forces by either the cement or the interdigitation interface. However, a posterior bone graft provides stability by acting as a band to resist tension in flexion (as do intact posterior ligaments), and in extension it resists compressive loads well. The posterior bone graft to augment the anterior spacer is therefore an advisable adjunct when feasible. If the posterior elements are destroyed or unable to function, use of the anterior spacer alone is very likely to leave a clinically unstable situation.

Since several complications of structural failure have been reported in patients with constructs containing methylmethacrylate,[46,120,174] it is suggested

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* Personal communication, K. Ono., M.D., December 1984.

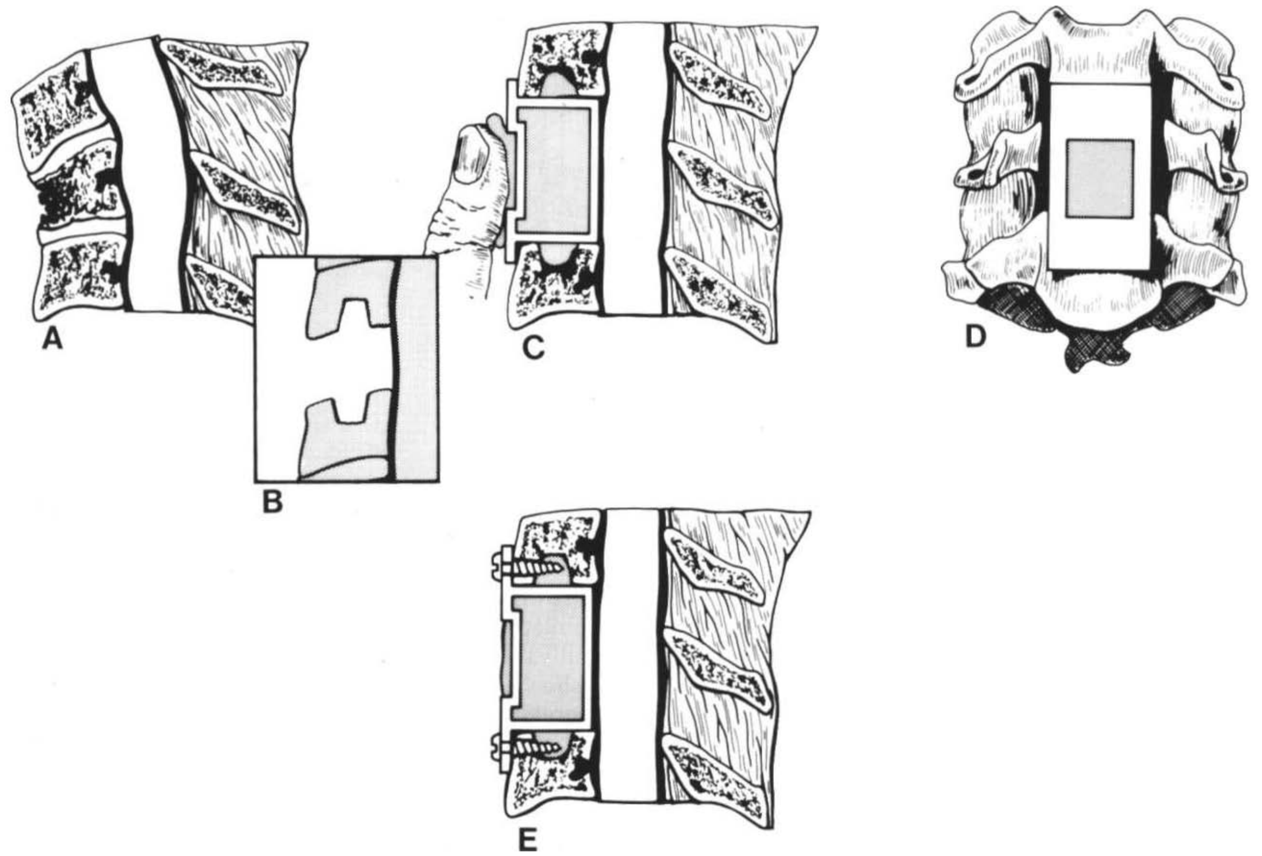

FIGURE 8-52 The Ono device. This stainless steel shell is used in conjunction with methylmethacrylate when the latter is used as a spacer. **(A)** The diseased vertebra is shown. **(B)** The vertebral body is resected, and a rectangular tunnel of bone is made in the two adjacent vertebrae. **(C)** The device is put in place, and the methylmethacrylate is packed in. **(D)** An anterior view of the construct with the cement in place. The device protects the neural structures from the cement and is designed so that the cement can be thoroughly packed. On the lateral view **(C)**, one can observe bone wedged between the methylmethacrylate and the outside rim of the shell. The locking of this little segment of bone tends to resist extension and make the construct more stable. **(E)** We suggest the following modification of this construct, which will offer additional immediate postoperative stability against extension. The rectangular steel device is altered so that two screws can be placed through the stainless steel shell, into the anterior portion of the vertebral body, and into the methylmethacrylate. This device has been subsequently modified by Ono and associates.[224] The modification involves the device as shown in **A**, **B**, **C**, and **D**, but made in a ceramic to facilitate osseous ingrowth.

that some attention be paid to the technical aspects of the use of this material. A review and analysis of the purposes for which the material is being used is helpful. If the surgeon is conversant with the information concerning general principles and analysis of the constructs presented here, he may be better equipped to use this material successfully as a spine implant.

It is of interest to note that methylmethacrylate has been used as a spacer for the intervertebral disc,[48] despite the striking paucity of biomechanical similarity between the two substances. The procedure consists of laminotomy, removal of the herniated disc fragment or the entire disc, and insertion of soft, freshly prepared methylmethacrylate into the interspace. Cleveland reports having performed this on 126 patients with "extremely satisfactory results."[48] We view this with respectful skepticism and look forward to a more detailed documentation of the effectiveness of this procedure.

Comment

With the advent and development of better spinal instrumentation and the availability of various types of allografts, we anticipate that the goals of immediate postoperative stability as well as long-term stability can be achieved. Thus, cement will be employed in fewer instances—perhaps only as a filler or a spacer, to enhance or reinforce screw purchase, and for salvage procedures in patients for whom the surgery must be done quickly.

SOME KEY FUNCTIONS IN SELECTING A GIVEN IMPLANT

Principle and rationale
Biomechanical validity
Component analysis
Combined functional evaluation
Status of patient's bone
Operative and postoperative risks
Experimental studies of implant
Clinical studies on use of implant
Costs • User friendliness • Availability

PART 4: BIOMECHANICAL CONSIDERATIONS IN THE ART AND SCIENCE OF SPINE INSTRUMENTATION

For a book on the clinical biomechanics of the spine, the following section is perhaps the most significant one from intellectual, academic, practical, basic science, and clinical perspectives. This section must bring together bioengineering information that relates to anatomy, technology, metallurgy, surgical technique, surgical judgment, clinical pathology, and clinical evaluation. We have assiduously avoided any tendency to produce an instrument catalog, a consumer report, or a "cookbook" on spinal instrumentation. Rather, our goal is to share with the reader our best effort to provide the necessary understanding and information to make the best clinical decisions about the selection of surgical constructs, given the currently available knowledge and instrumentation (implants).

The goal for this part of the chapter is to provide for the reader a framework—some information and suggestions that will be useful in bringing a biomechanical perspective to the clinical decision-making process of selecting and using spinal implants. Perhaps the crucial first questions for the clinician to answer are "What am I trying to achieve with the instrumentations?" and "Why?" The next question is "Given the experimental and clinical information available, what is the best implant to achieve what I must achieve at the least risk to my patient and (philosophically for some) the least cost to my society?"

How might one consider a given implant? A set of considerations is summarized in display above, and is discussed in detail below.

Principle and rationale: This refers to the basic advantage, purpose, and benefit of the device. For example, a Harrington distraction rod provides bending moments at the ends of a scoliosis deformity to straighten the curve.

Biomechanical validity: Does the advantage or benefit of the device make sense in view of reasonable validated biomechanical theory and the goal(s) the surgeon has identified for the surgical procedure?

Component analyses: Have the various components of the device been tested, and are they known to be mechanically sound and of appropriate mechanical capabilities? Have they been tested appropriately for stiffness, failure loads, fatigue, and pullout strength?

Combined functional evaluation: How does it all work when tested with a spine (*i.e.*, strength, fatigue, stiffness, efficacy of distraction and/or compression)?

Status of patient's bone: Bones materially or structurally altered by disease may be rendered inappropriate for certain implants.

Operative and postoperative risks: Certain implants clearly are associated with more complications or potential complications than others.

Experimental studies: The more thoroughly tested implants provide the surgeon with a better known entity and thus a more focused and accurate analysis of risks and benefits. *In vivo* animal studies can also provide valuable information.

Clinical studies: High-quality clinical studies provide the ultimate basis for the evaluation of risks and benefits.

Costs/User friendliness/Availability: Costs may be prohibitive or may be a factor in the selection among similar devices. User friendliness is a broad but important consideration that includes ease of application, quality of instrumentation used to implant a device, templates, safety checks, and accessibility of good instruction. Availability includes the status of a device as regards government and/or other regulatory agencies. Obviously, the foregoing analysis may point to one or more implants, and availability may determine the final selection.

WIRES, MESH, AND SCREWS

The use of stainless steel wire posteriorly in surgical spine fusion has been very useful. It may serve a large variety of functions, most frequently being used as a tension band to coapt and fix posterior elements. It is also employed to attacb, immobilize, and secure bone graft to the recipient site.

The Best Use of Wires

There are three excellent studies that provide the surgeon with useful information upon which to base decisions about stainless steel orthopedic implant wires.[110,219,258] Several of the basic wiring techniques are presented in Figures 8-53 and 8-54. Our selection of the most cogent information from these studies is presented in the above chart. For more details, the articles can be consulted.

Basic Information

- As expected, the larger wires are stronger.[258] Fastening twists are stronger than knots or the Association for Study of Internal Fixation (ASIF) bend techniques:[110,258] however, square knots are acceptable.
- Two twists are enough; additional twists add no additional strength.[110,258]
- Commercial wire twisters were better than ordinary pliers.
- Two single-wire loops are better than a continuous double-wire loop.[258]

Cautions

- Notching 1% of the diameter reduces fatigue resistance by 63% (bending, twisting, and knotting the wire had no serious effect).[219]
- Tension-equalizing loop opposite fastening loop weakened system 10–15%.[258]
- The wire wrap and the ASIF loop techniques were unacceptable.[110]
- "Gauge" numbers apparently are not standardized internationally.

Wires and Lamina Fixation

A number of spinal surgical procedures employ the passage underneath or anterior to the lamina. Biomechanical studies show that this is one of the most

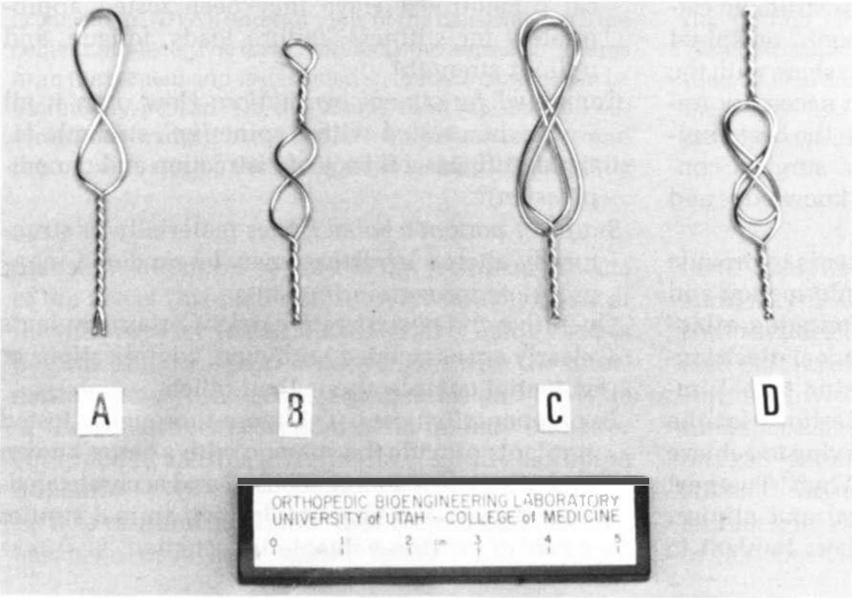

FIGURE 8-53 These four wire constructs are arranged in increasing order of load bearing to failure. **(A)** Single-strand figure 8 was the weakest of the four. **(B)** Single-strand figure 8 with a tension-equalizing loop opposite the fastening twist had a higher load to failure than did A. **(C)** Double figure 8 in a continuous loop was stronger than B. **(D)** A construct made of two single-strand figure 8s was the strongest. *(From Schultz, R. S., Boger, J. W., and Dunn, H. K.: Stainless steel surgical wire in various fixation modes. Clin. Orthop., 198:304, 1985.)*

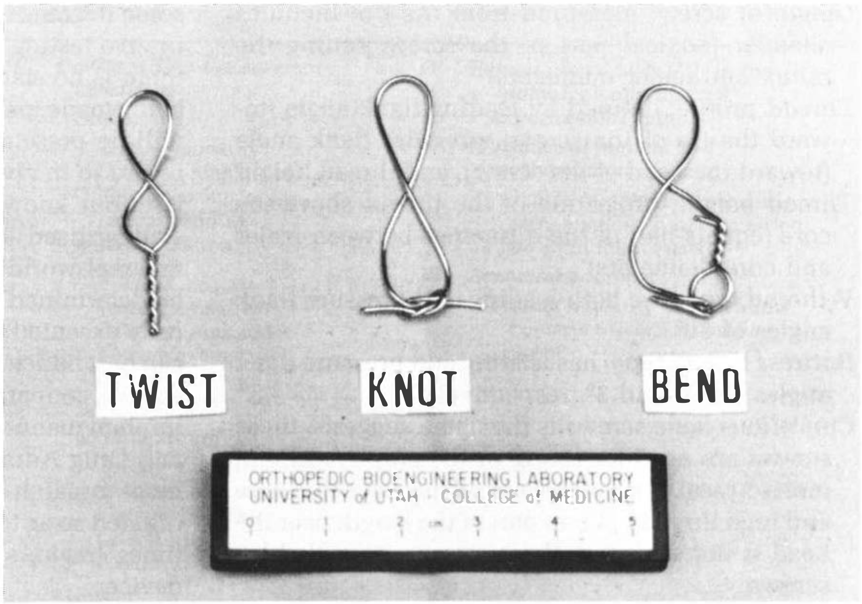

FIGURE 8-54 This group of three is listed in decreasing order of strength as measured by the maximal tensile load to failure. Twisting was 32% stronger than knotting and 56% stronger than bending ($p < .01$). (From Schultz, R. S., Boger, J. W., and Dunn, H. K.: *Stainless steel surgical wire in various fixation modes.* Clin. Orthop., 198:304, 1985.)

effective methods for achieving a very stiff or stable fixation. Unfortunately, there are significant liabilities in the process of lamina wiring. Nicastro and associates[212] showed that even when experienced surgeons used careful techniques during "surgical passage," the wires went further into the canal than would be ideal. Blackman and Toton[15] demonstrated that upon removal of a sublaminar wire, again the extent of intrusion into the canal was more than was desired or should be tolerated. Schrader and coworkers[257] showed in dog studies that during extraction of sublaminar wires there was an average of 47% indentation of the dural sac by the wires as observed on the myelographic column during cineroentgenography. Observed complications included hemorrhage—epidural, subdural, and intramedullary; epidural adhesions; dural lacerations; cord indentations; and neurologic damage. Moreover, there have been several clinical examples of either immediate[12,147] or delayed[216,248,325] complications from the use of interlaminar wire techniques.

Goll and associates[100] have offered some suggestions that may reduce some of the untoward effects of sublaminar wire passage. They recommend that: lateral passage be avoided, the radial curvature of the wire equal at least the width of the lamina, the bend of the tip be no greater than 45°, and the spinous process be removed routinely for midline passage.

We prefer to avoid techniques requiring the use of laminar wires, particularly in view of the fact that there are alternative implants and surgical constructions that, given current knowledge, are just as effective.

Mesh

Stainless steel wire mesh is available in several different sizes. We recommend that it be used whenever possible to reinforce methylmethacrylate that will be subjected to significant tensile and bending loads. The size suggested is the stiffest one that is compatible with a reasonable ease of handling. This mesh should also permit ease of penetration with an awl so that it can accept wires and pins of various sizes.

Screws

Before we describe the use of screws in the fixation of the spine, a short review of the salient features of a screw is provided.

Pitch: distance between adjacent threads
Major diameter: outside diameter of screw—used for defining screw size
Shank diameter: diameter of the unthreaded part of the screw
Minor or core *diameter:* diameter of the screw at the base of the thread (also, it equals the major diameter minus twice the thread height)

Length of screw: measured from the tip; includes chamfer (conical part of the screw joining the minor and major diameters)

Thread profile: defined by leading flank angle (toward the tip of the screw), pressure flank angle (toward the head of the screw), and thread height

Thread height: projection of the thread above the core (equals half of the difference between major and core diameters)

V-thread type: has both leading and pressure flank angles of 30°

Buttress thread type: has leading and pressure flank angles of 45° and 3°, respectively

Cancellous bone screw: as the name suggests, these screws are used for fixing in the epiphyseal and metaphyseal areas of the bone. They have thin core and high threads. When part of the length near the head is not threaded, these screws are called lag screws.

Cortical bone screw: these screws are used to fix a plate to the cortex of the bone. They have relatively shallow threads, which are over the entire length of the screw.

A technique involving the use of screws has been described for C1–C2 fusion. Screws have also been recommended in the lumbar spine.[23,158] With regard to their role in preventing displacement, screws can be presumed to be effective in the cervical spine in view of the magnitude of the physiologic loads exerted there. However, in the lumbar spine, either across a spondylolisthesis defect or across the facet joints, it is unlikely that screw fixation is an effective method of immobilization.[10,23,129] We do not believe that the screw techniques described for the lumbar spine have proved to be beneficial enough to expose the patient to the possible complications of hardware failure or neural or vascular irritation.

When a screw is intended to be anchored in cancellous bone, a cancellous bone screw should be employed, using the appropriate technique. If the screw is to be anchored in cortical bone, the screw should be properly tapped.

SPINAL IMPLANTS

Implant Testing

The issue of implant testing is of the utmost importance. The topic possesses complexity of monumental proportions. Nevertheless, it is crucial to achieve some degree of standardization of both in vitro and in vivo testing protocol. At the time of this writing, there is no standard protocol; however, progress is being made, particularly for the in vitro testing. This will be presented in summarized form here. With regard to in vivo experimentation and testing, there is to our knowledge little or no movement toward standardized testing. Ultimately, the overall accurate real-world risks/benefits of spinal implants can be determined only by astutely designed and carefully executed clinical studies. We believe that this can best be achieved through the considered, collaborative, cooperative efforts of investigative surgeons, implant manufacturers, third-party payers, the Food and Drug Administration, and private and government research funding agencies. This must be coordinated so as to avoid useless, inappropriate, sometimes tragic, sporadic, uncontrolled trials of new devices.

The value of standardized testing protocol is to provide a useful basis for comparison of present and future implants as well as to build a useful data bank for general knowledge about the biomechanics of spine instrumentation. Since there is not yet a generally accepted standard, we have elected to present here a synopsis and rationale for two somewhat similar approaches. The one presented by the authors is more of a conceptual framework, and the other (the development of which was sponsored by the United States Food and Drug Administration) is a detailed experimental protocol.

Our proposal is based largely on the work of Panjabi.[225] The ideal would be to have a comprehensive mechanical characterization of the device alone and in the spine construct. The types of tests to be done are shown in Table 8-5. The goal of the protocol is to evaluate the *multidimensional experimental construct stability*[E] of the intact and pathologically altered spine implant construct. This multidirectional analysis addresses the importance of the physiologic multidirectional kinematics and kinetics of the patient's spine. The flexibility method for testing the spine constructs is suggested because in this method a load is applied to a free vertebra and the displacements are measured. This method allows natural movement of the spine to take place.

Another important conceptual consideration in the protocol has to do with the method of load application to the surgical construct. The goal is to load a complex construct uniformly so that failure will occur at the weakest point rather than at some point

TABLE 8–5 Implants Testing Conceptual Protocol

- *I. Devices alone*
  - Stiffness
    - Testing modality[*]
    - Data and rank order[†]
  - *Flexibility*
    - Testing modality
    - Data and rank order
  - *Load-bearing capacity*
    - Testing modality
    - Data and rank order
    - Sites of failures
  - *Fatigue tolerance*
    - Testing modality
    - Data and rank order
    - Sites of failure
  - *Special features*
    - Plate–screw junctions
  - *Mechanical properties*
    - Of other fixtures
- ***II. Device and spine interface (hooks, screws)***
  - *Pull-out strength*
- ***III. Device within the surgical constructs***
  - *Stiffness*
    - Testing modality
    - Data and rank order
    - Flexion/extension
    - Lateral bending
    - Axial rotation
  - *Flexibility*
    - Testing modality
    - Data and rank order
    - Flexion/extension
    - Lateral bending
    - Axial rotation
  - *Load to failure*
    - Testing modality
    - Data and rank order
    - Flexion/extension
    - Lateral bending
    - Axial rotation
  - *Strain*
    - Testing modality
    - Data and rank order
    - Flexion/extension
    - Lateral bending
    - Axial rotation
  - *Fatigue tolerance*
    - Testing modality
    - Data and rank order
    - Flexion/extension
    - Lateral bending
    - Axial rotation
    - Sites of failures
- ***IV. Repeat above tests (III) in the "pathologically altered" experimental spine.***
- ***V. Mechanical models and mathematical models***
  - *Adjunctive potentially useful tests (explained and reported on individual basis)*
- ***VI. Recommendations***
  - Testing modality should allow construct to fail at weakest point.

[*] Testing modality refers to information about how the structure is tested (i.e., compression, tension, three-point bending, four-point bending, and so forth).

[†] Data and rank order refers to the actual quantitative findings as well as comparative sequential ranking of the implants tested.

where there is excessive load due to the method of load application. Figure 8-55 is a review of bending moment diagrams for the various loading techniques. The loading technique C in this figure, a pure bending moment, produces the bending moment diagram shown in Figure 8-55C. This provides constant bending moment at each level of the spine implant construct, and it will fail at the "weakest link" (weakest point) in the construct. This, of course, is a point that is important to identify for clinical, experimental, and design purposes.

The fatigue characteristics of the implant are of great clinical importance because of the classic race between the fatigue life of the implant and the maturation of the clinical fusion mass to the point that it can take over the forces and unload the implant system (Fig. 8-56). Specific *in vivo* precise stereophotogrammetric study shows that this takes 6 months.[222] We must also recognize that even after solid fusion, the implant may be subjected to some motion and loads.[222]

The conceptual outline for the testing protocol is presented in Table 8-5. Even though this is not an accepted standardized protocol, it is suggested that the reader review it as a summary of all the variables that may be tested and also as a working frame of reference for a review of experimental data on various available spinal implants.

The next outline for the organization of a test protocol that we would like to present is that devel-

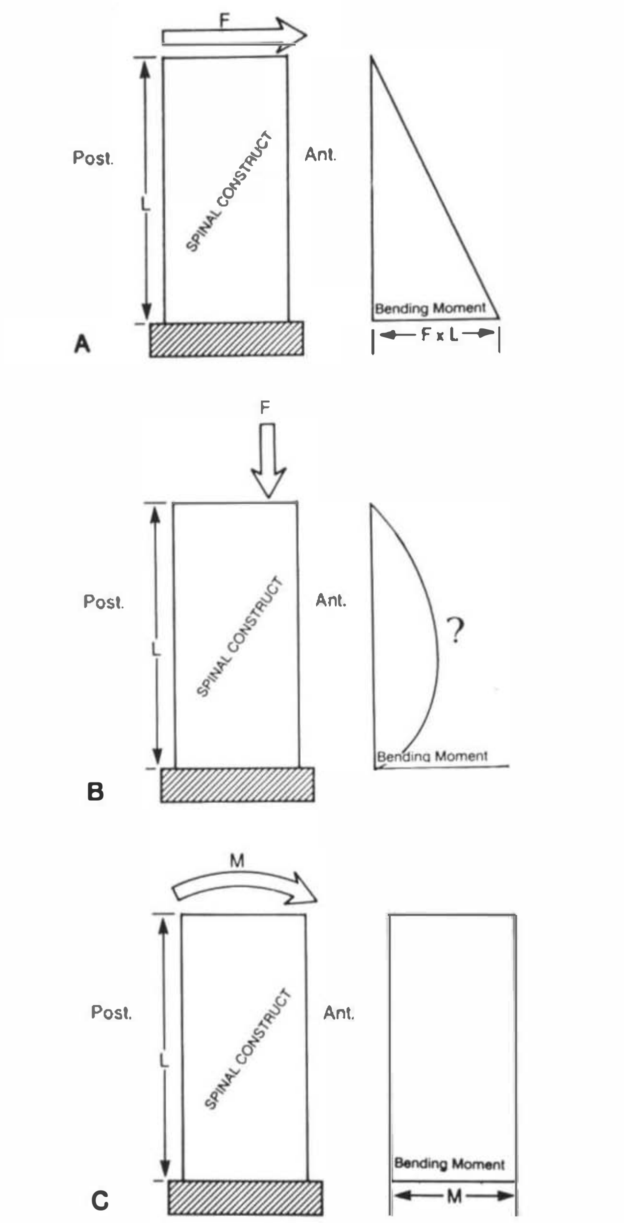

FIGURE 8-55 Bending moment diagrams for three different methods of producing flexion of the construct. **(A)** Shear force applied at the free end produces increasing bending moment towards the fixed end. **(B)** An anterior eccentric compression force produces variable bending moment as the construct deforms. (C) A pure bending moment results in the most uniform loading of the construct. This is the recommended loading method. *(From Panjabi, M. M.: Biomechanical evaluation of spinal fixation devices. I: A conceptual framework. Spine, 13:1129, 1988.)*

CLASSIFICATION OF SPINAL IMPLANT COMPONENTS

A. Longitudinal structural elements
  1. Rods
    a. Smooth
    b. Ratcheted
    c. Threaded
    d. Knurled
  2. Plates
    a. Notched
    b. Individual holes
  3. Cables
B. Elements that attach to vertebrae
  1. Screws
    a. Pedicle screws
    b. Anterior screws
  2. Hooks
    a. Laminar hooks
    b. Pedicle hooks
    c. Transverse process hooks
  3. Wires
    a. Sublaminar wires
    b. Transverse process wires
C. Elements that connect to the pelvis or sacrum
  1. Sacral screws
  2. Galveston technique
  3. Sacral hooks
D. Elements that connect longitudinal elements together
  1. DTT
  2. Wires
  3. Unitized rods
  4. Cross-links

From Ashman, R.B., Bechtold, J.E., Edwards, T., Johnston, C.E., McAfee, P.C., and Tencer, A.F.: *In vivo* spine implant mechanical testing protocol. Submitted for publication, Spine, 1989.

oped by Ashman and colleagues.[8] This outline was also influenced by Panjabi and associates, and contains some of the same elements. The detailed, well-referenced work of Ashman's group includes some excellent recommendations.

Virtually all implants can be broken down to basic components, and implant components are classified in the display above. The complexity of spinal implant testing is readily apparent when one reflects upon the numerous devices included on this list. The mechanical characteristics of these devices in relation to each other and to the bone should be quantified and compared. These authors have chosen the coordinate system that has been suggested in the past.[229, 316] The importance of testing for rotation about the three major planes is indicated. The ana-

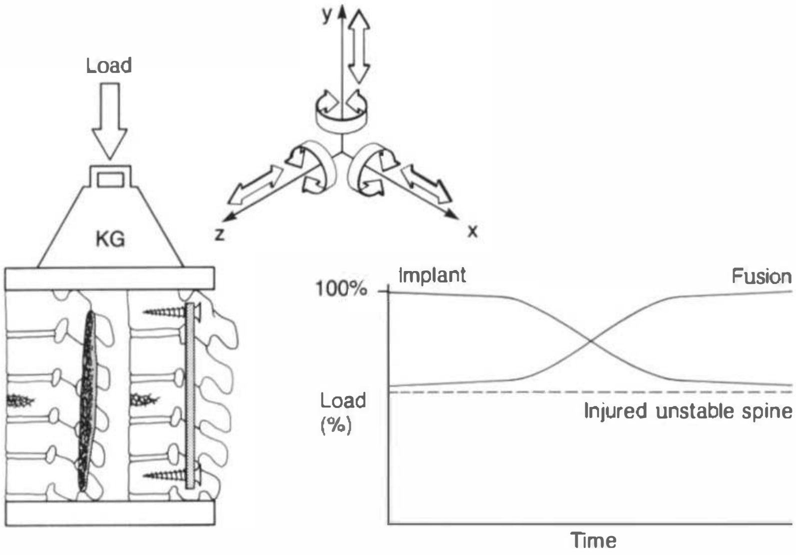

FIGURE 8-56 The spine implant composite is loaded with the various components of forces and moments shown on the coordinate system. The compressive load has been selected to demonstrate the concepts of load sharing and fatigue failure. The conceptual illustration shows that the load is shared by the injured/unstable spine, implant, and fusion mass. Initially, the fusion mass has no strength, and therefore the load is mainly shared by the implant and the spine. However, with time, the fusion mass gains stiffness and takes over more of the load, thus unloading the implant. If the magnitude and frequency of the spinal loads is such that the fatigue tolerance limit of the implant is exceeded, the implant will fail. Thus, there is a kind of race between the healing fusion and the fatiguing implant. The loss of bone support (*i.e.*, significant decrease in the load-bearing capacity of the injured/unstable spine) may significantly increase the implant load and cause premature failure. This diagram is presented to emphasize a conceptual point. The reality is unknown but is certainly more complex than we have depicted here. The relative magnitudes of the load sharing in time are not known, and it is unlikely that the "implant curve" and the "healing fusion curve" are mirror images of each other.

tomic (structural) dissimilarities in the various animal and human spines are addressed. Specific limitations in extrapolations from animal to human spines must be considered when test data are evaluated. The authors also point out the importance of the selection of the appropriate methodology for statistical analysis to be used with the experimental design. They emphasize that this selection is best done as an integral part of the design and planning of the experiment.

The protocol discusses the usefulness of normalizing data analysis. This is made possible by using an experimental design that compares mechanical properties of the same intact spine with itself after the introduction of some experimental variable. It is recommended that the design be clear about the experimental creation of various types of modes (*i.e.*, "burst fracture," "scoliosis," and so forth). This is crucial for the interpretation of results. For example, the decision to use one implant over another should not be based on the demonstrated superiority of that device in a "scoliosis" experiment protocol when the clinical decision to use it is going to apply to a patient with spond*ylolisthesis*. The outline for the proposed protocol is presented in a display to help the reader organize and evaluate experimental data on various spinal implants. We would also like to share the following quote from the article by Ashman and colleagues: "It is believed that the experimental conclusions from different institutions using in vivo and ex vivo specimens and animal models should be combined to initiate the basis for controlled clinical trials. In this manner the development of newer spinal implants can progress with the least chance of clinical complications."[8]

There are several additional concepts that we would like to present before the discussion of the various implants.

Ideal Rigidity

Theoretically, an *extremely* rigid plate *very* securely affixed to two or more vertebral bodies also spanned by a bone graft could result in failure of fusion because of stress protection. The healing grafted area would be exposed to too little load and/or too little motion, the graft would resorb, and no arthrodesis would occur. Here is another scenario. In a grossly unstable spine, two or more segments are wired loosely through the spinous processes and bone grafted. There is no external fixation and too much motion and/or too much shear and tension in the region. The bone graft fails to mature, it is absorbed, and there is a failure of fusion. We assert that somewhere between those two extremes there exists a range of stiffness of the abnormal spine/bone graft/implant construct that is ideal for bony fusion to become as strong as possible as soon as possible. This ideal stiffness (or ideal stiffnesses) is at the time of this writing unknown. We submit that this is an

ORGANIZATION OF TEST PROTOCOL

In vitro mechanical testing protocol

A. Coordinate system
B. Application of loads
    1. Directions of loads
    2. Load rates
    3. Mounting of specimens
C. Measurement of displacements
    1. Displacements at the load point
    2. Displacements between vertebral bodies
    3. Coupled displacements
D. Definition of spine model
    1. Human cadaver
        a. Relative quality of bone
    2. Animal models
        a. Bovine calf
        b. Porcine
        c. Canine
        d. Goat
        e. Others
E. Results of mechanical testing
    1. Vertebral alignment
        a. Static force analysis
        b. Stiffness
            Average stiffness
            Tangent stiffness
            Deflection at a certain load
            Load at a certain deflection
            Stiffness vs. load
    2. Maintaining alignment
        a. Implant component strength
            Static strength
            Fatigue strength
                In air vs. saline
        b. Construct strength
            Static strength
            Ultimate strength
            Yield strength
            Energy absorption to failure
            Fatigue strength
                Stress vs. number of cycles to failure
                Load vs. number of cycles to failure
                Load vs. stresses in implants
    3. Promote solid fusion
F. Discussion of results
    1. Relative comparison between different constructs with the same number of FSUs
    2. Normalization to intact and uninstrumented spine
    3. Normalization of bone quality
G. Specific test protocols
    1. Burst fracture
    2. Scoliosis
    3. Kyphosis
    4. Spondylolisthesis
    5. Postlaminectomy instability
    6. Ligamentous instability (fracture dislocations and flexion/seat belt injuries)

(Ashman, R.B., Bechtold, J.E., Edwards, T., Johnston, C.E., McAfee, P.C., and Tencer, A.F.: *In vitro* spine implant mechanical testing protocol. J. Spinal Disorders, 2:274, 1989.)

important question for the analysis of current and future spine implants.

Variable Rigidity

This is a hypothesis that suggests that the ideal rigidity is in fact a range of rigidities that should vary during different stages of healing. For example, during the early stages of healing, when delicate new vessels are forming to vascularize the fusion site, a more rigid construct is desirable. Then, as healing progresses and more bone is produced, it may be that a less rigid construct, which will allow more motion and/or loads to be applied to the developing fusion, is desired. The hypothesis presumes that mechanical stimulation through Wolff's law will facilitate the development of the fusion. Thus, the hypothesis of variable rigidity purports that we provide the healing arthrodesis with a highly rigid construct at the beginning of healing and a less rigid one in the latter stages.

Strain

Perren and Cordey introduced the concept of *interfragmentary strain* as a possible explanation for the production of different histologic types of fracture healing.[235] One experimental technique to consider in addressing the issue raised by the concept of *ideal rigidity* is the measurement of strain. Nagel[206] has introduced this into the design and interpretation of *in vivo* experiments of spine fusions in sheep. He noted that a 36% strain in the region of the bone graft resulted in nonunion, whereas a strain of only 10% allowed the bone graft to fuse. Strain in the region of the bone graft is considered a useful measure, even in the *in vitro* studies of spine implant constructs.[8,49]

CERVICAL SPINE IMPLANTS: C0–C7

Introduction

This section will review the most cogent experimental and clinical work on the topic of cervical spine implants and will offer comments and recommendations. We have not attempted to catalog all available implants; however, we have chosen to include certain implants based on the following criteria: (1) representation of various prototypes, (2) information available about the implant, (3) presumed frequency of use of the implant, and (4) our opinion of the overall importance of the implant. The traditional wire implants are presented in the section on page 580.

Experimental Studies

The literature as of mid-1988 contained few experimental studies of internal fixation of the cervical spine except for some on the use of wire.

Ulrich and colleagues[298] studied fresh cadaver cervical spines to compare anterior and posterior cervical spine fixation procedures. The investigators selected C5–C6 FSUs, applied a bending load to simulate flexion, and measured sagittal plane translation and rotation. Posterior ligaments were removed to simulate posterior instability, and the FSUs were fixed with (1) an H-plate anteriorly, (2) a hook plate posteriorly, or (3) sublaminar wires. Various combinations of these were also tested. At the end of these studies, the authors concluded that the hook plate alone appeared to provide stability. The use of interlamina wire alone or anterior H-plate fixation alone did not appear to provide adequate stability.

Hanson and associates,[117] in biomechanical studies of fresh human cadaver specimens, were able to compare C1–C2 Gallie-type posterior wiring with the Magerl bilateral screw fixation of the lamina of C2 to the lateral masses of C1. The stiffness and load to failure were studied in both flexion and axial rotation. The screw fixation was stiffer than the wiring technique.

Montesano and Juach[200] created unstable C2–C7 human cervical spine specimens by ablating the anterior elements. They then applied posterior cervical spine plates and the techniques of Magerl and Roy-Camille, respectively. The former employs two plates attached to the lateral masses by 20–22-mm 3.5 cancellous bone screws directed anteriorly, laterally, and cephalad. The latter technique uses a 12–14-mm screw attaching two plates to the lateral masses by directing the screws caudally toward the facet joint. The C5 vertebral body was removed, and one of the two plating techniques was used to stabilize C4–C6. The constructs were loaded first in compression nondestructively with 100 N and then to failure, and the results with the two techniques were compared. The Magerl technique was stiffer and had a higher load to failure ($p < .05$). The failure mode for the Magerl system was plate bending, while the failure mode for the Roy-Camille system was pullout of the caudal screw.

Clinical Studies

We will follow our usual patterns and sequence—the discussion of anterior procedures cephalocaudally and posterior procedures cephalocaudally.

Anterior Implants

The anterolateral bilateral screw fixation through the lateral masses was described on page 542. The anterior stabilization of a fractured dens as described by Böhler[17] has been presented on page 610.

Grossman and Selligson[106] reported on 13 patients treated with anterior vertebral body screw plate fixation (Fig. 8-37). There was good alignment and fusion rate. The challenges of attaining proper plate position and screw depth were discussed by the authors, who considered this a demanding method to be reserved for special indications.

Morscher and co-workers[202] developed a titanium anterior plate and screw system designed with a set screw in order to provide a stronger plate–screw interface and eliminate the need for the screw to go into the posterior cortex of the vertebral body. The screw length is too short to allow it to go through the posterior cortex of the vertebral body. The screws are perforated to permit bony ingrowth and consequently an increase in pullout strength and a decrease in loosening (Fig. 8-57).

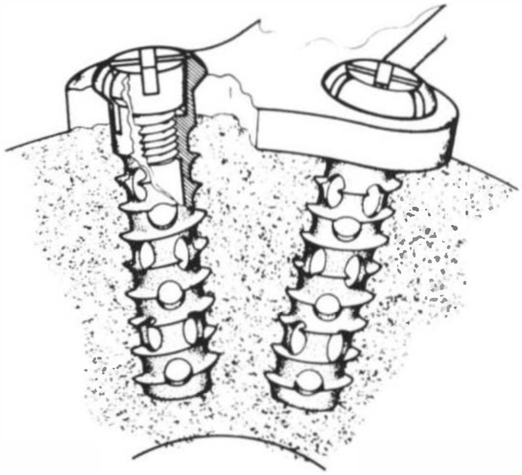

FIGURE 8-57 This anterior cervical plate has several cogent characteristics. It is made of titanium and has perforations. This allows for bony ingrowth and a greater possibility of bending with the bone. The screws have a blunt tip and a proximal inset locking set screw (see cutaway). The screw is designed to eliminate the need to penetrate the posterior cortex of the vertebra, and the set screw increases stiffness and thus reduces the problem of loosening of the screw–plate juncture. *(From Morscher, E., Sutter, F., Jenny, H., and Olerud, S.: Die vordere verplattung der halswirbelsäule mit dem hohlschraubenplattensystem aus titanium. Chirurg, 57:702, 1986.)*

Anterior cervical spine plates have also been designed by Louis, Roy-Camille, Fuentes, Benezech, Caspar, and others.*

Ono and co-workers[224] have devised a ceramic prosthesis for replacement of the vertebral body following extirpation of tumor. The prosthesis has the same design as the Ono prosthesis and also is intended to be used in conjunction with polymethylmethacrylate.

Posterior Implants

Itoh and associates have described a technique in which wires are used to attach a contoured rod to the occiput and the cervical spine using a sublaminar wiring technique in order to achieve an occipital cervical fusion.[141] The group reported successful treatment of 13 patients. No external postoperative immobilization was required.

An implant for posterior C1–C2 arthrodesis has been used by Roosen and colleagues[244] in three patients. The implant is used in pairs and with bone graft. The clamp puts the bone graft in the interlamina space under compressive load and provides enough stability to avoid the need for external immobilization.

Mitsui[196] has designed a compression clamp for posterior arthrodesis of C1–C2.

Holness and associates[135] have presented their experience with the Halifax interlamina clamp in a report of 51 patients. This clamp is designed for use at C1–C2 as well as other levels of the cervical spine. Only one clamp is used at each level to be fused. Bone grafting was not performed, and the patients were placed in a Philadelphia collar for 3 months. The authors report arthrodesis anteriorly

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*Various combinations of these implants are available and usable in different countries. We have provided cogent references when available; however, technical information on these products can best be obtained through local surgical implant distributions.

and/or posteriorly in all patients followed for 4 or more years. There was no instability or infection. Clamps slipped in two patients, both of whom underwent reoperation to replace the clamps. The surgeons considered this procedure safer than those involving passing wires under the lamina and quicker than those utilizing interlamina and interspinous wiring.

A clinical report of 11 patients with pathologic fractures of the cervical spine has been presented by Fidler.[78] The patients were treated with posterior fusions, using posterior instrumentation with plates and laminar wires supplemented with cement. Four of these patients had occipital cervical fusions with an implant that screws into the midline of the occiput and attaches to the cervical spine with laminar wires. This approach was successful in relieving pain and maintaining stability. There were no complications.

Cooper and associates[54] published a report of a series of 19 patients with posterior stabilization of the cervical spine using Roy-Camille plates screwed to the articular masses. The mean followup was only 9.2 months; however, at that point there was one failure in a patient with ankylosing spondylitis. The screws pulled out. The authors, perhaps prematurely, concluded that posterior plating is superior to techniques using wire. The study is nonetheless useful and important in that it is one of the first, if not the first, reports of cervical spine posterior plate screw fixation in North America.

Murphy and associates[204] have improvised with the Harrington compression rod and the technique of facet wiring to develop an implant system for immediate stabilization of the cervical spine. The surgeons used a large Harrington compression rod bent to a "U" shape and contoured as needed to control kyphosis/lordosis. They also used a small Harrington compression rod with modified hooks designed to rest on the lamina. The nuts can be positioned in relation to the hooks so that turning the nuts will apply either compression or distraction, depending on their position relative to the hooks. For ultimate fixation and stability, the Harrington compression rod, large or small, was secured in position by a wire through the facet joints using the Southwick-Robinson technique shown in Figure 8-40. Roy-Camille plates (and no hook plates) were also used, and case reports are included in the publication. Occipital cervical plates designed to be screwed onto the lateral masses and the occiput have been designed by Roy-Camille and also by Vlahovitch and Fuentes[304] with Benezech.

Comments and Recommendations

There is a distinct paucity of referred publications that describe experimental or clinical studies of cervical spine implants. The clear message, then, is to proceed conservatively with caution.

Probably the most pressing need in the field of cervical spine surgery implantation is for a good method of occiput–cervical fixation. We have described some early developments; however, this area of spine surgery is in its infancy. We're not "waiting for Godot" but for ingenious designs, laboratory tests, and clinical studies to meet this challenge. An anterior cervical plate that does not need to penetrate the region of the spinal cord, that has an ability to withstand the expected physiologic loads in the region of the plate–screw interface, and that includes a healthy fatigue life expectancy would be a useful addition to the surgical armamentarium. Experimental and clinical data on such a device may be forthcoming.

In view of the published complications by Nordt and Stauffer[216] and general knowledge, it is appealing to be able to effectively immobilize C1–C2 with a clamp and ovoid passing wires into the cervical canal.[135] The clinical series by Holness and colleagues shows that C1–C2 can be fused with few complications and no wires in the cervical canal. We view this as a promising addition to cervical spine surgical implants. Biomechanical tests to compare the clamp with wire constructs would be useful.

Since the rate of union in most cervical spine series is in the 85–90% range and the present wiring, bone grafting, external immobilization methodologies work satisfactorily, our basic suggestion is that we allow time for some experimental and clinical data to accrue before making major changes in traditional practices.

SYNOPSIS OF SPINE IMPLANTS: C0–C7

The display on pages 610–613, "Generalizations on Implant Prototypes," summarizes the advantages and disadvantages of selected cervical spine implants from the perspective of clinical biomechanics.

THORACIC, LUMBAR, AND LUMBOSACRAL SPINE IMPLANTS

Experimental Studies

At the time of this writing there is a paucity of published experimen*tal* studies of anterior fixation of the thoracic and lumbar spine. Anterior implants employed in conjunction with cement have been presented in a preceding section of this chapter (p. 576). Several devices have been described. Some of them will be presented in the clinical section that follows. There have been few experimental studies of anterior instrumentation devices in this region of the spine. Rezaian and associates[241] completed biomechanical and clinical studies on a device designed to replace a broken vertebra. The device has two parts; one is a turnbuckle distractor/spacer and the other component is a staple that can be used as a compressor/stabilizer. The goal is to allow immediate ambulation. Biomechanical studies using cadavers loaded to failure show the construct to be better than the intact spine in terms of load-bearing capabilities. The investigators report experience in the treatment of 22 patients with thoracolumbar fractures.

In contrast, there has been considerable testing of posterior fixation devices in the thoracic and lumbar region. A review of what we consider the most cogent of those studies, along with our views about their clinical and biomechanical implications, follows.

We will begin with the work of Fidler[78] because it provides a broad overview. Moreover, some of the findings from this relatively simple, basic model have been confirmed by subsequent animal and human spine surgical constructions. This work is recommended for the reader interested in experimental design as well as the key challenges and issues involved with studies of spine implants. The investigator created appropriately designed plastic models of the spine that represented very well a "standardized" spine. The great advantage of this methodology is that the individual variation in anatomic and material properties of human cadaver spines is eliminated, thus improving the controls in the experimental designs. This model was used to compare 14 different patterns of posterior instrumentation strategies to fit a grossly anteriorly destabilized spine model. The destabilization was removal of a vertebral body to simulate a pathologic fracture. The surgical constructions were then tested in compression to 3 mm and in axial torsion to 27°.

The study showed that of those constructions tested (Fig. 8-58), the most rigid was a well-fitted steel rectangle fastened to the vertebrae with laminar wires (see Fig. 8-59). The addition of bone cement significantly improved rigidity. This improvement was almost as great when the cement was placed in just one side as when it was placed on both sides. This is clinically relevant in that it suggests that one side can be left for bone grafting that, if

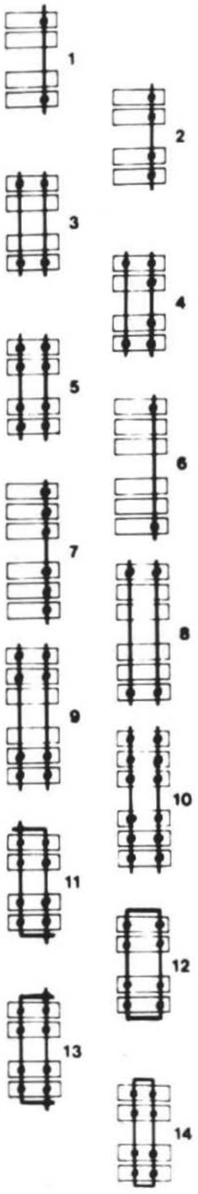


FIGURE 8-58 Diagrams of the 14 types of posterior instrumentation. The dots indicate the points of attachment of the instrumentation to the laminae. Constructions 1–10 are based on Harrington distraction rods, and the end dots thus represent hooks. The intervening dots in constructions 1–10 and all the dots in constructions 11–14 indicate laminar wires. Construction 11 = bilateral Luque "L" rods; 12 = broad welded rectangle; 13 = rectangle made from "C" rod and straight rod; 14 = narrow welded rectangle. (From Fidler, M. W.: Posterior instrumentation of the spine. An experimental comparison of various possible techniques. Spine, 11:367, 1986.)

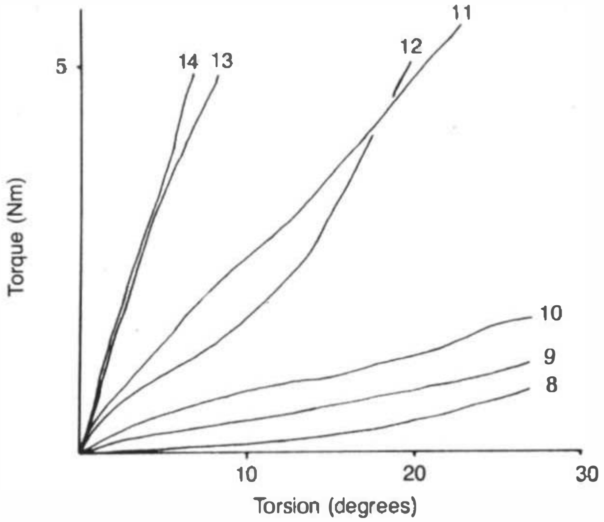


FIGURE 8-59 Load–deformation curves for seven types of posterior construction without additional cement. The numbers on the curves correspond to those in Figure 8-58. 8 = bilateral Harrington rods; 9 = bilateral Harrington rods and four laminar wires; 10 = bilateral Harrington rods and eight laminar wires; 11 = Luque "L" rods and laminar wires; 12 = broad welded rectangle and laminar wires; 13 = "C" rod–straight rod rectangle and laminar wires; 14 = narrow welded rectangle and laminar wires. *(From Fidler, M. W.: Posterior instrumentation of the spine. An experimental comparison of various possible techniques. Spine, 11:367, 1986.)*

successful, would provide long-term stability. Goel and colleagues studied cadaver L4–L5 posteriorly decompressed FSUs fixed with Luque rectangles wired to a remaining portion of the laminae of L4 and L5. This construction reduced motion at the decompressed level by 35%.[99]

There are some additional generalizations that seem tenable based on the Fidler study. Harrington distraction rods generally did not provide the same degree of rigidity as was observed with (1) the laminar wires combined with double Luque rods, (2) the C-rod with straight rod, or (3) the welded rectangle.

Double Harrington distraction rods generally provided a more rigid construction than a single one. The use of double or single Harrington rods fixing to two vertebrae above and below was better than one, and the use of three was better than two. Additional fixation in the case of Harrington distraction rods was with lamina wires. The more fixation points to vertebrae up or down, right or left, the more rigid the fixation. This study has its limitations; however, there are some useful generalizations that are well supported by its findings. Obviously, the clinical choice of instrumentation cannot be based solely on these or any other laboratory studies. Considerations unrelated to any limitations of this study are: (1) the passage of wires carries significant risks, and (2) we do not yet know the ideal rigidity that is required for spine arthrodesis.

Panjabi and associates, using fresh human cadaveric T9–L3 thoracolumbar specimens, studied the comparative multidirectional stability provided by eight long devices spanning five vertebrae to an injured specimen.[226] The study provided, for the first time, translatory motions at the site of injury. The results were presented for each device as a percentage change in the range of motion of the construct (injured specimen and the device) as compared to the range of motion of the intact spine. The Luque rods and the Luque rectangle provided the most stability in flexion, extension, and lateral bending, but none of the devices was able to restore stability to the values of the intact spine in axial rotation. The Harrington distraction/compression combination and the Harrington distraction with sleeves were the next most stable devices. The least stable devices were those which spanned three vertebrae: Luque short rectangle and the Dunn device.

Although the short device may have a biomechanical disadvantage as compared to the long device, there are some clinically valid reasons for the use of short devices. With this in mind, six short devices were compared for their multidirectional stability.

Abumi and associates, using five vertebral (T11–L3) fresh human cadaveric thoracolumbar specimens, first produced injury at the T12–L1 level by complete transection of the posterior elements, posterior longitudinal ligament, and posterior half of the disc, leaving only the anterior half of the disc intact.[1] Using pure bending and torsional moments to produce the motion and the stereophotogrammetric technique to measure the motion, they determined the translatory and rotatory multidirectional stabilities at the fracture site for six different short devices. Under physiologic flexion load, they found the most stable device to be the external spinal skeletal fixator (ESSF), followed by Harrington rod with reverse ratchets, Luque rectangle, and Kaneda device. Extension loads produced similar results, except that the Harrington and Luque exchanged places. The Kaneda device, being laterally placed, provided the most stability in lateral bending, but the ESSF was not far behind. The most difficult direction to stabilize was the axial rotation. None of the devices restored stability to the intact spine values. However, the most stable device, not surprisingly, was again the ESSF. A general finding of this study was that it is comparatively easier to restore rotatory stability than translatory stability at the site of injury for practically all types of physiologic loads. In other words, the injured spine may feel stable when considering the rotatory motions, but it may allow translatory motions in the transverse plane. The latter motions may impede fracture healing and fusion by allowing micromovement between the bony fragments. Thus, it is suggested that for comprehensive testing of spinal implants, the stability determinations may be made for rotatory and translatory motions at the fracture site.

Fatigue tests of spinal instrumentation are relatively rare. Nasca and associates[207] compared Harrington distraction rod constructions with Luque rod constructions in 12 swine spines. A pneumatic testing device was used, and the tests involved 10,000 displacement cycles of 2.54 cm of axial displacement. The force required to produce the 2.54 cm displacement was measured for the uninstrumented spine, Harrington construction, and Luque construction. The results were found to be respectively about 500 N, 800 N, and 1000 N. Thus, the Harrington rod construct was superior to the Luque rod construct. This is an interesting contrast to the previous experiment of axial and torsional rigidity in which the Luque system was better but in different test conditions. The two studies are very different; nevertheless, the importance of comprehensive testing to include fatigue tests is emphasized. The clinician must weigh the relative advantage of flexibility superiority versus fatigue superiority as well as laminar wiring risks versus lamina hook placement risks. We will discuss further the laminar wire instrumentation. Since we don't know the ideal rigidity, and since we do know that laminar wire risks are significant, we suggest that the best choice currently is to avoid the passage of laminar wires unless there is already complete loss of neurologic function or some compelling reason to use them.

The Harrington rods proved superior in the previously reported fatigue tests. It is interesting to note where these rods fail when they reach their fatigue tolerance. Cook and associates[52] completed biomechanical and metallurgical tests to evaluate eight clinically failed rods. All but one of the rods failed at the ratchet–shaft junction, as would be expected because of stress concentration. The one shaft failure was associated with a defect in the shaft. It is significant that five of the eight failed rods were associated with pseudarthrosis. One can speculate that there was probably a high elasticity fusion mass associated with the other three failures. It was suggested that choosing a rod that will place the ratchet–shaft juncture as close as is feasible to the hook and using the shortest feasible rod are both strategies that should reduce fatigue fracture of the rods.

Wittenberg and associates[329] recently completed a comparative fatigue study of four implants in the calf lumbar spine using a complete burst fracture model. The implant systems tested were as follows: (1) AO Fixateur Interne (with 5 mm Schanz screws); (2) Steffee plate (with 7 mm screws); (3) Harrington distraction rods (with Drummond wires to spinous processes); and (4) Luque plates (with 6.5 mm cannulated screws). The specimens were fixed in an Instron testing machine with a spring mounted posteriorly to simulate the erector spinae muscles. An axial load was applied to simulate the physiologic preload. Flexion compression loading was then applied both to determine stiffness and then cyclically at 3 Hz up to 100,000 cycles as tolerated.

The Ao system was significantly less stiff than the Luque system ($p<0.05$). There were no other significant differences in the stiffness of the implants. The ideal rigidity of spinal implants is, of course, yet to be determined. The Harrington Drummond wire system and the Luque system both allowed significantly less ($p<0.05$) posterior distraction (actual mm of motion instead of strain ($\Delta d/d$)) than did the AO Fixateur Interne and the Steffee systems. The ideal or allowable distraction or strain for a spinal implant construct during the fusion process is of course yet to be determined.

Hanley and colleagues[116] completed biomechanical studies in which they were able to quantitatively demonstrate the stabilizing capacity of Harrington instrumentation. Using cadaver material, they showed that laminectomy reduced lumbar spine stiffness 37.7% with flexion, 46.8% with extension, and 19.4% with axial torsion loads. With application of Harrington rods, the stiffness of the laminectomized spine increased 4266% in flexion, 1481% in extension, and 773% in axial torsion. It is also important to note that in comparison with an *intact* spine, the instrumented spine increased 1567% in flexion, 693% in extension, and 150% in response to axial torsion loads. These data help put in perspective the observation that Harrington rods do not control axial rotation as well as some other devices. Again, how much control is needed? What is the ideal stiffness? Is 150% of normal good enough? What are the risks and costs involved in getting more torsional rigidity? We need to know as much as possible about just what biomechanical goals we should be striving to achieve. We need to understand something about the experimental basis on which devices are studied, compared, and contrasted. We need to be as certain as we can that what we are striving for and achieving biomechanically is definitely worth the risks, complications, and expenses that are incurred.

It is useful to note here that Asher and associates[7] showed that a rigid cross-linkage between rods was better than wire in its ability to increase axial torsional resistance. Both wires and rods as cross-links were helpful, but the rods were more helpful. These findings were not based on fatigue tests.

Jacobs and associates have completed an extensive and useful experimental study of several types of spinal implants.[143] This extensive study utilized 40 cadaver spines and examined three simulated defects: (1) posterior ligamentous injury, (2) vertebral body injury, and (3) both injuries combined. These were tested in flexion, and the stabilizing effects of the following systems were compared: (1) Weiss springs, (2) Roy-Camille plates, (3) vertebral body plates, (4) Harrington compression rods, and (5) Harrington distraction rods.

For posterior defects, the Harrington compression rods provided stability similar to that of the intact spine. For anterior defects, the Harrington distraction rod spanning three vertebrae above and below the injured vertebrae gave stability comparable to that of the intact spine. With combined defects, the long distraction system was the most stable system, but it did not achieve the same local tolerance that was achieved with the other single defects.

As can be seen from some of the previous work, the number of levels to be included in the fixation and/or fusion above and below the defect is an important issue. This is because of the importance of purchase or anchoring for increasing rigidity in the construction. For example, a three-vertebrae fusion by Luque rectangle requires attaching to one vertebra on one side and to two on the other side of the defect. One-vertebra attachment by wire provides inadequate fixation (between the rod and the vertebra) to counteract the physiologic moments applied to the spine. Two vertebrae above and below the defect solve this problem. This argument is supported by the study presented earlier, where significant differences were found in the short versus long Luque rectangles, to the advantage of the longer device.[226]

The overall clinical biomechanical considerations here are as follows. The longer fusions apply more stress to the adjacent unfused segments. The clinical liabilities have been summarized on page 564. Also, longer fusion masses have a greater potential for pseudarthrosis and require more surgery. On the other side there must be enough of a fusion mass and/or extent of fixation to achieve adequate short- and long-term stability.

The next two studies are significant in regard to these questions. Purcell and associates[236] in biomechanical studies showed that hook pullout from the lamina could be prevented by implanting the Harrington hook three levels above the defect and two levels below it. To be more explicit, we are instructed to span two normal laminae and place the hook under the third (lamina or pedicle). Below we are advised to span one normal lamina and place the hook under the second one.

Based on the need for purchase, rigidity, and better fixation in the early postoperative period, the "fix long, fuse short" concept emerged. The fixation for stability as suggested above is employed after passing two normal vertebrae. After a period adequate for healing of the fusion, the rod is removed. A study by Kahanovitz and colleagues[148] suggests that this practice may prove hazardous to the temporarily immobilized unarthrodesed facet joints. This group reported that canine facet joints immobilized for 2 months showed chondrolysis, cloning, invasion of the tidemark, and loss of proteoglycans. These significant degenerative changes in these joints were not reversible 1.5–3 months following release from fixation. Moreover, Urban and coworkers showed that the immobilization of normal discs in animals caused water loss and reduction of metabolic activity within the disc.[299] A liberal extrapolation of these findings to the clinical setting portends that a temporarily immobilized FSU will undergo degenerative changes. These temporarily immobilized joints are particularly vulnerable because they are adjacent to a fused segment and are exposed to the sequelae of increased loads and motion.

The problem of hook pullout and the need for longer fixation to gain rigid fixation can be addressed to some extent by the more rigid pedicle screw and rigid plate fixation system or the Wisconsin wiring system.

Intrapedicular and transpedicular fixation of the spine was first developed and used by Professor Roy-Camille of France in 1963. Several implant devices have been designed to utilize the vertebral pedicles for fixation. The major ones will be discussed in this experimental section and in the clinical section. As suggested above, the pedicle fixation devices have an advantage in that they do not necessarily require the involvement of more than one vertebra above and below the level of the defect. The problems of hook and rod dislocation are eliminated. The pedicular fixation system eliminates the rather significant problems of laminar wire fixation.

Krag[163] completed some experimental studies that nicely support the rationale for the development of the Vermont Spinal Fixator (VSF). Although some experimental data are presented, the major portion of this work constitutes an exhaustive review of the literature on spinal implants. The authors also present new experimental data on vertebral dimensions. The information supports the conclusion that, anatomically speaking, there is ample room for safe pedicle fixation. They showed that an increase in depth of penetration from 50–80% is accompanied by a 30% gain in screw placement strength.

Skinner and associates[267] studied pullout strength, displacement before failure, and energy absorption before failure in several different pedicle screws in human cadaver pedicles. Those studied were ranked overall as follows: (1) Steffee, (2) 6.5-mm AO, (3) 5-mm AO, (4) Howmedica, and (5) Roy-Camille. An increase in major diameter was associated with greater pullout strength. As an interesting aside, the investigators noted that placement of the transpedicular screw into the vertebral end-plate significantly increased the pullout strength. They also observed that screws that penetrated the anterior cortex of the vertebral body had no greater pullout strength than those placed into the vertebral body. Although we consider pullout strength important in the clinical setting, it's the bending moments applied to the screw and its metal attachment that are more crucial.

For morphometric analysis of the thoracic and lumbar pedicles, we recommend the work of Zindrick and associates,[336] who completed 2,905 pedicle measurements. These include measurements important to the surgical implantation of pedicle screws.

In a recent study, Panjabi and co-workers determined the capacity of five pedicle screw systems and the facet screw fixation to provide multidirectional stability to an unstable L5–S1 joint.[230] Using fresh human cadaveric L2–sacrum specimens, the instability was produced at L5–S1 by transecting all posterior ligamentous elements and drilling a transverse hole through the disc. There was an average 50% loss of stability in various directions. Facet screw fixation, together with AO, ACE, C-D, Steffee, and Wiltse systems, were evaluated.

All devices provided more than adequate stability in flexion. The most stable device was the Wiltse double-rod system, followed by the Wiltse single-rod system and the Steffee plate. In extension, stability of facet fusion was less than that of the intact spine, while stability in all the devices was either equal to or more stable than that of the intact spine. Lateral bending was well controlled by all devices, the most effective being the Steffee plate, the ACE system, and the Wiltse system. Finally, the Steffee plate, the ACE with transverse wire, the C-D, and, surprisingly, facet fusion fixation were among the most stable constructs.

Magerl[183] tested the external spinal skeletal fixation (ESSF) system (pedicle screw fixation secured by an external clamp device) in cadaver spine constructions and compared it with (1) dorsal plate fixation, (2) the Jacobs distraction system (modified Harrington), and (3) the conventional Harrington distraction system. The ESSF system showed the greatest stiffness and stability (Fig. 8-60).

Olerud and colleagues[221] have reported a clinical series of external transpedicle spinal fixation performed for quite different purposes. The authors used Schanz screws in the pedicles attached to an external modified Hoffman fixation device to distract and immobilize displaced (unstable) lumbar and lumbosacral spine segments. This was employed as a clinical test to determine whether or not the patient's back and/or leg pain would be relieved. If so, the subluxation instability could be diagnosed as the cause of the pain. A spine fusion of the FSU would then be expected to relieve the pain. Eighteen patients were immobilized with this technique, and all but one were relieved of low back pain and often

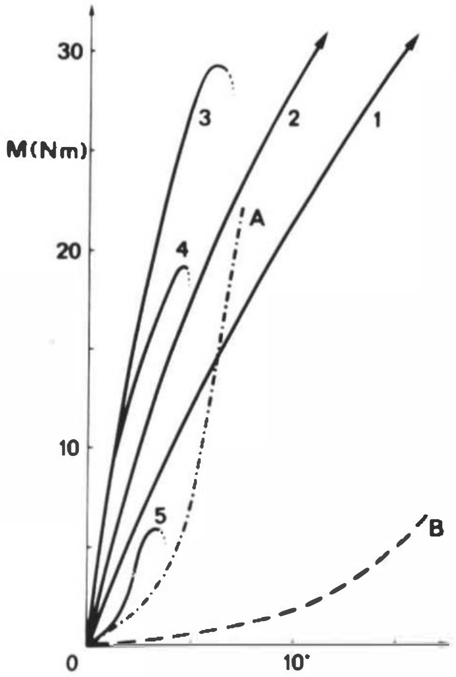


FIGURE 8-60 Stiffness and stability of the following fixation systems: ESSF (1) without and (2) with screw fixation of facet joints; (3) dorsal plate fixation; (4) Jacobs distribution system; and (5) conventional Harrington distraction system. The fracture model had a vertebral body defect and dissected posterior ligaments. Plot *A* describes the mechanical behavior of the intact spine and plot *B* the behavior of the dissected fracture model. The uninstrumented and instrumented cadaveric spines were exposed to a pure fixation moment (M), shown on the y-axis. The x-axis represents the degree of deformation in flexion. In (4) and (5), the cranial hooks disengaged, and in (3) the anchorage of end screws failed at the indicated flexion moments. ESSF remained stable over the entire range of 40 Nm, which was applied as a flexion moment (see text). *(Modified from Schlapfer, F., Worsdorfer, O., Magerl, F., and Perren, S. M.: Stabilization of the lower thoracic and lumbar spine: comparative in vitro investigation of an external skeletal and various internal fixation devices. In Uhthoff, H. K. (ed.): Current Concepts of External Fixation of Fractures, p. 367. Berlin, Heidelberg, Springer-Verlag, 1982.)*

associated leg pain. There were no complications. In addition to the contributions of this work to the body of knowledge on spine instrumentation, it is a milestone contribution to clinical knowledge and practice in regard to the management of patients with clinical instability and low back pain.

Zindrick and colleagues[337] studied the biomechanical aspects of a lumbosacral intrapedicular screw fixation. They noted in pullout strength studies that osteoporosis was a significant factor. They found that large-diameter screws engaging the cortex of the anterior vertebral body provided the most secure fixation. For the sacral fixation, they noted that screws directed straight anteriorly into the ala were not as strong as those aimed laterally into the ala at 45° or medially into the first sacral pedicle. They noted further that PMMA was helpful in restoring fixation to loose screws and that pressurized cement doubled the pullout force.

In a recent study, Wittenberg and colleagues[329a] observed the effects of several important variables on pull-out strength of transpedicular screws. They observed the effects of screw design (6.25 mm Steffee screw, 5 mm and 6 mm Schanz screws, and 5 mm and 6 mm Kluger screws). The effects of creating the pedicle hole with a probe vs. a drill was observed. And, perhaps most important, the effects of bone mineral equivalent densities as measured by quantitative computed tomography (QCT) were also observed.

The Steffee screws (deep thread, small core diameter) and the Kluger screw (shallow thread, large core diameter) did not show any significant pull-out stength from human vertebrae of similar mean equivalent density. However, in the calf bone with a much higher mean equivalent density the strength of the fixation in the bone for both pull-out and vertical loading was statistically significantly stronger than with the Kluger screws. This implies that the equivalent numerical density of the bone may be more important than the screw design. This has very important clinical implications. Certainly it will be useful to take into consideration a quantitative analysis of the quality of the bone upon which we depend for our pedicle fixation.

There was no significant difference in pull-out strength based on whether or not a probe or drill was used to make the entryway for the pedicle screw.

Kornblatt and co-workers[161] studied four lumbosacral fixation systems to determine their relative stiffness and strength in flexion loading. The systems studied were: (1) translamina facet joint screws, (2) Luque rectangle box, (3) Luque fixation to the pelvis through the Galveston technique, and (4) two-part pelvic spinal rod system. The stiffness of the four systems was compared to that of the normal intact spine with and without posterior ligament disruption. All four systems had similar strength. The facet screws were 20% stiffer and the pelvic spinal system was 2.3 times stiffer than the two Luque systems.

Keene and associates[154] compared loads to failure of the Harrington compression system and the Wisconsin system. The latter is a design with posteriorly slotted hooks to allow relative ease of application. The Harrington rods were 14% stronger; however, both systems failed at low loads because of failure of the bone. The Wisconsin required 2.25 times as long to fail, and it absorbed twice as much energy before failure. The ease of application of the Wisconsin system is reflected in the 50% reduction in operating time noted in a series of 30 patients.

Shaw and associates[262] designed biomechanical investigations of cadavers to evaluate the stabilizing effects of fixation techniques on simulated pelvic fractures. Malgaigne-type fractures with sacroiliac disruptions were created in four pelves. They were fixed first with either Slatis- or Pittsburgh-type anterior fixation systems. These constructions were tested in longitudinal and torsional loading. Then, posterior threaded compression rods connecting the posterosuperior iliac spines were applied, and the specimens were retested. Prior to the application of the posterior compressions screws, the construction was unstable, but following that the stabilization was improved.

Ashman and associates recently conducted a comprehensive study of the failure characteristics of several different pedicle-screw spinal fixation systems.[8a] They measured axial and torsional stabilities, stresses on the implant body and pedicle screws, and fatigue failures and theoretically computed stresses at the sites of failure on the screws.

For the evaluation of stability characteristics they used fresh cadaveric human thoracolumbar specimens. To simulate a burst fracture, they removed entire L1 vertebral body and transected the pedicles, thus destroying both the anterior and middle columns. They measured the axial translation of the upper vertebra when it was subjected to 450 N of compressive force, blocking other degrees of freedom. In addition, they conducted a torsion test in which a rotation was imparted to the top vertebra and torque developed and motions produced at the site of injury were measured. Stresses were measured in another study at the points of expected high stresses, both in the body of each implant and in the screw. Finally fatigue tests were conducted using plastic vertebrae in a worse case scenario in which the middle vertebra was completely missing. In these tests completely reversible loads were applied at a rate of 2 cycles per second. Only one load magnitude of ± 450 N was used. Under these experimental conditions the Steffee plate system failed in less than 5000 cycles, while the Zielke system lasted for about 300,000 cycles. Next were the AO fixateur interne, Luque plate, and AO notched plate systems, which each surpassed 1,000,000 cycles. The authors point out that even though the three systems mentioned exceeded one million cycles of loading, it does not mean that they would not fail in the clinical situation. The *in vivo* loads are presently not known and are probably higher than ±450 N, as clinical failures of some of these devices have been observed.

Clinical Biomechanical Considerations

Slotted Compression Plate

Humphries, Hawk, and Berndt reported on the use of a compression plate for anterior lumbar spine surgery (Fig. 8-61).[138] They designed plates for arthrodesis of one or two FSUs. Compression was applied with a clamp and a slotted mechanism, which was then subsequently tightened. The clamp was held in place by two noncancellous bone screws. The disc was removed and the space was packed with cancellous bone chips. The patient was permitted out of bed with no orthosis when he so desired. The mechanical weakness of this construct is due to the fact that cancellous bone chips in the disc space cannot support the large loads of three to four times body weight. The plate screw system, with its purchase in the cancellous bone of the vertebral bodies, should not be expected to hold. The screws are likely

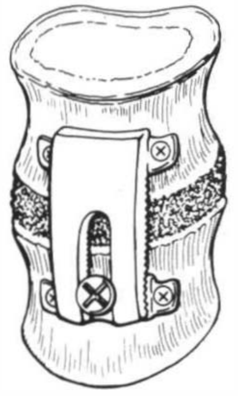


FIGURE 8-61 A slotted plate that is designed to fix and apply compression across an interbody fusion. The compressive force is applied and maintained by a clamp while the screws are inserted.

to pull or cut out of the cancellous bone of the vertebral body. Despite the above analysis, Humphries, Hawk, and Berndt reported no failures of the construct in 14 cases. It would seem desirable to study a larger series before using this construct (see also the Syracuse I-plate, p. 606).

The Prosthetic Lumbar Disc

Following discectomy, some patients will have either continued or new low back pain. This is sometimes associated with lumbar spondylosis involving disc space narrowing, facet arthropathy, and "instabilities." Many of these patients are treated with spine arthrodesis, which sometimes alleviates the pain. Spine fusion is, however, sometimes associated with a variety of undesirable sequelae (p. 564). A number of investigators have had the idea that a prosthetic disc could be developed. The purpose of such a prosthesis would be to maintain: disc space height (and the normal canal and nerve root foramenal space), "stability," normal motion, normal mechanical properties, and appropriate distribution of loads between the anterior and posterior elements.

We have elected to describe the "Carité disc endoprosthesis SB" because to our knowledge it is the one with which there is the greatest clinical experience.[37,338] The prosthesis consists of three parts: two central concave-molded titanium plates with spikes and a biconvex polyethylene "disc" as interspace filling material. There is a metallic wire ring around the polyethylene "disc" to serve as x-ray identification.

The reported clinical experience included 44 patients in whom 50 prosthetic discs were implanted through a left anterior retroperitoneal approach.[338] The prosthesis is inserted without cement, and the patient is up the following day. The indication was painful disc degeneration with or without instability and following conservative treatment. Apparently, back pain following discectomy that was unrelieved by conservative treatment was also an indication. Contraindications include: osteoporosis, spinal stenosis, severe spondylolisthesis, severe facet joint arthropathy, regional infections, uncooperative patients, and those with psychologic disturbances. Discography was done with the goal of identifying the painful disc.

Complications consisted of two deep-vein thromboses and one urinary tract infection. In two patients there was failure of the metal at the point of contact with the vertebral end-plate. The design was changed during the treatment of the initial series of patients.

Follow-up x-rays showed normal motion (Fig. 8-62). There was either absence of pain or a marked reduction in pain in 41 patients and persistence of pain in 3 patients. Patients return to work in 4 months on average. The follow-up at the time of the report was very short. Nevertheless, the procedure is worthy of careful attention and further study. Of course, it will be very important to focus on determining which patients are appropriate candidates for the procedure. Several years will be required to determine the best design, including a determination of the device that can best preserve or reestablish the normal kinematics and kinetics of the FSU as well as eliminate pain.

Laminar wires are effective spine immobilizers[78] but are not recommended because of their high risk/benefit ratio. The Harrington distraction rod system was superior to the Luque system in fatigue tests.[207] The Harrington distraction rods clinically fail at the ratchet–shaft junction. Surgical decisions that keep this junction as close as possible to the end of the rod decrease bending moments and therefore reduce the tendency for failure. Although the Harrington distraction system does not prevent axial rotation (y-axis) as well as other systems do, it increases axial torsional stiffness to 150% of normal.[116] Since we don't know how much is needed, this may or may not be enough.

The Harrington compression rods or distraction rods spanning three vertebrae above the defect provided stability comparable to that of the intact spine.[143] The placement of rods two or three levels above and below the defect provides better fixation and lessens the probability of hook dislodgement.[236] This surgical strategy, however, is associated with a longer fusion mass or probable initiation of degenerative changes[148] because of temporarily immobilizing facet joints. Consequently, the "rod long, fuse short" strategy carries a certain liability. A different instrumentation system (pedicular fixation) that allows one to immobilize and fuse only to the normal vertebrae above and below the defects may prove to be the solution to the "rod long, fuse short" dilemma. When pedicle screws are loose because of osteoporosis or some other reason, PMMA is helpful in improving the strength of fixation.[337]

Translaminar facet screws (see p. 615) were shown to constitute a lumbosacral spine fixation construction that was stiffer than both the Luque

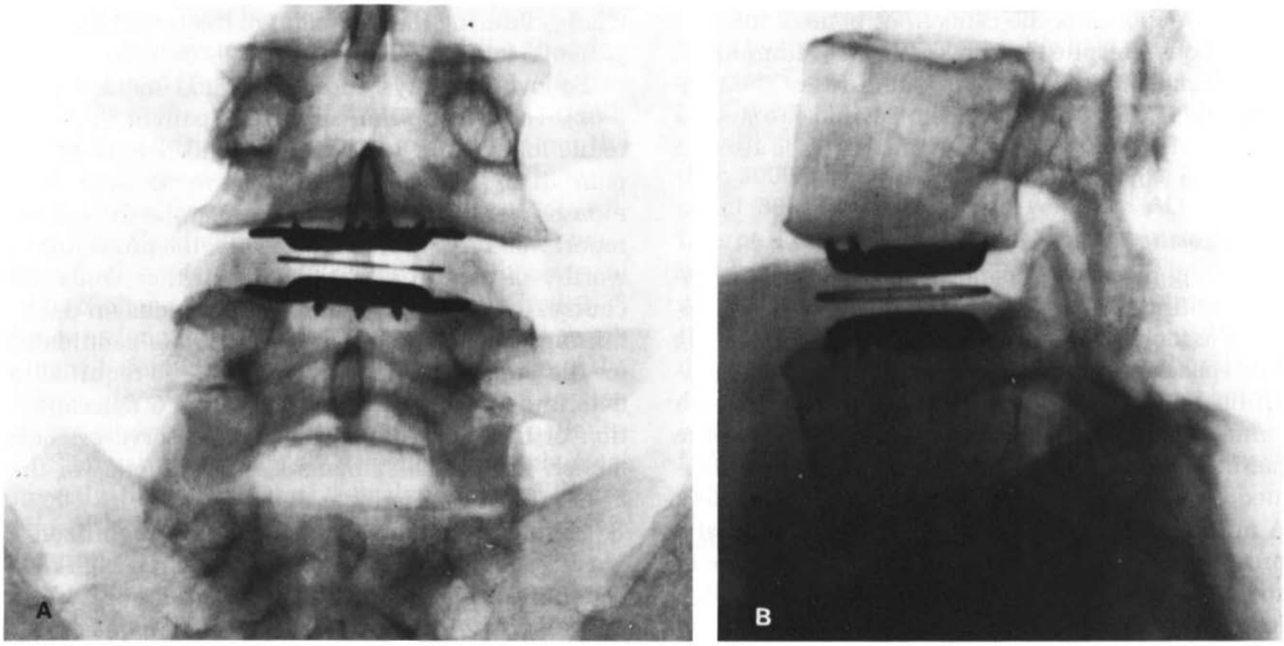


FIGURE 8-62 (**A**) Anteroposterior view of intervertebral disc prosthesis. This is basically composed of a plastic biconvex disc articulating with a concave metal component imbedded in each end-plate and/or subchondral bone. The metal discs can be seen, and the ring between them identifies the plastic component. (**B**) The components are shown here on a lateral radiograph. *(Courtesy of Professor H. Zippel.)*

rectangle box and the Luque fixation attached to the Galveston implant.[161]

This information, based on experimental studies, is best used in combination with directions indicated in some of the key clinical studies to be reported in the following section.

Gruca-Weiss Springs

This device was first developed by Gruca in 1956 for the correction of scoliosis.[109] In 1975, Weiss described its use for stabilization of the lumbar spine.[312] The instrument system consists of two heavy springs with hooks on both ends. The springs are under tension and are applied over the section of the diseased spine to be corrected or stabilized. The hooks are inserted in the transverse processes or the laminae at the top and bottom of the curve that is to be corrected. When treating kyphosis, they should be attached to the laminae. The two springs lie over the intervening laminae separated by the spinous processes. When scoliosis is being treated, the hooks are better placed over the transverse processes on the convex side of the curve to create a greater moment arm for correction of abnormal rotation in the frontal plane.

Figure 8-63A shows the combined forces applied to the spine by a pair of springs. The forces on the end vertebra are equal in magnitude to the tension in the two springs. Because of the curvature of the spine, the springs also apply small, radially directed forces on each of the vertebrae within the curve.

These forces produce two different kinds of bending moments on the spine. The two large forces produce bending moments that tend to correct the angular deformity, while the small radial forces have an opposite effect. The net effect is such that a modest correction occurs. In addition to the bending moments, a large compression is produced between the vertebrae.

Benzel and Larson[14] reported their experience with 90 patients in whom they were retrospectively able to compare the use of Harrington distraction rods with a modified Weiss spring technique in the treatment of post-traumatic thoracic spines. The Weiss spring modification involved connecting the pair of springs at two different points with Parham

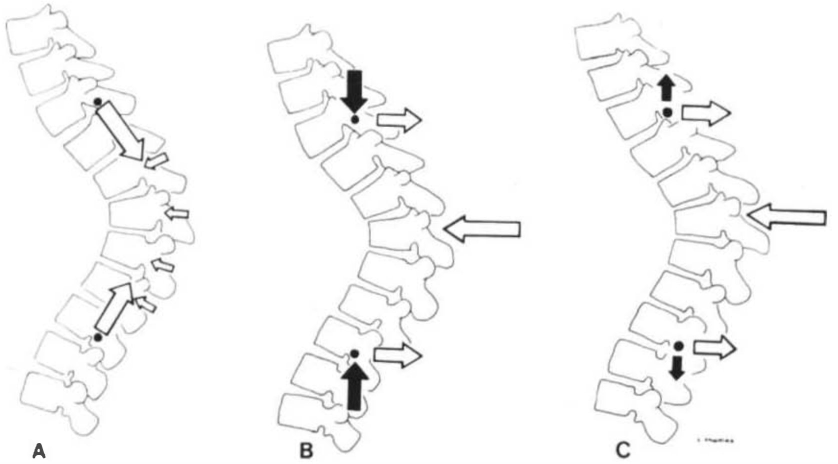


FIGURE 8-63 A simplified diagram showing the forces operating on the spine due to the three instrumentations. **(A)** Gruca-Weiss springs. The large arrows represent the forces equal to the tension in the springs, while the small arrows represent the pressure of the springs against the spine. **(B)** Harrington compression rods. The three light arrows represent the forces due to the three-point bending of the spine as the Harrington rod is inserted between the two hooks. The black arrows are the compressive forces applied by tightening the nuts. **(C)** Harrington distraction rods. The three-point bending forces are similar to those shown in **B** but larger in magnitude. The smaller black arrows represent the distraction forces, which are probably of lesser magnitude than the compression forces created by compression rods. *(White, A. A., Panjabi, M. M., and Thomas, C. L.: The clinical biomechanics of kyphotic deformities. Clin. Orthop., 128:8, 1977.)*

bands. Forty-seven patients were treated with Harrington distraction rods and 43 with Weiss springs. Eight (17%) of the Harrington instrumentations failed (*i.e.*, hook dislodgement, pain, excessive spine stiffness), and there was only one failure of the Weiss springs. The authors considered the Weiss spring instrumentation technique to be superior to the Harrington distraction rod instrumentation. The dynamic or nonrigid fixation afforded by the springs was purported to be the advantage. For this reason alone, we believe that this is a very important work. To our knowledge, it is the first to address with clinical data this highly cogent issue regarding the clinical biomechanics of the spine.

Harrington Compression Rod

This instrument system is designed to apply compression to the spine at the points where the hooks are inserted. Figure 8-63B shows the characteristic forces that are applied to the spine with these devices. As a result of the bending of the compression rods in the sagittal plane, there are modest transverse forces that tend to pull the hooks posteriorly and push the apex of the kyphotic deformity anteriorly.[317] This combination of forces (a pull at the ends and a push at the center) constitutes a three-point bending system. Because this system operates in the sagittal plane, it tends to correct the kyphotic deformity. The action of the compression force, however, is to *increase* the angulation of the deformity.

Harrington Distraction Rod

When the Harrington rod is used in the distraction mode, tension is applied to the spine in addition to the three-point bending previously described.[8] The tensile forces produce bending moments, so that the distraction rods are more efficient than the compression rods in correcting the deformity. These forces are illustrated in Figure 8-63C. It has been shown that a *combination* of axial distraction and transverse loading provides the most effective correction, regardless of the degree of the curve (see p. 138). This produces bending moments in addition to those produced by the three-point bending system. In other words, the angular correction is improved over that offered by the compression rod system. This, then, can be presumed to be useful instrumentation for correction of clinically stable kyphotic deformities. These considerations apply much more to post-traumatic conditions than to Scheuermann's kyphosis. In developmental kyphosis, the forces that must be resisted are significantly large. Bradford and colleagues operated on over 50 patients for this disease.[26] They found that the deformity was very resistant to correction. This is probably related to the observed hypertrophy of the anterior longitudinal ligaments, which increases the tensile resistance to the correctional forces. In addition, the basic deformity itself is considerably stiff. Bradford and colleagues found that these resting forces often required considerable bending of the distraction rod prior to correction. It is important to emphasize that in the correction of kyphosis the Harrington distraction rods *are not used primarily as distractors. They are used to produce bending* moments that apply correcting couples and then serve as internal splints. The distraction mechanism is then employed to lock the rods into place. The hook is fixed on the rod by the tension produced in the soft tissue, which locks the hook into the riveting mechanism of the rod. If there is gross disruption of the ligaments of an FSU, the distraction system may have certain limitations or even dangers. This may be an indication for the use of a combined system described on page 601.

Comparison of Clinical Biomechanics of the Three Systems

Although we do not know the quantity of forces applied in the three systems, a simple mechanical analysis generates some useful information. Measurements were made of the three-point bending stiffness of the compression and distraction Harrington rods.[317] The Harrington distraction rod was found to be 4.7 times as stiff as the compression rod of the same length. This implies that the three-point bending in the two procedures is approximately five times more effective with the Harrington distraction rod. The distraction rod has the additional advantage of producing a distraction force, which further increases its effectiveness by providing corrective bending moments. In summary, then, the distraction rods are clearly more efficient in correcting deformity, and the compression rods may be useful through their ability to provide some immediate clinical stability by means of impaction. We believe that the functioning of both of these instruments is most effective in the presence of an intact anterior longitudinal ligament.

The Gruca-Weiss spring has a different mode of action and is difficult to compare with the Harrington systems. We do not recommend the Gruca-Weiss spring for the corrective instrumentation of kyphotic deformity. Although this spring is not specifically designed to treat kyphosis, the experimental work of Jacobs and associates[143] supports our opinion (see p. 593).

Can these methods be applied without any consideration of the clinical stability of the spine? Bending moments in the sagittal plane, which tend to correct kyphotic deformity, produce tension in the anterior and compression in the posterior elements of the spine. Therefore, the efficiency of the surgical instrumentation in the correction of kyphosis is directly related to the ability of the anterior elements of the spine to withstand tensile loads. When the anterior elements are known to be disrupted, the Gruca-Weiss springs and the Harrington compression rods that are able to apply some anterior compressive force are relatively more attractive. This is due to the fact that the compressive forces can provide some stability. Additional details of this analysis are available.[317]

An experiment was performed by Stauffer and Neil on cadaver spines to study the relative stability provided by the three fixation procedures described above. The disruptive bending load applied to the spine to test its stability was a combination of flexion moment and axial torque. They found that Harrington compression rods provided the maximum stability of the three, followed by Harrington distraction rods and the Gruca-Weiss springs.[276] The compression rods provide stability through the vectors of force that impact or provide an element of compression between the upper and lower portions of the spine (see Fig. 8-63B). Similar studies by Meyer and colleagues confirmed this work.[197] These investigators also showed that the best instrument fixation of the disrupted spine sustained loads that were only 50% of the loads that the normal spine could bear.[197] We believe that when a clinically unstable spine is to be corrected, one or two Harrington compression rods should be considered. This view is well supported by the work of Jacobs and colleagues[143] described on page 593.

Combined Use of Harrington Distraction and Compression Rods

We have shown previously that the bending moments created by the distraction rod are the most effective in correction of the traumatized spine. It has also been shown that in situations in which clinical stability is a factor, especially with a nonfunctional anterior longitudinal ligament, the compression rods can be expected to contribute significantly to clinical stability through impaction at the disrupted spine segments. Therefore, it is reasonable that in situations in which strong correctional forces are needed in addition to compression for clinical stability, combined compression and distraction rods may be the treatment of choice (Fig. 8-64). Although one might be concerned about frontal plane rotation in this system, it is not significant for two reasons. First, the couple formed by the distraction and compression forces has a short lever arm, resulting in a small frontal plane bending moment. Second, this moment is adequately resisted by numerous anatomic constraints, particularly the spinous processes, which immediately buttress against the rod to prevent frontal plane rotation.

A Technique for Application of Harrington Rods in Kyphosis Correction

Probably the best instrumentation method for exerting correctional forces on a kyphotic deformity or burst fracture is transpedicular fixation with Schanz screws. Correctional forces can then be exerted with

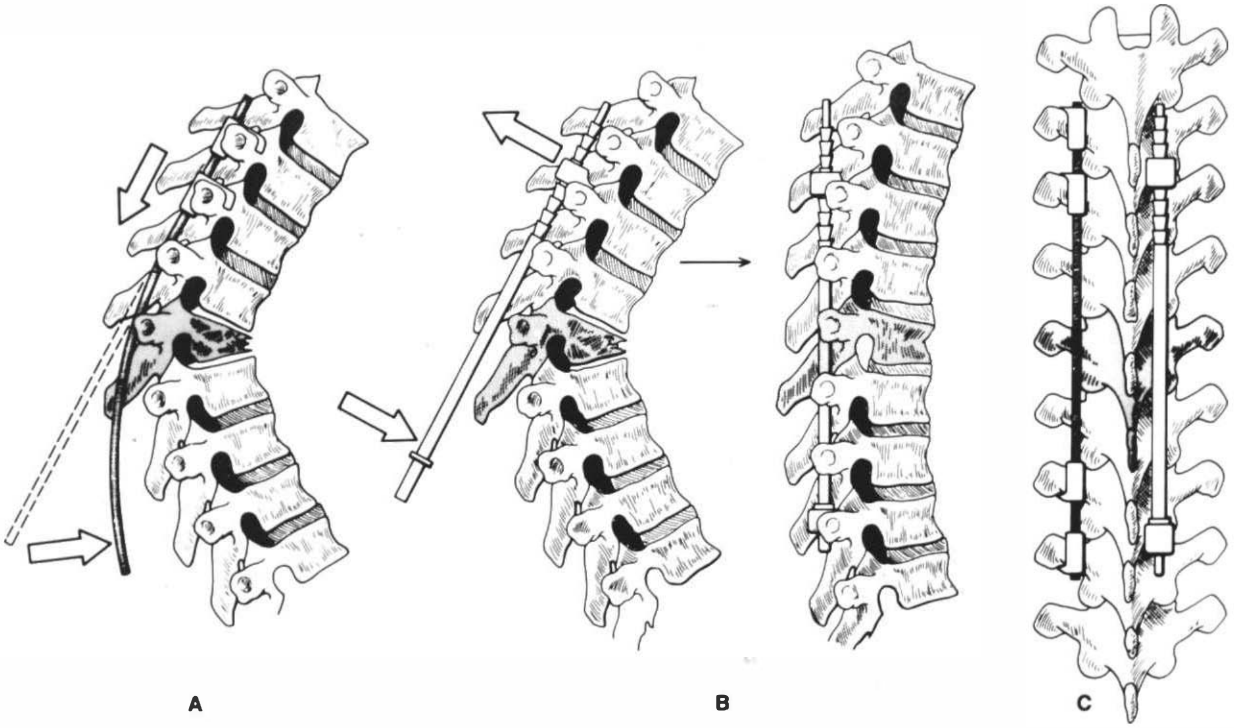


FIGURE 8-64 Rationale and biomechanics for the combined use of Harrington compression and distraction rods. (A) The compression rod has a relatively low stiffness and may not be helpful in correcting deformity. (B) The distraction rod is about five times as stiff as the compression rod. It applies a strong couple to the deformity and is likely to correct it. The rod is then attached and serves as a splint to maintain correction. It is not employed as a distractor. (C) The compression rod is then applied to stabilize the two parts of the kyphos in their corrected position. Frontal plane rotation is small because of the short distance between the two rods and is restricted by the buttressing of the spinous processes against the stiff Harrington distraction rod. Recently, implants that are better designed to correct kyphotic deformity have been developed (see p. 602).

the Dick fixateur interne or with the external fixator developed by Magerl.[183] The correctional capacity of the fixateur interne is shown in Figure 8-65.

The Hinge Principle in the Correction of Kyphosis

In the correction of a kyphotic deformity, when the angle between the two arms of the kyphos is increased by corrective displacement, a hinge of some sort is necessary. This was pointed out by O'Brien in the correction of tuberculous kyphosis.[217] His recommendation was that a portion of bone be left posteriorly as a hinge. The principle is analogous to aligning the position of a door in relation to the wall. If there is a *hinge*, it works beautifully. The door is closed and the 180° angle with the wall is readily achieved. If there is no hinge, the door simply pulls away from the wall. The two analogous situations are shown in Figure 8-66. The hinge does not have to be comprised of posterior bone. Other tissues, such as the intertransverse ligaments, the facet capsules, or the anterior longitudinal ligament, can also serve that function. It is suggested that the surgeon be aware of situations in which there is no hinge and that he or she take this important factor into consideration when correctional force vectors are applied. In addition to the correctional moments, there will be a need to apply some loads that will approximate and align the two limbs of the kyphotic deformity. Consequently, when there is no hinge, one should be provided by the appropriate instrumentation. We suggest a Harrington compression rod, wiring of the posterior elements, or Gruca-Weiss springs, in that order of preference, but depending upon the particular clinical problem.

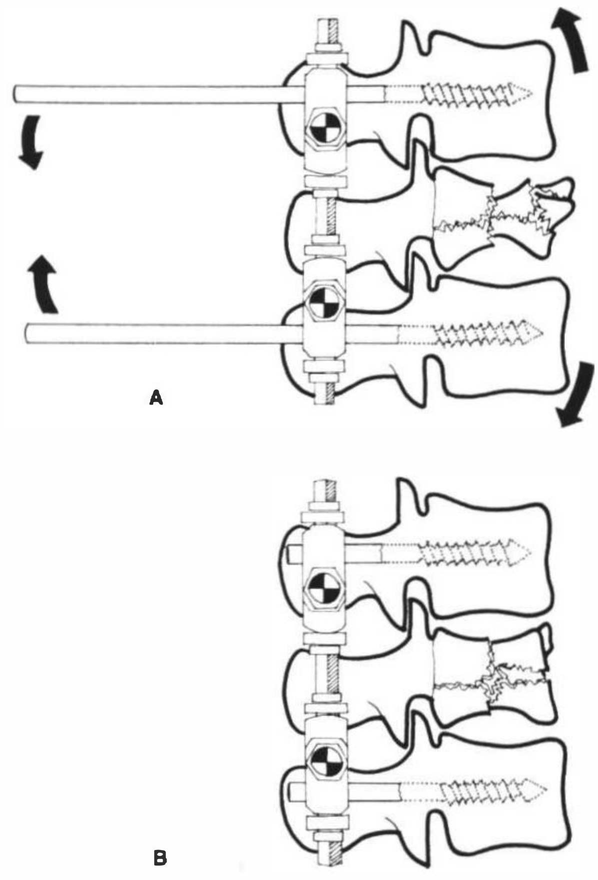


FIGURE 8-65 (A) The Schanz screws provide two strong correctional bending moments in the sagittal plane, if the two ends of the Schanz screws are approximated. The resulting bending moments are shown by arrows. (B) The corrected fracture is shown after the screws have been cut. A kyphotic deformity would be similarly corrected. (From *Dick, W.: Internal fixation of thoracic and lumbar spine fractures. Toronto. Hans Huber Publishers, 1989.)*

Scoliosis Instrumentation

The mechanics of the Harrington instrumentation for scoliosis has certain similarities to that of kyphosis. This is reviewed in Chapter 3, along with an analysis of the Dwyer instrumentation.

Pedicle Fixation

Several systems for intrapedicular fixation of the thoracic, thoracolumbar, lumbar, and lumbosacral spine are now in use. These include the AO Fixteur Interne, Steffee, Wiltse, Luque, Roy-Camille, Magerl, Louis, and others. At the time of this writing, there are relatively few completed publications of the documented clinical experience with these various implants.

Steffee and associates[279] have described the Steffee system and reported five patient examples. Other series have been presented at various meetings. Luque[180] reported 20 patients with a broad variety of diseases. These patients were treated with spine fusions using pedicle screws attached by wires to a contoured rod. There were no complications in the 20 patients. A review of what we consider the most cogent information available at the time of this writing follows.

A recent study by Bernhardt and colleagues[12a] raised some questions about transpedicular fixation. This study consisted of 47 patients who had lumbar

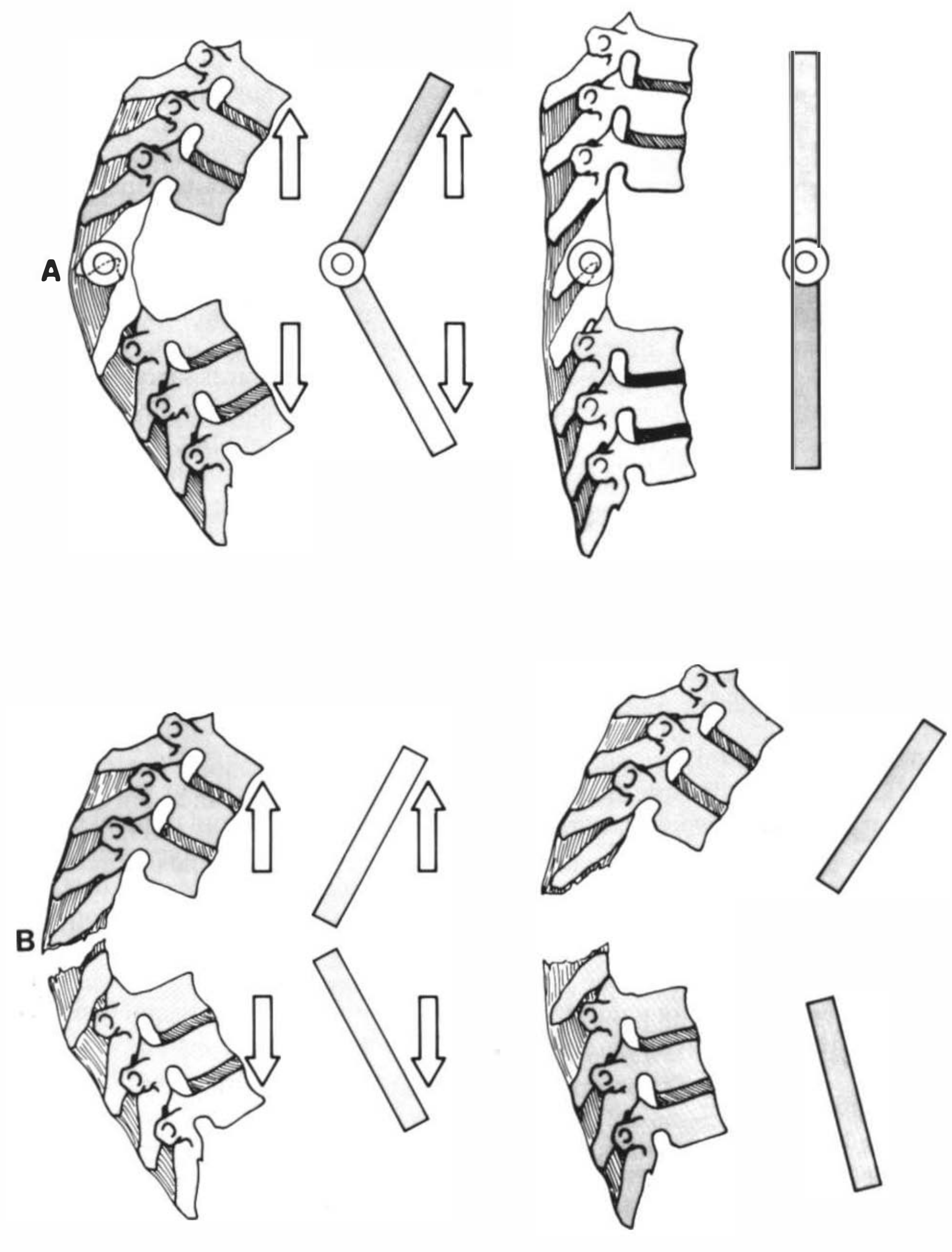


FIGURE 8-66 This illustrates the necessity for some type of hinge in order to correct a deformity. **(A)** The anterior longitudinal ligament, the intertransverse ligaments, bone, or the posterior ligamentous structures may serve as a hinge. For obvious reasons, the spinal cord should not be the hinge. **(B)** With no hinge, it is not possible with an axial force to change the angle between the limbs of the deformity. The two portions tend to separate.

or lumbosacral fusions with or without pedicle screw and plate fixation. Eighteen patients had pedicle screw plate fixation using the variable screw placement (VSP) system. There was a control group of 27 patients with the same surgeon and technique except that no internal fixation was employed. The rate of pseudarthrosis was essentially the same for the two groups—22% for the VSP group and 26% for the control group. The excellent results were 67% for the VSP group and 70% for the control group. Two of the instrumented patients had disabling postoperative leg dysesthesias. This complication was not observed in the control group. These data suggest that lumbar and lumbosacral spine fusion without transpedicular screw plate fixation is equally as effective as but safer than with transpedicular fixation. This was a retrospective study; however, its design was such that the issue of directionality was not considered to be a significant liability.

Magerl[183] described the external skeletal spinal fixation (ESSF) system for stabilization of the lower thoracic and lumbar spine and reported on 52 patients with a 1-year follow-up. Most had been treated for trauma and some for osteomyelitis. The results were considered encouraging, and there were no serious complications.

Posterior Lumbosacral Fixation Devices

Lumbosacral Spring Fixation

A construct has been described by Hastings and Reynolds in which a coiled spring with an outside diameter of 4.3 mm is employed in conjunction with a hook in lumbosacral spine fusions.[122] Small hooks that can be screwed into the coiled spring are inserted into or hooked over the lamina. The bone grafts lie between the spring and the host bed. The investigators present a chart that allows the determination of tension in the spring for various elongations. This careful measurement of forces applied to the body is a commendable principle. The tensed spring probably holds the bone graft in place effectively. However, we doubt that the tensile loads of 60–80 N (13–18 lbf) are very significant in fixing the FSUs against the large magnitude of forces (three to four times body weight) exerted at the lower lumbar spine.

We considered deleting the description of this spring in this second edition; however, it was included for an important reason. This device provides a dynamic, flexible, or low-rigidity fixation. This concept, we predict, will find its way to useful clinical application in future spine implant development.

Knodt Rods

These devices were designed for internal fixation of the spine, the theory being that it is somehow beneficial to fuse the FSU in some degree of flexion that opens the intervertebral foramina but also takes away lumbosacral lordosis. A report showing a 60% failure rate suggests that they are not particularly helpful.[66]

Selby[260] reported 92 patients operated on with Knodt rods for posterior fusions. The fusion rate was 93%; however, 33% of the patients required reoperation for removal of loose implants. Taylor and Gardner[287] reported on the use of Knodt rods in 36 patients. There was only a 64% fusion rate. Lee and deBari[172] completed a controlled study in which 24 patients were treated with laminectomies, foraminotomies, and posterior lateral fusion. Half had Knodt rods implanted, and the other half did not. No significant clinical benefit could be recognized for those patients having the rods implanted. Despite the mechanical effect of opening the intervertebral foramina, the clinical evidence does not, in our opinion, support a decision to use this implant.

Alar hooks with Harrington distraction rods; sacral bars with Harrington distraction or compression rods; Luque-Allen rods with insertion into the posterior iliac wings (Galveston technique); the Edwards sacral fixation device (with elliptical sleeves); and the René Louis lumbosacral plates and translaminar facet screws are all techniques for posterior lumbosacral fixation. There are a variety of presumed advantages and disadvantages of these various implants. Several of the implants do not maintain lumbosacral lordosis. Some can provide axial compression, some axial distraction, some both, and some neither. Except for translaminar facet screws, none appears to have much resistance to axial rotation. Some require more dissection than others, and two of them alter the mechanics of the sacroiliac joints.

In our view, there is not yet adequate experimental and clinical information upon which to base any definitive recommendations. The following study provides the best clinical evidence upon which to make a selection.

Kornblatt and co-workers conducted a clinical study in which 135 patients with posterior lateral lumbosacral fusions were reviewed to determine the factors that were associated with fusion rate and the time required for fusion. The use of facet joint screws or rods affixed to the pelvis decreased pseudarthrosis rate and time required for fusion at a statistically significant level ($p < 0.02$).[161] This work provides some justification for the use of implants to improve the rate of successful arthrodesis. More studies to support this assumption are needed.

Instrumentation for Thoracic and Lumbar Fractures

Since there has been considerable attention to instrumentation for this particular problem, we chose to present it separately. The management of these fractures is presented in Chapter 4. This section focuses primarily on implants with a view toward providing the reader with information that will help in the selection of a particular system.

The work of Tencer and colleagues[289] involved an experimental design in which they used a cadaver model and simulated a burst-type fracture with 35% spinal canal occlusion. The subsequent studies showed that distraction of the FSU an average of 5.2 mm would result in a significant increase in contact pressure of the canal material on the spinal cord. They noted that shortening an average of 3.2 mm did not have a significant effect, nor did flexion of 20°. The clinical implications are that in the case of a burst fracture with 35% canal occlusion, one would like to avoid overdistraction (>5.2 mm), compression of more than 3.2 mm, and flexion of more than 20°. This ought to be considered in the selection of an implant to treat the fracture.

The previously described work indicates that too much distraction is undesirable, yet we learn from Fredrickson and associates[89] that distraction is the major factor in the reduction of intracanal fragments. Obviously, overdistraction must be avoided. Here, the work of Andén and colleagues is useful. This work is presented in one of the following paragraphs.

McAfee and co-workers[191] completed cadaver spine biomechanical analyses involving compressions of three systems of posterior spinal instrumentation used for stabilization of simulated thoracolumbar fractures. The systems were: (1) conventional Harrington distraction instrumentation, (2) segmentally wired Harrington distraction rods, and (3) Luque segmental instrumentation. For experimentally created burst fractures, the segmentally wired Harrington distraction rods axially loaded were the stiffest ($p < 0.001$). For the experimentally simulated translational fracture dislocation, axially preloaded and axially rotated, the Luque segmental system was the stiffest ($p < 0.05$).

Ferguson and co-workers[75] completed in *vitro* biomechanical studies of destabilized fracture models in which they compared various forms of internal fixation. The surgical constructions were tested in axial torsion, flexion, lateral bending, and extension. The implants tested as follows in decreasing order of stiffness: Roy-Camille plates with six screws, wired Harrington rods, C-rods and J-rods, and Vermont Internal Fixator. The constructs with these same implants were tested for fatigue tolerance and failed in torsion and lateral bending but maintained themselves in flexion.

Andén and co-workers[6] showed that disruption of the anterior longitudinal ligament in unstable thoracolumbar fractures can be recognized at the time of surgery. Two factors can be monitored to recognize whether or not the ligament is intact. One is the forces required for distraction during the procedure; they will be greater when the ligament is intact. A force-indicating distraction is required to quantitate the resistance. The other is the pattern of vertebral displacement at the level of injury. When the ligament is intact, there will be sagittal plane angulation and rotation about an axis anterior to the vertebral bodies. When the ligament is disrupted, there is little or no angulation but vertical (y-axis) displacement with little or no sagittal plane angulation. The angulation can be readily recognized (*i.e.*, parallel or divergent displacement) if wires are put into those spinous processes which are adjacent to the fracture site. We applaud the measurement of therapeutically employed forces.

This certainly emphasizes the importance of analyzing the location and extent of instability in the selection of a particular implant. Obviously, an anterior and posterior combined instability should not be treated with distraction rods alone.

The report by Kaneda and colleagues[152] of 27 patients in whom there was an anterior retroperitoneal surgical approach, decompression fusion, and instrumentation provides a good sense of what can be achieved with this methodology. The fixation involves two screws in the vertebral body above and below the fracture. To these, two spiked plates are attached and are connected by two threaded rods with bolts that will allow distraction or compression. Zielke instrumentation was used in the first 12 patients, and the Kaneda device was used in the remaining 15. Of the patients with incomplete defects, 26 improved postoperatively, 19 by at least one Frankel subgroup.[87] This is presented as a satisfactory method for decompression and stabilization using just one operation anteriorly.

Kostuik[162] reported his experience with the treatment of 49 patients with fractures of the lumbar spine with and without neurologic involvement. Forty-two of these patients had anterior decompression. The patients had anterior fusion and internal fixation. The initial group had a modified Dwyer implant, and the first 31 patients were treated with a Kostuik-Harrington distraction system. The union rate in this series was 96%. In this group, 32 patients who had partial neurologic deficit were treated with anterior decompression and improved an average of 1.6 grades on the Frankel classification.[67] This, like the previously described series, suggests that this treatment program can provide successful improvement in neurologic status and can stabilize the patient with just one operation.

Yuan and associates[334] have presented their clinical experience with the Syracuse I plate in 16 patients. This is a low-profile metal plate that is attached to two vertebral bodies, each with fixation by two 6.5-mm cancellous screws. Patients included ten with acute burst fractures, four with metastatic lesions, and two with old healed fractures with deformity. Follow-up was 12–24 months, and complications were minimal.

We have reviewed three publications that are proponents of the anterior approach with decompression bone graft and internal fixation with an implant. There are other proponents also. However, we shall now present some of the advocates of the posterior approach. The cogent arguments for this approach have been presented by Jacobs and Casey,[142] based on a review of the literature and clinical experience with over 100 surgically treated thoracolumbar spinal injuries. They advocate internal fixation in order to mobilize the patient and protect neurologic structures. The patient should be clinically and radiologically evaluated to determine the presence of anterior and/or posterior instability and managed for protection of all instability. Neurologic recovery, they assert, can be expected with prompt and complete decompression through correction of deformity and malalignment along with the use of rigid internal fixation. They suggest further that the internal fixation should be able to provide compression for posterior injuries and distraction for anterior injuries and should also resist bending. They contend that the device should also be able to restore the normal contour of the spine. The approach they put forward to achieve all of this is the "rod long, fuse short" rationale (see p. 593).

Dick[63] has emphasized the desirability of maintaining as much lumbar motion as possible in patients who are paraplegic. Studies are in process to quantify the differences; nevertheless, he is certain that the elimination of lumbar motion significantly reduces the functional capacity of paraplegics. With lumbar motion, paraplegic patients can readily climb back into a wheelchair without assistance should they fall out. They can more readily lift objects from the floor while sitting in a wheelchair. They maintain a greater ability to bend forward longitudinally in the sitting position. A flexible lumbar spine helps in swinging the legs forward with the pelvis when walking. These are reasons for immobilizing as well as fusing the shortest segment of the spine that is compatible with adequate stabilization of the osseous and ligamentous injury. These considerations suggest another liability in the "rod long, fuse short" rationale and an advantage for instrumentation devices that allow adequate stabilization with a minimum number of FSUs being immobilized.

There have been problems with the use of Harrington instrumentation in the treatment of fractures of the spine. The deficiencies in the system have been analyzed and presented by Gertzbein and colleagues.[98] They noted a significant incidence of loss of reduction due to deficiencies of the anterior elements. Somewhat surprisingly, loss of reduction was not related to (1) the levels of instrumentation, (2) length of fusion, (3) severity of initial deformity, or (4) degree of initial correction. The recommendations were (1) to use a "C" clamp or wire on the lamina, (2) to use the lamina rather than the facet for hook placement, and (3) to use the compression system for fracture dislocations.

McAfee and co-workers have presented some dramatic complications involving the use of Harrington instrumentation.[186] They studied 40 patients in whom 45 Harrington instrumentation procedures were performed. Some of the complications may have been related to a failure to recognize or consider certain biomechanical principles. The complications were as follows: loss of fixation associated with dislodgement or disengagement, 16 patients; persistent neural compression due to inadequate distraction, 16 patients; inadequate reduction of translatory displacement, 9 patients; overdistraction, 4 patients; significant gibbus greater than 40°, 3 patients; Harrington rod breakage before fusion, 2 patients; and alar hooks posteriorly indenting the thecal sac, 2 patients. There were significant additional miscellaneous complications in 5 patients, and there was also 1 death. For prevention, the authors suggested monitoring distraction with x-rays or some other reliable technique. We believe that the information and material presented in this section on spinal implants will help the reader avoid these complications.

In recognition of the need for some compression or coaptation in grossly unstable thoracolumbar fracture dislocations, Floman and co-workers have employed a combination of interspinous wiring and

Harrington distraction rods (Fig. 8-67).[84] When all of the posterior elements are out, the complication of over*distraction* with posterior distraction instruments can occur. The spinous process wiring prevents that but nevertheless allows for the alignment and splinting that can occur with properly used Harrington distraction rods. This combination also creates bending moments that tend to restore disc height and correct the kyphosis.

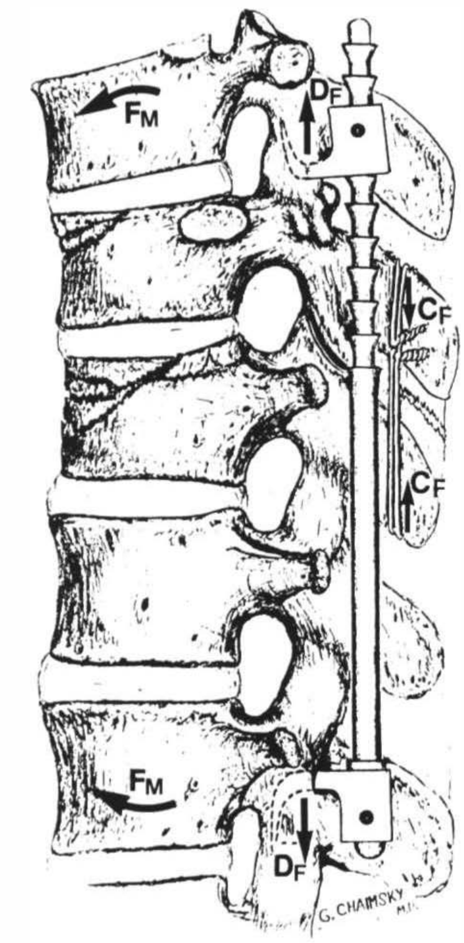


FIGURE 8-67 Forces involved when interspinous wiring is combined with Harrington distraction rods. The distraction forces are represented by arrows D_F. The flexion moment is indicated by arrow F_M. These are to some degree balanced by the force C_F, which is created by the posterior cerclage wiring that produces compression. The posterior elements, if intact, will allow a moment to be produced with the facet structures serving as a fulcrum. This creates the correctional forces and tends to reconstitute the discs and the vertebral heights. *(From Floman, Y., Fast, A., Pollack, D., Yosipovitch, Z., and Robin, G. C.: The simultaneous application of an interspinous compressive wire and Harrington distraction rods in the treatment of fracture-dislocation of the thoracic and lumbar spine. Clin. Orthop., 205:207, 1986.)*

Synopsis of Instrumentation for Fractures and Fracture Dislocations of the Thoracolumbar and Lumbar Spine

Our major clinical concerns are the patient's neurologic status and the condition of the spinal canal. Is there something in the spinal canal? Is the canal malaligned? We want to clear it of foreign material and align it. Overdistraction of >5.2 mm or underdistraction (i.e., shortening of <3.2 mm) may be a liability to the neural elements.[289]

The neurologic condition can be treated with posterior reduction and stabilization with or without exploration and removal of material within the canal. Burst fractures can be stabilized with Harrington distraction rods with posterior wiring, and fracture dislocations can be stabilized with the Luque segmental system[191] or spinous process wiring in association with Harrington distraction rods.[84] Complications of posterior instrumentation for these problems may be avoided by spinous process wiring, as described, or by careful monitoring of the distraction process so as to recognize disruption of the anterior and/or posterior elements.[6,186] When gross disruption of anterior and posterior ligaments is recognized, Harrington compression rods can be useful.

When the posterior approach is unsuccessful, an anterior approach may be required. Anterior approaches with decompression, fusion, and instrumentation have proved effective as one-stage procedures.[152,162]

It has been asserted that, all things considered, a posterior "rod long, fuse short" system with special laminar hook clamps is the system of choice.[142] The justification is that a *sole* anterior approach is likely to be inadequate if the posterior elements are nonfunctional. There have been significant complications with the regular Harrington system, particularly with hook dislodgement.[98,186] Several relatively new systems (Cotrel–Dubousset, Steffee, Dick fixateur interne, and others) are available that may prove useful in treating this problem.

The Cotrel–Dubousset instrumentation has been successfully employed in the treatment of unstable burst fractures. The rigidity of the surgical construction with this device allowed for shorter fusions and preservation of more FSUs.[58] A synopsis of the system is provided by Mubarak and associates.[203]

We do not think that there is adequate evidence upon which to base definitive recommendations.

A conservative approach, and the one that we recommend, is the use of Harrington distraction rods to monitor the displacement and to avoid overdistraction, a transpedicular decompression (see Fig. 8-11) if needed posterolateral approach to remove bone fragments from the canal. We recommend contouring the rods and fixing them two levels above and below with Drummond wires for secure fixation. If there is postoperative incomplete neurologic deficit and imaging evidence of material in the canal >30%, we think an anterior decompression and fusion is advisable.

A simplified comparison of some of the various posterior instrumentations is given in Figure 8-68. The normal spine was tested in flexion, and its stiffness was determined. A fracture model was created by making a defect in the vertebral body and dissecting the corresponding posterior ligaments. The fracture model was then tested alone as well as with various fixation devices. Figure 8-68 summarizes the data produced by Magerl.[183] We have presented it in this form to show the relative stability of the intact spine in comparison with the stability achieved with the various implants applied to a simulated fractured spine. The mode of failure is also indicated.

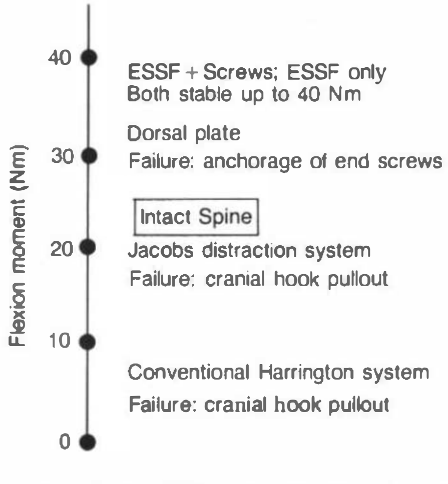


FIGURE 8-68 A simplified comparison of several posterior instrumentations applied to a simulated fracture model. The experimental fracture was developed by creating a defect in the vertebral body and dissecting the posterior ligaments. *ESSF* = external spinal skeletal fixation.

SACROILIAC JOINTS AND PELVIS

Sacroiliac and pelvic instability has been discussed in Chapter 5. The clinical biomechanics of the instrumentation of this region is the topic of Walheim's comprehensive thesis,[305a] on which we have based our presentation. The external skeletal fixation of the pelvis is achieved with the use of the Hoffman frame and its modifications. This device immobilizes the ilia but cannot control the sacroiliac joints. In order to achieve this either screw fixation across the joints or an internal plate and screw combination is required. The pubic symphysis can be immobilized with a plate–screw–bone graft combination or with the external skeletal fixator. All the plate screw combinations can be done with or without compression. These various implants are illustrated in figure 8-69.

SYNOPSIS OF SPINE IMPLANTS: T1–S1

The display on pages 613–615, "Generalizations on Implant Prototypes," summarizes the advantages and disadvantages of selected thoracic, lumbar, and lumbosacral spine implants from the perspective of clinical biomechanics.

TONGS AND TRACTION

The history of the use and development of tongs for the application of skeletal traction to the spine is interesting. The stimulus apparently was created in 1932, when a 22-year-old woman was in an automobile accident. She sustained, along with a number of other injuries, a compound fracture of the mandible and an open fracture dislocation of C2 on C3. The jaw injury obviated the usual head-halter treatment of the dislocation. A consulting physician, Dr. Coleman, suggested to the attending physician, Dr. Crutchfield, that extension tongs be applied to the skull. The sharp points were removed from the extension tongs and they were inserted into the skull, held together by a heavy elastic band. The treatment was successful and the case was reported in 1933.[56,57] The apparatus was subsequently mod-

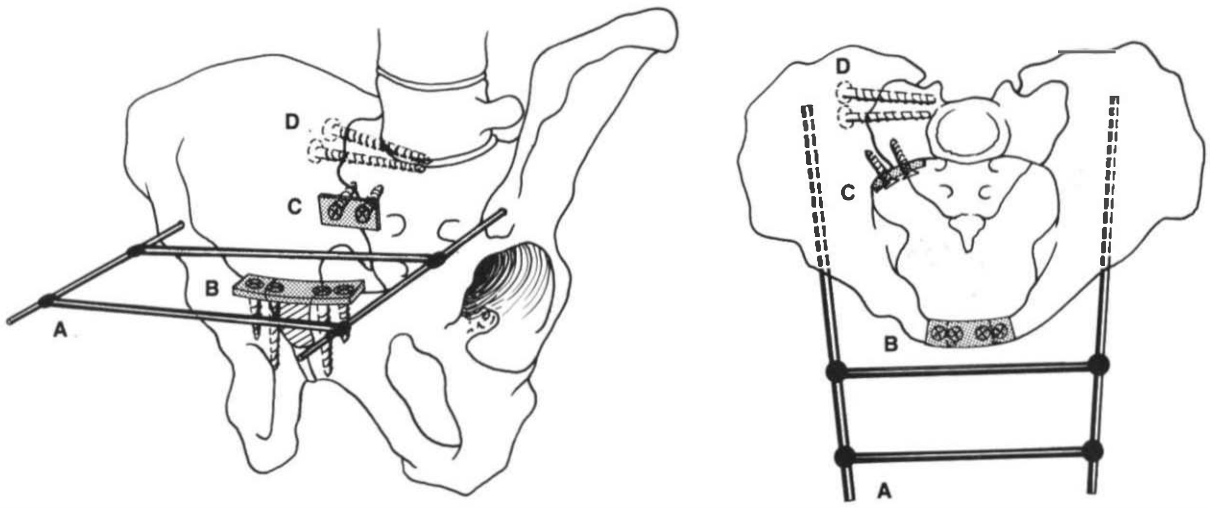


FIGURE 8-69 Four prototypical implants for the sacroiliac joints and pelvis: anterior oblique (*Left*) and cross-sectional (*Right*) views. *A:* External skeltal fixator of the ilia and pubic symphysis. B: This plate screw combination can be used in conjunction with a bone graft to achieve arthrodesis of the pubic symphysis. C: The sacroiliac joint can be instrumented from inside the pelvis with a plate and screw combination. Bone graft and/or compression may be included. *D:* The sacroiliac joints may be immobilized with two or more screws.

ified for clinical use and came to be known as Crutchfield tongs.

Crutchfield Tongs

These may even now be the most commonly used tongs. They are simple, effective, and easy to use. However, even in the hands of the experienced user, they sometimes slip out. Grundy's study showed that Crutchfield tongs either fell out or pulled out in 42 out of 104 patients after 21 days. There are two mechanical factors that are helpful in preventing slippage. The first relates to the magnitude of the axial load. The location of the tongs and their mechanism of attachment are such that they should not be employed when large loads are to be applied. If the physician wishes to use a 20- to 25-kg (40–50-lb) weight to reduce a facet dislocation or to do a stretch test, it is better to use the Vinke tongs. The second factor involves the relationship of the axial force vector to the cranium at the site of implant of each of the pins. This is best achieved by using the following guidelines: (1) Spread the tongs about 10–12 cm (4–5 in) apart for determining sites of insertion. (2) Since the preceding guideline may be affected by the shape of the cranium, we suggest these additional guidelines. Try to position the tongs such that the implanting pins are as close as possible to an angle of 90° to the table of bones of the skull and the line of pull of the traction. The skull pin angle is the more important. (3) Check the tongs daily and tighten them only when they loosen.

In the sagittal plane, it is recommended that the tongs be placed in line with the external auditory meatus. We suggest that the direction of pull of the traction and the positioning of the shoulders are much more important factors in determining the flexion/extension position of the neck than is the site of tong insertion.

Vinke Tongs

These are useful tongs. They are more versatile, and safe. The versatility results from the fact that very large loads of 20–25 kg (40–50 lb) may be applied with much less risk of the tongs pulling out. Vinke tongs are safe because they are less likely to penetrate the skull or to pull out than are other tongs. However, they require special insertion tools, skin incisions, and local shaving. Nevertheless, they are an improvement over the Crutchfield tongs in bio-

(Text continues on p. 616.)

GENERALIZATIONS ON IMPLANT PROTOTYPES

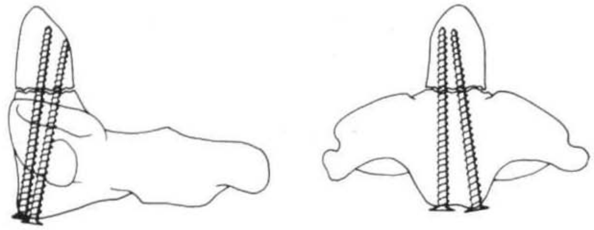
Spine Implants: C0–C7

I. The Halo

- A. The halo is an externally fixed implant for fixation of C0–T1
    - 1. *Advantages*
        - a. May be used without formal surgery
        - b. Can immobilize several levels
        - c. Can be used in an emergency
        - d. Is a low-stiffness, usually satisfactory immobilizer
        - e. Is relatively easy to apply
    - 2. *Disadvantages*
        - a. Cannot provide high-stiffness immobilization
        - b. Has certain complications and limitations

II. Anterior Implants

- A. Anterior screw fixation for dens fractures

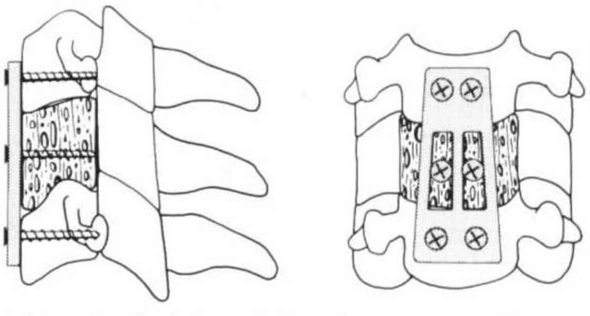


This anterior screw fixation is biomechanically stronger with two parallel screws. It is not known whether the risk/benefit of the "second screw" placement is in the patient's favor.

- 1. *Advantages*
    - a. Useful in preventing unwanted translation of fragment
    - b. Problem of fixation solved without eliminating axial rotation
- 2. *Disadvantages*
    - a. Relatively high and serious complication potential
    - b. Can be technically difficult with short neck or "barrel" chest

- B. Screw fixation of lateral masses and articular facets of C1–C2 (see Fig. 8-25)
    - 1. *Advantages*
        - a. Relatively high stiffness fixation without passing laminar wire
        - b. An alternative when the posterior elements or the posterior approach is not available
    - 2. *Disadvantages*
        - a. Two anatomically difficult exposures

- C. Plate and screw fixation (Caspar, Fuentes, Louis, Moscher, Roy-Camille)

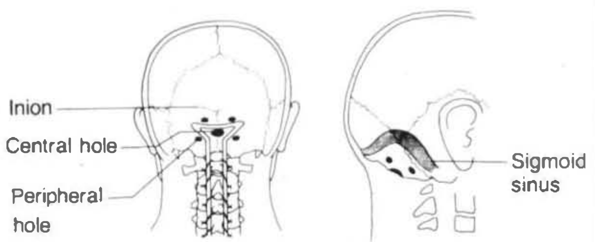


Although all of the risks/benefits are not yet known, there are situations in which the use of the plate to help stabilize a bone graft would be useful. The anterior plate and screw system alone may be inadequate to stabilize a grossly unstable cervical spine.

- 1. *Advantages*
    - a. May provide immediate postoperative stability, although additional posterior fixation may be needed to achieve it
- 2. *Disadvantages*
    - a. Loosening may occur
    - b. Problems of overpenetration of posterior cortex with screw may occur

III. Posterior Implants

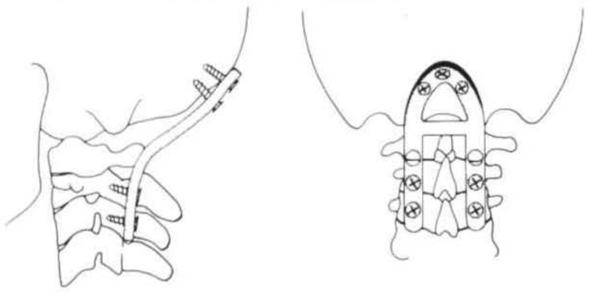
A. Wires occipital cervical to bone graft or implant (Robinson-Southwick, Itoh)



This technique, described by Itoh, et al.,[141] provides a stable occipital cervical fusion construct. The passage of sublaminar wires is a part of the procedure. We are in favor of avoiding this whenever possible. An alternative is to wire the rod to the facet joints, as described for bone graft wiring in Figure 8-40. A safe internal fixation of the occipital cervical region is important because it can obviate the necessity of prolonged use of the halo apparatus.

1. ***Advantages***
   - a. Stable internal fixation
   - b. High-stiffness resistance to all loading parameters
   - c. Patient comfort probably greater than with halo
2. ***Disadvantages***
   - a. Passage of occipital and cervical laminar wires provides significant risks

B. Plate and screws to occiput with plate wired or screwed to the vertebra (Fidler, Fuentes-Benezech)



These devices are designed to provide internal stabilization of the occipital cervical region. The safety, strength, design, and number of levels to be included by the implant all require considerable additional clinical and laboratory study.

1. ***Advantages***
   - a. Stable high-stiffness internal fixation
   - b. Screws in occiput may be safer than wires (authors' opinion, no data)
2. ***Disadvantages***
   - a. Risks if lamina wire passage is required

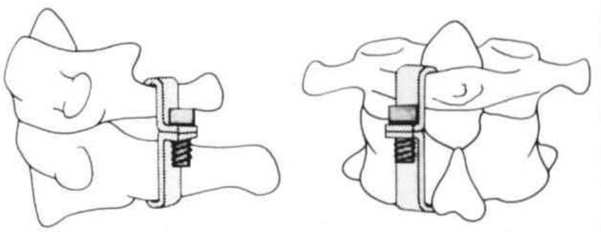
C. C1–C2 posterior wiring (see Figs. 8-27–8-30) (Brooks, various modifications; Gallie, various modifications)

1. ***Advantages***
   - a. Stable fixation in most parameters
2. ***Disadvantages***
   - a. Reduced stabilization against anterior translation
   - b. Risks of sublaminar wire passage
   - c. Decreased axial rotation

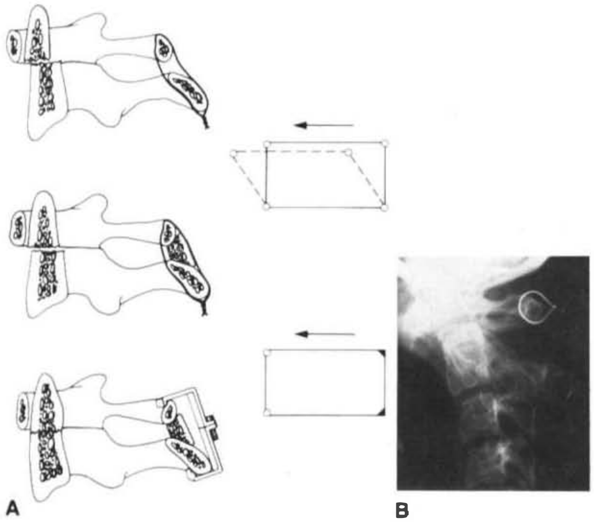
D. Screws, lateral masses C1–C2 (see Fig. 8-25) (Magerl, Barbour)

1. ***Advantages***
   - a. Can be used when posterior elements are absent or inadequate for fixation
   - b. Avoids passage of laminar wires
   - c. Provides better stabilization than the posterior wire fixation[117]
2. ***Disadvantages***
   - a. Risks of screw placement

E. Clamp C1–C2 (Halifax, Mitsui)



This device with or without an accompanying boneblock between the laminae is a relatively new implant that in our opinion deserves consideration and study.



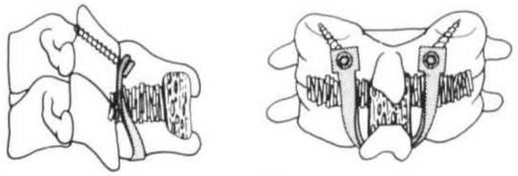
Parallelogram effect. (A) Here are several biomechanical considerations that are relevant to posterior C1-C2 fixation. If a posterior wire is used without bone graft wedges, there is no elasticity and a tendency for anterior instability due to what we've called the "parallelogram effect." This is due to the fact that the circular wire is not efficient in preventing translatory displacement. If a graft is wedged in as shown, there is more elasticity, which theoretically facilitates fusion. There is also more anterior translatory stability, and the "parallelogram effect" remains a hypothetical possibility. With the Halifax clamp and a wedge of bone graft, there is elasticity and greater resistance to anterior displacement and the "parallelogram effect." This procedure also prevents the passage of wire into the spinal canal with the associated risk of neurologic damage. (B) This patient was referred to us with this standard surgical construction. There was a chronic defect of the dens (post-trauma or developmental) associated with fatigue failure of the wire, nonunion, and anterior displacement of the atlas. A more stable posterior fixation might have prevented this.

1. ***Advantages***
   - a. Stable posterior fixation
   - b. No passage of laminar wires
   - c. Ease of application
   - d. No "parallelogram effect"
2. ***Disadvantages***
   - a. May cause reduction in extension
   - b. May slip off the lamina

F. Wires for lower cervical spine (spinous processes and facets) (see Figs. 8-39, 8-54)

1. ***Advantages***
   - a. Tried
   - b. Tested
   - c. Technically easy
   - d. Adequate stability
2. ***Disadvantages***
   - a. Relatively lower stiffness
   - b. May not be adequate for tumor surgery
   - c. Facet wiring technique unphysiologic for unfused adjacent joint

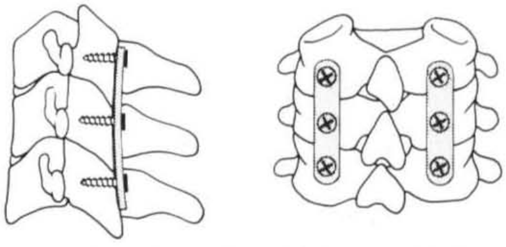
G. The clamp and hook plate in the middle and lower cervical spine (Magerl, Halifax; see the preceding diagram)



This implant has some of the same advantages as the Halifax. The risk of screw placement into the lateral mass and the relative biomechanical immobilizing parameters must be determined by further clinical observations and experimental studies.

1. ***Advantages***
   - a. These implants immobilize the cervical spine without passing laminar wires and without having to involve a normal facet joint, with the passage of facet wires
   - b. Stability of hook plate was superior to the interlaminar wire and also to the anterior bone graft

H. Screws and plates to lateral masses (Fuentes and Benezech, Louis, Magerl, Roy-Camille)



This type of implant will probably provide the most effective stabilization of all the posterior implants. At the time of this writing, the studies are not yet available.

Screws and plates to lateral masses

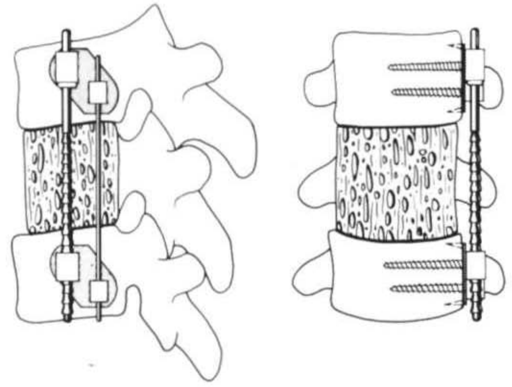
1. *Advantages*
    a. Can be used in the presence of extensive laminectomies
    b. Fixation without passage of laminar wires
    c. Magerl system is more stable than the Roy-Camille[200]
2. *Disadvantages*
    a. Technical risks
    b. More data needed on stiffness and load to failure

GENERALIZATIONS ON IMPLANT PROTOTYPES

Thoracic, Lumbar, and Lumbosacral Implants: T1–S1

I. Anterior Implants

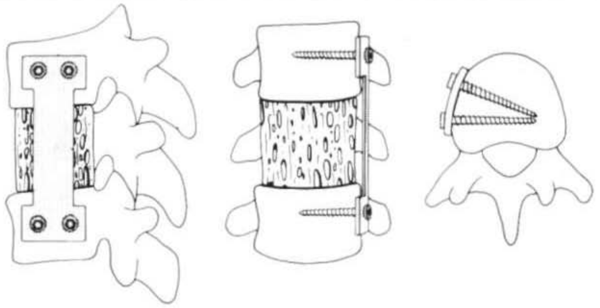
A. Vertebral body screw with plates and connecting rods (Kaneda, Kostuik)



These are adjustable anterior (attached to vertebral body) devices designed to distract and internally fix two or more vertebral bodies.

1. *Advantages*
    a. One-stage decompression and fusion
    b. Short-segment fusion
    c. Immobilization with adequate rigidity
    d. Some have both compression and distraction capability
2. *Disadvantages*
    a. More serious complications with anterior surgical approaches
    b. Implant may interfere with vascular structures

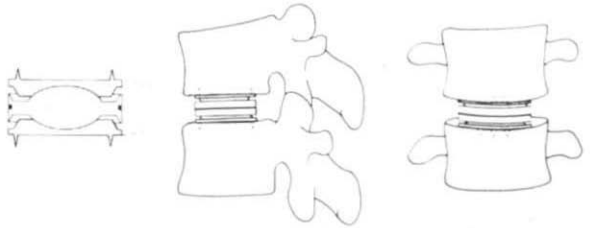
B. Vertebral body plates and screws (Syracuse-I plate) (Yuan)



A plate for internal fixation of the vertebral bodies.

1. *Advantages*
    a. One-stage decompression and fusion
    b. Short-segment fusion and immobilization
    c. Low profile
2. *Disadvantages*
    a. Incomplete control of compression and distraction

C. The artificial lumbar disc (Zippel)



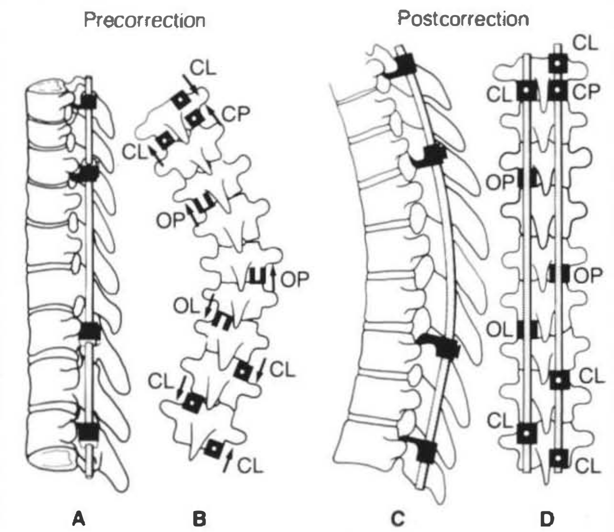
Because the implant cannot be completely visualized on x-ray, we have provided an artist's illustration of the components and their position in the vertebral body interspace.

The artificial lumbar disc

1. ***Advantages***
   - a. Pain relief without elimination of motion
   - b. Does not eliminate the possibility of subsequent arthrodesis
2. ***Disadvantages***
   - a. New, experimental
   - b. Requires anterior approach

II. Posterior Implants

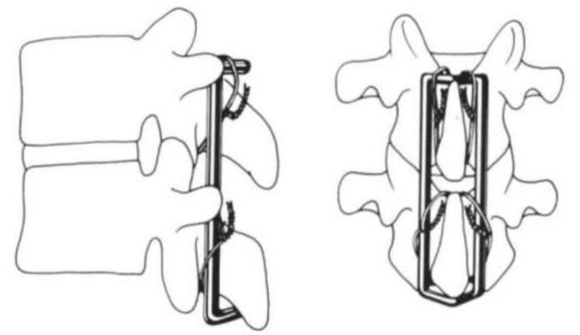
A. Distraction and compression rods with hooks (Cotrel-Dubousset, Edwards, Harrington*, Jacobs)



*This shows a scoliotic/lordotic spine, before and after correction with the CD system. CL = closed laminar; OP = open pedicular; OL = open laminar; CP = closed pedicular. The appropriately contoured, concave rod is placed as shown in **A** and rotated 90° about its longitudinal axis. This corrects the frontal plane scoliosis deformity, **B** to **D**, and also the sagittal plane lordotic deformity, **A** to **C**, converting it into a more anatomically normal thoracic kyphosis. This may be the most complex, versatile, and advanced posterior fixation device. It is generally capable of providing the most rigid surgical construction with the spine. However, we do not think there is enough evidence yet to analyze the risk/benefit or the cost/benefit relationships.*

1. ***Advantages***
   - a. Posterior approach has less potential for serious complications
   - b. Collectively, the systems offer a broad range of stiffnesses (C-D is the most rigid)
   - c. Harrington system is rigid enough to achieve fusion; also, distraction forces can be quantitatively monitored with this system
2. ***Disadvantages***
   - a. Generally require longer fusion and instrumentation than do the anterior devices
   - b. Harrington system has the risk of hook dislodgement
   - c. With burst fracture, anterior decompression may be required

B. Implants with segmental laminar wiring (Hartsill, Luque, Double "L", C-rod)

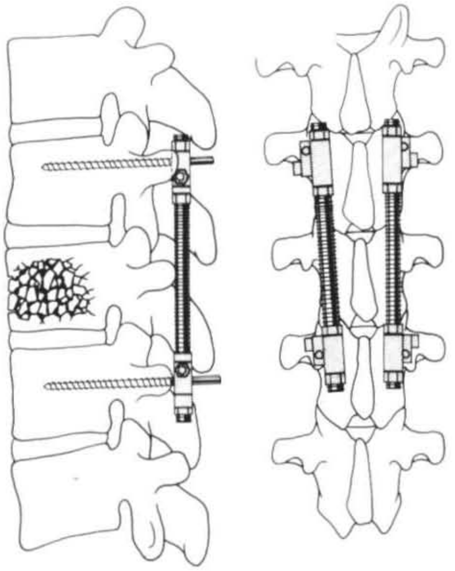


Segmental fixation with a wire that encircles the lamina and a portion of a metal rod. This forms a strong attachment, and it thoroughly immobilizes the spine. However, entry into the spinal cord canal has certain inherent risks.

*The stiffness of this system may be significantly enhanced by wiring the Harrington rod to the spinous process and also by using the Edwards sleeve. Laminar wiring to the Harrington rod will also significantly increase stiffness.

1. ***Advantages***
   a. A stiff system, particularly in axial rotation
2. ***Disadvantages***
   a. Danger to neural structures when laminar wires are used
   b. Cannot control compression or distraction
   c. Less "user friendly"

C. Intrapedicular fixation, with plates or rods (Cotrel-Dubousset, Dick, Edwards, Louis, Magerl, Olerud, Roy-Camille, Steffee, Vermont Spinal Fixation [Krag], Wiltse, others)

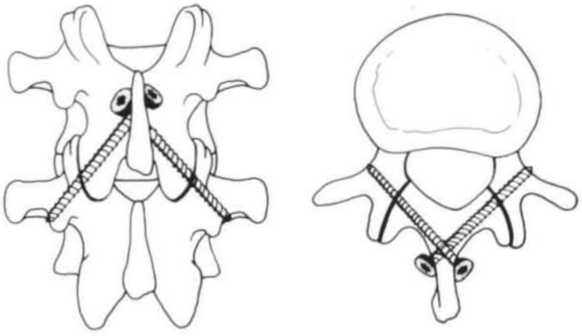


Transpedicular fixators provide a rigid construction. However, there is the risk of damage or irritation to nerve roots.

1. ***Advantages***
   a. High stiffness in all parameters of loading
   b. Most are relatively "user friendly"
   c. Some allow distraction and/or compression
   d. Some will allow reduction of abnormal translation
   e. Improvement of fixation of osteoporotic bone with pedicle screw and cement

   f. Can bone graft vertebral body through pedicle
   g. Can be used as a clinical test for pain and instability if Magerl external pedicular system is used
2. ***Disadvantages***
   a, Some systems may be too stiff
   b. Early reports show high complication rate (hardware failure, infection, nerve root damage or irritation, failure of fusion)

D. Laminar facet screws (King, Magerl)



This appears to be a simple and useful technique; however, more laboratory and clinical studies are needed. This simple spine fusion construction for fusion of the thoracolumbar spine has been shown to rigidly immobilize the spine[183] and also to be associated with a better fusion rate.[161] Because of the sagittal plane orientation of the facet joints, they can be readily transfixed with a translaminar approach. Professor Magerl indicates that the screws should not be applied as lag screws, because to do so would allow slipping.[183]

1. ***Advantages***
   a. Ease of application
   b. Adequate stiffness, including axial rotation
2. ***Disadvantages***
   a. Fusion and complication rate unknown

mechanical design and instrumentation. The value of the design is shown in Figure 8-70. The pin attachments are both automatically placed at about 90° to the line of pull and the table of the cranium at the site of implant. There is the added insurance against pullout that comes from the flange mechanism that spreads out between the two tables of the skull. This gives added protection against penetration through the inner table and makes it impossible to pull the pin out without tearing through the outer table of the skull or twisting the flange mechanism back to its original position. This feature alone justifies our strong preference for this instrument over the Crutchfield tongs, which have been associated with death from brain abscess due to skull penetration by the pins.[311] A variety of other complications can accompany the use of skull caliper traction. Two other rather dramatic ones are intracranial aneurysm and intracranial hemorrhage. The work of Hirsch[133] provides an excellent review of complications and some guidelines on the safe use of the devices.

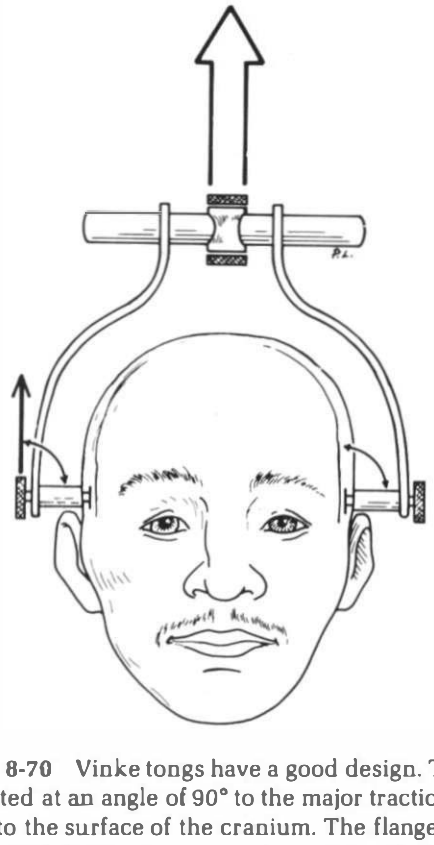


FIGURE 8-70 Vinke tongs have a good design. The pins are inserted at an angle of 90° to the major traction vector and 90° to the surface of the cranium. The flange mechanism prevents the tongs from coming out from between the tables of the skull.

Gardner-Wells Tongs

Gardner introduced the concept of spring-loaded points for cervical traction in 1973.[96a] Gardner-Wells tongs are probably the most widely used tongs in the United States, and we recommend them. They are simple to apply, and require no shaving, skin incision, or special equipment. The instructions appear on a metal plate that is attached to the tongs. They do have disadvantages. The pins protrude laterally, which makes lying on the side difficult. The single size hoop may not fit small crania.[163a] Rimel's modification[241a] of Gardner-Wells tongs (University of Virginia tongs) addresses these problems. Gardner-Wells tong application is straightforward. First, ensure that the points are sharp. Next, choose, cleanse, and anesthetize the pin insertion sites. The insertion site is 2 cm cephalad to the pinna in line with the tragus. This location is well below the maximum biparietal diameter.[163b] Center the hoop in the frontal plane and advance both pins equally. One pin is spring-loaded and has an indicator stem. Tighten the pins until the indicator stem protrudes 1 mm. The pins now have 25-lb compressive force against the skull.[163a] The cadaver study by Krag showed that the mean pull-off force was 137 ± 34 pounds when the indicator protrudes 1 mm. If the pins were undertightened such that the indicator stem protruded only 0.25 mm (one-quarter of the recommended distance), the minimum observed pull-off strength was still 60 pounds.[163a] These pull-off strengths provide adequate fixation for the suggested cervical traction weights in different clinical situations (see Tables 8-6 and 8-7).

Bow for Application of Traction

Wang and colleagues[307] have developed a device to make the application of traction more versatile and convenient in the hospital setting. The device is less bulky, does not require traction weights, does not interfere with an x-ray machine, and does not have to be removed for patient transport or for spinal surgery. Also, the device has a measuring gauge to quantitatively monitor forces applied.

The Halo Apparatus

The Halo device is clearly an implant into the cranium. It is also the ultimate orthotic device. Therefore, it is discussed also on page 488 in the chapter on orthoses.

TABLE 8–6 Cervical Traction Weights for Treatment of Fractures and Dislocations at Various Levels in the Cervical Spine

Level of Injury	Minimum Weight kg (lb)	Maximum Weight kg (lb)
C1	2.3 (5)	4.5 (10)
C2	2.6 (6)	5.4 (12)
C3	3.6 (8)	6.8 (15)
C4	4.5 (10)	9.0 (20)
C5	5.4 (12)	11.3 (25)
C6	6.8 (15)	13.6 (30)
C7	8.2 (18)	15.9 (35)

(Crutchfield, W. G.: Skeletal traction in the treatment of injuries to the cervical spine. J.A.M.A., 155:129, 154. Copyright © 1954, American Medical Association.)

TABLE 8–7 Suggested Guidelines for the Amount of Traction to Achieve Various Clinical Goals

Region	Anchor	Physiologic Alignment	Correction of a Deformity†
Cervical	4–8 kg (8–18 lb)	8–10 kg (18–22 lb)	10–28 kg (22–62 lb)
Upper thoracic	8–10 kg (18–22 lb)	10–15 kg (22–30 lb)	15–28 kg (30–62 lb)
Mid- and lower thoracic	10–15 kg (22–30 lb)	15–20 kg (30–44 kg)	15–28 kg (30–62 lb)
Lumbar*	10–15 kg (22–30 lb)	18–27 kg (40–60 lb)	23–36 kg (50–80 lb)

* Values were calculated for the lumbar region under the assumption that no antifriction device is employed.

† Reduction of difficult dislocation or fracture dislocation.

This device is responsible for a number of significant advancements in the surgical management of the spine.[214,217] It consists of a stainless steel ring with holes through which pins may be passed, implanted into the outer table of bones of the skull, and then fixed to the circumferential rim (halo). A discussion of prominent biomechanical characteristics follows.

Through its multiple attachment points, it is possible to gain excellent fixation of the cranium. This permits the application of large loads over a long period of time because alternate pin sites may be employed. We have found that it is useful in children who need rigid skull fixation, because the bone of a child's skull is relatively softer and more elastic, and therefore pin fixation of the skull is very difficult. The multiple points of attachment permitted with the halo apparatus reduce the stress at any one point (Fig. 8-71B). The multiple fixations allow for more precise control of the head in all three planes. This provides excellent indirect control of the cervical spine, which can be particularly helpful in a grossly unstable situation. With this apparatus, forces and moments can be applied to the spine to control flexion/extension, lateral bending, and axial rotation. The halo is a useful technique when one wishes to place the head and neck in the desired position with the patient awake. With this device there are the options of (1) simple skeletal traction, (2) attachment to a plaster body jacket, (3) attachment to a fabricated plastic body jacket, or (4) attachment to a pelvic ring attached to the iliac bones.

In order to guard against penetration of the pins into the skull, the pins are inserted with a torque wrench, and the torque should not exceed 0.625 Nm (5 in lb) in children or 0.65 Nm (5.5 in lb) in adults.[159,214] However, the recent work of Cloward[51] and Garfin[97] and their respective co-workers has shown that with 0.90 Nm (8 in lb) of torque there was less pin loosening and fewer pin tract infections.

This device provides the spine surgeon with a real asset for the care of a broad variety of spine problems. We would like to share also one of its limitations. Whitehill and co-workers[323] have reported five cases in which the use of halo fixation was associated with resubluxation. Three of the patients had unilateral facet dislocations (one was a fracture dislocation), and two were bilateral facet dislocations. The report indicates a need for caution and attention to precisely which cervical spine conditions can reliably be treated with the halo. Here we have cervical spine injuries generally thought to be relatively stable (unilateral facet dislocations) as well as quite unstable injuries (bilateral facet dislocations) both redislocating in a halo. The fixation points at the skull and the thoracic cage obviously allow too much motion in the *middle* cervical spine region. We know from Chapter 7 that studies of the halo show motion in this region.

We cannot resist speculation here. These clinical losses of reduction in the middle cervical spine may, as suggested, be due to excessive motion of the *middle* cervical vertebra. This motion is possible because of the distance of the vertebra from the two fixation points. However, it is also possible that they are related to the characteristic biomechanical properties of the middle cervical spine, which are discussed in Chapter 4.

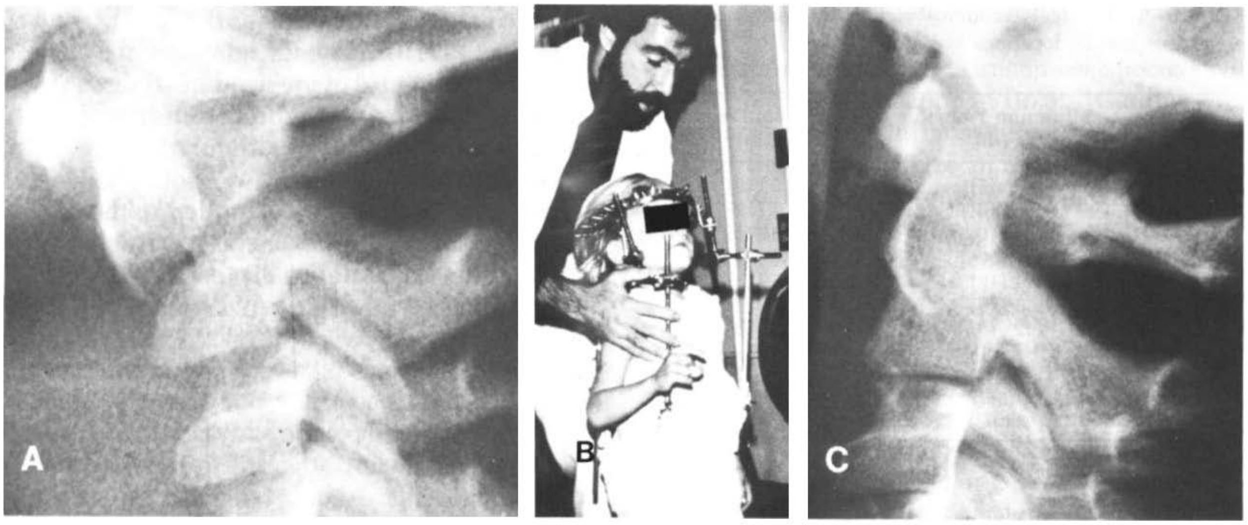

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FIGURE 8-71 The use of the halo apparatus, employing multiple pin fixation to diminish individual pin stress in the relatively soft bone of a 4-year-old boy with a fractured odontoid. **(A)** The initial displaced fracture of the odontoid is shown. **(B)** Application of the halo apparatus with extra pin fixation. **(C)** Fracture after 10 weeks of immobilization.

The complications with the use of the halo have been well documented.[97,274] The most common complication of pin loosening and infection can be greatly reduced by appropriate technical changes. These include proper pin placement, the use of a higher torque to insert the pins, and meticulous care of the pin site. Don't attempt to maintain a grossly unstable cervical spine in a halo vest, because it does have some limitations. Yet, despite all these potential problems, Ersmark[72a] found, in a well-done clinical study, that the halo vest was superior to collars, skull traction, and surgery in the treatment of cervical spine injuries.

How Much Traction?

This is a difficult problem that requires clinical judgment and careful radiographic monitoring. Traction for pain and for diagnostic evaluation is discussed in Chapters 6 and 5, respectively.

Here, traction is discussed with respect to immobilization of the spine and correction of a local or regional deformity.

Crutchfield suggested the guidelines shown in Table 8-6. His concern was that excessive distraction should not occur.

Traction is sometimes used for immobilization rather than distraction, but as an anchor. This is the situation in which a patient has an injury that is stable, but the patient needs to be held quiet with the spine protected from intrinsic muscles, loads imposed by gravity, movement, or low-magnitude forces that might be applied from the outside.

The next level is that in which alignment must be attained or maintained against moderate physiologic or deforming forces. This requires more of a force, and the amount of that force increases as the lesion moves caudally.

When a recalcitrant deformity must be overcome, as with scoliosis, kyphosis, or a fracture that is difficult to reduce, the ranges increase considerably. For the unique situation of a unilateral facet dislocation that is difficult to reduce, we suggest traction of up to one-third body weight, but not to exceed 32 kg (65 lb). For a summary of the traction guidelines, see Table 8-7. These figures are only guidelines. They should all be reduced by about 20 or 30% if an antifriction device is placed under the patient. These figures are our recommendations based on various data sources in the literature and our own research and clinical experience. For any given clinical situation, the guidelines may be employed initially. Then the traction is adjusted to fit the unique requirements of the individual patient. The adjustments are made with careful monitoring of the situation through checks for pain, neurologic status, and radiographic analysis.

PART 5:
AN ANALYSIS OF THE MECHANICS OF SPINE OSTEOTOMIES

BASIC OSTEOTOMY

The basic goals in the use of osteotomy are to gain correction, to achieve and maintain clinical stability, and to avoid damage to vital structures. The ideal site for osteotomy from the geometric and mechanical standpoint is at the apex of the curve of the deformity.

The basic design of spinal osteotomies has been that of a wedge cut through the posterior elements (Fig. 8-72). The base of the wedge of the osteotomy should be in the direction in which the apex of the curve is pointing. The apex of the wedge should reach at least as far anteriorly as the center of motion about which the correction is to take place. Thus, the apex of the wedge should be at or close to the posterior longitudinal ligament. This traditional design has been used for a number of years. A recent improvement is discussed on page 621.

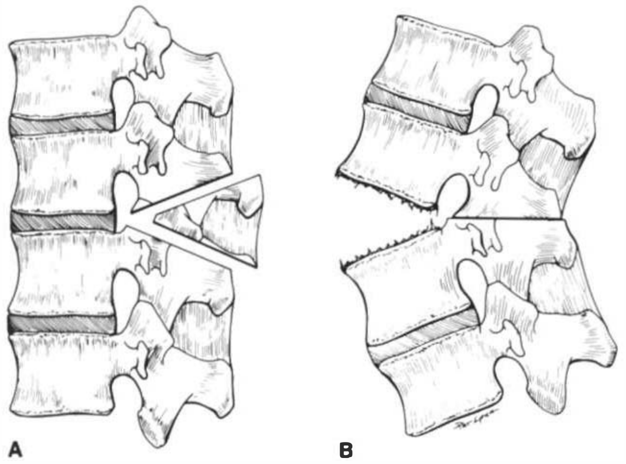

FIGURE 8-72 (A) This is a diagram of the configuration of an osteotomy. It applies to any region of the spine but is probably safest in the lumbar region. The apex of the wedge is in the region of the posterior longitudinal ligament, and the angle at the apex is the same as the angle of the correction. (B) The corrected spine with the posterior elements in apposition and the anterior elements distracted.

The size of the angle of the wedge is the same as the angular correction that will be achieved and usually is in the range of 40–60°.[196] The structure that is the greatest distance away from the center of motion opposite the osteotomy on the concavity of the curve must be cut, broken, or deformed. Care should be taken to construct the osteotomy so that the center of motion or the fulcrum about which the correction takes place is not behind the spinal canal, since the correctional rotation about the axes could cause excessive stretch and damage to the neural elements.[2] The literature shows that circumstances in which there is difficulty in rupturing the anterior ligamentous structures rarely occur. Some of the variations of the basic osteotomy and some of the relevant biomechanical factors are discussed.

CERVICAL AND CERVICOTHORACIC OSTEOTOMY

The actual site of deformity is generally in the cervicothoracic region. In terms of pure mechanics, this is the most logical place to carry out the corrective osteotomy. This site, at the lower cervical spine below C6, is used to avoid the vertebral artery area. Operating in this area carries a high risk of spinal cord damage. When all the posterior elements are transected and the anterior elements are ruptured, a clinically unstable situation is produced.[129] Posterior wedge osteotomy and section of vertebral body from the posterior exposure with alternate side-to-side retraction of the spinal cord has been reported.[164] There is relatively little extra space for the neural elements in this region; therefore, the risk of damage from either displacement or surgical encroachment is high. For these reasons, in cervical osteotomies we suggest a halo device[265] or some method of obtaining immediate postoperative stability, such as internal wiring. It is also advantageous to design the osteotomy to include ample posterior element resection so that there is room for the unobstructed posterior displacement of the spinal cord (Fig. 8-73).[129,265,300] Thoracic spine osteotomies have also been carried out.[129] The considerations are essentially the same as for the cervical spine.

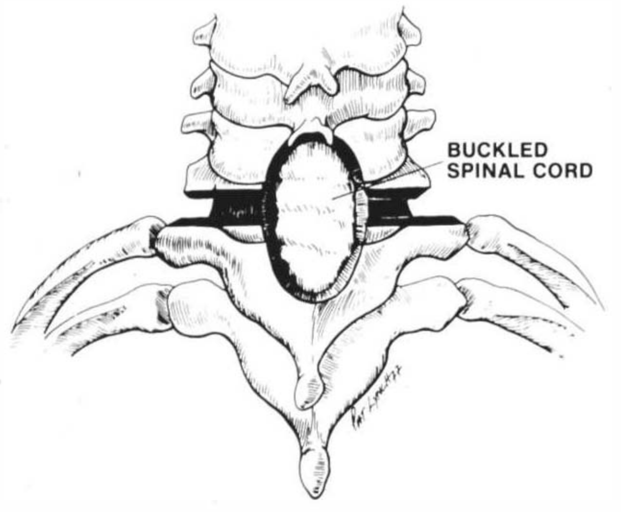

FIGURE 8-73 The Simmons construct for cervical osteotomy, in which there is also adequate laminectomy above and below to allow for any buckling or displacement of the cervical spine that may occur with the correction. The lateral resections are beveled toward each other so that opposing surfaces will be parallel and in apposition following extension osteotomy. *(Modified from Simmons, E. H.: The surgical correction of flexion deformity of the cervical spine in ankylosing spondylitis. Clin. Orthop., 86:132, 1972.)*

LUMBAR SPINE OSTEOTOMIES

In this region also there are problems of clinical instability and neural damage associated with osteotomy and the subsequent displacement of the spine. However, the risks are reduced because of the increased space for the neural elements and the relative clinical stability of the lumbar region. However, postosteotomy instability has been reported here, too.[129] With the exception of Briggs and Keates, who used Wilson plates,[30] most surgeons have used casts, traction, or recumbency rather than internal fixation for lumbar spine osteotomies. There have been a number of variations on the basic constructs described by Smith-Petersen and colleagues.[269]

In the frontal plane, the osteotomy may be transverse or V-shaped. The latter design, originally suggested by Smith-Petersen, is preferable. It provides good potential for correction and some postosteotomy stability against anteroposterior translation and axial rotation.

Anterior surgery to release the anterior longitudinal ligament and the annulus in addition to the posterior osteotomy has been suggested.[165] The purpose is to control the correction and avoid any damage that may come from the relatively imprecise directive "bend until there is a resounding snap." Osteotomies at two or more levels have also been recommended to reduce the stress at one level and/or to gain additional correction.[196] There have also been modest variations relevant to the exact configuration of the osteotomy.

Some of the more significant variations are discussed here. Briggs and Keates emphasized the importance of foraminotomy of the posterior portion of the intervertebral foramen in addition to a portion of the pedicle to avoid nerve root encroachment at the time of correction following osteotomy. Adams recognized the advantages of doing the procedure with the patient in the lateral position (see the list below).[2] He also designed an apparatus with which to correct the deformity on the operating table with gradual, controlled application of three-point bending. This device is described in his publication and would be useful to any operating theater doing more than an occasional spinal osteotomy.

The overall mortality from spinal osteotomies is about 10%.[129, 170] The complications are listed below.

There are other situations in which spinal osteotomies are indicated for kyphosis. Sharrard has described a procedure for congenital kyphosis in meningomyelocele.[261] Osteotomies have also been carried out in adolescents and adults with partial or complete paralysis associated with severe kyphosis.

ADVANTAGES OF THE LATERAL POSITION FOR CORRECTIVE SPINAL OSTEOTOMIES*

Facilitates positioning of the grossly flexed patient
Facilitates administration of anesthesia
Blood flows out rather than welling up
Eliminates injury risk to ankylosed cervical spine
Provides sturdier and more comfortable position for the surgeon

* Since most patients who are treated for spinal osteotomy have severe ankylosing spondylitis, it is important to keep in mind the fact that they are primarily abdominal breathers, because their costovertebral joints are generally ankylosed. The value of protecting the cervical spine from injury in these patients is not theoretic, as one death has been reported from fracture dislocation in this region associated with lumbar spine osteotomy in the prone position.[2]

COMPLICATIONS FROM SPINAL OSTEOTOMY

Ruptured aorta or inferior vena cava
Paralytic ileus (superior mesenteric artery syndrome)
Cervical fracture or fracture dislocation
Nerve compression from vertebral subluxation
Death from postoperative cervical instability

A variety of osteotomies have been performed for scoliosis associated with unilateral vertebral bars and hemivertebra. Sometimes in severe scoliosis it is necessary to perform an osteotomy on an iatrogenic or spontaneously fused segment to gain correction with either halo pelvic or halo femoral distraction.* The basic principles previously discussed apply in virtually any corrective osteotomies. They are resection at the location of maximal deformity, protection and preservation of neural structures, and the establishment of adequate postoperative clinical stability, with internal or external fixation as needed.

There are some recent works on spinal osteotomies for ankylosing spondylitis that we think are useful contributions. McMaster and colleagues[195] reported on 14 patients. They used the Smith-Petersen technique in conjunction with a compression device. The results were reported as good, with fusions occurring in 9 months. The authors offer a "pearl" with the suggestion that keeping the nasogastric tube in for 48 hours postsurgery will prevent mesenteric artery syndrome.

Thomasen[290] reported osteotomy in 11 patients. In addition to the posterior osteotomy, he elected to remove the pedicles and a portion of the posterior vertebral body that is then subjected to a "surgical" compression fracture posteriorly (Fig. 8-74).

Camargo and colleagues[41] reported on 66 patients. They had one death from a ruptured aorta that happened to be calcified. Perhaps a calcified aorta should be considered a relative or even an absolute contraindication to the procedure.

There are several advantages to the Thomasen technique. First, with the removal of the pedicles and posterior elements, nerve roots cannot be damaged. Second, with the deeper posterior wedge there is less stretch of the anterior structures. This could be important in regard to the aorta, particularly when it is calcified. Third, because there is cancellous to cancellous bone healing due to the surgical "compression fracture," healing may occur without bone graft and may possibly occur more rapidly. Then internal fixation is suggested, because 1 of the 11 patients with no internal fixation had an anterior vertebral body fracture associated with an L2 dislocation and neurologic deficit. This more recently developed technique seems to represent a significant advance.

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* Personal communication, J. P. Kostuik, M.D.

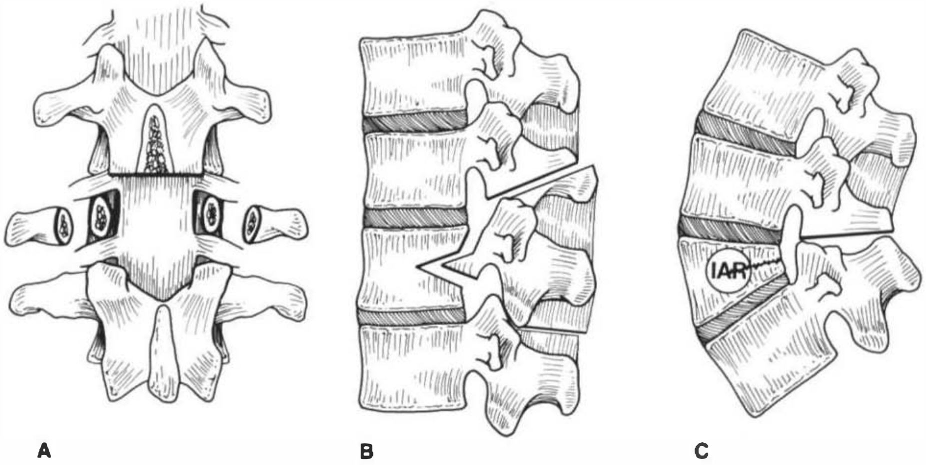

FIGURE 8-74 **(A)** Posterior view of removal of the posterior elements. This includes the inferior portion of the spinous process above the laminae, the inferior articulating process of the facets, the pedicles, and a posterior portion of the vertebral body. It is not necessary to remove the transverse processes. **(B)** Lateral view showing removal of these same structures. **(C)** The new position following the correcting osteotomy. The instantaneous axis of rotation (IAR) is more anterior with this technique, thus accounting for some of the advantages mentioned in the text.

■ CLINICAL BIOMECHANICS

Decompressions

- It is essential to use all available clinical and imaging information to localize the site of the offending pressure as accurately as possible.
- The site of pressure must be identified in relation to the spinal cord, the vertebral body, and the midline. The most appropriate decompression and surgical reconstruction can then be selected.
- There may be encroachment both anteriorly and posteriorly because of a contrecoup situation in which there is primary pressure or encroachment on one side that pushes the cord through the remaining free space against normal anatomic structures on the opposite side. Removal of the primary pressure initially is the best choice of surgical construct.
- Anterior exposure is recommended for anterior structures, with the selection of the appropriate procedure to cover situations in which the offending pressure is behind the vertebral body.
- The multilevel laminectomy is not the best choice for anterior pressure. It should be employed when the anterior approach is not possible or when there is a developmentally narrow canal. Laminectomy is the procedure of choice for primary posterior compression problems.
- Cervical canal size is a major factor in the clinical manifestations of cervical spondylotic myelopathy, ossification of the posterior longitudinal ligament, cervical disc disease, and cervical spine trauma.
- The considerations of decompression in the thoracic region are essentially the same as those in the other regions, except that posterior decompressions have been shown in *in vitro* biomechanical experiments to be less effective than anterior procedures for removing anterior pressure.
- The dentate ligaments may play a role in the clinical biomechanics of the spine.
- In the lumbar spine, the majority of neural encroachments may be thoroughly decompressed from the posterior approach.
- As in the cervical region, problems of disc disease, trauma, and spinal stenosis are more severe when there is a developmentally small lumbar spinal canal.

Spine Fusions

- Spine arthrodesis is generally employed to reestablish strength, maintain correction, prevent progression of deformity, and alleviate or eliminate pain by altering the regional mechanics.
- Spine fusions may be accompanied over time by several types of alterations to adjacent FSUs. Some of these changes are themselves diseases caused by abnormal mechanics that lead to symptoms.
- All things considered, allograft is almost as good as autograft for spine fusions in adults.
- The choice of bone graft material for a particular surgical construct involves some biomechanical considerations. The ilium is generally preferable and probably the most versatile. However, the ribs offer some appealing advantages. The use of the fibula and tibia has some liabilities. In some situations, allografts may be equally as effective, with some practical advantages.
- Positioning of bone grafts is important. If fusion is performed to provide clinical stability, then posterior positions with maximum leverage are important. If movement between the vertebral bodies is expected to affect the disc, then interbody placement of the graft is preferable.
- The placement of a bone graft on the compression side *away* from the neutral axes is most effective in a surgical construct used to treat kyphosis. However, the longer the bone graft, the more likely it is to succumb to buckling failure.
- The extent of fusion depends upon the clinical goals that the surgical construct is designed to achieve. Generally, it is best to fuse to the first adjacent normal FSU. With special constructs in the treatment of major deformities, additional normal FSUs should be included. In kyphosis, all vertebrae in the deformity should be included.
- As a first approximation, a reconstructive spine arthrodesis is carried out at the site of major destruction. If it is anterior, the reconstruction should be anterior; if it is posterior, the reconstruction should be posterior. There are exceptions.
- Anterior C0–C1 fusion is useful when the posterior route is not available. The transverse ligament should be intact to ensure clinical stability.
- Posterior C0–C1 or C0–C2 fusion may be achieved by a simple technique that can be clinically stabilized with a halo apparatus. A more complex and difficult wiring procedure is also available. The latter is useful when posterior elements are missing and immediate maximum stability is required.
- Bilateral anterior screw fixation of the lateral masses provides good fixation of the atlanto-axial joint for arthrodesis.
- Circumferential wiring of an iliac bone graft between the posterior elements of C1 and C2 (Brooks construct) provides a stable construct for atlanto-

axial fusion. This has been the construct of choice when posterior fusion of C1–C2 is indicated. With more clinical experience, the Halifax may prove to be preferable.

- Anterior cervical interbody fusion with the horseshoe graft (Smith-Robinson construct) removes the disc, exposes the medial and lateral aspects of the interspace, enlarges the neural foramen, elongates the yellow ligament, and resists collapse. We suggest this operation as the procedure of choice for cervical spondylosis.
- Anterior cervical trough fusion (Bailey-Badgley construct) removes all or part of the intervertebral disc, exposes midline portions of the cord behind the vertebral body, can enlarge the interspace and elongate the yellow ligament, resists collapse, provides immediate postoperative stability even in some difficult situations, and conveniently fuses one or more FSUs.
- Anterior dowel interbody fusion (Cloward construct) removes all or part of the intervertebral disc, nicely exposes the anterior region of the spinal cord, and has a risk of collapse.
- Mechanical, structural, and clinical changes may occur adjacent to a fusion mass (e.g., increased motion, degenerative changes, spinal stenosis and fracture dislocations, paradoxical motion, and restricted motion).
- Anterior keystone interbody fusion removes all or part of the intervertebral disc, exposes the midline portion of cord behind the vertebral body, can enlarge the neural foramen and elongate the yellow ligament, and provides a large surface area for contact. This construct in biomechanical tests did not extrude its graft before fracture and complete disruption of the spine. We suggest this procedure followed by the modified Bailey-Badgley construct as the surgical constructs of choice for multilevel anterior fusions.
- The notched fibula graft construct is designed to prevent extrusion and collapse. It is useful for multilevel fusions. The spiked horseshoe or keystone construct may be effectively used to replace an excised vertebral body.
- Posterior cervical spine fusion may have its immediate postoperative stability augmented by posterior wiring and the use of rib, fibula, or tibia bone graft.
- Posterior facet fusion is a very useful construct for situations in which immediate stability is important and the laminae have been removed.
- The use of rib grafts in an anterolateral or lateral trough provides a biomechanically sound construct for thoracic spine fusions associated with decompressions.
- Anterior interbody fusion of the lumbar spine is not often required. The benefits overshadow the liabilities only when the clinical goals cannot be achieved with a posterior arthrodesis.
- The anterolateral trough graft technique is a useful procedure for gaining some immediate postoperative stability in resisting sagittal plane translation.
- The interbody fusion can also be carried out from the posterior approach.
- Posterolateral fusion (transverse processes) is probably the most advantageous, and we recommend it as the construct of choice.
- With the use of sound biomechanical principles in conjunction with the proper implants and bone graft configurations, the vertebral column may be reconstructed following total spondylectomy.
- The progress of a fusion mass following surgery may be divided into four stages, which may be correlated with patient management and adjusted according to the unique considerations of each patient.

Surgical Constructs Employing Methylmethacrylate

- Methylmethacrylate has no adhesive qualities and forms a poor attachment to bone because of a distinct fibrous membrane that develops between it and the regional bone.
- Methylmethacrylate is most effective in resisting compressive loads. Its ability to resist tensile loads is improved by the incorporation of wire or wire mesh.
- Methylmethacrylate is used surgically in the spine as a spacer, an internal splint, or a fixation device.
- Biomechanical study of a methylmethacrylate surgical construct suggests that the weak link is the attachment of the cement to the bone.
- A conservative approach to the use of methylmethacrylate is recommended.
- A posterior construct is suggested for the achievement of maximum immediate clinical stability with methylmethacrylate. A modification using bone graft for long-range stability is also described.

Biomechanical Considerations in the Art and Science of Spine Instrumentation

- Implant testing is of considerable import and should be standardized as much as is feasible.
- *In vivo* biomechanical data are needed in or-

der to determine the optimum mechanical properties of the implant and its associated surgical construction.

- The biomechanical rationale should be critically reviewed for any given implant that the surgeon employs.
- When selecting wires, adequate tensile strength is the most important factor.
- Stainless steel mesh wire is recommended for use with methylmethacrylate that will undergo tensile loading.
- Wires under the lamina are nonanatomic, unphysiologic, and best avoided when other reasonable options are available.
- Screw fixation appears to be effective in cervical as well as lumbar spine fixations.
- Occipital cervical implants and C1–C2 posterior clamps may become some of the most useful cervical implants.
- The Gruca-Weiss springs have a modest correcting effect on a kyphotic curve and apply a significant amount of compression between the vertebrae that are spanned. They are of great interest because they represent a flexible form of spinal instrumentation.
- The Harrington compression rods have a net correctional influence on a kyphotic deformity through a three-point bending mechanism. This device is useful in the treatment of a clinically unstable kyphotic deformity.
- Distraction Harrington rods, because of their relative stiffness, are the most effective method of correcting post-traumatic kyphotic deformity when clinical stability is present.
- An analytical comparison of the two Harrington systems suggests that the distraction rod is the system of choice to correct a traumatic kyphotic deformity. However, in the presence of gross instability, the compression system provides some stability and is therefore preferable.
- It is sometimes advantageous to use a Harrington distraction rod in combination with a Harrington compression rod.
- A technique using paired Harrington distraction rods was shown to provide a mechanical advantage in kyphotic deformities that are difficult to correct.
- There are several implants that provide greater stiffness to a spine surgical construction than does the conventional Harrington distraction or compression system.
- These systems provide increased stiffness and stability and allow mobility and possible ambulation with no external immobilization. Some also exert more effective correctional forces.
- Rigid immobilizers include Cotrel-Dubousset, Steffee, AO fixation and other prototypes of pedicle fixation, Luque rods and rectangles, and spinous process wiring to Harrington rods, roughly in order of decreasing rigidity.
- The use of external pedicle fixation with altering rigidity as a pain-provocation test of instability is a milestone development that will help to clarify the biomechanics of clinical instability.
- The development and clinical use of a prosthetic lumbar disc like the preceding is a milestone development. At present, the mechanical design may be further advanced than the rationale and indications for its use.
- The appropriate amount of distraction is an important consideration in the management of thoracolumbar burst fractures.
- Crutchfield tongs are simple and easy to insert. They may pull out, especially if large traction forces of 20 kg or more must be applied.
- Gardner-Wells tongs are preferable from both the surgical and biomechanical standpoint.
- The halo apparatus, despite some important limitations, provides the best available mechanical control of the spine by external means. The screws should go into the outer table at 0.9 Nm (8 in lb) of torque.
- Axial traction is used in the spine to anchor, to gain alignment resisted by physiologic forces, or to correct resistant deformity. Guidelines for starting traction loads in different regions of the spine are provided.

Spine Osteotomies

- The basic spine osteotomy consists of removal of a wedge taken from the posterior elements; the apex of the wedge is at or near the posterior longitudinal ligament. A recently developed procedure that is theoretically better places the apex more anterior and resects the pedicles. In the frontal plane, the wedge is V-shaped. Beware the calcified aorta.
- In the cervical region, the osteotomy should include prophylactic decompression and postoperative halo fixation.
- There are significant advantages in the use of the side-lying position for lumbar spine osteotomy.

NOTES

[A] There is some confusion and disagreement about terminology with regard to laminectomy. We use laminotomy to indicate the removal of a portion of a given lamina. Complete laminectomy or bilateral laminectomy indicates removal of the spinous process and the entire lamina on each side of it. Hemilaminectomy or unilateral laminectomy indicates removal of the lamina on one side of the spinous process only.

[B] The biomechanical analyses are based upon the assumptions that the Harrington rods are not *permanently* bent during the operation and the compression rods do not have multiple hooks. The clinical studies, in which rods have been bent to accommodate a stiff curve or compression rods have been used with multiple hooks, may significantly differ from the conclusions of our analysis.

[C] This is a good construct designed by K. J. Keggi and associates; however, they now recommend something more similar to the construct shown in Figure 8-50, without steel mesh placed over the laminectomy site.

[D] Although canal size does not correlate with clinical pathophysiology as well as cross-sectional area of the dural sac, this should not be construed to mean that the canal size in the lumbar spine is not important. Studies have shown that disc disease is more common in those with small canals. Moreover, the margin of safety can be presumed greater than all the various potential encroaching factors in lumbar spinal stenosis when there is a normal or larger than average canal.

[E] *Experimental construct stability* is a term suggested by the authors to avoid confusion with the terms clinical stability, mechanical stability, and other uses of the term stability. Definition: the loss of the ability of an experimental spine implant construct to withstand the forces and displacements applied to it in a way that is not defined by the *experimental* protocol as failure.

Clinical construct stability is a term suggested by the authors and is defined as the ability of a spine implant construct to maintain its patterns of displacement in the clinical situation so as to avoid initial or additional neurologic deficit, severe deformity, or intractable pain.

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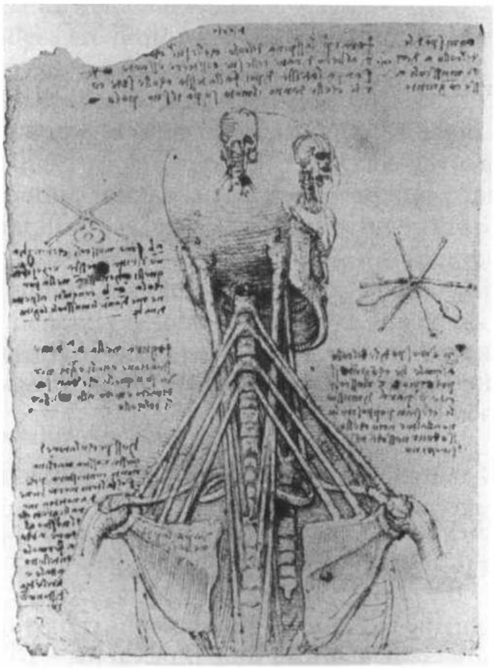
Biomechanics A to Z



O speculator on this machine of ours, let it not distress you that you give knowledge of it through anothers death, but rejoice that our Creator has placed the intellect on such a superb instrument.

Leonardo da Vinci (ca. 1500)

From Leonardo on the Human Body: Leonardo da Vinci. *Translations, text, and introduction by Charles D. O'Malley and J.B. de C.M. Saunders. New York, Dover Publications, Inc., 1983.*

9

This chapter contains most of the terms and engineering concepts that are applicable to orthopedic biomechanics. The authors feel that it provides a thorough understanding of the material presented in this book and of other literature on the subject.

The material here is presented in a way that is useful, understandable, and palatable to the clinician. The term is first defined in scientific prose, and units of measurement are given if applicable. This is followed by a description of the term, a familiar lay example, and an orthopedic example in most cases. Mathematical formulas are presented in the Explanatory Notes, along with an occasional discussion.

When units of measure are given, the new S.I. (*Systeme International d'Unites*) system has been adopted as a rule, while the presently used English/U.S.A. system is given in parentheses.

▶ Acceleration

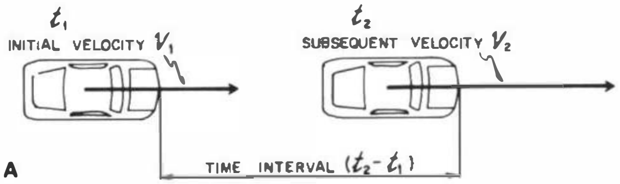
DEFINITION. The rate of change of linear velocity. The unit of measure of its magnitude is meters per second per second (feet per second per second).

DESCRIPTION AND EXAMPLES. Since acceleration is a vector quantity, changes in magnitude or direction may occur. When the driver of an automobile presses the accelerator and accelerates from a speed of 0 to 5 to 20 to 100 km/h (60 mph), the car undergoes linear acceleration. When the driver brakes, the car undergoes linear deceleration or retardation. These ideas are depicted in Figure 9-1A, and a mathematical derivation is given below.

Now, consider the rate of change of direction. A change in velocity direction with time without change in magnitude also produces acceleration. A passenger in a car taking a right turn is pushed to the left as the car negotiates the turn at a constant speed. This push is due to the change in the direction of the velocity vector. The concept is illustrated in Figure 9-1B, and the mathematical derivation is given below.

Generally, the term *acceleration* is used to represent linear acceleration. Another kind of acceleration is angular acceleration.

CHANGE IN VELOCITY MAGNITUDE



CHANGE IN VELOCITY DIRECTION

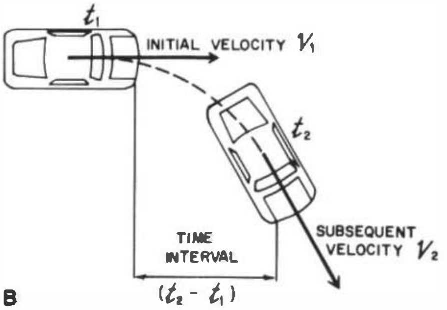


FIGURE 9-1 Linear acceleration (**A**) due to change in the magnitude (speed) of velocity and (**B**) due to change in the direction of velocity.

EXPLANATORY NOTES. First consider changes in magnitude (speed). If a body (automobile), shown in Figure 9-1A, at a certain point in time t_1 has speed V_1, and at another point in time t_2 has speed V_2, then the average acceleration is $(V_2 - V_1)/(t_2 - t_1)$. The instantaneous acceleration at time t is the average acceleration when the time interval $(t_2 - t_1)$ approaches zero. If the speed decreases during the time interval, there is deceleration or negative acceleration.

Referring to Figure 9-1B where the car is turning, the only change within time interval $(t_2 - t_1)$ is the change in the direction of the velocity vector from V_1 to V_2, the speed being the same. The acceleration is (vector V_2 − vector V_1)/$(t_2 - t_1)$. The resulting acceleration vector (see Vector) is directed toward the center of the turning circle of the car. In other words, to turn the car from the direction of vector V_1 to that of V_2, an acceleration directed toward the center of the circle must be applied. This is called centripetal acceleration. Because of the body inertia, the passenger feels a push directed opposite the direction of car acceleration (*i.e.*, away from the center). This push is called centrifugal force. The outward push in this case is similar to the backward push felt by a passenger sitting in a car that is accelerated forward.

▶ Allowable Stress

DEFINITION. A stress value that is higher than that due to the normal loads but is lower than the yield stress of the material. The unit of measure is newtons per square meter or pascals (pounds per square inch or psi). See Stress for its definition.

DESCRIPTION AND EXAMPLES. In designing structures for carrying mechanical loads it is necessary to allow for not only normal loads but also dynamic loads, accidental overloads, inaccuracies in material and workmanship, and other unknown variables. For these reasons, a margin of safety is generally provided by choosing a design or allowable stress much below the yield point (see *Yield Stress*) so that no permanent deformation can take place as a result of these unwelcome loads.

Bridges are built with allowable stress. They are built to carry greater loads than those to which they are expected to be subjected.

The same is true of the human skeleton. Our bones tolerate a broad range of physiologic loading. The routine human activity of walking, running, and jumping may be thought of as being in the physiologic range. When the pole vaulter, ski jumper, or paratrooper has an imperfect fall and does not break or permanently deform his skeleton, the bones have been overloaded but have stayed within the range of allowable stress.

▶ Angular Acceleration

DEFINITION. The rate of change of angular velocity. Since acceleration is a vector quantity, changes in magnitude or direction may occur. The unit of measure of its magnitude is radians per second per second (degrees per second per second).

DESCRIPTION AND EXAMPLES. The change in the angular velocity with time constitutes angular acceleration. In whiplash injury, when an auto is hit from behind, the trunk is linearly accelerated forward in relation to the head, which, because of its inertia, is slow to respond and is therefore angularly accelerated backward. Thus, the mechanism of injury is dependent upon the angular acceleration of the head as well as its inertia (see *Inertia*).

Because the angular velocity is a vector, a change in its direction with time without a change in its magnitude also produces acceleration. A gyroscope, as shown in Figure 9-2, consists of a heavy wheel rotating at a high speed. It is able to balance on the tip of a pen because if it tilts, its angular velocity vector changes direction, thus producing angular acceleration. This, by Newton's second law of motion, develops a counterbalancing moment that tends to bring the gyroscope back to its original position. Through this mechanism, a stable position is maintained as long as the gyroscope rotates. This gyroscopic "trick" intrigues the imagination because angular acceleration due to change in direction is not a part of everyday experience. However, the gyroscope obeys the same laws of mechanics as does a car being hit from behind.

FIGURE 9-2 Angular acceleration. A gyroscope balances on the tip of the pen because of the high rotatory speed of its wheel. If it tilts to the side, an angular acceleration and a moment develop, restoring it back to its original position.

EXPLANATORY NOTES. A mathematical expression for the angular acceleration can be derived. Consider a body, at a certain point in time t_1, that has angular speed W_1, and at another point in time t_2 has angular speed W_2. Then the average angular acceleration is $(W_2 - W_1)/(t_2 - t_1)$. The instantaneous acceleration at time t is the average acceleration when the time interval $(t_2 - t_1)$ approaches zero. If angular speed decreases during the time interval, the acceleration is negative and is called angular deceleration.

▶ Anisotropic Material

DEFINITION. A material is anisotropic if its mechanical properties vary with different spatial orientations.

DESCRIPTION AND EXAMPLES. If one takes a test sample of an anisotropic material, its mechanical properties such as strength and elasticity will vary according to relative orientation within the material. Some examples of anisotropic materials are wood, bone, ligaments, and cartilage.

Take out a cubic specimen of cancellous bone from a vertebra (Fig. 9-3). It is first loaded in an axial

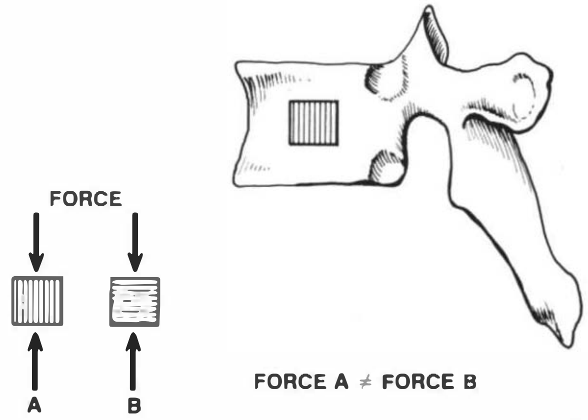


FIGURE 9-3 Anisotropic material.

direction (A), and then in a transverse direction (B). If the specimen is shown to be stronger or weaker during axial loading than during transverse loading, the anisotropic quality of the bone has then been demonstrated.

▶ Axis of Rotation

See *Instantaneous Axis of Rotation.*

▶ Bending

DEFINITION. When a load is applied to a long structure that is not directly supported at the point of application of the load, the structure deforms, and this deformation is called bending.

DESCRIPTION AND EXAMPLES. If a plastic ruler is bent as shown in Figures 9-4A and 9-4B, it is apparent that the ruler is stiffer and stronger when loaded as shown in Figure 9-4B than when loaded in the manner shown in Figure 9-4A. This is due to the fact that the material is further away from the center with respect to the bending mode in B (*see Sectional Moment of Inertia*).

The vertebral pedicle has a cross-section that is especially suitable for taking up bending loads in the sagittal plane. The moment of inertia of an elliptical cross-section is greatest for the bending loads in the direction parallel to its major axis, as shown on the right in Figure 9-4C. This is probably the reason that the pedicle cross-section is elliptical, with its long axis vertical as shown in Figure 9-4D. In other words, the structural design of the vertebra is capable of best resisting bending loads in the direction in which those loads are likely to be greatest.

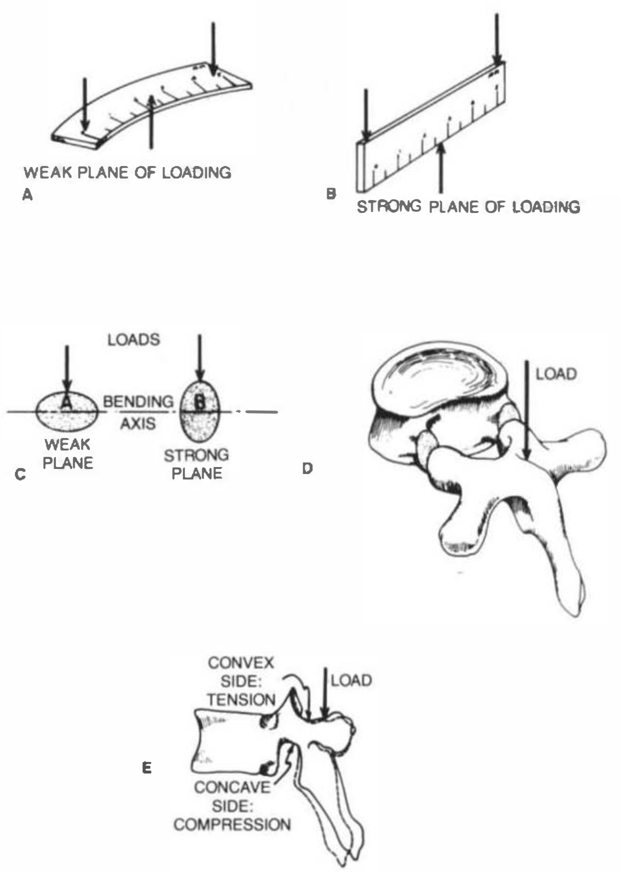


FIGURE 9-4 Bending. Resistance of a structure to bending loads is dependent upon its cross-sectional geometry in the plane of bending. **(A)** Weak plane of a ruler. **(B)** Strong plane of a ruler. **(C)** Weak and strong planes of a pedicle. **(D)** Pedicle cross-section is oriented to provide strong bending plane in the sagittal plane. **(E)** Sagittal plane bending results in tensile stresses on the convex side of bending and in compressive stresses on the concave side.

EXPLANATORY NOTES. Fibers on the concave side of the bent structure are compressed, while those on the convex side are elongated. Figure 9-4E shows a vertebra being loaded just posterior to the facets. The amount of fiber stress σ (sigma) is given by the following formula:

$$\sigma = \frac{M \times Y}{I}$$

where M = bending moment; Y = fiber distance from the neutral axis; and I = sectional moment of inertia. Note that the stress is not dependent upon material properties, such as the modulus of elasticity.

The radius of curvature R of the bent structure is given by another equation:

$$R = \frac{E \times I}{M}$$

where E = modulus of elasticity of the material.

▶ Bending Moment

DEFINITION. A quantity at a point in a structure equal to the product of the force applied and its lever arm (the perpendicular distance from the point to the force direction). The unit of measure is newton meters (foot poundforce).

DESCRIPTION AND EXAMPLES. A monkey sitting on a tree branch is subjecting the various sections of the branch to bending moments (Fig. 9-5). The bending moment changes in magnitude from zero under his seat to the maximum at the junction of the branch and the trunk of the tree. The same concept applies to the weight of the branch itself. The natural structure of the branch enables it to resist the progressively higher moments created at various sections from the tip to the base of the branch by the distribution of correspondingly more material with a larger sectional moment of inertia of the cross-sections.

It has been shown by *in vivo* disc pressure measurements that the disc pressure and the axial disc load in the lumbar region increase when a sitting subject lifts a telephone that is a fair distance away.[2] (In separate cadaver experiments, the disc pressure has been shown to be directly related to axial spinal load.) The mechanism of the increase of the axial load is as follows. The small weight of the telephone applies a substantial bending moment at the disc because of the large lever arm. This bending moment is counterbalanced by the bending moment provided by muscle and ligamentous forces, which have a much smaller lever arm and therefore must exert forces of very large magnitudes in order to maintain the equilibrium. It is this large muscle force that accounts for the large axial load and pressure in the disc.

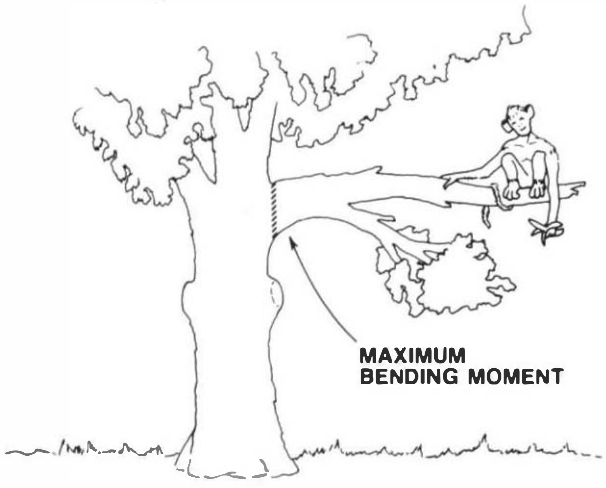


FIGURE 9-5 Bending moment. The thickness of the branch at any section is related to the bending moment at that section.

For this reason it is important to hold a weight as close to the body as possible.

▶ Bending Moment Diagram

DEFINITION. A diagram showing the amount of bending moment at various sections of a long structure subjected to bending loads.

DESCRIPTION AND EXAMPLES. If one knows the bending moment diagram and the dimensions of the structure and its material properties, it is possible to compute the normal stress, the shear stress, angulation, and deflection at every point of the structure when it is subjected to a given set of loads.

Figure 9-6A shows a portion of the spine in three-point bending. The three forces are F_1, F_2, and F_3. Figure 9-6B is the bending moment diagram. The shape of the diagram shows that the maximum bending moment will occur under the force F_1. Assuming that the spine structure and its material have properties that are the same along its entire length and that the weights of the vertebrae are negligible compared with the force F_1, the point under F_1 will be the point of highest stress and failure.

A bending moment diagram for any given load situation can be obtained by the simple method described below.

EXPLANATORY NOTES. The shape of the bending moment diagram for a given set of loading situations may be determined by the following procedure. Referring to Figure 9-6A, at a point X on the spine at distance A_x from the left support, the bending moment is:

$$\begin{aligned} M_x &= \text{force} \times \text{lever arm} \\ &= F_2 \times A_x \end{aligned}$$

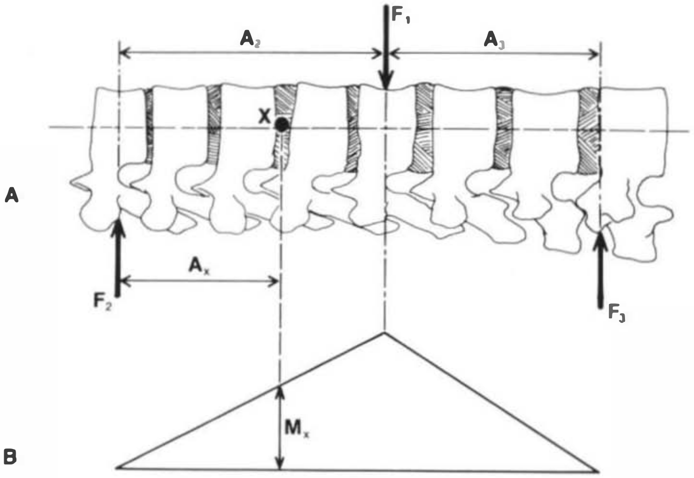


FIGURE 9-6 Bending moment diagram. **(A)** A spine specimen subjected to bending. **(B)** The bending moment at any section of the spine equals the height of the diagram at that section.

This is the height of the bending moment diagram under the point X (Fig. 9-6B). The complete bending moment diagram is obtained by moving point X from the left support to the right support and taking moments of all the forces to the *left of point X*. For a three-point bending load, the bending moment diagram is a triangle with its apex under the middle force. For other kinds of loads the bending moment diagram would have other shapes.

The highest bending moment under the force F_1 is as follows:

$$M_{max} = \frac{F_1 \times A_2 \times A_3}{A_2 + A_3}$$

▶ Biomechanical Adaptation

DEFINITION. Biologically mediated changes in the mechanical properties of tissues (material properties and/or structural changes) in association with the application of mechanical changes to those tissues.

DESCRIPTION AND EXAMPLES. A simple and fairly universal example of biomechanical adaptation is the common foot callus. When the feet are subjected to significant loads over normal or abnormal prominences, the skin and the subcutaneous tissues become harder and thicker, a material and a structural change, respectively.

Another example, demonstrating structural changes, is shown in Figure 9-7. Part A shows the lateral view of a normal ankle joint. In Part B, observe the build-up of a large triangular segment of bone at the distal anterior tibial eminence. This is a biologically mediated change in the ankle associated with the repeatedly applied forces and deformation in that area that are generated by the "push-off" activities of the athlete.

Wolff's law describes a type of biomechanical adaptation.

▶ Buckling

DEFINITION. A structure is said to buckle under axial load if there is a sudden "give" or lateral deformation of the structure.

DESCRIPTION AND EXAMPLES. Imagine a vertical rod of uniform cross-section fixed at its base (Fig. 9-8A). If the rod is centrally loaded by a small force, it will remain straight and shorten by a small amount. Now let the force be continuously increased. A point will be reached when the rod will suddenly deform (give way) laterally (Fig. 9-8B). This is the phenomenon of buckling, and the axial force at the time of buckling is called the buckling or critical load. This is to be distinguished from bending, which may be caused by an eccentrically applied force (Fig. 9-8C). In the case of bending, the rod will start bending immediately after the application of the force. There is no sudden buckling or giving way.

The buckling of a long structure was critically analyzed by Euler in the 18th century. He derived mathematical formulas that computed the buckling load (also called critical load). It was found to depend upon several factors, namely, length of the rod, its cross-sectional area, its sectional moment of inertia, and modulus of elasticity of the material of the rod. See *Elastic Stability* for further discussion.

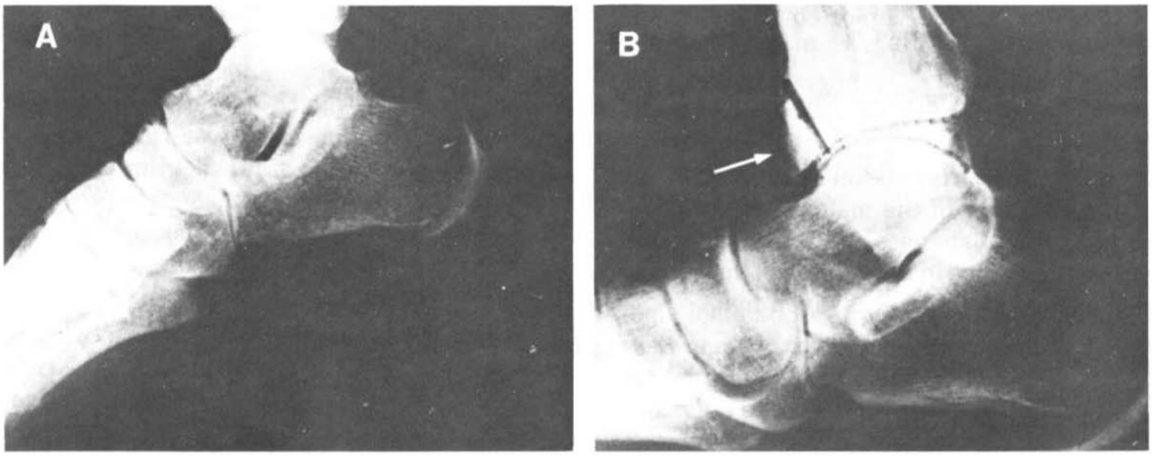


FIGURE 9-7 **(A, B)** Biomechanical adaptation. Notice the additional bone (*arrow*) in **B.** (*Courtesy of James Nicholas, M.D., New York, N.Y.*)

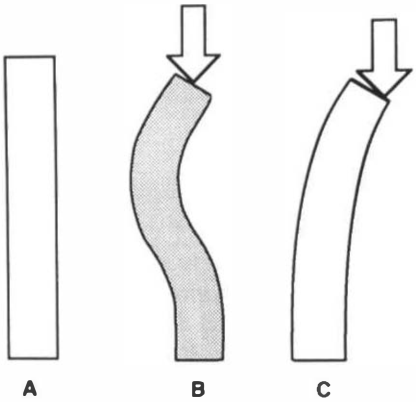


FIGURE 9-8 Buckling. A vertical rod of uniform cross-section is subjected to a vertical load. **(A)** Rod unloaded. **(B)** Rod loaded with a force applied at the center. The rod buckles when the applied force F equals or exceeds the critical load value for the rod. **(C)** Rod loaded with a force applied off center. This results in bending of the rod, irrespective of the magnitude of the applied force F.

▶ Bulk Modulus

DEFINITION. A ratio of hydrostatic stress to volumetric strain. Its unit of measure is newtons per square meter, N/m^2, or Pa (pound force per square inch or psi).

DESCRIPTION AND EXAMPLES. Consider a cube, 1 cm each side, made of rubber and lying on a table. We will use this cube as an example to define volumetric strain and hydrostatic stress. Now take this cube under water in a swimming pool to a depth of 10 m. Because of the hydrostatic pressure due to the water surrounding the cube on all sides, the cube will decrease in size (all three dimensions), although its shape will remain a cube. The change in volume divided by the original volume of the cube is called the volumetric strain. The hydrostatic stress acting on the cube is equal to the weight of the column of water above the face of the cube divided by the area of the face.

There are several examples in which hydrostatic stress is present. Compressive hydrostatic stresses exist in a rock at the bottom of the ocean or far under the earth's surface. Tensile hydrostatic stresses will exist in a tennis ball when placed inside a vacuum chamber. The same is true for the inside of a cube of steel that is heated uniformly all over its surface. See also *Strain* and *Stress*.

EXPLANATORY NOTES. The bulk modulus, generally represented by the letter K, by definition is given as:

$$K = \frac{\text{hydrostatic stress}}{\text{volumetric strain}} \ N/m^2$$

The bulk modulus, although a distinct quantity, is not an independent material parameter. It is dependent upon two fundamental characteristics of the material, namely, the modulus of elasticity and Poisson's ratio. If one knows these two material parameters, the bulk modulus of a material can be computed.[19] Without providing any derivations, we state that the strain in each of the three perpendicular directions of a cube subjected to hydrostatic stress is:

$$\varepsilon = \frac{\sigma(1 - 2\nu)}{E}$$

where σ is the stress, ν is Poisson's ratio, and E is the modulus of elasticity of the material of the cube.

Neglecting small changes, the volumetric strain (*i.e.*, the overall change in the volume of the cube as a ratio of the original volume) is given by:

$$1 + 3\varepsilon$$

Thus, by definition, the bulk modulus is given by:

$$K = \frac{E}{3(1 - 2} \text{ N/m}^2 \text{ or Pa}$$

For most materials, Poisson's ratio is 0.3, giving us the value for the bulk modulus as:

$$K = 0.83E \text{ Pa}$$

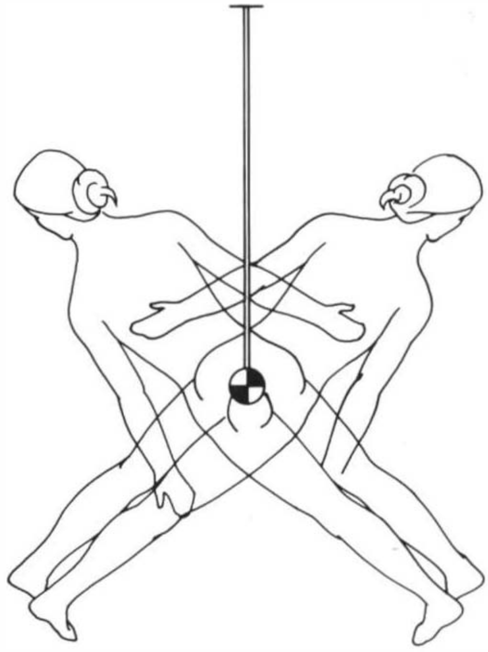


FIGURE 9-9 Center of gravity. A body suspended from its center of gravity may be oriented in any direction.

Center of Gravity

DEFINITION. The point in a body at which the body mass is centered.

DESCRIPTION AND EXAMPLES. If the body were hung from this point by a rope (Fig. 9-9), the body could then be oriented in any direction whatsoever, and it would remain in that orientation hanging freely.

The center of gravity of the body lies in the midsagittal plane (due to anatomic symmetry) and somewhat anterior to the upper sacral spine. It is reported to be 4 cm in front of the first sacral vertebra in the standing anatomic position. This probably is the reason for carrying a backpack on the back, which tends to bring the center of gravity more in line with the spine, thus reducing the bending stresses. It must be realized that the center of gravity is different for different body postures. The center of gravity refers to three-dimensional bodies (e.g., the entire human body or a single vertebra). An equivalent concept for areas is described under *Centroid*. The principle behind the concept of the center of gravity is further explained below.

EXPLANATORY NOTES. At the center of gravity, the sum of the moments due to weights of all the parts constituting the body is equal to zero. Therefore, when a body is hung from its center of gravity, the moments due to the body parts on the right-hand side of the center of gravity are exactly equal and opposite to those exerted by the body parts on the left-hand side. Hence, there is zero moment at the point of hanging and therefore also no tendency for the body to rotate.

Center of Mass

See *Center of Gravity.*

Center of Motion

See *Instantaneous Axis of Rotation.*

Centroid

DEFINITION. The centroid of an area is a point on which the total area may be centered.

DESCRIPTION AND EXAMPLES. One way to approximate the centroid of a given area is to do the following experiment. Draw the area whose centroid is required on a piece of thick paper and cut it out. The center of gravity of this piece of paper is the centroid of the area. To find the center of gravity, choose a point on the paper and hang the paper by a thread from this point. Orient the paper in an arbitrary plane and let it go. If it can maintain that orientation when hanging freely, then that point is the center of gravity for this particular piece of paper and the centroid for the section. Several trials may be required to find the right point. Results of this experiment can be obtained mathematically if the boundary can be described mathematically. The formulas are given below.

The centroid of a section is required, among other things, to determine bending strength of structures and other related items, such as the neutral axis and the sectional moment of inertia. The centroids of some simple cross-sections are shown in Figure 9-10A.

EXPLANATORY NOTES. In mathematical terms, the centroid of an area may be obtained in the following manner. Choose a stainless steel fixation plate, as shown in Figure 9-10B, as an example. The area to be analyzed is shown in an enlarged view in Figure 9-10C. The area is given by the integral:

$$A = \int y \, dx$$

where y and dx refer to a small strip of the area as seen in Figure 9-9C. The coordinates of the centroid of the area, C_x and C_y, are given by equations:

$$C_x = \int \frac{xy \, dx}{A}$$

$$C_y = \int \frac{xy \, dy}{A}$$

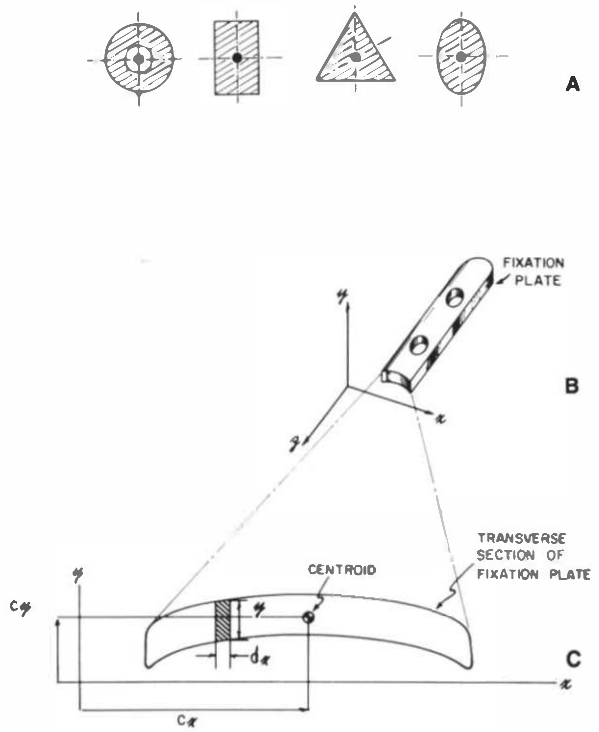


FIGURE 9-10 Centroid. **(A)** Centroids of some common sections. **(B)** A fracture fixation plate. **(C)** A close-up view of the plate section showing the method of computing the location of the centroid.

▶ Clinical Stability

DEFINITION. The ability of the spine under physiologic loads to limit patterns of displacement so as not to damage or irritate the spinal cord or nerve roots and, in addition, to prevent incapacitating deformity or pain due to structural changes.

DESCRIPTION AND EXAMPLES. Any disruption of the spinal components (ligaments, discs, facets) holding the spine together will decrease the clinical stability of the spine. When the spine loses enough of these components to prevent it from adequately providing the mechanical function of protection, surgical or other measures are taken to reestablish stability.

▶ Coefficient of Friction

DEFINITION. The ratio of tangential force to the normal interbody compressive force required to initiate a sliding motion between two bodies. This ratio has no units of measure.

DESCRIPTION AND EXAMPLES. A skater glides effortlessly on ice (Fig. 9-11A). The ratio of her effort (tangential force) to her body weight (normal force) is very small; thus, this interbody action has a rather low coefficient of friction. A boulder sitting on the road, on the contrary, requires considerable effort to

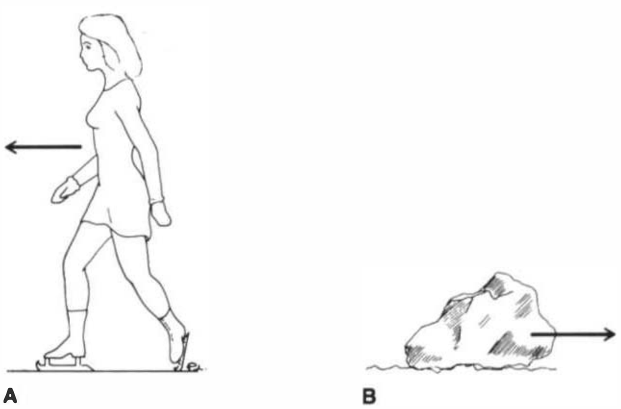


FIGURE 9-11 **(A)** A skater on ice has a low coefficient of friction. **(B)** In contrast, a boulder on the road has a high coefficient.

move it, denoting a high coefficient of friction between it and the road (Fig. 9-11B).

Some typical values of the coefficient of friction (static) for different contacting surfaces are as follows:

Stone on ground	0.75
High-density polyethylene against polished steel with lubricant*	0.1
Steel on ice	0.01
Bearing with lubrication	0.01
Animal joint	0.005

For further discussion, see *Joint Reaction Force*.

▶ Compression

DEFINITION. The normal force that tends to push together material fibers. The unit of measure is newtons (poundforce).

DESCRIPTION AND EXAMPLES. The weight of a building applies compression to its foundation.

The intervertebral disc is the main compression-carrying component in the spine. It is subjected to direct compression, even when a person is not carrying any loads. This compression is due to several causes: direct weight of the trunk, initial tension in other ligaments (e.g., ligamentum flavum), and additional tension in ligaments and muscles required to balance the eccentric trunk weight.

▶ Compressive Stress

See *Stress*.

▶ Conversion Table

Table 9-1 gives conversion factors for entities specified in the presently used English/U.S.A. system and the new S.I. (*Systeme International d'Unites*) system.

DESCRIPTION AND EXAMPLES. To obtain the English/U.S.A. measurement when the S.I. unit is given, multiply the S.I. quantity by the factor X. Use the factor Y instead to convert from the U.S.A. to the S.I. system. Examples are given below. The symbols used in the table are as follows:

degree = deg
foot = ft
inch = in
joule = J
kilogram = kg
meter = m
newton = N
pascal = Pa
pound = lb
poundforce = lbf
radian = rad
second = s

TABLE 9–1 Conversion Factors

Entity	S.I.	→X→ ←Y←	U.S.A.
Acceleration	m/s^{2}	3.2808 0.3048	ft/s^{2}
Angle	rad	57.296 0.0175	deg
Area	m^{2}	1550.0 0.000645	in^{2}
Density	kg/m^{3}	0.0000361 27680.	lb/in^{3}
Energy, work	Nm = J*	0.7376 1.3558	lbf ft
Force	N	0.2248 4.4482	lbf
Length	m	3.2808 0.3048	ft
	m	39.370 0.0254	in
Mass	kg	2.2046 0.4536	lb
Mass moment of inertia	kg m^{2}	23.730 0.0421	lb ft^{2}
Moment, torque	Nm	0.7376 1.3557	lbf ft
Polar moment of inertia, section moment of inertia	m^{4}	115.86 0.0086	ft^{4}
Pressure	N/m^{2} = Pa*	0.000145 6896.5	lbf/in^{2}
	Pa	0.0075 133.4	mm Hg
Sectional moment of inertia (area moment of inertia)	m^{4}	115.86 0.0086	ft^{4}
Stiffness	N/m	5.667 0.177	lbf/in
Stress (see Pressure)			
Velocity	m/s	3.2808 0.3048	ft/s
Volume	m^{3}	35.313 0.0283	ft^{3}
	m^{3}	61023. 0.0000164	in^{3}

* Officially recommended units of S.I.

Two examples are given below: (1) to convert moment from the U.S. to the S.I. system; and (2) to convert pressure from S.I. units to U.S. units.

Moment: $100 \text{ lbf ft} = 1.3557 \times 100 = 135.57 \text{ Nm}$

Pressure: $100 \text{ Pa} = 100 \times 0.000145$
$= 0.0145 \text{ lbf/in}^2$

▶ Coordinate Systems

DEFINITION. Reference systems that make it possible to define position and motion of rigid bodies in space or with respect to each other.

DESCRIPTION AND EXAMPLES. The motion of a body may be determined by knowing its position before and after a given time interval. The three-dimensional description of motion of an object requires a three-dimensional coordinate system. There are many types of coordinate systems available, but the following three are probably the most widely used: the cylindrical, the spherical, and the rectangular systems. The choice of a particular coordinate system depends upon the convenience it offers.

The cylindrical coordinate system is used for objects or motions with some circular symmetry about an axis. An egg has an axis of revolution. Any point P on its surface may advantageously be represented by cylindrical coordinates: r, θ, y (Fig. 9-12A).

The spherical system is preferable for situations in which spherical symmetry may be present. To define a point on earth that resembles a sphere, the spherical system is most convenient. The radius of the earth, and the longitude and latitude angles are the three required coordinates: r, θ, and ϕ, respectively (Fig. 9-12B).

The musculoskeletal components, in general, have no plane of symmetry. The rectangular coordinate system is most convenient here. The right-handed Cartesian orthogonal coordinate system, the proper name for the rectangular system most preferred, is defined as a system consisting of three

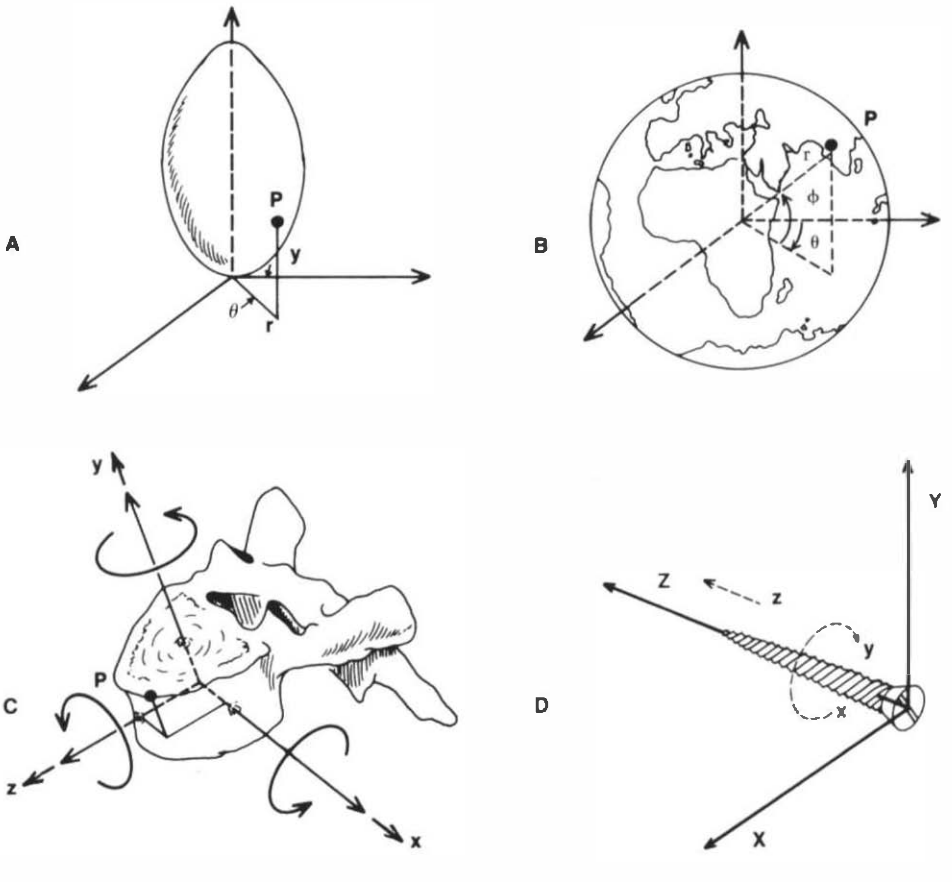


FIGURE 9-12
Coordinate systems.
(A) Cylindrical, P = (r, θ, y).
(B) Spherical, P = (r, θ, ϕ).
(C) Cartesian, P = (x, y, z).
(D) A screw, by its movement, defines the right-handed coordinate system.

straight lines mutually perpendicular and intersecting. These lines, called the axes, may be named x, y, and z. The point of intersection is called the origin.

A vertebra with the origin of the Cartesian coordinate system placed at the center is shown in Figure 9-12C. To define the mutual direction of the axes, imagine an ordinary (right-handed) screw placed along the z-axis with its tip pointing toward the +z-axis. Then rotation of the screw head from the +x-axis to the +y-axis will produce screw translation in the positive direction of the z-axis (Fig. 9-12D).

This right-handed system as opposed to the left-handed system is universally preferred by convention. The senses of motion are also defined by convention. Figure 9-12C shows positive translations and rotations about the three axes. Translatory motion along an axis toward its positive direction is called positive, while in the opposite direction it is called negative. A clockwise rotation about an axis, looking from the origin of the coordinate system toward the positive direction of the axis, is called positive rotation, while the counterclockwise rotation is termed negative.

An ordinary screw (with right-hand threads) will translate along the positive z-axis, when placed as shown in Figure 9-12D and turned clockwise. This is called the *right-hand screw rule*. It defines the positive direction of the rotation around an axis.

Recommendations have been offered for a standard use of the Cartesian coordinate system in the human body.[13,26]

▶ Couple

DEFINITION. A pair of equal and opposite parallel forces acting on a body and separated by a distance. The moment or torque of a couple is defined as a quantity equal to the product of one of the forces and the perpendicular distance between the forces. The unit of measure for the torque is the newton meter (foot poundforce).

DESCRIPTION AND EXAMPLES. A couple (of forces) is applied to the steering wheel of a car when it is turned (Fig. 9-13). This pair of equal and opposite forces creates a torque that turns the steering shaft.

EXPLANATORY NOTES. In the case of the steering wheel, the torque T is given by the following equation:

$$T = F \times D$$

where F is the force in newtons (poundforce) and D is the perpendicular distance in meters (feet).

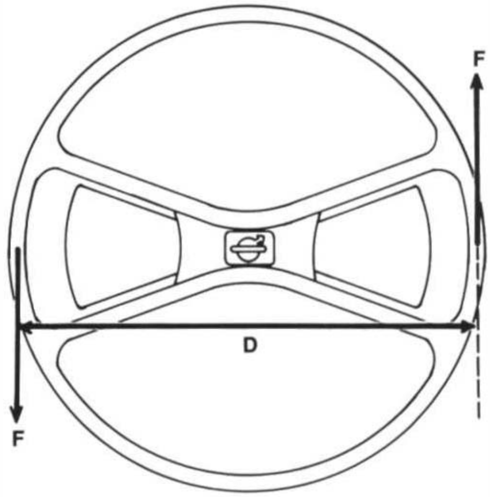


FIGURE 9-13 Couple. Two parallel, equal, and opposite forces *F* separated by distance *D* produce torque F × D.

▶ Coupling

DEFINITION. A phenomenon of consistent association of one motion (translation or rotation) about an axis with another motion about a second axis.

DESCRIPTION AND EXAMPLES. Vertebral motion, both in and out of the sagittal plane, produces other associated motions of translation and rotation.[11] Anterior translation of a vertebra produced by force F is always associated with flexion rotation (Fig. 9-14A). Similarly, axial rotation produced by axial moment M is consistently associated with lateral bending (Fig. 9-14B).

In scoliosis, lateral deformity is coupled with axial rotation, such that the posterior elements tend to rotate toward the concavity of the curve.[25]

▶ Creep

DEFINITION. A viscoelastic material deforms with time when it is subjected to a constant, suddenly applied load. The deformation–time curve approaches a steady-state value asymptotically. This phenomenon is called creep.

DESCRIPTION AND EXAMPLES. When an individual's height is measured in the morning and again at night after standing all day, the second measurement is found to be less than the first. The change in height (deformation) is due not to additional weight the person has gained but rather to creep. The same load applied over a period of time has caused a subsequent deformation and loss of height. This

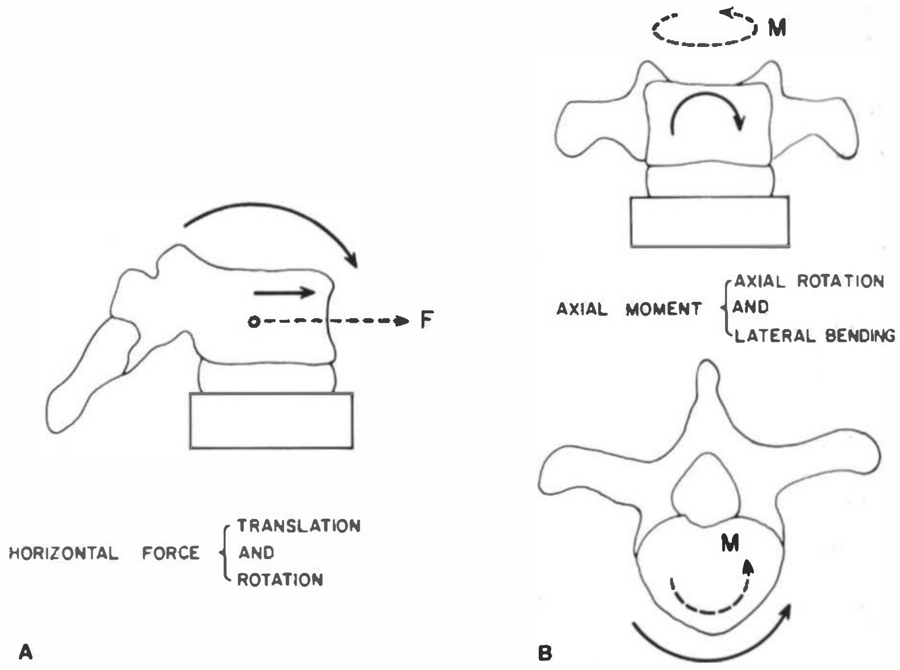


FIGURE 9-14 Coupling. **(A)** Anterior horizontal force *F* produces translation (main motion) and rotation (coupled motion). **(B)** Axial torque or moment *M* produces axial rotation (main motion) and lateral rotation (coupled motion).

time-dependent decrease in height is the result of creep due to viscoelastic properties of the disc.

In Figure 9-15, the creep test is performed on a functional spinal unit (FSU). On the left, the FSU is shown without load. In the middle, a sudden tensile load is applied, producing immediate deformation (see the deformation–time diagram). On the right, the same FSU is shown 1 hour later. Additional deformation has taken place within this time. Results of the creep test are plotted as a deformation–time curve. This is an important mechanical characteristic of the spine and other biologic structures.

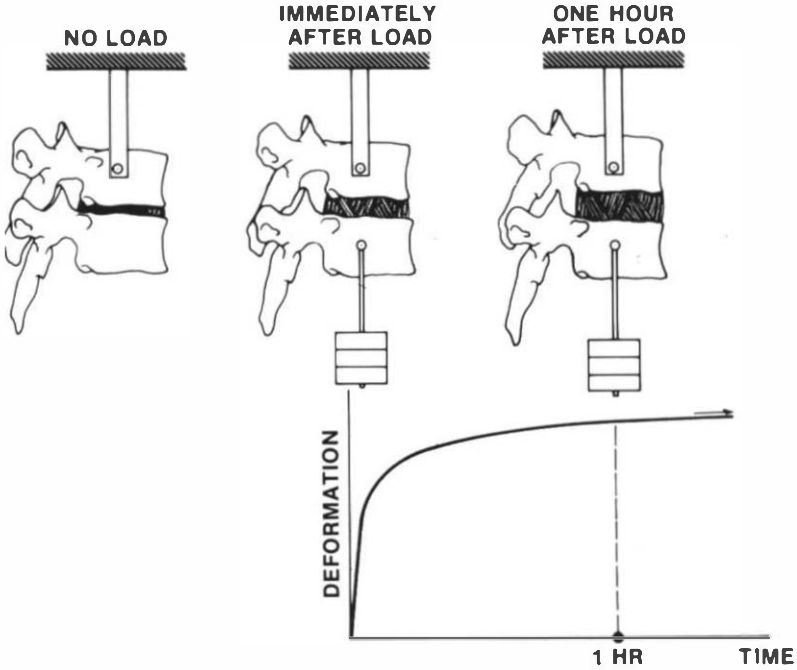


FIGURE 9-15 Creep. A deformation–time curve quantifies creep, after a constant force is applied and maintained.

The creep phenomenon in Figure 9-15 is the result of tensile loading, whereas the previous example was the result of compressive loading. However, both demonstrate the viscoelastic creep.

▶ Critical Load

See *Elastic Stability.*

▶ Cylindrical Coordinates

See *Coordinate System.*

▶ Damping

DEFINITION. A material property that constitutes resistance to speed.

DESCRIPTION AND EXAMPLES. To visualize the damping effect, consider a syringe (Fig. 9-16). A certain force applied to the plunger gives the plunger a certain speed. A slow movement of the plunger requires a smaller force, while a fast movement requires a considerable force. Figure 9-16 also shows the force–speed curve for the syringe. The slope of the curve is called the damping coefficient.

The shock absorber in a car uses the damping effect of the "fluid in a syringe" to smooth out the sharp vibrations of the wheels on a rough road and provide a smooth ride.

In engineering, the phenomenon of damping is represented by a mathematical model called the dashpot (see *Dashpot–Mathematical Element*). All biologic structures (bone, ligaments, joints, and the spine) exhibit damping properties in the form of viscoelastic behavior.

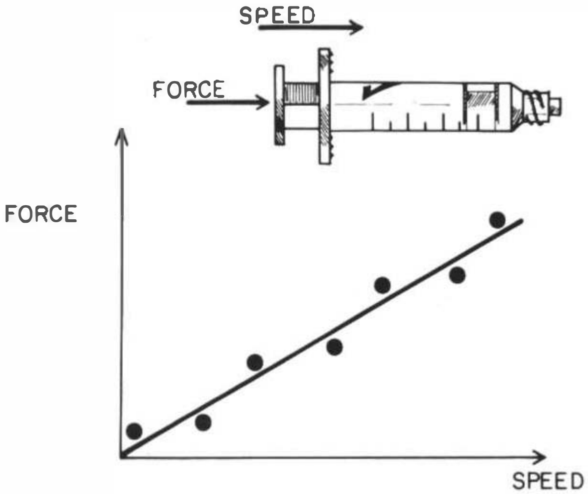


FIGURE 9-16 Damping. A force–speed curve quantifies damping.

EXPLANATORY NOTES. When the resistance offered by damping is proportional to the speed, it is called viscous damping. If a force is applied to deform a structure, the ratio of force exerted to the deformation speed produced is the measure of damping and is called the damping coefficient. The units of measure are newton second per meter (poundforce second per foot) for translatory motion and newton second per radian (poundforce second per degree) for rotatory motion.

▶ Dashpot–Mathematical Element

DEFINITION. A component used in building mathematical models of structures or materials that exhibit time-dependent viscoelastic behavior.

DESCRIPTION AND EXAMPLES. A boat on water symbolizes this element. The difficulty with which one must push or pull a boat across the surface of the water is dependent upon the speed of the movement. This phenomenon is due to the viscosity of the water.

The intervertebral disc has strong damping properties and is sometimes referred to as the shock absorber of the spine. Sudden motions of the lower part of the body are attenuated by viscera, skin, bones, discs, and vertebral bodies before reaching the head.

Most probably, the blood in the vertebral capillaries and sinusoids also offers resistance to deformation, thus acting as a dashpot. In rapid rates of loading, the blood cannot escape through the foramina rapidly enough and thus provides resistance. The system is viscoelastic. With lower loading rates, the blood offers much less resistance and the system is nearly elastic.

EXPLANATORY NOTES. The damping properties of a tissue or a system represented by the dashpot–mathematical element can be quantified. The damping characteristics of the tissue are quantified by the relationship between the load applied and the speed produced (Fig. 9-17). The slope of the load–speed curve is called the damping coefficient. The coefficient that varies with the load characterizes nonlinear damping. The area under the load–speed curve represents the rate of energy loss, usually in the form of heat, during the loading/unloading cycle. Also shown in Figure 9-17 is the dashpot symbol: a piston pushing on the fluid contained in a cylinder.

Also see Hysteresis.

▶ Deceleration

See *Acceleration* and *Angular Acceleration.*

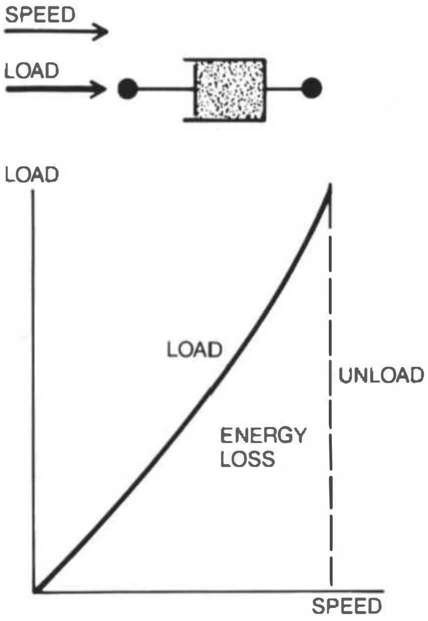


FIGURE 9-17 Dashpot. Time-dependent behavior is modeled by a dashpot. Resistance is created due to leakage of the fluid around the piston. The resistance (*Load*) offered by the dashpot is directly related to the speed.

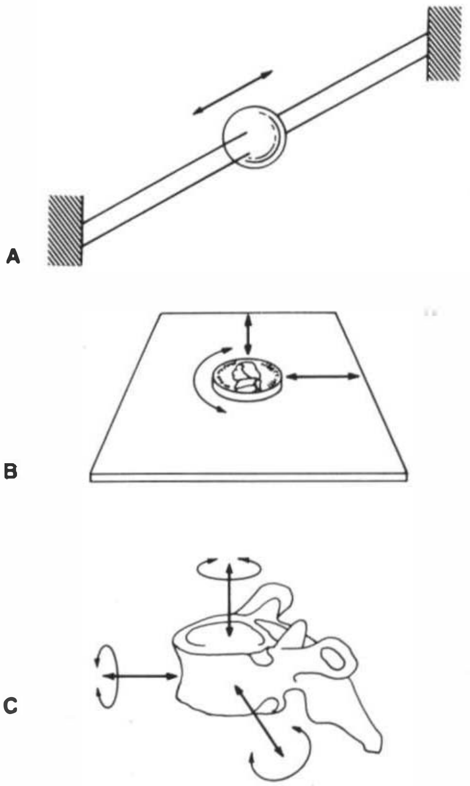


FIGURE 9-18 Degrees of freedom. (A) A bead on a track has one degree of freedom. (B) A nickel moving freely on a table has three degrees of freedom. (C) A vertebra is capable of all six degrees of freedom.

▶ Deformation

DEFINITION. The change in length or shape. Deformation is generally represented in the form of strain (see *Strain*).

▶ Degrees of Freedom

DEFINITION. The number of independent coordinates, in a coordinate system, needed to completely specify the position of an object in space.

DESCRIPTION AND EXAMPLES. The term is loosely applied to specify the independent motion components that are involved in the characteristic movements of a given rigid body. The motion of a rigid body in space has six degrees of freedom—three translations (expressed by linear coordinates) and three rotations (expressed by angular coordinates). When bodies are interconnected in a system, certain constraints are placed on the possible motions, and the number of degrees of freedom decreases. (See *Translation* and *Rotation*.)

A bead on tracks has a single degree of freedom, as shown in Figure 9-18A. A body moving in a plane has three degrees of freedom, two translations along mutually perpendicular directions in the plane and one rotation around an axis perpendicular to that plane. An example of such a body movement is a coin moving freely on a table (Fig. 9-18B).

A body has six degrees of freedom if it is allowed to move freely in three-dimensional space. The vertebra (Fig. 9-18C) is capable of performing all the six motions in space as the trunk is manipulated with respect to the pelvis. Thus, it has six degrees of freedom.

EXPLANATORY NOTES. It should be pointed out that although only three coordinates are required to completely define a point in space (see *Coordinate System*) a minimum of six coordinates is needed to specify position of a rigid body.

▶ Dry Friction–Mathematical Element

DEFINITION. A component used in building mathematical models of structures or materials that exhibit plastic behavior.

DESCRIPTION AND EXAMPLES. One cannot move a heavy anatomy book lying on a table by just blowing on it. If an increasing amount of force is applied, a threshold is reached, following which the book begins to move. It will continue to move without any subsequent increase in force. Upon removal of the force, the book will suddenly stop and will not go back to its original position. This is a characteristic of dry friction between bodies. There are other natural phenomena in which the relation between force and motion is similar. An example is the stretching of a ligament beyond its elastic limit, thus producing permanent deformation. The ligament is said to be plastically deformed. To describe this and other similar phenomena quantitatively, a mathematical model may be constructed wherein a dry friction–mathematical element may represent the actual behavior.

In Grade I spondylolisthesis, L5–S1, suppose the annulus and all the other supporting soft-tissue elements are removed. The patient then develops a moderately stable syndesmosis between the two vertebral bodies. Mild forces would not be strong enough to push L5 further forward with respect to S1. However, a large force could transcend the threshold of the dry friction offered by the syndesmosis, and L5 would slip indefinitely if it weren't for other clinical factors that create new dry friction thresholds and restrict further displacement.

The mathematical concept, as exemplified above, is used to represent those properties of a tissue which are characterized by a sudden displacement after a threshold load is reached and by permanent deformation at the removal of the load. In Figure 9-19, the relationship between the load and the deformation is shown. It is characterized by the threshold load and permanent deformation. Also, the dry friction symbol is shown: a block resting on a surface. See also *Plasticity*.

▶ Ductility

DEFINITION. Property of a material to absorb relatively large amounts of plastic deformation energy before failure.

DESCRIPTION AND EXAMPLES. Materials possessing large amounts of ductility are called ductile. In contrast, nonductile or brittle materials have a relatively small plastic energy–absorbing capacity (Fig. 9-20). Ductility of a material is quantified by either percentage elongation in length or percentage decrease in cross-sectional area at the time of failure. Generally, materials that exhibit less than 5% elongation are called brittle, while materials that exhibit more are called ductile.

Most metals are ductile, while ceramics, hard plastics, and cortical bone are brittle. Implants made of ductile materials can undergo large deformations and absorb substantial amounts of energy before failure. However, in general, they have lower ultimate tensile strength and therefore cannot take up overloads. Some examples of ductile and brittle materials are given in Table 9-2.

A spine with Ehlers-Danlos syndrome is more ductile than one with Marie-Strumpell's disease.

▶ Dynamic Load

DEFINITION. A load applied to a specimen is called dynamic if it varies with time.

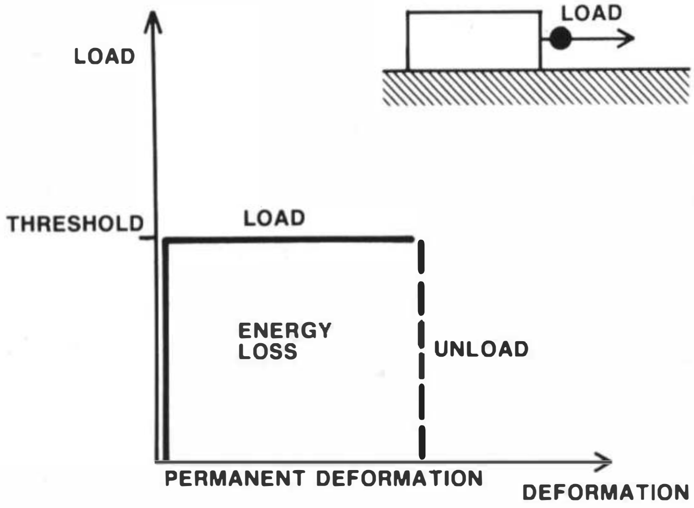


FIGURE 9-19 Dry friction. The load–deformation curve shows the motion of a block being pulled by a force. Because of friction, the block does not move until the load reaches a value (*Threshold*) slightly greater than the friction. On removal of the load, the block suddenly stops, leaving behind a permanent deformation.

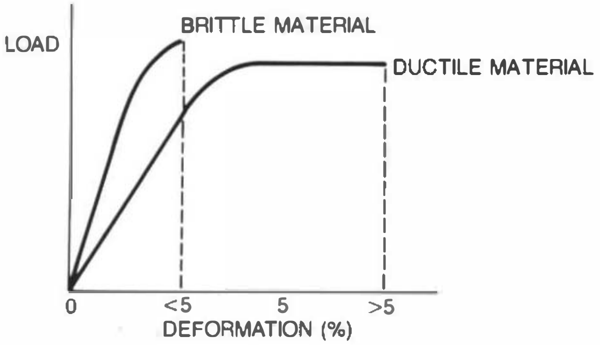


FIGURE 9-20 Ductility. A ductile material has greater deformation and absorbs larger amounts of energy before failure than a brittle material.

DESCRIPTION AND EXAMPLES. A dynamic load is the opposite of a static load. A dynamic load with a repetitive pattern of variation is called a cyclic load.

The lumbar spine of a pilot in a disabled high-speed aircraft is subjected to tremendous dynamic loads as he is ejected out of the craft by a rocket attached to his seat.

During normal gait, all body parts are subjected to dynamic loads. The head of the femur (Fig. 9-21) is stressed under varying degrees of dynamic compression as the load is transferred from one leg to the other.[15] This is a cyclic load because the loading pattern is repeated at each cycle of gait. In contrast, a static load is applied when a person is standing still.

Another example of a dynamic load with a repetitive pattern is vibrations. In an epidemiologic study of disc herniation in the lumbar spine, vibration has been implicated as a risk factor.[8] Recently, vibration transmission into the spine has been found to be most intense if the frequency of vibration is about 4.5 cycles per second.[10]

TABLE 9–2 Physical Properties of Some Materials

Material	Ultimate Strength		Elongation (%)	Property
	(MPa)	(lbf/in^2)		
Stainless steel, annealed[*]	517	75,000	40	Ductile
Stainless steel, cold worked[*]	862	125,000	12	Ductile
Cortical bone, wet[†]	81	11,800	1.2	Brittle
Cortical bone, dry[†]	107	15,500	0.66	Brittle

[*] American Society for Testing and Materials. Standards for Surgical Implants. Tab. 2. Philadelphia, 1971.

[†] Evans, F. G.: Mechanical Properties of Bone. Springfield, IL, Charles C Thomas, 1973.

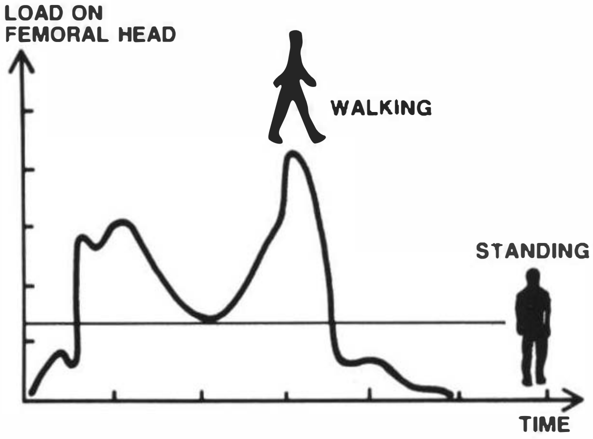


FIGURE 9-21 Dynamic load. The head of the femur is subjected to dynamic (varying with time) loads in walking and static (constant) loads in standing.

▶ Dynamics

DEFINITION. A branch of mechanics concerned with the study of the loads and motions of interacting bodies.

DESCRIPTION AND EXAMPLES. Gait analysis is a good example. Here, the loads applied by the muscles to the bones, the ground reactions (as measured by the force plate), and the various body motions produced are studied.

▶ Elasticity

DEFINITION. Property of a material or a structure to return to its original form following the removal of the deforming load.

DESCRIPTION AND EXAMPLES. Energy is stored during loading and released completely during unloading. Thus, no energy is lost in the process, and there is no permanent deformation. Stress and strain curves of an elastic material may be linear or nonlinear, but the loading and unloading curves are always the same.

All materials are elastic to a varying degree. A person jumping off a diving board uses the elastic properties of the board. He stores energy as he pushes the board downward by jumping on its unsupported end. The board in turn gives back the stored energy during the diver's push-off.

▶ Elastic Range

DEFINITION. A range of loading within which a specimen or a structure remains elastic.

DESCRIPTION AND EXAMPLES. When a specimen or a structure is subjected to a load, it deforms. If the deformation is such that the specimen or the structure returns to its preload shape upon release of the load, then the deformation is called elastic deformation. Figure 9-22A shows the load–deformation curve for a specimen. The elastic range is represented by line OB in the figure. Within the elastic range, the deformation may be proportional to the load (proportional or linear range), line OA, or it may vary (nonlinear range), line AB.

A rubber band or the old comic book character "Plastic Man" (Fig. 9-22B) demonstrates linear and nonlinear elasticity. Actually, this name is a biomechanical misnomer, since "Plastic Man" never exhibited any plastic deformation, always being in the elastic range. He might never have sold, however, under the correct engineering appellation of "Elastic Man."

Implants are designed so that the maximum stress remains within the elastic range during normal physiologic activity. Although the implants are capable of carrying much greater loads before failure, the elastic range is the only useful range, for once there is loading beyond this range, permanent deformation and implant failure will occur.

▶ Elastic Stability

DEFINITION. The ability of a loaded structure, given an arbitrary small elastic deformation, to return to its original position.

DESCRIPTION AND EXAMPLES. The stability of an elastic structure is a function of the geometry of the structure and the quantitative and qualitative characteristics of the applied load. The classic example of elastic instability is the axially loaded columns that were studied by Euler in the 18th century. A cylindrical bar with its lower end fixed in the ground and its upper end loaded with weights was investigated (Fig. 9-23A). Euler applied increasing loads on the column until the column was no longer able to maintain its straight vertical position. He called this

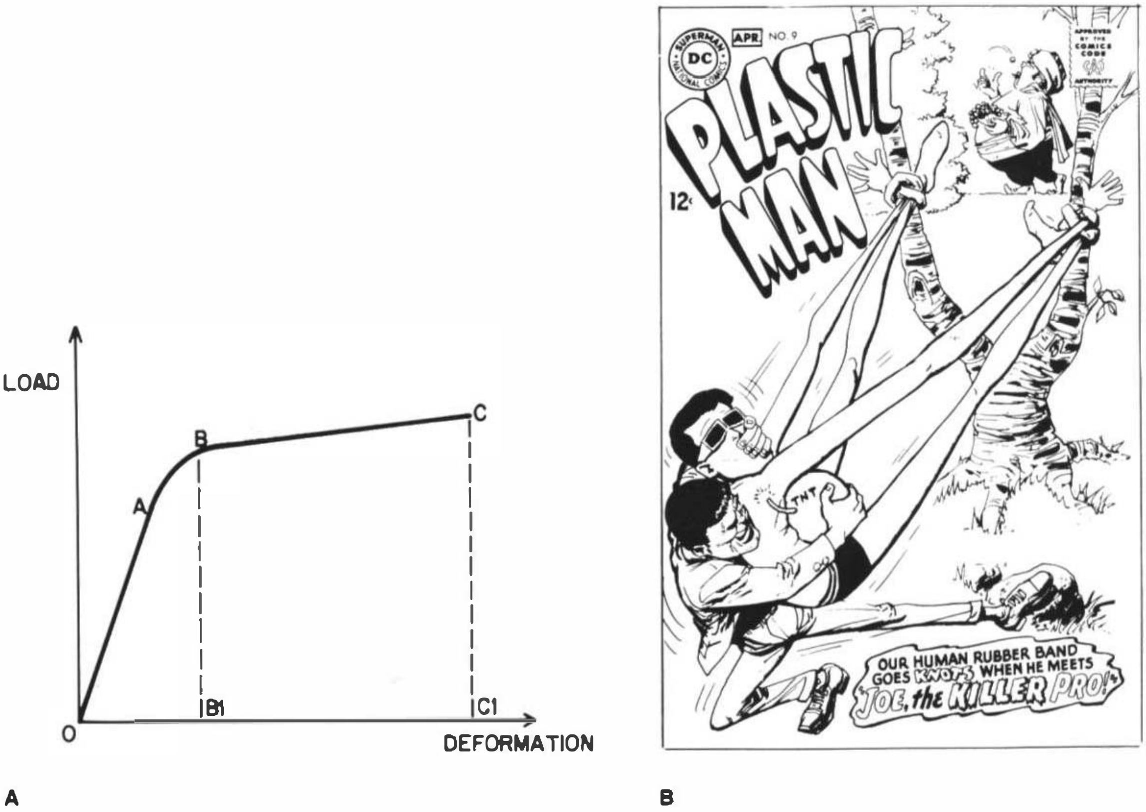


FIGURE 9-22 **(A)** Elastic range. OA = elastic, linear range; AB = elastic, nonlinear range; BC = plastic range. **(B)** "Plastic Man" demonstrates linear and nonlinear elasticity. *(Cartoon reproduced from the cover of* Plastic Man, *1968. © 1968 DC Comics Inc.)*

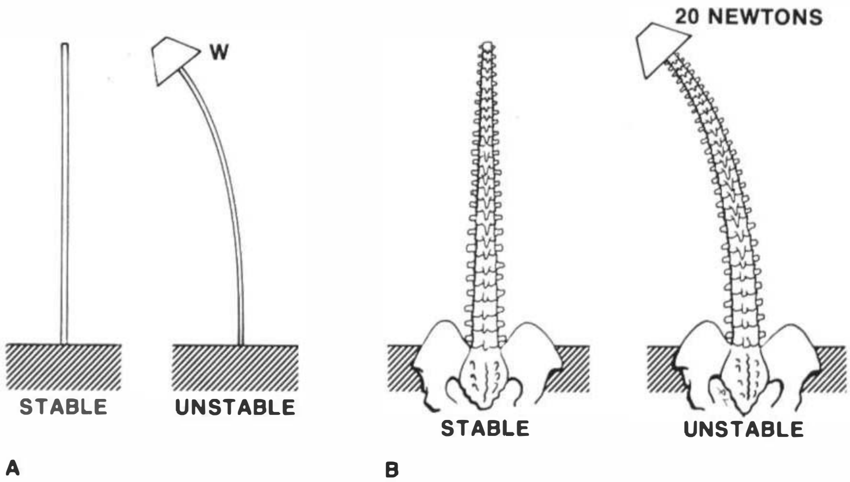


FIGURE 9-23 Elastic stability. **(A)** The column does not remain straight when W reaches a critical value. **(B)** The critical load for the human T1–S1 cadaver spine stripped of musculature is less than 20 N (4.5 lbf).

final load "the critical load." The mathematical formula for determining this load is given under Explanatory Notes.

It has been shown by cadaveric experiments that a spine specimen, T1 to pelvis, relieved of its musculature and the rib cage has a critical load of about 20 N (4.4 lb). Subjected to any load greater than the critical load, the spinal column is unstable and buckles like an elastic column (Fig. 9-23B).[9] This points out the importance of the spinal muscles and certain other anatomic structures in maintaining the elastic stability of the spine. Note that this elastic stability is distinctly different from what is referred to as clinical stability.

EXPLANATORY NOTES. Euler's formula for calculating the maximum load W in newtons (poundforce), the so-called critical load, is as follows:

$$W = \frac{\pi^2 EI}{4 L^2}$$

where $\pi = 3.1416$; E = modulus of elasticity in N/m^2 or MPa (lbf/ft^2); I = section moment of inertia in m^4 (ft^4); and L = length in m (ft).

▶ Elastic Zone

DEFINITION. A part of the range of motion of a body (e.g., a vertebra), starting from the beginning of some resistance offered by the joint to the end of the range of motion. The unit of measure is meter (foot) for translation, or radian (degree) for rotation.

DESCRIPTION AND EXAMPLES. During flexion/extension of the knee with relaxed muscles, there is very little resistance offered by the joint for most of its range of motion, except when the tibia approaches the ends of its range of motion. These parts of the physiological range of motion where the joint offers resistance to motion of the knee are the elastic zones.

Adding the elastic zone to the neutral zone gives us a measure of the range of motion. See also *Neutral Zone* and *Range of Motion*.

▶ Energy

DEFINITION. The amount of work done by a load on a body. The unit of energy is the newton meter or joule (foot poundforce).

If the load deforms or displaces the body, the corresponding energy is called *strain* or *potential energy*, respectively. If the load imparts motion to the body, the energy is called *kinetic* energy.

DESCRIPTION AND EXAMPLES. The strain or potential energy of a structure subjected to a load is represented by the area of its load–deformation diagram. Assume that the load–deformation diagram shown in Figure 9-24 is for a spine segment subjected to compression force. Then, if the spine has been elastically deformed to point B, the elastic energy stored is the area O-A-B-B1-O and is fully recoverable on removal of the load. Deformation from B to C is plastic (*i.e.*, because of high load, the structure is breaking down on a microscopic scale). If a fracture takes place at C, then the areas B1-B-C-C1-B1 and O-A-B-C-C1-O represent the plastic and total energies, respectively. The total energy has been expanded in plastically deforming vertebrae and ligaments, creating fracture surfaces and imparting kinetic energy to the fractured pieces. For further discussion, see *Potential Energy* and *Kinetic Energy*.

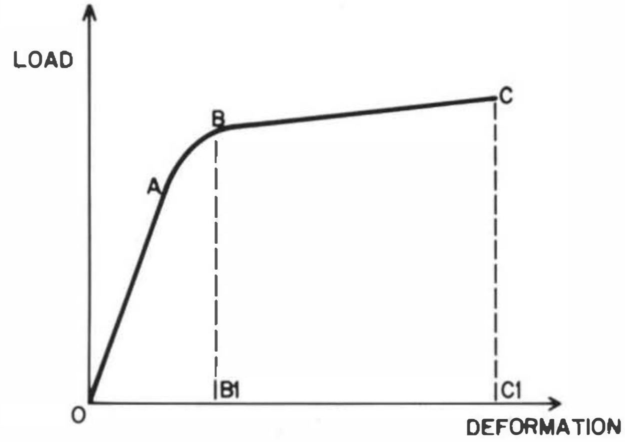


FIGURE 9-24 Energy. The area under the load–deformation curve represents the energy. Area ($OABB_1O$) represents elastic energy. Area ($B_1BCC_1B_1$) represents plastic energy.

EXPLANATORY NOTES. Mathematical expressions for the two kinds of energy are as follows:

$$\text{potential energy} = F \times D$$

$$\text{kinetic energy} = \frac{m \times V^2}{2}$$

where F = force (constant magnitude) in N (lbf); D = displacement of the point of force application in the direction of the force in m (ft); m = mass in kg (lb); and V = velocity (constant magnitude) in m/s (ft/s).

The above formulas take on integral forms (*i.e.*, can be represented by areas under the graph) if the force and velocity are not constant.

▶ Energy Absorption Capacity

DEFINITION. The mechanical energy absorbed by a structure loaded to failure. The unit of measure is the newton meter (foot poundforce).

Also called *total* energy, this energy is expanded during plastic deformation, during fracture surface generation, and in imparting motion to fractured fragments. It is conveniently given by the total area (O-A-B-C-C1-O) under the load–deformation curve shown in Figure 9-24.

DESCRIPTION AND EXAMPLES. Higher energy absorption capacity is generally synonymous with high ductility of materials. A stainless steel fixation plate, although designed for loads under its yield stress, may in an accident be subjected to high energy impact. If the plate has high energy absorption capacity, it may help the patient in one of two ways. It may deform considerably without failure because of its ductility, thus maintaining some of the alignment and eliminating additional complications, or it may deform and break by absorbing large amounts of impact energy, thus decreasing the amount of energy available to cause soft-tissue damage.

▶ Equilibrium

DEFINITION. A body is said to be in a state of equilibrium if it is at rest or in uniform motion under a given set of forces and moments.

DESCRIPTION AND EXAMPLES. The concept of equilibrium arises from Newton's second law of motion (see *Newton's Laws*). All forces and moments acting on a body must balance each other so that the body does not accelerate. There are two types of equilibrium: *static*, in which the body in equilibrium is at rest; and *dynamic*, in which the body in equilibrium is moving at a constant speed (uniform motion).

Figure 9-25A shows part of a lumbar spine and a horizontal bar carrying weight, representing the weight of the upper body. Because of eccentricity of the weight, the lumbar spine is subjected to forces as well as bending moments. To estimate the loads acting on the L4 vertebra when a person is lifting a weight, six equilibrium equations may be set up by the method of free-body analysis (see below). The vertebra, with possible forces and moments acting on it, is shown in Figure 9-25B. The solution to the equilibrium equations constitutes a calculation of

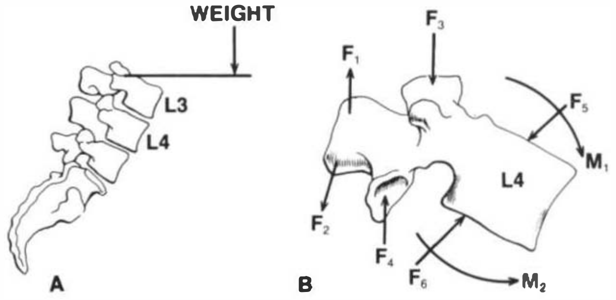


FIGURE 9-25 Equilibrium. Relationships between **(A)** external load (weight) on the spine, and **(B)** various internal forces and moments are given by equilibrium equations. F_1 and F_2 = muscle and ligamentous forces; F_3 and F_4 = facet forces; F_5 and F_6 = disc forces; M_1 and M_2 = disc bending moments.

the magnitude of forces and moments acting on the vertebra.

EXPLANATORY NOTES. For a body or a structure to be at rest, or in uniform motion, the two following conditions must be satisfied: (1) The sum of forces in all directions acting on it must be equal to zero, and (2) the sum of moments taken at any point of the body, around all axes, also must be equal to zero.

If the force and moment vectors are broken down into their components along the three axes of a coordinate system, then mathematically the following six equilibrium equations apply:

$$\Sigma F_x = 0, \quad \Sigma F_y = 0, \quad \Sigma F_z = 0$$
$$\Sigma M_x = 0, \quad \Sigma M_y = 0, \quad \Sigma M_z = 0$$

where F = forces at the point on the body and M = moments at the point on the body.

Subscripts x, y, and z refer to the axes of an orthogonal coordinate system at that point. The symbol Σ (sigma) stands for summation of all the forces and moments. The six equations of equilibrium, as given above, are probably one of the most important biomechanical tools for the mechanical analysis of the musculoskeletal system. They may be used in any situation: a single force or a complex combination of forces and moments in three-dimensional space acting at different points on a body. A simple graphical solution may be used for forces acting in one plane, such as forces exerted by various traction devices. For the three-dimensional loading situations, the algebraic solution of the six equations given above is probably the most efficient method.

▶ Fatigue

DEFINITION. The process of the growth of cracks in structures subjected to repetitive load cycles. The load is below the failure load of the structure.

DESCRIPTION AND EXAMPLES. When a fatigue crack reaches a certain size, the stress in the rest of the structure becomes so high that the structure fails. This is *fatigue failure*. Another way to look at this phenomenon is to consider it as a summation effect. As soon as the structure is subjected to a repetitive load, however small, the "fatigue clock" starts ticking. The speed of the clock is in proportion to the magnitude of the load. The higher the load, the faster runs the clock. The life of a given structure may then be measured by its "fatigue clock." When the structure has lived its full life, as measured by the "clock," it fails.

The magnitude of the cyclic load is within the elastic range and is far below the failure load of the structure. For steel, cyclic loads of magnitudes as low as 20% of the failure load will cause fatigue failure in a reasonable time interval. A similar figure for a sample of cortical bone is 35%. In implants and bone, it is the combination of somewhat higher physiologic loads and their cumulative repetition that brings about the failure. The method by which fatigue failure may be calculated is given below.

Most probably, in living bone, the fatigue limit is relatively higher than 35%, since reparative biologic processes may compensate for the propagation of cracks. However, fatigue fractures in bone do occur, indicating either that loads above the fatigue limit have been applied for a sufficient period of time or that the bone-healing process failed to repair the minute fatigue cracks at a sufficiently rapid rate. Fatigue fractures have often been called "stress (or march) fractures," which is a biomechanical misnomer. All fractures are created by excessive stress. What is special about fractures due to fatigue is the repetitive nature of the loads of relatively low magnitude applied over a certain period of time.

EXPLANATORY NOTES. The process of fatigue is documented by the Wohler or fatigue curves. The load is plotted on the ordinate, while the number of load cycles to failure, on the logarithmic scale, is plotted on the abscissa (Fig. 9-26A). The fatigue limit, also called the endurance limit, is the lowest load that will cause fatigue failure. Loads lower than the fatigue limit never cause a fatigue failure within a reasonable time. A single load that causes failure is called the ultimate load. Generally, a straight line connects the ultimate load and the endurance limit points. This line is also referred to as the S-N curve.

For the purpose of standardization, the fatigue curve is generally obtained for cyclic loads that vary with time from a maximum in one direction to a maximum in the opposite direction, as shown in Figure 9-26B. Note that the *in vivo* loads are not fully reversing (e.g., load on the heel during walking—the compressive force varies between zero and body weight.

▶ Flexibility Coefficient

DEFINITION. The flexibility coefficient of a structure is defined as the ratio of the amount of displacement produced to the load applied. It is a quantity that characterizes the responsiveness of a structure to the applied load. Units of measure are meters per

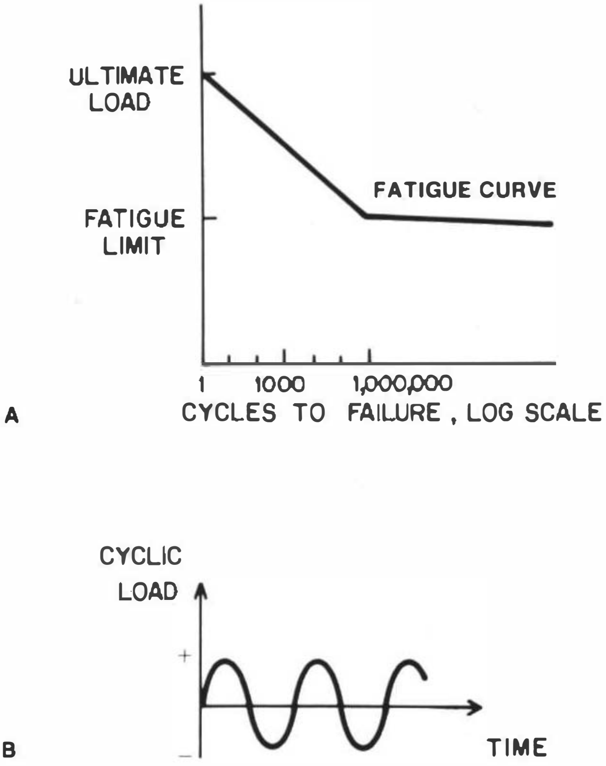


FIGURE 9-26 Fatigue. **(A)** The fatigue curve. The number of cycles to failure in logarithmic scale are plotted against the load. **(B)** Cyclically varying loads are applied to the test specimens to determine the fatigue curve. Although a fully-reversing load is shown, this is not always the case.

newton (feet per poundforce) for linear displacement and radians per newton meter (degrees per foot poundforce) for angular displacement.

DESCRIPTION AND EXAMPLES. For a structure with a linear load–displacement curve, the flexibility coefficient is a constant and is the inverse slope of the curve. For more complex structures, the flexibility coefficient may vary with the magnitude of the load.

A supple scoliotic spine has a relatively high flexibility coefficient. In such a spine, only small forces are required to produce large deformations. Thus, such a spine might be expected to respond well to treatment with a Milwaukee brace.

EXPLANATORY NOTES. Mathematically speaking, the flexibility coefficient f is related to the applied load F and the displacement D by the following formula:

$$f = \frac{D}{F}$$

The inverse of the flexibility coefficient is generally called the stiffness coefficient k:

$$k = \frac{1}{f}$$

It should be pointed out that, strictly speaking, in complex structures such as the human spine, with true three-dimensional motions that are coupled, the simple relationship of reciprocity between the flexibility and stiffness coefficients does not hold. In such instances, the two coefficients can be meaningfully related by means of matrix inversion only, a much more complex mathematical operation.

Also, the spine, and most biologic structures, exhibit significant nonlinear behavior. Therefore, the flexibility coefficient (a single number) is not an accurate description of a complex nonlinear behavior.

▶ Force

DEFINITION. Any action that tends to change the state of rest or motion of a body to which it is applied. The unit of measure for the magnitude of force is newtons (poundforce).

DESCRIPTION AND EXAMPLES. A woman sitting in a chair is at rest under the action of two equal and opposite forces (Fig. 9-27A). The earth's gravitational field is trying to accelerate her toward the center of the earth. The chair is applying an exactly equal force in the opposite direction, thus preventing her motion. If one suddenly removes the chair (Fig. 9-27B), the gravitational force (mass times acceleration) will quickly change her position and attitude of rest.

Force is a vector quantity and is completely spe-

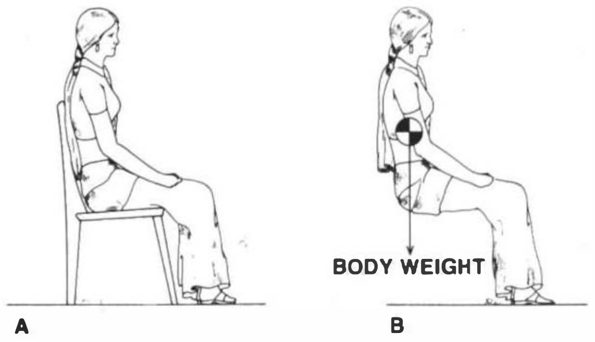


FIGURE 9-27 Force. **(A)** Balanced equal and opposite forces hold the body in equilibrium. **(B)** Unopposed gravitational force accelerating the body towards the floor.

cified by its magnitude, direction, point of application, and sense. A hospital bed on wheels serves as an example. To specify the force that will be applied, the magnitude of force in newtons (poundforce) must be defined. The orientation of the direction of the force must be discerned (e.g., is the force applied in a vertical direction, in a horizontal direction, or in any other direction in space?). The point at which the force is being applied must also be specified. Finally, the type of force, push or pull, must be determined. Once all four parameters are defined and the force is applied to the bed, it will move in a certain direction. A change in any of the parameters will produce a different motion of the bed.

▶ Four-Point Bending

DEFINITION. A long structure is loaded in four-point bending when two transverse forces are applied on one side and two on the other.

DESCRIPTION AND EXAMPLES. If all the forces are equal and arranged symmetrically (Fig. 9-28A), a unique situation results, so that the structure between the inner pair of forces is subjected to a *constant* bending moment or stress. The mathematical derivation for this is given below. Since the bending moment is constant along the length B, a constant corrective effect is obtained along the corresponding region of the spine. This may be useful in certain clinical situations. Three-point bending, in contrast, has a varying bending moment with a peak just under the middle force.

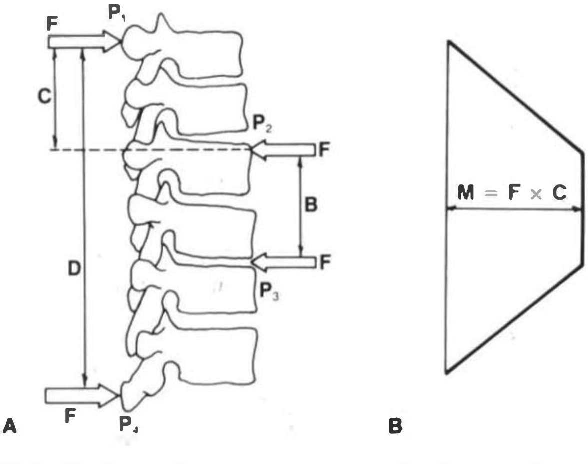


FIGURE 9-28 Four-point bending. (A) Spine subjected to four forces (P_1 to P_4). (B) Bending moment diagram with constant bending moment generated between points P_2 and P_3.

EXPLANATORY NOTES. As shown by the bending moment diagram in Figure 9-28B, the equation for the maximum bending moment is as follows:

$$M = F \times C$$

It is interesting to note that dimensions D and B are not included in the above formula. The reason is that the bending moment at a point of a long structure is the summation of all the moments on one side of the point. Above the point P_2 there is only force F located at P_1 a distance C away (Fig. 9-28A). As one travels from P_2 to P_3, there are equal and opposite contributions toward the bending moment from the two F forces located at P_1 and P_2. Thus, the bending moment remains constant between P_2 and P_3.

▶ Free-Body Analysis

DEFINITION. A technique used for determining the internal stresses at a point in a structure subjected to external loads.

DESCRIPTION AND EXAMPLES. This is an important method in the analysis of internal forces in a structure subjected to external loads (forces and moments).

The part of the structure to be analyzed is isolated or cut away from the rest by an imaginary boundary. At the boundary, internal stresses are represented by forces and moments as if they were the loads applied to the isolated portion of the structure by the rest of the structure. Equilibrium equations (see *Equilibrium*) are then applied to the isolated portion of the structure to evaluate the internal stresses at the boundary in terms of the external loads. This method is based on the fact that the isolated structure must be in complete balance with respect to all the forces and moments applied to it. This process is called free-body analysis, and the isolated portion of the structure is referred to as the free-body diagram.

In studying the stresses in the C7–T1 disc when a person is bending forward, so that the cervical spine is in the horizontal plane (Fig. 9-29A), the imaginary boundary is drawn between C7 and T1. The isolated structure consists of the head and the whole cervical spine. The external load is the weight of the head and the whole cervical spine (Fig. 9-29B). The internal forces and moments at the boundary are F_1, F_2, F_3 and M. They have replaced the interaction of the rest of the structure. By applying six equilibrium equations to the isolated structure, the disc force F_3 and disc bending moment M are obtained. If one knows the disc loads and the

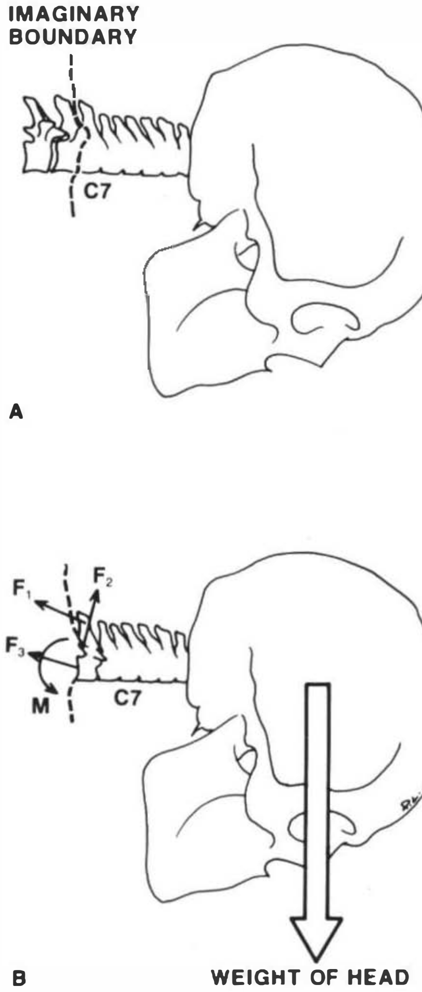


FIGURE 9-29 Free-body analysis. **(A)** A structure (e.g., cervical spine and head) is isolated for analysis. **(B)** The external load and possible internal forces and moments are shown. Equilibrium equations are used to calculate the internal force and moments.

geometric and material properties of the disc, the required stresses may be computed. (See *Equilibrium*.)

▶ Friction

See *Coefficient of Friction.*

▶ Functional Spinal Unit

See *Motion Segment.*

▶ Helical Axis of Motion

DEFINITION. A unique axis in space that completely defines a three-dimensional motion of a rigid body from position one to position two. It is analogous to the instantaneous axis of rotation for plane motion.

DESCRIPTION AND EXAMPLES. In three-dimensional motion, a rigid body is displaced from one position to another position in space. According to the laws of mechanics, a rigid body may always be moved from position one to position two by a rotation about a certain axis and a translation along the *same axis.* This constitutes helical motion. Six numbers are required to define three-dimensional motion: four define the position and orientation of the helical axis, and two define the amount of rotation about and translation along it.

The helical axis of motion is one of the most precise ways to define the three-dimensional motion of a rigid body. This method of presentation is well suited for describing motion of irregular bodies, such as anatomic structures upon which it is difficult to consistently and accurately identify reference points. Because the helical axis of motion describes any kind of general motion, there are many illustrative examples.

If one throws a perfect "bullet" pass with an American football, as shown in Figure 9-30, then instantaneous motion of the ball is defined by an axis that runs through the center and is oriented along the longitudinal axis. The ball is translating along that axis and is rotating about that same axis. This is the instantaneous helical axis of motion of the ball at the instant of observation.

When a screw is driven into the bone to fix a fracture, the screw translates into the bone as its head is rotated. The motion of the screw is a helical motion, and the axis of the screw is the helical axis of motion. It is for this reason that the helical axis is sometimes called a screw axis of motion.

The helical axis has been used in two instances in orthopedics, namely, to define intervertebral motions in the thoracic spine[24] and to define the motions of the metacarpophalangeal joint.[12] A potential use for the helical axis is to define precisely the movement that has taken place in the transition from a normal spine to a scoliotic spine for each vertebra.

▶ Hysteresis

DEFINITION. Hysteresis is a phenomenon associated with energy loss exhibited by viscoelastic materials when they are subjected to loading and unloading cycles.

DESCRIPTION AND EXAMPLES. It is known that the area under a loading curve in a load–deformation diagram represents the energy of deformation (see

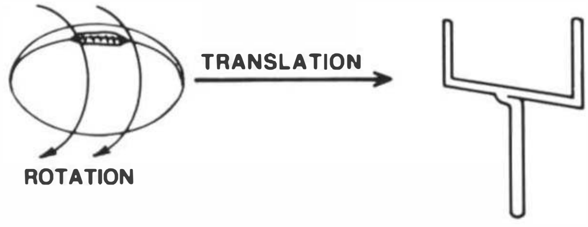


FIGURE 9-30 Helical axis of motion.

Energy). If the unloading curve is exactly the same as the loading curve, then the energy of deformation is completely regained during unloading. On the other hand, if the unloading curve is *below* the loading curve, then the energy regained is less than the energy put in. The area enclosed between the two curves represents the energy lost and is called hysteresis loss.

Figure 9-31 shows the results of a tension test experiment performed on a cruciate ligament of a rabbit.[22] Note that the load–deformation curve during the unloading cycle is below the curve for the loading cycle. Measurements from the diagram show that 17% of the total energy is lost during a single load/unload cycle.

▶ Impulse

DEFINITION. Linear impulse of a force is the product of the force and the time interval of force application. The unit of measure is newton seconds (poundforce seconds).

Angular impulse of a moment is defined as the product of the moment and the time interval of moment application. The unit of measure is newton meter seconds (foot poundforce seconds).

DESCRIPTION AND EXAMPLES. When one pushes a stalled car on the road, it gains speed slowly. At any point in time, the speed of the car is in direct proportion to the magnitude of the force and the time interval for which it is applied. According to the definition, this is impulse. If a larger impulse is applied to the same car, or if the same impulse is applied to a smaller car, both cars will achieve higher speed.

In trauma, the destruction of the tissue (hard as well as soft) depends not only upon the magnitude of the force but also upon the time duration of force application. It has been shown in spinal cord trauma experiments, in which a weight is dropped from a certain height, that the amount of damage to the cord is directly related to the magnitude of impulse of the falling weight.[4]

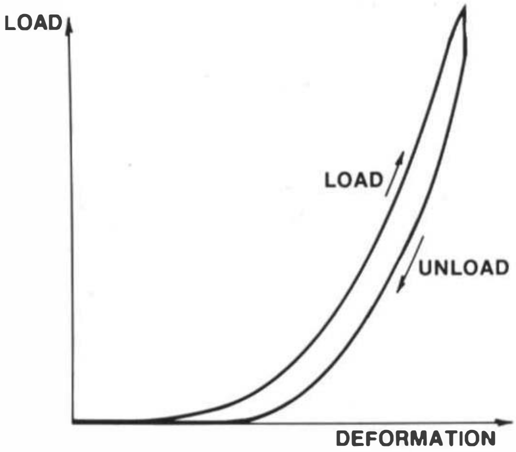


FIGURE 9-31 Hysteresis. The area between the load and unload curves is the energy loss in one load/unload cycle.

▶ Inertia

DEFINITION. The property of all material bodies to resist change in the state of rest or motion under the action of applied loads.

DESCRIPTION AND EXAMPLES. A bicycle has mass, and therefore inertia. One must apply forces to the pedals to get the bike moving from its state of rest. To slow down, it is necessary to apply braking forces to alter the state of motion to a slower one.

The concept of inertia is important in the analysis of trauma to the spine. When an acceleration is imparted to the lower portion of a resting spine in a rear-end collision (whiplash), the inertia of the head and upper portion of the body resists the change. This resistance imparts potentially injuring forces to the spine and adjacent structures (see *Angular Acceleration*).

▶ Instantaneous Axis of Rotation

DEFINITION. When a rigid body moves in a plane, at every instant there is a point in the body or some hypothetical extension of it that does not move. An axis perpendicular to the plane of motion and passing through that point is the instantaneous axis (center) of rotation (IAR) for that motion at that instant.

DESCRIPTION AND EXAMPLES. Plane motion is fully defined by the position of the instantaneous axis and the magnitude of rotation about it. Figure 9-32 shows a graphical technique of determining the instantaneous axis of rotation when a body moves from position 1 to position 2. The axis is found to be at the intersection of the two perpendicular bisectors

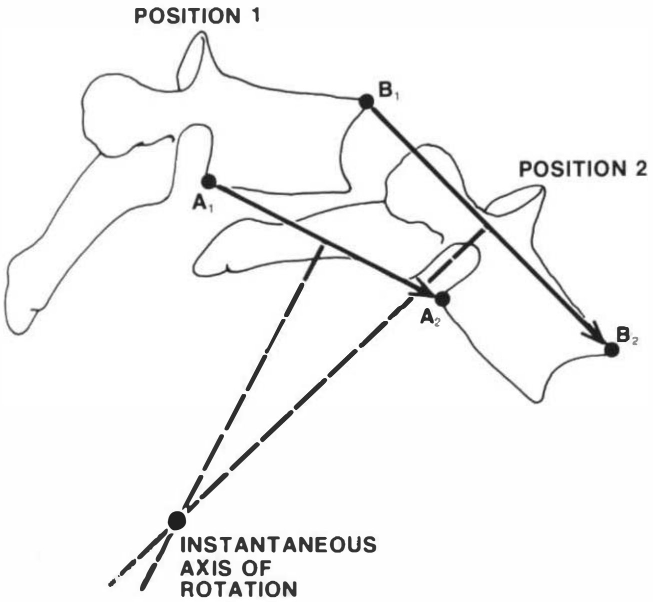


FIGURE 9-32 Instantaneous axis of rotation. A construction for determining the IAR is shown. A_1A_2 and B_1B_2 are translation vectors of points A and B.

of translation vectors A_1 A_2 and B_1 B_2 of any two points A and B on the body.

Vertebrae undergo plane motion during flexion/extension of the spine. Each vertebra has instantaneous axes of rotation in relation to an outside frame of reference (e.g., the ground) as well as in reference to each of the other vertebrae. During motion from full flexion to full extension or vice versa, different anatomic components (ligaments, muscles, and portions of facet articulation) come into play as motion progresses. In other words, the structure of the FSU changes. Since an IAR is related to a certain structure, it also changes as a function of the degree of bending.

The beauty of the concept of the instantaneous axis of rotation is that any kind of plane motion may be described: translation, rotation, or a combination of the two. For a detailed study of complex plane motion, one may regard the motion as being made up of smaller steps. Thus, a set of instantaneous axes of rotation may be established to represent the total motion. This pattern of IAR has been successfully used in the clinical evaluation of knee injuries.[7]

One should note that the location of the IAR is highly sensitive to measurement errors.[9a] Therefore, the experiments should be well designed and the results carefully evaluated.

▶ Instantaneous Velocity

DEFINITION. The average velocity when the time interval approaches zero. The unit of measure is meters per second (feet per second).

Velocity is linear when the motion is translation and angular when it is rotation. It is a vector quantity and therefore has magnitude (speed) and direction.

DESCRIPTION AND EXAMPLES. A car traveling from Yale to Harvard (Fig. 9-33), for some unfathomable reason, has an average speed (velocity magnitude) of 100 km/h (60 mph). At a certain instant in time, the speed is probably higher or lower than the average (i.e., the instantaneous speed varied). At another instant in time, the direction of the speed probably differs from the direction of Yale to Harvard. A complete description of velocity of the car, therefore, requires full documentation of its instantaneous velocity vectors, which includes changes in both speed and direction throughout its journey from excellent toward good. On the straight portions of the road (Fig. 9-33A), if the speed changes, the velocity vector varies. On the curved portions of the road (Fig. 9-33B), the speed may remain constant, but still the velocity vector may vary because of the continually changing direction.

▶ Integration, Integral

DEFINITION. *Integration* is an incremental summation process. *Integral* is the mathematical description of integration.

DESCRIPTION AND EXAMPLES. Integration is used for finding lengths, areas, and volumes of complex objects by adding together all the small bits into which the object could be divided.

To find the volume of a whole tibia, it may be cut into transverse sections that are 1 mm (0.04 in) thick. By measuring the areas of all cross-sections, multiplying these by the thickness, and adding them all together, the tibia volume is obtained. This physical process can be done mathematically if information regarding variation of the cross-sectional area as a function of axial length can be expressed mathematically. Then,

$$\text{tibia volume} = \int_{l_1}^{l_2} A\,dl$$

where $\int$ is the integral. The mathematical function that describes the variation of the cross-sectional area with length is denoted by letter A, and dl is the symbol denoting that the integration (summation) is

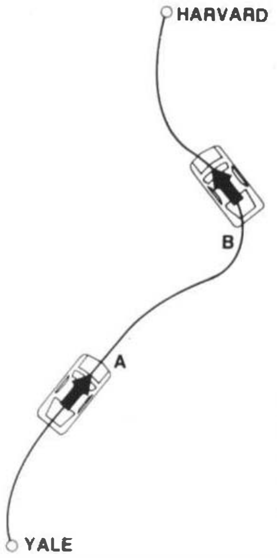


FIGURE 9-33 Instantaneous velocity. Although the speed of the car at points A and B is the same, the velocity, which includes direction, is different at each point.

to be performed along the length l. Quantities l_1 and l_2 are the boundaries of the length l.

The right-hand side of the expression above is the integral, and the process of computation is called integration. It should be read as follows: "integral of A with respect to l between the limits l_1 and l_2." Sometimes the limits l_1 and l_2 may be missing from the expression, which implies that these are either understood or have not yet been defined.

▶ Isotropic Material

DEFINITION. A material is called isotropic if its mechanical properties are the same in all directions.

DESCRIPTION AND EXAMPLES. In other words, if one takes a sample of the isotropic material for testing, then the values of its mechanical properties (strength, modulus of elasticity, and so forth) will be the same regardless of the orientations of the test samples.

Metals, hardened methylmethacrylate, and ice are examples of isotropic materials. Wood and bone, in contrast, are not isotropic because they have fibers (collagen and cellulose, respectively) oriented in preferred directions. There is, to our knowledge, no isotropic tissue in the body, because every tissue is highly specialized to resist loads optimally in a certain direction only. Therefore, a tennis ball is used here rather than an organ to demonstrate isotropic properties.

Wherever the surface of the tennis ball is hit, its mechanical properties are the same, provided that the force vector remains constant. When the ball is hit, its subsequent motion depends upon the load–deformation characteristics in the direction of the force vector at the time of the ball–racket contact. This consistency of mechanical response is isotropy.

Experiments have shown that cortical bone is highly sensitive to the direction of force application and is therefore not isotropic.[18]

▶ Joint Reaction Force

DEFINITION. If a joint in the body is subjected to external forces in the form of external loads and/or muscle forces, the internal reaction forces acting at the contact surfaces are called *joint reaction forces*. The unit of measure is newtons (poundforce).

DESCRIPTION AND EXAMPLES. The contact surface of the metacarpal portion of a metacarpophalangeal joint (Fig. 9-34A) has two components of joint reaction: one that is perpendicular to the contact surface, called the normal component, and one that is parallel to the surface, called the tangential or frictional component (Fig. 9-34B). The perpendicular component is always compressive. The tangential component in the healthy joints is generally very small, about 1% of the normal component. This is due to very low joint friction. The direction of the tangential component is always opposite the direction of the sliding motion. The ratio of the two components (tangential to normal) is the coefficient of friction of the joint.

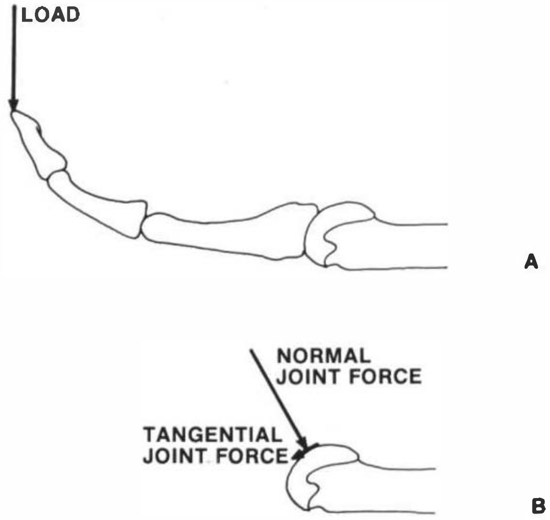


FIGURE 9-34 Joint reaction force. **(A)** External load. **(B)** Joint reaction forces.

▶ Kilopond

DEFINITION. A metric measure of force. It is equal to the gravitational force applied to 1 kg of mass at the earth's surface.

DESCRIPTION AND EXAMPLES. If a 1-kg mass is held in the hand, a downward force of 1 kp is applied to the hand by the earth's gravity. Therefore, there is a one-to-one relation between mass in kilograms (or pounds) and its force in kiloponds (or poundforce). Unfortunately, this relationship is strictly earthbound. It does not apply to forces on the moon, for example. Because the moon's gravity is only one-sixth of that of the earth, a 1-kg (1-lb) mass on the moon applies a downward force on the hand of only one-sixth of a kilopond (or poundforce). This is one of the reasons that in the newly adopted Systeme International d'Unites, the unit of force is newtons (see *Newton*).

The kilopond is abbreviated as kp, and conversion factors to units of force are as follows:

1 kilopond (kp) = 9.806 newtons
1 kilopond (kp) = 2.205 poundforce
1 newton (N) = 0.1020 kilopond
1 newton (N) = 0.2249 poundforce
1 poundforce (lbf) = 0.4536 kilopond
1 poundforce (lbf) = 4.448 newtons

▶ Kinematics

DEFINITION. A division of mechanics (dynamics) that deals with the geometry of the motion of bodies (displacement, velocity, and acceleration) without taking into account the forces that produce the motion.

DESCRIPTION AND EXAMPLES. Range and pattern of motion of various anatomic joints are good examples of kinematic studies. When a scoliotic deformity, as measured by Cobb's method, is compared at different times, a kinematic study of the disease is being performed.

▶ Kinetic Energy

DEFINITION. The energy that a body possesses because of its velocity. The unit of measure is newton meters or joules (foot poundforce).

DESCRIPTION AND EXAMPLES. An automobile weighing 2500 kg (3290 lb) traveling at 50 km/h (30 mph) possesses a kinetic energy of 241,127 newton meters (167,000 foot poundforce) (Fig. 9-35A). Doubling the speed makes the kinetic energy four times greater (Fig. 9-35B). Therefore, a collision at twice the speed is potentially four times more damaging because of the kinetic energy that must be dissipated at the time of collision.

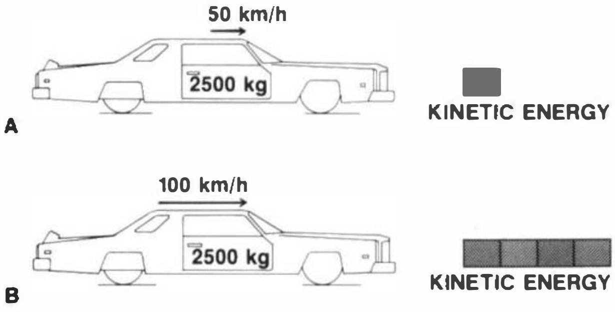


FIGURE 9-35 Kinetic energy.

EXPLANATORY NOTES. Mathematically, the kinetic energy T in joules is expressed as follows:

$$T = \frac{m \times V^2}{2}$$

where m = mass of the body in kilograms (pounds) and V = velocity (speed) in meters per second (feet per second).

▶ Kinetics

DEFINITION. A branch of mechanics (dynamics) that studies the relations between the force system acting on a body and the changes it produces in the body motion.

DESCRIPTION AND EXAMPLES. Observe an athlete starting for the 100-meter dash. From a standstill, he reaches his cruising speed in a few seconds. His muscles apply forces at appropriate points on the bones to produce maximum acceleration of the whole body. A study involving relationships between the various muscle forces applied and the body acceleration produced is a kinetic study of the mechanics of short-distance running.

A study of the forces acting on a scoliotic spine to move it from a deformed position to a more corrected position is another example of kinetics.

▶ Linear

See *Nonlinear*.

▶ Load

DEFINITION. A general term describing the application of a force and/or moment (torque) to a structure. The units of measure are newtons (poundforce) for the force and newton meters (foot poundforce) for the moment.

Because the force and moment are three-dimensional vectors, each having three components, the load may be thought of as a six-component vector.

DESCRIPTION AND EXAMPLES. When a person lifts a weight (Fig. 9-36), the spine is loaded. The L5 vertebra is subjected to weights (forces) from the upper body and the lifted weight. It is also subjected to the bending moment caused by the forces because these forces are away from the center of the L5 vertebra. Thus, the load vector at L5 completely describes all the forces and moments acting on it.

The term *load* is loosely used, causing significant ambiguity. To say that the lumbar spine carries a load of two times body weight implies that the load on the lumbar vertebrae is a *compressive force* of two times body weight, although there are, in addition, a shear force and a flexion bending moment.

▶ Load–Deformation Curve

See *Stress–Strain Diagram*.

▶ Mass

DEFINITION. The quantitative measure of inertia for linear motion. The unit of measure is kilograms (pounds).

DESCRIPTION AND EXAMPLES. Imagine a cart of mass M kilograms (pounds) on frictionless wheels on a table, as shown in Figure 9-37. Apply a given force F newtons (poundforce) to the cart by means of a string. The cart will accelerate into motion. If the mass of the cart is doubled and the same force is applied again, the cart starts moving rather slowly, at precisely half the acceleration. The additional mass doubles the total inertia, thus reducing the acceleration.

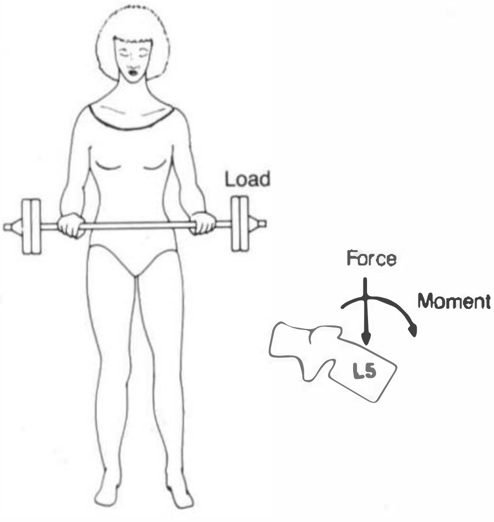


FIGURE 9-36 Load is equivalent to forces and/or moments.

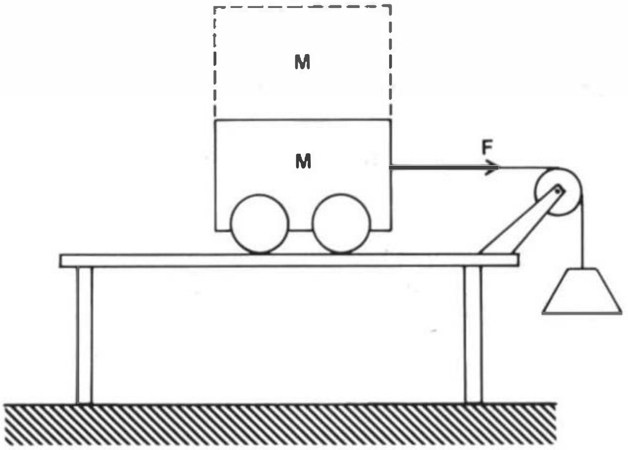


FIGURE 9-37 Mass. The same force F applied to twice the mass will produce half the acceleration.

The above phenomenon was first critically observed by Newton and forms the basis of Newton's second law of motion, which states that a body is accelerated in direct proportion to the amount of force applied. The constant of proportionality between the acceleration and the force is called mass (see also *Newton's Laws*).

Weight of a body is a term that is loosely used. However, it is defined as a measure of the *force* applied to the mass of the body by the earth's gravity. Therefore, when 1 kg of sugar is held in one's hand, the sugar is being pulled downward with a force of 1 kilogramforce (1 kp or 9.81 N).

▶ Mass–Mathematical Element

DEFINITION. In mathematical modeling, mass is used to represent the inertia of bodies to linear motion.

DESCRIPTION AND EXAMPLES. Consider a model of a passenger–automobile system involved in a collision. This is a dynamic situation. Relatively heavy parts of this system (e.g., the automobile, trunk, head, arms, and legs) offer considerable resistance to change in motion because of their mass. Representation of this behavior in a mathematical model is done by the mass–mathematical elements.

▶ Mass Moment of Inertia

DEFINITION. The quantitative measure of inertia for change in angular velocity. The unit of measure is kilogram meter squared (pound foot squared).

DESCRIPTION AND EXAMPLES. A bicycle wheel off the ground, if given a certain speed, continues to rotate for a long time. It does this because of the mass moment of inertia of its rim. This inertia is equal to the mass of the rim times the square of the radius of the wheel. If the same rim mass is concentrated into a small disc around the wheel axle and is given the same speed, it slows down much faster because of its lower radius and the mass moment of inertia. It is this mass moment of inertia of the wheels that keeps a bicycle stable in its upright position when it is in motion.

Another example of this phenomenon is the figure skater who is spinning with both arms abducted out to the sides. Gradually bringing the arms in causes an increase in the spin (Fig. 9-38). This may be explained by the fact that the angular momentum (mass moment of inertia times the spin speed) of the body remains the same. As the arms are brought in, there is a decrease in the radius of gyration and the mass moment of inertia. Because the angular momentum is constant, there is a corresponding increase in the spin speed.

EXPLANATORY NOTES. Mathematically, the mass moment of inertia, called I, is given by the following:

$$I = m \times R^2$$

where m = mass of the body in kilograms (pounds) and R = radius of gyration in meters (feet).

This term is not to be confused with "sectional moment of inertia." See *Moment of Inertia of an Area*.

▶ Materials Testing Machines

See *Testing Machines*.

▶ Mathematical Model

DEFINITION. A set of mathematical equations that quantitatively describes the behavior of a given physical system.

DESCRIPTION AND EXAMPLES. There are situations that cannot be duplicated experimentally. Examples are human spine behavior during pilot ejection from disabled aircrafts, whiplash injury in automobile collision, and landing of the lunar module on the moon. However, these situations can be simulated by a computer using the technique of mathematical modeling.

Consider the simulation of the mechanism of whiplash injury (Fig. 9-39A). The simplest model represents the human body–automobile system as consisting of three masses: the head (M_1), the trunk (M_2), and the automobile including the rest of the body (M_3), as shown in Figure 9-39B. As a first approximation, the cervical spine and the hip joint

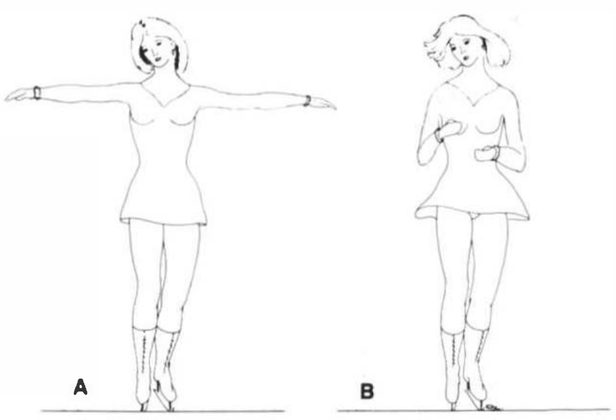


FIGURE 9-38 Mass moment of inertia. **(A)** In this position, the skater is spinning at a slow speed and has great inertia. **(B)** Bringing the arms toward the sides decreases the inertia and increases the speed of spinning.

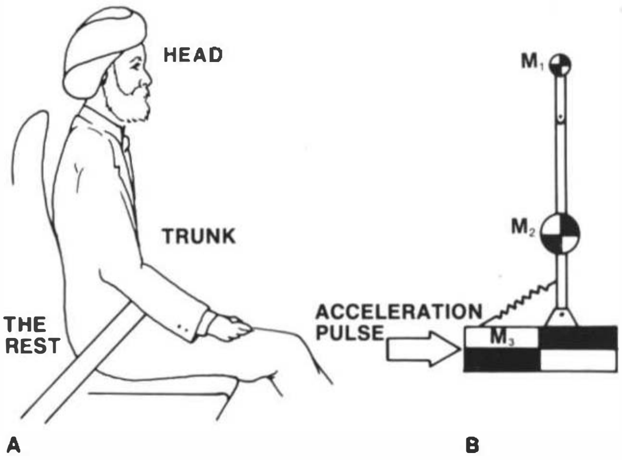


FIGURE 9-39 Mathematical model. **(A)** The physical system may be divided into three entities: the head, the trunk, and the rest (remaining body and all of the automobile). **(B)** A simple mathematical model consists of three masses, M_1, M_2, and M_3, representing the three entities and springlike elements joining the masses.

may be represented as hinges and the seat belt as a spring. The rear-end collision is simulated by a sudden acceleration applied to the mass M_3.

A simple idea of the mechanism may be obtained by building a physical model in wood. However, for more accurate and detailed studies, a complex mathematical simulation in a computer is required. In an analysis of this kind, the effects of such variables as design of the passenger seat, stiffness of the seat belt, body weight, viscoelastic properties of the cervical spine, and severity of the collision may easily be simulated. Complex simulations utilize the capacity of computers to deal with massive quantities of information at very high speed.

A note of caution. Mathematical models are highly versatile tools, and their use is to be encouraged. However, simply stated, they are a set of mathematical equations. Depending upon the data (physical properties of tissues and anatomy of structures) that are put in a model, the same set of equations may represent elastic behaviors of a ligament or a piece of metal. There are three important aspects to a mathematical model. First, a model represents only a part of the physical world it is attempting to model; it incorporates several assumptions (e.g., a vertebra may be considered a rigid body). It is essential to know the set of assumptions that is built into the model. Second, the data set it uses must be of good quality and relevant to the model. For example, the physical cadaveric properties should have been obtained from fresh cadaveric specimens to represent spinal column behavior. Third, it is necessary to check the validity of a model by independent experiments in which model prediction is checked against experimental observations.

▶ Mechanical Stability

See *Elastic Stability*.

▶ Modulus of Elasticity

DEFINITION. The ratio of normal stress to normal strain in a material. The unit of measure for the modulus of elasticity (E) is newtons per square meter or pascals (poundforce per square inch).

DESCRIPTION AND EXAMPLES.. The modulus of elasticity defines the mechanical behavior of the *material* of a structure. It is a measure of the stiffness of the material. Cortical bone has a high modulus of elasticity, and subcutaneous fat, sometimes quite aesthetically, has a low modulus of elasticity. The higher the value, the stiffer the material.

To appreciate the biomechanical interaction between the components of an orthopedic design, the moduli of elasticity of some relevant materials are given in Table 9-3.

The modulus of elasticity is generally obtained by calculating the slope of the linear elastic part of the stress–strain diagram of a material.

▶ Moment

See *Couple*.

▶ Moment of Inertia of an Area

DEFINITION. A measure of the distribution of a material in a certain manner about its centroid. This distribution determines the strength in bending. The unit of measure is meters to the fourth power (inch to the fourth power).

DESCRIPTION AND EXAMPLES. Consider an object with circular cross-section. How should its material be distributed to make it strong against bending and torsional loading? The concept of moment of inertia of an area is useful here. This quantity is also called *sectional moment of inertia*.

As an example, take the comparison of a solid rod and a hollow rod, both with the same cross-sectional area. A rod 10 mm (0.394 in) in diameter, shown in Figure 9-40A, has a sectional moment of inertia about its diameter of 490 mm^4. Redistributing this same material into a hollow tube of 1-mm (0.039-in) thickness results in an outer diameter of 26 mm (1.024 in) and moment of inertia of 3256 mm^4 (0.0078 in^4). Now the bending strength (the moment of inertia divided by the radius; see *Bending*) of the two rods may be calculated. The 26-mm, thin tube is 2.56 times as strong as the 10-mm rod. If similar calculations are made for torsion loading (see Tor-

TABLE 9–3 Moduli of Elasticity

	$\times 10^9$ N/m^2	$\times 10^6$ psi
Stainless Steel*	200	29
Cortical Bone†		
Longitudinal	8.9	1.3
Tangential	4.3	0.6
Radial	3.8	0.5
Methylmethacrylate	1.0	0.15
Cancellous Bone†	0.14	0.02

* American Society for Testing and Materials. Standards for Surgical Implants. Tab. 2. Philadelphia, 1971.

† Evans, F. G.: Mechanical Properties of Bone. Springfield, IL, Charles C Thomas, 1973.

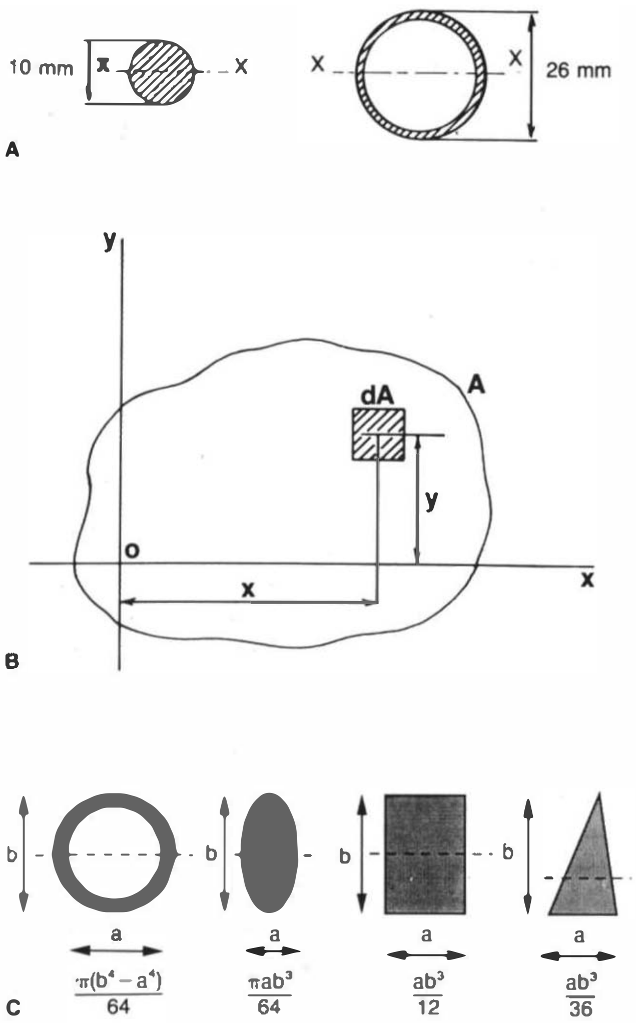


FIGURE 9-40 Area moment of inertia. **(A)** The hollow tube is 2.56 times stronger in bending and 5.12 times stronger in torsion than the solid tube. Both tubes have the same amount of material. **(B)** Mathematical derivation of the equation for the area moment of inertia. **(C)** Area moment of inertia of four cross-sections.

sion), the mass distribution effect is even more dramatic. The corresponding strength ratio is 5.12 in torsion.

One of the best examples of the above concepts is the construction of human bones. They are hollow and cancellous on the inside and hard and cortical on the outside. This provides maximum "strength" for weight and also some neat space for making blood cells.

EXPLANATORY NOTES. The moment of inertia for a given section of a long structure is calculated by the formulas given below. Figure 9-40B shows a sectional area A. Its moments of inertia about the x-axis and y-axis passing through point O are given by formulas:

$$I_{xx} = \int y^2 \, dA$$

$$I_{yy} = \int x^2 \, dA$$

where integration is done over the whole area A. For a circular section, as depicted in Figure 9-41, the following applies:

$$I_{xx} = I_{yy} = \frac{\pi d^4}{64}$$

where $\pi \simeq 3.14$ and d is the diameter.

▶ Momentum

DEFINITION. Linear momentum of a particle or rigid body is the product of its mass and its velocity. The unit of measure is kilogram meters per second (pound feet per second).

Angular momentum of a particle or rigid body is defined as the product of its mass moment of inertia and its angular velocity. The unit of measure is kilogram meter squared per second (pound foot squared per second).

▶ Motion

DEFINITION. The relative displacement with time of a body in space, with respect to other bodies or some reference system.

DESCRIPTION AND EXAMPLES. A general displacement of a rigid body consists of rotation about a certain axis combined with translation along a certain direction. The body that is not constrained has six degrees of freedom. One method used to describe the general motion of a rigid body is to measure translation vectors of any three non-colinear (not lying on one straight line) identifiable points on it. Another method is to break down the observed motion into its two natural components, namely, a translation vector of a chosen point and rotation of the body about an axis through that point. Finally, the most elegant method is to determine the helical axis of motion, which does not require any points on the body (see *Helical Axis of Motion*).

A special case of general motion is motion in which the body moves in a plane and has only three degrees of freedom. This motion may be described by translation vectors of any two points on a rigid

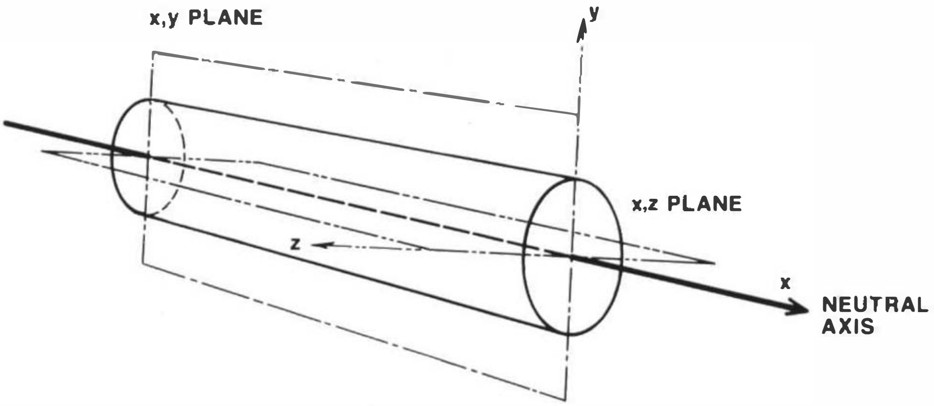


FIGURE 9-41 Neutral axis. The x,z (horizontal) and x,y (vertical) planes shown here are the neutral planes for bending. Their intersection is the neutral axis.

body. Alternatively, a translation vector of a chosen point and a rotation about that point are sufficient. More concisely, it is defined by its instantaneous axis of rotation and the angle of rotation (see *Instantaneous Axis of Rotation*).

Another simplified version of general motion is out-of-plane motion. Again, there are three degrees of freedom: two rotations about mutually perpendicular axes and a translation perpendicular to the plane formed by the axes.

▶ Motion Segment

DEFINITION. A unit of the spine representing inherent biomechanical characteristics of the ligamentous spine at one spinal level.

Physically, it consists of two adjacent vertebrae and the interconnecting soft tissue, devoid of musculature. In the thoracic region, two articulating heads of ribs with their connecting ligaments are also included.

DESCRIPTION AND EXAMPLES. *Motion* segment is the term most commonly used to explain this concept. However, it is grammatically incorrect. An acceptable alternative would be to use the hyphenated form motion-segment. The term *functional spinal unit* (FSU), which is discussed in Chapter 1, is the term of choice, because it adequately describes the concept and is grammatically correct.

▶ Neutral Axis

DEFINITION. A longitudinal line in a long structure where normal axial stresses are zero when the structure is subjected to bending.

DESCRIPTION AND EXAMPLES. If a long structure with symmetrical cross-section is subjected to bending loads in its plane of symmetry, it develops a curvature. The fibers on the convex side of this curvature are then in tension, while those on the concave side undergo compressive stresses. Somewhere in between these two layers is a layer of fibers that has zero normal stress. This is the neutral plane.

When the bending takes place in the vertical (x, y) plane (see Fig. 9-41), the fibers in the horizontal (x, z) plane will be stress-free. This is the neutral plane for this loading. If bending loads were applied in the x, z plane, the x, y plane would be the neutral plane. The line of intersection of the two neutral planes is called the neutral axis. It should be noted that although the normal stress is zero at the neutral axis, shear stresses may be present because of transverse forces.

If the long structure is subjected to torsion about the neutral axis, again the fibers at the neutral axis are unique and have zero shear stress.

From the above discussion, we conclude that the fibers at and around the neutral axis have very low stresses compared with the fibers at the periphery during bending and torsional loads. In human bones, the development of hollow structures, with cortical bone distributed toward the periphery where the stresses are highest, may be an example of biomechanical adaptation.

▶ Neutral Zone

DEFINITION. A part of the range of motion of a body (e.g., vertebra), starting from the neutral position up to the beginning of some resistance offered by the joint. The unit of measure is the meter (foot) for translation, or the radian (degree) for rotation.

DESCRIPTION AND EXAMPLES. Load–displacement curve of most biological tissues and structures is nonlinear. Generally, starting from the neutral position, there is large deformation due to application of a small load. After this "easy" deformation, also called "free-play," there is increasing resistance offered by the tissue. Thus, the motion that takes place between the neutral position and the beginning of significant resistance is the neutral zone.

There are many examples of neutral zones. When the tibia is moved in the sagittal plane with respect to the femur, most of the motion is neutral zone. Only near the ends of the range of motion is there resistance offered by the ligaments. In the upper cervical spine, there is about 40° of axial rotation to each side at the C1–C2 joint. It has been measured experimentally that about 75% of this motion is neutral zone.[12a] See also *Elastic Zone*.

▶ Newton

DEFINITION. The unit of force in the Systeme International d'Unites. One newton is the amount of force required to give a 1-kg mass an acceleration of 1 meter per second per second.

DESCRIPTION AND EXAMPLES. A newton is the best unit of force because it is not dependent upon the earth's gravitational field for its definition, in contrast to the kilopond and poundforce. Thus, it is the universal measure of force. It is based upon Newton's second law of motion, which states that a force in newtons equals the mass in kilograms multiplied by acceleration in meters per second per second. The name has been chosen to honor the man who laid the foundation of modern mechanics.

The abbreviation is N, and the conversion factors to other units of force are as follows:

1 newton (N) = 0.2248 poundforce
1 poundforce (lbf) = 4.48 newtons
1 newton (N) = 0.1020 kilopond
1 kilopond (kp) = 9.806 newtons
1 newton (N) = 100,000 dynes

▶ Newton's Laws

DEFINITION. Isaac Newton (1642–1727) postulated three laws that form the basis of mechanical engineering science. These are based on his observations, and since their inception they have been shown to be in agreement with other observations. The laws and their simple interpretations are as follows:

1. A body remains in a state of rest or uniform motion in a straight line until it is acted upon by a force to change that state. In other words, a book on a table will stay there forever, and a golf ball once hit will keep traveling with constant velocity (assuming no air resistance or gravity) until some force interferes.
2. The rate of change of momentum is equal to the force producing it. Stated differently, force equals mass times acceleration. For rotatory motion, moment equals mass moment of inertia times angular acceleration.
3. To every action there is an equal and opposite reaction. A classical example is a rocket. The exhaust gases are pushed toward the rear (action) while the rocket is pushed forward (reaction).

▶ Nonlinear

DEFINITION. The concepts used in characterizing the relationship between two variable quantities.

If the ratio of one variable quantity to the other variable quantity is constant through a defined range of values, then the relationship is said to be linear within that range. Any deviation from linearity is defined as nonlinear behavior. If this relationship is plotted on graph paper, only the linear relation will be a straight line.

DESCRIPTION AND EXAMPLES. A tracing of an actual load–deformation curve* of a motion segment including two vertebrae subjected to compressive loading is shown in Figure 9-42. The initial portion of the curve, 0–3 mm of deformation, is highly nonlinear, which may be demonstrated by drawing a straight line from the origin to the 3-mm point on the curve. If a similar line is drawn from the 3-mm point to the 5-mm deformation point on the curve, it is seen that the deviation from linearity of the actual curve is rather small. Therefore, for all practical purposes, the latter portion of the curve is considered linear. For more precise quantification of nonlinearity, see the explanation below.

EXPLANATORY NOTES. If one needs to be precise, the percentage of nonlinearity may be defined. This is the percentage ratio of maximum deviation from the straight line (f) to the highest range (F), as shown in Figure 9-42. In this example, the 0–3-mm range has nonlinearity of 44%, and the corresponding figure for the 3–5-mm range is 3.7%.

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* Unpublished data, Biomechanics Laboratory, Yale University School of Medicine, New Haven, CT.

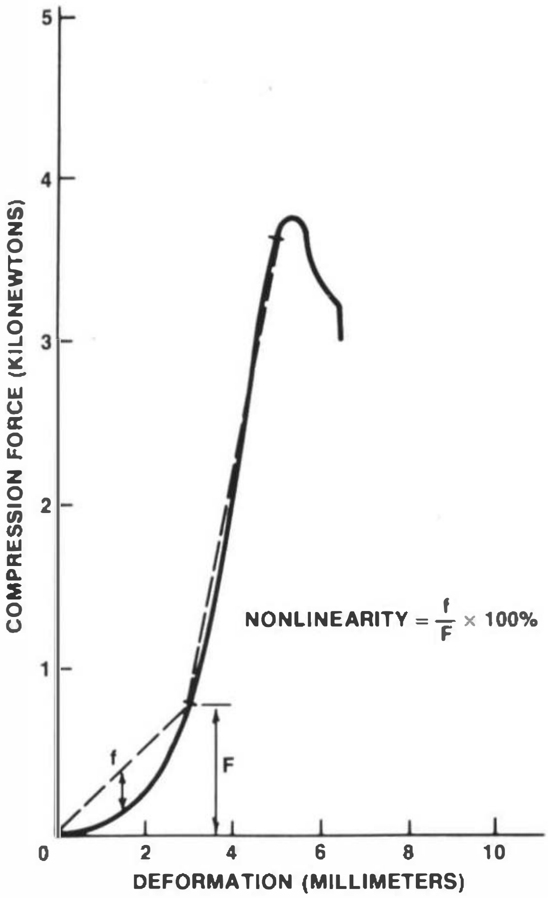

FIGURE 9-42 Nonlinear deformation curve. The quantity *nonlinearity* specifies the amount of departure from a straight line.

▶ Normal Stress

DEFINITION. The intensity of force perpendicular to the surface on which it acts. The unit of measure of normal stress is newtons per square meter or pascals (poundforce per square inch).

DESCRIPTION AND EXAMPLES. When a structure such as a long bone or an implant is subjected to tension, compression, and/or bending, the axial fibers of the structure are subjected to normal stress. Figure 9-43A shows a bone being subjected to bending moments. Normal stress acts perpendicular to a surface. It can be either a positive normal stress, as in tension, or a negative normal stress, as in compression. In bending, there is compressive stress on the concave side and tensile stress on the convex side of the neutral axis (Fig. 9-43A, B). There is also a normal stress in fibers oriented 45° to the longitudinal axis when a tubular structure is subjected to torsional loading. This is the mechanism involved in spiral ski fractures. A simple equation for the normal stress is given below.

EXPLANATORY NOTES. The common symbol for denoting the normal stress is the Greek letter σ (sigma). Mathematically, the normal stress is given by the following formula:

$$\sigma = \frac{F}{A}$$

where F is the applied force in newtons (poundforce) and A is the area in square meters (square feet).

▶ Out-of-Plane Motion

DEFINITION. A motion of a rigid body in which the body does not move in a single plane.

DESCRIPTION AND EXAMPLES. Out-of-plane motion is a combination of translation and rotation. It has three degrees of freedom: rotation about two mutually perpendicular axes, forming a plane, and translation perpendicular to that plane.

The lateral bending of the spine exemplifies this type of motion. A vertebra in the spine undergoing lateral bending is shown in Figure 9-44. The vertebra rotates about a horizontal axis (anteroposterior) and translates out of the sagittal plane into the horizontal plane. It goes from position 1 to 2 to 3, as shown in Figure 9-44. (Because of the coupling,

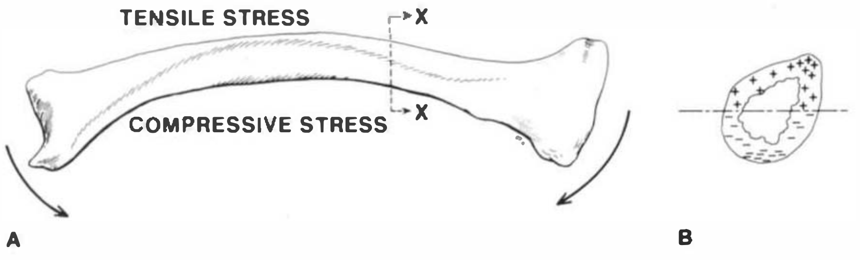

FIGURE 9-43 Normal stress. **(A)** A bone subjected to bending moment. **(B)** Cross-section at x-x. The tensile stresses (+) are on the convex side, while compressive stresses (−) are on the concave side of the bend.

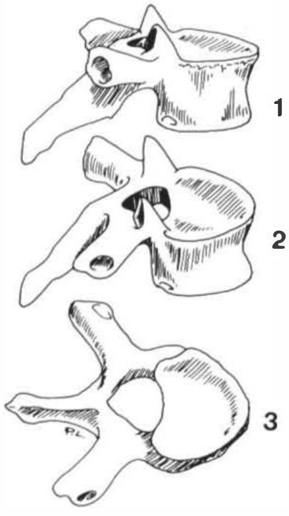

FIGURE 9-44 Out-of-plane motion.

there may also be axial rotation, which is not shown here.) In contrast to this motion, plane motion is depicted in Figure 9-45.

Plane Motion

DEFINITION. A motion in which all points of a rigid body move parallel to a single plane.

DESCRIPTION AND EXAMPLES. This motion is a combination of translation and rotation. It has three degrees of freedom: translations along two mutually perpendicular axes and rotation about an axis perpendicular to the other two axes.

Flexion/extension of the spine (Fig. 9-45) translates a vertebra in the horizontal and vertical directions. At the same time it rotates the vertebra about an axis perpendicular to the sagittal plane, from position 1 to 2 to 3. The motion takes place in a single plane. This is in contrast to out-of-plane motion (see *Instantaneous Axis of Rotation*).

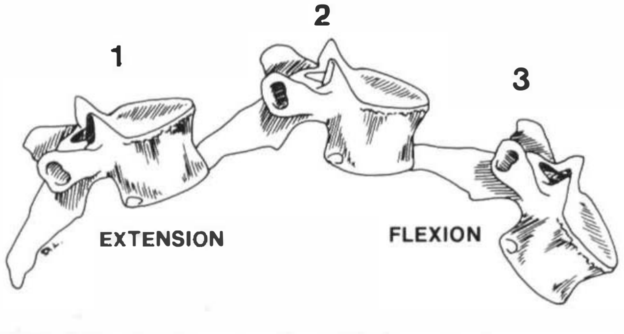

FIGURE 9-45 Plane motion. All the particles constituting the vertebra move in paths that lie in parallel planes.

Plasticity

DEFINITION. The property of a material to permanently deform when it is loaded beyond its elastic range.

DESCRIPTION AND EXAMPLES. The stress–strain curve for a material is shown in Figure 9-46. When the material is loaded beyond its elastic range (AB), it enters the plastic range (BE). Unloading within the plastic range, as shown by the line CD, always produces permanent deformation, shown as AD. On reloading, the material generally passes back into the plastic range as if unloading and reloading had not taken place.

To understand and visualize plastic behavior, engineers make use of a simple model consisting of a friction block connected in series with a spring, as shown in the lower portion of Figure 9-46. The motion of the free end of the spring describes the behavior of the material subjected to a force.

The application of the force produces deformation of the spring but no motion of the friction block. This corresponds to the elastic material behavior

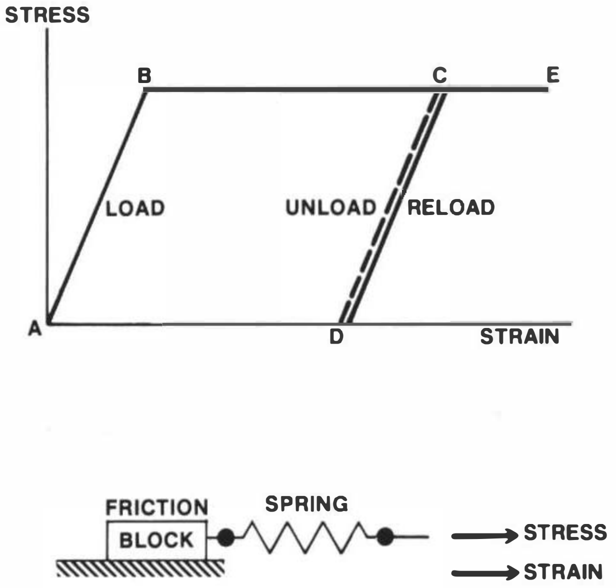

FIGURE 9-46 Plasticity. A block being pulled by a force via a spring is used to explain the elastic and plastic behaviors. AB = elastic loading (spring); BC = plastic deformation (friction); CD = elastic unloading (spring); AD = permanent deformation (friction); DCE = reloading cycle; DCE ≡ ABC.

(AB). This behavior continues until the force reaches a value that is just sufficient to move the friction block. This corresponds to the yield point B for the material. A small additional force produces motion of the entire model with no additional deformation of the spring. This behavior corresponds to the plastic behavior of the material (BC). On release of the load (at C) the spring recoils (CD), but the friction block does not go back to its original position, thus producing permanent deformation (AD). Reloading (DCE) duplicates the loading behavior (ABC).

The permanent deformation of a ligament after it has been subjected to greater than 40% of its ultimate load is one of the examples of plastic behavior.[21] Under such high load, collagenous fibers glide over one another, in the manner of the friction block.

▶ Plastic Range

DEFINITION. If a specimen is loaded beyond its elastic range, it enters the plastic range.

DESCRIPTION AND EXAMPLES. In Figure 9-46, AB is the elastic range, and BE is the plastic range. Modeling the elastic and plastic behavior by a friction block connected in series with a spring is a simple way to visualize what is happening. The larger the plastic range until failure, the higher the ductility and energy absorption capacity of the material.

When Dizzy Gillespie first began playing his trumpet as a young man, the deformation of his cheeks was in the elastic range, and they returned to their normal size. In later years, with strong forces and perhaps some alterations in the tissues, his cheeks went into the *plastic range of deformation* (Fig. 9-47).

FIGURE 9-47 Plastic range. *(Photograph by Ozier Muhammad. In JET Magazine, November 10, 1977. Copyright © 1977 by Johnson Publishing, Inc., Chicago.)*

▶ Poisson's Ratio

DEFINITION. The ratio of transverse to axial strain. It is generally represented by the Greek letter ν (nu). Since it is a ratio, there is no unit of measurement.

DESCRIPTION AND EXAMPLES. Take a rubber band and stretch it. A careful observation shows that it gets thinner when stretched. The reverse will happen if a piece of rubber such as a pencil eraser is compressed. In both instances, the changes in the transverse dimensions take place because, in general, the material is incompressible (this is not true of gases). The volume remains constant. An increase or decrease in the length is accompanied by a corresponding decrease and increase, respectively, in the transverse dimensions. Poisson's ratio quantifies this material behavior.

When a screw is employed to fix a fracture, Poisson's ratio is in action. As the screw is tightened, the bone is compressed and the screw is lengthened. Because of Poisson's effect, the thread diameter in the bone expands and the screw diameter decreases. Therefore, a screw that is all right under no load may become loose when tightened. The chances of this happening are small if the standard hole is drilled for a given screw. (This phenomenon is different from the mechanism of a screw pullout, in which the bone between the threads is stripped by driving the screw too hard.)

This effect was first discovered by Poisson in the early 19th century. The theoretic upper limit for this ratio for any material is 0.5. Bone and steel have a Poisson's ratio of approximately 0.3.

EXPLANATORY NOTES. If a rubber eraser, cylindrical in shape, is subjected to compression, the axial strain is the change in unit length $(L_1 - L_2)/L_1$, and the transverse strain is the change in unit diameter $(D_2 - D_1)/D_1$. The ratio of the first to the second strain is Poisson's ratio.

▶ Polar Moment of Inertia

DEFINITION. A property of the cross-section of a long structure that quantifies the distribution of the material about its long axis and is a measure of its torsional strength. The unit of measure is meter to the fourth power (inch to the fourth power).

DESCRIPTION AND EXAMPLES. When a long structure is subjected to torsion, the maximum shear stress and angle of rotation are functions of the torque applied, the material properties, and the geometry of the cross-sections. The last quantity is characterized by the polar moment of inertia of the section. The more distant the mass is with respect to the axis of torsion, the greater is the polar moment of inertia. The general expression for determining the polar moment of inertia for a given section is given below.

The juncture of the middle and distal thirds of the tibia fractures more frequently than any other area. This is the shaded area shown in Figure 9-48A. Fracture is common largely because the polar moment of inertia at this distal section is minimal compared with the rest of the tibia. (In reality, it is the polar moment of inertia divided by the radius that determines the shear stress in the tibia; see *Torsion*.) The diagram shows the actual polar moment of inertia of a human tibia on the vertical axis (ordinate) and the distance from the distal end on the horizontal axis (abscissa).[17]

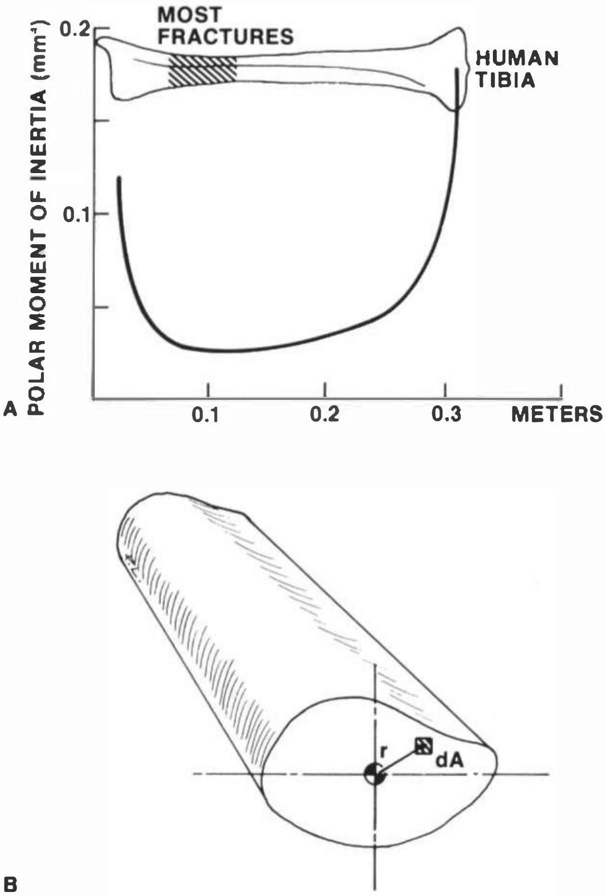

FIGURE 9-48 Polar moment of inertia. (A) Experimentally determined values for the human tibia. (B) Mathematical derivation.

EXPLANATORY NOTES. Referring to Figure 9-48B, the mathematical expression for the polar moment of inertia is as follows:

$$J = \int r^2 \, dA$$

where J = polar moment of inertia of an area; dA = small area away from axis; r = radius to the center of dA; and ∫ = integration over the whole section.

▶ Potential Energy

DEFINITION. Energy that may be stored within a structure as a result of deformation or displacement of that structure. The unit of measure is newton meters or joules (foot poundforce).

DESCRIPTION AND EXAMPLES. By testing a spring in a testing machine, its load–deformation graph may be drawn as shown in Figure 9-49A. The shaded area under the graph is the potential energy stored in the spring for the given deformation. This energy is recoverable in the form of useful work. A clock with a spring-wound motor works on this principle. Other mechanical clocks work by storing the potential energy in the form of displacement of the weights against gravity.

Pulling back a bow stores the potential energy of deformation of the bow (Fig. 9-49B). On release of the arrow, this energy is converted into the kinetic energy of the arrow (see *Kinetic Energy*).

EXPLANATORY NOTES. Referring to Figure 9-49A, the formula for potential energy is as follows:

$$U = \frac{F \times D}{2}$$

where U = potential energy in Nm or J (lbf ft); F = force of deformation in N (lbf); and D = deformation in m (ft).

The above formula is valid for a force proportional to deformation, as shown in Figure 9-49A. For other force–deformation relationships, the equation will be in an integral form.

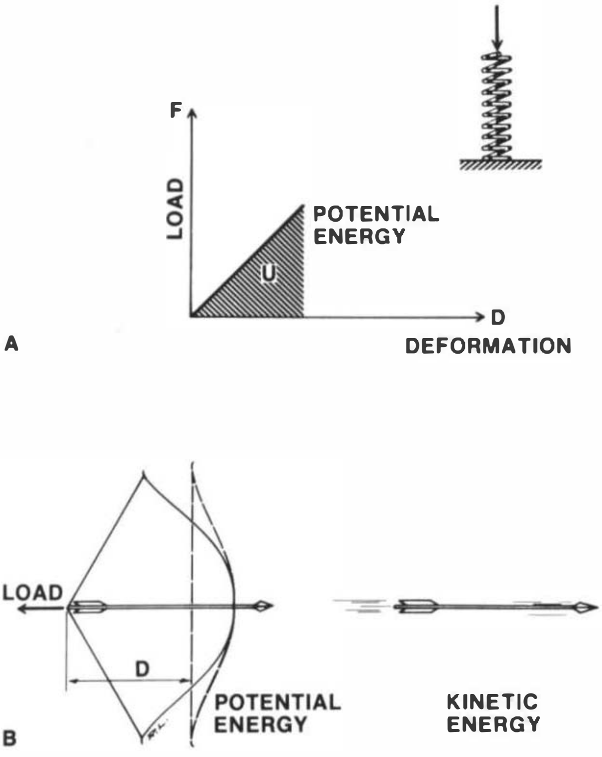

FIGURE 9-49 Potential energy. (A) The deformation of a spring stores potential energy. (B) Potential energy may be converted into kinetic energy.

▶ Principal Planes

DEFINITION. Planes in a structure where the shear stress is zero and only the normal stresses are present.

When a structure is subjected to loads, stresses are created within it. Looking at a single point in the structure, innumerable planes may pass through that point in various directions. On each of these planes there are stresses perpendicular to the planes (normal stresses) and parallel to the planes (shear stresses). The proportion of normal to shear stresses varies with different planes. Those planes in which there are only normal stresses (*i.e.*, no shear stresses) are called the principal planes (by definition). Because the material generally fails because of excessive tensile, normal stresses, determination of directions along which these stresses are highest (principal planes) is of great practical importance. At a given point in a structure subjected to a load, there are always three principal planes, and they are perpendicular to each other. One or two of these planes may not be of any interest in simple loading situations, but they are still there. A few examples are described below.

DESCRIPTION AND EXAMPLES. If a cylindrical test specimen were taken from a bone and were subjected to tension, there would be only one principal plane of interest, the plane perpendicular to the axis, as shown in Figure 9-50A. The other two planes are in axial directions.

One of the mechanisms of ski fracture injuries in the tibia is that of torsional loading, applied by a transverse force to the tip of the ski (Fig. 9-50B1). If one were to look for the principal planes in the tibia subjected to torsional loading, there would be two planes mutually perpendicular to each other and at

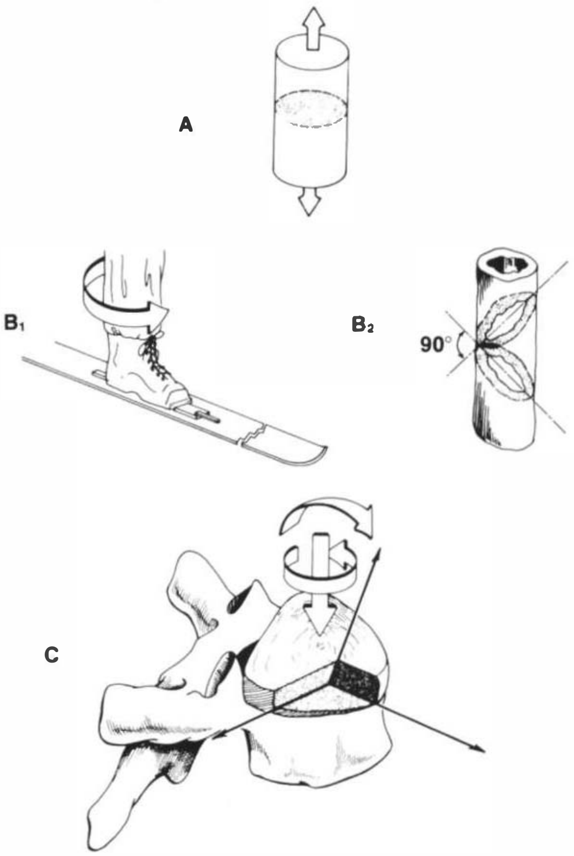

FIGURE 9-50 Principal planes. At any point, there are always 3 planes, which are mutually perpendicular and carry only normal stresses. Some planes have zero normal stresses. (A) Tensile force—one plane perpendicular to the force. (B) Torsion—two planes at 45° and 135° to the torsion axis. (C) Complex load—three planes arbitrarily oriented.

45° and 135° to the tibial axis (Fig. 9-50B2). The third plane is in the axial direction.

A more complex loading situation is that of the disc when the spine undergoes lateral bending (Fig. 9-50C). The disc carries at least a vertical load, a bending moment, and an axial torque. In such a complex situation, all three principal planes are of interest. Mathematical techniques are available to determine these planes, given the structure and the loading situation.

▶ Principal Stresses

DEFINITION. The stresses normal to the principal planes are called the principal stresses. The unit of measure is newtons per square meter or pascals (poundforce per square inch).

DESCRIPTION AND EXAMPLES. At a point in a three-dimensional body subjected to complex loads, it is always possible to find three mutually perpendicular planes in which shear stress is zero. Such is the case for a vertebra when the spine is subjected to bending. For a biaxial stress field (loading in one plane), there are two principal stresses of interest. An example is a ski fracture. In a uniaxial stress field (tension or compression applied to a bar) there is only one principal stress of interest. (For further explanation, see *Principal Planes*.)

▶ Radius of Curvature

DEFINITION. The radius of a circle that fits a given curve at a point as snugly as possible. The unit of measure is meters (feet).

The radius of curvature is a measure of smoothness or crookedness of a curve. The greater the radius of curvature, the smoother or less crooked the curve.

Cobb's angle for measuring scoliotic spines does not give the most accurate quantification of the spinal curvature. Two scoliotic curves that are markedly different, as shown in Figure 9-51A, have the same Cobb's angle measurement ($\theta_1 = \theta_2$). A more precise and descriptive quantification may be given by also measuring the radius of curvature at the apex of the curves (R_1 and R_2 in Fig. 9-51A). If this method is adapted to a clinical situation, templates for measuring the radius of curvature could be used.

DESCRIPTION AND EXAMPLES. To find the amount of curvature at a point X of a curve (Fig. 9-51B), mark two points A and B, one on each side at an equal distance from X. Draw a circle through the three points. (This is always possible, and it will have a unique radius.) Now move the points A and B closer to X and draw a new circle. Keep repeating this procedure until the points A and B are practically the same as X. The radius of the circle at that moment, R, is the radius of curvature of the curve at that point X. The curvature is defined as the reciprocal or inverse of the radius of curvature.

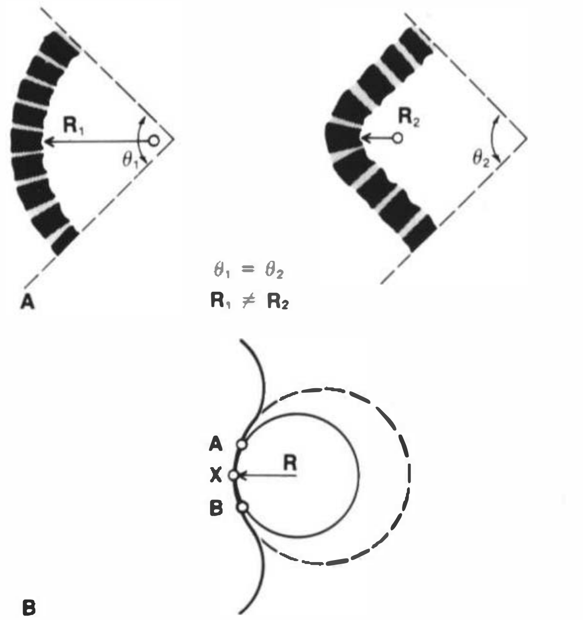

FIGURE 9-51 Radius of curvature. **(A)** Two scoliotic spines may have the same Cobb's angle but different shapes. The latter may be defined by the radius of curvature. **(B)** Mathematical derivation.

▶ Range of Motion

DEFINITION. Displacement from one extreme to the other extreme of the physiologic range of translation or rotation of a joint, for each of its six degrees of freedom. The units of measure are meters (feet) and degrees, respectively.

DESCRIPTION AND EXAMPLES. In an FSU of the cervical spine, one range of motion would be the number of degrees rotated or translated between the points of full active voluntary extension and full active voluntary flexion. See also *Elastic Zone* and *Neutral Zone*.

▶ Rate of Deformation

See *Stress and Strain Rates*.

▶ Rate of Loading

See *Stress and Strain Rates*.

Relative Motion

DEFINITION. Between two moving objects, the motion of one object observed from the perspective of the second object.

DESCRIPTION AND EXAMPLES. Consider the following. A person sitting in a car traveling at 50 km/h (30 mph) in front of a camera set up on the ground with an open shutter holds out his arm through the car window and drops a ball from his hand. Now look closely at the relative motions of the ball with respect to the car and to the camera. Assuming that there is no air resistance to the falling ball, its motion outside the car is the same as if it were dropped inside the car. It is known that the ball inside the car will drop straight downward. Therefore, the relative motion of the ball with respect to the car is a vertical, downward, straight-line motion. Now, consider the picture taken by the camera. The ball has traced a parabolic path on the picture. This is due to the fact that the ball has a horizontal velocity of 50 km/h with respect to the camera at the start of the fall. This motion is being supplemented by the increasing vertical velocity of the ball as it falls. The result is a parabolic motion, just like that of a projectile fired horizontally.

Therefore, the relative motion of an object is dependent upon the motion of the observer or the frame of reference.

The spine is a collection of vertebrae connected in a chainlike structure. When a person bends forward, the head moves with respect to the ground. This is sometimes termed an *absolute* motion. The motion of the head with respect to the C1 vertebra is the *relative motion.*

Relaxation

DEFINITION. The decrease in stress in a deformed structure with time when the deformation is held constant.

DESCRIPTION AND EXAMPLES. Let a specimen of viscoelastic material be stressed and then its deformation fixed. The internal stresses decrease with time exponentially, reaching a lower value (zero at infinite time). This phenomenon is *relaxation.*

If one jumps up, grabs a branch of a nearby great tree, and pulls it so that the tip of the branch touches the ground but does not break, a certain force is required. The force necessary to hold the branch tip to the ground diminishes with time. This diminution of force is due to relaxation.

As another example, we may take an FSU that has been instrumented with a force-measuring transducer (Fig. 9-52A). The clamping vise is tightened to produce a certain deformation and internal stress in the FSU. Let the deformation be kept constant (Fig. 9-52B). The force transducer reading then indicates a decrease with time (Fig. 9-52C). The curve of force versus time is called the relaxation curve. It is the manifestation of the viscoelastic properties of the FSU.

Retardation

See *Acceleration* and *Angular Acceleration.*

Rigid Body

DEFINITION. A collection of particles joined together rigidly.

Theoretically speaking, a rigid body when sub-

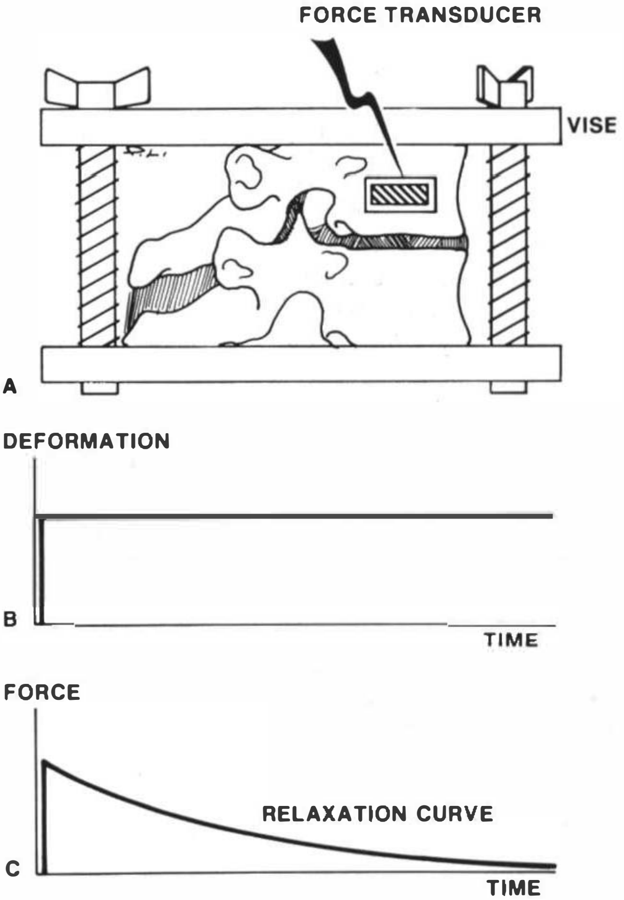

FIGURE 9-52 Relaxation. (A) Experimental setup. (B) A constant deformation is applied. (C) The force in the specimen decreases with time.

jected to finite loads must not deform. However, for practical purposes, the following definition is used: A body is said to be rigid if its deformation as compared with the other bodies (the so-called flexible bodies) in the system is small within a given range of the loads applied.

DESCRIPTION AND EXAMPLES. During flexion of the spine, the motion of the head with respect to the pelvis can be fully accounted for by the deformations of the discs, the vertebral arches, the spinous processes, and the ligaments. Deformation of the bodies of the vertebrae is negligible, comparatively. Thus, under these loading conditions, the vertebral bodies may be considered as rigid bodies in the spine system.

▶ Rotation

DEFINITION. Motion of a rigid body in which a certain straight line of the body or its rigid extension remains motionless. This line is the axis of rotation. The unit of measure is radians or degrees.

DESCRIPTION AND EXAMPLES. The spin of a tennis ball is a rotation about an axis through one of its major diameters. A bicycle wheel rotates about its hub axis.

All joints of the body have predominantly rotatory motions. The axis of motion, however, may vary during the complete range of motion. The variation may be in location as well as in orientation. Consider the knee. During the first 70–80° of extension, the axis of rotation of the tibia with respect to the femur is approximately perpendicular to the femur axis. However, its position changes in a well-defined pattern as extension progresses. In the last 10–20° of extension, it also undergoes axial rotation. This implies that near the end of extension, the axis of rotation lies at an angle different from 90° to the femur axis.

EXPLANATORY NOTES. Rotation is not really a vector, because it does not obey the basic vector rules. However, small rotations (5°) may be approximated as vectors for ease of mathematical considerations. In such a case, the axis of rotation becomes the direction of the vector. The length of the vector then represents the magnitude of rotation.

▶ Scalar

DEFINITION. A quantity that is completely defined by its magnitude.

DESCRIPTION AND EXAMPLES. Room temperature is a scalar that may be measured and defined in Celsius or Fahrenheit scales. Unlike vectors, scalars do not have direction, and therefore they are not dependent upon a coordinate system.

Other examples of scalar quantities are volume, density of a material, and the energy absorption capacity of bone and muscle mass.

▶ Sectional Moment of Inertia

See *Moment of Inertia of an Area.*

▶ Shear Modulus

DEFINITION. The ratio of shear stress to the shear strain in a material. The unit of measure is newtons per square meter or pascals (poundforce per square inch).

DESCRIPTION AND EXAMPLES. Shear modulus is a material property. The relation of shear modulus to shear is analogous to the relation of modulus of elasticity to tension and compression. As an example, take a cube of rubber with convenient measurements of 1 cm in length, width, and depth (Fig. 9-53). Application of shear force (parallel to the upper surface) deforms the cube. Because the length and area of the chosen cube are unity, the shear force and deformation are also the shear stress and strain, respectively. The shear modulus is the ratio of the two (by definition). However, it is not an independent property of a material but is related to the modulus of elasticity and Poisson's ratio. Its exact relationship is described by the formula below.

Materials with a low modulus of elasticity (rubber, ligament) have a lower shear modulus, and

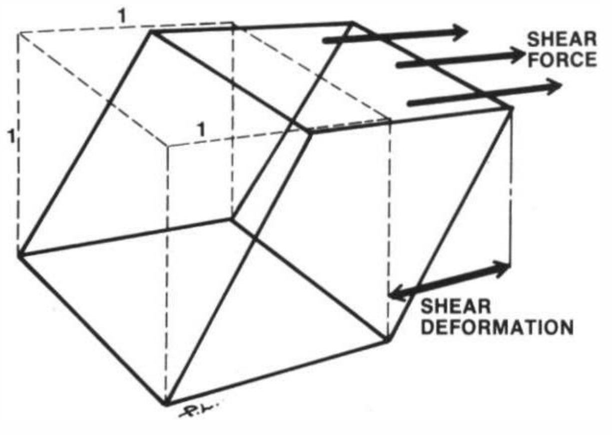

FIGURE 9-53 Shear modulus. Shear force deforms a cube into a parallelepiped. The ratio of the shear force to the shear deformation is a measure of the shear modulus of the material.

those with a higher modulus of elasticity (steel, bone) have a higher shear modulus. The shear modulus is generally 37–40% of the modulus of elasticity.

EXPLANATORY NOTES. The relationship between the shear modulus G and the other two material constants is given by the following formula:

$$G = \frac{E}{2(1 + \nu)}$$

where E = modulus of elasticity, N/m^2 (psi) and ν = Poisson's ratio. For most materials, G = 0.38 E.

Shear Stress

DEFINITION. The intensity of force parallel to the surface on which it acts. The unit of measure is newtons per square meter or pascals (poundforce per square inch).

DESCRIPTION AND EXAMPLES. Scissors, sometimes called shears, are in fact just that, instruments that operate by producing shear stress (Fig. 9-54A). They function effectively in cutting the material. Their mechanism consists of the use of equal and opposite forces applied to the material by means of the two blades to create nearly pure shear stress in the material (Fig. 9-54B).

Shear stress is somewhat synonymous to torsion, but the failure mode in torsion may not be due to shear. An example to illustrate this point is the ski fracture. Torsion to the tibia is applied because of rotation of the foot with respect to the knee. The transverse and axial sections of the tibia are subjected to shear stresses, while sections +45° and −45° to the long axis are under normal stress (tension and compression; Fig. 9-54C).

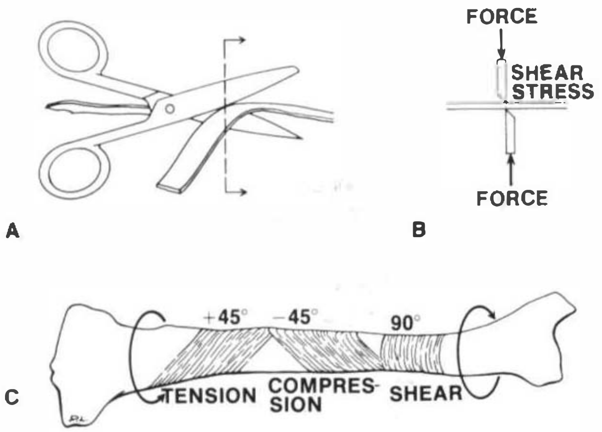

FIGURE 9-54 Shear stress. (A, B) Scissors cut the material by producing shear stresses. (C) Torsion produces different types of stresses in different directions.

It can be shown theoretically that the magnitudes of the shear and the two normal stresses are the same. The observation that torsional loading generally produces spiral fractures implies that bone is weaker in tension than in shear.

In the spine, shear is particularly important at the L5–S1 level because of the lumbar lordosis. A compressive load results in shear load at this joint.

S.I. Units

DEFINITION. The International System of Units consists of a set of basic units. For biomechanical purposes, these are: meter (m) for length, kilogram (kg) for mass, seconds (s) for time, and radian (rad) for angle. For other units and the conversion factors between the S.I. and the English/U.S.A. units, see *Conversion Tables*.

Sine

See *Trigonometric Functions*.

Spherical Coordinates

See *Coordinate System*.

Spring–Mathematical Element

DEFINITION. An elastic mathematical element. It is used in conjunction with other elements to mathematically represent observed phenomena in which elastic behavior is present.

DESCRIPTION AND EXAMPLES. Elastic behavior is depicted graphically in Figure 9-55A. The loading and unloading curves (opposite arrows) for such a behavior are exactly the same. The stiffness coefficient, the slope of the curve, quantifies the spring behavior. If it is constant, the spring element is a linear spring and the load–deformation curve is a straight line. If the coefficient varies with load, it is a nonlinear spring and the curve is no longer a straight line (Fig. 9-55A).

A model representing the action of shooting an arrow from an archer's bow basically consists of a spring. The spring, in this case, is most probably nonlinear and represents the combined properties of the bow and the string in the direction of the pull.

All biologic materials are viscoelastic in their mechanical behavior. In order to mathematically simulate this, a combination of viscous and elastic elements is used. The load–deformation curve of a

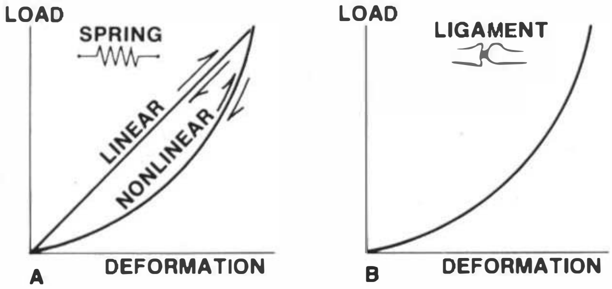

FIGURE 9-55 Spring. **(A)** Representation of the elastic (springlike) behavior of a material by load-deformation curve. The behavior can be linear or nonlinear. **(B)** Such a curve can represnt the behavior of a ligament.

rabbit cruciate ligament is shown in Figure 9-55B.[22] Note that the concavity of the curve is toward the load axis. This implies that the spring element is nonlinear, and its stiffness coefficient increases with load. In other words, the cruciate ligament of the rabbit becomes stiffer with increasing load and thus provides greater stability as it tightens. The experiment was performed at a certain loading rate. For a different loading rate, there would be a different curve because of the viscous part of the viscoelastic behavior of the cruciate ligament.

▶ Stability

See *Elastic Stability* and *Clinical Stability*.

▶ Statics

DEFINITION. The branch of mechanics that deals with the equilibrium of bodies at rest or in motion with zero acceleration.

DESCRIPTION AND EXAMPLES. This is probably the most useful part of mechanics for solving day-to-day orthopedic biomechanical problems. The tool most often used for solving the problem is free-body analysis, using equilibrium equations (see *Free-Body Analysis* and *Equilibrium*).

To apply traction to the spine of a patient lying in bed, force is applied to the head on one side and to the femur on the opposite side (Fig. 9-56). If 180 N (40.2 lbf) are applied to the head and 250 N (55.9 lbf) to the femur, how much of the force is being applied to the spine? Answers to such questions come from the science of statics. In this example, the friction forces are generated between the body and the bed as the traction is applied. The friction forces under the pelvis, back, and head are assumed to have values and directions as shown in Figure 9-56. From the laws of statics, in this example the cervical spine is subjected to a traction of 170 N (38 lbf). Of course, if there were no friction forces, the two tractions and the spine force would all be equal.

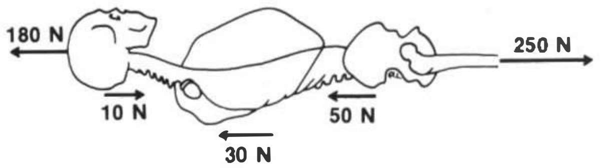

FIGURE 9-56 Statics. Various forces are generated between the body and the bed when the spine is subjected to traction. Quantitative analysis of the forces is a study in statics.

▶ Static Load

DEFINITION. A load applied to a specimen is called static if it remains constant with respect to time. Its antonym is dynamic load.

DESCRIPTION AND EXAMPLES. Imagine a person standing on a nice sunny beach. The depth of his footprints is an indication that he is applying a static load to the earth. If he jumps up and comes down on the sand, he is applying a dynamic load. The footprints will be deeper, indicating higher loads. In most cases, dynamic loads are additions to static loads already present, and therefore they produce higher stresses.

A smooth recovery from anesthesia after a Harrington rod procedure for scoliotic correction is of great advantage in keeping the *static* distraction force low. However, if the recovery is violent because of coughing, additional *dynamic* forces are applied to the system. A distraction hook may penetrate through the bone as a result.

▶ Stiffness

DEFINITION. A measure of resistance offered to external loads by a specimen or structure as it deforms. This phenomenon is characterized by the stiffness coefficient.

DESCRIPTION AND EXAMPLES. Stiffness and elasticity are two similar but quite different concepts. The former represents mechanical behavior of a *structure* including the material, shape, and size, while the latter is a pure *material* property. For example, stainless steel has a higher modulus of elasticity as a material than cortical bone. This is indicated by the stress–strain curves in Figure 9-57 A; the steel has a

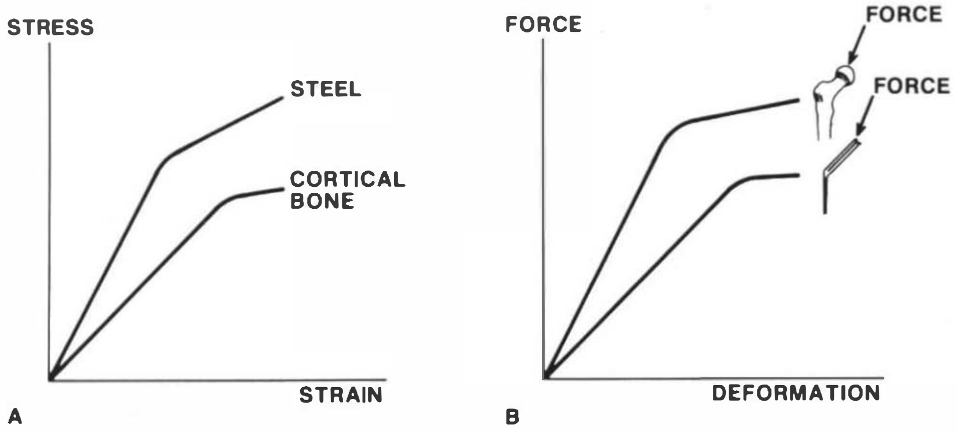

FIGURE 9-57 Stiffness. (A) Steel is a stiff material as compared to bone. It has a higher modulus of elasticity. (B) The femoral neck is *stiff*er than any hip nail. This is due to advantageous distribution of material in the femoral neck.

higher slope value. A hip nail has lower stiffness than the neck of the femur in the characteristic loading patterns, as shown in Figure 9-57B. This discrepancy is explained by an analysis of the amount and distribution of the two materials. The nail has a smaller cross-section, and its material is relatively near to its axis. In contrast, the femur neck has a bigger cross-sectional area, and its material is distributed farther away from the axis, thus providing much more resistance to bending through the larger moment of inertia of its cross-section.

A spine with ankylosing spondylitis is a structure that has a high stiffness. The supple spine of a newborn has a low stiffness.

▶ Stiffness Coefficient

DEFINITION. The property of a structure defined by the ratio of force applied to the deformation produced. It quantifies the resistance that a structure offers to deformation.

For a particular structure, the slope of its load–displacement curve is the stiffness coefficient. When the curve is linear, the slope and therefore the stiffness coefficient is a constant. For a specimen with nonlinear stiffness behavior, the stiffness coefficient varies with the magnitude of the load. The unit of measure is either newtons per meter (poundforce per foot) or newton meters per radian (foot poundforce per degree).

DESCRIPTION AND EXAMPLES. One may take a rubber band and hang a 100-g (0.22-lb) weight from its end. If the rubber band is stretched 1 cm (0.033 ft), then one may calculate the stiffness coefficient of the rubber band as 98.1 N/m (6.67 lbf/ft). (A 100-g [0.22-lb] weight applies 0.981 N [0.22 lb] of force.) If an additional 100 g (0.22 lb) produces another centimeter (0.033 ft) of stretch, the rubber band exhibits linear behavior; if not, the behavior is nonlinear.

Take an entire spine from a *patient with ankylosing spondylitis* and fix it at the sacrum. Then apply a pull by way of a spring balance at C7 until the spring balance registers a force F of 20 N (4.47 lbf; Fig. 9-58A). Let the distance that the vertebra C7 moves

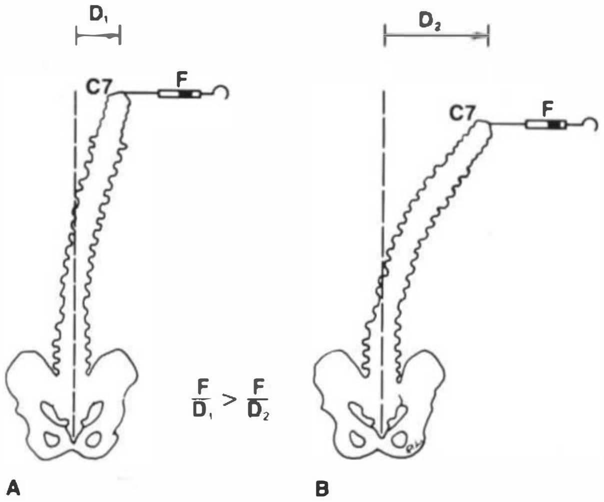

FIGURE 9-58 Stiffness coefficient. (A) Stiff spine, ankylosing spondylitis. (B) Supple spine in an agile adolescent.

be D_1. Now repeat the same experiment with a supple spine of an agile adolescent. Apply the same amount of force and measure the motion of C7 again. Its value is D_2 (Fig. 9-58B). The ratio of the force to the displacement is the coefficient of stiffness. It is found that F/D_1 is greater than F/D_2, and therefore the stiff spine has a higher stiffness coefficient. The inverse of stiffness is termed *compliance*. Thus, a highly stiff structure will have low compliance.

EXPLANATORY NOTES. Mathematically, the stiffness coefficient k is given by the following formula:

$$k = \frac{F}{D}$$

where F = load applied (force or moment) and D = displacement produced (translation or rotation).

Strain

DEFINITION. The change in unit length or angle in a material subjected to load.

There are two types of strain: normal, symbolized by the Greek letter ϵ (epsilon), and shear, symbolized by γ (gamma). The former is defined as the change in length divided by the original length. The normal strain can be tensile or compressive. The latter is defined as the change in angle. There are no units of measure.

DESCRIPTION AND EXAMPLES. Figure 9-59A shows a long specimen, say a bar made of rubber, being deformed. In Figure 9-59A, A1 has become elongated by applying tension. It may be compressed by applying a compressive force, as shown by A2. The ratio of the change in length to the original length is the *normal* strain.

Figure 9-59B shows a square specimen being deformed by a horizontal force. The shape of the specimen is changed from a square to a parallelogram. The upper right-hand corner of the specimen has been displaced. The *shear* strain is defined as this displacement divided by the height. If this ratio is small, then the shear strain may be conveniently represented by the angle of the inclined face of the specimen, as shown. However, the angle must be measured in radians.

A tennis ball hit normally to the wall is compressed as it comes in contact with the wall, producing normal strain. However, if the ball is hit nearly tangential to the wall, a large amount of shear strain is produced. The two cases are shown in Figure 9-59C.

Strains and Sprains

DEFINITION. Clinical terms that characterize injury to capsular, ligamentous, or musculotendinous structures.

DESCRIPTION AND EXAMPLES. In reality, they are often used in clinical situations when there is no possibility of determining exactly what has happened to the injured structure. The terms are not applied when a complete rupture or failure of the structure is clearly involved. They are not normally used when there is a clinically demonstrable plastic deformation or laxity of the structure. With strains and sprains, the structures have been loaded and deformed to a point at which pain is produced, and this is about all that can be determined. What has

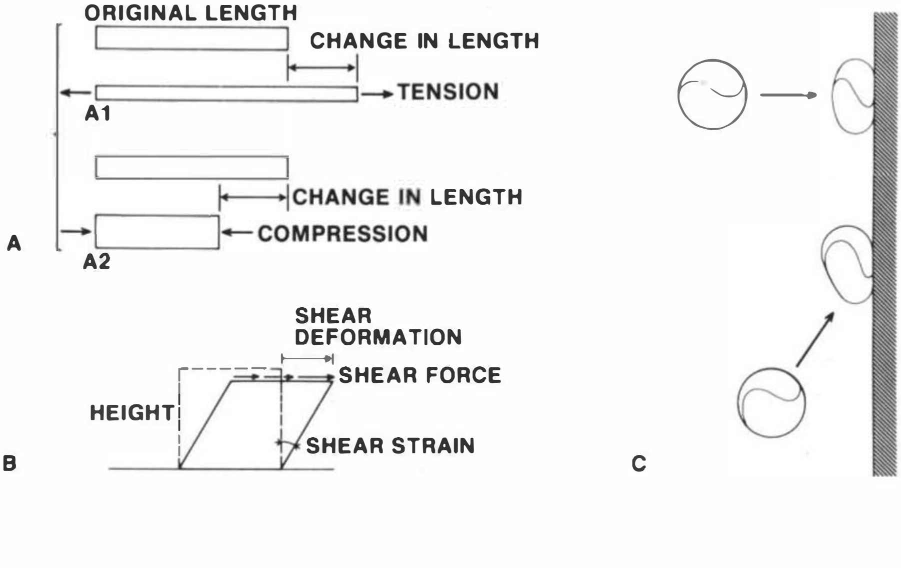

FIGURE 9-59 (A) Normal strain. (B) Shear strain. (C) *Normal* and *shear* strains in a tennis ball.

happened biomechanically when these terms are used is not known.

▶ Stress

DEFINITION. The force per unit area of a structure and a measurement of the intensity of the force.

There are two kinds of stress: normal, symbolized by the Greek letter σ (sigma), and shear, symbolized by τ (tau). The normal stress is perpendicular to the plane of a cross-section (Fig. 9-60A). The normal stress can be tensile or compressive. Shear stress is parallel to the cross-section (Fig. 9-60B). The unit of measure is newtons per square meter or pascals (poundforce per square inch).

DESCRIPTION AND EXAMPLES. When a structure is loaded with forces or moments, stresses are created throughout within the body. How much of the normal and shear stresses are present at a given point in the body depends on the orientation of the plane to which the stresses are referred. Changes in orientation of the cross-sections through a point in a structure alter the ratio between the normal and shear components, although the total stress remains the same. As an example, take the case of ski fractures. The tibia is subjected to torsion and fails with a spiral fracture (Fig. 9-60C). Torsion produces shear stress in a cross-section normal to the axis. At ±45° to the axis, the stress is no longer composed of shear but is pure tensile or compressive, depending upon the direction of torsion. At cross-sections between these two planes, there is a combination of shear and normal stresses.

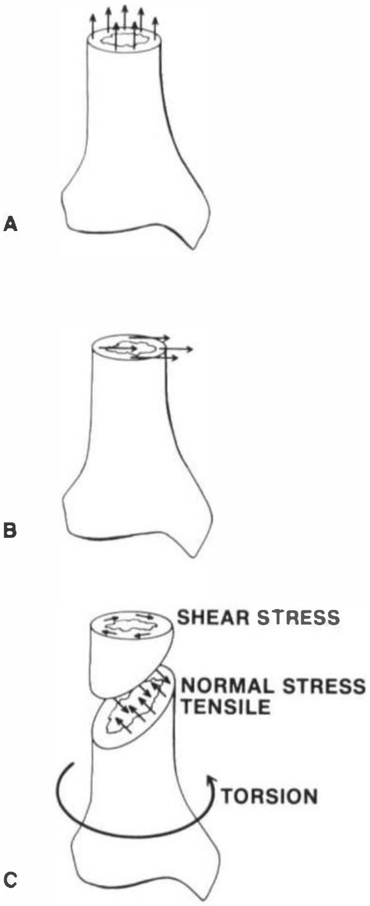

FIGURE 9-60 (A) Normal stress. (B) Shear stress. (C) Stresses during a ski fracture of the tibia.

EXPLANATORY NOTES. For torsional loading of a long structure, the three principal planes are oriented at +45°, −45°, and 0° to the long axis (the shear stresses are zero in those planes; see *Principal Planes*). Any other plane has a combination of normal as well as shear stresses. The transverse and axial planes are somewhat special. Here the normal stress is zero, and therefore only the shear stress is present. This can be verified by applying free-body analysis.

▶ Stress–Strain Diagram

DEFINITION. The plot of stress, usually on the ordinate or y-axis, versus strain, usually on the abscissa or x-axis. The relationship represents mechanical behavior of a *material*.

DESCRIPTION AND EXAMPLES. The stress–strain diagram of cancellous bone under compression, taken from the middle of the vertebral body along the direction of the longitudinal axis of the spine, serves as an example. First, a suitable specimen is prepared, and its length and cross-sectional area are measured (Fig. 9-61A). In a testing machine, an axial compressive load is applied. The load applied and the deformation produced are continuously measured and plotted on a graph paper. This is the *load–deformation* curve. Stress is obtained by dividing the load by the original cross-sectional area. Strain is obtained by dividing the deformation by the original length. Thus, the load–deformation curve is converted to the stress–strain diagram shown in Figure 9-61B. Segment OA is the linear elastic range within which stress and strain are proportional. Also, in this range, on removal of the load, the specimen returns to its original length and shape. The segment AB is the nonlinear elastic range within which stress and strain are no longer proportional to each other.

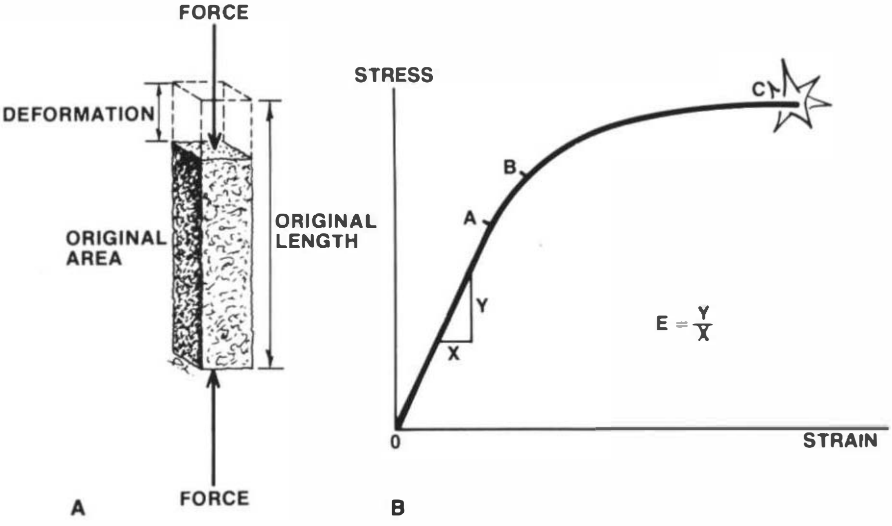

FIGURE 9-61 Stress–strain diagram. **(A)** A cancellous bone specimen being subjected to compression. **(B)** The results of this test are plotted as a stress—strain curve. OA = linear elastic range; AB = nonlinear elastic range; BC = plastic range; **E** = modulus of elasticity.

However, the specimen still returns to its original shape on removal of the load. Segment BC is the plastic range in which excessive deformation takes place with a very small increase in load, and the specimen no longer returns to its original shape on removal of the load; a residual permanent deformation is produced. At point C there is a sudden decrease in stress without additional strain, representing failure.

One of the important characteristics of a stress–strain diagram is the slope of its linear elastic range. It is called the modulus of elasticity (E) and is depicted in Figure 9-61. See *Modulus of Elasticity.*

▶ Stress and Strain Rates

DEFINITION. The rate of change of load per unit area with time is called the stress rate. Similarly, the rate of change of deformation per unit length with time is called the strain rate. The respective units of measure are newtons per square meter per second (poundforce per square inch per second) for the stress rate and per second (per second) for the strain rate. Sometimes the strain rate is given as meter/meter/second, which is dimensionally identical to per second.

DESCRIPTION AND EXAMPLES. All materials are sensitive to the rate of loading to a certain degree. This phenomenon is more predominant in viscoelastic materials such as plastics and biologic tissue than in metals. "Silly Putty" is a plastic of chewing gum consistency, and its inherent sensitivity to loading rates is meant to intrigue both the child and the adult. A slow and mild pull on the putty can produce a deformation of as much as several thousand percent before fracture. However, a quick, strong pull will break the putty with less than 10% deformation. So, if one were to describe the mechanical properties of "Silly Putty" or, for that matter, any viscoelastic material, it would be silly not to mention the rate of loading.

It has been well established in the biomechanics literature that bone, ligaments, tendons, and passive muscles are viscoelastic and are therefore sensitive to the rate of loading. Nevertheless, one still finds data published on the mechanical properties of vertebrae, discs, and various ligaments with no mention of the rate of loading. Such data are not of much use.

An example of the dependency of energy absorption capacity of a rabbit femur on the rate of deformation is shown in Figure 9-62. Note that the bone strength increases with the rate of deformation and seems to reach a maximum at about 1 rad/s.[14]

▶ Stress Concentration

DEFINITION. A sudden change in material and/or structure that creates a localized stress peak that cannot be predicted by simple strength of material theory.

DESCRIPTION AND EXAMPLES. Take two strips of metal, as shown in Figure 9-63. Strip II has a small hole but is wider than Strip I, so that they both have the same net cross-sectional area. Now apply tensile forces. The resulting stress in Strip I is uniform and equal to the force divided by the area. Strip II has a local high stress at the edge of the hole, nearly three

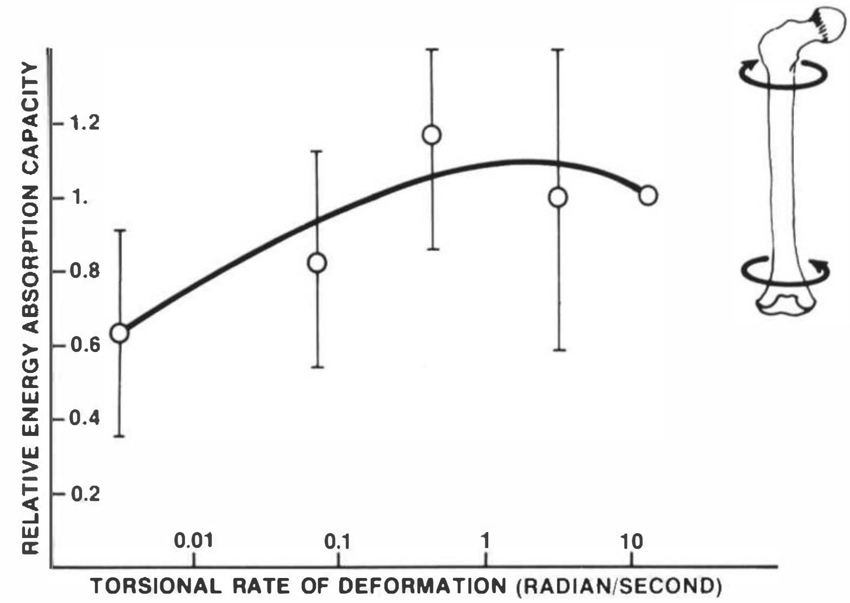

FIGURE 9-62 Stress and strain rates. Torsional rate of deformation alters the energy absorption capacity of a bone subjected to torsional failure.

times in magnitude as compared with Strip I.[20] This high stress cannot be predicted by the simple strength of material theory. However, it is calculable by the so-called Theory of Elasticity. If the strips are made of a ductile material, and the load is increased gradually, the strength of the two strips is found to be about the same. This is due to the fact that although the material at the hole edge does yield at one-third the load for Strip I, the effect is only local. However, if the load is cyclic or the material is brittle, there is a different situation. Under these circumstances, the failure load for Strip II will actually be one-third of that for Strip I.

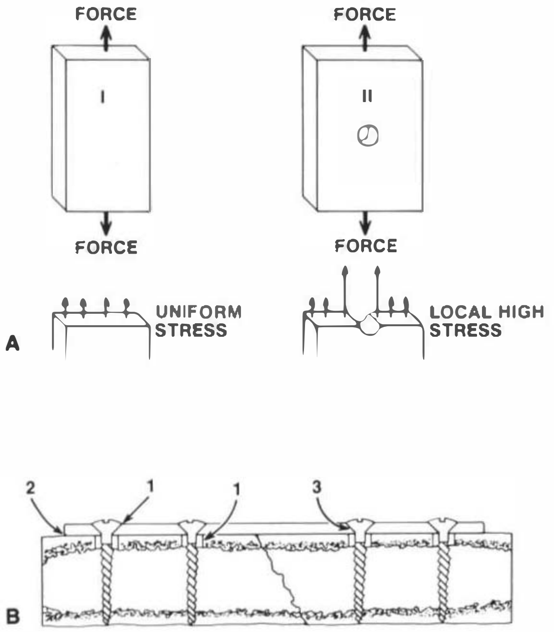

FIGURE 9-63 Stress concentration. **(A)** In a plate with a hole. **(B)** In a plate–bone–screw system due to (1) sudden changes in cross-sections, (2) mismatch of material properties, and (3) load application points.

There are many instances of stress concentration in orthopedic constructions. An example of a fixation plate is shown in Figure 9-63B. Three possible causes of stress concentration of the plate system are depicted: (1) sudden change in the cross-section of the plate, (2) junction of two or more dissimilar components of the system with mismatch of their mechanical properties, and (3) local stress at the points of application of loads.

▶ Subluxations and Dislocations

DEFINITION. A subluxation may be defined as a partial dislocation.

It is any pathologic situation in which there is not a normal physiologic juxtaposition of the articular surfaces of a joint. Such situations should be reliably demonstrable radiographically.

Dislocation is the term that is employed when there is no longer any degree of contact between the articulating surfaces.

DESCRIPTION AND EXAMPLES. A femoral head that is totally out of the socket and lying posterior to the acetabulum is completely dislocated.

These clinical terms are employed frequently but are not always clearly defined.

▶ Tangent

See *Trigonometric Functions*.

▶ Tangential

See *Joint Reaction Force*.

▶ Tensile Stress

See Stress.

▶ Tension

DEFINITION. A normal force that tends to elongate the fibers of a material. The unit of measure is newtons (poundforce).

DESCRIPTION AND EXAMPLES. When a rubber band is stretched, tension is applied. The rubber fibers are elongated. If there are any cuts or other weak spots on the surface, fracture cracks will initiate from these. If sufficient tension is applied, these cracks increase in size until the rubber band fails.

Tension is also manifested in the fibers on the convex side of a long structure when the structure is bent (see *Normal Stress*), as well as in fibers at 45° to the long axis when torsion is applied (see *Shear Stress*).

When the spine is flexed, ligaments posterior to the instantaneous axis of rotation are subjected to tension. When axial rotation occurs in the spine, the disc is subjected to torsion and some fibers of the annulus are subjected to tension.

▶ Testing Machines

DEFINITION. Standardized machines capable of applying standardized loads or displacements to a test specimen and measuring displacements or loads.

DESCRIPTION AND EXAMPLES. Materials testing machines, as the name implies, were originally designed to test material properties (e.g., of the stainless steel or high-density polymers). A test specimen of standardized shape and size was made of the material to be tested. These testing machines were generally capable of applying compression or tension to the test specimen.

For the tensile test, one end of the specimen is held in a stationary chuck (holding jig) attached to the base of the machine, while the other is held in a moving chuck attached to the moving cross-head. A load cell, provided between the chuck and cross-head, registers the force applied to the specimen. The force signal is recorded by a chart recorder moving with a speed proportional to the speed of the cross-head. Thus, the graph recorded on the chart recorder is a load–displacement curve of the specimen. The older machines had several restrictions (e.g., applied only translatory displacements to the specimen, often at slow rates).

Modern testing machines are much more versatile. They can either apply translatory displacement (and record force) or apply force (and measure translatory displacements). Some can even apply rotatory load (axial torque) or displacement (axial torsion) with the help of optional attachments. Another major feature of the modern testing machines is their speed. Because of the hydraulic power source, they are capable of applying load or displacement at very high speeds, simulating high-speed trauma. This particular aspect is very important for all biomechanical tests because the biological tissue, especially ligaments and discs, is highly viscoelastic and exhibits time-dependent behavior.

Among the U.S. manufacturers, the Instron in Massachusetts and the MTS in Minnesota are the most popular companies, with a wide variety of models available.

▶ Three-Dimensional Motion

DEFINITION. The most common kind of motion of a rigid body.

DESCRIPTION AND EXAMPLES. The body may move in any of six possible degrees of freedom. The motion is a combination of translation along any direction and rotation about any axis in space. Most of the human body joints have three-dimensional motion.

A body performing plane motion may always be brought from one position to another by pure rotation about an axis, the so-called instantaneous axis of rotation. Similarly, a body performing three-dimensional motion may always be moved from one position to another by defined amounts of rotation about and translation along an axis, called the helical axis of motion (HAM). Thus, a step of three-dimensional motion is fully defined by the position and direction of the instantaneous helical axis of motion and the magnitude of translation along and rotation about this axis (see *Helical Axis of Motion*).

Lateral bending produces translation and rotation of the vertebrae in the coronal plane as well as axial rotation because of inherent properties of the FSU. This is not plane motion because various points on the vertebrae do not travel in parallel planes. It is a three-dimensional motion. Vertebrae in a scoliotic spine have undergone three-dimensional displacement from a normal spine to a scoliotic curve.

▶ Three-Element Model

DEFINITION. A mathematical model consisting of a spring-element connected in parallel with a dashpot-element. The two are then further connected in series with a second spring-element. The three-element model is shown in Figure 9-64A. It is used to symbolize and mathematically simulate time-dependent mechanical behavior of certain viscoelastic materials.

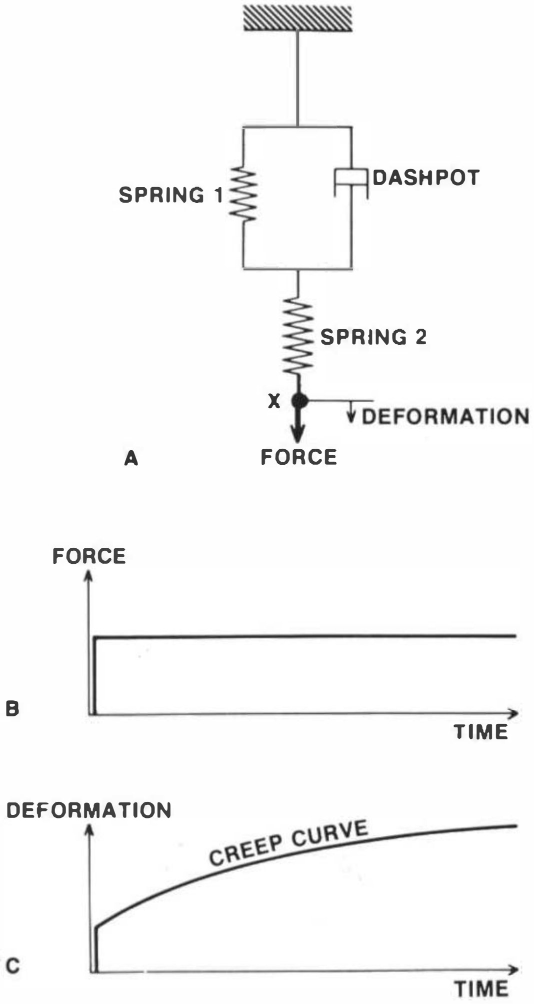

FIGURE 9-64 Three-element model. (**A**) Representation of a three-element model. (**B, C**) Creep behavior.

DESCRIPTION AND EXAMPLES. The creep phenomenon is often used to characterize viscoelastic behavior of biologic materials. The behavior of the three-element model under creep is studied here in detail to see if it can mimic the actual viscoelastic behavior. The creep phenomenon may be defined as a sudden application of a constant force, as shown in Figure 9-64B. After the force is applied, the deformation as a function of time is measured. Two things happen when creep is performed on the three-element model by applying sudden force. Referring to Figure 9-64A, with the sudden application of force the dashpot produces infinite resistance and locks in, but Spring 2 elongates, which produces immediate displacement of point X. Then, as time passes, with the force being held constant, the resistance of the dashpot decreases. This lets Spring 1 elongate at a rate defined by the dashpot and the stiffness properties of Spring 1. Thus, there is immediate elastic deformation with the application of a sudden force followed by an additional deformation as a function of time. The rate of deformation decreases with time, producing the characteristic creep curve (Fig. 9-64C).

Most biologic materials are viscoelastic in nature and their uniaxial behavior is adequately simulated by a single mathematical three-element model. By assigning different values to the stiffness coefficients of the two springs and the damping coefficient of the dashpot, time-dependent mechanical behavior of ligaments, tendons, skin, cancellous and cortical bone, and cartilage can be simulated.

During traction application to a scoliotic spine, immediate deformation of the spine may be represented by Spring 2, and the additional time-dependent deformation by a combination of Spring 1 and the dashpot. If suitable values are assigned to the three elements to represent the spine behavior, then it is theoretically possible to estimate the optimum time duration for traction.

▶ Three-Point Bending

DEFINITION. A structure is loaded in three-point bending with a single force applied on one side and two forces applied on the other side acting in the opposite direction.

DESCRIPTION AND EXAMPLES. A femur being subjected to three-point bending for determining its strength is shown in Figure 9-65A. The bending moment produced varies along the length of the structure, being zero under the end forces and maxi-

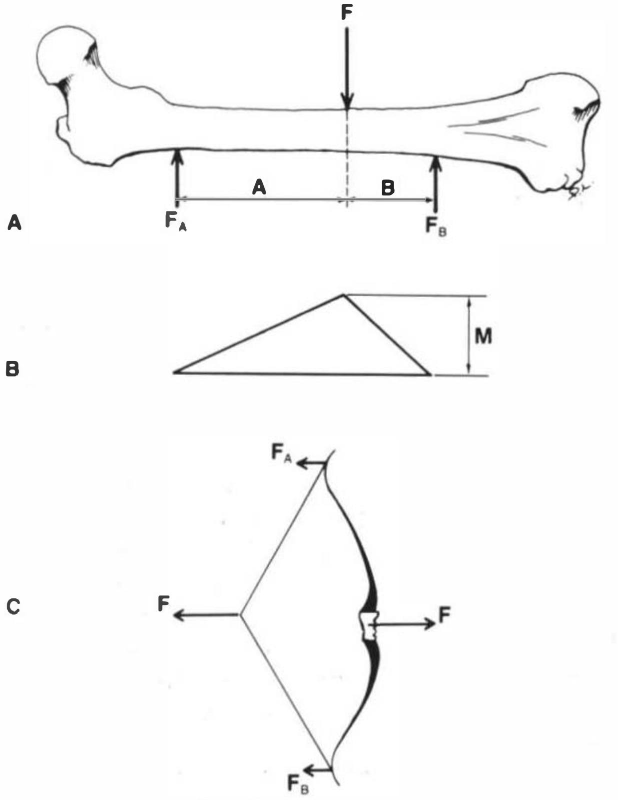

FIGURE 9-65 Three-point bending. **(A)** A femur being subjected to three-point bending. **(B)** Triangular bending moment diagram. **(C)** A bow loaded in three-point bending.

mum under the middle force. This is represented by the triangular bending moment diagram (Fig. 9-65B). The quantitative expression for maximum bending moment is given below.

When the archer draws the bow string, the bow is loaded in three-point bending (Fig. 9-65C). The bow has maximum cross-section at its middle because the bending moment is highest under the middle force.

The Milwaukee brace is an example in which three-point bending forces are employed in addition to axial tension to obtain angular correction of the spine.

EXPLANATORY NOTES. The maximum bending moment is given by the following equation:

$$M = \frac{F \times A \times B}{A + B}$$

where F is the middle force and A and B are the distances of the two end forces from it (Fig. 9-65B).

Torque

See *Couple* and *Torsion*.

Torsion

DEFINITION. A type of load that is applied by a couple of forces (parallel and directed opposite each other) about the long axis of a structure.

The load is called torque. It produces relative rotation of different axial sections of the structure with respect to each other. For a straight structure, all the sections are subjected to the same torque. However, in a curved structure, loaded by a torque on its ends, each cross-section is subjected not only to torque but also to bending. The magnitude of bending depends upon the orientation of the particular cross-section with respect to the torque axis.

DESCRIPTION AND EXAMPLES. In a straight bar (Fig. 9-66A), shearing stress is produced in cross-sections that are perpendicular and parallel to the torque axis. These are called circumferential and longitudinal shear stresses, respectively. On the other hand, normal stresses, tension, and compression are produced at ±45° with respect to the torque axis. These four stresses at a point are shown in Figure 9-66A. The results are based upon stress analysis of a cylindrical structure. All four stresses are equal in magni-

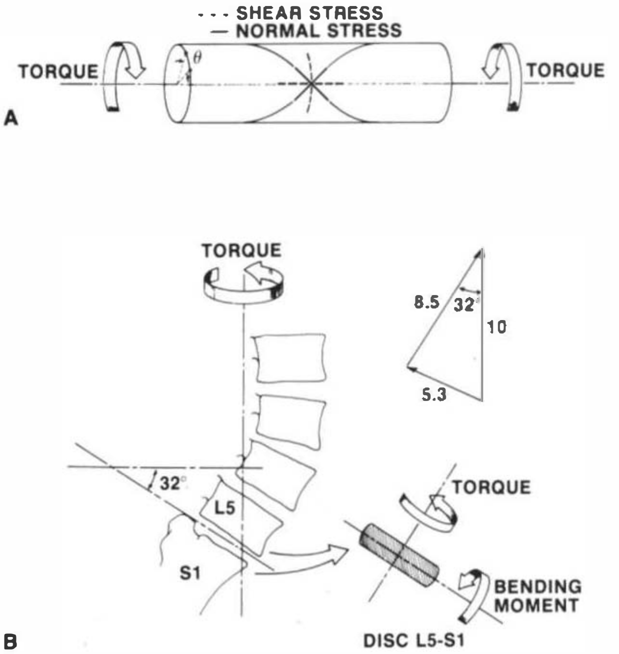

FIGURE 9-66 Torsion. **(A)** Application of torque produces shear and normal stresses. **(B)** Rotation of the trunk produces torsion and bending of the L5–S1 disc.

tude. The relationships between the stresses and the dimensions of the structure are given below.

A piece of ordinary chalk, when subjected to torque, breaks along a plane about 45° to the long axis where tensile stresses are maximum. From Figure 9-66A, it is known that all four stresses produced at a point are equal in magnitude. Therefore, tensile stress failure indicates that the chalk material is weakest in tension, as compared with shear and compression.

According to some researchers, disc failure in low back pain is due to combined torsion and bending loads.[6] Since the lumbar spine is a curved structure, it may be shown to be subjected to these combined torsion and bending loads by simple axial rotation of the trunk with respect to the pelvis. The L5–S1 disc typically has an angle of 32° with the vertical axis (Fig. 9-66B). Therefore, when an axial torque of 10 N-m (7.3 ft lbf) is applied about the vertical axis to the spine, the disc is subjected to a torque of 8.5 N-m (6.2 ft lbf) and a lateral bending moment of 5.3 N-m (3.9 ft lbf). These numbers were obtained by a free-body analysis (see *Free-Body Analysis*) of the disc and are shown in Figure 9-66B.

EXPLANATORY NOTES. When a straight structure is subjected to a torque, the shear and normal stresses (Fig. 9-66A) are manifested and the two ends of the structure rotate with respect to each other. These stresses in newtons per square meter or pascals (poundforce per square inch) and the angular deformation are given by the following formulas:

$$\text{shear stress} = \text{normal stress} = \frac{T \times R}{J}$$

$$\text{deformation angle in radians} = \frac{T \times L}{G \times J}$$

where T = torque in Nm (lbf ft); R = cylinder radius in m (ft); J = polar moment of inertia in m^4 (ft^4); L = cylinder length in m (ft); and G = shear modulus of the material in N/m^2 (lbf/ft^2).

▶ Torsional Rigidity

DEFINITION. The torque per unit of angular deformation. The unit of measure is newton meters per radian (foot poundforce per degree).

DESCRIPTION AND EXAMPLES. Torsional rigidity means rotatory stiffness. An example of this is the resistance felt when turning the steering wheel of an automobile (Fig. 9-67). The torsional rigidity of the steering wheel system can be measured by applying a defined torque and recording the angular displacement of the steering wheel before the tires turn on the pavement.

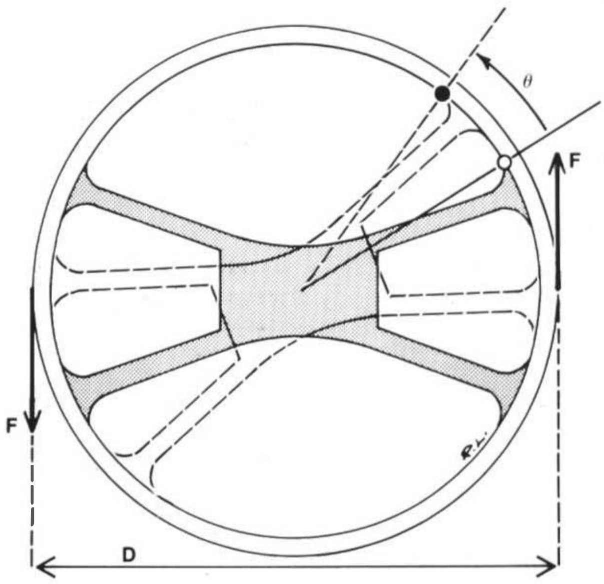

FIGURE 9-67 Torsional rigidity, as exemplified by the effort required to turn a steering wheel to a certain angle. Torsional rigidity = (D × F)/θ.

Torsional rigidity is an important quantity in characterizing body joints. For the analysis of the mechanism of ski fractures, it is essential to know the values of torsional rigidity of the joints involved. These values may be obtained by experiments. To calculate torsional rigidity of simpler structures, such as a cylindrical specimen of bone, the formula given below may be used.

Average values of torsional rigidity of the joints of the lower extremity have been measured:[17]

Hip	1.3 Nm/rad (230 in lbf/rad)
Knee	2.0 Nm/rad (350 in lbf/rad)
Ankle	2.1 Nm/rad (360 in lbf/rad)

EXPLANATORY NOTES. Mathematically, the following formulas apply:

$$\text{torsional rigidity} = \frac{T}{\theta}$$

where T = torque in Nm (lbf in) and θ = angular displacement in radians.

For a cylindrical structure, the torsional rigidity in terms of its basic structural properties can be described:

$$\text{torsional rigidity} = \frac{G \times J}{L}$$

where G = shear modulus of the material in N/m^2 (lbf/ft^2); J = polar moment of inertia in m^4 (ft^4); and L = cylinder length in m (ft).

▶ Translation

DEFINITION. Motion of a rigid body in which a straight line in the body always remains parallel to itself. The unit of measure is meters (feet).

DESCRIPTION AND EXAMPLES. If a boat is pushed smoothly in a straight line from position 1 to position 2 without pitching or rolling, it moves in pure translation (Fig. 9-68). A straight line joining two points X and Y will always remain parallel to itself in any two instantaneous positions of its motion. The line joining the two positions of the same point is the translation vector of the body.

During gait, the head moves forward, sideways, and up and down with respect to the ground. Neglecting any minor angular motions, the head may be said to go through translatory motions in three-dimensional space throughout its gait cycle.

Translation is a vector quantity. It has magnitude as well as direction. Motion of a point in space may be represented by a single translation vector. The motion of a rigid body may require one translation vector if the body is undergoing pure translation. For a rigid body moving in a plane, the translation vectors of two points must be known. Finally, for a rigid body performing three-dimensional motion, translation vectors of three points are required.

▶ Trigonometric Functions

DEFINITION. Sin (θ), the sine of angle θ, is the ratio of length BC to length AB.

Cos (θ), the cosine of angle θ, is the ratio of length AC to length AB.

Tan (θ), the tangent of angle θ, is the ratio of length BC to length AC.

DESCRIPTION AND EXAMPLES. There are three basic trigonometric functions: sine, cosine, and tangent. Their short forms are sin (θ), cos (θ), and tan (θ), where θ is a given angle in degrees or radians.

The functions are best described by referring to a right-angle triangle, one with one 90° angle (Fig. 9-69).

EXPLANATORY NOTES. From the above definitions, tan (θ) can be obtained by dividing sin (θ) by cos (θ).

Because the trigonometric functions are ratios of one length to the other, they have no units of measure.

▶ Ultimate Load

DEFINITION. The largest load a structure can sustain without failure. The unit of measure is newtons (poundforce) if the load is a force and newton meters (foot poundforce) if the load is a torque or moment.

DESCRIPTION AND EXAMPLES. For a simple structure subjected to a uniaxial load exemplified by a well-machined tensile test specimen of bone, there is a well-defined unambiguous point of failure. This is the maximum load point on the load–deformation curve (Fig. 9-70A). For complex structures subjected to a simple uniaxial load, the maximum load may not be called the ultimate load. This is well illustrated by the compressive load–deformation curve of an intact FSU.

The tracing of an actual experiment carried out in a testing machine is shown in Figure 9-70B. The compressive load increased and reached a maximum at point X and then decreased to Y and started increasing again. At point Z the load reached the limit of the transducer measuring load, but it was

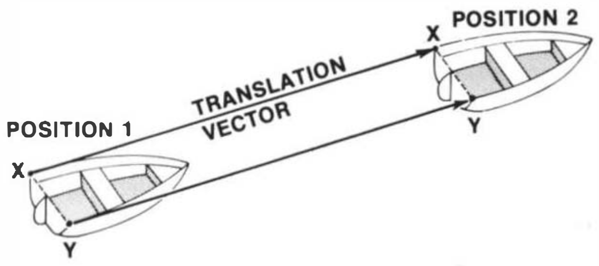

FIGURE 9-68 Translation. If line XY moves parallel to itself, then, and only then, the motion is pure translation.

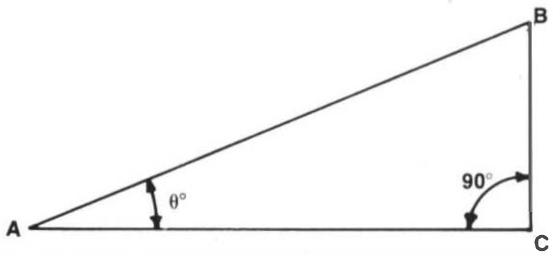

FIGURE 9-69 Trigonometric functions.
Sin (θ) = BC/AB
Cos (θ) = AC/AB
Tan (θ) = BC/AC

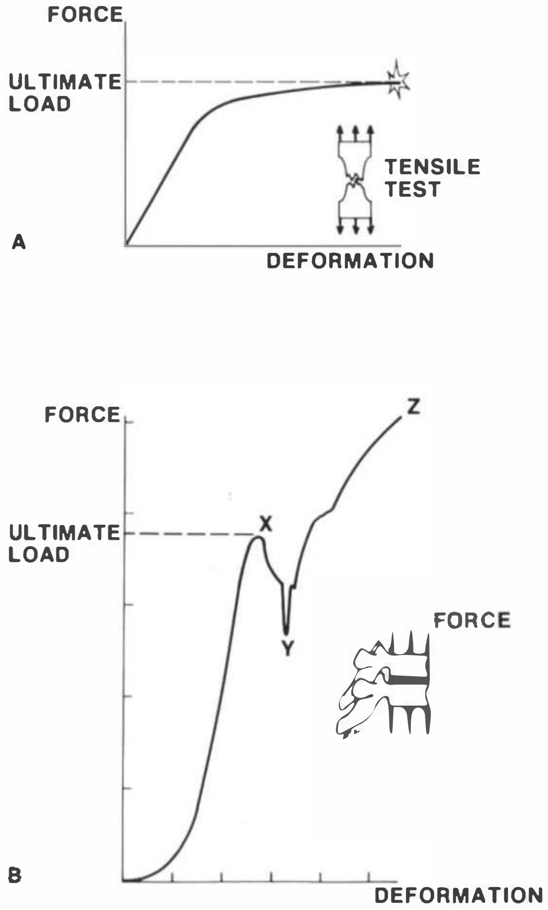

FIGURE 9-70 Ultimate load. **(A)** In tensile tests, the ultimate load is unambiguous. **(B)** The loading pattern observed in a real compression test situation is more difficult to analyze.

still increasing. What is the ultimate load? For the purpose of this experiment, point X was selected because certain structures (probably the end-plates) within the FSU failed and caused the decrease in the load-sustaining capacity. After point X, the FSU is not and never can be the same as it was in its initial pretest state. Therefore, any load that peaks after point X does not belong to the original FSU. It is important in reporting load-bearing capacity to indicate exactly where on the load–deformation curve the actual failure point was read.

Ultimate load divided by the original cross-sectional area is called ultimate stress. In the case of force, the ultimate stress is a normal stress if the area under consideration is normal to the force, and it is a shear stress if the area is parallel to the force. The units of measurement are newtons per square meter (poundforce per square inch).

▶ Units

See Conversion *Table*.

▶ Unit Vector

DEFINITION. A vector with unit magnitude. It is a mathematical quantity and is used to define a direction.

In three-dimensional space, a unit vector is made up of three numbers representing the inclination of its direction with respect to the three axes of a coordinate system.

DESCRIPTION AND EXAMPLES. Figure 9-71A shows a sailboat in rough sea. How can one make use of this concept in defining the orientation of the sailboat? As shown in Figure 9-71A, let vector N be parallel to

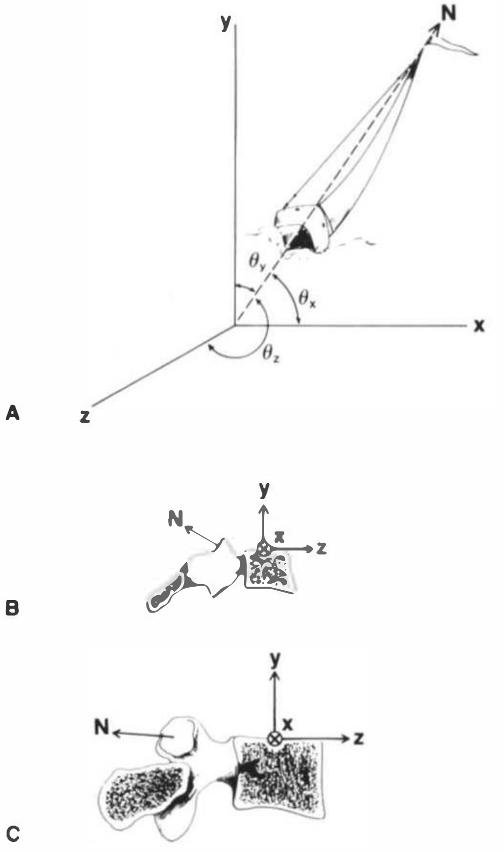

FIGURE 9-71 Unit vector. The concept of the unit vector helps define orientation of **(A)** a boat mast, **(B)** the facet plane of a cervical vertebra, and **(C)** the surface of a lumbar facet.

the mast. It makes three angles with the axes: θ_x, θ_y, θ_z. The vector N is made up of these three angles, or rather the cosines of these angles, as shown in Figure 9-71A (see *Trigonometric Functions*).

Facet orientation of the vertebrae varies with the level of the spine. In the cervical spine, the plane of the facets is approximately perpendicular to the sagittal plane and tilted about 45° to the vertical direction. If one knows these two angles, one can calculate, by the formula given below, the angles as shown in Figure 9-71B. In the lumbar region (around L3), the orientation of the facets is more complex. The facet joints are not simple planes, but they describe moderately curved surfaces. However, the unit vector concept can still be used to represent slopes of this complex surface at different points. For one such point in the middle of the facet, measured on a cadaver specimen, the angles of the unit vector components were found to have the following values: $\theta_x = 150°$, $\theta_y = 80°$, and $\theta_z = 118°$.

EXPLANATORY NOTES. Referring to Figure 9-71A, if θ_x, θ_y, θ_z are the angles made by the direction of the unit vector N with the respective axes x, y, z, then the three components of the unit vector N are as follows:

$$N = \begin{bmatrix} \cos\theta_x \\ \cos\theta_y \\ \cos\theta_z \end{bmatrix}$$

Furthermore, because the length of the unit vector is unity, the following applies:

$$(\cos\theta_x)^2 + (\cos\theta_y)^2 + (\cos\theta_z)^2 = 1$$

The two equations above define the unit vector completely once two of the three angles are known.

▶ Vector

DEFINITION. A quantity that possesses both a magnitude and a direction.

All vectors obey the parallelogram rule of addition, which states the following: If a parallelogram is constructed so that the two vectors to be added are adjacent sides, then the resultant is represented by a certain diagonal of the parallelogram (Fig. 9-72A).

DESCRIPTION AND EXAMPLES. All traction techniques in orthopedics are based upon the fundamental rule of the parallelogram. Traction forces may be represented by vectors. By single or multiple application of the rule, one can precisely determine the resulting force and its direction applied to the body part.

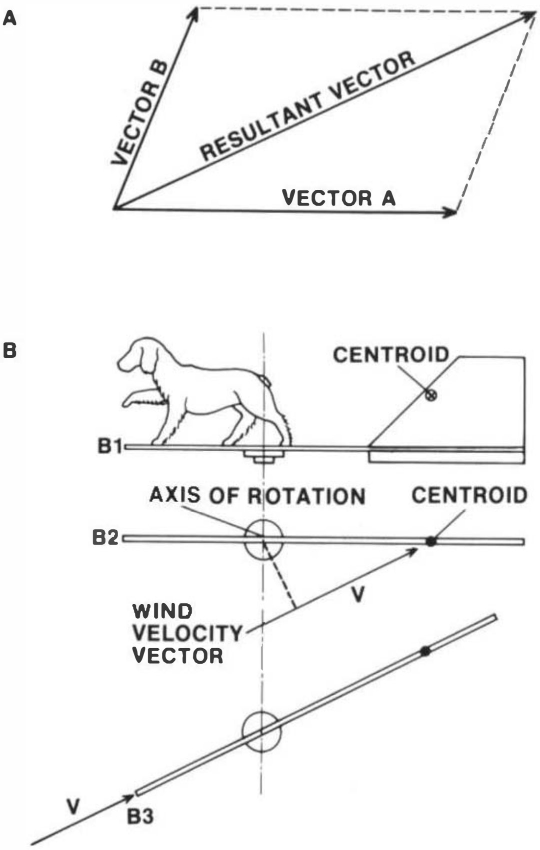

FIGURE 9-72 Vector. (A) Parallelogram rule for addition of two vectors. *(B1)* Side view of a weather vane. *(B2)* Top view. *(B3)* Vector diagram. The weather vane rotates until it aligns itself with the wind velocity vector.

Weather vanes are mounted atop old farmhouses to indicate wind direction (Fig. 9-72B1). Vector V represents the wind and its force acting at the centroid of the vane (Fig. 9-72B2). The centroid is eccentric with respect to the axis of rotation. Therefore, the wind velocity vector V produces a torque about the rotation axis. The weather vane will rotate, because of this torque, until it is in line with the wind velocity (Fig. 9-72B3). This reduces the torque to zero, producing a stable direction for the weather vane. When the wind changes direction, the stable direction of the weather vane will also change. Thus, the weather vane always indicates the direction of the wind velocity vector. How does the dog distinguish if the wind is coming from the front or the back? Consider it as an exercise.

A mathematical definition of a vector in three-dimensional space is extremely useful in analysis of complex loads or motions and is described below.

EXPLANATORY NOTES. In a three-dimensional space, with respect to a Cartesian coordinate system, a vector has three components, and they are depicted as follows:

$$V = \begin{bmatrix} V_x \\ V_y \\ V_z \end{bmatrix}$$

where V, the vector, is generally written in bold letters, and V_x, V_y, and V_z are its components or projections along the three coordinate axes. The magnitude of the vector (written with two vertical bars) is given by the following equation:

$$|V| = \sqrt{V_x^2 + V_y^2 + V_z^2}$$

Its direction is given by another set of equations:

$$\cos \theta_x = \frac{V_x}{|V|}$$

$$\cos \theta_y = \frac{V_y}{|V|}$$

$$\cos \theta_z = \frac{V_z}{|V|}$$

where θ_x, θ_y, and θ_z are the angles made by the vector with the respective coordinate system axes.

▶ Velocity

DEFINITION. The rate of change of position of a point with respect to a coordinate system. It is a vector quantity. Its magnitude is called speed. The velocity may be linear or angular, depending upon the type of motion. Correspondingly, the unit of measure is meters per second (feet per second) or radians per second.

DESCRIPTION AND EXAMPLES. A tennis ball traveling in midair, at any instant, has linear velocity. It may also have angular velocity if it spins.

The femur of a person running changes its linear and angular positions with time (Fig. 9-73). Therefore, the femur has linear as well as angular velocities. (See also *Instantaneous Velocity*.)

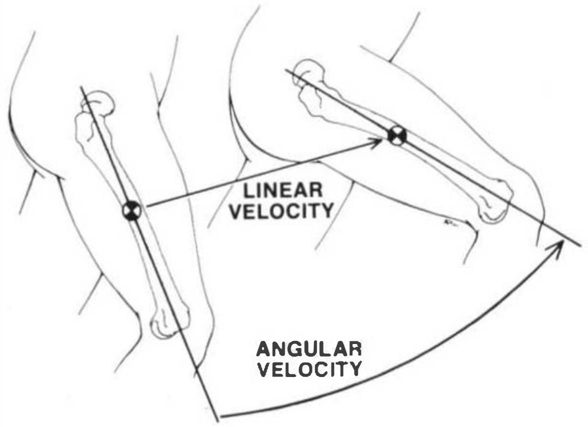

FIGURE 9-73 Velocity. During walking, the femur has linear as well as angular velocity.

▶ Vibrations

DEFINITION. Vibration is an oscillatory motion of a particle or structure with a certain periodicity.

DESCRIPTION AND EXAMPLES. Pluck a string on a guitar. It will vibrate. Observe that the motion of a point on the string is perpendicular to the length of the string. Let us plot this motion as a function of time (Fig. 9-74). The curve marked 1 starts at time zero, the instant when the string is at its undisplaced original position. The displacement reaches its maximum value and then decreases, passes the zero-value, reaches the opposite maximum, and then returns to the zero-value on the time axis. If there were no friction in the string, this cycle of vibration would repeat itself forever, but because of the energy loss in the string, the vibratory displacement will die down with time.

Vibrations of the string of a guitar provide a pleasant experience. But most often, vibrations are considered undesirable. Riding in a car without shock absorbers on a rough road is not a pleasant experience. In fact, long exposure to vibrations of a certain kind may be directly harmful. White fingers in the case of a pneumatic hammer operator is a well-known example in which the blood supply to the fingers is compromised because of vibration transmission into the hands. It has also been documented that long exposure to vibrations in a motor vehicle over time may significantly increase the risk of disc herniation.[8] Although the precise mechanism of this phenomenon is not known, it is believed that the

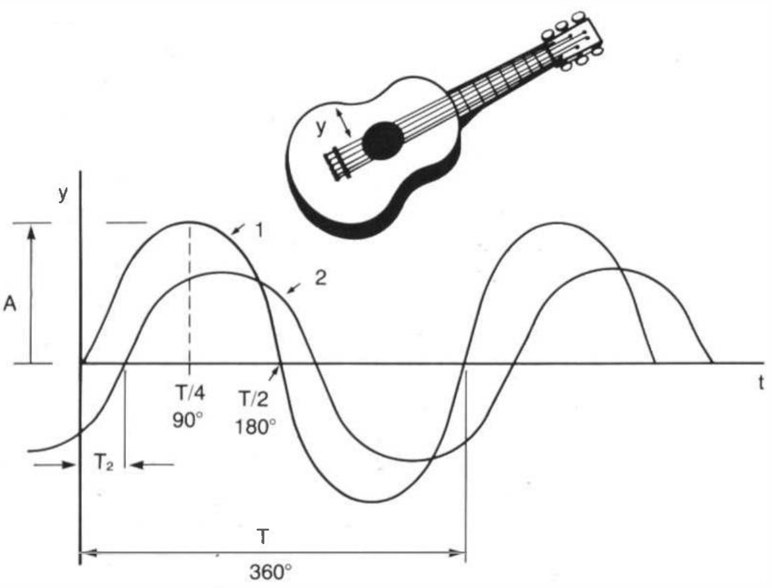

FIGURE 9-74 Vibrations of the string of a guitar may be represented by a sine curve. Such a curve represents displacement Y of the string, which increases with time for T/4 seconds, decreases for the next T/2 seconds, and then increases again for T/4 seconds, reaching its starting position. Here, T is the time required to complete one vibration cycle. A second string (curve 2) is plucked T_2 seconds after the first one (curve 1). This time lag is often represented in terms of degrees (considering T to be equal to 360°) and is called phase angle.

road vibrations are transmitted via the tires, car suspension, and seat suspension and reach the spinal column. A vibrating structure has highest amplitude at its resonance frequency. Therefore, certain transmitted vibrations with frequency near the resonance frequency of the spinal column may be responsible for the disc herniation, through yet unknown mechanisms. The spinal column resonance frequency has been found to be about 4.5 Hz.[10]

EXPLANATORY NOTES. The vibratory displacement, as shown by curve 1 in Figure 9-73, is a sinusoidal curve (*i.e.*, it is mathematically represented by a sine function). Two parameters, the amplitude A and the time period T, completely define a sine function. The amplitude A defines the height of the sine curve, while T is the time taken by the curve to complete one vibratory cycle. The reciprocal of this time period (*i.e.*, 1/T) is the frequency of the vibrations. Its unit of measurement is cycles per second or the more popular Hz (hertz).

Because the vibration is a repetitive process, it is often represented in terms of a circle. A given point on a circle is repeated after traveling 360° around the circle. Thus, one vibration cycle is equated to 360° (one-fourth and one-half vibration cycles are respectively represented by 90° and 180°).

There is another factor that is needed to more completely define the vibratory motion. This is the phase angle or the point in time at which the vibration starts. To further explain this, consider another vibrating string of the guitar that is plucked shortly after the first one. Its vibratory motion may be represented by curve 2. Notice that it starts a certain time T_2 after curve 1. This time lag is often represented by an angle in degrees, using the circle concept described above. Thus, curve 2 lags behind curve 1 by the phase angle $\theta 2$ in degrees. (The phase angle $\theta 2$ equals T_2 divided by the time period T and multiplied by 360°.)

▶ Viscoelasticity

DEFINITION. The time-dependent property of a material (e.g., hysteresis, creep, relaxation) to show sensitivity to rate of loading or deformation.

DESCRIPTION AND EXAMPLES. As the name suggests, two basic components of viscoelasticity are viscosity and elasticity. The behavior of a viscoelastic material is a combination of these two fundamental properties. Creep and relaxation are two phenomenological characteristics of viscoelastic materials and are used to document their behavior quantitatively. During creep tests, the load is suddenly applied and is kept constant thereafter; the resulting displacement is recorded against time. In relaxation tests, a deformation is produced and then fixed; the resulting decrease in load is recorded as a function of time.

There are two other practical phenomena that are typical of viscoelastic materials. A load–deformation curve of a viscoelastic material is dependent upon the rate of loading. The higher the rate of loading, the steeper the resulting curve. The other phenomenon involves the loading and unloading cycle. A viscoelastic material shows hysteresis (loss of energy in the form of heat during each cycle).

It has been experimentally determined that bone, ligaments, tendons, and passive muscles are viscoelastic, and their behavior can be reasonably simulated by the three-element model.

EXPLANATORY NOTES. Actual behavior of real-life materials such as bone, soft tissue, and plastics is very complex. However, their main characteristics can be simulated mathematically and represented by models that combine the basic elements of mathematical modeling (spring and dashpot) in a well-defined manner. Three of these basic combinations are the Maxwell, Kelvin, and three-element models. The models and their corresponding creep and relaxation curves are shown in Figure 9-75.

The Maxwell model is a series combination of a spring S and a dashpot D (Fig. 9-75A1). If a creep test is performed on it, then the motion of the point X, with point Y being fixed, as a function of time is given by the graph shown in Figure 9-75A2. There is an immediate displacement followed by a proportionately increasing displacement with time. Results of the relaxation test are shown in Figure 9-75A3. The force decreases exponentially (continuously, at an ever-decreasing rate, to zero).

The Kelvin model is a *parallel* combination of a spring S and a dashpot D (Fig. 9-75B1). The creep curve shows that the displacement of the point X is continuously increasing, but with an ever-decreasing rate (exponentially; Fig. 9-75B2). The relaxation is immediate but incomplete (Fig. 9-75B3). In other words, the force immediately decreases and then remains constant, never becoming zero.

The three-element model derives its name from the three mathematical components it is made of: two springs, S_1 and S_2, and a dashpot, D (Fig.

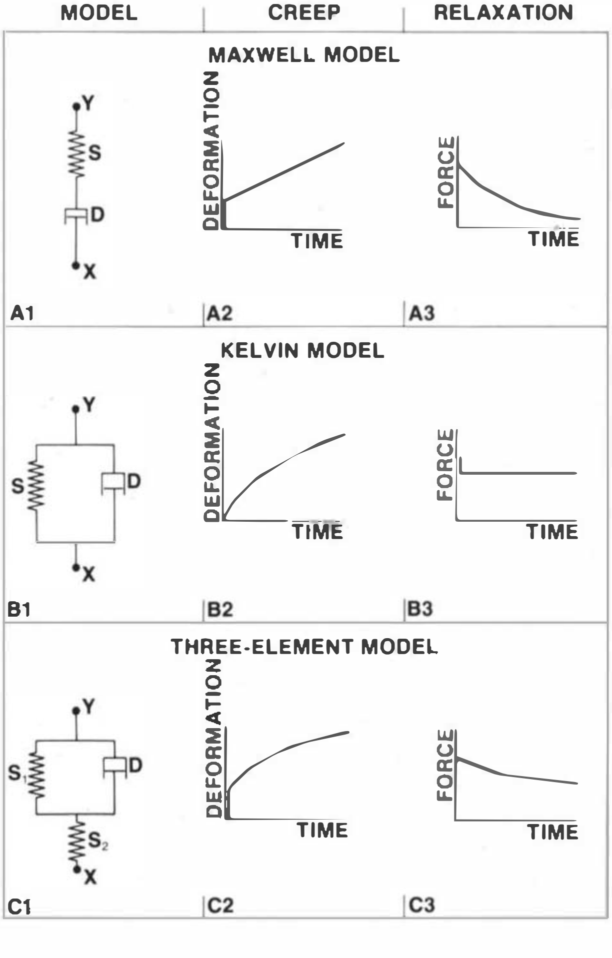

FIGURE 9-75 Viscoelasticity. Representation of viscoelasticity by model and by creep and relaxation behaviors are shown. **(A)** Maxwell model for fluids. **(B)** Kelvin model for solids. **(C)** Three-element model for most biologic tissues.

9-75C1). Results of the creep test are an immediate displacement followed by an exponential displacement with time (Fig. 9-75C2). The relaxation is exponential with time but is not complete (force never becomes zero [Fig. 9-75C3]; see also *Three-Element Model*.

▶ Viscoelastic Stability

DEFINITION. The type of stability in which the critical load is a function of time as well as the geometric and material properties of the structure.

DESCRIPTION AND EXAMPLES. Certain structures made of viscous and elastic elements when subjected to constant load may exhibit accelerating deformation behavior with time. Like a purely elastic structure, a viscoelastic structure does not have a critical load. It has a critical time period for a given load. Within this time period, the system is stable, and beyond it, it is unstable. This phenomenon is called the viscoelastic stability, and it is in contrast to the elastic stability, in which the critical factor is the load, with no dependency on time whatsoever.

There are plastics and organic materials that exhibit viscoelastic instability. Glue is one of these. When a heavy piece of material, such as a picture frame, is fixed to the wall with a piece of tape and falls off after a few hours or days, time-dependent stability has been exemplified.

Biologic structures are viscoelastic and therefore have time-dependent stability. Living bodies are much more complex. They are able to respond to unstable situations by altering the structure so as to re-create structural stability.

▶ Viscosity

DEFINITION. The property of materials to resist loads that produce shear. Viscosity is the ratio of shearing stress to shearing strain rate, or shearing stress to velocity gradient. It is commonly represented by η (eta) or μ (mu). The units of measure are newton seconds per square meter (poundforce per square foot) and poise (1 poise = 0.1 newton seconds per square meter).

DESCRIPTION AND EXAMPLES. In lubrication of joints, the viscosity of the fluid plays a very important role. If it is too high, it will resist motion. If it is too low, it will have less friction, but it can support only small loads before the thin lubricating film breaks down.

Viscosity of water does not vary with the rate of shear strain or the velocity gradient. The synovial fluid, on the other hand, has viscosity that varies inversely with the velocity gradient: 100 poise at velocity gradient of 0.1 per second and 1 poise at 100 per second. Figure 9-76A shows the variation of viscosity of water as well as the synovial fluid as functions of the velocity gradient. The two variables are plotted on logarithmic scales.

EXPLANATORY NOTES. Mathematically, the viscosity (η) is as follows:

$$\eta = \frac{\text{shear stress}}{\text{shear strain rate}}$$

$$\eta = \frac{\text{shear stress}}{\text{velocity gradient}}$$

The former definition is used with viscous solids, while the latter is used with fluids. Stress and strain rates are defined elsewhere (see *Stress and Strain Rates*). Velocity gradient is the variation of fluid velocity with fluid depth. Take two glass plates with a fluid between them (Fig. 9-76B). If one plate is moved with respect to the other, a velocity gradient

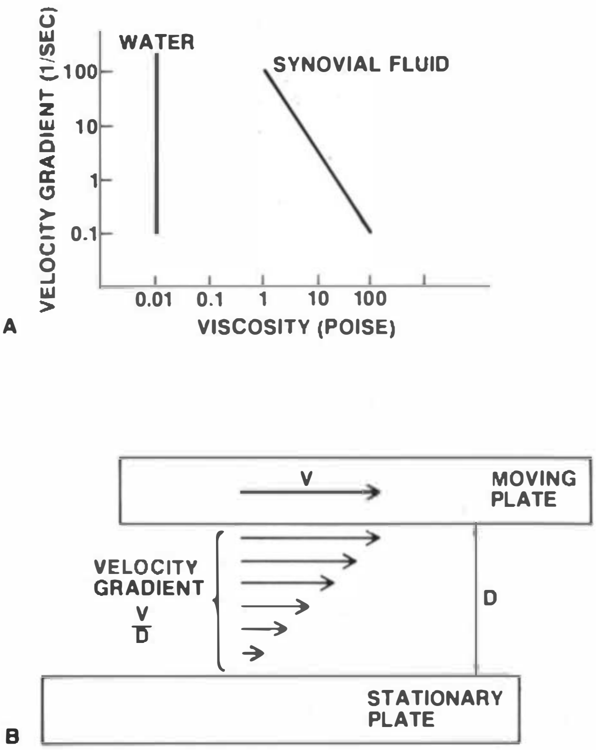

FIGURE 9-76 Viscosity. **(A)** The viscosity of water remains constant, while that of the synovial joint fluid decreases with an increase in joint velocity. **(B)** Interpretation of the velocity gradient.

is created. The fluid layer attached to the moving plate has velocity V, the same as the moving plate, while the fluid layer attached to the stationary plate is at rest. The layers between have intermediate velocities. If D is the distance of the plates, then V divided by D is the velocity gradient, assuming that the variation is linear.

▶ Work

DEFINITION. The amount of energy required to move a body from one position to another. Mechanical work is defined as the product of force applied to the distance moved in the direction of the force. The unit of measure for work is newton meters or joule (foot poundforce).

DESCRIPTION AND EXAMPLES. A woman weighing 60 kg (132 lb) climbs a flight of stairs that is 3 m (9.8 ft) high (Fig. 9-77A). How much work did she do? She worked 1766 Nm (1302 ft lbf) against earth's gravity. This is the amount of potential energy she possesses. To return back to the ground floor, the energy may be used positively to do some useful work, or it may be dissipated as heat.

During normal gait, the center of gravity of a person goes up and down approximately 6 cm (2.4 in; Fig. 9-77B).[16] The energy expended by a 60-kg (132-lb) person is about 32.3 Nm (26 ft lbf) per gait cycle. Fifty such gait cycles will be required to equal the energy expended in climbing the stairs. Actual energy loss will be higher as additional energy is needed to accelerate and decelerate other parts of the body because of inertia effects. A person with abnormal gait may have to move his center of gravity up and down a larger amount. He would then consume energy at a higher rate.

EXPLANATORY NOTES. The mathematical definition of work is as follows:

$$W = F \times D$$

where W = work in N-m (ft lbf) or joule; F = force in N (lbf); and D = distance moved in the direction of force in m (ft).

Here, the assumption of constant force is made. If the force varies, then $W = \int F\,dD$.

In the above example of a woman weighing 60 kg and climbing 3 m, the amount of work is as follows:

$$\begin{aligned} W &= (60 \times 9.81) \times 3 = 1766 \text{ Nm or Joule} \\ &= 1302 \text{ ft lbf} \end{aligned}$$

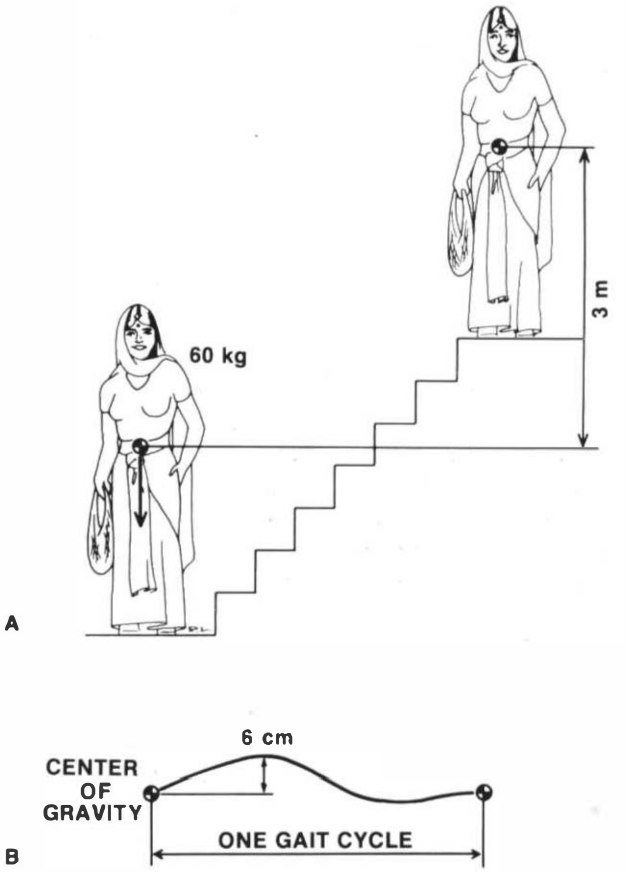

FIGURE 9-77 Work. **(A)** Climbing stairs requires work. **(B)** Walking results in up/down motion of the body weight, requiring work.

The amount of energy expended during a gait cycle is as follows:

$$\begin{aligned} W &= (60 \times 9.81) \times 0.6 = 35 \text{ Nm} \\ &= 26 \text{ lbf ft} \end{aligned}$$

Note that 9.81 is the value of the gravitational acceleration in meters per second per second.

▶ Yield Stress

DEFINITION. Magnitude of stress on the load–deformation curve at which appreciable deformation takes place without any appreciable increase in load. The unit of measure is newtons per square meter (poundforce per square inch).

In other words, yield stress is the stress of a material subjected to a load when plastic deformation has just started (Fig. 9-78).

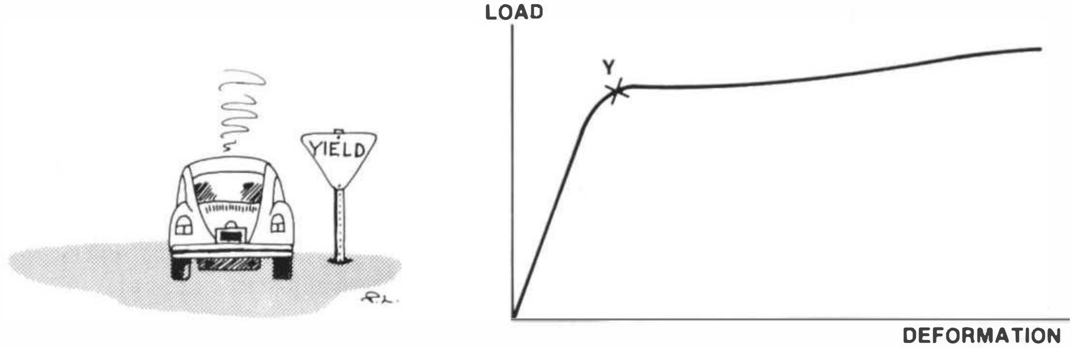

FIGURE 9-78 Yield stress.

Beyond the yield stress, the load–deformation curve is nearly a horizontal line. All deformation after the yield stress is permanent and is manifested at the time of removal of the load. Figure 9-78 shows a tracing of an actual load–deformation curve of a bone specimen.[3] Yielding probably started at point Y.

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