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现代骨科疾病诊断与治疗

CURRENT
Diagnosis & Treatment in ORTHOPEDICS

LANGE

second edition

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CURRENT
Diagnosis & Treatment in ORTHOPEDICS

Second Edition

Edited by

Harry B. Skinner, MD, PhD
Professor and Chairman
Department of Orthopedic Surgery
University of California, Irvine
College of Medicine
Professor of Mechanical and Aerospace Engineering
School of Engineering
University of California, Irvine

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Authors

Robert L. Barrack, MD
Professor of Orthopedic Surgery; Director, Adult Reconstructive Surgery, Tulane University School of Medicine, New Orleans
Internet: rbarrac@tmcpop.tmc.tulane.edu
Basic Science in Orthopedic Surgery

Michael S. Bednar, MD
Associate Clinical Professor, Department of Orthopedic Surgery and Rehabilitation, Loyola University Chicago, Stritch School of Medicine, Maywood, Illinois
Internet: mbednar@luc.edu
Hand Surgery

George D. Carlson, MD
Assistant Professor, Department of Orthopedic Surgery, University of California, Irvine Medical Center, Orange
Internet: gcarlson@uci.edu
Disorders, Diseases, & Injuries of the Spine

Stephen D. Cook, PhD
Lee C. Schlesinger Professor, Department of Orthopedic Surgery; Director of Orthopedic Research, Tulane University School of Medicine, New Orleans
Internet: scook2@mailhost.tcs.tulane.edu
Basic Science in Orthopedic Surgery

Edward Diao, MD
Assistant Professor, Department of Orthopedic Surgery; Chief, Division of Hand, Upper Extremity, and Microvascular Surgery, University of California, San Francisco
Internet: diaoe@orthosurg.ucsf.edu
Musculoskeletal Trauma Surgery

Richard J. Friedman, MD, FRCS(C)
Clinical Professor of Orthopedic Surgery, Medical University of South Carolina, Charleston; Director, Carolina Center for Orthopedic Surgery, Charleston
Internet: rjfriedman@home.com
Sports Medicine

Richard A. Gosselin, MD, FRCS(C), FAAOS
Private Practice, Merrittt Island, Florida
Internet: froggydoc@aol.com
Musculoskeletal Trauma Surgery

Serena S. Hu, MD
Associate Professor, Department of Orthopedic Surgery, University of California, San Francisco
Internet: hus@orthosurg.ucsf.edu
Disorders, Diseases, & Injuries of the Spine

James O. Johnston, MD
Professor of Orthopedic Oncology, University of California, San Francisco
Internet: johnston@orthosurg.ucsf.edu
Tumors in Orthopedics

Mary Ann E. Keenan, MD
Professor of Orthopedic Surgery and Professor of Physical Medicine and Rehabilitation, Temple University School of Medicine, Philadelphia; Director, Neuro-Orthopedic Program, Department of Orthopedic Surgery, Albert Einstein Medical Center, Philadelphia
Internet: drkeenan@aol.com
Rehabilitation

Terry R. Light, MD
Dr. William M. Scholl Professor and Chairman, Department of Orthopedic Surgery and Rehabilitation, Loyola University Chicago, Stritch School of Medicine, Maywood, Illinois
Internet: tlight@luc.edu
Hand Surgery

David W. Lowenberg, MD
Associate Professor of Clinical Orthopedic Surgery, University of California, San Francisco; Attending Surgeon and Chief of Fracture Service, California Pacific Medical Center, San Francisco
Musculoskeletal Trauma Surgery

Jeffrey A. Mann, MD
Private Practice, Oakland, California
Foot & Ankle Surgery

Roger A. Mann, MD
Associate Clinical Professor, Department of Orthopedic Surgery, University of California, San Francisco; Director, Foot Fellowship Program, Oakland, California
Foot & Ankle Surgery

Patrick J. McMahon, MD
Assistant Professor, Department of Orthopedic Surgery, Divisions of Sports Medicine, Shoulder and Elbow Surgery, University of Pittsburgh
Internet: pjmcmaho@yahoo.com
Sports Medicine

Keith D. Merrill, MD
Assistant Professor, Department of Orthopedics, Medical University of South Carolina, Charleston
Internet: merrillk@musc.edu
Sports Medicine

Robert S. Namba, MD
Associate Clinical Professor of Orthopedics, University of California, Irvine, College of Medicine; Attending Surgeon, Southern California Permanente Medical Group, Irvine
Internet: robert.s.namba@kp.org
Adult Reconstructive Surgery

William Petty, MD
Professor, retired, Department of Orthopedics, University of Florida College of Medicine, Gainesville; Chairman and Chief Executive Officer, Exactech, Inc, Gainesville, Florida
Internet: bill.petty@exac.com
Orthopedic Infections

George T. Rab, MD
Ben Ali Shriners Professor of Pediatric Orthopedics and Chief, Pediatric Orthopedics, University of California, Davis, School of Medicine; Consulting Physician, Shriners Hospitals for Children, Northern California
Internet: george.rab@ucdmc.ucdavis.edu
Pediatric Orthopedic Surgery

R. Lor Randall, MD
Assistant Professor, Department of Orthopedics, University of Utah Health Sciences Center; Director, Sarcoma Service, Huntsman Cancer Institute; Attending Physician, Primary Children's Medical Center; Consultant, Shriner's Hospital Intermountain, Salt Lake City
Internet: r.lor.randall@hsc.utah.edu
Tumors in Orthopedics

Harry B. Skinner, MD, PHD
Professor and Chairman, Department of Orthopedic Surgery, University of California, Irvine, College of Medicine; Professor of Mechanical and Aerospace Engineering, School of Engineering, University of California, Irvine
Internet: hskinner@msx.ndc.mc.uci.edu
Basic Science in Orthopedic Surgery; General Considerations in Orthopedic Surgery; Musculoskeletal Trauma Surgery; Sports Medicine; Adult Reconstructive Surgery

Douglas G. Smith, MD
Associate Professor, Department of Orthopedic Surgery, University of Washington School of Medicine, Seattle; Co-Principle Investigator, The Prosthetics Research Study, Seattle
Internet: dgsmith@u.washington.edu
Amputations

Clifford B. Tribus, MD
Assistant Professor, Division of Orthopedics, University of Wisconsin Medical School, Madison
Internet: tribus@surgery.wisc.edu
Disorders, Diseases, & Injuries of the Spine

Robert L. Waters, MD
Clinical Professor of Orthopedics, University of Southern California School of Medicine, Los Angeles; Medical Director, Ranchos Los Amigos National Rehabilitation Center, Downey, California
Internet: rwaters@dhs.co.la.ca.us
Rehabilitation

Preface

This *Current Diagnosis & Treatment in Orthopedics* is the second edition of the orthopedic surgery contribution to the Lange CURRENT series of books. It is intended to fulfill a need for a ready source of up-to-date information on disorders and diseases treated by orthopedic surgeons and related physicians. It follows the same format as other Lange CURRENTs with an emphasis on major diagnostic features of disease states, the natural history of the disease where appropriate, the work-up required for definitive diagnosis, and finally, definitive treatment. Because the book focuses on orthopedic conditions, treatment of the patient from a general medical viewpoint is de-emphasized except when it pertains to the orthopedic problem. Pathophysiology, epidemiology, and pathology are included when they assist in arriving at a definitive diagnosis or in understanding the treatment of the disease or condition.

References to the current literature were carefully chosen for the first edition and updated for the second edition so that the reader can investigate topics to greater depth than would be possible in a text of this size. Selected references to the older literature are also included when those articles are landmarks in the advancement of the understanding of orthopedic diseases and conditions.

INTENDED AUDIENCE

Students will find that the book encompasses virtually all aspects of orthopedics that they will encounter in classes and as sub-interns in major teaching institutions.

Residents or house officers can use the book as a ready reference, covering the majority of disorders and conditions in emergency and elective orthopedic surgery. Review of individual chapters will provide house officers rotating on subspecialty orthopedic services with an excellent basis for further, in-depth study.

For emergency room physicians, especially those with medical backgrounds, the text provides an excellent resource in managing orthopedic problems seen on an emergent basis.

Family practitioners and internists will find the book particularly helpful in the referral decision process and as a resource to explain disorders to patients.

Lastly, practicing orthopedic surgeons, particularly those in subspecialties, will find the book a helpful resource in reassuring them that their treatment in areas outside their subspecialty interests is current and up-to-date.

ORGANIZATION

The book is organized primarily by anatomic structure. Because of the natural subspecialization that has occurred in orthopedic surgery over the years, strict anatomic divisions are not always possible and in those cases subspecialties are emphasized. Thus, there is some overlap and some artificial division of subjects. The reader is encouraged to read entire chapters or, for more discrete topics, to go directly to the index for information. For example, the house officer rotating onto the foot and ankle service would find reading the foot and ankle chapter to be a prudent method of developing a baseline knowledge in foot surgery. A knee problem might be best approached by looking in the sports medicine chapter or in the adult reconstructive surgery chapter.

The first chapter serves as a basis for the rest of the book because it summarizes current basic information that is fundamental in understanding orthopedic surgery. Chapter 2 introduces aspects of interest in the perioperative care of the orthopedic patient. Management of orthopedic problems arising from trauma is covered in Chapter 3, while Chapter 4 deals with sports medicine with emphasis on the knee and the shoulder. Chapter 5 covers all aspects of spine surgery including degenerative spinal problems, spinal deformity, and spinal trauma.

Chapter 6 provides comprehensive coverage of tumors in orthopedic surgery, including benign and malignant soft tissue and hard tissue tumors. Adult joint reconstruction, including the disorders that lead to joint reconstruction, are covered in Chapter 7. In Chapter 8, infections with their special implications for orthopedic surgery are covered. Chapter 9 discusses foot and ankle surgery and Chapter 10, hand surgery. Chapter 11 covers diseases in orthopedics unique to children. The final two chapters deal with amputation and all aspects of rehabilitation fundamental to orthopedic surgeons in returning patients to full function.

OUTSTANDING FEATURES

- Careful selection of illustrations maximizes their benefits in pointing out orthopedic principles and concepts.
- The effect of changes in imaging technology on optimal diagnostic studies is emphasized.
- Bone and soft tissue tumor differential diagnosis are simplified by comprehensive tables that categorize tumors by age, location, and imaging characteristics.
- Concise, current, and comprehensive treatment of the basic science necessary for an understanding of the foundation of orthopedic surgery patient care is given.

NEW TO THIS EDITION

This edition has seen many changes that enhance its teaching potential. The key features have been chosen to help the student organize the important aspects of the diseases and allow easy recall. For example, the adult reconstruction chapter now includes graphs that help the student diagnose hip and knee problems based on the age of the patient at presentation. This chapter also includes summaries of the important aspects for prescribing the many available NSAIDs, including the new COX-2 inhibitors. Similarly the orthopedic oncology chapter has added additional tables to bring the myriad of orthopedic tumors into a cohesive and organized system for making the diagnosis of benign or malignant, primary or metastatic, tumor.

This edition also provides more guidance in the management of various conditions. For example, the trauma chapter has been expanded and includes information on the management of external fixators that are taking an expanding role in orthopedic trauma care. Similarly, the amputation chapter provides more guidance for the post-operative care of lower extremity amputations. The rehabilitation chapter provides information to allow prediction of function, such as ambulatory capability after spinal cord injury.

In order to provide the background to understand advances in orthopedic surgery, many additions have been made to the basic science chapter. It now includes new information on materials that have recently come onto the market for joint replacement, including the new polyethylenes. There is also new information on growth factors for the stimulation of bone formation.

Taken as a whole, all of the new features, including those mentioned above, combined with a review and update of the entire text and references make this edition a significant improvement over the last.

Harry B. Skinner, MD, PhD

Orange, California
April 2000

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Basic Science in Orthopedic Surgery

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Stephen D. Cook, PhD, Robert L. Barrack, MD, & Harry B. Skinner, MD, PhD

BIOMECHANICS & BIOMATERIALS

Stephen D. Cook, PhD,
& Robert L. Barrack, MD

Orthopedic surgery is the branch of medicine concerned with restoring and preserving the normal function of the musculoskeletal system. As such, it focuses on bones, joints, tendons, ligaments, muscles, and specialized tissues such as the intervertebral disk. Over the last half century, surgeons and investigators in the field of orthopedics have increasingly recognized the importance that engineering principles play both in understanding the normal behavior of musculoskeletal tissues and in designing implant systems to model the function of these tissues. The goals of the first portion of this chapter are to describe the biologic organization of the musculoskeletal tissues, examine the mechanical properties of the tissues in light of their biologic composition, and explore the material and design concepts required to fabricate implant systems with mechanical and biologic properties that will provide adequate function and longevity. The subject of the second portion of the chapter is gait analysis.

BASIC CONCEPTS & DEFINITIONS

Most biologic tissues are either **porous materials** or **composite materials.** A material such as bone has mechanical properties that are influenced markedly by the degree of porosity, defined as the degree of volume that is void in the material. For instance, the compressive strength of osteoporotic bone, which has increased porosity, is markedly decreased in comparison with the compressive strength of normal bone. Like composite materials, **alloyed materials** consist of two or more different materials that are intimately bound. While composite materials can be physically or mechanically separated, however, alloyed materials cannot.

Generally, composites are made up of a matrix material, which absorbs energy and protects fibers from brittle failure, and a fiber, which strengthens and stiffens the matrix. The performance of the two materials together is superior to that of either material alone in terms of mechanical properties (eg, strength and elastic modulus) and other properties (eg, corrosion resistance). The mechanical properties of various types of composite materials differ, based on the percentage of each substance in the material and on the principal orientation of the fiber. The substances in combination, however, are always stronger for their weight than is either substance alone. Microscopically, bone is a composite material consisting of hydroxyapatite crystals (the fibers) and an organic matrix that contains collagen.

The mechanical characteristics of a material are commonly described in terms of stress and strain. **Stress** is the force that a material is subjected to per unit of original area, and **strain** is the amount of deformation the material experiences per unit of original length in response to stress. These characteristics can be adequately estimated from a **stress-strain curve** (Figure 1–1), which plots the effect of a uniaxial stress on a simple test specimen made from a given material. Changes in the geometric dimensions of the material (eg, changes in the material's area or length) have no effect on the stress-strain curve for that material.

Mechanical characteristics can also be estimated from a **load-elongation curve,** in which the slope of the initial linear portion depicts the **stiffness** of a given material. Although similar in appearance to the stress-strain curve, the load-elongation curve for a given material can be altered by changes in the material's diameter (cross-sectional area) or length. For instance, doubling the diameter of a test specimen while maintaining the original length will double the stiffness because the increased diameter doubles the **load to failure** (that is, it doubles the amount of stress that a material can withstand in a single application) without changing the total elongation. Conversely, doubling the length of the test specimen while maintaining the original diameter will decrease the stiffness by half because doubling the length in turn doubles the elongation without changing the load to failure.

Because of this difference between the stress-strain curve and load-elongation curve, any comparison of

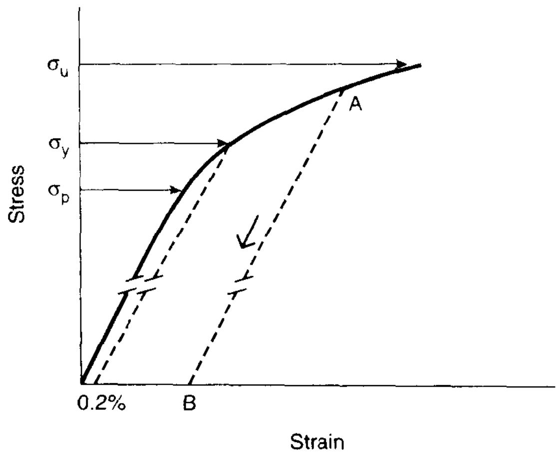

Figure 1–1. A generalized stress-strain diagram illustrating the mechanical properties of a material subjected to stress. The proportional limit (σ_p) of a material is the stress at which permanent or plastic deformation begins. Since the proportional limit is difficult to measure accurately for some materials, a 0.2% strain offset line parallel to the linear region of the curve is constructed. The stress corresponding to this line is defined as the yield stress (σ_y). If stress is removed after the initiation of plastic deformation (point A), only the elastic deformation denoted by the linear portion of the stress-strain curve is recovered. The ultimate tensile strength (σ_u) is the maximal stress that a material can withstand in a single application before it fails.

the characteristics of specimens requires that the same type of curve be used in the evaluation. If the load-elongation curve is used, the geometric dimensions of the specimens must also be the same. In this chapter, subsequent discussions will pertain to the stress-strain curve, although differing terminology in the load-elongation curve will be noted parenthetically.

The initial linear or elastic portion of the stress-strain curve (Figure 1–1) depicts the amount of stress a material can withstand before permanently deforming. The slope of this line is termed the **modulus of elasticity** (stiffness) of the material. A high modulus of elasticity indicates that the material is difficult to deform, whereas a low modulus indicates that the material is more pliable. The modulus of elasticity is an excellent basis on which different materials can be compared. When materials such as those used in implants are compared, however, it is important to remember that the modulus of elasticity is a property only of the material itself and not of the structure. Implant stiffness—or, more correctly, flexural rigidity—is a function both of material elastic modulus and of design geometry.

The **proportional limit,** or σ_p, of a material is the stress at which permanent or plastic deformation begins. The proportional limit, however, is difficult to measure accurately for some materials. Therefore, a 0.2% strain offset line parallel to the linear region of the curve is constructed, as shown in Figure 1–1. The stress corresponding to this line is defined as the **yield stress,** or σ_y. If stress is removed after the initiation of plastic deformation (point A in Figure 1–1), only the elastic deformation denoted by the linear portion of the stress-strain curve is recovered. The **ultimate tensile strength** (failure load), or σ_u, is the maximal stress that a material can withstand in a single application before it fails.

When subjected to repeated loading in a physiologic environment, a material may fail at stresses well below the ultimate tensile strength. The **fatigue curve,** or **S-N curve,** demonstrates the behavior of a metal during cyclic loading and is shown in Figure 1–2. Generally, as the number of cycles (**N**) increases, the amount of applied stress (**S**) that the metal can withstand before failure decreases. The **endurance limit** of a material is the maximal stress below which fatigue failure will never occur regardless of the number of cycles. Fatigue failure will occur if the combination of local peak stresses and number of loading cycles at that stress are excessive. Environmental conditions strongly influence fatigue behavior. The physiologic environment, which is corrosive, can significantly reduce the number of cycles to failure and the endurance limit of a material.

Materials can be evaluated in terms of ductility, toughness, viscoelasticity, friction, lubrication, and wear. These properties will be introduced here, and many of them will be explored in detail in subsequent sections.

Ductility is defined as the amount of deformation that a material undergoes before failure and is charac-

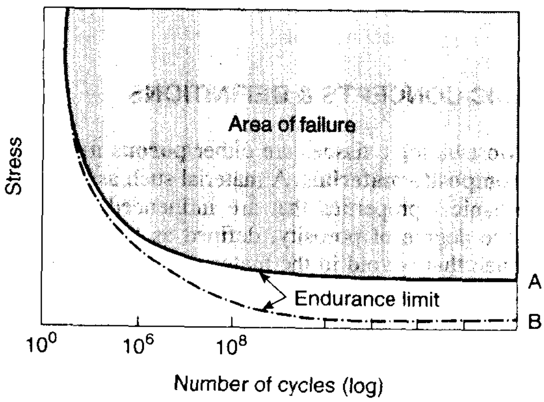

Figure 1–2. A generalized diagram comparing two fatigue curves, or S-N curves, for the same material. Curve A illustrates the material's endurance limit in a noncorrosive environment, while curve B illustrates its endurance limit in a corrosive environment. The body is an example of a corrosive environment for implant materials.

terized in terms of total strain. A brittle material will fail with minimal strain caused by propagation because the yield stress is higher than the tensile stress. A ductile material, however, will fail only after markedly increased strain and decreased cross-sectional area. Polymethylmethacrylate (a polymer) and ceramics are brittle materials, while metals exhibit relatively more ductility. Environmental conditions, especially changes in temperature, can alter the ductility of materials.

Toughness is defined as the energy supplied to a material to cause it to fracture and is measured by the total area under the stress-strain curve.

Since all biologic tissues are viscoelastic in nature, a thorough understanding of **viscoelasticity** is essential. A viscoelastic material is one that exhibits different properties when loaded at different strain rates. Thus, its mechanical properties are time-dependent. Bone, for example, absorbs more energy at fast loading rates, such as in high-speed motor vehicle accidents, than at slow loading rates, such as in recreational snow skiing.

There are three important properties of viscoelastic materials: hysteresis, creep, and stress relaxation. When a viscoelastic material is subjected to cyclic loading, the stress-strain relationship during the loading process differs from that during the unloading process (Figure 1–3). This difference in stress-strain response is termed **hysteresis.** The deviation between loading and unloading processes is dependent on the degree of viscous behavior. The area between the two curves is a measure of the energy lost by internal friction during the loading process. **Creep,** which has also been called **cold flow** and is observed in polyethylene components, is defined as a deformation that

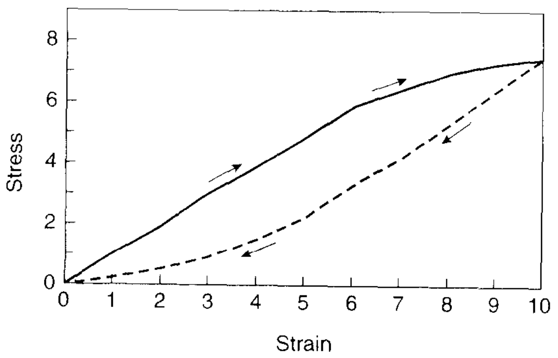

Figure 1–3. When a viscoelastic material is subjected to cyclic loading, the stress-strain curve during the loading process (solid lines) differs from that during the unloading process (dotted lines). This difference in stress-strain response is called **hysteresis.** The area between the two curves is a measure of the energy lost by internal friction during the loading process.

occurs in a material under constant stress. Some deformation is permanent, persisting even when the stress is released. The constant strain associated with a decrease in stress over time can result in **stress relaxation,** a phenomenon evident, for example, in the loosening of fracture fixation plates. The time necessary to attain creep or stress relaxation equilibrium is an inherent property of the material.

Friction refers to the resistance between two bodies when one slides over the other. Friction is greatest at slow rates and decreases with faster rates. This is because the surface asperities (peaks) tend to adhere to each other more strongly at slower rates. Mechanisms of **lubrication** reduce the friction between two surfaces. Several lubrication mechanisms are present in articular cartilage to overcome friction processes in normal joint motion. Similarly, mechanisms are present in polyethylene-metal articulations to overcome friction in joint replacements.

Wear occurs whenever friction is present and is defined as the removal of surface material by mechanical motion. Wear is always observed between two moving surfaces, but lubrication mechanisms act to reduce the detrimental effects of excessive wear. Three types of wear mechanisms are apparent in normal and prosthetic joint motion: abrasive, adhesive, and three-body wear. **Abrasive wear** is the generation of material particles from a softer surface when it moves against a rougher, harder surface. An example of the product of abrasive wear is sawdust, which results from the movement of sandpaper against a wood surface. The amount of wear depends on factors such as contact stress, hardness, and finish of the bearing surfaces.

Adhesive wear results when a thin film of material is transferred from one bearing surface to the other. In prosthetic joints, the transfer film can be either polyethylene or the passivated layer of metal. Regardless of the material, wear occurs in the surface that loses the transfer film. If the particles from the transfer film are shed from the other surface as well, they behave as a third body and also result in wear.

Three-body wear occurs when another particle is located between two bearing surfaces. Cement particles act as third bodies in prosthetic joints. Implant designers continue to search for compatible substances that reduce friction at articulating surfaces and thereby reduce the amount of wear debris generated. Wear of polyethylene is the dominant problem in total joint replacement today because the wear debris generated is biologically active and leads to osteolysis.

BIOMECHANICS IN ORTHOPEDICS

An analysis of the factors that influence normal and prosthetic joint function requires an understanding of free body diagrams as well as the concepts of force, moment, and equilibrium.

Force, Moment, & Equilibrium

Forces and moments are vector quantities—that is, they are described by point of application, magnitude, and direction. A force represents the action of one body on another. The action may be applied directly (such as via a push or a pull) or from a distance (such as via gravity). A normal tensile or compressive force is applied perpendicular to a surface, whereas a shear force is applied parallel to a surface. A force that is applied eccentrically produces a moment.

The force generated by gravity on an object is the center of gravity. An object that is symmetric has its center of gravity in the geometrically centered position, whereas an object that is asymmetric has its center of gravity closer to its "heavier" end. The center of gravity for the human body is the resultant of the individual centers of gravity from each segment of the body. Therefore, as the body segments move, the center of gravity changes accordingly and may even lie outside the body in extreme positions, such as encountered in gymnastics. A moment is defined as the product of the quantity of force and the perpendicular distance between the line of action of the force and the center of rotation. A moment usually results in a rotation of the object about a fixed axis.

Newton's first law states that a body (or object) is in equilibrium if the sum of the forces and moments acting on the body are balanced; therefore, the sum of forces and moments for each direction must equal zero. The concept of equilibrium is important in understanding and determining force-body interactions, such as the increased joint reaction force occurring in an extended arm because of an external weight and such as the increased joint reaction force occurring in the hip at a specific moment during walking.

Free Body Diagrams

A free body diagram can be used to schematically represent all the forces and moments acting on a joint. The concepts of equilibrium can be extended to determine joint reaction or muscle forces for different conditions, as demonstrated in the following two examples.

Example No. 1: Determine the force on the abductor muscle of a person's hip joint (the abductor force, or F_{AB}) and the joint reaction force (the F_J) when the person is standing on one leg. The weight of the trunk, both arms, and one leg is 5/6 of the total weight (w) of the person. As illustrated in Figure 1–4, this weight will tend to rotate the body about the femoral head and is counteracted by the pull of the abductor muscles on the pelvis. The necessary equation to solve for the abductor force, F_{AB}, is as follows:

$$F_{AB} \cdot a = \tfrac{5}{6} w \cdot b$$

In solving the equation, assume that a = 5 cm and that b = 15 cm.

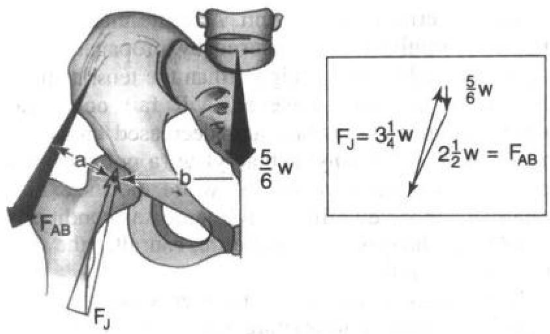

Figure 1–4. A free body diagram and force triangle illustrating the method for determining the force of the abductor muscle of a person's hip joint (F_{AB}) and the joint reaction force (F_J) when the person is standing on one leg and the total weight (w) of the person is known. See the discussion of example no. 1 in the text.

After this equation is solved, two of the three forces are known. The remaining force (the F_J) can be determined from a force triangle (Figure 1–4), because according to Newton's first law, the sum of forces must equal zero.

Example No. 2: Determine the force on a person's deltoid muscle (the deltoid force, or F_D) and the force of the joint acting about the shoulder (the joint force, or F_J) when the person holds a metal weight (w) at arm's length (Figure 1–5). The weight of the arm is ignored because only the increase in forces about the shoulder caused by the metal weight is to be determined. F_D is determined by summing the moments about the joint center. The necessary equation is as follows:

$$F_D \cdot a = W \cdot b$$

In solving the equation, assume that a = 5 cm and that b = 60 cm.

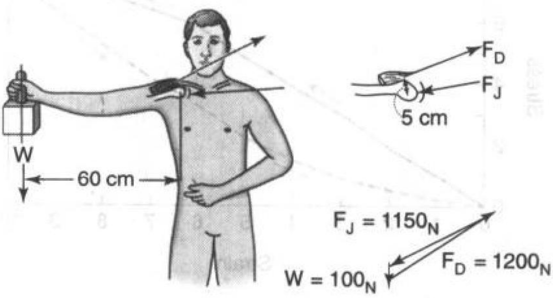

Figure 1–5. A free body diagram and force triangle illustrating the method for determining the force of a person's deltoid muscle (F_D) and the force of the joint acting about the shoulder (F_J) when the person holds a metal weight (w) at arm's length. See the discussion of example no. 2 in the text.

After this equation is solved, a joint reaction force of 1150 newtons is determined using a force triangle (Figure 1–5).

Moments of Inertia

The orientation of the bone or implant cross-sectional area with respect to the applied principal load also greatly influences the biomechanical performance. Bending and torsion occur in long bones and are important considerations in the design of implants. In general, the farther that material mass is distributed from the axis of bending or torsion while still retaining structural integrity, the more resistant the structure will be to bending or torsion. The **area moment of inertia** is a mathematical expression for resistance to bending, while the **polar moment of inertia** is a mathematical expression for resistance to torsion. Both types of moment of inertia relate the cross-sectional geometry and orientation of the object with respect to the applied axial load. The larger the area moment of inertia or the polar moment of inertia is, the less likely the material will fail. Figure 1–6 summarizes the area moments of inertia for representative shapes important to orthopedic surgery. Creating an open slot in an object will significantly decrease the polar moment of inertia of the object.

Knowledge of moments of inertia is important for understanding mechanical behavior in relation to object geometry. For instance, the length of the long bones predisposes them to high bending moments. Their tubular shape helps them resist bending in all directions, however. This resistance to bending is attributable to the large area moment of inertia because the majority of bone tissue is distributed away from the neutral axis. The concept of moment of inertia is crucial in the design of implants that are exposed to excessive bending and torsional stresses.

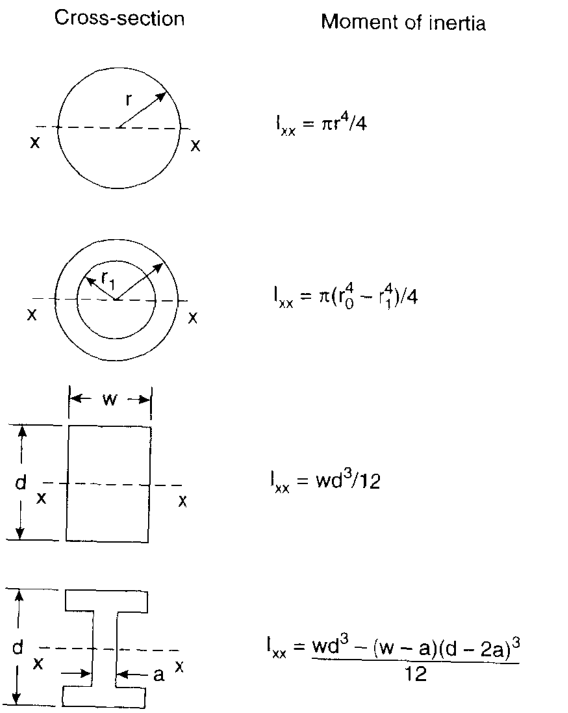

Figure 1–6. Summary of the area moments of inertia for representative shapes important to orthopedic surgery.

BIOLOGIC TISSUES IN ORTHOPEDICS

The functions of the musculoskeletal system are to provide support for the body, to protect the vital organs, and to facilitate easy movement of joints. The bone, articular cartilage, tendon, ligament, and muscle all interact to fulfill these functions. The musculoskeletal tissues are integrally specialized to perform their duties and have excellent regenerative and reparative processes. They also adapt and undergo compositional changes in response to increased or decreased stress states. Specialized components of the musculoskeletal system, such as the intervertebral disk, are particularly suited for supporting large stress loads while resisting movement.

Bones

Bones are dynamic tissues that serve a variety of functions and have the ability to remodel to changes in internal and external stimuli. Bones provide support for the trunk and extremities, provide attachment to ligaments and tendons, protect vital organs, and act as a mineral and iron reservoir for the maintenance of homeostasis.

A. Structural Composition: Bone is a composite consisting of two types of material. The first material is an organic extracellular matrix that contains collagen, accounts for about 30–35% of the dry weight of bone, and is responsible for providing flexibility and resilience to the bone. The second material consists primarily of calcium and phosphorous salts, especially hydroxyapatite $[Ca_{10}(PO_4)_6(OH)_2]$, accounts for about 65–70% of the dry weight of bone, and contributes to the hardness and rigidity of the bone. Microscopically, bone can be classified as either woven or lamellar.

Woven bone, which is also called **primary bone,** is characterized by a random arrangement of cells and collagen. Because of its relatively disoriented composition, woven bone demonstrates isotropic mechanical characteristics, with similar properties observed regardless of the direction of applied stress. Woven bone is associated with periods of rapid formation, such as the initial stages of fracture repair or biologic implant fixation. Woven bone, which has a low mineral content, remodels to lamellar bone.

Lamellar bone is a slower-forming, mature bone that is characterized by an orderly cellular distribution and regular orientation of collagen fibers (Figure 1–7). The lamellae can be parallel to each other or concentrically organized around a vascular canal called a **haversian system** or **osteon.** At the periphery of each osteon is a cement line, a narrow area containing ground substance primarily composed of glycosaminoglycans. Neither the canaliculi nor the collagen fibers cross the cement line. Biomechanically, the cement line is the weakest link in the microstructure of bone. The organized structure of lamellar bone makes it anisotropic, as seen in the fact that it is stronger during axial loading than it is during transverse or shear loading.

Bone can be classified macroscopically as cortical tissue and cancellous (trabecular) tissue. Both types are morphologically lamellar bone. Cortical tissue relies on osteons for cell communication. Because trabecular width is small, however, the canaliculi can communicate directly with blood vessels in the medullary canal. The basic differences between cortical tissue and cancellous tissue relate to porosity and apparent density. The porosity of cortical tissue typically ranges from 5% to 30%, while that of cancellous tissue ranges from 30% to 90%. The apparent density of cortical tissue is about 1.8 g/cm, and that of cancellous tissue typically ranges from 0.1 to 1.0 g/cm. The distinction between cortical tissue and cancellous tissue is arbitrary, however, and in biomechanical terms the two tissues are often considered as one material with a specific range in porosity and density.

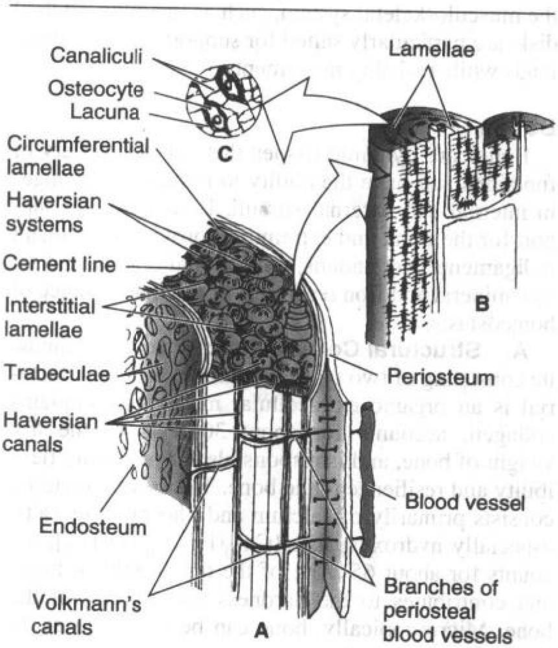

Figure 1–7. The structure of bone. **A:** A section of the diaphysis of a long bone, depicted without inner marrow. Each osteon is bounded by a cement line. **B:** Each osteon consists of lamellae, concentric rings composed of mineral matrix surrounding the haversian canal. **C:** Along the boundaries of the lamellae are small cavities known as lacunae, each of which contains a single osteocyte. Radiating from the lacunae are tiny canals, or canaliculi, into which the cytoplasmic processes of the osteocytes extend. (Reproduced, with permission, from Nordin M, Frankel VH: Biomechanics of bone. In Nordin M, Frankel VH [editors]: Basic Biomechanics of the Musculoskeletal System. Lea & Febiger, 1989.)

The organization of cortical and cancellous tissue in bone allows for adaptation to function. Cortical tissue always surrounds cancellous tissue, but the relative quantity of each type of tissue varies with the functional requirements of the bone. In long bones, the cortical tissue of the diaphysis is arranged as a hollow cylinder to best resist bending. The metaphyseal region of the long bones flares to increase the bone volume and surface area in a manner that minimizes the stress of joint contact. The cancellous tissue in this region provides an intricate network that distributes weight-bearing forces and joint reaction forces into the bulk of the bone tissue.

B. Biomechanical Behavior: The mechanical properties of cortical bone differ from those of cancellous bone. Cortical bone is stiffer than cancellous bone. While cortical bone will fracture in vivo when the strain exceeds 2%, cancellous bone will not fracture in vivo until the strain exceeds 75%. The larger capacity for energy storage (area under the stress-strain curve) of cancellous bone is a function of porosity. Despite different stiffness values for cortical and cancellous bone, the following axiom is valid for all bone tissue: the compressive strength of the tissue is proportional to the square of the apparent density, and the elastic modulus or material stiffness of the tissue is proportional to the cube of the apparent density. Therefore, any increase in porosity, as occurs with aging, will decrease the apparent density of bone, and this in turn will decrease the compressive strength and elastic modulus of bone.

Variations in the strength and stiffness of bone also result from specimen orientation (longitudinal versus transverse) and loading configuration (tensile, compressive, or shear). Generally, the strength and stiffness of bone are greatest in the direction of the common load application (longitudinally for long bones). With regard to orientation, cortical bone (Figure 1–8) is strongest in the longitudinal direction. With regard to loading configuration, cortical bone is strongest in compression and weakest in shear.

Tensile loading is the application of equal and opposite forces (loads) outward from the surface. Maximal stresses are in a plane perpendicular to the load application and result in elongation of the material. Microscopic studies show that the tensile failure in bones with haversian systems is caused by debonding of the cement lines and pull-out of the osteons. Bones

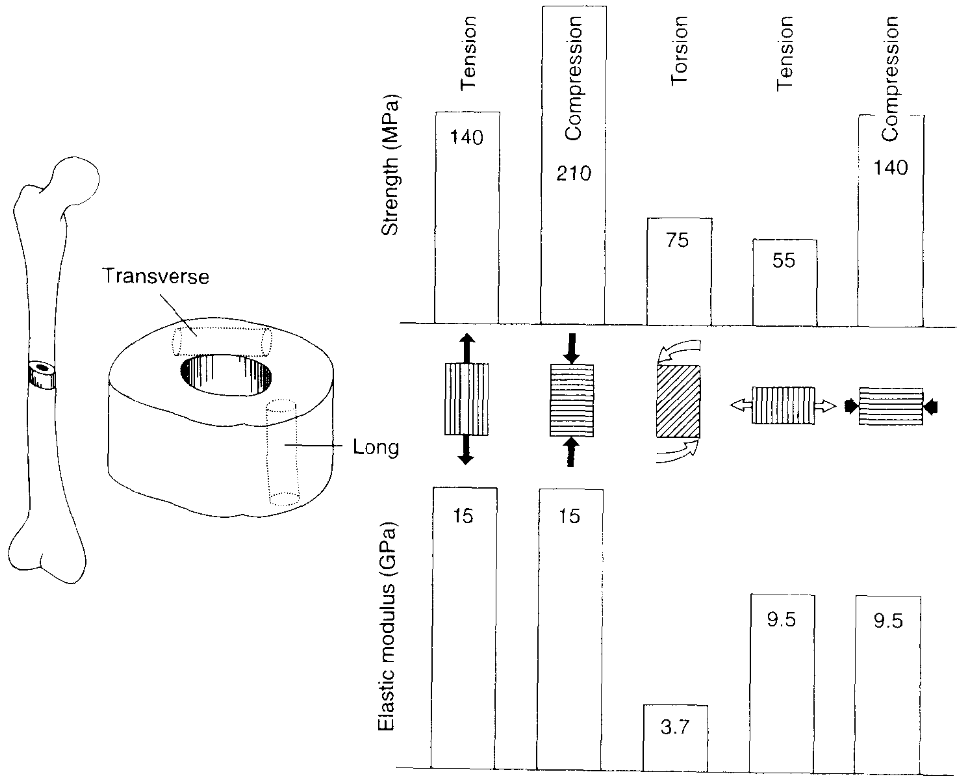

Figure 1–8. The effects of specimen orientation and loading configuration on the strength and elastic modulus of cortical bone from the diaphyseal region of a long bone.

with a large percentage of cancellous tissue demonstrate trabecular fracture with tensile loading.

The converse of tensile loading is **compressive loading,** which is defined as the application of equal and opposite forces toward the surface. Under compression, a material shortens and widens. Microscopic studies show that compressive failure occurs by oblique cracking of the osteons in cortical bone and by oblique cracking of the trabeculae in cancellous bone. Vertebral fractures, especially associated with osteoporosis, are associated with compressive loading.

The application of either a tensile load or a compressive load produces a shear stress in the material. **Shear loading** is the application of a load parallel to a surface, and the deformation is angular. Clinical studies show that shear fractures are most common to regions with a large percentage of cancellous bone, such as the tibial plateau.

Bone is a viscoelastic material, and its mechanical behavior is therefore influenced by strain rate. Bones are approximately 50% stiffer at high strain rates than at low strain rates, and the load to failure nearly doubles at high strain rates. The result is a doubling of the stored energy at high strain rates. Clinical studies show that the loading rate influences the fracture pattern and the associated soft tissue damage. Low strain rates, characterized by little stored energy, result in undisplaced fractures and no associated soft tissue damage. High strain rates, however, are associated with massive damage to the bone and soft tissue owing to the marked increase in stored energy.

Bone fractures can be produced either from a single load that exceeds the ultimate tensile strength of the bone or from repeated loading that leads to fatigue failure. Since bone is self-repairing, fatigue fracture occurs only when the rate of microdamage resulting from repeated loading exceeds the intrinsic repair rate of the bone. Fatigue fractures are most common during strenuous activity when the muscles have become fatigued and are therefore unable to adequately store energy and absorb the stress imposed on the bone. When the muscles are fatigued, the bone is required to carry the increased stress.

C. Remodeling Mechanisms: Bone has the ability to alter its size, shape, and structure in response to mechanical demands. According to Wolff's law regarding bone remodeling in response to stress, bone resorption occurs with decreased stress, bone hypertrophy occurs with increased stress, and the planes of increased stress follow the principal trabecular orientation. Thus, bone remodeling occurs under