FIFTH EDITION

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DeLee, Drez, & Miller's

Orthopaedic Sports Medicine

Principles and Practice

Mark D. Miller
Stephen R. Thompson

Volume 1

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DELEE, DREZ, & MILLER'S ORTHOPAEDIC SPORTS MEDICINE, FIFTH EDITION

ISBN: 978-0-323-54473-3

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International Standard Book Number: 978-0-323-54473-3

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Design Direction: Ryan Cook

Printed in China

Last digit is the print number: 9 8 7 6 5 4 3 2 1

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1600 John F. Kennedy Blvd.
Ste 1600
Philadelphia, PA 19103-2899

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To sports medicine professionals in every discipline and to athletes at every level and every sport. Without them there would be no sports medicine.

And to Drs. Jesse DeLee and David Drez. Thank you for entrusting us to keep your vision alive.

MARK D. MILLER

For Linden Ellis: You are too young to read this now, and you may choose never to read this later, but it is nonetheless still for you.

And, for Shannon. As always, thank you for everything you do, including enabling me to do what I do.

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PREFACE

What's in a name? In developing and editing the fifth edition of DeLee, Drez, & Miller's Orthopaedic Sports Medicine: Principles and Practices, we considered removing "orthopaedic" from the title. In an effort to keep pace with the rapidly growing specialty of sports medicine that includes internal medicine, pediatrics, rehabilitation medicine, athletic training, as well as orthopedics and many other disciplines, it is essential to expand the book's focus beyond orthopedics and address a more all-inclusive vision of sports medicine. Regardless of how sports medicine touches your individual practice, this updated version of this classic textbook remains the most comprehensive in the field.

As sports medicine continues to evolve, the need to include additional topics is essential. For this fifth edition, one focus is addressing problems of revision surgery. Important new chapters have been added on revision shoulder instability, revision rotator cuff, and revision anterior cruciate ligament surgery. Additionally, we have fine-tuned the organization of chapters under the dutiful watch of section editors. The hand section has been revised to include more comprehensive chapters on sports-related injuries. The hip section has been updated to reflect the current thinking on the hottest new topics in sports medicine, including new chapters on posterior hip pain and peritrochanteric lesions. And the non-operative sections have been extensively edited and expanded, to include new information on sports nutrition, psychological adjustment to athletic injury and genitourinary trauma in the athlete.

Incredibly, there are over 300 contributors to this new edition, many of whom are widely regarded as the foremost experts in their fields. To each of them, we extend heartfelt gratitude for their willingness to participate and share their knowledge. Similarly, we would be remiss if we did not thank our outstanding section editors: Drs. Bedi, Lesniak, Khodaee, Deu, Hart, Brockmeier, Kakar, Gwathmey, McCarty, Kadakia, Aiyer, Shen, and Milewski. They did the heavy lifting in the publishing process and we are deeply appreciative of their efforts.

It remains a distinct honor and pleasure to continue the tradition of Drs. Jesse DeLee and David Drez, who first produced this text in 1994. We hope it enables practitioners to remain on the cutting edge of sports medicine to the benefit of the widest possible spectrum of athletes and patients under our collective care.

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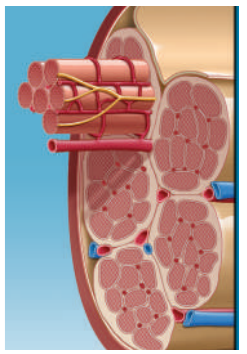

Physiology and Pathophysiology of Musculoskeletal Tissues

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TENDON AND LIGAMENT

Structure

Tendons and ligaments are both dense, regularly arranged connective tissues. The surface of the tendon is enveloped in a white, glistening, synovial-like membrane, called the *epitenon*, which is continuous on its inner surface with the *endotenon*, a thin layer of connective tissue that binds collagen fibers and contains lymphatics, blood vessels, and nerves. In some tendons, the epitenon is surrounded by a loose areolar tissue called the *paratenon,* which functions as an elastic sheath through which the tendon can slide. In some tendons, the paratenon is replaced by a true synovial sheath or bursa consisting of two layers lined by synovial cells, called the *tenosynovium*, within which the mesotendon carries important blood vessels to the tendon.[1] In the absence of a synovial lining, the paratenon often is called a *tenovagina*. Together the epitenon and the paratenon compose the *peritenon* (Fig. 1.1). The blood supply to tendons has several sources, including the perimysium, periosteal attachments, and surrounding tissues. Blood supplied through the surrounding tissues reaches the tendon through the paratenon, mesotenon, or vincula. Vascular tendons are surrounded by a paratenon and receive vessels along their borders; these vessels then coalesce within the tendon. The relatively avascular tendons are contained within tendinous sheaths, and the mesotenons within these sheaths function as vascularized conduits called *vincula*. The muscle-tendon and tendon-bone junctions, along with the mesotenon, are the three types of vascular supply to the tendon inside the sheath. Other sources of nutrition[2] include diffusional pathways from the synovial fluid, which provide an important supply of nutrients for the flexor tendons of the hand, for example. The nervous supply to a tendon involves mechanoreceptors located near the musculotendinous junction, which provide proprioceptive feedback to the central nervous system.

Ligaments grossly appear as firm, white fibrous bands, sheets, or thickened strips of joint capsule securely anchored to bone. They consist of a proximal bone insertion, the substance of the ligament or the capsule, and a distal bone insertion. Because most insertions are no more than 1 mm thick, they contribute only a small amount to the volume and the length of the ligament. Bundles of collagen fibrils form the bulk of the ligament substance.[3–5] Some ligaments consist of more than one band of collagen fibril bundles. For example, the anterior cruciate ligament (ACL) has a continuum of fiber lengths; different fibers become taut throughout the range of motion.[6] The alignment of collagen fiber bundles within the ligament substance generally follows the lines of tension applied to the ligament. This is in contrast to the alignment of collagen fiber bundles within the tendon, which is generally parallel to its longitudinal axis. In addition, thinner collagen fibrils extend the entire length of the tendon. Light microscopic examination has shown that the collagen bundles have a wave or crimp pattern. The crimp pattern of matrix organization may allow slight elongation of the ligament without incurring damage to the tissue.[6] In some regions, the ligament cells align themselves in rows between collagen fiber bundles, but in other regions, the cells lack apparent orientation relative to the alignment of the matrix collagen fibers. Scattered blood vessels penetrate the ligament substance, forming small-diameter, longitudinal vascular channels that lie parallel to the collagen bundles. Nerve fibers lie next to some vessels, and, like tendon, nerve endings with the structure of mechanoreceptors have been found in some ligaments.[4,7,8]

Tendon and ligament insertions vary in size, strength, angle of the ligament collagen fiber bundles relative to the bone, and proportion of ligament collagen fibers that penetrate directly into bone.[4,5,9] Based on the angle between the collagen fibrils and the bone and the proportion of the collagen fibers that penetrate directly into bone, investigators group tendon and ligament insertions into two types: direct and indirect. Direction insertions typically occur at the apophysis or epiphysis of bone, often within or around a synovial joint, and consist of sharply defined regions where the collagen fibers appear to pass directly into the cortex of the bone.[9,10] Although the thin layer of superficial collagen fibers of direct insertions joins the fibrous layer of the periosteum, most of the tendon or ligament insertions consist of deeper fibers that directly penetrate the cortex, often at a right angle to the bone surface. The deeper collagen fibers pass through four zones with increasing stiffness: ligament substance, fibrocartilage, mineralized fibrocartilage, and bone.[9,10] This four-zone interface is known as the fibrocartilaginous enthesis.[11] Dissipation of force is achieved effectively through this gradual transition from tendon to fibrocartilage to bone. A larger area of fibrocartilage can be found on one side of the insertion, which is thought to be an adaptation to the compressive forces experienced by the tendon or ligament on that side.[12] Conversely, indirect or oblique insertions, such as the tibial insertion of the medial collateral ligament of the knee or the femoral insertion of the lateral collateral ligament, typically occur at the metaphysis

Abstract

Musculoskeletal structures contain tissue-specific cells, extracellular matrix, and fiber arrangements which impart their unique biologic and mechanical properties. Tendon, ligament, meniscus, articular cartilage, and bone all have different structures that determine their specific function. Furthermore, the healing potential and response to injury of these tissues is highly variable and dependent on a number of factors, including the presence of a surrounding vasculature and the ability of intrinsic cells to replicate and remodel the injured matrix. In this chapter, we review the structure, biology, healing response to injury, and potential augmentative therapies of tendon, ligament, meniscus, articular cartilage, and bone.

Keywords

tendon
ligament
meniscus
cartilage
articular
bone
physiology

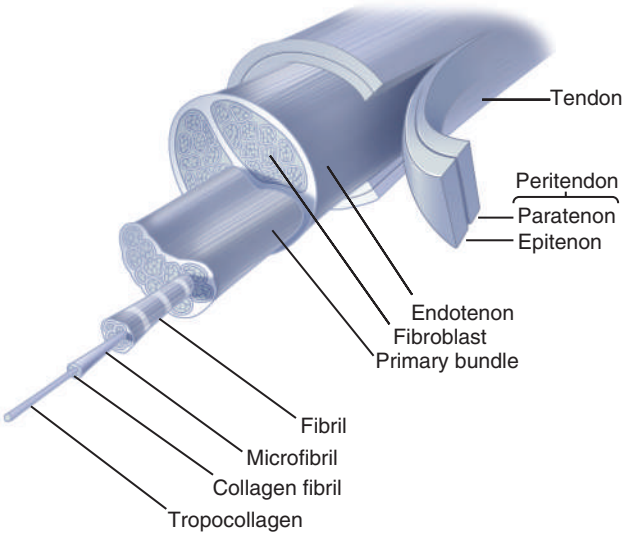

Fig. 1.1 Structural organization of tendon.

or diaphysis of bone without an intervening fibrocartilage zone. They usually cover more bone surface area than do direct insertions, and their boundaries cannot be easily defined because the collagen fibers pass obliquely along the bone surface rather than directly into the cortex.

Extracellular Matrix

Tendons and ligaments consist of relatively few cells and an abundant extracellular matrix primarily containing collagen, proteoglycans, and water. Tenocytes (tendon-specialized fibroblasts) are the dominant cell of tendons, whereas fibroblasts are the dominant cells of ligaments. Tenocytes and fibroblasts form and maintain the extracellular matrix. Within ligaments, fibroblasts vary in shape, activity, and density among regions of the same tissue and with the age of the tissue.[4,5,9,13] Both tenocytes and fibroblasts are spindle shaped, with fibroblasts being rounder, and extend between the collagen fibrils.[14] Endothelial cells of small vessels and nerve cell processes are also present.[4,5,9,13] Studies have shown that tendon and ligament contain a small population of resident stem cells which function to maintain tissue homeostasis during growth and repair.[15–17]

Type I collagen, which is the major component of the molecular framework, composes more than 90% of the collagen content of ligaments. Type III collagen constitutes approximately 10% of the collagen, and small amounts of other collagen types also may be present. Ligaments have a higher content of type III collagen than do tendons.[18] All types of collagen have in common a triple helical domain, which is combined differently with globular and nonhelical structural elements. The triple helix conformation of collagen is stabilized mainly by hydrogen bonds between glycine residues and between hydroxyl groups of hydroxyproline. This helical conformation is reinforced by hydroxyproline-forming and proline-forming hydrogen bonds to the other two chains. The physical properties of collagen and its resistance to enzymatic and chemical breakdown rely on covalent cross-links within and between the molecules.

Elastin is a protein that allows connective tissues to undergo large changes in geometry while expending little energy in the process. Tendons of the extremities possess small amounts of this structural protein, whereas most ligaments have little elastin (usually less than 5%), although a few, such as the nuchal ligament and the ligamentum flavum, have high concentrations (up to 75%). In most tendons, elastin is found primarily at the fascicle surface,[19] comprising less than 1% of the tendon by dry weight, and it is responsible for the crimp pattern of the tendon when viewed by a light microscope. Elastin forms protein fibrils or sheets, but elastin fibrils lack the cross-banding pattern of fibrillar collagen and differ in amino acid composition, including two amino acids not found in collagen (desmosine and isodesmosine). In addition, unlike collagen, elastin amino acid chains form random coils when the molecules are unloaded. This conformation of the amino acid chains makes it possible for elastin to undergo some deformation without rupturing or tearing and then, when the load is removed, to return to its original size and shape.

Approximately 1% of the total dry weight of tendon and ligament is composed of ground substance, which consists of proteoglycans, glycosaminoglycans, structural glycoproteins, plasma proteins, and a variety of small molecules. Most ligaments have a higher concentration of glycosaminoglycans than do tendons, due to the functional need for more rapid adaptation.[18] Proteoglycans and glycosaminoglycans both have important roles in organizing the extracellular matrix and control the water content of the tissue.[4,20–23] Tendon and ligaments contain two known classes of proteoglycans. Larger proteoglycans contain *long* negatively charged chains of chondroitin and keratan sulfate. Smaller proteoglycans contain dermatan sulfate. Because of their long chains of negative charges, the large articular cartilage-type proteoglycans tend to expand to their maximal domain in solution until restrained by the collagen fibril network. As a result, they maintain water within the tissue and exert a swelling pressure, thereby contributing to the mechanical properties of the tissue and filling the regions between the collagen fibrils. The small leucine-rich proteoglycans usually lie directly on the surface of collagen fibrils and appear to affect formation, organization, and stability of the extracellular matrix, including collagen fibril formation and diameter. They may also control the activity of growth factors by direct association.[21,24]

Although noncollagenous proteins contribute only a small percentage of the dry weight of dense fibrous tissues, they appear to help organize and maintain the macromolecular framework of the collagen matrix, aid in the adherence of cells to the framework, and possibly influence cell function. One noncollagenous protein, fibronectin, has been identified in the extracellular matrix of ligaments and may be associated with several matrix component molecules and with blood vessels. Other noncollagenous proteins undoubtedly exist within the matrix, but their identity and their functions have not yet been defined. Many of the noncollagenous proteins also contain a few monosaccharides and oligosaccharides.[4,5]

Injury

Acute strains and tears to tendons and ligaments disrupt the matrix, damage blood vessels, and injure or kill cells. Damage to cells, matrices, and blood vessels and the resulting hemorrhage

start a response that leads to a sequential process of inflammation, repair, and remodeling.[25,26] These events form a continuous sequence of cell, matrix, and vascular changes that begins with the release of inflammatory mediators and ends when remodeling ceases.[25] As with any injury to biologic tissue, acute inflammation lasts 48 to 72 hours after the injury and then gradually resolves as repair progresses. Some of the events that occur during inflammation, including the release of cytokines or growth factors, may help to stimulate tissue repair.[25] These mediators promote vascular dilation and increase vascular permeability, leading to exudation of fluid from vessels in the injured region, which causes tissue edema. Blood escaping from the damaged vessels forms a hematoma that temporarily fills the injured site. Fibrin accumulates within the hematoma, and platelets bind to fibrillar collagen, thereby achieving hemostasis and forming a clot consisting of fibrin, platelets, red cells, and cell and matrix debris. The clot provides a framework for vascular and fibroblast cell invasion. As they participate in clot formation, platelets release vasoactive mediators and various cytokines or growth factors (e.g., transforming growth factor-β [TGF-β] and platelet-derived growth factor). Polymorphonuclear leukocytes appear in the damaged tissue and the clot. Shortly thereafter, monocytes arrive and increase in number until they become the predominant cell type. Enzymes released from the inflammatory cells help to digest necrotic tissue, and monocytes phagocytose small particles of necrotic tissue and cell debris. Endothelial cells near the injury site begin to proliferate, creating new capillaries that grow toward the region of tissue damage. Release of chemotactic factors and cytokines from endothelial cells, monocytes, and other inflammatory cells helps to stimulate migration and proliferation of the fibroblasts that begin the repair process.[25]

Overuse tendon injury is one of the more common forms of musculoskeletal injury and clinical causes of pain, although controversy exists in the literature about a universal classification and the responsible pathologic entities. A classification of Achilles tendon disorders[27] provides a guide to the structural manifestations of overuse injury as follows: (1) peritendinitis, or inflammation of the peritenon; (2) tendinosis with peritendinitis; (3) tendinosis without peritendinitis; (4) partial rupture; and (5) total rupture. Other classifiers have added a sixth category, tendinitis, in which the primary site of injury is the tendon, with an associated reactive peritendinitis.[28] The classification is not universal because some tendons lack a paratenon and instead have synovial sheaths; furthermore, it is unclear if certain histopathologic conditions are actually separate entities. For instance, human biopsy studies have been unable to show histologic evidence of acute inflammation within the tendon substance.[29] Because of uncertainty regarding the histologic features of these conditions, several authors have suggested use of the term *tendinopathy* rather than *tendinitis*.[30,31]

Studies have shown that in cases of chronic tendinosis, the pathologic lesion is typical of a degenerative process rather than an inflammatory one and that this degeneration occurs in areas of diminished blood flow. Several authors have documented the existence of areas of marked degeneration without acute or chronic inflammatory cell accumulation in most of these cases.[32–34] These changes are separate and distinct from the site of rupture. A review of patients with chronic tendinitis syndrome revealed similar findings of tendon degeneration.[27,35] Nirschl[35] described the pathology of chronic tendinitis as "angiofibroblastic hyperplasia." A characteristic pattern of fibroblasts and vascular, atypical, granulation-like tissue can be seen microscopically.[35,36] Cells characteristic of acute inflammation are virtually absent. These observations suggest that factors other than mechanical overuse play an important role in the pathogenesis of these tendon lesions.

In several studies, a correlation between age and the incidence of chronic tendinopathy has been identified.[37,38] In vitro studies have shown decreased proliferative and metabolic responses of aging tendon tissue.[39] Other causative factors include the lack of blood flow in certain areas (e.g., supraspinatus and Achilles tendon) that may predispose a tendon to rupture or may result in chronic tendinopathy.[40] Biopsy specimens of young patients with symptoms of chronic tendinopathy have revealed a change in the morphology of tenocytes adjacent to areas of collagen degeneration.[28]

Repair

Tendons and ligaments may possess both intrinsic and extrinsic capabilities for healing, and the contribution of each of these two mechanisms probably depends on the location, extent, and mechanism of injury and the rehabilitation program used after the injury. Several studies[2,41–46] have suggested that the inflammatory response is not essential to the healing process and that these tissues possess an intrinsic capacity for repair. Recent research has isolated intrinsic stem cells within tendon and ligament, although their in vivo identities, niche, and role in healing remain controversial.[17,47] Lindsay and Thomson[43] were the first to show that an experimental tendon suture zone can be isolated from the perisheath tissues and that healing progressed at the same rate as when the perisheath tissues were intact. Later, in isolated segments of profundus tendon in rabbits, these researchers found anabolic and catabolic enzymes, which showed that an active metabolic process existed in the isolated tendon segments.[44]

As in other areas in the body, tendon healing proceeds in three phases: (1) an inflammatory stage, (2) a reparative or collagen-producing stage, and (3) a remodeling phase.

Inflammatory Phase

Tendon and ligament healing begins with hematoma formation and an inflammatory reaction that includes an accumulation of fibrin and inflammatory cells. A clot forms between the two ends and is invaded by cells resembling fibroblasts and migratory capillary buds. Within 2 to 3 days of the injury, fibroblasts within the wound begin to proliferate rapidly and synthesize new matrix. They replace the clot and the necrotic tissue with a soft, loose fibrous matrix containing high concentrations of water, glycosaminoglycans, and type III collagen. Inflammatory cells and fibroblasts fill this initial repair tissue. Within 3 to 4 days, vascular buds from the surrounding tissue grow into the repair tissue and then canalize to allow blood flow to the injured tissue and across small tissue defects. This vascular granulation tissue fills the tissue defect and extends for a short distance into the surrounding tissue but has little tensile strength. The inflammatory phase is evident until the 8th to 10th day after injury.

Reparative Phase

As the repair progresses during the next several weeks, proliferating fibroblasts continue to produce fibrous tissue containing a high proportion of type III collagen. Collagen synthesis reaches its maximal level after approximately 4 weeks, and at 3 months, collagen synthesis continues at a rate 3 to 4 times that of normal tissue. Over time, water, glycosaminoglycan, and type III collagen concentrations decline, the inflammatory cells disappear, and the concentration of type I collagen increases. Newly synthesized collagen fibrils increase in size and begin to form tightly packed bundles, and the density of fibroblasts decreases. Matrix organization increases[48–51] as the fibrils begin to align along the lines of stress, the number of blood vessels decreases, and small amounts of elastin may appear within the site of injury. The tensile strength of the repair tissue increases as the collagen concentration increases.

Remodeling Phase

Repair of many tendon and ligament injuries results in an excessive volume of highly cellular tissue with limited mechanical properties and a poorly organized matrix. Remodeling reshapes and strengthens this tissue by removing, reorganizing, and replacing cells and matrix.[25] In most tendon and ligament injuries, evidence of remodeling appears within several weeks of injury as fibroblasts and macrophages decrease, fibroblast synthetic activity decreases, and fibroblasts and collagen fibrils assume a more organized appearance. As these changes occur in the repair tissue, collagen fibrils grow in diameter, the concentration of collagen and the ratio of type I to type III collagen increase, and the water and proteoglycan concentrations decline. During the months after the injury occurs, the matrix continues to align, presumably in response to loads applied to the repair tissue. The most apparent signs of remodeling disappear within 4 to 6 months of injury. However, removal, replacement, and reorganization of repair tissue continue to some extent for years.[50,52,53] The mechanical strength of the healing tendon and ligament increases as the collagen becomes stabilized by cross-links and the fibrils assemble into fibers.

Factors Affecting Healing

Among the most important variables that affect healing of tendon and ligament are the type of tendon or ligament, the size of the tissue defect, and the amount of load applied to the repair tissue. For example, injuries to capsular and extra-articular ligaments stimulate production of repair tissue that will fill most defects, but injuries to intra-articular ligaments, such as the ACL, often fail to produce a successful repair response. Treatments that achieve or maintain apposition of torn tissue and that stabilize the injury site decrease the volume of repair tissue necessary to heal the injury, which can benefit the healing process. Such treatments may also minimize scarring and help to provide near-normal tissue length. For these reasons, avoidance of wide separation of ruptured tendon or ligament ends and selection of treatments that maintain some stability at the injured site during the initial stages of repair are generally desirable.

Early excessive loading in the immediate postoperative period may have a deleterious effect on tendon and ligament healing by disrupting the repair tissue, leading to gap formation and ischemia, adverse changes in tendon matrix, and possible rupture.[4,25,54–56] However, controlled loading of tendon and ligament repair tissue can promote healing and enhance the mechanical and biologic characteristics of tendon-to-bone healing.[57] The optimal amount of tension necessary to promote an acceptable clinical response is currently not well understood and depends on the type of tissue and healing environment, but it is clear that remodeling of collagen scar tissue into mature tendon tissue depends on the presence of tensile forces.[58,59] The concept of immediate passive mobilization after flexor tendon repair in the hand was introduced by Kleinert and coworkers,[60] who showed that, during limited active extension, reciprocal relaxation of the flexor tendons occurs, allowing passive extension of the repaired tendon. This controlled passive motion was found to be effective experimentally and clinically in decreasing the tethering effect of adhesions and in improving the rates of tendon repair, gliding function, and strength of the tendon.

Methods for Augmentation of Tendon and Ligament Healing

A large body of research has demonstrated the potential for growth factors to improve tendon and ligament tissue healing by stimulation of cell proliferation, chemotaxis, matrix synthesis, and cell differentiation (summarized in Table 1.1). In addition to multifunctional cytokines such as TGF-β and platelet-derived growth factor, work has focused on recapitulating the cellular and molecular signals that are expressed during embryonic tendon development, such as scleraxis and TGF-β3.[61] However, challenges in the delivery of these growth factors, specifically regarding the optimal carrier vehicles and proper dosing regimen, to the desired site still remain.

Platelet-rich plasma (PRP), an autologous blood concentrate, can be used to locally deliver a high concentration "cocktail" of cytokines and has gained popularity as a treatment modality for tendon and ligament injuries. Recent studies have reported potentially promising results with the use of PRP to augment healing of rotator cuff repair[62–64] and patellar tendinopathy.[65] However, the results of PRP for augmentation of tendon and ligament healing have been variable, which can partially be attributed to the lack of understanding of the optimal PRP formulation for different tissues and pathologies, as well as the tremendous variability in the methods of PRP production among commercial systems.[66,67] To complicate matters further, within a given separation technique, there is a high degree of intersubject and intrasubject variability in the composition of PRP produced.[68]

Cell-based approaches appear promising for tendon and ligament tissue engineering and improvement of healing. Therapies using mesenchymal stem cells (MSCs) derived from adipose and bone marrow to augment tendon and ligament healing have garnered the most attention due to their multipotent potential and ability to exert a paracrine effect to modulate and control inflammation, stimulate endogenous cell repair and proliferation, inhibit apoptosis, and improve blood flow.[14,69] However, like PRP augmentation therapy, continued research is needed to identify the optimal cell source and the ideal treatment protocol needed to drive differentiation of these or neighboring

TABLE 1.1 Growth Factors in Soft Tissue Repair

Biologic Factor	Functions	Reference
TGF-β	Influx of mononuclear cells and fibroblasts	Lee J et al: *Iowa Orthop J* 1998
	Enhanced collagen deposition	Spindler KP et al: *J Orthop Res* 2002
		Spindler KP et al: *J Orthop Res* 2003
		Kashiwagi K et al: *Scand J Plast Reconstr Surg Hand Surg* 2004
		Kim HJ et al: *Connect Tissue Res* 2007
		Kim HM et al: *Connect Tissue Res* 2011
		Manning CN et al: *J Orthop Res* 2011
		Kovacevic D et al: *Am J Sports Med* 2011
GDF 5/6/7	Influx of mononuclear cells and fibroblasts	Wolfman NM et al: *J Clin Invest* 1997
	Enhanced collagen deposition	Aspenberg P et al: *Acta Orthop Scand* 1999
		Rickert M et al: *Growth Factors* 2001
		Forslund C et al: *J Orthop Res* 2003
		Virchenko O et al: *Scand J Med Sci Sports* 2005
		Fealy S et al: *Am J Sports Med* 2006
		Dines JS et al: *J Shoulder Elbow Surg* 2007
		Saiga K et al: *Biochem Biophys Res Commun* 2010
		Date H et al: *J Orthop Res* 2010
IGF-1	Proliferation of fibroblasts	Abrahamsson SO et al: *J Orthop Res* 1991
	Enhanced collagen deposition	Abrahamsson SO et al: *J Orthop Res* 1996
		Kurtz CA et al: *Am J Sports Med* 1999
		Dahlgren LA et al: *J Orthop Res* 2002
		Dahlgren LA et al: *J Orthop Res* 2005
		Provenzano PP et al: *BMC Physiol* 2007
PDGF-B	Influx of mononuclear cells and fibroblasts	Lee J et al: *Iowa Orthop J* 1998
	Enhanced angiogenesis	Hildebrand KA et al: *Am J Sports Med* 1998
	Enhanced collagen deposition	Nakamura N et al: *Gene Ther* 1998
		Kobayashi M et al: *J Shoulder Elbow Surg* 2006
		Uggen C et al: *Arthroscopy* 2010
		Hee CK et al: *Am J Sports Med* 2011
bFGF	Proliferation of fibroblasts	Lee J et al: *Iowa Orthop J* 1998
	Enhanced collagen deposition	Cool SM et al: *Knee Surg Sports Traumatol Arthrosc* 2004
		Saiga K et al: *Biochem Biophys Res Commun* 2010
		Date H et al: *J Orthop Res* 2010
HGF	Enhanced angiogenesis	Ueshima K et al: *J Orthop Sci* 2011
	Enhanced collagen deposition	
PRP	Enhanced angiogenesis	Murray MM et al: *J Orthop Res* 2006
	Enhanced collagen deposition	Murray MM et al: *J Orthop Res* 2007
		Joshi SM et al: *Am J Sports Med* 2009
VEGF	Enhanced angiogenesis	Boyer MI et al: *J Orthop Res* 2001
	Enhanced collagen deposition	Petersen W et al: *Arch Orthop Trauma Surg* 2003
BMP-12	Enhanced ossification	Aspenberg P et al: *Scand J Med Sci Sports* 2000
	Enhanced angiogenesis	Lou J et al: *J Orthop Res* 2001
	Enhanced collagen deposition	Seeherman HJ et al: *J Bone Joint Surg Am* 2008

bFGF, Basic fibroblast growth factor; *BMP-12,* bone morphogenetic protein-12; *GDF,* growth/differentiation factor; *HGF,* human growth factor; *IGF-1,* insulin-like growth factor-1; *PDGF-β,* platelet-derived growth factor-β; *PRP,* plasma-rich protein; *TGF-β,* transforming growth factor-β; *VEGF,* vascular endothelial growth factor.

cells into mature tenocytes and fibroblasts. Recent studies have identified resident tissue-specific stem cells in the perivascular regions of native tendon and ligament that detach from vessels in response to injury, migrate into the interstitial space, and deposit extracellular matrix,[70,71] although their precise potential for use in augmenting tendon and ligament healing remains to be elucidated.

Research has also investigated scaffold materials to augment tendon repair and ligament reconstruction. Porcine-derived small intestine submucosa has been used as a collagen scaffold to augment Achilles tendon and rotator cuff tendon repair. However, negative clinical results have been reported, including inflammatory/immunologic response to the small intestine submucosa material believed to be due to residual porcine DNA in the implant.[72,73] Various other allografts and xenografts, such as collagen allograft matrices and porcine dermal xenografts, are commercially available and differ from porcine small intestine submucosa in both biologic and mechanical composition.[74,75] Nanomaterials are promising for tendon and ligament tissue engineering because the microstructure of the material mimics

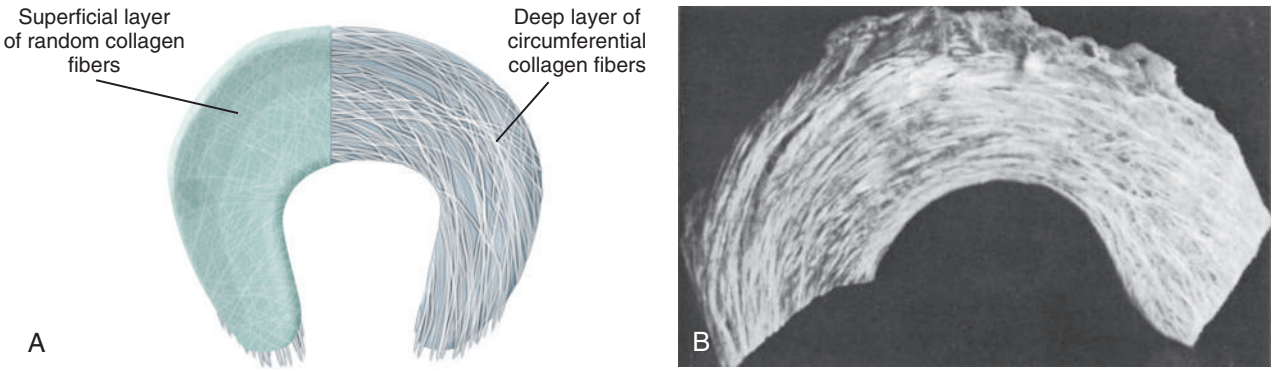

Fig. 1.2 (A) Diagram of collagen fiber architecture throughout the meniscus. Collagen fibers of the thin superficial sheet are randomly distributed in the plane of the surface and are predominantly arranged in a circumferential fashion deep in the substance of the tissue. (B) Macrophotograph of bovine medial meniscus with the surface layer removed, showing the large circumferentially arranged collagen bundles of the deep zone. ([A] Modified from Bullough PG, Munuera L, Murphy J, et al. The strength of the menisci of the knee as it relates to their fine structure. *J Bone Joint Surg Br.* 1970;52:564–570. [B] From Proctor CS, Schmidt MB, Whipple RR, et al. Material properties of the normal medial bovine meniscus. *J Orthop Res.* 1989;7:771–782.)

native extracellular matrix. Multiphasic scaffolds are being used to create bone-ligament composites.[76] In addition to various scaffold materials and cell types, it has become clear that mechanical stimulation of the neotissue is also critical to optimize the structure and composition of the tissue.[77] The specific scaffold can be modified in vitro by seeding marrow stromal cells on the scaffold and applying cyclic stretching to increase the alignment of cells, as well as to improve the production and orientation of collagen. When applied in vivo, such a tissue-engineered scaffold could serve to accelerate the healing process, ultimately helping to make a better neoligament or tendon.

MENISCUS

Structure

Human menisci are semilunar in shape[78] and consist of a sparse distribution of cells surrounded by an abundant extracellular matrix.[79–81] The meniscus functions to optimize force transmission and provide stability to the knee. The medial meniscus is the dominant secondary stabilizer in an ACL-deficient knee during the Lachman maneuver,[82] whereas the lateral meniscus is the dominant secondary stabilizer in an ACL-deficient knee during the pivot shift maneuver.[83] Within the meniscus lies an anisotropic, inhomogeneous, and highly ordered arrangement of collagen fibrils. The meniscal surface is composed of a randomly woven mesh of fine collagen type II fibrils that lie parallel to the surface. Below this surface layer, large, circumferentially arranged collagen fiber bundles (mostly type I) spread through the body of the tissue (Fig. 1.2).[84,85] These circumferential collagen bundles give menisci great tensile stiffness and strength parallel to their orientation.[85] The collagen bundles insert into the anterior and the posterior meniscal attachment sites on the tibial plateau, providing for rigid and strong attachment sites. Fig. 1.2A illustrates these large fiber bundles and the thin superficial surface layer. Fig. 1.2B is a photograph of a bovine medial meniscus with the surface layer removed, showing the large collagen bundles of the deep zone.

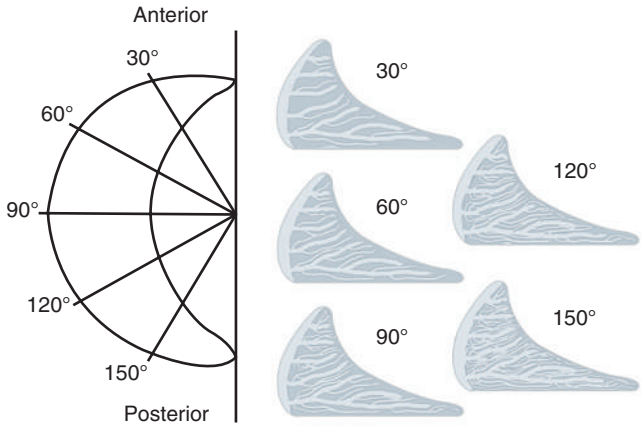

Fig. 1.3 Radial collagen fiber bundles of the meniscus. Radial tie fibers consisting of branching bundles of collagen fibrils extend from the periphery of the meniscus to the inner rim in every radial section throughout the meniscus. They are more abundant in the posterior sections and gradually diminish in number as the sections progress toward the anterior region of the meniscus. (Modified from Kelly MA, Fithian DC, Chern KY, et al. Structure and function of meniscus: basic and clinical implications. In: Mow VC, Ratcliffe A, Woo SL, eds. *Biomechanics of Diarthrodial Joints.* Vol 1. New York: Springer-Verlag; 1990.)

Radial sections of meniscus (Fig. 1.3) show radially oriented bundles of collagen fibrils, or "radial tie fibers," among the circumferential collagen fibril bundles, weaving from the periphery of the meniscus to the inner region.[85,86] These tie fibers help to increase the stiffness and the strength of the tissue in a radial direction, thereby resisting longitudinal splitting of the collagen framework. In cross section, these radial tie fibers appear to be more abundant in the posterior sections than in the anterior sections of the meniscus.[87]

Unlike articular cartilage, the peripheral 25% to 30% of the lateral meniscus and the peripheral 30% of the medial meniscus[78,88–90] have a blood supply, and the peripheral regions of the meniscus, especially the meniscal horns,[91,92] have a nerve supply as well. Branches from the geniculate arteries form a capillary

plexus along the peripheral borders of the menisci, with the medial inferior geniculate artery supplying the peripheral medial meniscus and the lateral inferior geniculate artery supplying the peripheral lateral meniscus.[88,89] Small radial branches project from these circumferential parameniscal vessels into the meniscal substance.[90] The central aspects of the meniscus do not have a direct arterial supply and instead receive nutrients primarily through synovial fluid diffusion.

Extracellular Matrix

The mechanical functions of the menisci depend on a highly organized extracellular matrix consisting of fluid and a macromolecular framework formed of collagen (types I, II, III, V, and VI), proteoglycans, elastin, and noncollagenous proteins, along with the cells that maintain this matrix.

Based on morphologic characteristics, two major types of meniscal cells exist.[80,93] Near the surface, the cells have flattened ellipsoid or fusiform shapes and are considered more fibroblastic; in the deep zone, the cells are spherical or polygonal and considered more chondrocytic. The superficial and the deep meniscal cells appear to have different metabolic functions and perhaps different responses to loading.[94] Like most other mesenchymal cells, these cells lack cell-to-cell contacts. Because most of the cells lie at a distance from blood vessels, they rely on diffusion through the matrix for transport of nutrients and metabolites. The membranes of meniscal cells attach to matrix macromolecules through adhesion proteins (e.g., fibronectin, thrombospondin, and type VI collagen).[80] The matrix, particularly the pericellular region, protects the cells from damage due to physiologic loading of the tissue. Deformation of the macromolecular framework of the matrix causes fluid flow through the matrix[94,95] and influences meniscal cell function. Because meniscal tissue is more fibrous than is hyaline cartilage, some authors have proposed that meniscal cells be called fibrochondrocytes.[80,96]

Water comprises 65% to 75% of the total weight of the meniscus.[94,95,97] Some portion of this water may reside within the intrafibrillar space of the collagen fibers.[98,99] Most of the water is retained within the tissue in the solvent domains of the proteoglycans due to both their strong hydrophilic tendencies and the Donnan osmotic pressure exerted by the counter ions associated with the negative charge groups on the proteoglycans.[94,100,101] Because the pore size of the tissue is extremely small (<60 nm), very large hydraulic pressures are required to overcome the impact of frictional resistance when forcing fluid flow through the tissue. These interactions between water and the macromolecular framework of the matrix significantly influence the viscoelastic properties of the tissue.

Some meniscal regions have a proteoglycan concentration of up to 3% of their dry weight.[80,81,95,102] Like proteoglycans from other dense fibrous tissues, meniscal proteoglycans can be divided into two general types. The large, aggregating proteoglycans expand to fill large volumes of matrix and contribute to tissue hydration and the mechanical properties of the tissue. The smaller, nonaggregating proteoglycans usually have a close relationship with fibrillar collagen.[80,103,104] The large aggregating meniscal proteoglycans have the same structure as the large aggregating proteoglycans from articular cartilage.[81,104] The concentration of large aggregating proteoglycans suggests that they probably contribute less to the properties of meniscus than to the properties of articular cartilage.[94,95] As with the quantitatively minor collagens, the smaller nonaggregating meniscal proteoglycans may help to organize and stabilize the matrix, but currently their exact function remains unknown.

Noncollagenous proteins also form part of the macromolecular framework of the meniscus and may contribute as much as 10% of the dry weight of the tissue in some regions.[80] Two specific noncollagenous proteins, link protein and fibronectin, have been identified in the meniscus.[80] Link protein is required for the formation of the stable proteoglycan aggregates that are capable of forming strong networks.[105,106] Fibronectin serves as an attachment protein for cells in the extracellular matrix.[107] Other noncollagenous proteins such as thrombospondin[108] may serve as adhesive proteins in the tissue, thus contributing to the structure and the mechanical strength of the matrix; however, the exact details of their composition and function in the meniscus remain largely unknown.

Finally, elastin contributes less than 1% of the dry weight of the meniscus.[80] The contribution of elastin to the mechanical properties of meniscal tissue is not well understood because the sparsely distributed elastic fibers are unlikely to play a significant role in the organization of the matrix or in determining the mechanical properties of the tissue.

Injury

Traumatic meniscal tears occur most frequently in young, active people. Tension, compression, or shear forces that exceed the strength of the meniscal matrix in any direction can lead to tissue failure. Acute traumatic injuries of normal meniscal substance usually produce longitudinal or transverse tears, although the morphology of these tears can be highly variable, including oblique, radial horizontal, bucket-handle, and complex tears. The configuration of tears due to overloading of normal meniscal tissue depends strongly on the direction of the load and the rate of stretch.[94] Unlike acute traumatic tears through apparently normal meniscal tissue, degenerative meniscal tears occur as a result of age-related changes in the tissue. These degenerative tears are most common in persons older than 40 years. Often, these persons do not recall a specific injury, or they recall only a minor load applied to the knee. Degenerative tears often have complex shapes or may appear as horizontal clefts or flaps, as though they were produced by shear failure. Multiple degenerative tears often occur within the same meniscus. These features of degenerative meniscal tears suggest that they result more from age-related changes in the collagen-proteoglycan solid matrix than from specific acute trauma.

The response of meniscal tissue to tears depends on whether the tear occurs through a vascular or an avascular portion of the meniscus.[89] The peripheral, vascular regions respond to injury as other vascularized, dense fibrous tissues do. The tissue damage initiates a sequence of cellular and vascular events including inflammation, repair, and remodeling[109] that can result in healing and restoration of tissue structure and function. Although tears through the vascular regions of the meniscus can typically heal, tears through the avascular regions do not typically heal

spontaneously, resulting in tissue deficiency.[78,89] Therefore strategies for meniscal repair in the avascular zone are continuously being explored.[110]

Meniscal Repair

Factors Affecting Healing

Repair in vascular regions of the meniscus. Partial meniscal resection through the peripheral vascularized region or complete meniscal resection initiates production of repair tissue that can extend from the remaining peripheral tissue into the joint.[111–116] Although the repair cells usually fail to replicate normal meniscal tissue, many authors have referred to this phenomenon as *meniscal regeneration*.[114,116] Some repaired menisci grossly resemble normal menisci, but the functional capabilities and mechanical properties of this "regenerated" meniscal tissue have not been comprehensively studied. Surgeons have reported meniscal regeneration in many clinical situations. Investigators have also examined the tissue produced by meniscal regeneration in animals. Meniscal regeneration can occur repeatedly in the same knee[115] and occasionally occurs after total knee replacement.[114] In rabbits, meniscal regeneration occurs more frequently on the medial side of the knee than on the lateral side, and development of degenerative changes in articular cartilage after a meniscectomy is inversely correlated with the extent of meniscal regeneration.[111–113] Synovectomy appears to prevent meniscal regeneration, which suggests that synovial cells contribute to the formation of meniscal repair tissue. The mechanisms and conditions that promote this type of repair, its functional importance, and the factors related to the predictability and frequency of meniscal regeneration remain unknown.

Repair in avascular portions of the meniscus. The response of meniscal tissue to tears in the avascular portion resembles the response of articular cartilage to lacerations in many respects. Experimental studies show that a penetrating injury to the avascular region of the meniscus causes no apparent repair or inflammatory reaction. Meniscal cells in the injured region, like chondrocytes in the region of an injury limited to the articular cartilage, may proliferate and synthesize new matrix, but they appear to be incapable of migrating to the site of the defect or producing enough new matrix to fill it. The ineffective response of meniscal cells in the avascular region of the meniscus has led investigators to develop novel methods to stimulate repair. Some promising approaches include creation of a vascular access channel to the injury site and stimulation of cell migration to the avascular region using implantation of a fibrin clot, an artificial matrix, or growth factors.[117] Synovial abrasion has also been shown to stimulate proliferation of the synovial fringe into the meniscus and allows blood vessels to enter the avascular regions. Although early results appear promising, the quality of the repair tissue, its biomechanical properties, and the long-term results of these methods have not been evaluated.

Augmentation of Meniscus Healing

Given the well-established poor intrinsic healing potential of the meniscus, particularly in avascular regions, intense interest exists regarding methods to augment healing using cytokines, exogenous cells, and scaffolds. Fibroblast growth factor-2 and connective tissue growth factor have been evaluated in rabbit models,[118,119] and vascular endothelial growth factor has been tested in a sheep meniscus tear model.[120,121] Although these cytokines appear to have a positive effect on basic meniscal fibrochondrocyte biology, the challenge at this time is to identify the optimal carrier vehicles and dosage to translate these preclinical data to clinical trials. Although some studies have suggested that PRP, as a source of cytokines, may confer some benefit in meniscus healing,[122,123] other studies have demonstrated no differences in outcomes or reoperation rates.[124] In addition, in an animal model of meniscus injury, PRP treatment increased hypertrophic fibrous tissue rather than meniscal cartilage.[125] Thus further investigation is necessary to better elucidate the role of growth factors and PRP in meniscal healing.

Cell-based approaches have also been evaluated for augmentation of biologic healing mechanisms. Various sources of both autogenous[126] and allogeneic cells[127] have been evaluated using different carrier materials. Both differentiated cells, such as chondrocytes, and undifferentiated cells, such as MSCs,[128,129] have been tested in animal models. Few human studies investigating the role of MSCs for meniscal repair have been performed.[130–133] Although the authors suggest that MSCs may be effective in repairing meniscal tears, these studies are limited, and more rigorous, placebo-controlled trials are necessary.[134]

Scaffolds. The use of scaffold materials to replace a portion of the damaged meniscus or to replace the entire structure is an appealing option and has the theoretical benefit of providing mechanical stability to the injured site while allowing for cell attachment and proliferation.[135,136]A collagen-based scaffold (Collagen Meniscus Implant, Menaflex, ReGen Biologics, Glen Rock, NJ) and a resorbable porous polyurethane-based scaffold (Actifit, Orteq Sport Medicine, London, United Kingdom) have demonstrated satisfactory clinical outcome in up to 80% of cases at up to 10 years and 2 years of follow-up, respectively.[137,138] Both of these devices are designed for partial meniscus replacement. Although it remains unclear whether the use of such scaffolds can affect the long-term sequelae of meniscectomy, early results are promising and may represent a new horizon in the treatment of these complex injuries. Further optimization of these materials may occur by incorporating undifferentiated cells into the scaffold.

ARTICULAR CARTILAGE

Synovial joints allow the rapid controlled movements necessary to support joint motion and to participate in sports. Normal function of these complex diarthrodial structures depends on the structural integrity and macromolecular composition of articular cartilage. Sports-related traumatic disruptions of cartilage structure and alterations in the macromolecular composition or organization change the biomechanical properties of the tissue, compromise joint function, and can lead to progressive pain and disability.

The specialized composition and organization of hyaline articular cartilage impart its unique biomechanical properties that permit normal synovial joint function. In the joint, cartilage distributes the loads of articulation, thereby minimizing peak

stresses acting on the subchondral bone. The tensile strength of the tissue provides its structural integrity under such loads. Alterations in the mechanical properties of cartilage due to injury, disease, or increasing age have not been well defined, but the available information shows that these properties change with age and loss of structural integrity. Cartilage from skeletally immature joints (open growth plates) is much stiffer than cartilage from skeletally mature joints (closed growth plates).[139] Older cartilage and fibrillated cartilage have much lower tensile stiffness and strength.[109] Participation in sports often subjects the articular cartilage to intense repetitive, compressive high-energy impact forces that can cause tissue injury. These abnormally large forces generate high shear stresses at the cartilage-subchondral bone junction, causing matrix disruption and death of the articular chondrocytes that may lead to early osteoarthritis.[140,141] Because cartilage is aneural, patients with pure chondral injuries can remain asymptomatic.

Structure and Composition of Articular Cartilage

Like the dense fibrous tissues and meniscus, articular cartilage consists of cells, matrix water, and a matrix macromolecular framework.[142] Unlike the most dense fibrous tissues, cartilage lacks nerves, blood vessels, and a lymphatic system. The composition, organization, and mechanical properties of the matrix of articular cartilage and the cell morphology and function vary according to the depth from the articular surface (Fig. 1.4).[143,144] Matrix composition, organization, and function also vary with distance from the cell.[109] Morphologic changes in articular cartilage cells and matrix from the articular surface to the subchondral bone make it possible to identify four zones or layers of articular cartilage.

Zones of Articular Cartilage

Superficial zone. The thinnest zone, the superficial tangential zone, has two layers. A sheet of fine fibrils without cells covers the joint surface (see Fig. 1.4). On phase-contrast microscopy, it appears as a narrow bright line, the "lamina splendens." In the next layer of the superficial zone, flattened ellipsoid chondrocytes are arranged so that their major axes are parallel to the articular surface (see Fig. 1.4). They synthesize a matrix that has a high collagen concentration and a low proteoglycan concentration relative to the other cartilage zones. Water content is the highest in this zone, averaging 80%.[143] In addition, a specific protein, called lubricin (or PRG4) is also only produced in this zone. Lubricin plays an important role in joint lubrication and in allowing frictionless articulation.[145–147]

Transitional zone. The transitional (middle) zone has several times the volume of the superficial zone (see Fig. 1.4). The cells of this zone assume a spheroidal shape and synthesize a matrix with collagen fibrils of a larger diameter and a higher concentration of proteoglycans than is found in the superficial zone. In this zone, the proteoglycan concentration is higher than in the superficial zone, but the water and the collagen concentrations are lower.[148]

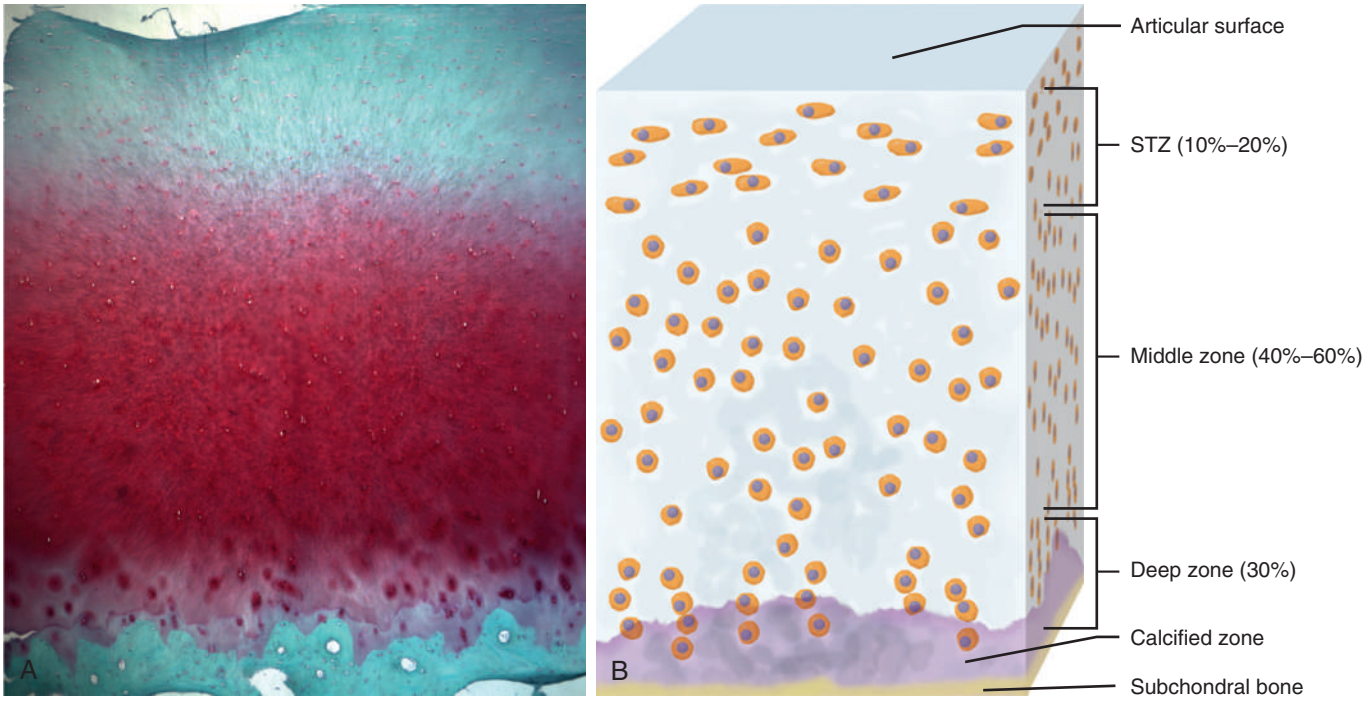

Fig. 1.4 Normal articular cartilage structure. Histologic (A) and schematic (B) views of a section of normal articular cartilage. The tissue consists of four zones: the superficial tangential zone *(STZ),* the middle zone, the deep zone, and the calcified zone. Notice the differences in cell alignment among zones. The cells of the superficial zone have an ellipsoidal shape and lie with their long axes parallel to the articular surface. The cells of the other zones have a more spheroidal shape. In the deep zone, they tend to align themselves in columns perpendicular to the joint surface. (Schematic from Nordin M, Frankel VH. *Basic Biomechanics of the Musculoskeletal System.* 2nd ed. Philadelphia: Lea & Febiger; 1989.)

Deep zone. The chondrocytes in the deep zone resemble those of the middle zone, but they tend to align in columns perpendicular to the joint surface (see Fig. 1.4). This zone contains the collagen fibrils with the largest diameter, the highest concentration of proteoglycans, and the lowest concentration of water. The collagen fibers of this zone pass through the tidemark (a thin basophilic line seen on light microscopic sections that marks the boundary between calcified and uncalcified cartilage)[141] into the calcified zone.[149]

Calcified cartilage zone. A zone of calcified cartilage lies between the deep zone of uncalcified cartilage and the subchondral bone. The calcified layer plays an integral role in securing the cartilage to bone by anchoring the collagen fibrils of the deep zone to the subchondral bone. In this zone, the cell population is scarce and chondrocytes are hypertrophic. Type X collagen is present in the calcified cartilage.

Chondrocytes

The chondrocyte is the predominant cell in cartilage. Chondrocytes contribute only 5% or less to the total volume of cartilage. Like other mesenchymal cells, chondrocytes surround themselves with their extracellular matrix and rarely form cell-to-cell contacts. In normal cartilage, they are isolated. Because the tissue lacks blood vessels, the cells depend on diffusion through the matrix for their nutrition and rely primarily on anaerobic metabolism.

After completion of skeletal growth, chondrocytes rarely divide, but throughout life they synthesize and maintain the extracellular matrix that gives cartilage its essential material properties. Synthesis and turnover of proteoglycans are relatively fast, whereas collagen synthesis and turnover are very slow.[150,151] The limited potential for chondrocyte replication contributes to the limited inherent capacity of cartilage to regenerate or heal after injury.

Extracellular Matrix

Water contributes up to 80% of the wet weight of articular cartilage. The interaction of water with the matrix macromolecules, particularly the large aggregating proteoglycans, significantly influences the material properties of the articular cartilage.[109,143,152,153] This tissue fluid contains gases, small proteins, metabolites, and a high concentration of cations that balance the negatively charged proteoglycans.[152,154] The interaction between proteoglycans and tissue fluid significantly influences the compressive stiffness and resilience of articular cartilage.[139,152]

Collagens contribute approximately 60% of the dry weight of cartilage, proteoglycans contribute 25% to 35%, and the noncollagenous proteins and glycoproteins contribute 15% to 20%. Collagens are distributed relatively uniformly throughout the depth of the cartilage except in the collagen-rich region near the surface. The collagen fibrillar meshwork and cross-linking give cartilage its form and tensile strength.[139,155] Proteoglycans and noncollagenous proteins bind to the collagenous meshwork or become mechanically entrapped within it, and water fills this molecular framework. Proteoglycans give cartilage its stiffness in compression and its resilience. Some noncollagenous proteins organize and stabilize the matrix macromolecular framework, whereas others bind chondrocytes to the macromolecules of the matrix.

Cell-Matrix Interactions

Maintenance of cartilage depends on continual complex interactions between chondrocytes and the matrix they synthesize. Normal degradation of matrix macromolecules, especially proteoglycans, requires that the chondrocytes continually synthesize new molecules.[143,144] If the cells did not replace the lost proteoglycans, the tissue would deteriorate. Mechanical loading also affects cartilage homeostasis.[156,157]

Chondrocytes respond to changes in patterns of matrix deformation due to persistent changes in joint use. Both mechanical and physicochemical events during matrix deformation likely play significant roles. A chondrocyte embedded in the charged extracellular matrix may exist in either an undeformed state or a deformed state. Deformation during compression alters the charge density around the cells and induces a streaming potential throughout the tissue. These physicochemical effects vary according to proteoglycan concentration relative to depth from the surface in the different zones of the charged collagen-proteoglycan extracellular matrix[143,144] and are important in modulating chondrocyte proteoglycan biosynthesis.[152,158–161] In addition to these events, biochemical agents such as growth factors,[162] cytokines, and enzymes are also potent stimulators of chondrocytes. Altogether, studies addressing these important questions offer great challenges for the future.

Relevance for Articular Cartilage Repair

The typical tissue response of vascularized connective tissues to injury follows a cascade of inflammation, repair, and scar remodeling, which is facilitated by cells and other mediators brought in from the surrounding vasculature. However, because hyaline cartilage is avascular, this vital response cannot be generated, and the intrinsic reparative ability of cartilage is very low.[163–165] In healthy cartilage, a homeostasis of extracellular matrix metabolism balances the degradation of macromolecules with their replacement through newly synthesized products. Insult to the cartilage can lead to imbalance of this equilibrium and a shift toward degradation, leading to progression of the chondral defect and potentially, osteoarthritis. Factors associated with this physiologic imbalance and repair response include joint loading, depth of the defect, size of the defect, and patient age.

Joint Loading

Acute or repetitive direct blunt trauma or abnormal loading can cause a spectrum of cartilage injuries ranging from those isolated to microscopic matrix damage to those that lead to visible fissures causing matrix disruption and chondrocyte death. Abnormal loading can be caused by mechanical malalignment or concomitant ligament injury that leads to excessive focal stresses on the cartilage. Loss of proteoglycans typically occurs before other signs of tissue injury[166,167] and may be due to either increased degradation or altered synthesis of the molecules, including the collagen fibrils, thus increasing the vulnerability of the tissue to damage from further impact loading.[166–168] Disruption of the surface collagen matrix leads to increased hydration, fissuring in the cartilage, and thickening of subchondral bone. A study of the response of human articular cartilage to blunt trauma

showed that impact loads exceeding 25 N/m^2 (25 megapascal [MPa]) caused chondrocyte death and cartilage fissures.[149]

Depth of the Defect

Articular cartilage defects are generally classified as chondral or osteochondral, depending on the depth. Chondral defects can be further classified into partial thickness or full thickness (i.e., down to subchondral bone). The repair response depends on whether the injury extends down to the subchondral vascular bone marrow. For partial-thickness chondral injuries, the local response depends entirely on chondrocytes near the injury site, which proliferate and increase the synthesis of matrix macromolecules.[109,169] However, the newly synthesized matrix and the proliferating chondrocytes are unable to fill the tissue defect, and soon after injury, the increased proliferative and synthetic activity ceases. Because chondrocytes cannot repair these matrix injuries, the fissures either remain unchanged or progress. Injuries that extend down into the subchondral bone marrow lead to migration of osteoprogenitor cells into the defect region and synthesis of a new fibrocartilaginous tissue. However, this repair tissue is biomechanically and structurally inferior to hyaline cartilage and thus prone to breakdown with time and loading.[170,171]

Size of the Defect

Smaller defects are less likely to affect the stress distribution on the subchondral bone and progress in size, whereas larger defects are more likely to progress due to increased rim stresses and an inadequate repair response. A study in horses revealed that defects less than 3 mm in diameter may lead to complete repair after 9 months, whereas those larger in size (up to 21 mm in diameter) failed to heal.[172]

Age

Articular cartilage undergoes significant structural, matrix composition, and mechanical changes with age.[104,139,173,174] As with most type of cells in the body, mitotic and synthetic activities of chondrocytes decline with age.[175] These changes are responsible for higher incidence of chondral lesions and development of osteoarthritis in older patients. As a result, any reparative response or ability to maintain extracellular matrix homeostasis decreases with older age. Animal studies in rabbits have demonstrated a better reparative response for chondral defects in younger animals compared with older ones.[176,177]

Clinical Relevance and Further Developments

Small, symptomatic chondral defects may be treated by marrow-stimulating techniques such as microfracture. However, the resultant fibrocartilage repair tissue after microfracture is histologically different and biomechanically inferior to native hyaline cartilage.[178] For larger lesions, a myriad of options is available, including osteochondral autografts or allografts, autologous chondrocyte implantation (ACI), or matrix-induced autologous chondrocyte implantation (MACI). Osteochondral grafts are able to treat defects that extend into the subchondral bone, whereas ACI and MACI require well-preserved bone stock at the base of the chondral defect. The use of particulated juvenile cartilage allograft, which may have increased proliferative and restorative potential, has demonstrated promising early results for treatment of cartilage defects of the knee.[179,180] Future developments include improved scaffolds,[178,181] augmentation with therapeutic factors such as proteins or genes,[182] and the use of MSCs.[182,183] Furthermore, small molecules and activation of endogenous repair (homing of intrinsic progenitors/MSCs) are potential forthcoming therapeutic avenues.[184,185]

BONE

Types of Bone

Normal bone is lamellar and can be classified as cortical or cancellous. Immature bone and pathologic bone are woven and, in contrast to lamellar bone, have more random orientation with more osteocytes, increased turnover, and inferior integrity. Lamellar bone is stress oriented, whereas woven bone is not stress oriented.

Cortical bone (compact bone) (Fig. 1.5) makes up 80% of the skeleton and is composed of tightly packed osteons or haversian systems which are connected by haversian (or Volkmann) canals. These canals contain arterioles, venules, capillaries, nerves, and possibly lymphatic channels. Interstitial lamellae lie between the osteons. Fibrils frequently connect lamellae but do not cross cement lines. Cement lines define the outer border of an osteon and represent the area where bone resorption has stopped and new bone formation has begun. Nutrition occurs through the intraosseous circulation, which involves networks of canals and

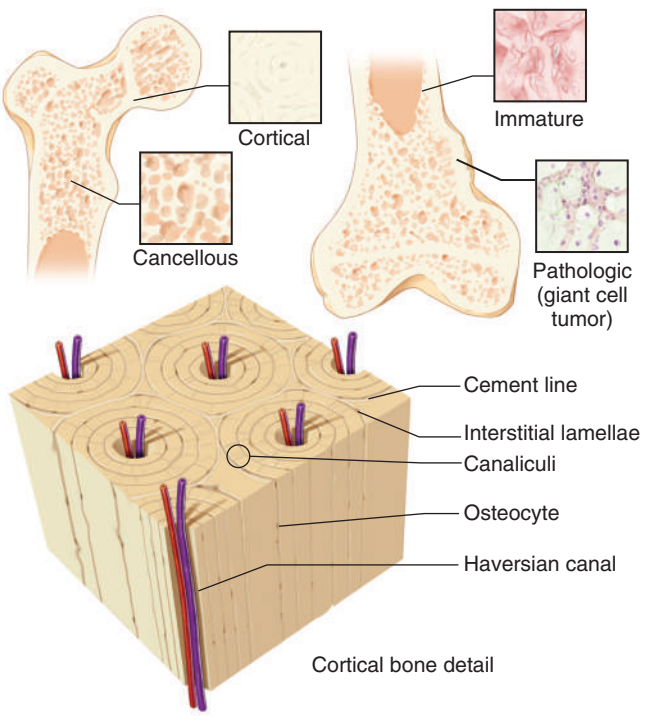

Fig. 1.5 Types of bone. Cortical bone consists of tightly packed osteons. Cancellous bone consists of a meshwork of trabeculae. In immature bone, unmineralized osteoid lines the immature trabeculae. In pathologic bone, atypical osteoblasts and architectural disorganization are seen. (From Brinker MR, Miller MD. *Fundamentals of Orthopaedics*. Philadelphia: WB Saunders; 1999.)

canaliculi. Radiating processes of bone osteocytes, also known as filopodia, project into the canaliculi and allow for osteocyte interaction.

Cortical bone is characterized by a slow turnover rate, a relatively high Young's modulus (E), and a high resistance to torsion and bending. Cancellous bone (spongy or trabecular bone) (see Fig. 1.5) is less dense than cortical bone and undergoes more remodeling according to lines of stress (Wolff's law). Cancellous bone has a higher turnover rate and a smaller Young's modulus and is more elastic than cortical bone.

Cellular Biology

Osteoblasts are responsible for bone formation and are derived from undifferentiated mesenchymal cells. More differentiated, metabolically active cells line bone surfaces, and less active cells in "resting regions" or entrapped cells maintain the ionic milieu of bone. Disruption of the lining cell layer activates these cells.

Osteocytes (see Fig. 1.5) make up 90% of the cells in the mature skeleton and serve to maintain bone. These cells consist of former osteoblasts that have been trapped within newly formed matrix, which they help to preserve. Osteocytes are not as active in matrix production as are osteoblasts.

Osteoclasts are responsible for bone resorption.[186] These multinucleated, irregularly shaped giant cells originate from hematopoietic tissues.[187] Bone resorption occurs in depressions known as Howship lacunae and is more rapid than bone formation; however, bone formation and resorption are linked ("coupled").

Osteoclasts have specific receptors for calcitonin, osteoprotegerin, and other molecules which allow them to directly regulate bone resorption. Interleukin-1 (IL-1) is a potent stimulator of osteoclastic bone resorption in nonphysiologic situations and has been found in the membranes surrounding loose total joint implants. IL-10 suppresses osteoclast formation.[188]

Bone Matrix

Bone matrix is composed of both organic and inorganic components. The organic components make up 40% of the dry weight of bone. Organic components include collagen (mainly type I) proteoglycans, noncollagenous matrix proteins (glycoproteins, phospholipids, and phosphoproteins), growth factors, and cytokines. Collagen is responsible for the tensile strength of bone.

Proteoglycans are partially responsible for the compressive strength of bone. Matrix proteins include osteocalcin,[189] osteonectin, osteopontin, and others. Growth factors and cytokines, which are present in small amounts in bone matrix, include TGF-β; insulin-like growth factor; interleukins (e.g., IL-1 and IL-6); and bone morphogenetic proteins. These proteins aid in bone cell differentiation, activation, growth, and turnover. The inorganic or mineral component of bone matrix makes up 60% of the dry weight of bone. Calcium hydroxyapatite [$Ca_{10}(PO_4)_6(OH)_2$] is responsible for the compressive strength of bone. Calcium hydroxyapatite makes up most of the inorganic matrix and is responsible for matrix mineralization.

Bone Remodeling

Bone remodeling is affected by mechanical stress according to Wolff's law. Removal of external stresses can lead to significant bone loss, but this situation can be reversed to varying degrees on remobilization. In addition to remodeling in response to stress, bone remodels in response to piezoelectric charges. The compression side is electronegative, stimulating osteoblasts and bone formation; the tension side is electropositive, stimulating osteoclasts and bone resorption.

Both cortical bone and cancellous bone are continuously remodeled by osteoclastic and osteoblastic activity. Bone remodeling occurs in small packets of cells known as basic multicellular units. This bone remodeling is modulated by systemic hormones and local cytokines. Bone remodeling occurs throughout life. The Hueter-Volkmann law (i.e., compressive forces inhibit growth and tensile forces stimulate growth) suggests that mechanical factors can influence longitudinal growth, bone remodeling, and fracture repair. Cancellous bone remodels by osteoclastic resorption followed by osteoblastic bone formation.

Bone Circulation

As an organ system, bones receive 5% to 10% of the cardiac output. The long bones receive blood from three sources: the nutrient artery system, the metaphyseal-epiphyseal system, and the periosteal system. Bones with a tenuous blood supply include the scaphoid, the talus, the femoral head, and the odontoid.

The nutrient artery enters the diaphyseal cortex through the nutrient foramen and then enters the medullary canal. In the medullary canal, the nutrient artery branches into ascending and descending small arteries, which, in turn, branch into arterioles which penetrate the endosteal cortex to supply the inner two-thirds of mature diaphyseal cortex through vessels that traverse the haversian system. The metaphyseal-epiphyseal system arises from the periarticular vascular plexus. The periosteal system is composed primarily of capillaries that supply the outer one-third of the mature diaphyseal cortex. Although the nutrient artery system is a high-pressure system, the periosteal system is a low-pressure system.

At the site of bony injury, the initial response is decreased flow to a fracture as a result of disruption of the nutrient artery system at the fracture site.[190] However, within hours to days, bone blood flow increases (as part of the regional acceleratory phenomenon) and peaks at approximately 2 weeks. Blood flow returns to baseline between 3 and 5 months. The arterial system of bone has great potential for vasoconstriction (from the resting state) and much less potential for vasodilation. The vessels within bone possess a variety of vasoactive receptors, which may be useful in the future for pharmacologic treatment of bone diseases related to aberrant circulation (e.g., osteonecrosis and fracture nonunion).[191]

Tissue Surrounding Bone

The periosteum is the connective tissue membrane that covers bone. It is more highly developed in children because of its role in the deposition of cortical bone, which is responsible for growth in bone diameter. The inner, or cambium, layer of periosteum is loose and more vascular and contains cells that are capable of becoming osteoblasts. These cells are responsible for enlarging the diameter of bone during growth[192] and forming periosteal

callus during fracture healing; the outer, fibrous layer is less cellular and is contiguous with joint capsules.

Bone marrow is the soft, gelatinous tissue within the interior of bones, which consists of both stromal and progenitor cells. Red marrow is more commonly found in flat bones, contains hematopoietic stems cells and MSCs, and slowly changes to yellow marrow with age. Yellow marrow is most commonly found in the long bones, contains primarily fat cells, and has a lower water content than red marrow.

Types of Bone Formation

Endochondral Bone Formation and Mineralization

In the process of endochondral bone formation, undifferentiated cells secrete cartilaginous matrix and differentiate into chondrocytes. This matrix mineralizes and is invaded by vascular buds that bring in osteoprogenitor cells. Osteoclasts then resorb calcified cartilage, and osteoblasts form bone. Bone replaces the cartilage model. Examples of endochondral bone formation include embryonic long-bone formation, longitudinal growth (physis), development of fracture callus, and the formation of bone via the use of demineralized bone matrix.

Intramembranous Bone Formation

Intramembranous bone formation occurs without a cartilage model. Undifferentiated mesenchymal cells aggregate into layers (or membranes). These cells differentiate into osteoblasts and deposit organic matrix that mineralizes to form bone. Examples of intramembranous bone formation include embryonic flat-bone formation (e.g., the pelvis, clavicle, and vault of the skull), bone formation during distraction osteogenesis, and blastema bone formation (which occurs in young children with amputations).

Appositional Ossification

In the process of appositional ossification, osteoblasts align themselves on an existing bone surface and lay down new bone. Examples of appositional ossification include periosteal bone enlargement (width) and the bone formation phase of bone remodeling.

Biology of Fracture Healing

Overview

Fracture healing involves a series of cellular events: inflammation, fibrous tissue formation, cartilage formation, and endochondral bone formation.[193] The cellular events of fracture healing are influenced by undifferentiated cells in the vicinity of the fracture, osteoinductive growth factors released into the fracture environment, and the mechanical loading environment.

Fracture Repair

The response of bone to injury can be thought of as a continuum of histologic processes, beginning with inflammation, proceeding through repair (soft callus followed by hard callus), and finally ending in remodeling. Fracture repair is unique in that healing is completed without the formation of a scar. Fracture healing may be influenced by a variety of biologic and mechanical factors.[194–196]

In the inflammation phase, bleeding from the fracture site and surrounding soft tissues creates a hematoma, which provides a source of hematopoietic cells capable of secreting growth factors. Subsequently, fibroblasts, mesenchymal cells, and osteoprogenitor cells accumulate at the fracture site, and fibrovascular tissue forms around the fracture ends. Osteoblasts from surrounding osteogenic precursor cells, fibroblasts, or both proliferate.

In the repair phase, primary callus response occurs within 2 weeks. If the bone ends are not in continuity, a bridging (soft) callus occurs. Fibrocartilage develops and stabilizes the bone ends. The soft callus (fibrocartilage) later is replaced by woven bone (hard callus) through the process of endochondral ossification. Another type of callus—medullary callus—supplements the bridging callus, although it forms more slowly and occurs later in the repair process. The amount and type of callus formation are dependent upon the method of treatment (Table 1.2).[197] Primary cortical healing, which resembles normal remodeling, occurs with rigid immobilization and anatomic (or near-anatomic) reduction with the bone ends in continuity. With rigidly fixed fractures (such as with a compression plate), direct osteonal or primary bone healing occurs without visible callus formation. In contrast, "endochondral healing," with periosteal bridging callus formation, occurs in the setting of nonrigid fixation.

The remodeling phase begins during the middle of the repair phase and continues long after the fracture has clinically healed (can be for several years). Remodeling allows the bone to assume its normal configuration and shape based on the stresses to which it is exposed (Wolff's law). Throughout the process, woven bone formed during the repair phase is replaced with lamellar bone. Fracture healing is complete when repopulation of the marrow space occurs.

TABLE 1.2 Type of Fracture Healing Based on Type of Stabilization

Type of Immobilization	Predominant Type of Healing	Comments
Cast (closed treatment)	Periosteal bridging callus	Enchondral ossification
Compression plate	Primary cortical healing (remodeling)	Cutting cone–type remodeling
Intramedullary nail	Early: periosteal bridging callus Late: medullary callus Dependent on extent of rigidity Less rigid: periosteal bridging callus More rigid: primary cortical healing	Enchondral ossification
Inadequate	Hypertrophic nonunion	Failed endochondral ossification; type II collagen predominates

From Brinker MR. Basic sciences. In: Miller MD, Brinker MR, eds. *Review of Orthopaedics*. 3rd ed. Philadelphia: WB Saunders; 2000.

TABLE 1.3 Factors Affecting Bone and Bone Healing

Factor	Example	Action	Clinical Comment
Growth factors	BMPs	Osteoinductive, induce metaplasia of mesenchymal cells into osteoblasts	Used clinically to stimulate bony healing in challenging injuries such as open tibial fractures[42] and nonunions
	TGF-β	Induces osteoblasts to synthesize collagen	
	IGF-I and IGF-II and IGFBPs 1–6	Regulate proliferation and differentiation of bone-forming osteoblasts	
	Platelet-derived growth factor	Attracts inflammatory cells to the fracture site	
Ultrasound	Low-intensity pulsed ultrasound	Accelerates fracture healing and increases the mechanical strength of callus, including torque and stiffness	Decreases consolidation time during distraction osteogenesis and stimulates healing of nonunions
Electricity	Direct current	Stimulates an inflammatory-like response	
	Alternating current	Affects collagen synthesis and calcification	
	Pulsed electromagnetic fields	Initiates calcification of fibrocartilage	Does not induce calcification of fibrous tissue

BMP, Bone morphogenetic protein; *IGF,* insulin-like growth factor; *IGFBP,* insulin-like growth factor–binding protein; *TGF-β,* transforming growth factor-β.

The different factors affecting bone are summarized in Table 1.3.

For a complete list of references, go to ExpertConsult.com.

SELECTED READINGS

Citation:
O'Keefe R, Jacobs JJ, Chu CR, et al, eds. *Orthopaedic Basic Science*. 4th ed. American Academy of Orthopaedics; 2013.

Level of Evidence:
V

Summary:
An excellent textbook covering all aspects of the basic science of musculoskeletal tissues.

Citation:
Andarawis-Puri N, Flatow EL, Soslowsky LJ. Tendon basic science: Development, repair, regeneration, and healing. *J Orthop Res.* 2015;33(6):780–784.

Level of Evidence:
V

Summary:
A review focusing on the challenges and important questions regarding tendinopathies and tendon healing.

Citation:
Proffen BL, Perrone GS, Roberts G, et al. Bridge-enhanced ACL repair: A review of the science and the pathway through FDA investigational device approval. *Ann Biomed Eng.* 2015;43(3):805–818.

Level of Evidence:
V

Summary:
A review of the approaches for bioenhanced ACL repair and the group's journey to carry their novel technology from bench to bedside.

Citation:
Eleftherios MA, Gomoll AH, Malizos KN, et al. Repair and tissue engineering techniques for articular cartilage. *Nat Rev Rheumatol.* 2015;11(1):21–34.

Level of Evidence:
V

Summary:
A comprehensive review of the biology of cartilage repair and the science supporting current reconstructive surgical techniques and future tissue engineering endeavors.

Citation:
Erggelet C, Mandelbaum B. *Principles of Cartilage Repair*. Heidelberg: Springer-Verlag; 2008.

Level of Evidence:
V

Summary:
Clinical overview of current techniques for cartilage repair with detailed descriptions and graphics.

Citation:
Einhorn TA, Gerstenfeld LC. Fracture healing: mechanisms and interventions. *Nat Rev Rheumatol.* 2015;11(1):45–54.

Level of Evidence:
V

Summary:
A comprehensive review of the biology of fracture healing at the tissue, cellular, and molecular levels.

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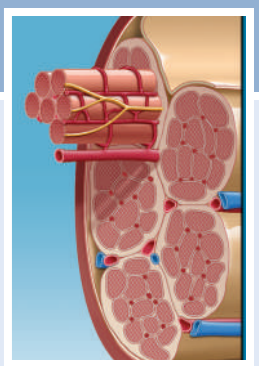

Basic Concepts in Biomechanics

Richard E. Debski, Neel K. Patel, Jason T. Shearn

Biomechanics is an interdisciplinary field that uses the principles of mechanics to improve the human body through design, development, and analysis of equipment, systems, and therapies. This biomechanical knowledge can help in understanding the loading of the musculoskeletal system and its mechanical responses, which can be used to determine normal function, predict changes, and propose interventions. More specifically, basic biomechanics explores forces and moments required for movement and balance of the human body by muscle recruitment and the consequence of internal forces on loading of soft tissues. This chapter explores biomechanics in terms of four topics: statics, dynamics, mechanics of materials, and applications. Additionally, examples of these concepts are provided throughout the chapter to illustrate how they can be directly applied to sports medicine.

BASIC CONCEPTS

Units of Measure

Understanding the basic units of measure for biomechanics is important in describing the dimensional or spatial analyses. There are three primary dimensions—length, time, and mass—from which secondary dimensions are subsequently derived (Table 2.1). Metric unit measurements are commonly used in the field of biomechanics to describe such dimensions and are also used in this chapter. One measurement not described in Table 2.1 is that used for angular descriptions. The unit for angles is typically defined in terms of radians or degrees.

Scalars and Vectors

Most of the physical quantities encountered in mechanics are either scalars or vectors. Scalar quantities describe only the magnitude of an element and are used for concepts such as time, length, speed, mass, temperature, volume, work, power, and energy. In contrast, both a magnitude and a direction are associated with vector quantities. Examples of vector quantities are displacement, force, moment, velocity, and acceleration. The magnitude of an object's velocity is identical to its speed, but the velocity vector also contains information about the direction of motion. Graphically, a vector is represented as an arrow, with the orientation of the arrow indicating its line of action or direction. For example, the force of gravity is a vector quantity that can be represented by a downward arrow at an object's center of mass. Typically vectors are represented in a Cartesian coordinate system.

Coordinate Systems

A coordinate system is a frame of reference for a structure or system. Coordinate systems are spatial tools that are useful in describing body position and orientation and the directionality of all vector quantities. To apply mechanical principles to a system, all vector quantities must be expressed with respect to the same coordinate system. Several types of coordinate systems are available, but the most commonly used in sports medicine research is the Cartesian coordinate system. This system can be either two- or three-dimensional. All axes are perpendicular to one another (a condition known as orthogonality) and have both positive and negative components. A vector quantity contains a component in each dimension corresponding to the magnitude associated with each axis. In a two-dimensional system, the horizontal and vertical axes are typically labeled "x" and "y," respectively. In a three-dimensional system, the "z"-axis is added. This axis lies perpendicular to both the x- and y-axes, with its positive side determined by the right-hand rule. This rule states that if you align the index finger of your right hand with the positive x-axis and the middle finger with the positive y-axis, your thumb will be pointing in the positive direction of the z-axis. Similarly, the right-hand rule can be used to determine the positive and negative directions of angles and moments about a specified axis. While the thumb of your right hand is aligned with the rotational axis, your fingers will curl in the positive direction of rotation. These rules are very important in establishing the correct spatial quantities within a designated coordinate system.

Anatomic Coordinate System

An anatomic coordinate system is a Cartesian coordinate system with each axis representing anatomic directions such as superior-inferior, medial-lateral, and anterior-posterior. The corresponding rotations about these axes are represented by internal-external rotation, flexion-extension, and abduction-adduction, respectively. Researchers and clinicians use the anatomic coordinate system whenever possible because it is the most clinically relevant for diagnosis and treatment.

Degrees of Freedom

Degrees of freedom are the total number of independent movements needed to completely describe the position and orientation of a body in a given coordinate system. The musculoskeletal system has numerous degrees of freedom through which countless

Abstract

Biomechanics is an interdisciplinary field that uses the principles of mechanics to improve the human body through design, development, and analysis of equipment, systems, and therapies. This biomechanical knowledge can help in understanding the loading of the musculoskeletal system and its mechanical responses, which can be used to determine normal function, predict changes, and propose interventions. More specifically, basic biomechanics explores the forces and moments required for movement and balance of the human body by muscle recruitment and the consequence of internal forces on loading of soft tissues. This chapter explores biomechanics in terms of four topics: statics, dynamics, mechanics of materials, and applications. Additionally, examples of these concepts are provided throughout the chapter to illustrate how they can be directly applied to sports medicine.

Keywords

biomechanics
forces motion
material properties
viscoelasticity

TABLE 2.1 Dimensional Quantities and Corresponding Units in the International System of Units

Dimensions	SI Unit
Primary	
Length	Meter (m)
Time	Second (s)
Mass	Kilogram (kg)
Secondary	
Area	m^2
Volume	m^3
Velocity	m/s
Acceleration	m/s^2
Force	Newton (N) = kg • m/s^2
Pressure/stress	Pascal (Pa) = N/m^2
Moment/torque	N • m
Work/energy	Joule (J) = N • m
Power	J/s

SI, International System of Units.

movements are accomplished. These movements can be performed in three-dimensional space through a set of translations and rotations and are typically described in an anatomic coordinate system. For example, the glenohumeral joint has a total of six degrees of freedom: three translations (superior-inferior, medial-lateral, and anterior-posterior), and three rotations (internal-external, abduction-adduction, and flexion-extension). However, depending on the analysis, the glenohumeral joint can be assumed to be a ball-and-socket joint, thus reducing the number of degrees of freedom to three rotations by constraining the translations. Similarly, the elbow only allows for one degree of freedom when it is considered a simple hinge joint.

Degrees of freedom are important to consider when describing body motion and position numerically. Joint motion may be constrained by the shape of the articular surfaces or by stabilizing ligaments, reducing the degrees of freedom and simplifying the description. However, some joints, such as the knee and ankle, also allow for small amounts of translation that are important to consider when interpreting a clinical examination. With an anterior cruciate ligament (ACL) injury, the knee will have an increased level of anterior translation, thereby altering the anterior degree of freedom. Using reduced degrees of freedom is strategic when modeling or characterizing joints of the body. For example, when a person picks up an object from the floor and subsequently returns to an erect posture, each vertebra moves in three-dimensional space to achieve the desired posture, representing a large number of degrees of freedom. To evaluate lower back injuries using modeling techniques, the degrees of freedom can be significantly reduced by only considering the L5–S1 joint and assuming that the motion occurs only in the sagittal plane, resulting in a two-dimensional analysis. Similarly, in assessing the tibiofemoral joint during a kicking motion, it could be assumed that the tibia moves in only two dimensions within the sagittal plane (Fig. 2.1A). These reductive assumptions often overestimate the translations, rotations, forces, and moments required to perform the motion.

TABLE 2.2 Newton's Laws of Motion Applied to Static and Dynamic Analyses

Law	Definition	Equation
First (inertia)	An object at rest tends to stay at rest and an object in motion tends to stay in motion with the same speed and in the same direction unless acted upon by an external unbalanced force.	$\Sigma F = 0$ $\Sigma M = 0$
Second (acceleration)	The rate of change of the moment of a body is directly proportional to the applied force and takes place in the direction in which the force acts.	$F_{NET} = m \cdot a$ $M_{NET} = I \cdot \alpha$
Third (action-reaction)	For every action there is an equal but opposite reaction.	

Newton's Laws

Three basic rules of physics—Newton's laws of motion—are used to describe the relationship between the forces applied to the body and the consequences of those forces on human motion (Table 2.2).

Newton's First Law (Inertia)

Newton's first law states that an object at rest tends to stay at rest and an object in motion tends to stay in motion with the same speed and in the same direction (velocity) unless acted upon by an unbalanced force. This law is also commonly referred to as the *law of inertia*. Therefore a nonzero resultant force must act on a rigid body to change its velocity. For example, headrests are placed in cars to prevent whiplash injuries during rear-end collisions by stopping the motion of the head, which illustrates decreasing linear velocity and acceleration by application of an unbalanced force. Furthermore, if the resultant moments and forces are zero, then the body will also have no rotational or linear acceleration.

Newton's Second Law (Acceleration)

Newton's second law pertains to the behavior of objects for which all forces are not balanced. Therefore the resultant force is not equal to zero, and acceleration in the direction of the applied force will occur. The magnitude of the acceleration is proportional to the magnitude of the resultant forces or moments applied to the body. Thus, in essence, the first law is a special case of the second law:

$$\Sigma F = m \cdot a : \Sigma M = I \cdot \alpha$$

where "F" is the resultant force, "m" is the mass of the body being acted upon, "a" is the acceleration of the body due to the unbalanced external forces, "M" is the resultant moment, "I" is the mass moment of inertia (resistance of a body to rotation), and α is the angular acceleration.

Linear acceleration can exist as an extremity is sent through a range of motion (ROM) during a task (see Fig. 2.1F). In sports, an athlete frequently controls his or her mass moment of inertia or center of mass of the entire body by altering the positioning

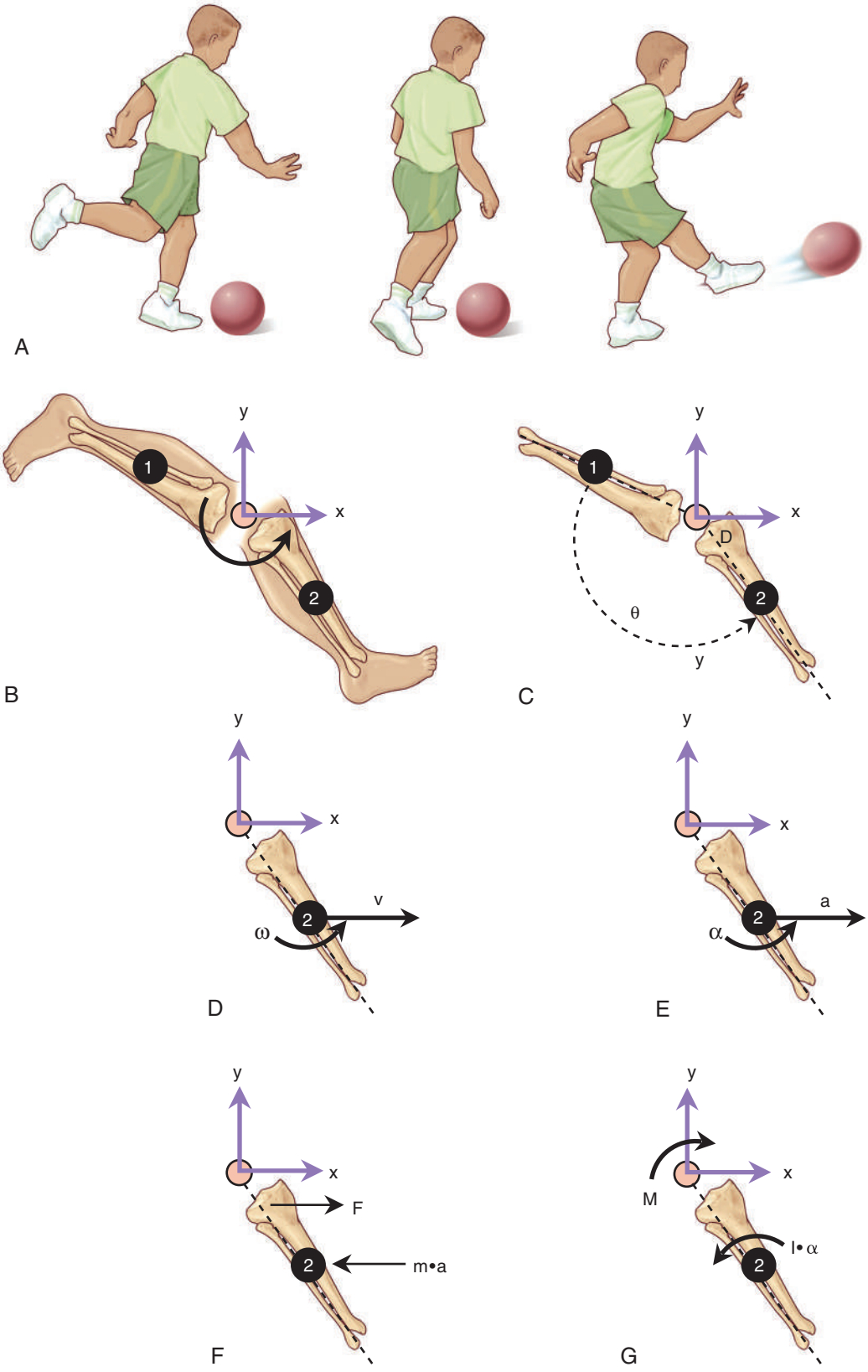

Fig. 2.1 (A) Sequence of leg positions while kicking a ball. (B) The motion of the leg in the acceleration phase of kicking in the sagittal plane at positions 1 and 2. (C) Displacement *(D)* and rotation *(θ)* of the tibia during the motion. (D) The linear *(v)* and angular *(ω)* velocity vectors. (E) The linear *(a)* and angular *(α)* acceleration vectors in the acceleration phase of kicking. (F) The linear and angular inertial forces, with the applied force *(F)* resisted by the inertial force *(m • a)*. (G) The applied moment *(M)* resisted by the angular inertia *(I • α)*.

of the individual body segments to achieve stability or a particular motion. Gymnasts and divers use this concept to achieve multiple somersaults while in the air by tucking in the head and limbs closer to the center of the body, which is an example of angular acceleration due to changes in angular inertia (see Fig. 2.1G).

Newton's Third Law (Action-Reaction)

Newton's third law states that for every action there is an equal but opposite reaction. This law explains the idea that if a person pushes against a wall, it will in essence push back. Forces of action and reaction are equal in magnitude, but in the opposite direction. This concept is important in examining the principle of equilibrium and when using tools that assess forces being applied to the body of interest. Reaction forces act to constrain motions by reacting to an applied force. A daily application of this concept is simple walking and running. Every time a person places a foot on the ground, a force is exerted from the foot on the ground throughout the gait cycle. Simultaneously, however, a force of equal magnitude is exerted in the opposite direction from the ground up to the foot, which is termed a *ground reaction force*. Ground reaction forces are the forces applied by the ground because of the weight of the body. Similarly, within joints, muscles and connective tissues are found that create a *joint reaction force* (Fig. 2.2). Both passive and active stabilizers maintain the humeral head within the glenoid fossa of the glenohumeral joint,[1] which has implications for surgical repair of Bankart lesions by changing the arc of the glenoid and tendon transfers in changing the line of muscle action. If these joint structure changes alter the natural forces and/or the direction of these forces, joint instability would result, potentially leading to long-term degenerative changes.

When a body is balanced by equal but opposite reactions for every action, the system is considered to be in a condition of equilibrium, which means that the net effect of the applied forces is zero as well as the net moment about any point. Although the sum of the force and moment vectors is zero, the body may still be moving at a constant velocity. Assuming a state of static equilibrium with a set of known force vectors, it is possible to determine unknown forces within a system. To do so, each force vector can be resolved into its individual components, such as the vertical (Fy) and horizontal (Fx) components for a two-dimensional coordinate system (Figs. 2.3 and 2.4D). Therefore the resulting equations for force and moment equilibrium in three dimensions are:

$$\Sigma Fx = 0; \Sigma Mx = 0$$

$$\Sigma Fy = 0; \Sigma My = 0$$

$$\Sigma Fz = 0; \Sigma Mz = 0$$

For static equilibrium, no linear or rotational motion occurs.

STATICS

Using the aforementioned equations of static equilibrium, static analyses evaluate the external effects of forces on a rigid body at rest or during motion with a constant velocity. Applied to the body, static analyses are used to further determine the magnitude and direction of forces at joints and in the muscles. Forces provide both mobility and stability to the body but also introduce the potential to deform and injure the body. Typically, healthy tissues are able to withstand changes in their shape, but a tissue structure that has been injured by disease or trauma may not be able to adequately sustain the same loads required to perform activities of daily living. To perform static analyses, we must be able to

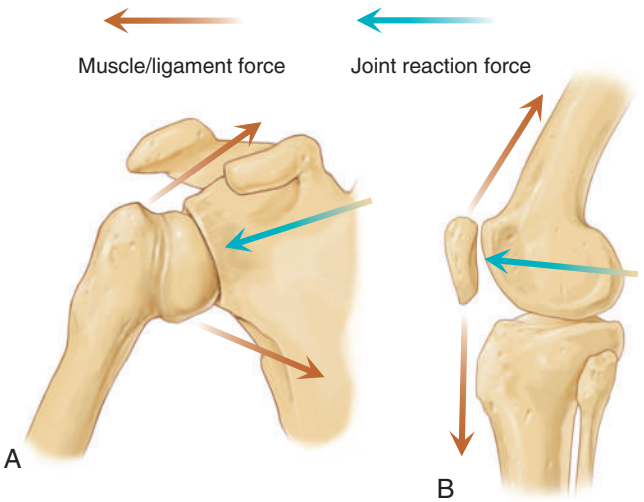

Fig. 2.2 Resulting joint reaction force at the glenohumeral (A) and patellofemoral (B) joints.

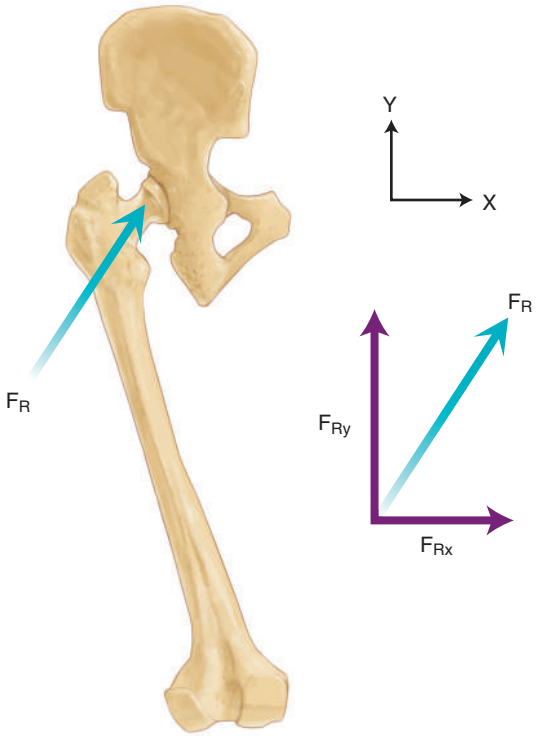

Fig. 2.3 Force vector (F_R) representing the joint reaction force between the femoral head and acetabulum, and F_R resolved into its individual component forces (F_{Rx} and F_{Ry}) in the *x* and *y* directions.

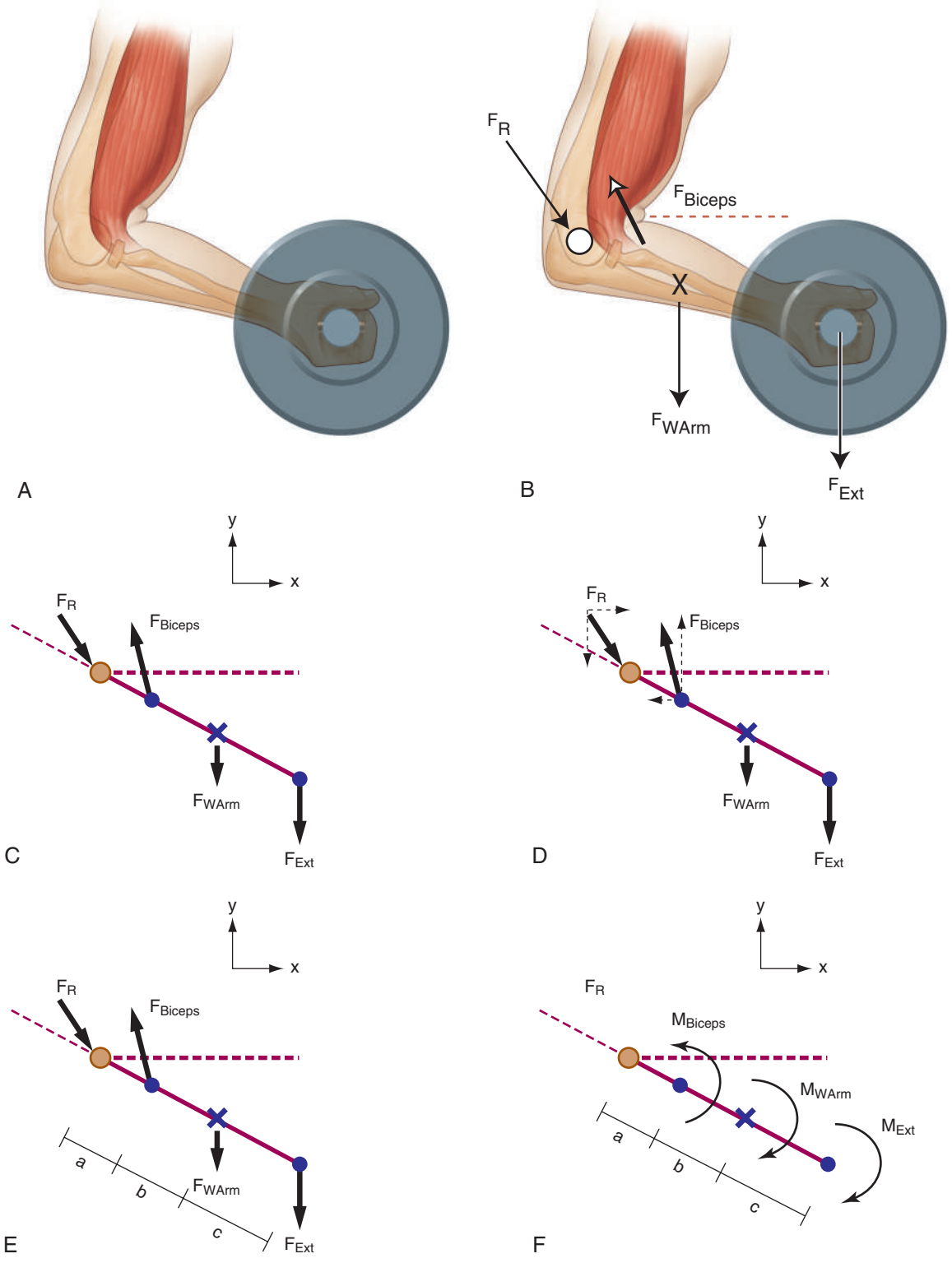

Fig. 2.4 (A) Simulation of a person performing a biceps curl. (B) The dumbbell applies an external force *(F_{Ext})* downward in addition to the downward force due to the weight of the arm *(F_{WArm})*. The biceps muscle generates a force *(F_{Biceps})* to the forearm and causes a joint reaction force *(F_R)* at the elbow to keep the joint stabilized. (C) A free-body diagram of the forearm representing each force as an *arrow*, with the head of the *arrow* pointing in the direction of the applied force. (D) These forces are then decomposed into component vectors. (E) The varying moment arms for each force vector are identified with respect to the axis of rotation at the elbow. (F) The biceps muscle creates a counterclockwise moment *(M_{Biceps})* to resist the clockwise moments due to the weight of the arm *(M_{WArm})* and dumbbell *(M_{Ext})*.

represent the complex interactions of forces and moments acting on a body through the use of vectors and free-body diagrams.

Force Vectors

The most common vector quantity in mechanical systems is a force. A force is applied on an object to create either a pushing or pulling response. Depending on the original state of the object, a force can cause a stationary object to move or alter the state of an object already in motion. Such forces can be either internal or external. Internal forces include those that hold a rigid body together, such as muscle tension generated within an extremity, whereas external forces are those applied to a rigid body. An example of an external force is the weight of an object being held in a person's hand. Forces can be further categorized into two subgroups: contact (tension, friction, external, and internal) and distance (gravitational and magnetic) forces. Distance forces act at a distance from the object or body with no direct physical contact. Additionally, forces can contain components acting in both the normal (perpendicular) and tangential (parallel) directions of the surface to which they are being applied.

Moment/Torque Vectors

Vector quantities can represent not only translational motions in response to an applied force but also the rotation, twisting, and bending of an object. A moment (or torque) is determined by the magnitude of the force acting about a point and the length of the shortest distance between the point and line of action of the force. This distance is known as the *moment* or *lever arm*. A larger moment arm requires less force to achieve equivalent angular motion about the axis of rotation, which means that if the force remains the same but the lever arm increases, the moment will be of greater magnitude. Although a moment can realistically be calculated about any point, typically it is calculated about a joint axis of rotation during biomechanical analyses. For example, when tension is generated in the biceps muscle during a biceps curl exercise, the tendon pulls on the forearm at a distance from the elbow axis of rotation (see Fig. 2.4). This tension from the biceps muscle generates a positive moment about the elbow that can either help maintain the posture of the elbow or increase the amount of flexion, depending on whether the biceps muscle force is equal to or greater than the applied external forces of the weight.

Free-Body Diagrams

To better evaluate a biomechanical system, such as forces being applied to a specified part of the body, free-body diagrams are an effective tool to simplify a complex analysis. Free-body diagrams allow for visualization and ease of calculation by properly identifying all the forces and moments acting on the body of interest in order to successfully achieve equilibrium. These diagrams are used by first drawing the body of interest and then isolating the body from its environment, only including the forces acting on the body. Again using the example of a biceps curl exercise to evaluate the elbow joint, the system should be drawn as only the forearm (with the radius and ulna combined as one), because this segment is the body segment of interest, with the elbow as the axis of rotation (see Fig. 2.4B). It is important to note the position and orientation of the object by defining a coordinate system.

Applied forces are then identified and considered. Arrows representing the force vectors are drawn at every point where two (or more) bodies interact or join (see Fig. 2.4C). Recalling that internal forces are those that hold a rigid body together, two examples within the human body are muscle contractions and bony contact at a joint line. Internal forces are further examined with the concept of stress, which is introduced in a later section of the chapter. External forces can be the weight of an object, friction, or gravity applied at the center of mass of a body segment. In a simplified situation, flexion of the elbow is counteracted by the weight of the forearm itself and the weight of the dumbbell. The point of application of the forearm weight is at the segment's center of gravity. Commonly this point is determined from anthropometric data. The point of application of the biceps force on the forearm is its tendon insertion. The remaining force is the result of contact with the distal end of the humerus and the proximal end of the forearm, whereby the point of contact is approximately at the joint surface of the bones. These forces can then be further resolved into their respective components along the previously identified x- and y-axes (see Fig. 2.4D).

In addition to forces, the forearm experiences various applied moments. Because the points of application for the forces are known, the distances of each force along the forearm from the axis of rotation—the elbow—is also known (see Fig. 2.4E). Using these distances as moment arm values, the next step is to identify moments created by each force. The direction of the moment is relative to the applied force and its relationship to the axis of rotation. No moments are generated by the forces at the elbow because their moment arms are zero (see Fig. 2.4F). Once all the forces and moments applied to the body or body segment of interest are identified, they can be summed to determine the resultant force and moment. When the body is in a state of static equilibrium, the resultant forces and moments sum to zero, according to Newton's third law of motion.

Ligament and Joint Contact Forces

These concepts of static analyses can be applied not only to whole-body analyses but also at the joint and tissue level. For the typical joint, forces can be related to compression and shear. For example, when a person is standing with the knee in full extension, the tibial plateau and femoral condyles experience compressive forces in the normal direction to each articular surface. However, an increased shear force is experienced during an anterior drawer test or when stopping quickly in a sporting event. Shear forces are experienced in the tangential direction along the tibial plateau. These forces are not only experienced between the bony structures but also transmitted through the soft tissue structures. During an anterior drawer test or quick stop, the ACL can become significantly loaded to resist anterior tibial translation and provide stability at the joint. The ACL becomes loaded as a result of a generated tensile force within the ligament to maintain equilibrium. However, after an injury such as an ACL rupture, opposing shear forces cannot be transferred through the ligament, and thus equilibrium cannot be

maintained at the joint without excessive translation, resulting in anterior instability.

DYNAMICS

Based on the equations from Newton's second law of motion, dynamic analyses evaluate bodies in motion and can be divided into two subgroups: kinematics and kinetics. For dynamic systems, the forces and moments do not have a net value of zero, which violates the principle of equilibrium, and thus a different approach must be taken. Kinematics simply describes the motion of bodies, without regard to the factors that cause or affect the motion, by characterizing the geometric and time-dependent aspects of the motion. Conversely, kinetics is based on kinematics but includes the effects of forces and moments. Motion analyses and sports mechanics typically involve dynamic systems.

Kinematics

Simply stated, kinematics deals with motions without regard to forces and moments. These motions include translations and rotations. Translations are simply the linear motions in which all the parts of a rigid body move simultaneously in the same direction as every other point in that body and at the same velocity. Rotations are the angular motions of a rigid body along a circular path and about an axis of rotation. During passive knee flexion, the tibiofemoral joint undergoes both linear and angular motions. As the knee progresses through the ROM, the tibia experiences rotation in the sagittal plane, and simultaneously, the point of contact between the tibial and femoral articular surfaces translates in the posterior direction. Although flexion is the primary angular motion, internal tibial rotation also occurs, which exemplifies the complex motion that can occur at a single joint.

Linear and Angular Kinematics

The spatial components used to describe linear kinematics are position, distance, and displacement. Position is a vector that simply defines the location of the object in space relative to a reference frame, commonly the center of a joint or a point of contact. Once the initial position changes, the object has moved a specified distance and displacement. Distance is a scalar quantity that describes the length of the entire path traversed. Displacement is a vector quantity (d) defining the shortest distance (straight line) between the starting and ending positions of the object independent of the path taken (see Fig. 2.1C). Additionally, temporal descriptions of linear kinematics are speed, velocity, and acceleration. Velocity is a vector quantity (v) describing the rate of change of the position with respect to time, whereas speed is a scalar quantity equal to the magnitude of the velocity vector (see Fig. 2.1D). Furthermore, acceleration is also a vector quantity (a) that describes the rate of change of velocity with respect to time, but it is commonly used to express an increase or decrease in speed (see Fig. 2.1E).

Angular kinematics is experienced when a change in angular position occurs such that the body undergoes rotational motion about an axis of rotation. The angular distance through which the body moves is equal to the length of the angular path. Angular kinematics can describe rotatory motion for a body segment, such as the lower leg during a kick (see Fig. 2.1A), in addition to the body as a whole, such as for a gymnast swinging on the bar to perform a giant circle. Descriptors used for these motions with respect to time are *angular position, displacement, distance, velocity,* and *acceleration*. These terms are similar to those used for linear kinematics, but they apply to a rotatory motion (see Fig. 2.1C–E). The lower leg undergoes large changes in angular velocity (ω) and acceleration (α) throughout the ROM for activities such as running, swimming, or kicking a soccer ball. Most general movements in the body are a combination of both linear and angular motion. For example, during a normal gait cycle, the lower extremities translate and rotate. Similarly, during athletics, combined motions are clearly illustrated during pitching, as the glenohumeral joint creates an instantaneous axis of rotation when the ball is swung along an imaginary arc about the shoulder in addition to being propelled forward.

Relative Motion

When considering forces and moments applied to a body, it is important to know the relative relationship in terms of both position and orientation for the individual bodies to each other in addition to the overall environment. An illustration of this concept is that a moving object may appear to have a different motion for two different observers depending on their location to the moving object. Knowing relative relations allows for an accurate description of the system when solving for unknown forces by establishing a frame of reference, or coordinate system. If a frame of reference is not identified, the measurements become irrelevant. A frame of reference describes the position of one body with respect to another whereby a relative measurement can be performed. This measurement is made by comparing the change in position and orientation of the object to the reference frame. For example, the position and orientation of the femur can be defined relative to the pelvis when describing motions of the hip or of the tibia with respect to the femur during kicking (see Fig. 2.1B).

Joint Motions

Motion at an articular surface can be described in terms of three motions that exist as a result of convex surfaces moving on a concave surface (Fig. 2.5). A *rolling motion* occurs when the bone with the convex surface rotates to cause a change in the point of contact for both articular surfaces in addition to a corresponding linear motion. A ball experiences this type of motion when it is rolled across a smooth surface. A *sliding motion* is experienced when one articular surface translates across the other with no rotation and progressively changes the point of contact. Sliding occurs when a box is pushed across a smooth surface. A *spinning motion* occurs when there is a single point of contact on the fixed surface and the point of contact changes on the rotating surface that does not undergo any linear motion. Spinning can be experienced on an automobile tire when the tire is rotating but the automobile is not moving forward or backward because of ice on the road. For certain joints, a combination of two or all three of these motions occurs. The motion achieved at a joint is based on the shape of the articular surface.

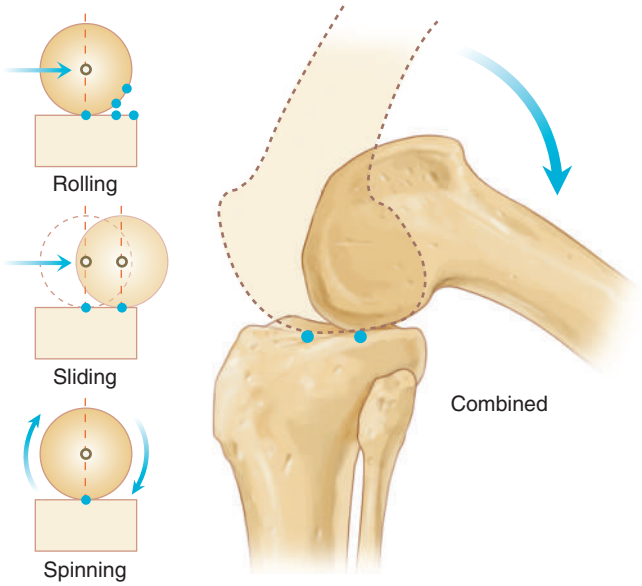

Fig. 2.5 Three fundamental motions that occur between articular surfaces. The point of contact changes on both articular surfaces during a rolling motion. The point of contact on the moving surface remains constant during a sliding motion. A single point of contact occurs on the fixed surface during a spinning motion. Some joints, such as the tibiofemoral joint, experience up to all three of these motions simultaneously.

For example, at the tibiofemoral joint during flexion, the knee experiences femoral condylar rollback on the tibial plateau. This phenomenon encompasses all three motions (rolling, sliding, and spinning) through flexion rotation, posterior translation, and external rotation, respectively, of the femur relative to the tibia. Although it is important and beneficial to understand the translations and rotations occurring at a joint, it can also be beneficial to understand the forces and moments associated with the observed translations and rotations, thus introducing the necessity of kinetic analyses.

Kinetics

Kinetics is the branch of mechanics that describes the effect of forces and moments on the body by utilizing Newton's second law of motion. More specifically, by using kinetics, it is possible to determine the forces and moments on a joint produced by mass, muscle tension, soft tissue loading, and externally applied loads. The appropriate range of joint loading that can be considered noninjurious can also be determined using kinetic analyses. Conversely, situations can be identified that produce excessively high moments and forces that exceed these limits, thus leading to musculoskeletal injuries. If an excessive valgus moment is experienced at the knee because of contact during a sport such as football, the medial collateral ligament may become overloaded and rupture in an attempt to maintain stability and resist the applied load. High forces can also be applied internally by overloading tendons as a result of extreme muscle tension.

CLASSES OF MECHANICS

Mechanics is the study of forces and their effects on motion. Depending on the system being studied, different branches of mechanics may apply. When examining the behavior of solid bodies, such as the forces acting on human limbs and their motions, rigid body mechanics may apply. When examining the effects of a body moving through air or water, fluid mechanics are applicable.

Rigid Body Mechanics

In rigid body mechanics, it is assumed that any deformation caused by forces acting on a body is negligible. By assuming that a body is nondeformable—that the distance between any two points in the body will remain constant for any given translation or rotation—any changes in the shape of the body due to an applied force can be ignored, making kinematic analyses much simpler. No material in the human body can truly be considered a rigid body because all tissues undergo some degree of deformation, and thus it is important to note when the rigid body assumption is applicable. If the deformations experienced by a body are much smaller than the translations or rotations of that body, then the rigid-body principle can be applied. For example, when analyzing gait, the translations and rotations of the lower extremity will be much greater than any deformations experienced by the segments of the lower extremity, allowing each segment to be treated as a rigid body. Similarly, when using training equipment such as during weight lifting (see Fig. 2.4), the bar and limb segments of the upper extremity can be considered rigid bodies to allow for kinematic and kinetic analyses because any translations or rotations are much greater than any deformation in the equipment or limbs. Additionally, if one material is much stiffer than another, the stiffer material can be considered to be a rigid body that undergoes no deformation. For example, when performing mechanical testing of a joint complex, such as the femur–medial collateral ligament–tibia complex, the bones can be considered rigid bodies because they are much stiffer than the ligament tissue.

Fluid Mechanics

Fluid mechanics takes the concepts of statics and dynamics for solid bodies and applies them to fluids in order to study their behavior under applied forces and displacements. The term *fluid* refers to both gases and liquids and can be defined as a material that flows freely as the result of the application of a force and cannot support shear stresses. The most important parameter describing the behavior of a fluid is its viscosity (μ), a measure of how a fluid resists flow because of an applied force. Certain fluids, such as blood, have higher viscosities, whereas other fluids, such as air, have very low viscosities that can be considered to have negligible resistance. The viscosity of blood depends on its cellular content; when hematocrit levels rise above normal, blood viscosity rises and flow becomes more resistant to the forces pumping it through the body, leading to higher blood pressure as the heart works harder to pump blood through the body. During respiration, viscosity contributes to airflow patterns through the trachea and bronchi, relating to resistance of airflow. Although the viscosity of the air does not change, changes in the airway diameters due to disease can increase resistance to airflow, leading to respiratory pathology.

Additionally, movement of a solid object through fluid can result in the phenomenon of "drag," or fluid resistance, which reduces the velocity of the body. Drag is dependent on both the surface area exposed to the fluid in the direction of movement and on the velocity of the object: the greater the velocity and surface area, the greater the fluid resistance. At the whole-body level, water resists the movement of the body while swimming, and air resists the movement of sprinters traveling at high speeds. To reduce the effects of drag, athletes position their bodies during running or sprinting to reduce the surface area of their bodies perpendicular to the direction of motion.

Fluids can be considered to be either Newtonian or non-Newtonian. Newtonian fluids, such as water or air, exhibit a linear relationship between force applied to the fluid and the fluid deformation rate. Non-Newtonian fluids do not exhibit this relationship, an example being blood, as a result of its composition of particles (cells) suspended in aqueous plasma. The importance of Newtonian versus non-Newtonian flow relates to being able to accurately predict the behavior of fluids in the body, particularly blood. Depending on the size of the vessel, Newtonian blood flow may be assumed in some cases even if the fluid is non-Newtonian. The accurate prediction of blood flow patterns is important for understanding how abnormal blood flow can lead to atherosclerosis or low blood perfusion in tissues.

Viscoelasticity

A tissue is viscoelastic when it possesses to some degree both solid-like characteristics, such as elasticity and strength, and liquid-like characteristics, such as flow depending on temperature, time, rate, and amount of loading. Most tissues within the musculoskeletal system demonstrate at least some degree of viscoelasticity. After plotting the load-elongation results of a nondestructive tensile load, there remains a region between the loading and unloading curves that is known as hysteresis and clearly shows the time-dependent effects that viscoelasticity introduces (Fig. 2.6). Preconditioning with repeated loading and unloading of the tissue decreases this area of hysteresis and maximizes elongation of the tissue, which is why athletes precondition the tissues in their bodies by completing repetitive stretching activities.

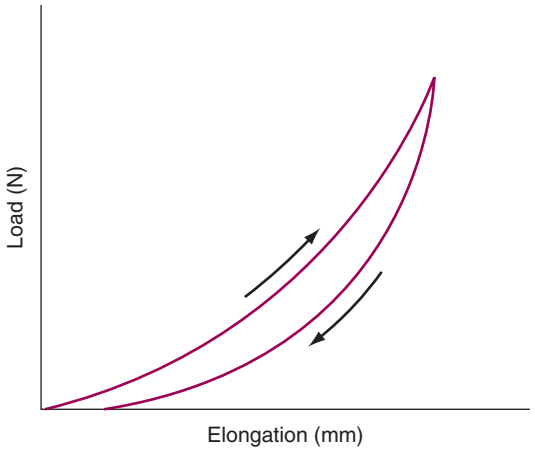

Fig. 2.6 Load-elongation curve of a biologic soft tissue in response to the application of a tensile load. The area between the loading and unloading curves represents the energy absorbed by the tissue during this loading regimen, or hysteresis.

A viscoelastic material experiences the phenomena of creep and stress-relaxation. *Creep* describes a progressive increase in elongation of a material when exposed to a constant load over time (Fig. 2.7A). In simple terms, creep can be considered the tendency of a material to move or deform in response to a constant stress. Conversely, with stress-relaxation, a decrease in load occurs over time upon application of a constant elongation (see Fig. 2.7B). Stress-relaxation is the phenomenon that occurs in a material to relieve stress under a constant strain due to the liquid-like characteristic of viscoelasticity. After a period of time, for both creep and stress-relaxation, the tissue will reach an equilibrium state of elongation and load, respectively.

A practical use of both stress-relaxation and creep in the clinic is for initial graft tensioning of ligament or tendon reconstructions. Over time it is impractical to expect that the initial graft tension will be maintained after being fixed. The length of the graft will inevitably increase from its original length because of a creep response. Moreover, the rate of loading is important because a faster loading rate will result in greater stiffness. Viscoelasticity is also experienced in the articular cartilage of the knee. For example, as the rate of compressive loading increases, such as during running activities, the tissue becomes much stiffer.[2] This increase in stiffness allows for improved protection to the underlying bone while forces at the joint are high. Overall there is not a substantial strain rate dependency on tendons and ligaments; however, bones exhibit large changes in stiffness with increases in the rate of loading. In addition, bone has a greater compressive strength than tensile strength, which is advantageous during sporting activities when a sudden powerful blow may occur to a portion of the body.

BIOMECHANICAL METHODS

In traditional approaches to evaluating both static and dynamic systems, it is assumed that each object in the system is a rigid body. Biomechanics accounts for the biologic materials of the musculoskeletal system in a more realistic manner by considering that load is experienced at the tissue level and that these tissues are deformable. These biologic materials include both soft (articular cartilage, tendons, ligaments, capsular tissues) and hard (bone) tissues. Tissue structures that have been weakened by disease or trauma may not be able to adequately resist the loads that are applied.

Structural Properties

Structural properties are represented by the mechanics of a structure that may incorporate multiple materials or tissue types, including the response of this complex to tensile, shear, and compressive loading. For example, the structural properties of the femur-ACL-tibia complex can be determined in response to a tensile load to assess its load-elongation behavior. To do this, a tensile force (F) is applied to the bone-ligament-bone complex, causing the tissue to become stretched until the complex ruptures. While loading is applied, the corresponding increase of length

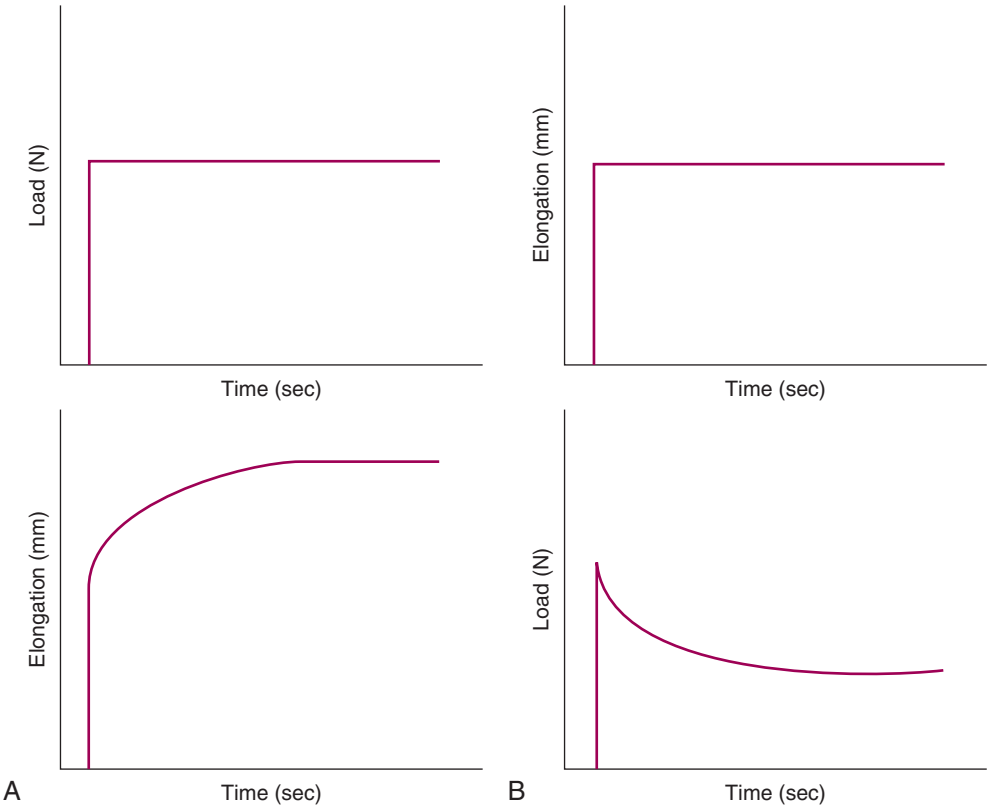

Fig. 2.7 Viscoelastic phenomena exhibited by biologic tissues include creep (A), that is, response due to a constant applied load, and stress-relaxation (B), that is, response due to a constant applied elongation over some time.

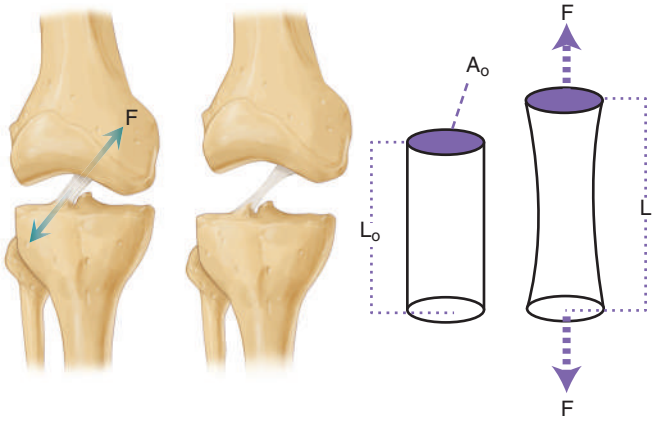

Fig. 2.8 Tensile loading along the longitudinal axis of the anterior cruciate ligament causes a change in length from its initial length *(L_0)* to an elongated state *(L)*. *A_0,* Cross-sectional area; *F,* applied force.

is measured (Fig. 2.8). The resulting nonlinear load-elongation curve that is typical of biologic soft tissues consists of four primary regions (Fig. 2.9A). As the tensile load is first applied, the relationship between load and elongation is nonlinear and is referred to as the *toe* region. This level of loading is typically experienced during a clinical examination as normal fiber recruitment occurs. However, with increasing loads, the relationship becomes more linear and represents loading experienced during daily and sporting activities. The slope of this linear region of the curve is known as the *stiffness*. When the soft tissue begins to sustain more load than the structure can support, plastic deformation begins to occur. Finally, the entire structure approaches the failure region and completely ruptures. The point at which failure occurs is known as the *ultimate load*. Another structural property is the energy absorbed as a result of loading of the tissue, which is determined by the area under the load-elongation curve.

Mechanical Properties

It is also important to understand the mechanical response of an individual tissue or material, which is independent of specimen geometry, specifically of the cross-sectional area and initial length, by using normalized load and deformation parameters. Mechanical properties can be used to evaluate the quality of the tissue when making comparisons between normal, injured, and healing states and are represented by the stress-strain relationship. *Stress* is defined as the amount of force applied per unit area and is one of the most basic engineering principles. *Strain* can be considered a dimensionless measure of the degree of deformation and is defined as the change in length per unit length. Specimens with a greater length can withstand more total deformation, whereas specimens with a larger cross-sectional area can carry loads of greater magnitude. The mechanical properties can be derived from a plot of the stress and strain data and may include modulus, ultimate strength, ultimate strain, and strain energy density. Stress-strain relationships are obtained

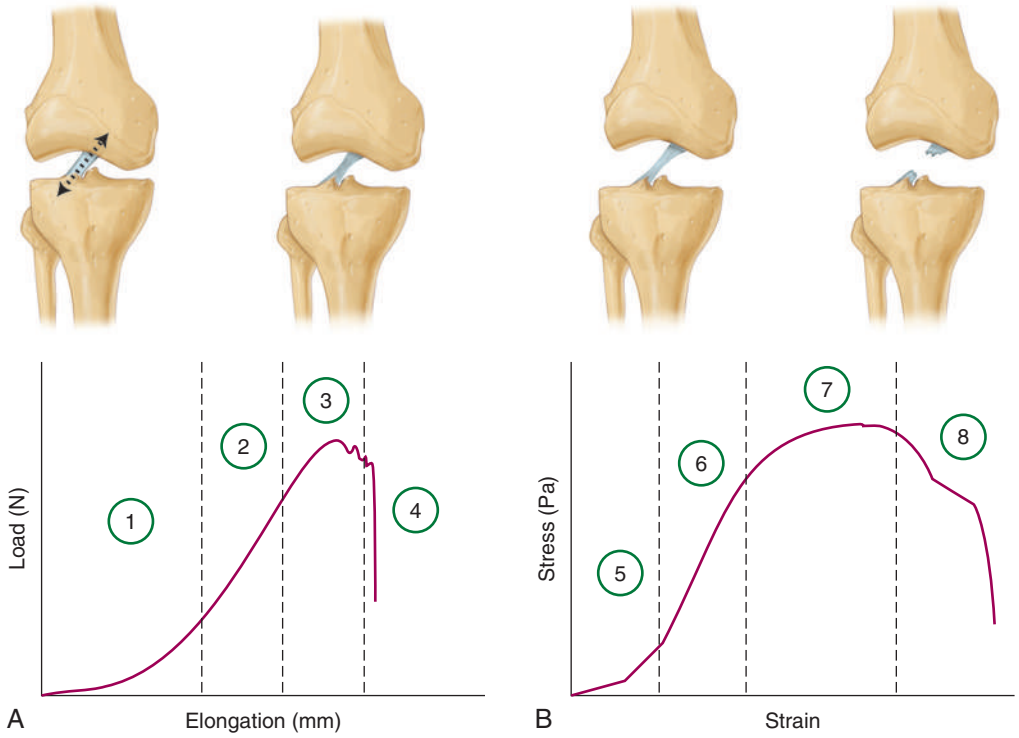

Fig. 2.9 (A) Load-elongation curve of a bone-ligament-bone complex in response to tensile loading characterizing its structural properties. Regions 1, 2, 3, and 4 correspond to the toe region, linear region, partial failure of the complex, and complete rupture of the complex, respectively. The loading of the tensile-bearing structure during activities of daily living will remain in regions 1 and 2. (B) Stress-strain curve of the ligament substance that characterizes its mechanical properties. Regions 5, 6, 7, and 8 correspond to the toe region, linear region, partial failure of the material, and complete rupture of the material, respectively. *Pa,* Pascal.

experimentally during tensile, compressive, or shear loading of the excised tissue.

There are four distinct regions along a stress-strain curve for biologic tissues (see Fig. 2.9B). A typical stress-strain curve begins with a nonlinear toe region (region 5). Stretching of the crimped collagen fibrils occurs within this region as the fibers are being drawn taut before significant tension can be measured. Strain becomes linearly proportional to stress in region 6, and the slope of the curve in this region can be calculated to determine the tangent modulus of the tissue. The area under the curve within this region can be referred to as the *strain energy density*. The tissue returns to its original length or shape once the stress is removed within this zone, which is usually reached during most daily activities. Furthermore, the energy used to deform the tissue is released when the applied stress is removed. When the tissue experiences extreme and abnormally large strain, the tissue undergoes only a marginal increase in corresponding stress (region 7). It is at this point that the tissue begins experiencing microscopic failures. The area under this region of the curve represents plastic deformation energy. Once the tissue undergoes this amount of deformation, the tissue does not recover and return to its original state in its entirety upon release of the deforming stress. If the tissue continues to deform, it will eventually experience complete failure (region 8). The mechanical properties of a tissue can be used to evaluate injured and healing states. When ligament repair is being evaluated, the medial collateral ligament has much lower modulus and maximum stress mechanical properties during the healing process after rupture, although it shows some improvement over time without full recovery.[3]

To determine the mechanical properties of a knee cruciate ligament, a tensile load is applied along the long axis of the excised tissue sample after measuring the original cross-sectional area of the tissue (see Fig. 2.8). When a load is applied in the direction perpendicular to the tissue structure such that there is only a linear change in deformation, the load and deformation are considered to be axial stress and strain, respectively. As the load is applied along the long axis of the ACL in Fig. 2.8, a decrease in the cross-sectional area at the midsubstance occurs in response to an increase in the overall length (L) of the ligament from the initial length (L0). This response is important to consider in converting load data, a structural property, to stress data, a mechanical property, because the structure is changing dimensions throughout the testing. In calculating the mechanical properties, these changes are often ignored and the original, initial conditions are used. Compressive loading is commonly used to test bone or articular cartilage, whereas tensile and shear testing may be used for connective structures (e.g., tendons, ligaments, and capsular tissues).

Shear stress and strain act parallel to the surface of the tissue (Fig. 2.10), and the body is observed to deform into a rhomboid with sides equal to 1 if the body is originally a unit cube. Shear strain can be measured as the angular change (γ) at any point