

Chapter 2

Mechanical Properties of Biological Materials

2.1 Introduction

When we want to design, that is, prepare a drawing and fabricate something to replace a desired body part or malfunctioning organ(s) due to disease processes, trauma, or surgical removal, it is necessary to understand the real nature and biomechanical characteristics of those anatomical parts, e.g., tissues and organs. Anthropometry is the science and practice of measuring the size and shape of the human body and its parts. To measure the properties of biological materials and tissues are also anthropometry's tasks. Biomechanics, in turn, studies the structure and function of biological systems using the methods of mechanics. The composition and behavior of bones, cartilages, and ligaments have been studied for many years. However, although we know much about these tissues, newer and better measurement techniques continuously improve the available data. It should be remembered that there are biological variations and environmental factors that significantly affect the mechanical properties of biological tissues.

This chapter introduces the terms and procedures involving biomaterials and tries to identify the last consensus data regarding the hard, but at the same time deformable, tissues relevant in the study of the human joints' behavior. In this group, we include skeletal bones, articular cartilage, and ligaments.

2.2 Structural Versus Material Properties

A biological tissue is often described in terms of its structural and material properties. **Structural properties** characterize the tissue in its intact form. Important structural properties are represented by a relationship between force and deformation, or stress and strain, and must be understood in order to predict how a tissue will behave in vivo.

Material properties characterize the behavior of the material comprising the tissue and to a first approximation are independent of the size of the tissue. The material

properties are usually expressed in terms of the stress–strain relationship of the material. The strength of a material, which is the breaking or ultimate strength under different modes of loading, such as tension, compression, torsion, or bending, will be different, as will the corresponding modulus of elasticity or stiffness, except bending.

The **stiffness** of a material represents the material's ability to resist deformation. Stiffness is commonly characterized by the slope of the linear region of a stress–strain curve, also referred to as **Young's modulus** when tested **under tension**. To describe the slope of other regions of the stress–strain curve, a tangent modulus is often defined. A **tangent modulus** should have associated with it a strain value or a range of strains. There can be different kinds of moduli depending on the loading types (e.g., shear modulus, compression modulus). The larger the stiffness, the greater the force required to cause a given deformation. If the stress in a material is directly proportional to the strain for strains up to the elastic limit, the material is called a **Hookean** material.

2.2.1 *Anisotropy and Nonhomogeneity*

Ideal materials are isotropic and homogeneous. A material is called isotropic when its properties are the same in each of three coordinate axes' (x , y , z) direction. Tensile and compressive properties may be different, but each respective property must be the same in three directions. A material is said to be homogeneous if it is made of the same material throughout. Biological tissues are **anisotropic** and **nonhomogeneous**.

2.2.2 *Viscoelastic Properties*

Biological tissues are viscoelastic materials; their behavior is both viscous, meaning time- and history-dependent, as well as elastic. A viscoelastic material possesses characteristics of **stress-relaxation**, **creep**, **strain-rate sensitivity**, and **hysteresis**. Force-relaxation (or **stress-relaxation**) is a phenomenon that occurs in a tissue stretched and held at a fixed length. Over time the stress developed within the tissue continually declines. Stress-relaxation is force- or strain-rate-sensitive. In general, the higher the strain or loading rate, the larger the peak force/stress and subsequently the greater the magnitude of the force-relaxation. In contrast to stress-relaxation, which occurs when a tissue's length is held fixed, is **creep**. **Creep** occurs with time when a constant force/stress is applied across the tissue. If subjected to a constant tensile force, then a tissue elongates with time. The general shape of the displacement-time curve depends on the past loading history (e.g., peak force, loading rate).

Another time-dependent property is **strain-rate sensitivity**. Different tissues show different sensitivities to strain rate. For example, there may be little difference in the stress–strain behavior of ligaments subjected to tensile tests varying in strain rate over 3 decades, while bone properties may change considerably

Additionally, the loading and unloading curves obtained from a force-deformation test of biological tissues do not follow the same path. The difference in the calculated area under the loading and unloading curves is termed the area of **hysteresis** and represents the energy lost due to internal friction in the material. The amount of energy liberated or absorbed during a tensile test is defined as the integral of the force and the displacement. Hence, the maximum energy absorbed at failure equals the area under the force–displacement curve.

Y. C. Fung [1], in his text *QLV (Quasilinear Viscoelastic) Theory*, suggested that if a step increase in elongation is imposed on the specimen, the stress developed will be a function of time (t) as well as of the material's stretch ratio (λ). The history of the stress response, called the relaxation function $[K(\lambda, t)]$, is assumed to be of the form $[K(\lambda, t) = G(t) * T(\lambda)]$ in which $G(t)$ is a normalized function of time, called the reduced relaxation function, and $T(\lambda)$ is a function of the stretch ratio alone, called the elastic response. Fung also proposed a function for defining the elastic response of the material under tension conditions.

2.2.3 Viscosity

The viscosity of a fluid is a measure of the fluid's resistance to flow. The viscosity of water is used as a reference to calculate other fluids' viscosity and is considered to be 1. The capsule of diarthrodial joints is normally filled with a fluid of viscosity 10 called synovial fluid. This fluid helps to reduce friction and wear of articulating surfaces. Just for comparison, the viscosity of olive oil, for example, is 84 [2].

2.3 Testing Procedures

Structural properties of biological tissues are usually determined through some form of mechanical testing (e.g., tensile tests, compressive tests, bending and torsion tests). Customized workstations utilizing force transducers, clamps, and an actuator to control the distance between clamps are commonplace. Commercial systems are also available and vary in design depending on the type of tissue being studied (e.g., macroscopic vs. microscopic, hard tissue vs. soft tissue, etc.) and the type of loading rates required. Instron and MTS are the two most common suppliers of mechanical testing systems. Currently, one UK-based company, McMesin, is also supplying such a machine. Most systems allow either force control or length control. See the pictures in Fig. 2.1.

Mechanical testing of tissue *in vivo* is very difficult and hence not commonly performed. Some of the techniques that have been utilized include (1) buckle transducers to monitor tendon and ligament forces, (2) telemetried pressure sensors to measure joint contact pressure, and (3) strain gauges to quantify bone and ligament strain. Some noninvasive approaches have also been employed. Ultrasound techniques

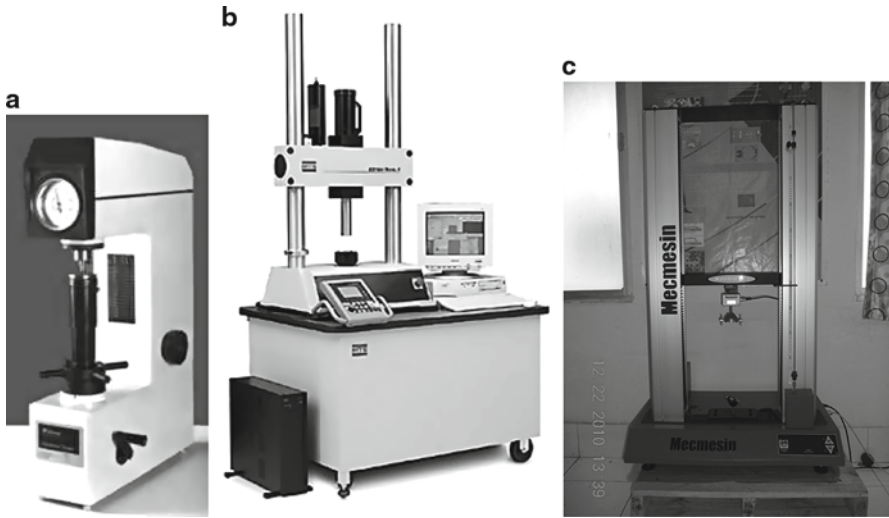


Fig. 2.1 Mechanical testing system: (a) hardness tester; (b) and (c) axial and bending load tester

have been used to detect changes in the speed of sound in different tissues, and these changes have been correlated with the tissue's elastic properties.

Various imaging techniques have also been used to quantify tissue geometry and deformation [2].

2.4 Bones

2.4.1 Composition

Bone is a composite material consisting of both fluid and solid phases. Two main solid phases, one organic and another inorganic, give bones their hard structure. An organic extracellular collagenous matrix is impregnated with inorganic materials, especially hydroxyapatite $\text{Ca}_{10}(\text{PO}_4)_6(\text{OH})_2$ (consisting of the minerals calcium and phosphate). Unlike collagen, apatite crystals are very stiff and strong. However, a bone's strength is higher than that of either apatite or collagen because, similar to what happens with concrete, the softer component prevents the stiff one from brittle cracking, while the stiff component prevents the soft one from yielding. The organic material gives bone its flexibility, while the inorganic material gives bone its resilience.

Calcium and phosphate account for roughly 65–70% of a bone's dry weight. Collagen fibers compose approximately 95% of the extracellular matrix and account for 25–30% of the dry weight of bone. Surrounding the mineralized collagen fibers is a ground substance consisting of protein, polysaccharides, or glycosaminoglycans (GAGs), primarily

in the form of complex macromolecules called proteoglycans. The GAGs serve to cement together the various layers of mineralized collagen fibers. Water accounts for up to 25% of the total weight of bone, with about 85% of the water being located in the organic matrix around the collagen fibers and ground substance. The other 15% is located in canals and cavities that house the bone cells.

2.4.2 Structure

Bone is identified as either **cancellous** (also referred to as trabecular or spongy) or **cortical** (also referred to as compact); see Figs. 2.2a and b. Cortical bone is roughly four times the mass of cancellous bone, in any long bone. The basic material comprising cancellous and compact bone appears identical; thus, the distinction between the two is the degree of porosity and the organization. The porosity of cortical bone ranges from 5 to 30%, while cancellous bone's porosity ranges from 30 to 90%. Bone porosity is not fixed and can change in response to altered loading, disease, and the aging process.

The fibrous layer covering all bones is the **periosteum**. This membrane covers the entire bone except the joint surfaces, which are covered with articular cartilage.

There are numerous terms used to describe the complex architecture of bone at a finer resolution. Both cortical and cancellous bone may contain two types of basic architecture, **woven** and **lamellar**. Bone can also be described as **primary** or **secondary** bone; regions within cortical bone are often described as either **haversian** or **lamellar**. Details about this may be found in any textbook on the biomechanics of bones.

In the human **femur**, there is a remarkable adaptation of the inner structure of the bone to the mechanical requirements due to the load on the femur head. The various parts of the femur taken together form a single mechanical structure wonderfully well adapted for the efficient, economical transmission of the loads from the acetabulum to the tibia. The bony material is arranged in the paths of the maximum internal stresses, which are thereby resisted and transmitted with the greatest efficiency, and hence with a maximum economy of material. The inner structure and external form of human bone are closely adapted to the mechanical conditions existing at every point in the bone. The inner architecture of normal bone is determined by definite and exact requirements of mathematical and mechanical laws to produce a maximum of strength with a minimum of material.

The cancellous bone of the upper femur to the lower limit of the lesser trochanter is composed of two distinct groups of trabeculae arranged in a nonlinear path: One has its origin in the medial (inner) side of the shaft and curving upward in a fan-like radii to the opposite side of the bone; the other originates in the outer portion of the shaft and runs upward and medially to end in the upper surface of the greater trochanter, neck, and head. These two systems cross each other at 90° angles.

In the shaft, the inner architecture is configured in order to economize for resisting shearing stresses, bending moment, and axial stress. Its structure serves to secure great strength with a relatively small amount of material (Fig. 2.3).

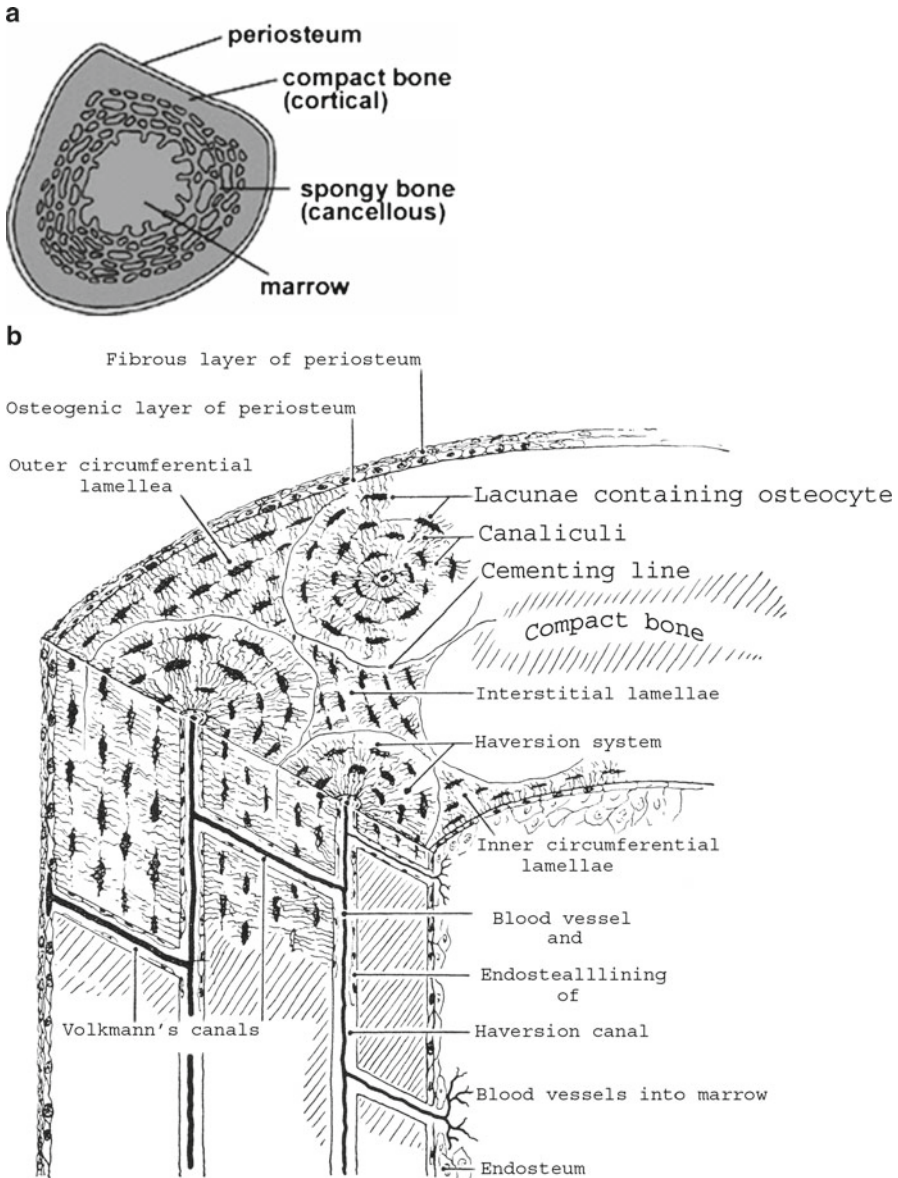


Fig. 2.2 (a) Cancellous and compact bone; (b) the conceptual structure of compact bone adopted from Ham with modification (1969)

In the **pelvis**, the thicker parts of the bone consist of cancellous tissue, enclosed between two layers of compact tissue; the thinner parts, as at the bottom of the acetabulum and the center of the iliac fosse, are usually semitransparent and composed entirely of compact tissue. It may be noted that the properties of bone vary from species to species, race to race, region to region, male to female, young to old,

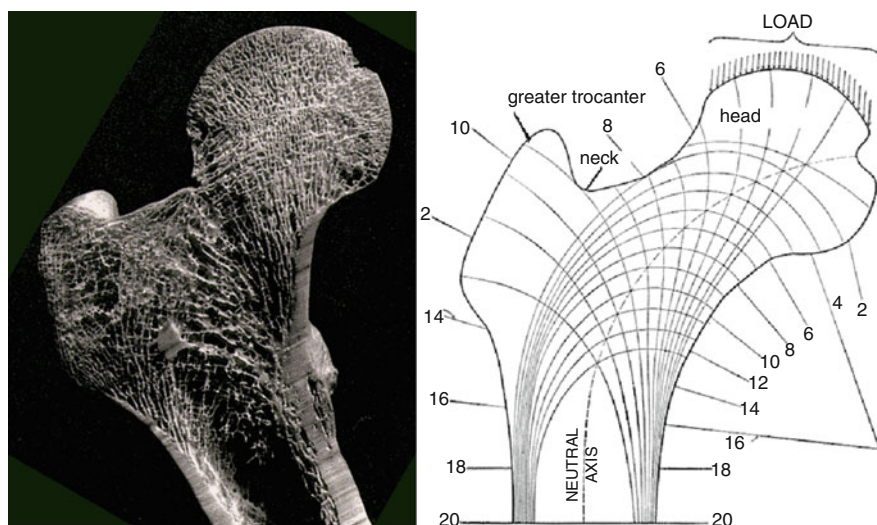


Fig. 2.3 Photograph of the upper femur in coronal section (*left*), and lines of stress (*right*), based upon the mathematical analysis of the right femur [7]

fresh to dry or embalmed, and direction to direction. In the same body there is a regional variation, and bone remodels itself according to the stress generated within it during activities. To illustrate these facts, Table 2.1 shows some illustrative values of the compact bone properties of various species in wet conditions and when loaded parallel to the axis.

2.5 Material Properties and Related Behavior

The mineral content of bone affects its mechanical property. Higher mineralization makes the bone stronger and stiffer (higher modulus of elasticity), but it lowers the toughness; that is, it is less capable of absorbing shock and strain energy. The organic phase makes it more pliable and shock-absorbing, which are desirable for athletes.

Cancellous bone is actually extremely anisotropic and nonhomogeneous. Cortical bone, on the other hand, is approximately linear elastic, transversely isotropic, and relatively homogenous. The material properties of bone are generally determined using mechanical testing procedures; however, ultrasonic techniques have also been employed. Force-deformation (structural properties) or stress-strain (material properties) curves can be determined using tests. However, the properties of bone and most biological tissues depend on the freshness of the tissue. These properties can change within a matter of minutes if allowed to dry out in the open. Cortical bone, for example, has an ultimate strain of around 1.2% when wet and about 0.4% if the water content is not maintained. Thus, it is very important to keep bone specimens wet in lactated Ringer's solution or normal saline water during testing.

Table 2.1 Mechanical properties of wet compact bone in various mode of loading parallel to axis

Bone	Horses	Cattle	Pigs	Human (20–39 years)
Ultimate tensile strength (MPa)				
Femur	121±1.8	113±2.1	88±1.5	124±1.1
Tibia	113	132±2.8	108±3.9	174±1.2
Humerus	102±1.3	101±0.7	88±7.3	125±0.8
Radius	120	135±1.6	100±3.4	152±1.4
Ultimate percentage elongation				
Femur	0.75±0.008	0.88±0.020	0.68±0.010	1.41
Tibia	0.70	0.78±0.008	0.76±0.028	1.50
Humerus	0.65±0.005	0.76±0.006	0.70±0.033	1.43
Radius	0.71	0.79±0.009	0.73±0.032	1.50
Modulus of elasticity in tension (GPa)				
Femur	25.5	25.0	14.9	17.6
Tibia	23.8	24.5	17.2	18.4
Humerus	17.8	18.3	14.6	17.5
Radius	22.8	25.9	15.8	18.9
Ultimate compressive strength (MPa)				
Femur	145±1.6	147±1.1	100±0.7	107±4.3
Tibia	163	159±1.4	106±1.1	
Humerus	154	144±1.3	102±1.6	
Radius	156	152±1.5	107±1.6	
Ultimate percentage contraction				
Femur	2.4	1.7±0.02	1.9±0.02	1.85±0.04
Tibia	2.2	1.8±0.02	1.9±0.02	
Humerus	2.0±0.03	1.8±0.02	1.9±0.02	
Radius	2.3	1.8±0.02	1.9±0.02	
Modulus of elasticity in compression (GPa)				
Femur	9.4±0.47	8.7	4.9	
Tibia	8.5		5.1	
Humerus	9.0		5.0	
Radius	8.4		5.3	
Ultimate shear strength (MPa)				
Femur	99±1.5	91±1.6	65±1.9	54±0.6
Tibia	89±2.7	95±2.0	71±2.8	
Humerus	90±1.7	86±1.1	59±2.0	
Radius	94±3.3	93±1.8	64±3.2	
Torsional modulus of elasticity (GPa)				
Femur	16.3	16.8	13.5	3.2
Tibia	19.1	17.1	15.7	
Humerus	23.5	14.9	15.0	
Radius	15.8	14.3	8.4	

Bone shows a linear range in which the stress increases in proportion to the strain. The slope of this region is defined as Young’s modulus, or the elastic modulus. An illustration of the material properties of bone relative to other materials is shown in Fig. 2.4.

Fig. 2.4 Comparative properties of different materials

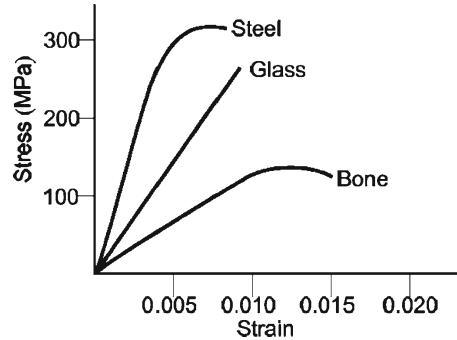
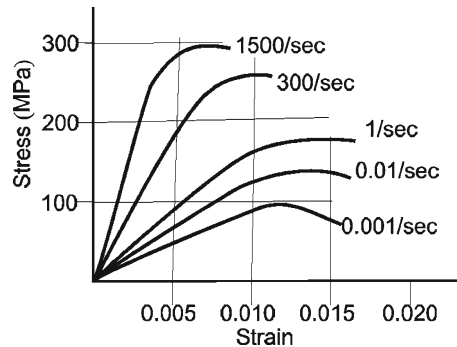


Fig. 2.5 Strain-rate sensitivity of cortical bone. As the strain rate increases, the ultimate strength increases and the ultimate strain decreases



Bone is strain rate-sensitive (Fig. 2.5) and tends to be more strain rate-sensitive than other biological tissues. This has implications for bone–ligament and bone–tendon injuries. The optimal strain rate for energy absorption is around 0.1–1 per second.

Material properties of the two types of bone differ. Cortical bone is more than 2 decades stiffer than cancellous bone. It can sustain greater stress but less strain before failure. Cancellous bone can sustain strains of 75% before failing *in vivo*, but cortical bone will fracture if the strain exceeds 2%. Cancellous bone has a greater capacity to store energy compared to compact bone since it is porous and filled with fluid, including blood, marrow, and body fluid.

Three important parameters that characterize some of the mechanical properties of bone—ultimate force, maximum deformation to failure, and the energy that it can store before failing—can be obtained from a force-deformation curve. The ultimate force represents the maximum load that the bone can sustain before it breaks. The ultimate force varies depending on the type of load applied (e.g., tensile, compressive, shear) and the loading rate. The deformation at failure is self-explanatory and also depends on the loading rate and direction. The energy absorbed before failing can be calculated from the area under the force-deformation curve and therefore depends on both the ultimate force and the ultimate strain. Children’s bones tend to absorb more energy before failure compared to adults (as much as 45% more).

Children's bones are weaker but more compliant (children's bones can be 68% as stiff as adult bone).

2.6 Cartilage

2.6.1 Composition

Articular cartilage, also called hyaline cartilage, is made of a multiphasic material with two major phases: a fluid phase composed of water (68–85%) and electrolytes, and a solid phase composed of collagen fibrils (primarily type II collagen) (10–20%), proteoglycans and other glycoproteins (5–10%), and the chondrocytes (cartilaginous cells). Thirty percent of all cartilage water resides in this interstitial fluid, and this amount does not vary with age. However, there is a significant increase in the total amount of water in degenerating cartilages [3].

This multiphasic system allows fluid flowing from the tissue to the solution surrounding the tissue, and vice versa, through the pores of the collagen–proteoglycan solid matrix. As the fluid passes to the pores, the force exerted on the walls of the pores causes more compaction. Thus, it becomes more and more difficult to squeeze fluid from the tissue with prolonged compression. This nonlinear flow-induced compression effect is very important in the physiology of cartilage not just because it determines cartilage compressive viscoelastic behaviors, but also because it provides the mechanism for energy dissipation (Fig. 2.6).

The thickness of articular cartilage varies with the particular joint and the location within the joint. Generally, it ranges from 0.5 mm in rabbit knee joints to 10.0 mm in the patellofemoral groove of bovine knee joints, and in humans it is thickest over the ends of femur and tibia, ranging from 2–4 mm [4].

The distribution and arrangement of cartilage components are not uniform. Instead, each layer has different biochemical, structural, and cellular characteristics. Some authors consider articular cartilage to have three distinct layers (superficial, 10–20%; middle, 40–60%; and deep, 25–35%) along its depth. Others prefer to divide articular cartilage into four zones: superficial, middle, deep, and calcified (Fig. 2.7). The *superficial zone* is characterized by flattened chondrocytes, relatively low quantities of proteoglycan, and high quantities of collagen fibrils arranged parallel to the articular surface. The *middle zone*, in contrast, has round chondrocytes, the highest level of proteoglycan among the four zones, and a random arrangement of collagen. The *deep zone* is characterized by collagen fibrils that are perpendicular to the underlying bone, and columns of chondrocytes arrayed along the axis of fibril orientation. The *calcified zone* is partly mineralized and acts as the transition between cartilage and the underlying subchondral bone. Considering either three or four layers, based on the depth-related differences in the structural, biochemical, and cellular compositions, it is reasonable to assume that the intrinsic mechanical properties of articular cartilage vary with depth.

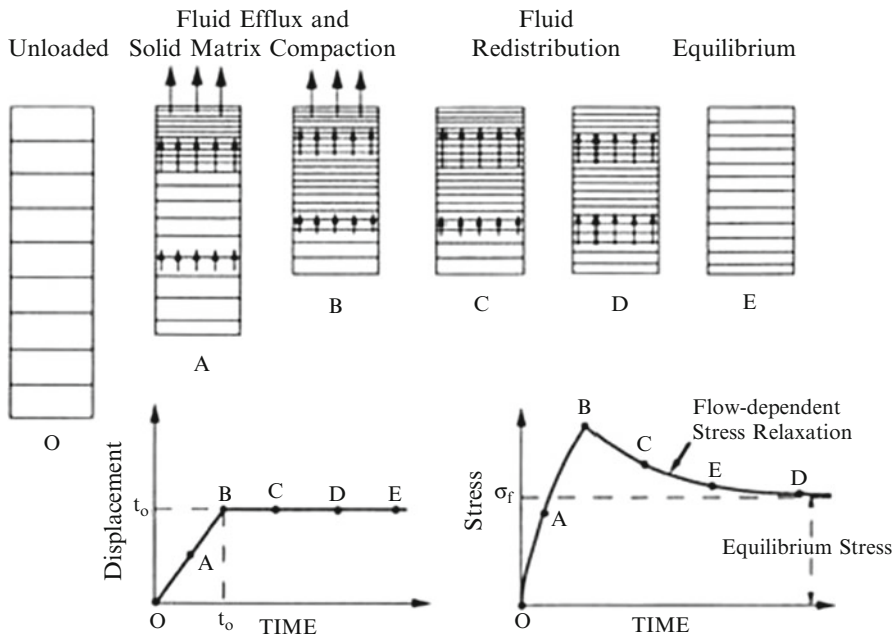


Fig. 2.6 Viscoelastic behavior of cartilage

2.7 Material Properties and Related Behavior

Interactions take place among the fluid, proteoglycan molecules, and various electrostatic charges, providing superior quality of lubrication and shock absorption. The cartilaginous tissue is extremely well adapted to glide. Its coefficient of friction is several times smaller than that between ice and an ice skate. There are electrostatic attractions between the positive charges along the collagen molecules and the negative charges that exist along the proteoglycan molecules. Hydrostatic forces also exist as forces are applied to cartilage and the fluid tries to move throughout the tissue. It is the combined effect of all these interactions that gives rise to the mechanical properties of the material.

Like bone, cartilage is an anisotropic material. The anisotropy results in part from the structural variations noted above. Because of its structure, cartilage is rather porous, allowing fluid to move in and out of the tissue. When the tissue is subjected to a compressive stress, fluid flows out of the tissue. Fluid returns when the stress is removed.

The mechanical properties of cartilage change with its fluid content, thus making it important to know the stress–strain history of the tissue to predict its load-carrying capacity. The material properties also change with pathology. The compressive aggregate modulus for human articular cartilage correlates in an inverse manner with the water content and in a direct manner with proteoglycan content per wet weight. There is no correlation with the collagen content, thus suggesting that proteoglycans are responsible for the tissue's compressive stiffness.

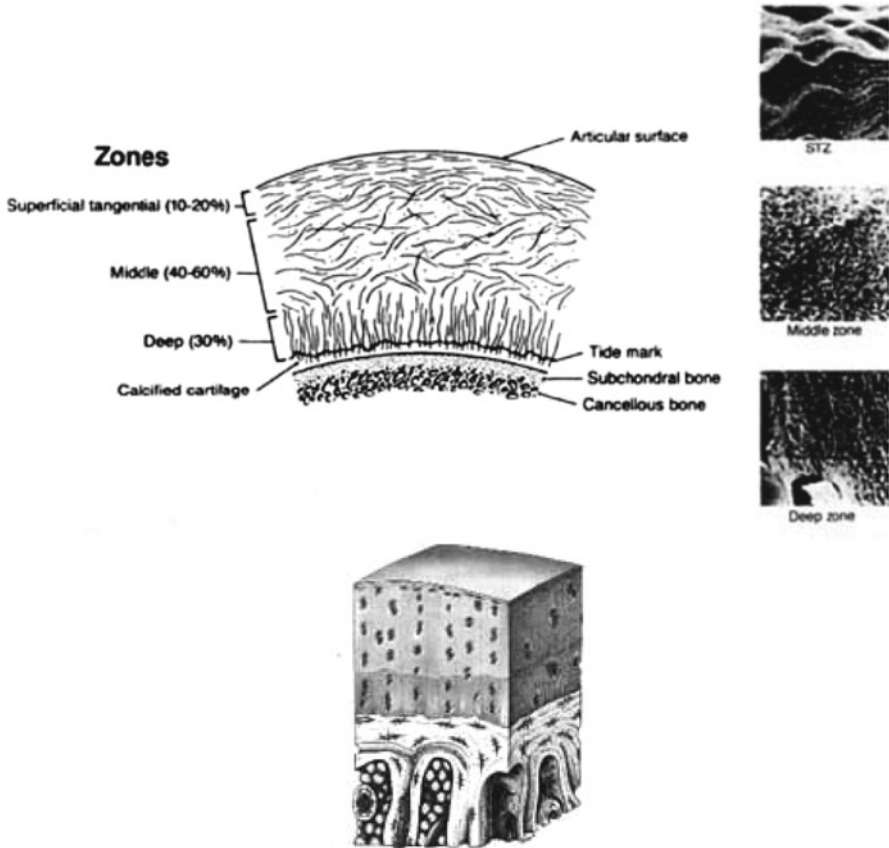


Fig. 2.7 Cartilage zones or layers

2.8 Ligaments

2.8.1 Composition

The major constituents of ligaments are collagen, elastin, glycoproteins, protein polysaccharides, glycolipids, water, and cells (mostly fibrocytes). The greatest quantities of constituents found in ligaments are collagen and ground substance. For practical purposes, the physical behavior of ligaments is usually predicted based on the content and organization of these substances alone [5].

Collagen constitutes 70–80% of the dry weight of ligament, the majority being type I collagen, which is also found in tendon, skin, and bone. Collagen has a relatively long turnover rate, with its average half-life being 300 and 500 days, which is slightly longer than that of bone. Therefore, several months may be required for a ligament to alter its structure to meet changes in physical loading conditions or to repair itself after injury. Water makes up about 60–80% of the wet weight of ligaments.

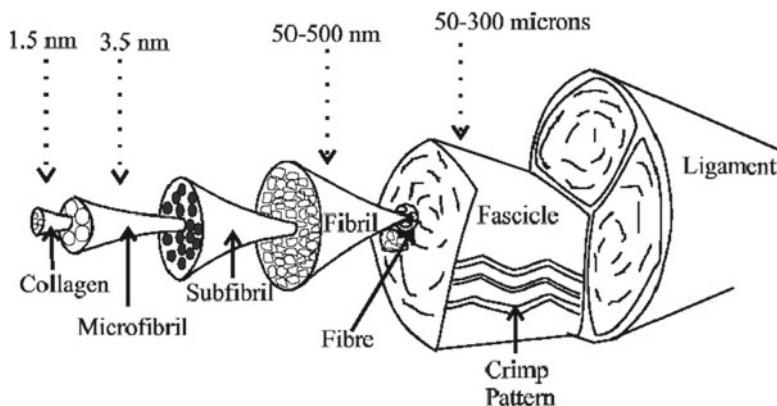
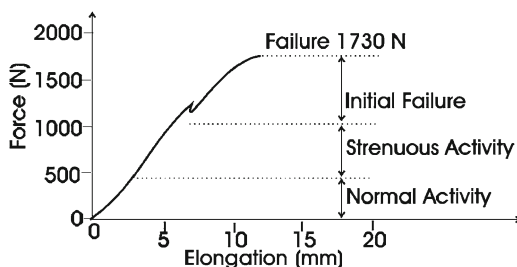


Fig. 2.8 Hierarchical structure of ligaments. The basic structural element is the tropocollagen molecule

Fig. 2.9 Force-elongation diagram obtained during tensile test of ligaments



A significant amount of this water is associated with the ground substance. On a dry weight basis, the ground substance comprises only about 1% of the total tissue mass. The ground substance likely provides lubrication and spacing, which aid in the sliding of fibers. In addition, the presence of ground substance is a source of ligament viscoelastic behavior.

Closely packed, parallel collagen fiber bundles are oriented to provide motion and stability for the musculoskeletal system (Fig. 2.8). Properties can change according to strain rate, temperature, hydration, maturation, aging, immobilization, exercise, and healing.

The structural properties of isolated ligaments and bone–ligament–bone preparations are normally determined via tensile tests. In such a test, a ligament, tendon, or bone–ligament–bone complex is subjected to a tensile load applied at constant rate. A typical force–elongation curve can be obtained from a tensile test, as shown in Fig. 2.9. The force–elongation curve is initially upwardly concave, but the slope becomes nearly linear in the prefailure phase of tensile loading. The force–elongation curve represents structural properties of the ligament. That is, the shape of the curve depends on the geometry of the specimen tested (e.g., tissue length and cross-sectional area).

2.9 Material Properties and Related Behavior

Although significant advances have been made in the biology, biochemistry, and mechanics of soft tissue, there is still much work left to be done. There is limited information available on in vivo tissue mechanical characteristics and behavior. Without accurate values of such in vivo information, extrapolations from animal and human in situ bone–ligament–bone testing to the function of intact human ligaments cannot be made confidently. Currently, we know that ligaments are composite, anisotropic structures exhibiting nonlinear time- and history-dependent viscoelastic properties. Described in this section are the mechanical behavior of ligamentous tissue, the physiological origin of this behavior, and the implications of such properties to ligament function during normal joint motion.

As seen above, the force–elongation curve represents structural properties of the ligament. Material properties, in turn, are more generally expressed in terms of a stress–strain relationship (Fig. 2.10).

2.9.1 Ligaments Have Characteristics of Strain-Rate Sensitivity, Stress-Relaxation, Creep, and Hysteresis

Ligaments exhibit significant time- and history-dependent viscoelastic properties. Time-dependent behavior means that during daily activities, ligaments are subjected to a variety of load conditions that affect their mechanical properties. For example, they become softer and less resistant after some minutes of running, returning to normal hardness when the exercise is interrupted. History dependency, in turn, means that frequent intense activities will change the tissue properties on a medium-term basis. For example, the ligaments of an athlete, after 6 months of daily training, will become softer and thus more adapted to the intense exercise, even when he or she is not training. In the same way, if the activities are interrupted for some months, the ligament properties will go back to normal levels. Figure 2.11 illustrates ligament softening, a decrease in peak loads occurring during cyclic testing of ligaments to a constant strain and at a constant strain rate.

Ligaments are also temperature-sensitive, with peak stresses increasing with decreased temperatures. Bone–ligament–bone preparations tested cyclically at

Fig. 2.10 Stress–strain relationship for human ligament

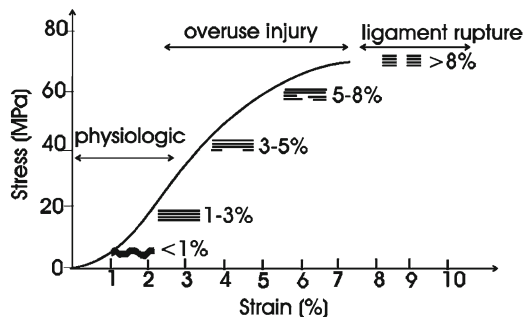
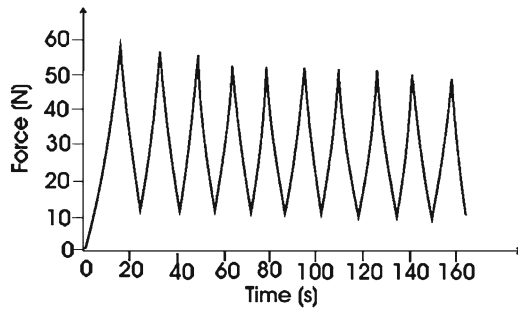


Fig. 2.11 Ligament response to cyclic loading and unloading. Peak loads decreased with each cycle, indicating ligament softening (Woo et.al. 1982)



21 °C show 30% greater peak loads than the same preparation tested at 37 °C. It has been suggested that the temperature of superficial tissues *in vivo* may be within 2 °C of the skin temperature, which can be 10 °C lower than the body temperature.

There are two age-related processes, maturation and aging, that also affect bone–ligament properties. During maturation, the structure and mechanical properties of collagenous tissues change. The stabilization of collagen with maturity enhances tissue strength, while the loss of water and elastin reduces tissue plasticity [6]. Aging connective tissue undergoes a generalized decrease in water content, which results in a reduction in tissue compliance. The elastic elements become coarser and more easily fractured. However, it is very difficult to distinguish aging effects from effects created by other factors such as disease or changes in activity levels. It has been estimated that regular exercise may retard the physiologic decline associated with aging by as much as 50% [6].

2.10 Correlation Between Structure and Function

The “crimp pattern” and the interaction and cross-linking of elastic, reticular, and collagen fibers of ligaments are critical for normal joint mobility. These features allow ligaments to have a limited range of strains over which they produce minimal resistance to movement. As a result, joints may easily be moved in certain directions and over certain ranges. Additionally, if a joint is displaced toward the outer limit of some normal range of motion, the strain in specific ligaments of that joint increases, causing recruitment of collagen fibers from their “crimp” state to a straightened condition. Fiber recruitment causes the ligament to quickly increase its resistance to further elongation, hence stabilizing the joint.

Another feature of ligaments that may be important for maintaining joint integrity is their neural network. Ligaments contain a variety of sensory receptors that may detect joint position, velocity, and acceleration. This feature may indirectly contribute to maintaining joint integrity by initiating the recruitment (or decruitment) of dynamic stabilizers such as muscles. More work is needed in this area to determine the role of these neural components.

Table 2.2 Comparative properties of biological materials

Density	Bone 1,810 kg/m ³	Cartilage 1,100 kg/m ³	Ligament
Strength (variation factor)	1.35 from the weakest to the strongest place	3 MPa (tensile-stress-strain curve becomes nonlinear)	
Strength (correlation coefficient)	0.4–0.42		
Young's modulus	18 GPa (intermediary between apatite = 165 GPa and collagen = 1.24 GPa)	1–10 MPa (tension) 1 MPa (compression)	1.2–1.8 GPa
Shear modulus	3.5 GPa		
Viscoelasticity	Yes	Yes, time- and history-dependent	Yes, time- and history-dependent
Anisotropy	Cancellous = extremely anisotropic Cortical = transversely isotropic	Anisotropic	Yes, for load deformation or stress-strain behavior (primarily oriented to resistance of the tensile loads)
Nonlinear behavior	Cancellous = nonlinear Cortical = approximately linear		Nonlinear
Ultimate stress (at failure)	135 MPa (tension) 200 MPa (compression)		50–150 MPa (tension)
Ultimate strain (at failure)			13–18% (varies among different materials)

Collected from many sources from Reading List

2.10.1 Ligament–Bone/Tendon–Bone Insertions

Tendons have large parallel fibers that insert uniformly into the bone. Ligament fibers are of smaller diameter than the tendon fibers, which can be either parallel or branching and interwoven. Ligament insertion sites are well suited for dissipating force. As the ligament passes through the insertion site, it is transformed from ligament to fibrocartilage and then to bone.

The bone–ligament junction of younger animals is consistently weaker than that of the ligament substance. The reverse is true for mature animals. This suggests an asynchronous rate of maturation between the bone–ligament junction and that of the ligament substance.

Two different types of insertions exist: *direct*, which is more common, where the tendon or ligament crosses the mineralization front and progresses from fibril to

fibrocartilage (usually less than 0.6 mm), to mineralized fibrocartilage (less than 0.4 mm), and finally to bone; *indirect*, which is less common, where it inserts into bone through the periosteum, with short fibers that are anchored to the bone. For quick reference, we have included a table for comparative properties of biological materials (Table 2.2).

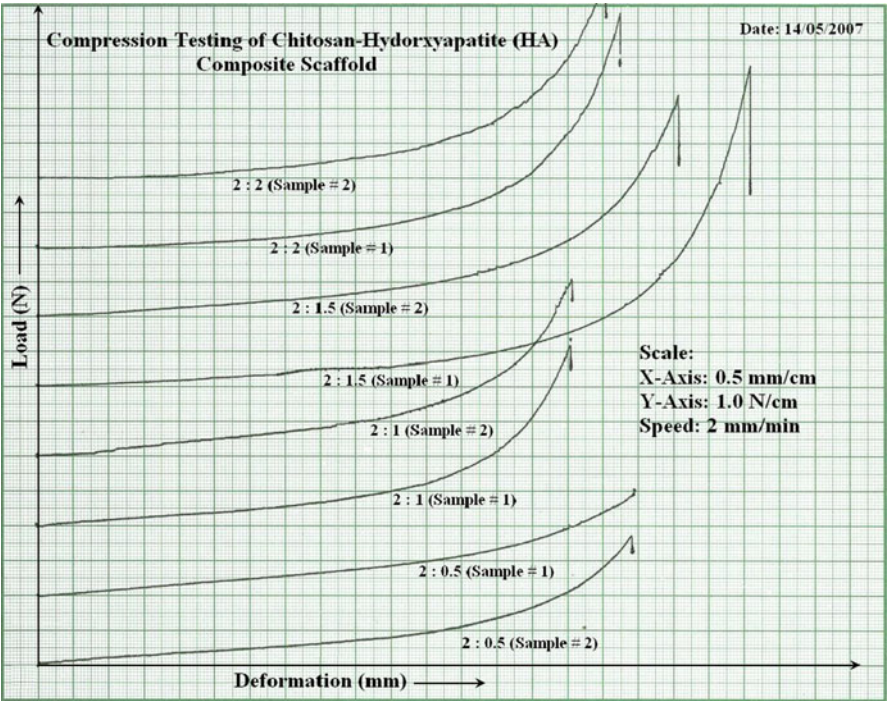
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Problems

1. List a total of five metallic, polymeric, ceramic, and composite biocompatible materials.
List their compositions and mechanical properties, such as ultimate strength, yield strength, ultimate strain, and hardness. In a single plot, draw their stress–strain diagram under tensile load with approximate scale indicating the yield point, ultimate strength, and modulus of elasticity.
2. List names of some hard and soft tissues in the human body. How do we characterize them? Show their stress–strain diagram.
3. Compare the hardness of different components of teeth. What materials will be suitable for the replacement of teeth?
4. The following figure shows the load deformation diagram of a scaffold of chitosan–hydroxyapatite composite mixed in different proportions. Discuss the effect of

HA on the ultimate strength and the primary and secondary moduli of elasticity. The cross-sectional area may be taken as $15 \times 10 \text{ mm}^2$ (based on a 2007 biomedical engineering master's thesis at Jadavpur University). The starting point of each plot is (0, 0).



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