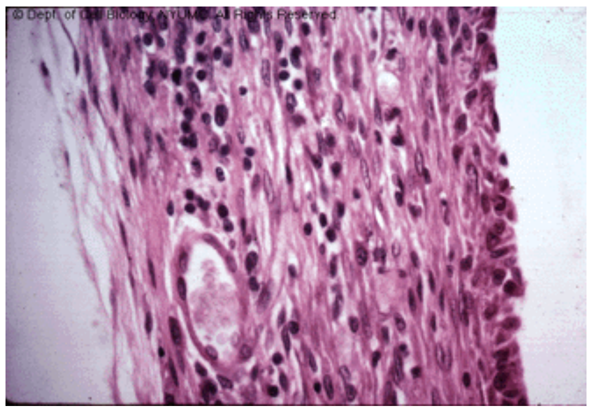
Chapter 4
Ligament

Purpose:
1. To describe ligament structure.
2. To describe the mechanical properties of ligaments.
3. To relate ligament structure to function.
4. To describe how ligaments fail and how they are repaired.
5. To describe how ligament change with aging and adapt to exercise and disuse.



Ligament Anatomy

General Structure and Function

Ligaments, lying internal or external to the joint capsule, bind bone to bone and supply passive support and guidance to joints. They function to supplement active stabilizers (i.e. muscles) and bony geometry (Akeson, Woo, Amiel and Frank, 1984). Ligaments are generally named according to their position in the body (e.g. collateral) or according to their bony attachments (e.g. coracoclavicular).

Well suited for their functional roles, ligaments offer early and increasing resistance to tensile loading over a narrow range of joint motion. This allows joints to move easily within normal limits while causing increased resistance to movement outside this normal range.

Ligaments and tendons are collagenous tissues with their primary building unit being the tropocollagen molecule (Viidik, 1973). Tropocollagen molecules are organized into long cross-striated fibrils that are arranged into bundles to form fibers. Fibers are further grouped into bundles called fascicles which group together to form the ligament (Figure 1).

Collagen fiber bundles are arranged in the direction of functional need and act in conjunction with elastic and reticular fibers along with ground substance, which is a composition of glycosaminoglycans and tissue fluid, to give ligaments their mechanical characteristics. In unstressed ligaments, collagen fibers take on a sinusoidal pattern. This pattern is referred to as a "crimp" pattern and is believed to be created by the cross-linking or binding of collagen fibers with elastic and reticular fibers.

The greatest number of human ligament injuries occurs to ligaments of the knee. Because of their clinical significance, knee ligaments are referred to throughout this chapter to illustrate various concepts. Four major ligaments assist in stabilizing the knee. Basic knee anatomy is illustrated in Figure 2. Two of these ligaments, the anterior cruciate (ACL) and posterior cruciate (PCL), are located within the joint capsule. The ACL is attached to the posterior side of the lateral femoral condyle and to the anterior intercondylar fossa of the tibia. The PCL is attached to the anterior portion of the intercondylar notch on the femur and to the posterior intercondylar fossa of the tibia. The ACL and PCL touch as they span the joint with the ACL passing anterior and lateral to the PCL. The other two ligaments, the medial collateral (MCL), also called the tibial collateral, and the lateral collateral (LCL), also called the fibular collateral, are located external to the joint capsule lying medially and laterally to the joint respectively.

For each plane of knee mobility, there are primary and secondary ligament stabilizers. The primary and secondary knee stabilizers that resist valgus stresses (opening of the medial side of the knee) are the superficial MCL and the ACL respectively. Varus knee stresses (opening of the lateral side of the knee) are resisted by a combination of knee structures depending on the angle of knee flexion. The LCL is the primary stabilizer at the mid-range of knee flexion and the PCL at 90 degrees of flexion. At full extension there is no single structure that acts as the primary stabilizer.

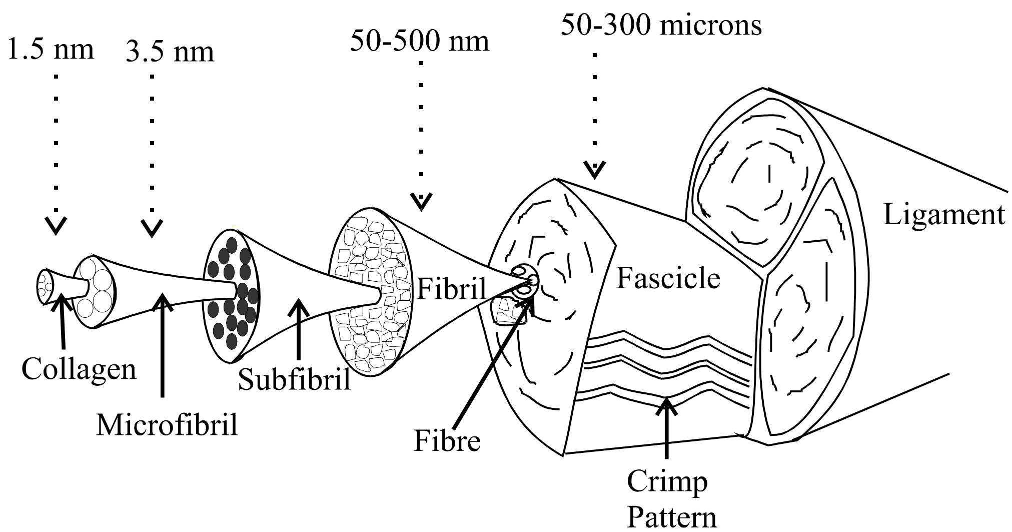


Figure 1 -A schematic diagram of the structural hierarchy of ligament is shown. Ligament is composed of smaller and smaller fiber bundles. The basic structural element is the tropocollagen molecule.

The ACL is the primary check against anterior displacement of the tibia relative to the femur, with the PCL being the primary stabilizer preventing posterior displacement. In knee flexion, the superficial MCL is the first defense against external rotation with the ACL acting as a secondary restraint. In knee extension, the ACL and superficial MCL act together as primary stabilizers against external rotation. With the knee flexed, internal rotation is prevented first by the cruciate ligaments and secondly by the LCL. In extension, the ACL is the primary stabilizer and the LCL the secondary (Marshall and Rubin, 1977).

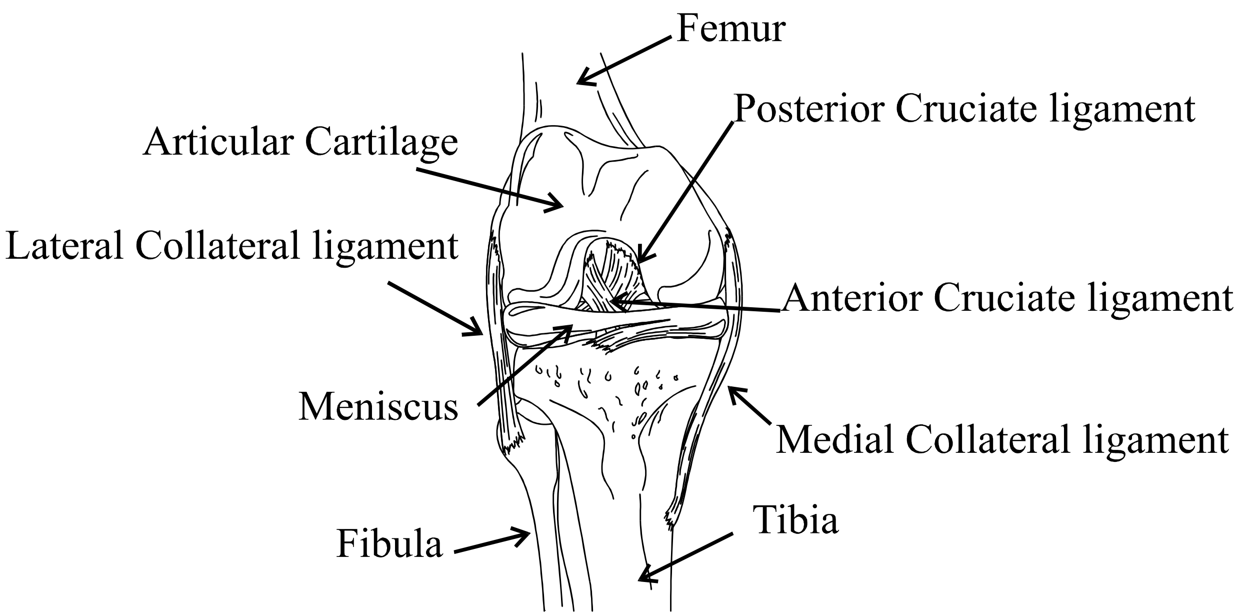


Figure 2 - Illustrated is a schematic representations of the structures of the human knee viewed from an anterior position. The three bony structures are the femur, tibia, and fibula. The four major ligaments supporting the knee are the anterior cruciate, posterior cruciate, medial collateral (or tibial collateral), and the lateral collateral (or fibular collateral).

Biochemical Constituents

The major constituents of ligaments are collagen, elastin, glycoproteins, protein polysaccharides, glycolipids, water, and cells (mostly fibrocytes) (Akeson et al., 1984). The greatest quantities of constituents found in ligaments are collagen and ground substance. For practical purposes the physical behavior of ligaments can be predicted based on the content and organization of these substances alone (Akeson et al., 1984).

Collagen constitutes 70 percent 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, 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 to 80 percent of the wet weight 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 one percent of the total tissue mass. The ground substance likely provides lubrication and spacing that aid in the sliding of fibers. In addition, the presence of ground substance is a source of ligament viscoelastic behavior.

Blood Supply

Blood is usually supplied to ligaments through periarticular arterial plexuses (Akeson et al., 1984; Butler, Grood, Noyes, and Zernicke, 1978; Dye and Cannon, 1988). The extent of the vascular supply varies among ligaments and may depend on their location relative to the joint capsule. Intraligament vasculature is rather limited indicating that some degree of diffusion is necessary to supply the inner ligament fibers with needed nutrients. The blood supply to a ligament is important for the synthesis of new collagen (Butler et al., 1978). As such, the extent of vascular damage that occurs during ligament trauma is a significant factor affecting the ability of the ligament to heal itself. Ligament surgical and rehabilitative procedures should be designed to maintain or enhance the tissue's vascular and blood supplies. Many rehabilitation programs utilize these ideas by using continuous passive motion (CPM) which moves the joint continuously and passively over a limited range of movement for a designated time period. CPM is a means to facilitate blood flow to damaged tissues and promote tissue synthesis.

Neural Structures

In addition to a vascular network, ligaments also contain a variety of neural elements (Kennedy, Alexander, and Hayes, 1982; Schultz, Miller, Kerr, and Micheli, 1984; Halata, Badalamente, Dee, and Propper, 1984; Akeson et al., 1984; Zimny, Schutte, and Dabezies, 1986; Schutte, Dabezies, Zimny, and Happel, 1987; Dye and Cannon, 1988). Nerves present within ligaments originate from nerves innervating muscles. Reflex pathways appear to exist which may allow ligament strain to be communicated to the central nervous system (CNS). The CNS can respond by stimulating specific muscles causing them to contract and prevent further joint displacement. The free nerve endings present in ligament are believed to detect joint position, speed, and movement direction (Akeson et al., 1984).

Biomechanics

Structural/Material Properties

Ligaments are composite, anisotropic structures exhibiting non-linear 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.

The structural properties of isolated tendons, 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 obtained from a tensile test of a rhesus monkey ACL is shown in Figure 3. 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).

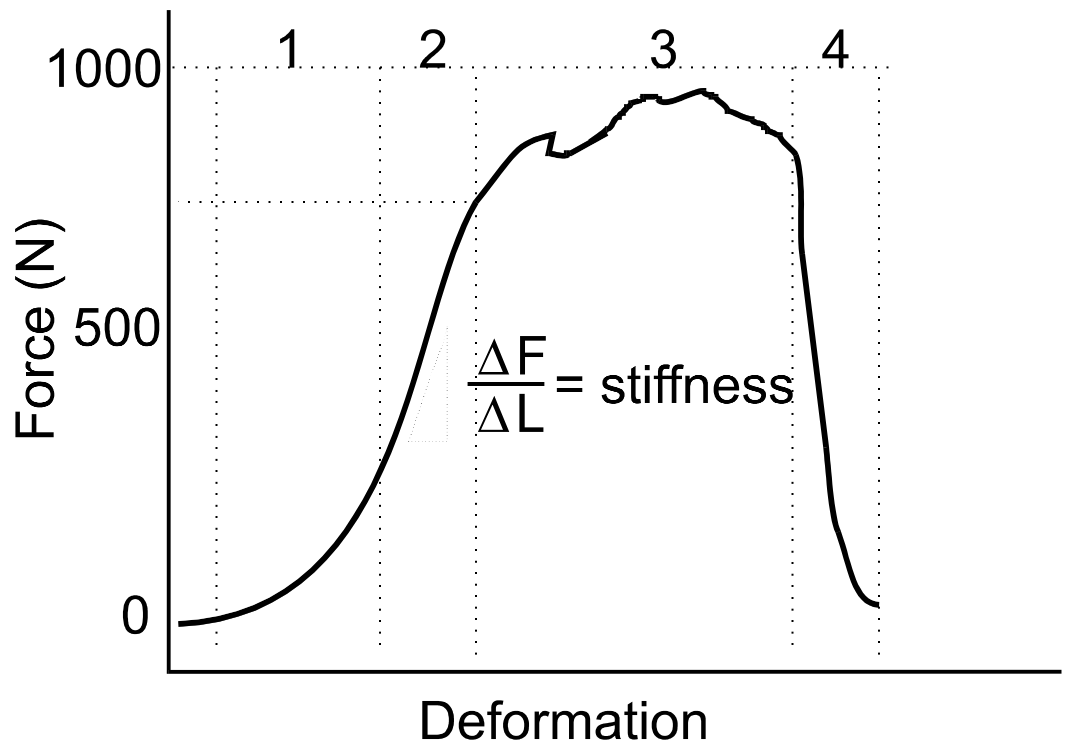


Figure 3 -A force-elongation curve obtained from a tensile test to failure of a rhesus monkey femur-ACL-tibia preparation is shown. There are four regions that are commonly used to describe a force-elongation or stress-strain curve. Region 1 is termed the "toe region" and elicits a non-linear increase in load as the tissue elongates. Region 2 represents the linear region of the curve. In Region 3, isolated collagen fibers are disrupted and begin to fail. In Region 4, the ligament completely ruptures. (Modified from Butler, D.L., Grood, E.S., Noyes, F.R., and Zernicke, R.F., Biomechanics of ligaments and tendons. Exercise and Sports Science Reviews, 6, 125-181, 1984.)

The material properties of the ligament are expressed in terms of a stress-strain relationship. Stress is defined here as the force divided by the original specimen cross-sectional area. Strain is defined as the change in length of the specimen relative to its initial length, divided by its initial length. Hence, a tissue's material properties may be obtained from force-elongation data by dividing the recorded force by the original cross-sectional area to give stress, and by dividing the difference between the specimen length and its original length by its original length to give strain. The advantage of constructing a stress-strain diagram is that to a first approximation the stress-strain behavior is independent of the tissue dimensions.

Stress-strain or force-elongation curves are typically described in terms of four regions. These four regions are illustrated in Figure 3. Region 1 is referred to as the "toe region". The non-linear response observed in this region is due to the straightening of the "crimp" pattern resulting in successive recruitment of ligament fibers as they reach their straightened condition (Abrahams, 1967; Diamant, Keller, Baer, Litt, and Arridge, 1972). As the strain increases, the "crimp" pattern is lost and further deformation stretches the collagen fibers themselves (Region 2). As the strain is further increased, microstructural damage occurs (Region 3). Further stretching causes progressive fiber disruption and ultimately complete ligament rupture or bony avulsion at an insertion site (Region 4). A hypothetical curve relating the "crimp" pattern and collagen fiber stretch to various portions of a stress-strain curve is illustrated in Figure 4

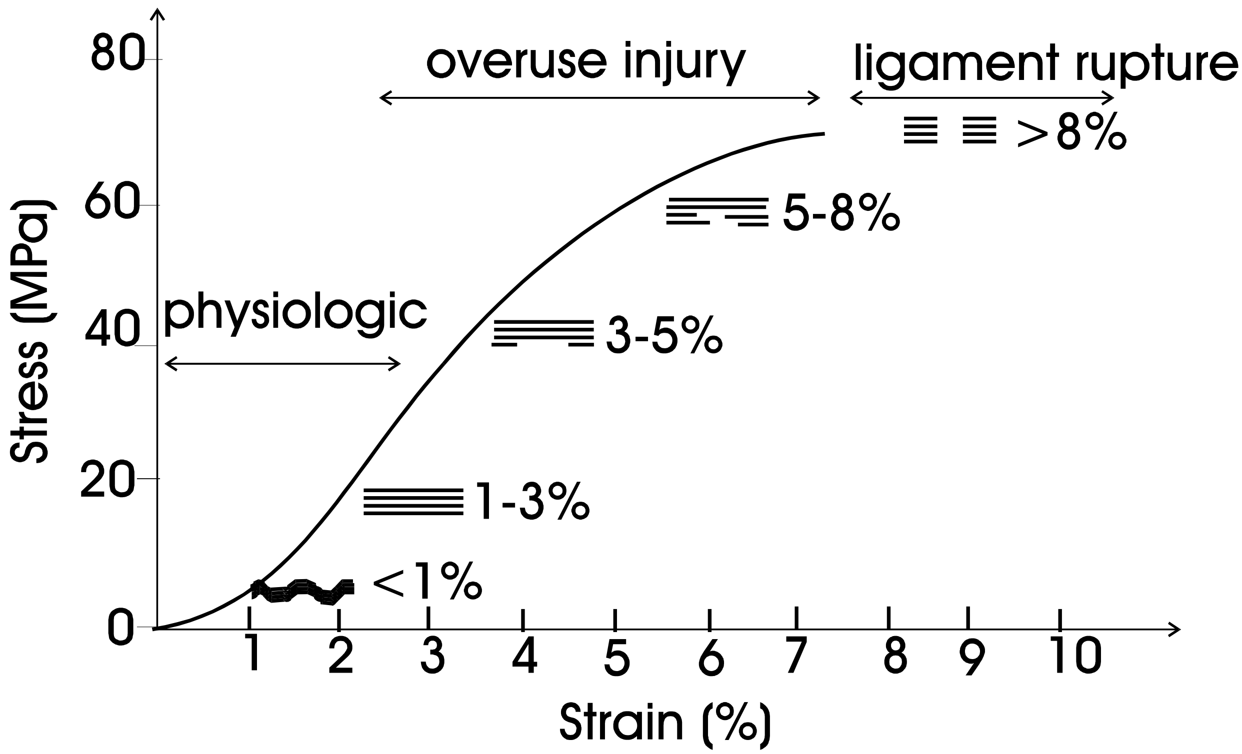


Figure 4 -A stress-strain curve illustrating the relationship between changes in the collagen "crimp" pattern, or stretch, and ligament mechanical properties is shown. Increases in ligament strain in the "toe region" of the curve results in straightening of the "crimp" pattern. During the linear portion of the curve the collagen fibers are stretched. As the ligament is further strained isolated ligament fibers begin to rupture and if deformation continues, then complete ligament fail occurs. (Modified from Butler, D.L., Grood, E.S., Noyes, F.R., and Zernicke, R.F., Biomechanics of ligaments and tendons. Exercise and Sports Science Reviews, 6, 125-181, 1984.)

Clinically, it is important to know the normal operating conditions of ligaments acting within the body. Such information is needed to relate isolated ligament-bone test data to that of ligaments acting in-vivo. This issue is considered again as more factors related to ligament function are presented throughout this chapter. Presently attention is given to the normal force and deformation levels experienced by ligaments in-vivo. A hypothetical force-elongation curve for the human ACL as postulated by Noyes, Keller, Grood and Butler (1984) is shown in Figure 5. Levels of daily activity are shown along the right vertical axis and hypothetical loading levels for the ACL on the left vertical axis. This curve suggests that during daily activities (such as walking or light jogging) the ACL operates along the "toe region" of the force-elongation curve. It is believed that ligaments are not generally loaded above one-fourth of their ultimate tensile load during these daily activities. The early part of Region 2 is considered the upper operating range of the ACL during strenuous activities as might be experienced during fast cutting or pivoting while running. Loading of the ACL beyond Region 2 results in ligament damage and may be incurred during events like clipping in football, a ski accident, or an incorrect landing during a gymnastics floor exercise.

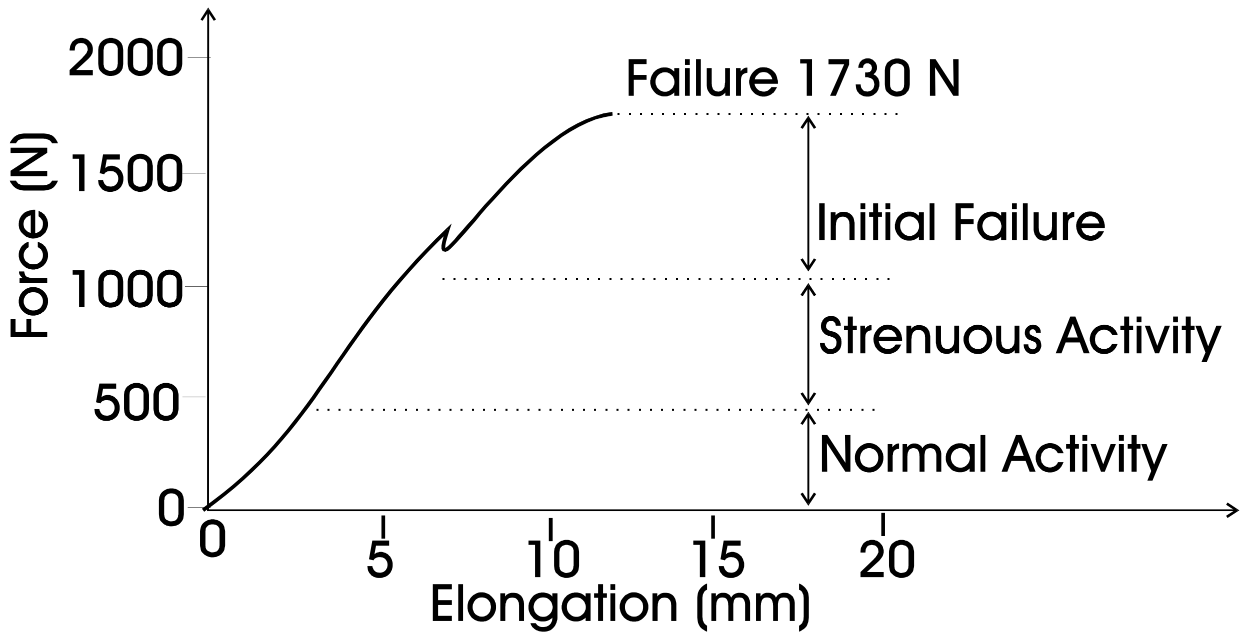


Figure 5 -A hypothetical force-elongation curve for a human ACL-bone complex is illustrated along with daily activities that correspond to specific loading levels. During routine daily activities such as walking and standing, ligaments are loaded to less than one fourth their ultimate tensile load. During strenuous activities such as fast cutting during intense running, loading levels may enter into region 3 where isolated fiber damage takes place. (Modified from Noyes, F.R., Butler, D.L., Grood, E.S., Zernicke, R.F., and Hefzy, M.S., Biomechanical analysis of human ligament grafts used in knee ligament repair and reconstruction. Journal of Bone and Joint Surgery, 66A(3), 344-352, 1984.)

Viscoelastic Properties

Ligaments exhibit significant time and history dependent viscoelastic properties. Ligaments have characteristics of strain rate sensitivity, stress relaxation, creep, and hysteresis as described in Chapter 1. One important aspect to consider is the relationship between the viscoelastic property of strain-rate sensitivity between ligament and bone. Very little difference has been reported in the stress-strain behavior of ligaments subjected to tensile tests varying in strain rate over three decades (Akeson et al., 1984). However, when failure modes are considered for bone-ligament-bone complexes, strain rate becomes a significant factor due to the greater strain rate sensitivity of bone. A hypothetical diagram illustrating the relationship between the probability of failure mode of bone-ligament-bone units and the rate of loading is shown in Figure 6. During slow loading rates, bony avulsion failure is common. As the loading rate increases, the bone becomes stronger than the ligament substance and the ligament substance fails first.

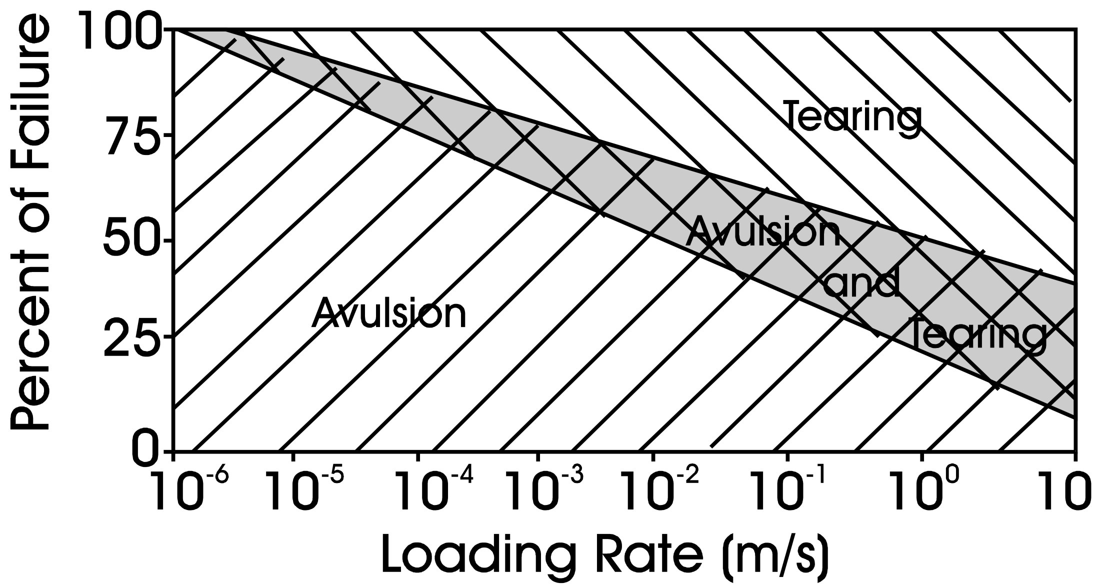


Figure 6 -Experimental results are presented indicating the probability of specific failure modes of bone-ligament-bone units as a function of loading rate. At slower loading rates bony avulsion failures have the greatest probability of occurring. At fast loading rates mid-substance failures are to be expected. (Modified from Crowninshield, R.D. and Pope, M.H., The strength and failure characteristics of rat medial collateral ligaments. Journal of Trauma, 16(2), 99-105, 1976)

Ligaments are subjected to a wide variety of loading conditions during daily activities that affect their mechanical properties. The time-dependent behavior of ligaments may be important during a variety of daily cyclic activities such as walking, running, cycling and other forms of exercise. Figure 7 illustrates ligament softening, a decrease in peak loads occurring during cyclic testing of ligaments to constant strain and at constant strain rate (Woo, Gomez, Woo, and Akeson, 1982). Also, ligament deformation increases slightly during early cycles to a constant load (Woo et al., 1982). Similar results have been observed for intact joints.

A commercial knee laxity testing device was used to quantify anterior and posterior laxity in basketball players after 90 minutes of practice and in runners after a 10 kilometer race. Anterior/posterior laxity increased by more than 18 percent in both cases (Steiner, Grana, Chillag, and Schelberg-Karnes, 1986). In this same study it was demonstrated that muscle relaxation was not a factor in the laxity measurements since knee laxity did not vary appreciably for normal knees before and during general anesthesia. The implications of ligament softening to athletic performance are not yet known.

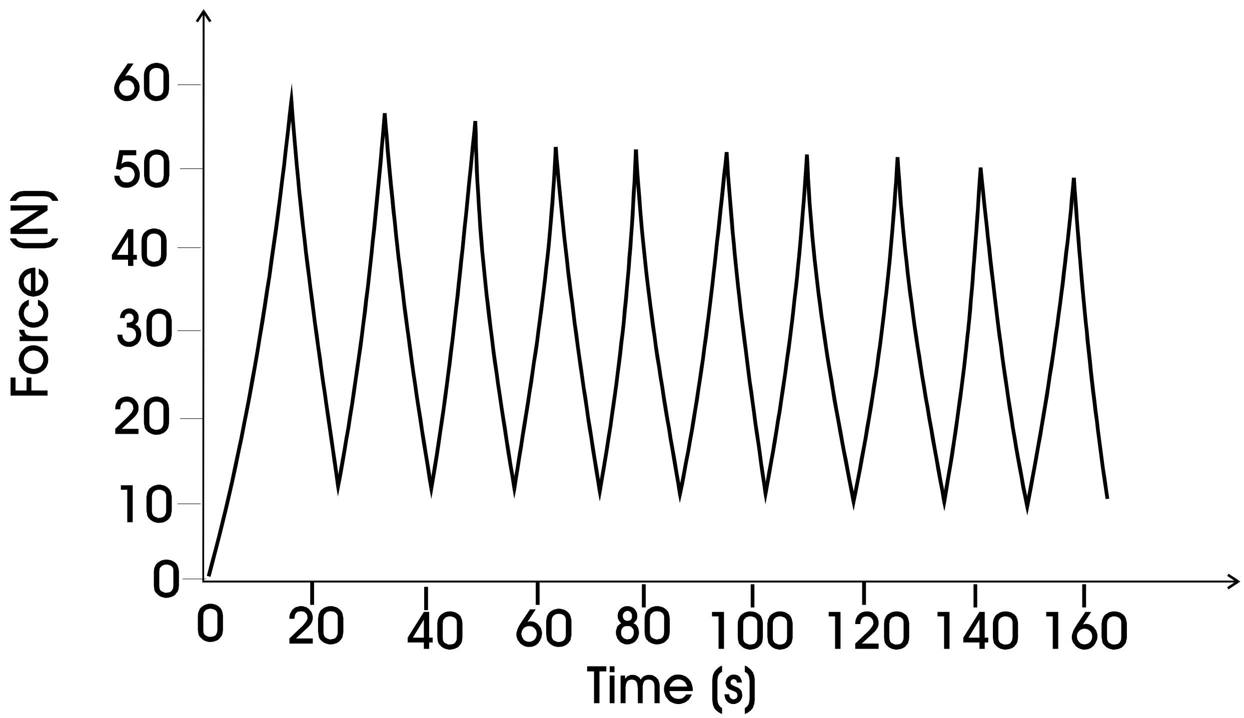


Figure 7 -Shown is a typical ligament response to cyclic tensile loading and unloading. Peak loads decrease with each cycle indicating ligament softening.

Temperature Sensitivity

Ligaments are temperature sensitive with peak stresses increasing with decreased temperature. Bone-ligament-bone preparations tested cyclically at 21 degrees Centigrade show 30 percent greater peak loads than the same preparation tested at 37 degrees Centigrade (Akeson et al., 1984). In-vivo, it has been suggested that the temperature of superficial tissues may be within 2 degrees of the skin temperature which can be 10 degrees lower than body temperature. Temperature effects may also become important under pathological conditions such as trauma, infection, or inflammation. Certainly this factor is important to consider when conducting soft tissue experiments or comparing data from tests performed at different temperatures.

Correlation Between Structure and Function

The "crimp pattern" and the interaction and cross-linking of elastic, reticular, and collagen fibers is 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 increase 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.

A second 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.

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 transition areas are less susceptible to disruption than the extremes on either side (bone or peri-insertional ligament substance).

Individual ligaments act synergistically with other ligaments, bones, and muscles to maintain joint integrity. To understand the functional role of ligaments, all structures of the joint must be considered. Ligaments will alter their structure and mechanical properties in response to environmental changes. All these things need to be considered prior to the development of training, injury treatment, or rehabilitation programs.

Failure Mechanisms

The mechanisms of ligamentous injury may be quite different depending on the conditions during which the injury was incurred. For example a skier quickly twisting his knee during a fall may rupture the mid-substance of both the MCL and ACL, while a football player slowly having his knee bent inward under the weight of several tacklers may experience a bony avulsion of the MCL.

Many factors have been found to influence the failure properties of ligamentous tissue and ligament-bone units. Three principal failure modes have been observed in ligament-bone preparations (Butler et al., 1978). The first type is a ligamentous failure, characteristic of fast loading rate failures. This failure mode results in a "mop end" appearance of the disrupted ligament ends. The bundles of fibers fail at different locations due to shear and tensile mechanisms between fibers.

The second mode of failure is bony avulsion fracture, which occurs during slow loading rate failures. Failure occurs through cancellous bone beneath the insertion site (Butler et al., 1978).

The third failure mode is cleavage (or pull-out) at the ligament-bone interface. This mode of failure is less common than the first two due to the efficient force dissipation that occurs through the insertional zone. When a failure of this type does occur, it generally occurs through the mineralized fibrocartilage (Butler et al., 1978).

Scanning electron microscopy has been used to evaluate the microstructural damage created in bone-ACL-bone preparations loaded to one-half their normal failure force (Grood, Noyes, Butler, Suntay, 1981). Electron-micrographs of these ligaments revealed that submaximal loading disrupted some fibers. In addition, some collagen fibers lost their wavy appearance suggesting that permanent deformation had occurred. The maximum loads utilized in the testing by Grood et al. (1981) were larger than those suggested to be the upper limit under daily physiological conditions (one-fourth the ultimate tensile load). However, results from their study suggest that ligaments may continually experience microstructural damage during strenuous activities. This may be an important factor contributing to overuse injuries. Injuries of this type occur when an individual does not allow enough "reduced activity" time between exercise bouts to allow tissues to repair themselves. As a consequence, the tissue continues to be damaged and weakened, resulting in debilitating pain.

Effects of Aging

Numerous studies have examined the relationship between age related processes and the structural and mechanical properties of collagenous tissues (Elliott, 1965; Benedict, Walker, and Harris, 1968; Diamant et al., 1972; Vogel, 1974; Hall, 1976; Noyes and Grood, 1976; Tipton, Matthes, and Martin, 1978; Woo, Orlando, Gomez, Frank, and Akeson, 1986). However, very few have dealt specifically with ligaments (Noyes and Grood, 1976; Tipton et al., 1978; Woo et al., 1986) and even fewer with bone-ligament complexes (Woo et al., 1986). There are two age related processes, maturation and aging, that affects bone-ligament properties.

During maturation the structure and mechanical properties of collagenous tissues change. Increases in collagen cross-linking, collagen glycosaminoglycan, and collagen-water ratios have been observed (Butler et al., 1978; Menard and Stanish, 1989). In addition, an increase in mechanical properties occurs during maturation. In studies of young and mature rabbits, Woo et al. (1986) showed that the mechanical properties of bone-ligament units of younger rabbits are inferior to those of mature rabbits. During the early growth period the strength of the bone-ligament complex rises quickly. Thereafter strength changes occur at a slower rate. The changes in mechanical properties that occur during maturation are certainly related to the changes in cross-linking and water content that occur during this same period. The stabilization of collagen with maturity enhances tissue strength while the loss of water and elastin reduces tissue plasticity (Menard and Stanish, 1989).

Also of interest for understanding the effects that maturation has on bone-ligament units is the difference in the mode of failure. The bone-ligament junction of the 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. A hypothetical curve illustrating the asynchronous rates of maturation for both the bone-ligament junction, and the MCL substance is shown in Figure 8. Ligament substance matures earlier than the ligament-bone junction (Woo et al., 1986). However, with maturity the bone-ligament junction strength increases to a value above that reached by the ligament substance.

It is very difficult to distinguish aging effects from effects created by other factors such as disease, or changes in activity levels. A study of human femur-ACL-tibia complexes revealed that specimens from older donors show lower stiffness and smaller force and deformation at failure than other specimens (Noyes and Grood, 1976). A higher incidence of bony avulsion failures for specimens fifty years and older compared to others was also revealed. 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. Thus the alterations observed in connective tissue structure manifest themselves as changes in tissue mechanical properties (Menard and Stanish, 1989). It is difficult to say with certainty that the observed structural and mechanical changes are strictly the result of aging.

In summary, during maturation, there appears to be an asynchronous rate of strength increase between ligament-bone substance and ligament substance, with the ligament having the faster rate. The slower rate of strength increase of the bone-ligament unit may be due to the more complex structure of the bone-ligament interface hence requiring more time to develop. It is difficult to speculate as to the exact effects that aging have on bone-ligament units once maturity has been reached. Studies suggest a decrease in stiffness and smaller force and deformation at failure with increased age. However, these results may reflect changes in other factors other than age. It must be emphasized that the effects of aging are highly individual and depend on factors such as genetics, past disease, and lifestyle. It is likely the observed alterations in the bone-ligament complex that occur in later years in life are due to a combination of aging factors, disease, and decreased physical activity.

For the young athlete, maturation translates into less compliant joint structures. The ligaments stabilizing joints become stiffer resulting in a reduction in overall joint compliance. The increased bone-ligament junction strength that occurs with maturation tends to equalize the strength between this region and the ligament substance. For the older athlete, further reductions in tissue compliance and hence joint compliance can be expected. It has been estimated that regular exercise may retard the physiologic decline associated with aging as much as 50 percent (Menard and Stanish, 1989).

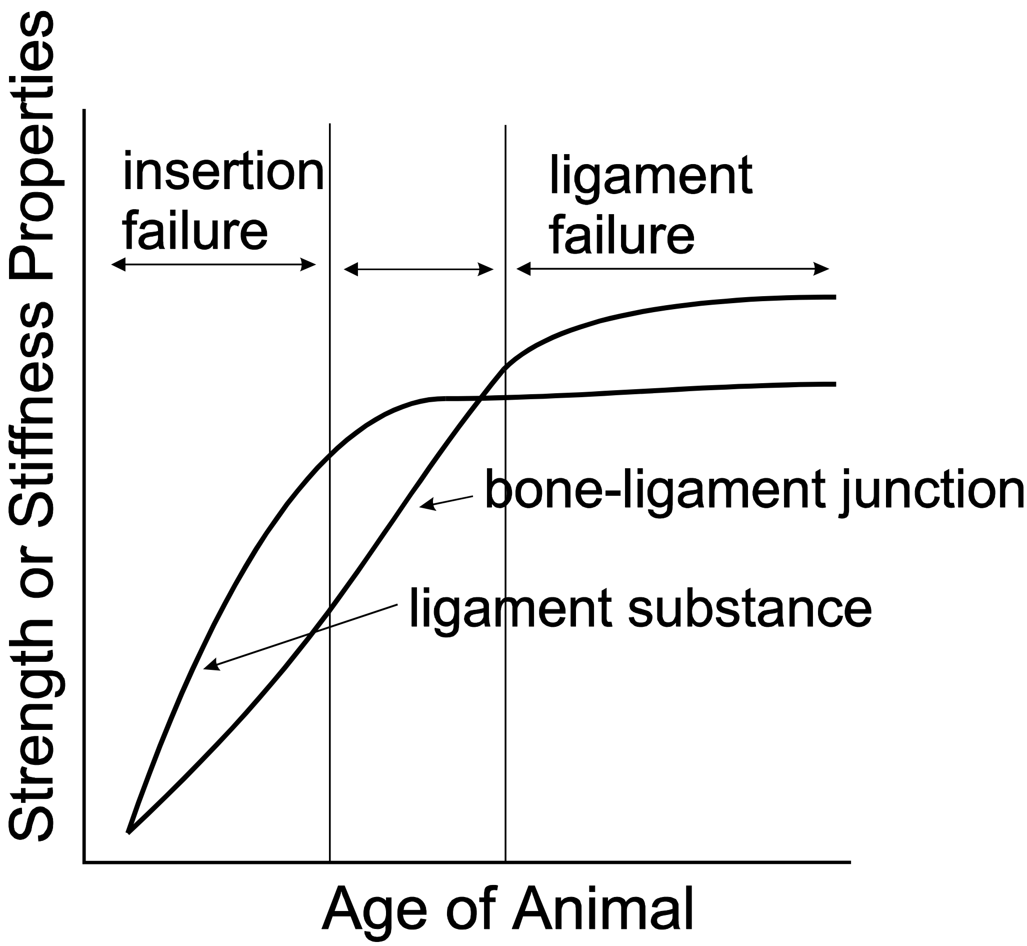


Figure 8 -A hypothetical diagram relating the asynchronous rates of maturation for both the ligament substance and the ligament-tibia junction in terms of strength and stiffness characteristics is shown. The strength of the ligament substance initially increases faster than that of the bone-ligament junction. The strength of the bone-ligament junction eventually surpasses that of the ligament substance. (Modified from Woo, S.L-Y., Orlando, C.A. Gomez, M.A., Frank, C.B., and Akeson, W.H., Tensile properties of the medial collateral ligament as a function of age. Journal of Orthopaedic Research, 4(2), 133-141, 1986.)

Exercise and Disuse

Professional and recreational athletes experience periods of increased and decreased physical activity. These cycles are, in part, dictated by competition schedules and injuries. Alterations in activity levels can have profound effects on the structural and mechanical properties of ligaments.

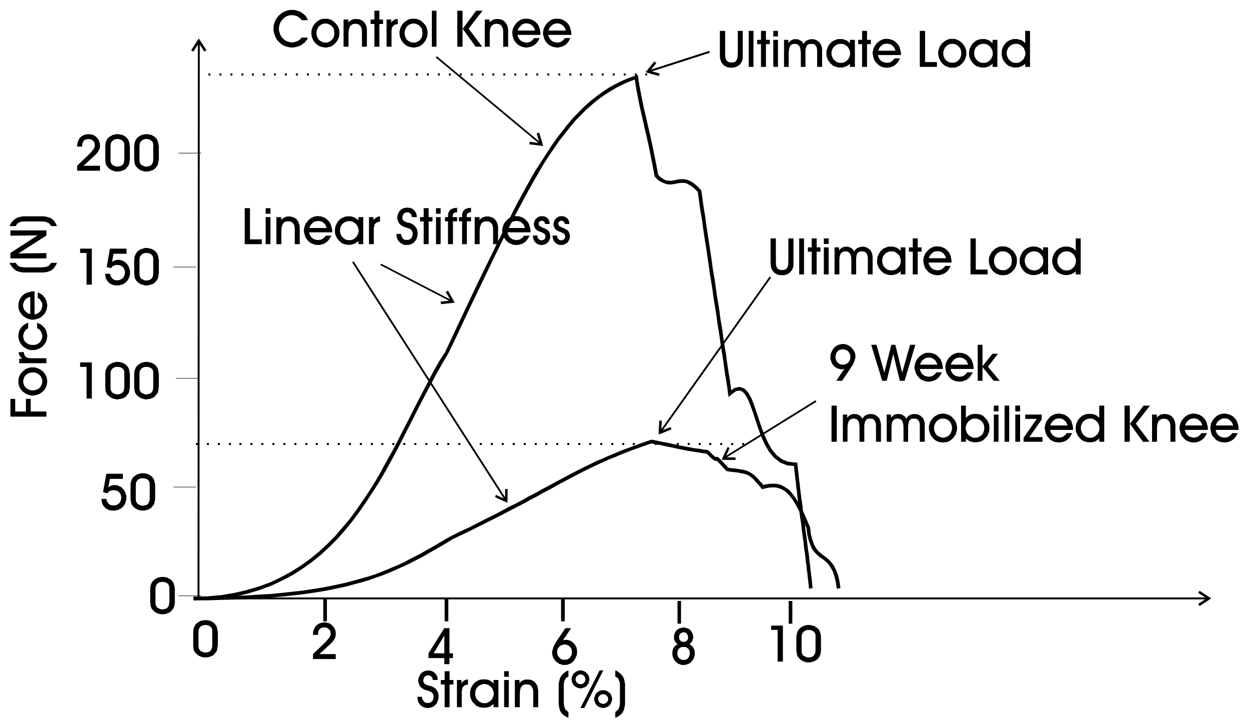


Figure 9 - Effects of 9 weeks of joint immobilization on the rabbit medial collateral ligament. Ultimate failure load decreases along with tissue stiffness and the energy absorbed prior to failure.

Several studies have been conducted to determine the effects that joint immobilization and exercise have on the structure and mechanical characteristics of bone-ligament complexes (Tipton, James, Mergner, and Tcheng, 1970; Laros, Tipton, and Cooper, 1971; Noyes, Torvik, Hyde, and DeLucas, 1974; Tipton, Matthes, Maynard, and Carney, 1975; Akeson, Amiel, Mechanic, Woo, Harwood, and Hamer, 1977; Noyes, 1977; Akeson et al., 1984; Woo et al., 1982; Woo, Gomez, Sites, Newton, Orlando, and Akeson, 1987; Wilson and Dahners, 1988). The effects of stress deprivation, induced by bed rest or joint immobilization, on intra-articular and extra-articular ligaments are profound (see Figure 9). On a gross scale, ligaments appear less glistening and grainier on dissection. Histologically there is an increased randomness of fibers, cells, and matrix organization (Akeson et al., 1984). Some general changes that occur in bone-ligament units due to immobilization include, (1) a reduction in the failure strength and energy absorption at failure (Tipton et al., 1970; Laros et al., 1971; Noyes et al., 1974; Tipton et al., 1975; Akeson et al., 1977; Noyes, 1977; Akeson et al., 1984; Woo et al., 1982; Woo et al., 1987), (2) bone resorption at the insertion sites (Laros et al., 1971; Noyes et al., 1974; Tipton et al., 1975; Akeson et al., 1977; Noyes, 1977; Akeson et al., 1984; Woo et al., 1982; Woo et al., 1987), and (3) increased intramolecular cross-links (Akeson et al., 1977; Akeson, Amiel, and

Woo, 1980). In general these changes translate into increased joint stiffness and increased susceptibility to ligament or bone-ligament damage.

Joint immobilization, using casting or bracing, is often prescribed for an individual after an injury has been sustained. However, joint stiffness may result from this treatment as a result of ligament contracture (shortening of the structure), periarticular tissue adhesions, and decreased joint lubricity. Joint stiffness is of clinical interest for obvious reasons. As discussed previously, ligaments become less stiff with immobilization, which is contrary to what might be postulated from observed increases in joint stiffness after immobilization. However, if ligament contracture takes place while the joint is immobilized, then for a given joint angle that ligament will function on a stiffer portion of the force-elongation curve. So, though the general stiffness of the ligament has decreased, its contribution to joint stiffness may actually have increased. Wilson and Dahners (1988) demonstrated that ligaments do in fact contract during stress deprivation. It is difficult to determine exactly where on the force-elongation curve a ligament operates in-vivo. Without such information it is difficult to assess how changes in ligament properties affect joint stiffness.

Of clinical relevance is the rate at which the properties of a bone-ligament complex return to normal values after mobilizing a previously immobilized joint. Several studies have shown that considerably more time is needed to regain original strength than was taken to deteriorate it (Noyes et al., 1974; Tipton et al., 1975; Noyes, 1977). Even after five months of recovery following eight weeks of immobilization in which the joint was pinned, rhesus monkey ACLs showed only partial recovery toward normal failure strengths, the maximum load at failure being 79 percent of controls (Noyes et al., 1974; Noyes, 1977). The rates of recovery for ligament substances and bone-ligament junctions vary, with the insertion zones having slower recovery rates. As stated in the section on aging, these different rates may be due to the more complex structure of the insertional zone.

The effects of exercise on ligaments have been less well defined due to inconsistent exercise protocols and methodological problems associated with quantifying physiologic proof of exercise intensity. Increases in ligament mass, fiber diameter, and cross-sectional area have been reported from long term exercise programs (Tipton et al., 1970; Tipton et al., 1975). The absolute separation force of bone-ligament-bone complexes usually is increased by exercise (Tipton et al., 1970; Tipton et al., 1975). Most researchers have attributed this improvement to changes in the insertion zones rather than changes in the ligament substance.

A hypothetical curve suggesting the relationship between duration of loading and the mechanical properties of ligaments is shown in Figure 10. Bone-ligament alterations due to exercise are less dramatic than those caused by stress deprivation. However, due to the slow collagen turnover rate, the resulting effects are probably long lasting in an animal that maintains normal activity. As with stress deprivation, the extent to which exercise effects a bone-ligament complex depends on several factors including age, sex, species, nutrition, hormones, and mode of exercise.

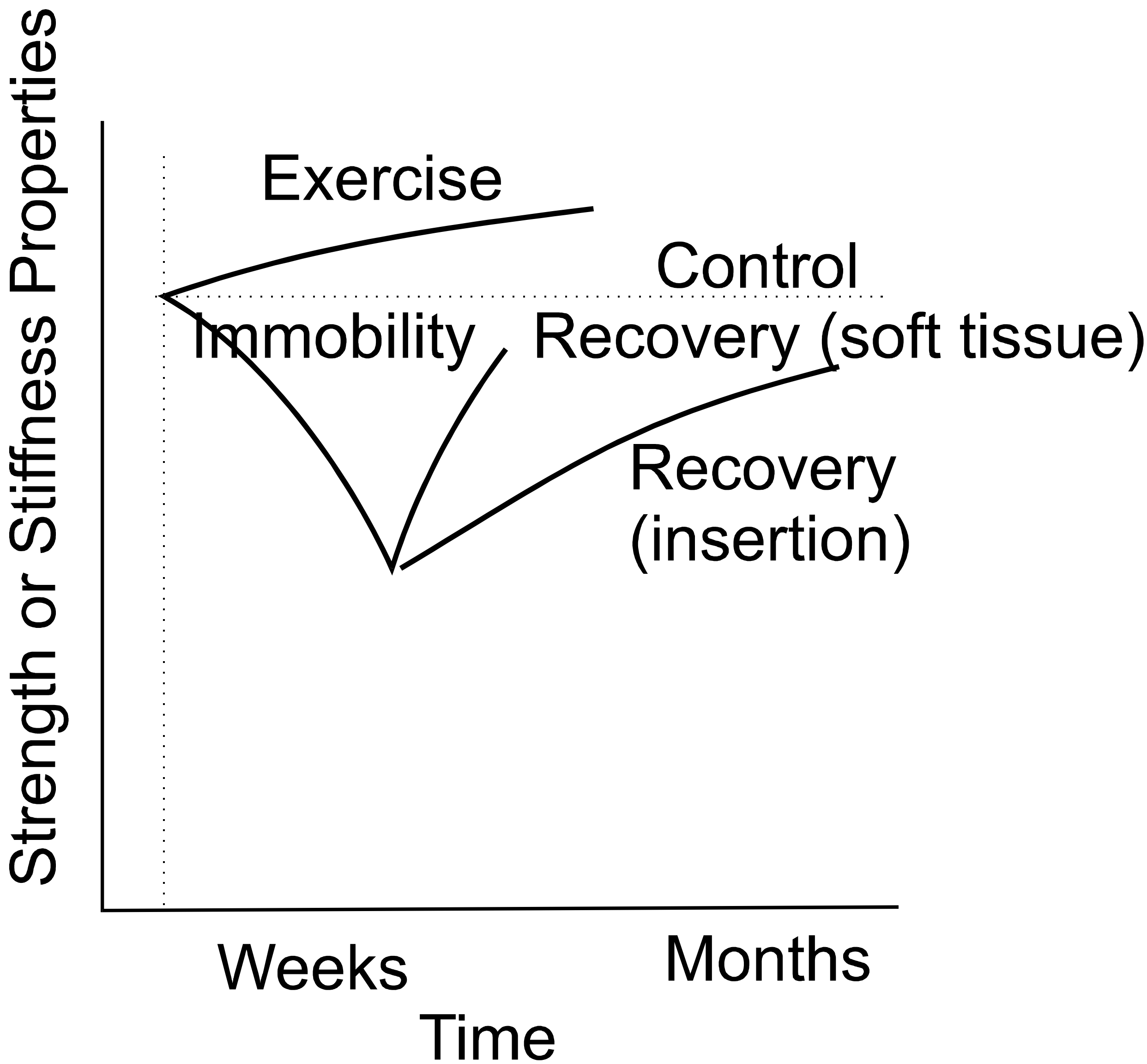


Figure 10 -A hypothetical curve demonstrating the nonlinear relationship between soft tissue mechanical properties (and/or mass) and the level (and/or duration) of tissue stress and strain is shown. During stress deprivation or immobilization the mechanical properties can decrease rapidly. Increased exercise levels above that of daily activities causes the mechanical properties to increase, but less dramatically than the decrease observed during stress deprivation. (Modified from Woo, S.L-Y., Gomez, M.A., Sites, T.J. et al., The biomechanical and morphological changes in the medical collateral ligament of the rabbit after immobilization and remobilization. Journal of Bone and Joint Surgery, 69A, 1200-1211, 1987.)

Trauma

Ligament trauma occurs when loading of the collagenous tissue causes microscopic or macroscopic structural damage. Microscopic damage results when a few isolated fibers are torn or stretched. Macroscopic damage occurs when major portions of the tissue are disrupted or completely torn. Injury to collagenous tissue invokes an inflammatory and healing response.

The initial response of hemorrhage and inflammation usually dominate for the first one to seven days following trauma. Following the initial response, a proliferation of connective tissue cells takes place with peak levels reached two to three weeks after injury (Akeson et al., 1984). Three to six weeks after injury the number and size of fibroblasts decrease and their nuclei begin to align with the long axis of the ligament. Remodeling and cell alignment continue thereafter (Frank, Woo, Amiel, Harwood, Gomez, and Akeson, 1983). Progressive changes in cell number, size, distribution, and orientation occur throughout stages of inflammation, proliferation, and remodeling. Variability in the extent and duration of these changes depends on a number of systematic factors (age, nutrition, hormones) and local factors (extent of trauma, blood supply, infection, mechanical stress, temperature, chemical environment).

From a clinical perspective, these processes are indicative of the amount of time required for a ligament to heal. Remodeling and cell alignment do not begin until three to six weeks post trauma. It may be several more months before the remodeling process is complete. Ligament healing is a slow process and should be recognized and treated as such.

Ligament trauma may or may not be perceived by an individual and may or may not affect joint function. Microstructural damage and remodeling often take place without being recognized by the individual. However, repeated minor traumas may compound if sufficient healing time is not given between injuries. Under these conditions inflammation and pain may become noticeable and hinder joint function. Clinically these conditions show up as lateral epicondylitis (tennis elbow) and other collagenous overload injuries.

Drugs

Frequently, corticosteroids are prescribed for the treatment of inflammation. Various studies have investigated the effects of steroid injections on the mechanical properties of ligaments (Hollander, 1972 and 1974; Mankin, 1974; Noyes, 1977). In a study by Noyes (1977), multiple intra-articular injections of methylprednisolone acetate, a corticosteroid, were administered into the knees of rhesus monkeys. Animals were assigned to a control group, a sham group in which saline was substituted for the steroid, a 20 mg (large dosage) group studied at 6 and 15 weeks, or a 4 mg group (small dosage) studied at 15 weeks following injection. Femur-ACL-Tibia preparations were tested to failure at a rapid rate. Statistically significant decreases in load at failure occurred in both the large dosage (11 percent at 6 weeks, 20 percent at 15 weeks) and small dosage (9 percent at 15 weeks) groups compared to controls. An 11 percent decrease in energy to failure was present at 15 weeks in the large dosage group, and an 8 percent decline occurred at 8 weeks in the small dosage group. Further work in this area reported by Noyes, Keller, Grood, and Butler (1984) indicate that the deleterious effects of single injections of methylprednisolone acetate are greater than those for multiple intra-articular injections. Results from these studies suggest that single intraligamentous or multiple intra-articular steroid injections have the potential to cause deleterious effects on the mechanical properties of ligaments. Corticosteroids are known to have inhibitory effects on the synthesis of glycosaminoglycans, proteins, and collagen. However, the specific mechanisms responsible for the observed deleterious effects are not known.

Clinical Considerations

There was a time when the incidence of ligamentous injuries was small. However, in today's health conscious society, with more and more people partaking in fitness programs and strenuous sports, such is not the case. Ligament injuries are on the rise and the number of clinical repairs and replacements is considerable. Clinically there are four major areas where biomechanics have contributed to the treatment of ligament injuries. These areas include injury diagnosis, ligament repair, rehabilitation, and joint bracing. Advances that have taken place in each of these areas are the result of coordinated interdisciplinary efforts.

Diagnosis

Determination of the nature and degree of ligament injury is one of the most important factors in successful treatment of the damaged structure. Diagnosis of a damaged ligament should begin with a history of the trauma. The activity during which the injury occurred, the sensation of a "pop", swelling, and joint instability are all indicators of the structures involved and the extent of injury. Beyond this step physical examination of the injured joint is required.

Numerous test protocols have been developed for diagnosing knee ligament injuries (anterior drawer, Lachman, MacIntosh pivot shift, Losse, Slocum, "jerk", and recurvatum). For a detailed discussion of these tests and others refer to the articles by Hughston, Andrews, Cross, and Moschi (1976); Torg, Conrad, and Kalen (1976); Galway and MacIntosh (1980); and Losse (1985). In short, during these tests a clinician attempts to produce a displacement of the joint in a given direction by applying a prescribed force. The resulting displacement indicates the joint structures damaged. For example, during an anterior drawer test, the patients knee is flexed to 90 degrees, the tibia is held in neutral rotation, and the clinician attempts to produce an anterior displacement of the tibia relative to the femur. The amount of displacement indicates whether the ACL has been damaged.

To understand the significance of these clinical procedures one must understand the role various structures play to stabilize the joint. Studies have been conducted in which intact cadaver knees were loaded or displaced a specific amount in specially designed fixtures. Individual ligaments were then cut and the change in force or joint position recorded (Nielson, Ovesen, and Rasmussen, 1984; Noyes et al., 1984). Resulting changes in the load or joint position suggest the role the cut structure played in resisting the applied load or displacement. For example, if an initial anterior displacement is applied to the tibia with the femur held fixed and all ligaments of the knee intact, a restraining force can be measured. If the ACL is subsequently cut the restraining force will decrease. The amount this force decreases is indicative of the contribution the ACL makes to stabilize the joint in the configuration tested.

Several interesting findings have been reported from work similar to that described above. The knee has primary and secondary restraints (Noyes et al. 1984). The ACL is the primary restraint of anterior drawer when the knee is flexed between 30 and 90 degrees. The PCL is the primary restraint of posterior drawer at similar knee angles. Nielson et al. (1984) propose that injuries caused by external rotation of the tibia, as might occur in skiing when the inside of a ski edge hits a bump and causes the tibia to rotate externally, damage the MCL first followed by the ACL, whereas internal rotation, as might occur as a football player attempts to pivot on one foot while that foot is held fixed to the ground, injures the ACL first followed by the MCL. This information indicates the diagnostic usefulness of obtaining the history of the trauma; specific movements are likely to damage specific structures. These data also suggests that protocols or test devices that examine the amount of anterior tibial displacement as a function of knee angle can be useful in determining ACL insufficiency.

Devices to quantify knee laxity have been devised to assist in the diagnosis of ligament injuries. These devices generally apply some force or torque to the joint. The resulting linear or angular displacement is recorded, compared to "normals" or the contralateral joint, and used to evaluate the extent of specific ligament damage. Two such knee devices are the Stryker Knee Laxity Tester (KLT) and the Medmetric KT-1000 or 2000. The usefulness of these devices may be limited by the fact that under the low loads (90 N) generally applied when using these devices, secondary stabilizers may be sufficient to prevent joint displacement and hence mask the injury to the primary stabilizer. However support for use of the Stryker KLT has been given by Boniface, Fu, and Ilkhanipour (1986), and both the Stryker KLT and Medmetric KT-2000 by Daniel, Malcom, Losse, Stone, Sachs, and Burks (1985). The support stated above is based on the successful use of these devices in predicting ACL damage as indicated by corresponding arthroscopic studies. In most cases where an ACL tear was present, the laxity measurements performed in the clinic were suggestive of a pathologic anterior laxity.

In addition to passive knee laxity tests, some researcher advocate dynamic tests (Tibone, Antich, Fanton, Moynes, and Perry, 1986; Tegner, Lysholm, Lysholm, and Gillquist, 1986). During dynamic tests such as walking, stair climbing, one legged hopping, running and cutting, parameters such as muscle activity recorded using electrodes, and joint forces determined using a force plate, cinematography, and inverse dynamics, may indicate abnormal responses and hence be useful for diagnosing injuries. Information from dynamic tests may also be beneficial for determining when an injured knee is sufficiently rehabilitated to withstand strenuous activities.

Treatments

Damaged ligaments may be treated conservatively or surgically by primary suture, augmentation, and synthetic or allograft replacement. For a review of the history of ACL repair techniques refer to Burnett and Fowler (1985).

The selection of treatment procedure depends on the needs of the individual. A conservative approach may be adequate for mildly active individuals who can compensate for the damaged tissue with other ligaments, muscles, or braces. However, as shown in a study by McDaniel and Dameron (1980) untreated ruptures of the ACL may result in increased anterior laxity, rotatory instability, and meniscal tears. In addition, osteoarthritis may result from long term joint laxity.

Some form of ligament repair may be necessary for individuals with severe joint instability or those desiring to participate in activities placing the injured ligaments and joint at further risk. In these cases, ligament repair is performed with the goal of returning the joint to near normal functional integrity. To do this requires an understanding of the mechanical properties of the normal ligament, the repair process and its effect on ligament mechanical properties, immune responses to implant materials, joint structure interactions, and the effect of attachment geometry and ligament tension on joint motion and stability.

A ligament substitute should possess similar mechanical characteristics to that of the original tissue. Ligament mechanical properties obtained from testing bone-ligament-bone preparations have been used to assess the adequacy of possible ligament substitutes (allograft or synthetic). Noyes et al. (1984) compared the ultimate load of the ACL with various allografts that have been used to replace a ruptured ACL. The medial third of the bone-patella-bone tendon has been found to have an ultimate load 168 percent that of the ACL. The quadriceps-patella-bone tendon, in comparison, has an ultimate load 36 percent that of the ACL. Both of these structures have been used as ACL replacements. However, from simple tensile test experiments it is evident that the quadriceps-patella-bone tendon is inadequate as an ACL replacement, whereas the medial third of the bone-patella-bone tendon has a greater likelihood of success.

Ultra-high-molecular-weight polyethylene (UHMWPE) was once used as an ACL replacement. However, UHMWPE was shown to possess poor creep characteristics compared to a normal ACL (Chen and Black, 1980). Therefore, as the artificial ligament was repeatedly stressed it lengthened and hence lost its joint stabilizing effect. UHMWPE implants demonstrated a high failure rate clinically, and were removed from the market. Early biomechanical testing may have prevented the use of these implants clinically.

Carbon fiber is another material that has been used in the construction of synthetic ligaments. Studies of the use of these implants have shown that carbon fibers fragment within the body and subsequently migrate to other areas (Zoltan, Reinecke, and Indelicato, 1988; Amis, Kempson, Campbell, and Miller, 1988). The long term effects of these fragments and increased carbon concentrations are not yet known.

Polyester ligaments have been shown to have similar mechanical properties to the human ACL (Amis et al., 1988). ACLs replaced using this material have shown good success clinically, but long term data is still being compiled.

These few examples illustrate the importance of properly testing and evaluating ligament substitutes before allowing their use clinically. Unfortunately, rigorous biomechanical testing of replacement ligaments has often taken place after their clinical use has begun.

Other important factors contributing to the successful replacement of a ligament are the attachment locations and preload. Errors in either locating the ligament attachment sites, or preloading the ligament will cause joint stiffness or laxity. Arms, Pope, Johnson, Fischer, Arvidsson, and Eriksson (1984) showed that misplacement of ligament attachment sites causes the ligament to strain more during normal joint motion than if it were located properly. Incorrect ligament attachment may lead to excessive ligament strain and early failure. Likewise, even if a ligament substitute is attached in the proper location, if the preload is too large, then the ligament will limit the joint range of motion causing excessive loading and possibly failure. If the preload is too small and the ligament is slack, then the ligament will not lend proper support to the joint.

Suture of the original ligament is the least used surgical procedure. If the attachment sites are intact, then the major issues are uniting the ruptured ends and restoring proper tension. However these are not simple issues to resolve. Ligaments repair themselves by scar formation followed by cell differentiation and alignment. The scar material, which has different material properties then ligament, may stretch during healing and result in a slack ligament. This may lead to joint laxity and conditions causing cartilage degeneration. Early controlled joint mobilization appears to enhance the healing rate of sutured ligaments by increasing cellularity, collagen content, and tensile strength (Noyes et al., 1984; Amis et al., 1988). Whether early mobilization reduces ligament stretching during healing needs further investigation.

Rehabilitation

Biomechanical study of knee ligaments has contributed significantly to the development of rehabilitation protocols. Arms et al. (1984) placed a strain gage on the anterio-medial aspect of the ACL of cadavers and measured the change in strain during knee flexion. During passive knee flexion the ACL is minimally strained when the knee is flexed between 30 and 45 degrees (0 degrees being full extension). Simulated quadriceps muscle forces increased the strain in the ACL above that of passive knee flexion for angles less than 60 degrees of flexion and decreased the strain for larger flexion angles. Therefore, according to these data, to reduce ACL strain after reconstruction or replacement, knee motion should be allowed only within the range 30-45 degrees of flexion during rehabilitation programs calling for no muscle activity. If isometric quadriceps muscle contractions are used during rehabilitation, to prevent muscle atrophy, then motion should be limited to angles greater than 45 degrees of flexion.

Bracing

There are three primary brace categories, (1) prophylactic, (2) rehabilitative, and (3) functional. Prophylactic braces are intended as preventative devices and hence are used prior to any ligament or joint injury. Rehabilitative braces are designed to allow a limited range of joint motion during recovery from a joint injury or a surgical procedure. Functional braces are intended to assist in stabilizing joints that may be prone to injury due to joint laxity or ligament weakness. The efficacy of these braces, however, is controversial.

Prophylactic knee braces consist of either a lateral bar or both medial and lateral bars suspended by straps and/or taping. They are intended to prevent or reduce the severity of injuries by reducing medial joint opening. Yet clinical studies of knee injuries sustained by football players wearing and not wearing prophylactic braces indicate that prophylactic braces do not reduce the incidence of knee injury, and in some cases increase the incidence (Hewson, Mendini, and Wang, 1986; Rovere, Haupt, and Yates, 1987; Garrick and Requa, 1987). Increased injuries may, in fact, be a result of brace design. Some braces have lateral bars that may act as fulcrums causing greater displacement of the tibia relative to the femur during certain loading conditions. The mechanics of the brace may provide only a partial explanation for increased injury. Braces may give athletes a false sense of security causing them to act more aggressively than they would without the braces.

In addition to the clinical investigations of braces, biomechanical studies have been conducted. Paulos, France, Jayaraman, Abbot, and Rosenberg (1987); Baker, VanHanswyk, Orthotist, Bogosian, Werner, and Murphy (1987); and Baker, VanHanswyk, Orthotist, Bogosian, Werner (1989) looked at static and dynamic response characteristics of prophylactic braces on cadaver knees. They reported that prophylactic braces provide little if any protective effect. In general, there is no evidence to support the use of prophylactic braces at this time (Millet and Drez, 1988).

Rehabilitative knee braces are those designed to allow protected motion of injured knees treated operatively or nonoperatively by allowing controlled joint motion that may be beneficial during ligament healing. Rehabilitative knee braces consist of hinges, brace arms, and thigh and calf enclosures. Eight commercially available rehabilitative knee braces were compared using a mechanical surrogate leg (Cawley, France, and Paulos, 1989). Most of the braces tested reduced both lower limb translations and rotations compared to the non-braced limb. Although these braces appear to be effective in various treatment programs, it is notable that these braces provide little anterior/posterior stability and that more knee joint motion may occur than prescribed by the device (due to motion of the knee relative to the brace).

Functional braces are designed to assist or provide stability for unstable joints. There are two basic construction types available. Both types use hinges and posts. Differences exist in whether limb enclosures or straps for suspension are employed. Several studies have been performed to determine the effectiveness of these braces (Beck, Drez, Young, Cannon, and Stone, 1986; Colville, Lee, and Ciullo, 1986; Knutzen, Bates, and Schot, 1987; Wojtys, Goldstein, Redfern, Trier, and Matthews, 1987). Beck et al. (1986) tested several functional braces on ACL deficient knees with commercially available knee laxity testers. It was concluded that braces constructed with the enclosures rather than straps perform better to prevent anterior tibial displacement. It was also noted that as the force increased, the effectiveness of the braces in controlling anterior tibial displacement decreased. Similar results have been obtained by other investigators (Colville et al., 1986; Wojtys et al., 1987). During athletic competition where conditions of high loading exist, the ability of functional braces to control pathologic anterior laxity is minimal (Millet and Drez, 1988).

Knee bracing continues to be a complex and controversial topic. Further research is needed in this area to develop brace designs that may provide the support and flexibility required by normal functional joints.

Summary

Although significant advances have been made in the biology, biochemistry, and mechanics of soft tissue biomechanics, there is still much work to be done. There is limited information available pertaining to 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 can not be made confidently.

Presented in this chapter are current ideas related to ligament structure and function. Ligaments play a significant role in stabilizing joints and specifying joint motion. They are commonly injured in sports and, therefore, an understanding of their form and function is relevant for the exercise/sports scientist. Such an understanding can be fundamental to both design training programs that minimize the risk of ligament damage, and recognize early signs of ligamentous injuries.

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Sample Problems:

1. Describe the general structural organization of ligament and tendon.

2. Draw a knee ligament force-deformation curve and identify on this curve the general range of physiological loading that the ligament might experience during daily activities such as walking, running, strenuous obstacle running, and a fall during skiing in which the knee is twisted.

3. Describe the failure mode that you would expect to observe in the LCL of a child that had a large load slowly applied to the knee to cause it move into a severe varus position. Explain the rationale for your answer.

4. Describe what changes you would expect to observe in the knee ligaments of a person exposed to 6 weeks of leg casting for a bone fracture (both structurally and mechanically).

5. What factors might contribute to a person having considerable joint laxity?