2. Review of Literature

2.1 Introduction

This section provides background information that is pertinent to the dissertation. Gross anatomy of the knee is introduced in terms of human structures, and comparative porcine anatomy is then presented. A description of the function of the anterior cruciate ligament (ACL) follows. Finally, literature related to the incidence, etiology, and treatment of ACL injuries is reviewed.

2.2 Gross Anatomy of the Knee

Gross anatomy involves the study of the structure of the body on a macroscopic level. In this section, gross anatomy of the knee is discussed first in terms of human characteristics. Then, comparative porcine anatomy is presented.

2.2.1 Human Anatomy

Except where noted, the following descriptions of the components of the human knee are based on information from *Gray's Anatomy* [35].

2.2.1.1 Bony Architecture

The knee consists of three bony articulations combining to form one joint (Figure 2.1). Two of these connections are condyloid joints, one occurring between each femoral condyle and the related meniscus and tibial condyle. The third articulation, occurring between the patella and femur, is partially arthrodial.

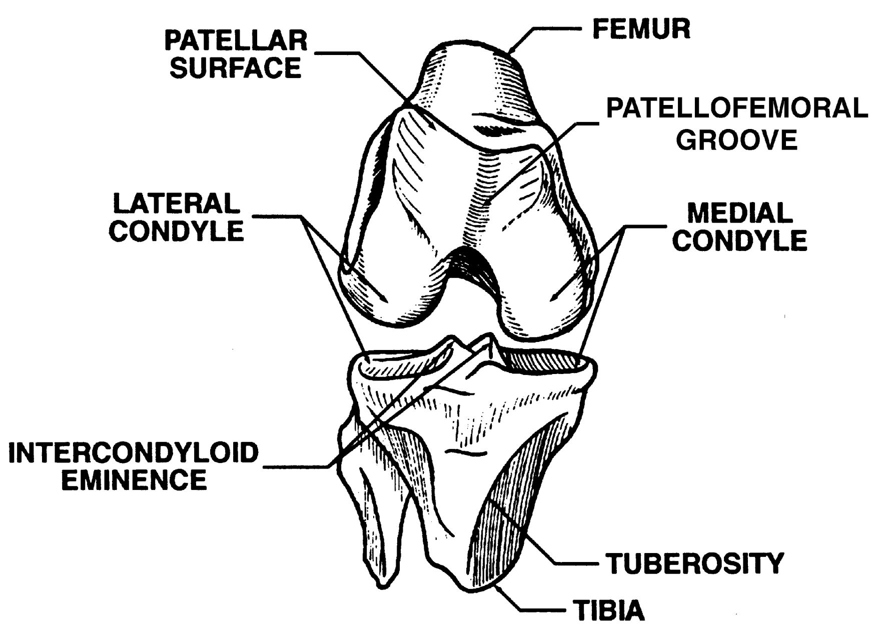

Figure 2.1 Bony Architecture of the Knee

2.2.1.1.1 Femoral Condyles

The inferior end of the femur consists of two oblong, convex eminences known as the femoral condyles. On the anterior side, the condyles are extended anteriorly to form a smooth shallow articular depression called the patellar surface. The central part of the patellar surface is typically referred to as the patellofemoral groove. In the posterior direction, the condyles extend to form a deep fossa known as the intercondylar notch. The lateral condyle is the larger of the two in the antero-posterior and transverse directions, while the medial condyle is the larger in the distal direction. This feature allows the lower surfaces of the condyles to lie in the same horizontal plane when the femur is aligned in its natural oblique position. The axes of the condyles from anterior to posterior edges are not parallel; that of the lateral condyle runs nearly antero-posteriorly, while that of the medial condyle is inclined medially.

2.2.1.1.2 Tibial Condyles

The superior end of the tibia consists of two large eminences known as the tibial condyles. The lateral condyle is nearly circular, while the medial condyle is oval. In the medio-lateral direction, both condyle surfaces are concave. In the antero-posterior direction, however, the medial condyle surface is concave, while the lateral condyle surface is convex. Located between the condyles are two depressions, the anterior and posterior intercondylar fossae, and a spine, the intercondylar eminence. The anterior surfaces of the condyles meet to form a large flat area. At the inferior edge of this area is the tibial tubercle.

2.2.1.1.3 Patella

The patella is a flat, triangular bone located at the front of the knee. The anterior surface is convex and marked with striations. The posterior surface contains a smooth articular area divided into two facets by a vertical ridge, which corresponds to the patellar surface on the femur. Inferior to the articular area is a convex non-articular area. The thick, superior edge of the patella is called the base, and the pointed, inferior edge is called the apex.

2.2.1.2 Soft Tissues

Soft connective tissues link the three bones in the knee joint (Figure 2.2). The major soft tissues in the knee are: (a) the articular capsule, (b) the quadriceps and patellar tendons, (c) the lateral collateral ligament, (d) the medial collateral ligament, (e) the anterior cruciate ligament, (f) the posterior cruciate ligament, and (g) the lateral and medial menisci.

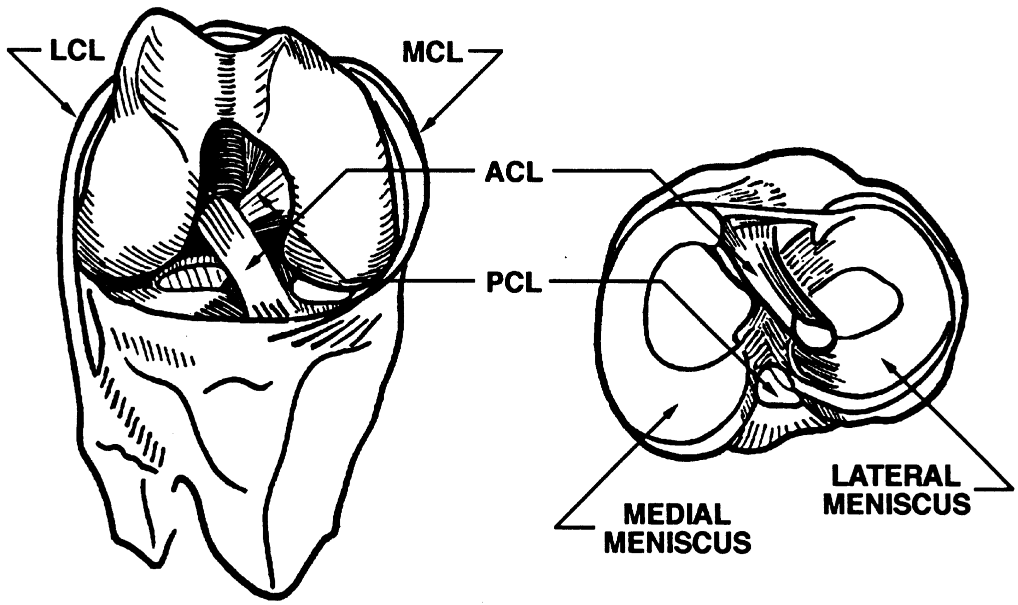

Figure 2.2 Soft Tissue of the Knee

2.2.1.2.1 Articular Capsule

The articular capsule of the knee is the largest and most extensive in the body. It consists of a strong, thin fibrous synovial membrane which forms a closed sac around the joint. Inseparable bands derived from the fascia lata and from the tendons surrounding the joint are connected throughout almost the entire extent of the synovial membrane to further strengthen the capsule. The inner lining of the synovial membrane consists of a free surface which secretes viscous synovial fluid for lubrication and nutrient supply.

2.2.1.2.2 Quadriceps and Patellar Tendons

The quadriceps tendon is the common termination of the four quadriceps femoris muscles and inserts into the base of the patella. Fibers from the central portion of the quadriceps tendon passes over the patella to merge with the patellar tendon. The medial and lateral sections of the quadriceps tendon pass on either side of the patella and blend into the joint capsule to form the medial and lateral patellar retinacula. The patellar tendon inserts into the tibial tubercle to connect the patella and the tibia.

2.2.1.2.3 Lateral Collateral Ligament

The lateral collateral ligament is a strong, round fibrous cord attached between the lateral femoral condyle and the lateral side of the fibular head. This ligament has no connections to the lateral meniscus.

2.2.1.2.4 Medial Collateral Ligament

The medial collateral ligament is a flat fibrous band attached on one end to the medial femoral condyle and on the other end to the medial tibial condyle and medial surface of the tibia. The deep fibers of this ligament adhere to the medial meniscus.

2.2.1.2.5 Anterior Cruciate Ligament

The ACL is located in the middle of the joint. The inferior end of the ACL merges with the anterior horn of the medial meniscus and attaches in front of the tibial intercondylar eminence. From this point, the ligament passes in a superior, posterior, and lateral direction to attach to the medial part of the lateral femoral condyle. The fibers of the ACL form three twisted bands: the anteromedial band, the intermediate band, and the posterolateral band [48]. Although the ACL is intracapsular, the synovial fold covers the ligament, thereby maintaining an extrasynovial environment.

2.2.1.2.6 Posterior Cruciate Ligament

The posterior cruciate ligament is also located in the middle of the joint, crossing the ACL to form an “X.” The inferior end of the posterior cruciate ligament blends with the posterior horn of the lateral meniscus and attaches to the tibial posterior intercondylar fossa. From this point, the ligament passes in a superior, anterior, and medial direction to attach to the lateral part of the medial femoral condyle. The fibers of the posterior cruciate ligament form two twisted bands: the anteromedial band, and the posterolateral band [48]. Like the ACL, the posterior cruciate ligament is covered by the synovial fold and is considered to exist in an extrasynovial environment.

2.2.1.2.7 Lateral and Medial Menisci

The menisci are two crescent lamellae located between the femoral and tibial condyles. The upper surfaces of the menisci are concave and contact the femoral condyles. The lower surfaces of the menisci are flat and rest on the tibial condyles. The peripheral border of each meniscus is thick, convex, and attached to the interior surface of the joint capsule. The opposite border is thin, concave, and free of connections.

2.2.2 Comparative Porcine Anatomy

Although the overall appearance of the pig is noticeably different from that of the human, gross anatomy of porcine and human knees are similar [12] (Figures 2.3, 2.4). Slight differences between the species do exist. The femur and tibia of the pig are relatively more massive and heavier than human bones. Also, to provide a greater moment arm for the quadriceps tendon, the porcine tibial tubercle protrudes more anteriorly than the human version.

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Figure 2.3 Skeletal Anatomy of the Porcine Pelvic Limb (Lateral View)

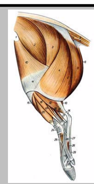

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Figure 2.4 Muscular Anatomy of the Porcine Pelvic Limb (Lateral View)

Minor soft tissue differences also exist between the porcine and human species. The porcine ACL consists of two distinct morphological units, while the human ligament is only one unit [30,31]. Hence, the porcine ligament contains only two bands: the anteromedial band and the posterolateral band. Function of the ACL in both species is similar.

The overall range of motion of the porcine knee is similar to that of the human knee [31]. Range of knee flexion in both species is about 150°. However, in the human, full extension occurs at about 0°, or when the longitudinal axes of the femur and tibia are approximately parallel. In the pig, full extension corresponds to about 30° human flexion, and full flexion is about 180°.

2.3 Anterior Cruciate Ligament Function

Ligaments constrain the relative motion of the bones in the knee joint [11]. The ACL primarily resists anterior tibial displacement through all flexion angles [13,16,28,34,58,80,82]. Secondarily, the ACL resists: (a) hyperextension [29,34], (b) internal and external tibial rotation [29,34,58,84,86], and (c) abduction and adduction [38,58,70].

2.3.1 Kinematic Behavior

The fibers of the ACL rotate about their insertion sites to control motion of the knee [75]. According to O'Connor and Zavatsky [75], Weber and Weber [92] first described the motion of the femur on the tibia during flexion and extension (Figure 2.5). If the femur rolled without sliding, the femoral condyle would lose contact with the tibial plateau at a certain flexion angle. On the other hand, if sliding occurred without rolling, the femur would prematurely contact the tibia, thus limiting motion. Hence, both rolling and sliding must occur during flexion. A four-bar linkage can theoretically model this behavior (Figure 2.6) [74,75]. Because motion of the knee occurs primarily in the sagittal plane, the model assumes the joint to be a two-dimensional single degree-of-freedom linkage. The ACL and PCL are considered to be two of the links, while the tibia and femur are the remaining two links.

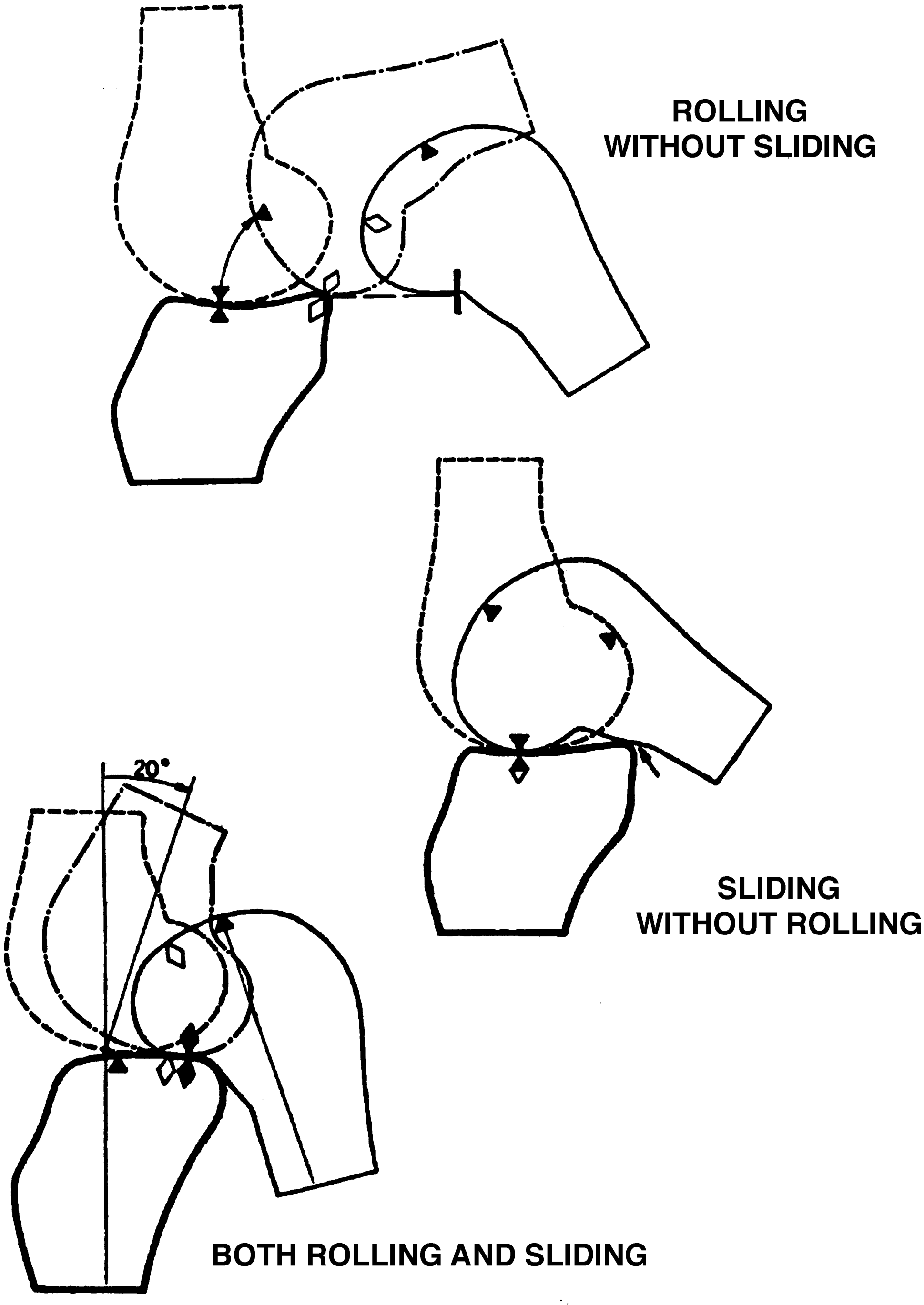

Figure 2.5 Glide-Roll Relationship of the Knee

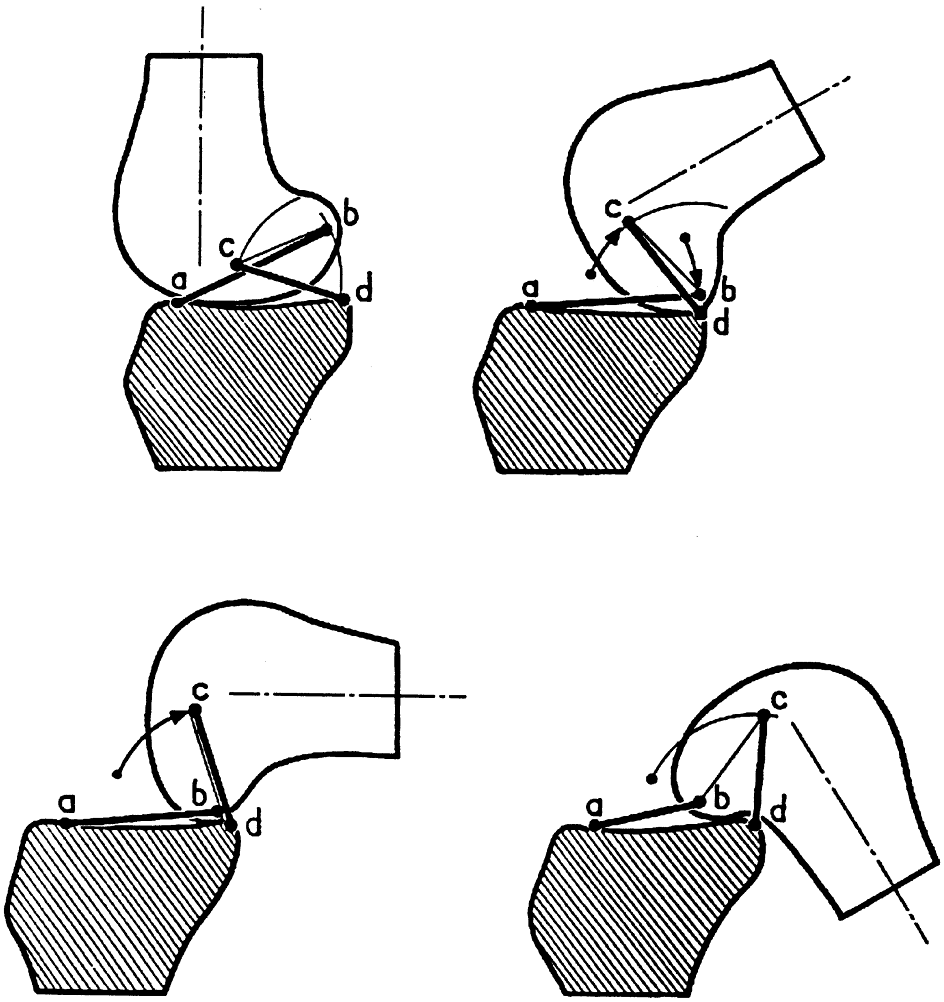

Figure 2.6 Four-Bar Linkage Model of the Knee

2.3.2 Kinetic Behavior

The ACL resists and controls knee motion by developing a tensile force when stretched [75]. Researchers have used indirect [42] and direct methods [53] to determine the force generated in the ACL during various movements.

2.3.2.1 Indirect Measurements

Researchers described an indirect method based on changes in knee kinematics after ACL transection to estimate ACL force in human cadaver knees [42,89]. A six degree-of-freedom instrumented spatial linkage was used to record relative motion between the femur and tibia as loads were applied to the joint. A load versus length curve was determined by digitizing the ACL insertion sites while performing an anterior drawer test on the specimen with only the ACL intact.

The relative movement between the ACL insertion sites was calculated for the intact knee test, and the load versus length curve was used to determine ACL load throughout the motion.

2.3.2.2 Direct Measurements

Four methods have been suggested to more directly measure ACL load: using a buckle transducer [1,52,54], bone block transducers [57], modified pressure transducers [41], and a robotic positioning system [27,55,56,81].

2.3.2.2.1 Buckle Transducer

Lewis et al. [53] described the buckle force transducer as consisting of strain gauges mounted on a rectangular metal frame (Figure 2.7). A removable cross-bar is used to place the transducer on a ligament. As the ligament tensile force increases, the bending moment on the frame increases. The output from the strain gauges is calibrated using known ligament forces to permit the measurement of future, unknown loads. Advantages to using buckle transducers to measure ACL force include: (a) The transducer is easily applied directly to the ligament; (b) the transducer directly measures ligament force, not ligament strain; and (c) the transducer is rugged enough to withstand in vivo implantation. Disadvantages of the buckle transducer include: (a) The transducer may contact other surfaces, leading to deceptive results; (b) viscoelastic creep of the ligament may alter the calibration constant after insertion of the transducer; (c) the calibration constant may be affected by stress distribution in the ligament; and (d) the ligament shortening that occurs when applying the transducer may affect the load distribution in the joint.

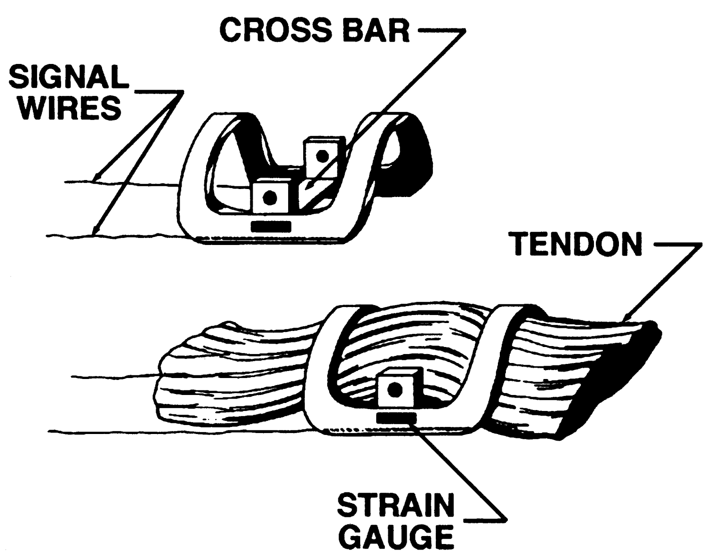

Figure 2.7 Buckle Transducer

2.3.2.2.2 Bone Block Transducer

Markolf et al. [57] introduced a method to directly measure force in the in vitro ACL using an externally mounted load cell. A hole was drilled from the anteromedial surface of the tibia in line with the axis of the ACL to a point just below the tibial plateau. Screws were placed in the subchondral bone under the ACL tibial insertion site. A threaded steel rod centered in a cylindrical liner was inserted into the hole, and acrylic cement was poured into the gap. After the cement had cured, the liner was removed. One end of a three axis load cell was fixed to the body of the tibia, and the other end was attached to the threaded rod in the cement. The bone core was then completed to free the ACL tibial insertion. Hence, the ACL maintained its anatomical position, and all ligament forces during subsequent knee motion were transmitted to the load cell. The authors noted an error in load values of up to nine percent due to motion of the bone plug in the tunnel.

2.3.2.2.3 Modified Pressure Transducer

Holden et al. [41] described the use of a modified pressure transducer to measure the in vivo force in the goat ACL during standing, walking, and trotting. A Teflon button was fastened to the diaphragm of the pressure transducer. When the device was implanted into the anteromedial band of the ACL, the ligament fibers pressed the button against the diaphragm. Increasing the axial force of the ligament increased the transverse force against button, and hence, raised the pressure sensed by the transducer. Use of the modified pressure transducer presents similar advantages and disadvantages as the buckle transducer.

2.3.2.2.4 Robotic Positioning System

Fujie et al. [27] described the use of a robotic positioning system to study joint kinematics. A 6-axis articulated robotic manipulator with 6 degrees-of-freedom motion controlled the position of a knee via a universal force sensor. Livesay et al. [55,56] used this system to determine the in situ forces in the ACL during knee motion constrained to 1 or 5 degrees-of-freedom. To accomplish this, the intact joint was moved through a range of motion while the universal force sensor monitored the external applied forces and moments. Following transection of the ACL, the robotic manipulator moved the knee precisely through the same path of motion while the universal force sensor again monitored external applied forces and moments. Because the path of motion was identical during both trials, the principle of superposition could be used to determine the force in the ACL during this motion (equal to the difference in external applied forces and moments between the two trials). Sakane et al. [81] used this technique to determine the forces in the bundles of the ACL when an anterior load was applied to the tibia.

2.4 Anterior Cruciate Ligament Injuries

Knee injuries are extremely common, and injuries to the ACL constitute a majority of these [40,46]. Research into the incidence and etiology of ACL injuries is reviewed, followed by a discussion of literature describing various treatments for these injuries.

2.4.1 Incidence and Etiology

ACL injuries occur in the general population in large numbers. In 1983, Johnson [47] reported 1.2% of a college entrance population had experienced an injury to the ACL at some time. An estimated 87,500 ACL ruptures occur annually [66]. Hirshman et al. [40] analyzed evaluations of patients that visited the San Diego Kaiser Knee Injury Clinic during a 3 year period. Of 1833 patients with acute knee injuries, 500 involved severe injuries of knee ligaments. Of these 500, 63% concerned the ACL. The majority of these ACL injuries were sports-related; football, baseball, soccer, skiing, and basketball accounted for 52% of the injuries, while other sports accounted for 15%. Vehicular and miscellaneous accidents accounted for the remainder of the injuries.

2.4.2 Treatment

Stark [88] attempted one of the earliest reported treatments for an ACL injury. Since that time, a variety of treatments have been suggested, and much controversy surrounds the clinical management of a patient with an ACL injury [3].

2.4.2.1 No Repair

Without treatment, the prognosis after an ACL injury may be poor. Based on a study of 223 patients with ACL injuries, Fetto and Marshall [23] declared that the ACL-deficient knee "appears to invariably embark upon a course of progressive deterioration and dysfunction." An untreated ACL injury can lead to increased anterior and rotational instabilities and to an increased frequency of meniscal injuries [71]. Researchers noted radiographic changes in 80% and osteoarthritis in 10% of patients with untreated ACL injuries at 14 year follow-up [61]. Noyes et al. [72] evaluated 103 untreated, athletic patients an average of 5.5 years after injury. Eighty-five of the patients participated in sports after injury, and 53 of these sustained a serious reinjury within one year of the initial injury. Only 11 patients reported having no serious limitations with respect to strenuous sports participation. Furthermore, 32 patients reported disability in walking, while 45 reported disability in activities of daily living.

Based on a study of patients with lacerated ACLs, Palmer [78] concluded that impromptu healing of a fully ruptured human ACL does not occur. He noted that the ACL had in fact absorbed after injury. A ruptured ACL can disappear entirely within 8 months [91], and possibly as soon as 6 weeks after injury [50]. O'Donoghue et al. [77] studied the response of partially and fully transected ACLs in a canine model. The ends of a fully lacerated ACL either retracted or absorbed, and a partially lacerated ACL showed little evidence of healing. The authors suggested that the "hostile" synovial environment surrounding the injured ligament deterred healing. Similar results have been noted using a rabbit model [39].

2.4.2.1.1 Effects of Synovial Fluid

Researchers have investigated the effect of the synovial environment on ACL and patellar tendon tissues [4]. Ordinarily, a synovial membrane covers the ACL, and separates the ligament from synovial fluid in the joint [6,14]. When the ACL ruptures, the membrane also tears [4]. Synovial fluid then immerses the injured ends of the ligament, and exposes the ligament to "potentially destructive enzymes" [4]. Following transection, canine ACL displayed an inflammatory response [77]. Synovial fluid also adversely affected ACL fibroblasts in vitro [5]. The number of fibroblasts populating the tissue culture decreased when exposed to synovial fluid. However, Nickerson et al. [69] found that synovial fluid stimulated rabbit ACL cell proliferation in vitro. The authors suggested that different cell types, synovial fluid sources, concentrations, and growth media may have caused the conflicting results. Hence, synovial fluid, under some conditions, may be a source of nutrition to ligament cells [69].

2.4.2.2 Primary Repair

Because an injured ACL apparently does not heal spontaneously, practitioners have attempted primary repair [4]. O'Donoghue et al. [77] found poor healing in fully transected canine ACLs that were repaired with sutures, although an unpredictable percentage of the ligaments would heal. These results were corroborated using canine and monkey models [18]. Amiel et al. [4] suggested that this phenomenon occurred because the injury model disrupted the ACL at an insertion site.

Cells from the adjoining bone could then produce the healing response. Because more than 80% of ACL ruptures occur in the proximal 20% of the tissue [85], this model does not accurately represent ACL healing in practice [4].

Feagin et al. [21] reported good initial results from the primary repair of 30 ruptured ACLs. Two years after surgery, 25 of 30 patients had "good to excellent" results. However, in a 5 year follow-up study of these patients, Feagin and Curl [22] discovered progressive degeneration of the knee. The authors considered these results to be "similar to the natural history of the unrepaired ACL." Other researchers have noted similar findings [8,76].

2.4.2.3 Reconstruction

Because of the problems associated with the innate healing response and with primary repair of injured ACLs, reconstruction of the ligament for patients with a highly active lifestyle or with functional disability is recommended [10]. Reconstruction involves substituting a graft for the injured ligament [10]. An appropriate replacement for the ACL must be selected, considering the structural and mechanical properties of normal ACL and the graft material [10]. Allografts, autografts, xenografts, and prosthetic devices have been suggested as potential replacement materials [25,26,45,60,90]. It is beyond the scope of this literature review to discuss every type of graft and reconstruction procedure. Because a bone-patellar tendon-bone autograft is arguably the most common structure used for ACL reconstruction [10,45], this procedure is reviewed in detail in the following section.

2.4.2.3.1 Patellar Tendon Autograft

ACL reconstruction with a patellar tendon autograft involves replacing the ligament by a graft taken from the injured knee. Except where noted, the following arthroscopic surgical procedure was described by Daniel [20]. An anterior medial incision is made to expose the patellar tendon and its insertion sites. Tissue is dissected along the medial side of the patellar tendon and across the front of the patella. The width of the patellar tendon is then measured. Daniel suggested that the smaller of one-third of the patellar tendon width or 10 mm determines the graft width. However, the width of the graft varies with the surgeon's preference [45]. An outline of the desired graft is drawn on the central portion of the patellar tendon. A scalpel is used to cut around the medial and lateral sides of the graft. A 20 x 9 x 7 mm bone plug from the patella and a 30 x 10 x 10 mm bone plug from the tibia constitute the ends of the graft (Figure 2.8). As with the graft width, the size and shape of the bone plugs vary with the surgeon's preference [45]. Using a sagittal saw, the bone plugs are harvested, thereby freeing the graft. The bone plugs are trimmed so that the patellar end passes through a 9 mm sizer and the tibial end passes through a 10 mm sizer. Sutures are placed in both bone plugs to provide a means to pull the ends of the graft. The graft is wrapped in a saline-soaked cover and fixed under tension for ten minutes to condition the tissue. To connect the graft across the joint, a 10 mm diameter tibial tunnel is drilled from the graft-harvesting incision, and a 9 mm diameter femoral tunnel is drilled from an incision lateral and superior to the knee. Jackson and Lemos [45] described a method in which the femoral tunnel is drilled starting from the interior of the joint. The tibial tunnel is positioned with reference to the anterior side of the anterior tibial spine and to the roof of the intercondylar notch when the knee is in full extension [36]. The femoral tunnel is positioned with reference to the lateral wall and roof of the intercondylar notch and to the posterior cruciate ligament. The surgeon can select a location that corresponds to the center of the ACL femoral attachment or one that is posterior and superior to that point [36]. The graft is pulled by the patellar end sutures through the tibial tunnel and into the femoral tunnel. The bone plug is fixed in the femoral tunnel with an interference screw, or by tying the sutures to a screw and washer (Figures 2.9, 2.10). Tension is applied to the graft by pulling on the sutures in the free bone plug, and the plug is fixed in the tibial tunnel in a similar manner as the femoral end. The incisions are closed with sutures, and the motion of the knee is tested. If the response is not satisfactory, the graft may be repositioned or retensioned. After a period of healing, the patient may follow a rehabilitation program to help restore normal function and regain muscle strength [79].

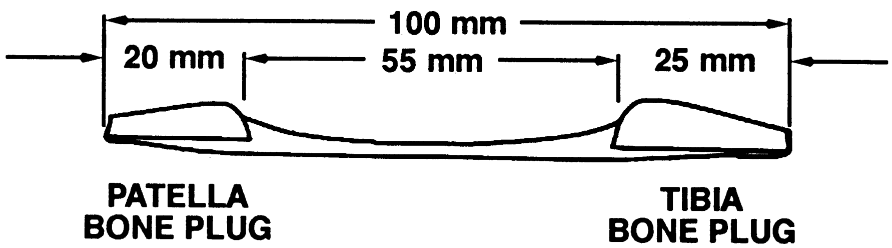

Figure 2.8 Bone-Patellar Tendon-Bone Graft

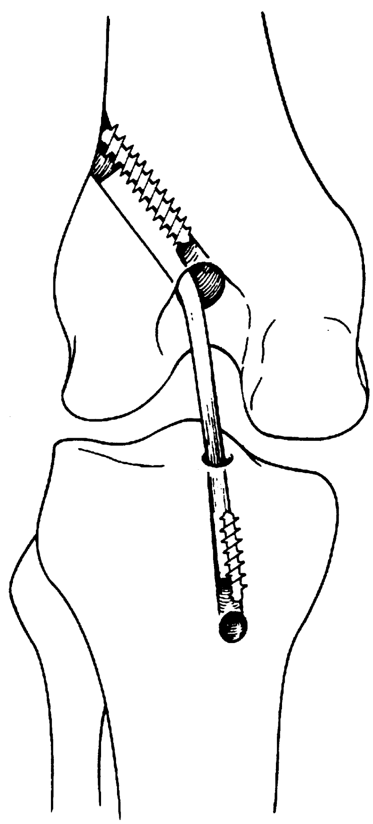

Figure 2.9 Completed ACL Reconstruction With Interference Screws

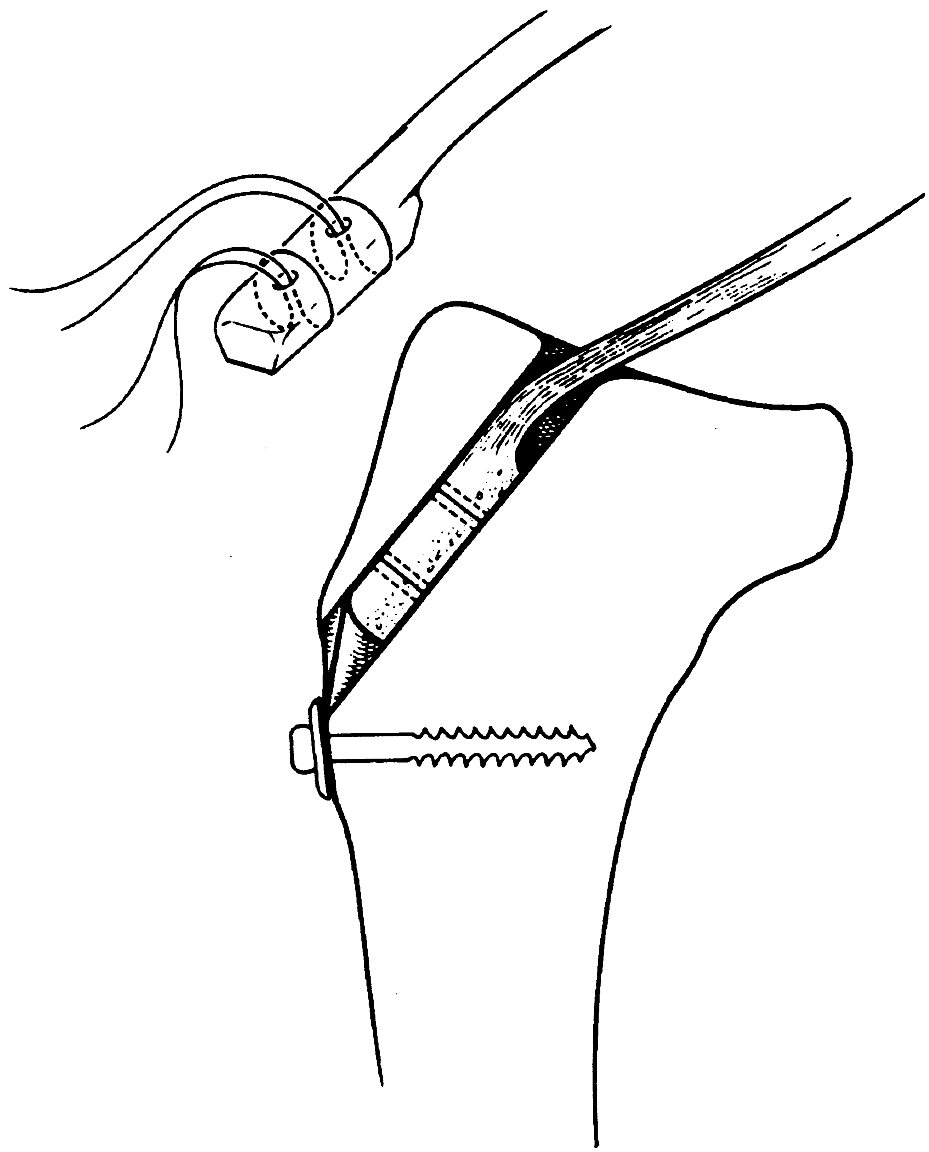

Figure 2.10 Screw and Washer Fixation of Graft

2.4.2.3.2 Follow-up Results

The outcome of reconstruction depends on the graft material selected, as well as other variables, such as graft intraarticular position, graft fixation, and initial graft tension [10]. Follow-up studies of reconstruction patients have reported varying levels of success and are difficult to interpret [33]. Gillquist [33] reviewed studies of follow-up patient examinations and reported the literature showed "a substantial number of abnormal" results. The literature reported increased knee laxity was frequently noted in patients [33]. Gillquist also reviewed studies that described "almost normal or normal knee function" in about 75% of patients after various methods of reconstruction.

Because different procedures were expected to yield different results, Gillquist attributed the promising success rates to “publication pressures”. He suggested surgeons believe the 75% level to be acceptable for publishing a procedure, and may manipulate the statistical analysis of their short-term retrospective follow-up examinations to reach that threshold. Many techniques that are initially advocated, quietly disappear without a demonstration of inefficiency [33]. According to Gillquist, surgeons advocate a procedure until they evaluate mid-term follow-up examinations of patients and find unacceptable results. They then begin to promote a new technique without determining reasons for the inadequacy of the previous method, or even reporting the failure. Because of this phenomenon, Gillquist stressed the need for 10 year prospective studies to analyze reconstruction techniques.

2.4.2.3.3 Biomechanical Changes

After implantation, the biomechanical properties of the graft change [68]. Although initially stiffer and stronger than the femur-ACL-tibia complex, researchers have shown that the structural properties of the femur-patellar tendon graft-tibia complex drop drastically immediately upon implantation [9,62,63]. The stiffness of the graft complex was determined to be 3% to 15% of a control femur-ACL-tibia complex, while the ultimate load was found to be 1% to 7% of the control. The properties improved after a period of healing, but they never approached pre-surgery levels. The specific amount of improvement varied from study to study. Ballock et al. [9] found the graft stiffness and ultimate load to be only 13% and 11% of control, respectively, after 52 weeks. After a healing period of the same duration, McPherson et al. [63] found the graft stiffness and ultimate load to be 35% and 45% of control, respectively. Ballock et al. suggested that the graft may need longer than 1 year to heal properly, but McPherson et al. found the properties decreased during the period between 1 and 2 years postoperative.

2.4.2.3.4 Biochemical Changes

Researchers have documented changes of the collagen type and content in the patellar tendon graft over time, indicating a graft remodeling process [3]. Amiel et al. [2] used a rabbit model to study the collagen content in a patellar tendon graft over 30 weeks post-surgery. Type III collagen was detected in the graft as early as 2 weeks, and reached a maximum level at 6 weeks. Type I collagen content decreased during this time to reach a minimum level at 6 weeks. Over the next 24 weeks, the Type III collagen content declined slightly, while the Type I collagen content increased slightly, so that at the completion of the study, the levels of both collagen types in the graft were similar to the amounts in the ACL. Oakes [73] reported collagen fibril diameters based on the analysis of biopsies from 39 human patellar tendon grafts at various times after surgery. The biopsy from a graft that was less than 6 months postoperative contained a large number of small diameter fibrils and none of the large diameter fibrils seen in normal patellar tendon. A similar trend was seen in biopsies from grafts that were from 1 to 5 years postoperative, although the number of larger diameter fibrils increased over time. Biopsies from grafts that were 5 to 9 years postoperative showed an increase in fibrils with a diameter greater than 100 nm.

2.4.2.3.5 Histological Changes

Researchers using histological analysis of patellar tendon grafts have detected four stages of transformation: avascular necrosis, revascularization, cellular proliferation, and tissue remodeling [3]. Kleiner et al. [49] studied the early histologic changes in rabbit patellar tendon grafts. Two days postoperative, the researchers noted a decrease in tissue cellularity, and the fibroblasts had become noticeably round. Seven days postoperative, the number of cells further decreased, so that fibroblasts were only noted in the midsubstance sporadically. Fourteen days postoperative, the tissue midsubstance was acellular; the only fibroblasts in the tissue were observed on the periphery. Amiel et al. [2] studied the histologic changes in rabbit patellar tendon grafts from 2 to 30 weeks postoperative. At 2 weeks postoperative, researchers noted a similar cell distribution to the one identified by Kleiner et al. The tissue crimp was similar to that of normal patellar tendon. Three weeks postoperative, the tissue experienced cellular proliferation, although the relative number of fibroblasts was less than normal patellar tendon. The tissue crimp remained similar to that of normal patellar tendon. Four weeks postoperative, the tissue cellularity had increased and approached that of normal ACL. Tissue crimp at this time was much less distinct. Six weeks postoperative, the tissue was more cellular than either normal patellar tendon or ACL, and the crimp occurred only in some sections. By 30 weeks postoperative, the number of fibroblasts had decreased, so that the cellularity was similar to that of normal ACL. Researchers have noted similar trends in graft remodeling using different animal models, although the duration of each stage varied slightly [7,19]. Rougraff et al. [81] studied histological changes in human patellar tendon grafts ranging from 3 weeks to 6.5 years postoperative. Focal areas of degeneration were present in grafts less than 3 years postoperative, and seemed to be most conspicuous in grafts that were 2 to 10 months postoperative. The degeneration was noted in grafts as late as 30 months postoperative, but none of the grafts that were over 3 years postoperative displayed it. These researchers classified fibroblasts as one of the following shapes: linear, spindle, oblong, or round. Normal patellar tendon was described as containing mostly linear cells. Normal ACL also was described as containing mostly linear cells, but relatively large numbers of spindle and oblong cells were also noted. The 3 week specimen contained spindle cells on the periphery and linear cells in the midsubstance. Spindle and oblong cells were more common than linear cells in the 6 week graft. In grafts older than 2 years, the oblong fibroblasts became much less numerous. Grafts that were older than 3 years contained primarily linear fibroblasts and were similar to normal ACL. The number of fibroblasts in the 6 week graft was twice that of the tissue at implantation. By 4 months, the number of fibroblasts reached eight times that of the tissue at implantation. Over the next 4 years, the number of cells decreased gradually to return to a similar level of normal ACL.

2.4.2.3.6 Graft Position

The intraarticular position of the graft determines the change in length of the graft as the knee moves through its full range of motion [36]. Bylski-Austrow et al. [17] found the femoral tunnel position significantly affects graft force as the knee is flexed. Moving the tunnel location anteriorly increased the force, while placing it distally decreased the force. A posterior location decreased the force slightly. A tibial tunnel that is placed too anteriorly can cause graft impingement [43]. Some surgeons try to achieve an isometric placement in which tunnel locations are selected to maintain a constant graft length as the knee moves through its full range of motion [36]. However, researchers have not been able to determine truly isometric sites [65,83,87].

2.4.2.3.7 Graft Fixation

Researchers have demonstrated that failure of a graft shortly after implantation may be due to failure of a fixation site [51,59]. During tensile tests along the tibial axis, Kurosaka et al. [51] found a significantly greater ultimate load when a 9.0 mm diameter interference screw was used versus a suture, screw, and washer fixation. Furthermore, all failures occurred at a fixation site. However, using a similar sized graft, Matthews et al. [59] found no significant difference in ultimate load between the two fixation techniques. Graft failure in this study was more likely to be a tissue rupture instead of fixation site failure. The authors attributed the different findings to strain rate differences. Matthews et al. tested specimens at a strain rate of 51 cm/min., while Kurosaka et al. tested with a strain rate of 180 cm/min. Because the graft is exposed to a variety of strain rates postoperative, it is not clear which strain rate best approximates in vivo graft failure [59]. Although failure modes were different for the two studies, ultimate loads were similar.

2.4.2.3.8 Graft Pretension

Graft pretension affects graft force [17], knee kinematics [15,17,24,37,44,64,67,93], and tissue remodeling [93]. Increasing the pretension in a flexible cable graft from 22 to 44 N led to significantly greater graft forces and posterior tibial displacements at all flexion angles [17]. However, using patellar tendon grafts, Gertel et al. [32] found no significant effect of pretension on knee kinematics or graft force. Yoshiya et al. [93] studied the effects of graft pretension in vivo using a canine model. Patellar tendon grafts were pretensioned to 1 or 39 N. In the immediate postoperative period, significantly greater knee laxity was noted in the lower tensioned group. After a 3 month healing period, however, no significant difference in laxity existed. Furthermore, no significant difference in structural properties was detected, although the authors noted a trend toward greater strength and stiffness in the lower tensioned grafts. After this healing period, the lower tensioned grafts had more vascularity than the higher tensioned grafts. Histologic examination showed the higher tensioned grafts exhibited focal degeneration and collagen fibers were replaced by a myxoid, extracellular matrix. The lower tensioned grafts displayed no apparent degeneration.

2.4.2.3.9 Flexion Angle at Pretensioning

Flexion angle when pretensioning the graft affects the magnitude of graft force and knee kinematics [17,32]. Using a flexible cable graft, Bylski-Austrow et al. [17] found pretensioning the graft at 30° flexion significantly increased graft load over a full range of motion as opposed to pretensioning at 0° flexion. The pattern of graft force did not vary between the two groups. Pretensioning the graft at 30° also significantly displaced the tibia posteriorly when compared to pretensioning at 0° flexion. Gertel et al. [32] verified these results using a patellar tendon graft in a cadaveric model.