3. Examination of Knee Kinematics In Vitro

3.1 Introduction

This chapter details two in vitro studies that were conducted to better understand the kinematic behavior of the knee. First, the role of graft pretension in the restoration of intact knee kinematics was investigated. Second, the kinematics of human and porcine knees were compared and contrasted.

3.2 Tensioning of the Anterior Cruciate Ligament to Restore Kinematics

3.2.1 Introduction

The anterior cruciate ligament (ACL) is the most commonly torn ligament in the human knee. ACL disruption leads to increased laxity of anterior tibial translation and internal tibial rotation [19,23,46]. This increased laxity can lead to a feeling of instability and is associated with subsequent articular cartilage damage. One of the goals of ACL reconstruction with a suitable graft is to restore the kinematics (e.g. the glide-roll relationship of the femur and tibia) and laxity to that of the pre-injury knee. However, achievement of this goal has not been documented, and researchers continue to investigate methods to improve the procedure and outcome.

Graft pretension has been shown to significantly affect kinematics following an ACL reconstruction [9,11,24,27,32]. Additionally, pretension directly influences the magnitude of force carried by the graft throughout the range of knee motion [24-26] and may have an effect on the biological remodeling of the graft [50]. However, the best pretension is still unknown.

Several techniques to select the best graft pretension have been described in the literature. In the first approach, the knee is positioned at a given flexion angle, an anterior force is applied to the tibia, and the graft is pretensioned to reproduce anterior translation of the intact knee [6]. As a second approach, Hunter et al. [22] suggested pretensioning a graft to match the force in an intact ACL at 30° flexion with an applied 90 N anterior tibial load. Finally, Nabors et al. [34] recommended pretensioning the graft using a one-handed maximal sustained pull on the graft with the knee at full extension. These studies consider the restoration of only one variable (A/P position or graft load) when the knee is held in a fixed position. However, these approaches may not restore kinematics throughout the range of knee motion.

A variety of tissues have been proposed as substitute grafts for a ruptured ACL [10]. However, none of these tissues exhibit the structural and mechanical properties of the intrinsic ACL [7,36,37,48]. These dissimilarities lead to the development of graft forces that differ in magnitude and direction from the forces in the ACL. One can not determine if the inability to restore kinematics are due to the graft or to the pretensioning technique. To isolate effects of the pretensioning technique, it is desirable to use a graft that matches the structure and properties of the intrinsic ACL.

To investigate how pretension affects kinematics throughout knee motion, this study had two goals. The first goal was to evaluate whether restoration of knee kinematics was possible by using an ideal graft—the anatomically oriented, native ACL. The second goal was to develop a method to determine the optimal graft pretension considering all kinematic variables over a range of knee motion.

3.2.2 Materials and Methods

The optimal pretension was established by three steps. First, the kinematics of an intact knee were measured during a simulated active extension motion. Then, kinematics of the same knee were measured following reconstruction with the ACL set to five different pretensions. Finally, a least-squares minimization technique was applied over a range of flexion to calculate the pretension that best restored intact-knee kinematics.

3.2.2.1 Specimen Preparation

Five unilateral fresh-frozen lower extremities with no outward signs of joint degeneration were used in this study. The absence of degenerative joint disease was verified by dissection at the conclusion of the experiment. Donor age ranged from 28 to 73 years, with a mean of 51.

In preparation for testing, a knee was isolated at mid-femur and mid-tibia and was dissected to the joint capsule. The quadriceps tendon remained intact. A 14.3 mm (9/16 in.) diameter solid aluminum rod was fixed in the intramedullary canal of the femur using two carriage bolts oriented perpendicularly to the rod and in mutually exclusive directions. A 12.7 mm (1/2 in.) diameter hollow aluminum rod was fixed in the intramedullary canal of the tibia in a similar fashion. The femoral rod was clamped horizontally to position the knee as shown in Figures 3.1 and 3.2.

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Figure 3.1 Knee Specimen, Lateral View

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Figure 3.2 Knee Specimen, Oblique View

3.2.2.2 Isolating the Anterior Cruciate Ligament

To use the intrinsic ACL as a graft, a controlled way to release and hold the tibial insertion site of the ACL was developed. This coring procedure preserved the orientation of the ACL within the joint while allowing a given pretension to be applied to the ligament.

The interior of the joint was visualized through a parapatellar incision. To serve as a guide for cannulated drill bits used during the coring procedure, a 2.4 mm diameter Steinmann pin was placed in the tibial insertion site of the ACL. To attach a device for pretensioning the ACL, a section of the tibia was removed with a modified Forstner bit (Figure 3.3). Using a 20.0 mm (outer diameter) hole saw, a core was started (Figure 3.4). Following the path started by the hole saw, a crescentic oscillating saw blade was used to cut the tibial plateau at the anterior and posterior borders of the ACL insertion site. Bridges of bone remained intact along the medial and lateral borders. At this point, the parapatellar incision was sutured.

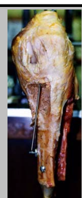

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Figure 3.3 Tibia With Removed Section

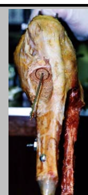

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Figure 3.4 Tibial Core Formation

To limit rotation of the insertion site, the pretensioning device was designed with a spline bearing (model SSPF 25; NB Corporation of America, Elk Grove Village, IL) as the integral part. A bearing of this type permits linear translation of the shaft, but prevents shaft rotation (Figure 3.5). An aluminum cap was secured to the subchondral bone under the insertion site using three 3.5 mm cortical bone screws, and the shaft of the bearing was attached to the cap (Figures 3.6, 3.7). The casing of the bearing was then fixed to the tibia with two 4.5 mm cortical bone screws. A 500 N load cell (model ELF-26-500; Entran, Fairfield, NJ) and a tensioning nut were connected between the bearing shaft and casing. The assembled device is shown attached to the tibia in Figures 3.8-3.11.

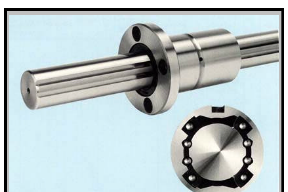

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Figure 3.5 Spline Bearing

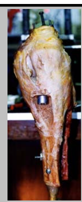

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Figure 3.6 Bone Connection to Cap

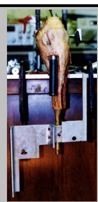

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Figure 3.7 Cap Connection to Bearing Shaft

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Figure 3.8 Pretensioning Device On Tibia, View 1

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Figure 3.9 Pretensioning Device On Tibia, View 2

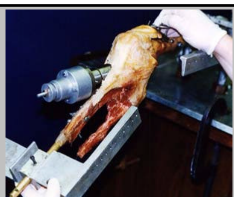

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Figure 3.10 Pretensioning Device On Tibia, View 3

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Figure 3.11 Pretensioning Device On Tibia, View 4

3.2.2.3 Kinematic and Load Measurements

A quadriceps-directed motion was selected to simulate active extension for this study. Using a clamp, the quadriceps tendon was attached via a load cell (model LCCB-100; Omega, Stamford, CT) to a permanent magnet DC motor (model Maxi-TorQ Premium; Dayton Electric, Chicago, IL) (Figure 3.12). To measure relative motion between the femur and tibia, a six degree-of-freedom instrumented spatial linkage (accuracy of 0.5 mm and 0.1° for translations and rotations, respectively) was connected between the intramedullary rods [31] (Figures 3.13, 3.14). A joint coordinate system as defined by Grood and Suntay [18] was used to describe tibia motion relative to the femur in clinical terms of three rotations and three translations. During an experimental trial, signals from the kinematic linkage and the quadriceps tendon load cell, along with the signal from the ACL load cell during pretension trials, were digitally acquired at 28 Hz as the knee was moved from 90° flexion to full extension by the motor.

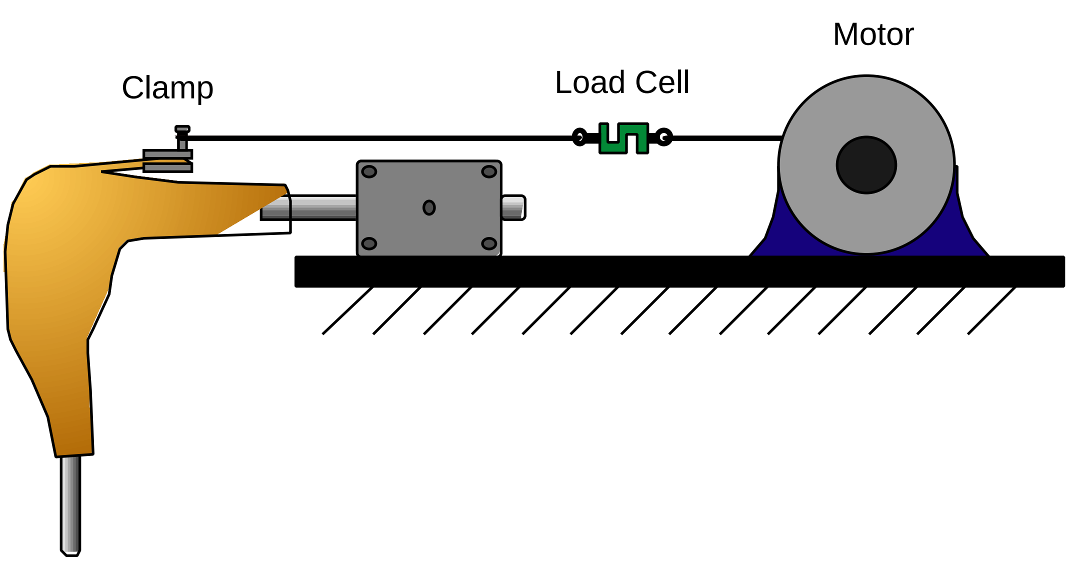

Figure 3.12 Active Extension Simulator

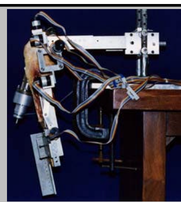

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Figure 3.13 Instrumented Spatial Linkage On Knee, Lateral View

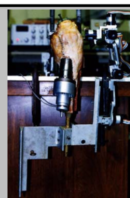

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Figure 3.14 Instrumented Spatial Linkage On Knee, Anterior View

With the pretensioning device in place, the protocol was to collect data for the intact knee. To access the ACL insertion and release the ligament, the parapatellar incision was re-opened. With the pretensioning device maintained in place, the core was completed by cutting the remaining bone bridges from inside the joint. The parapatellar incision was sutured again, and the ACL could now be pretensioned to a given level.

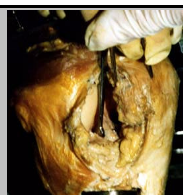

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Figure 3.15 Releasing the Tibial Plug

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Figure 3.16 Released Tibial Bone Plug, Interior View

3.2.2.4 Anterior Cruciate Ligament Pretensioning

To pretension the ACL, the knee was passively placed in 20° flexion (Figure 3.17, 3.18). During preliminary experiments, varus/valgus and internal/external rotations were seen to greatly affect

ACL tension. Therefore, when pretensioning the ACL, these rotations were held at values identical to those of the intact knee at 20° flexion. Position of the knee during pretensioning was monitored using real-time signals from the instrumented spatial linkage.

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Figure 3.17 Knee Positioned For Pretensioning, Lateral View

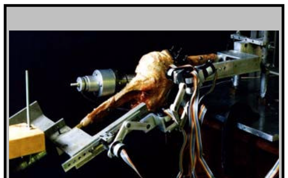

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Figure 3.18 Knee Positioned For Pretensioning, Oblique View

A given pretension was applied to the ACL, and the knee was passively flexed and extended ten times. The knee was returned to the pretensioning position, and, if the tension was within 1 N of the set value, kinematic data was collected during the full extension movement. Otherwise, the ACL was re-tensioned and the knee cycled again until the set pretension was maintained. At the end of data collection, the pretension was again verified in a similar manner. At the conclusion of the pretensioning trials the ACL was transected, and data for the ACL-deficient knee was collected.

For analysis purposes, it was necessary that at least one selected pretension would underconstrain the joint, while another would be overconstraining. Based upon preliminary experiments, 0, 10, 20, 30, and 40 N were selected as the pretensions for this study with the expectation that this range would satisfy the given requirements for any knee.

3.2.2.5 Least-Squares Analysis

To determine the pretension that best restored intact knee kinematics, a minimization technique was applied to the kinematic data. A quantity, defined as the squared difference in a kinematic variable (such as anterior/posterior translation) between an intact and pretensioned knee, was calculated at all flexion angles for all five pretensions. These quantities were then summed over the range of zero to forty degrees flexion for each pretension. A quadratic curve was fitted to the resulting points, and the minimum was taken as the optimal pretension.

3.2.3 Results

3.2.3.1 Quadriceps and Anterior Cruciate Ligament Loads

During the simulated knee extension, little variation in quadriceps load existed between the intact, pretensioned, or ACL-deficient conditions, except when full extension was approached (Figure 3.19). In this region, approximately 0-10° flexion, greater pretensions led to greater quadriceps loads. As pretension was increased from 0 N to 40 N, quadriceps load at full extension increased by an average of 25 N.

For a given pretension, ACL load typically varied only over the range of 0-40° flexion (Figure 3.20). In this range, ACL load increased with decreasing amounts of flexion. Greater pretensions also led to greater ACL loads throughout this range.

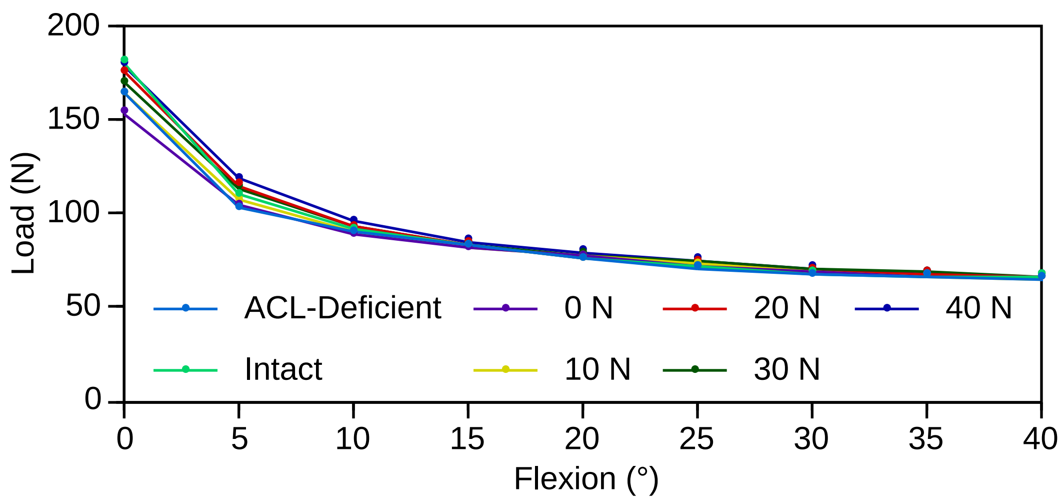

Figure 3.19 Quadriceps Loads

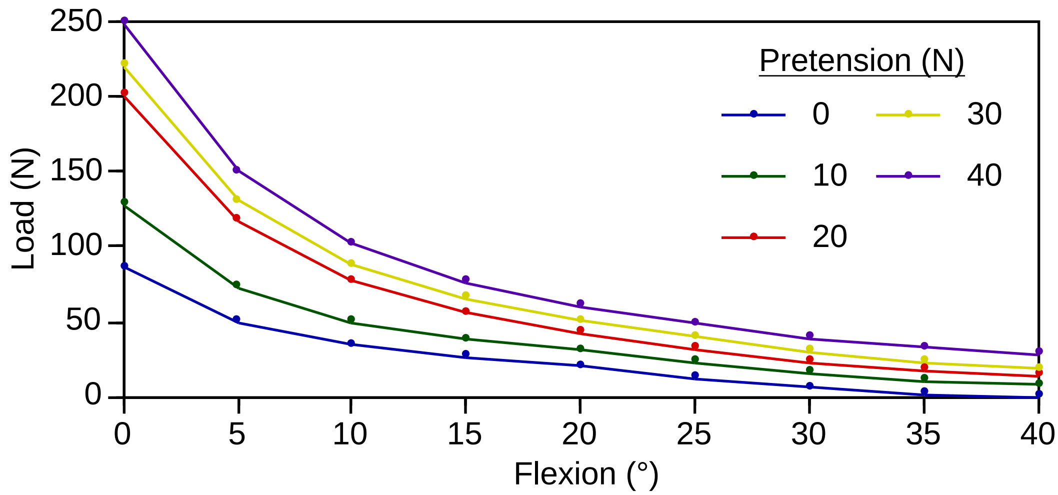

Figure 3.20 Anterior Cruciate Ligament Loads

3.2.3.2 Joint Kinematics

The kinematic variables that were most affected by ACL initial condition were anterior/posterior (A/P) translation and internal/external (I/E) rotation. Because loads in the ACL were significant only over the range of 0-40° flexion, changes in the kinematic variables were also considered only in this flexion range.

Kinematics of the intact knees during the simulated extension followed anticipated patterns (Figures 3.21, 3.22). For example, the anterior drawer effect of the quadriceps load and the external rotation of the screw-home mechanism were noted. Also as expected, transecting the ACL led to increased anterior translation and internal rotation throughout the range of flexion when compared to the intact knee.

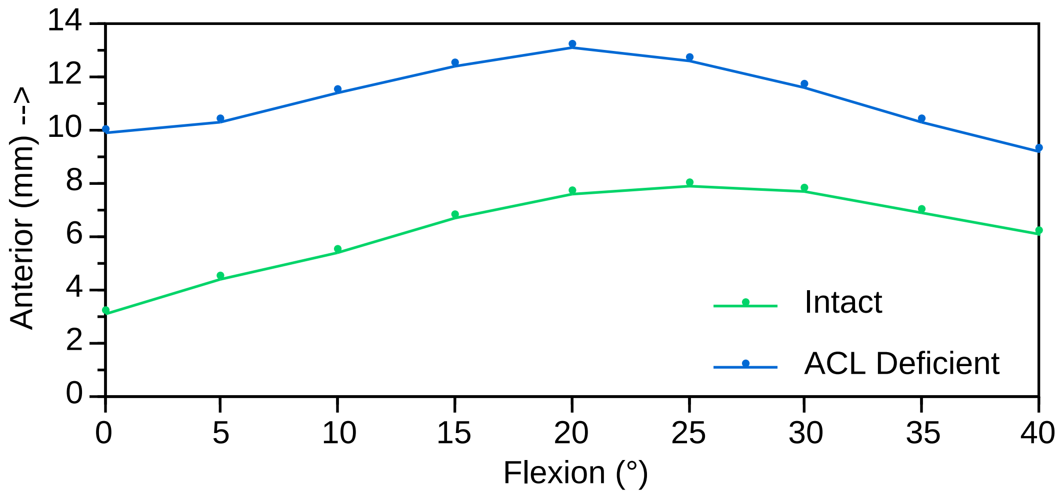

Figure 3.21 Anterior/Posterior Translation Versus Flexion

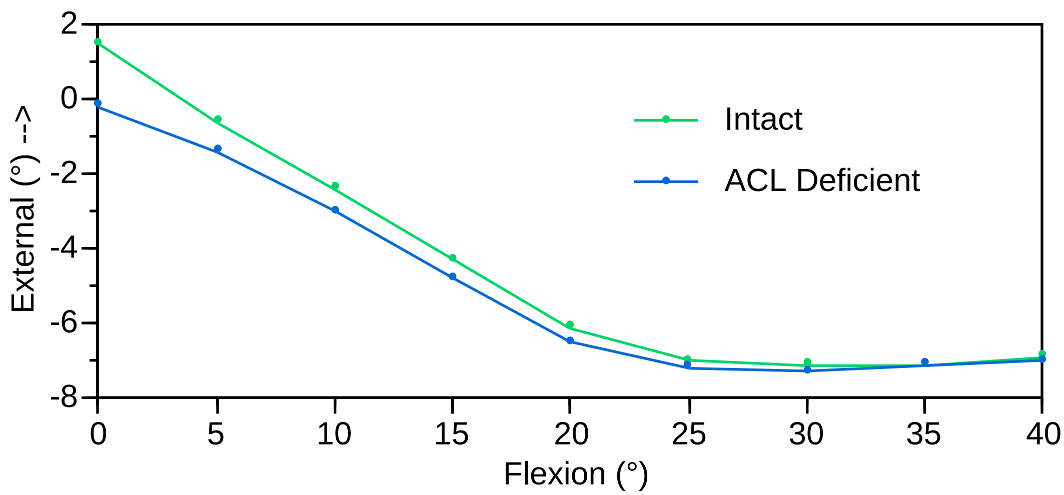

Figure 3.22 Internal/External Rotation Versus Flexion

The relationship between A/P translation and flexion for a representative specimen with different pretensions is shown in Figure 3.23. Increasing the pretension led to restricting the anterior translation of the tibia. The curves relating to the various pretensions all displayed a similar shape, and a continuous progression from curve to curve was seen as pretension was changed. Figure 3.24 shows the corresponding relationship between I/E rotation and flexion for the same specimen. Increasing the pretension led to restricting the internal rotation of the tibia. Although differences among pretensions are not as distinguished as for A/P translation, particularly between 30° and 40° flexion, a progression from curve to curve as pretension was changed did exist. The curves relating to the various pretensions again displayed a similar shape.

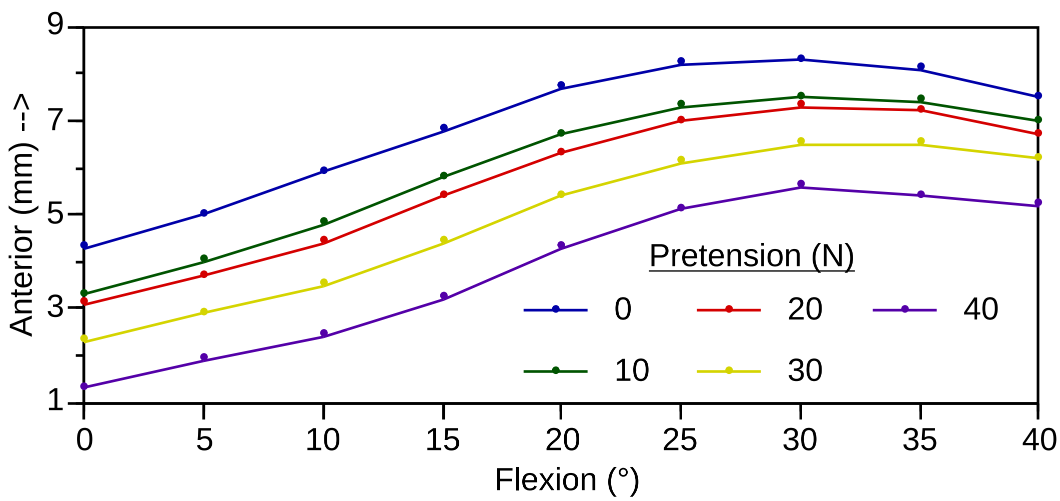

Figure 3.23 Anterior/Posterior Translations With Varying Pretensions

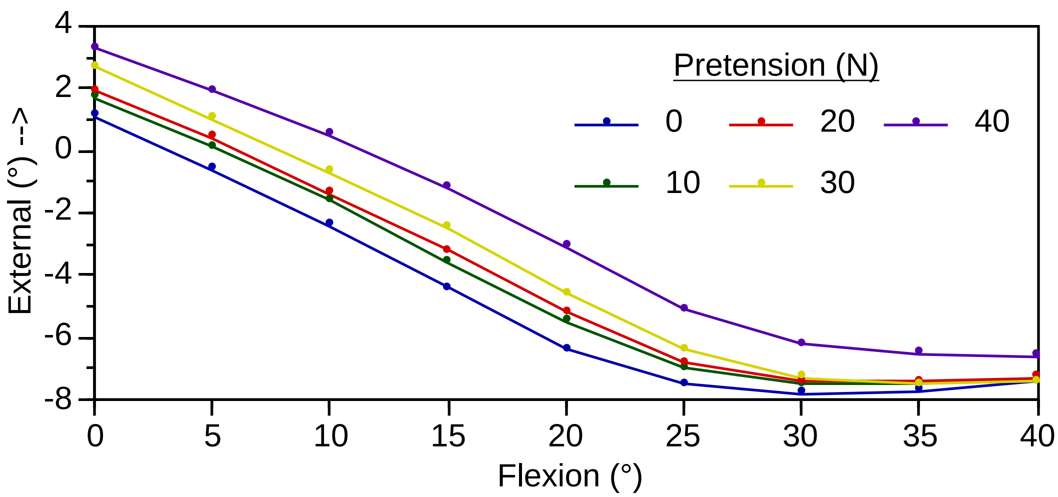

Figure 3.24 Internal/External Rotations With Varying Pretensions

3.2.3.3 Optimal Pretensions

An unexpected result arose when comparing A/P translation and I/E rotation between the pretensioned knee and the intact knee (Figures 3.25, 3.26). While the curves for the various pretensions displayed similar shapes, the curve from the intact knee followed a distinctly different path. No one pretension restored the intact knee kinematics throughout the entire range of motion. Still, with the goal of determining the optimal pretension, the minimization technique was applied (Figure 3.27).

The optimal pretensions to restore A/P translation and I/E rotation are listed in Table 3.1. A wide range of optimal pretensions existed across specimens, extending from 0 to 40 N. It should be emphasized that using a 0 N pretension would not result in an ACL-deficient knee because the pretension was applied with no loading of the quadriceps tendon. After pretensioning, when a quadriceps load was applied, the graft developed a counteracting force.

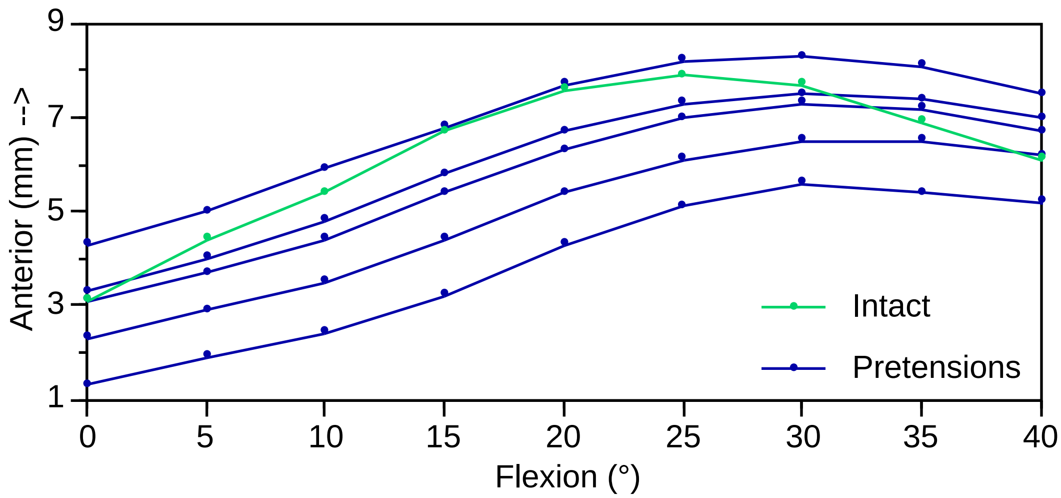

Figure 3.25 Anterior/Posterior Translations Superimposed

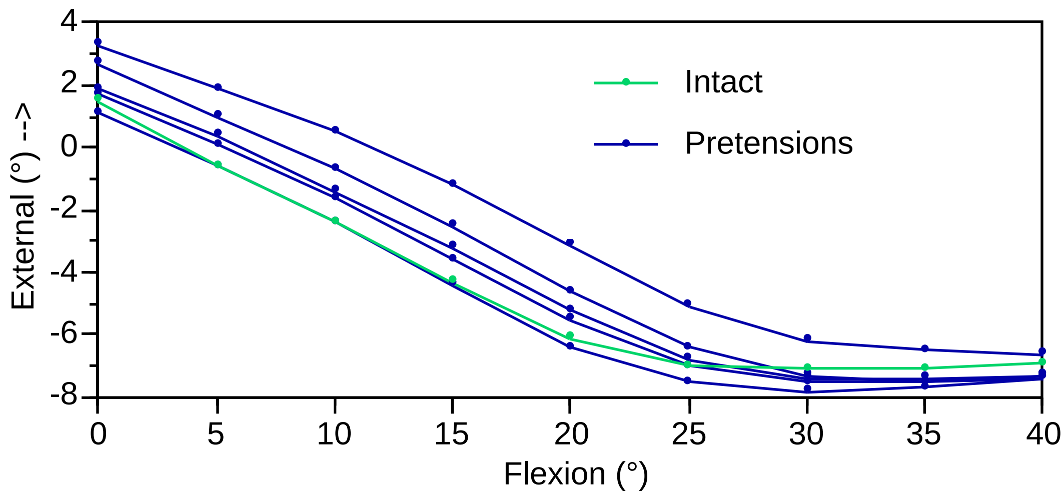

Figure 3.26 Internal/External Rotations Superimposed

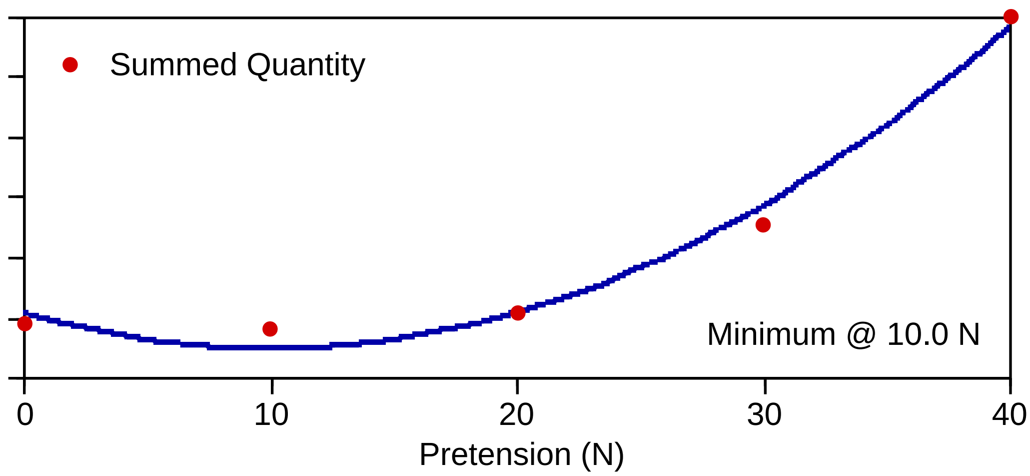

Figure 3.27 Example of Minimization Quadratic Fit

Table 3.1 Optimal Pretensions To Restore Intact Knee Kinematics

	Optimal Pretension (N)	
Specimen	For A/P Translation	For I/E Rotation
1	24.0	25.4
2	1.7	40.0
3	10.0	9.3
4	28.6	0.0
5	0.0	0.0

3.2.4 Discussion

One goal of ACL reconstruction, the restoration of intact knee kinematics, was investigated in this study. No study to date has reported attaining this goal. In previous attempts, researchers have used a variety of substitute grafts, ranging from bone-patellar tendon-bone complexes to carbon fiber composites [10]. To eliminate confounding variables introduced by differences in tissue structure, alignment, and material properties, the intrinsic ACL in its original anatomic orientation was used as a graft in the current study. Although great care was taken to maintain this orientation, intact knee kinematics were still not restored.

In addition to the graft, several other factors in the pretensioning procedure that may affect resulting joint kinematics were controlled in this study. First, the position of the joint during pretensioning was maintained to a much greater degree than in previous studies. Besides flexion angle, I/E rotation and ab/adduction were also held constant while the graft was pretensioned. Next, because different external loads on the joint would cause dissimilar kinematic motion, the same equipment was fixed to the specimen for all data collection trials. Additionally, quadriceps load variations were small and occurred only near the end of a trial, after the kinematic differences had appeared. Finally, integrity of the bony architecture of the joint was maintained. In particular, dissection following testing showed that the intercondylar eminence was not damaged by the coring procedure.

In spite of carefully controlling the variables in the pretensioning procedure, intact knee kinematics were not restored. Thus, it may not be possible to restore kinematics after releasing the ACL insertion site. It may also be possible that current tensioning techniques do not allow the restoration of kinematics. This finding indicates that the restoration of kinematics is not achievable, and that other considerations, such as cellular repopulation of the graft, dictate the “ideal” pretension. Further research into various techniques used to pretension a graft during ACL reconstruction and the effects of this pretension on biological remodeling is warranted.

A review of the literature revealed no studies that reported the load in the ACL or in a graft following ACL reconstruction during the simulated active extension motion used in the current study. However, other modes of motion, such as passive flexion/extension or a constant quadriceps load coupled with a resisting posterior tibial force to extend the knee, have been studied [9,12,16,20,22,26,28,29,32-34,38,47,51]. In support of results from previous studies using the ACL or an anatomic graft placement, the current study found a similar pattern in ACL load over flexion. Little load was seen in the ACL at larger flexion angles, while a rapid increase in load occurred near terminal extension. This finding also agrees with studies of ACL elongation. The ACL has been reported to be elongated at flexion angles less than 45°, thus suggesting the ligament supports a greater load in this region [3,12,21,38,40,45]. Contrary to the findings of the current study, Shoemaker et al. [43] reported a pattern of graft tension versus flexion resembling an inverted “U” with a maximum force at 45°. This difference is likely due to differing methods of generating knee motion. The current study simulated an active knee extension from a sitting position, while the prior study mimicked a standing position with quadriceps stabilization.

It should be emphasized that the term "ACL load" as used in the current study does not refer to the resultant force of the ACL. Instead, it refers to the component of ACL force measured along the longitudinal axis of the bearing shaft. Because this axis may not be identical in different specimens, absolute values of force are not compared between specimens. However, for a given specimen, the axis was fairly constant and thereby provided a means to compare ACL load within a specimen.

The current study attempted to determine an optimal graft pretension based on all kinematic variables through a range of joint motion. Burks and Leland [6] found mean graft tensions of 16, 38, and 61 N were required to return the tibia to the A/P position of the intact knee under an 89 N anterior drawer with bone-patellar tendon-bone, semitendinosis, and iliotibial band grafts, respectively. The optimal pretensions determined in the current study ranged from 0 to 40 N. Variations between the results from the studies may be due to graft tissue type. Grafts with dissimilar structural properties would be expected to require different pretensions to meet a given criterion. Additionally, the criterion used to determine the “ideal” pretension may affect the outcome. The previous study considered only one variable (A/P position) and held the knee in a fixed position. The current study considered three-dimensional motion of the knee under conditions more representative of activity.

The optimal pretensions determined for three specimens were similar for both A/P translation and I/E rotation. However, the remaining two specimens required different pretensions. This indicated that considering only one variable when attempting to restore kinematics may not always be sufficient. For example, if a graft was to be pretensioned using A/P translation considerations, it is possible to negatively affect I/E rotation behavior, and vice-versa. Therefore, when selecting graft pretension, it is important to consider multiple kinematic variables over a range of motion.

3.3 Comparison of Human and Porcine Knee Kinematics

3.3.1 Introduction

In vitro and computer models used to study ACL reconstruction provide considerable understanding of the surgical technique [e.g. 9,51]. However, these studies provide a glimpse at how the joint responds only at the instant of graft implantation. Because the graft is living tissue that remodels over time [1,2,35], subsequent joint response can not be predicted using these models.

Since the complex environment of the graft inhibits the use of in vitro models to predict biological and mechanical responses to ACL reconstruction, animal models are commonly used [4]. The suitability of a given animal model to represent the human response to ACL reconstruction involves several factors. Based on one such factor, force distribution in the ACL, the pig may be an appropriate model [49]. While other considerations, such as large joint size and convenience, also encourage the use of the pig, little scientific information about the model exists in the literature. The purpose of this investigation, then, was to increase understanding of the porcine model for ACL reconstruction in humans. In this regard, the kinematics of human and porcine knees were compared during a simulated active extension movement.

3.3.2 Materials and Methods

The kinematics of seven human and seven porcine knees were evaluated under intact and ACL-deficient states. A previous study (see Section 3.2) was used as a source of data for the human knees. Data for the porcine knees was acquired through additional testing.

3.3.2.1 Porcine Specimen Preparation

Porcine knees were obtained with the tissue stripped and isolated at the hip and mid-tibia. The joint capsule remained intact. The femur was cut at the midpoint to receive a 14.3 mm (9/16 in.) diameter solid aluminum rod in the intramedullary canal. The rod was fixed in place using poly(methyl methacrylate) and two carriage bolts oriented perpendicularly to the rod and in mutually exclusive directions. A 12.7 mm (1/2 in.) diameter hollow aluminum rod was fixed in the intramedullary canal of the tibia in a similar fashion. The femoral rod was clamped horizontally to position the knee as in the previous study.

3.3.2.2 Porcine Kinematic and Load Measurements

The same motor-load cell-cable system as in the previous study was used to simulate an active extension movement. However, because the knees were obtained without intact quadriceps tendons, another method for applying the extension moment was required. A 6.35 mm (1/4 in.) Steinmann pin was placed transversely through the proximal end of the patella. The cable from the load cell was then connected to both ends of the pin to form a "Y." In this manner, a force could be applied to the patella as if the quadriceps tendon had been pulled.

To measure relative motion between the femur and tibia, the same six degree-of-freedom instrumented spatial linkage as in the previous study was connected between the intramedullary rods [31]. A joint coordinate system as defined by Grood and Suntay [18] was again used to describe tibia motion relative to the femur in clinical terms of three rotations and three translations. During an experimental trial, signals from the kinematic linkage and the load cell were digitally acquired at 28 Hz as the knee was moved from 90° flexion to full extension by the motor.

Data was first collected for the intact knee. A medial parapatellar incision was then made to visualize the interior of the joint. The ACL was transected, and after the incision was sutured, kinematic data for the ACL-deficient knee was collected.

3.3.2.3 Kinematic Comparisons

Because of interspecimen variation, establishing an average model of knee motion for each species was not a useful method of comparison. Instead, gross kinematic patterns were identified for intact knees of each species, and changes in these patterns attributed to transecting the ACL were evaluated.

3.3.3 Results

3.3.3.1 Quadriceps Load

During the simulated extension of intact knees, quadriceps load varied over the range of flexion in a consistent manner. The pattern for each species differed slightly. In human specimens, quadriceps load increased slowly as the knee was extended from 90° to approximately 60° flexion (Figure 3.28). Quadriceps load then remained fairly constant until approximately 15° flexion, when the load sharply increased. Terminal extension was reached between 0° and 5° flexion. In porcine specimens, no region of constant quadriceps load was noted (Figure 3.29). Instead, the load

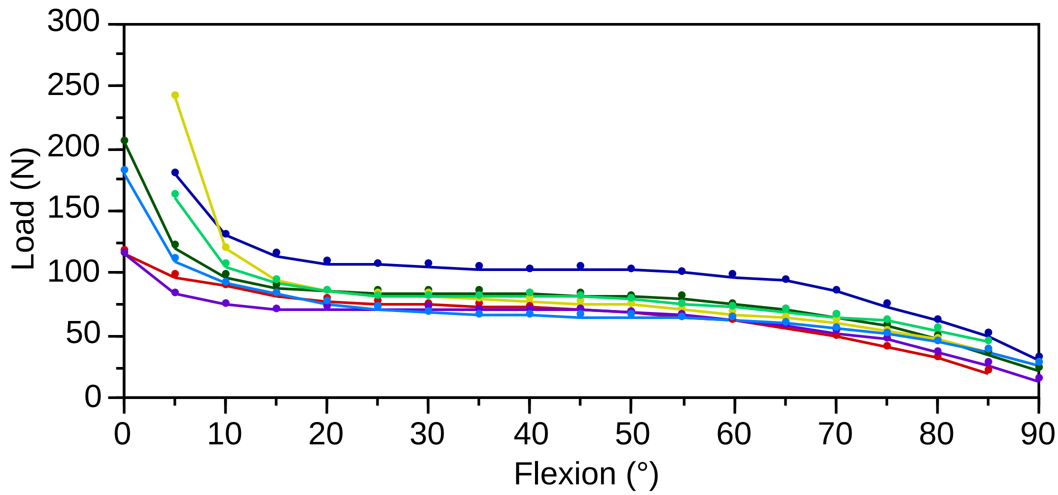

Figure 3.28 Quadriceps Load Versus Flexion For Intact Human Knees

increased slowly as the knee was extended from 90° to approximately 25° flexion, at which point a sharp increase in load was seen. Terminal extension occurred between 10 and 15° flexion.

Similar patterns of quadriceps load were seen after transecting the ACL (Figures 3.30, 3.31). The only difference between these patterns and those of the intact specimens occurred as terminal extension was approached. Transecting the ACL increased the amount of extension possible by approximately 5°, and less quadriceps force was required at a given flexion angle.

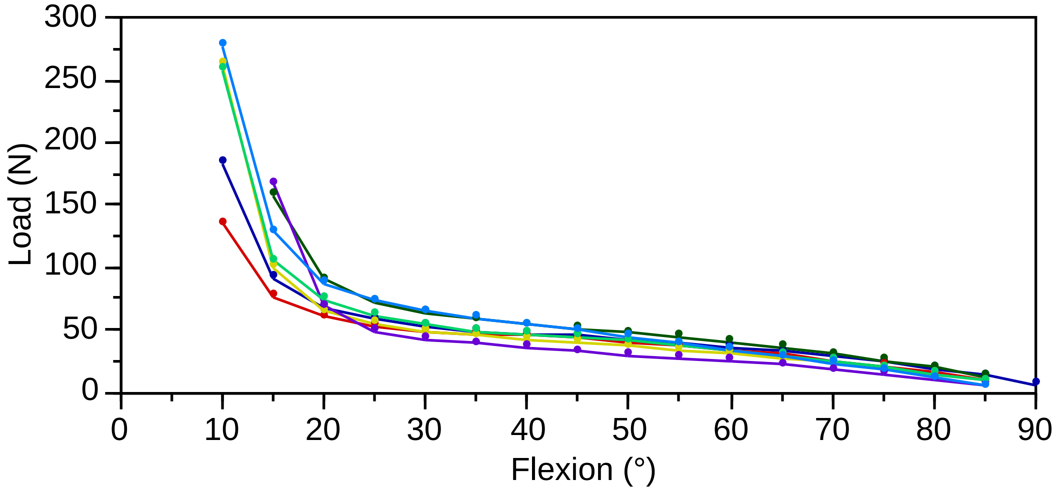

Figure 3.29 Quadriceps Load Versus Flexion For Intact Porcine Knees

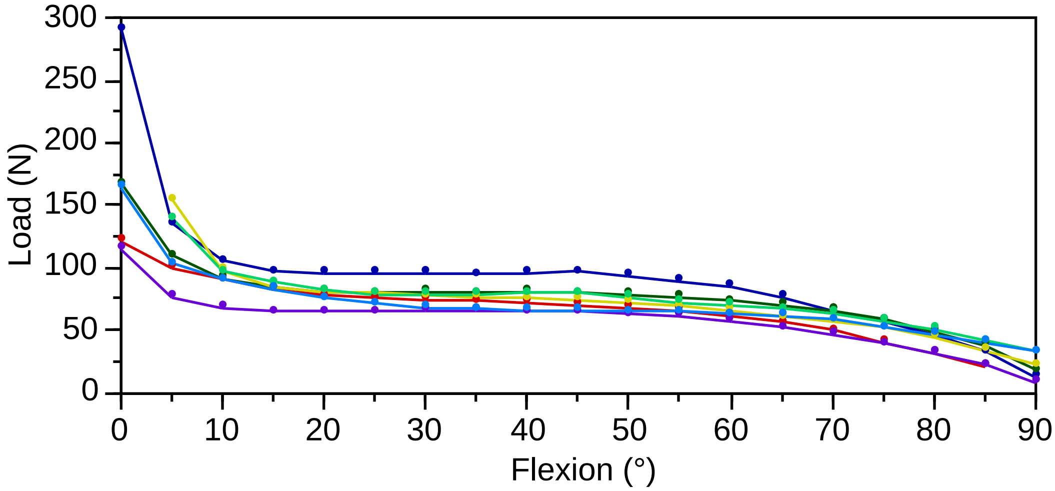

Figure 3.30 Quadriceps Load Versus Flexion For ACL-Deficient Human Knees

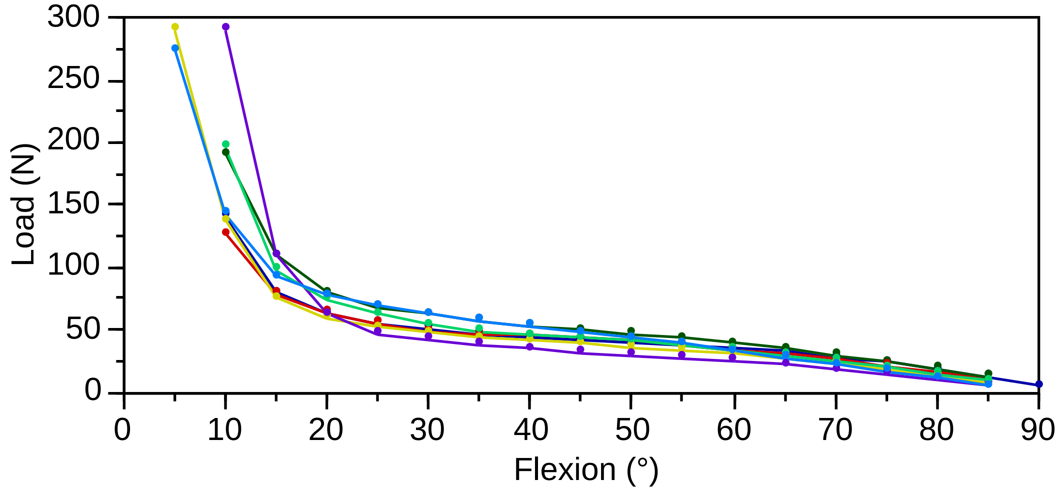

Figure 3.31 Quadriceps Load Versus Flexion For ACL-Deficient Porcine Knees

3.3.3.2 Joint Kinematics

3.3.3.2.1 Intact Human Versus Porcine Knees

The kinematic behavior of intact human and porcine knees was similar in some regards, and different in others. In particular, the species displayed similar kinematic patterns in medial/lateral (M/L) translation and compression/distraction. Differences became apparent when comparing abduction/adduction, I/E rotation, and A/P translation.

Differences between the species in terms of abduction/adduction were marked (Figure 3.32). Human specimens began the movement near 90°, defined as a condition where longitudinal axes of

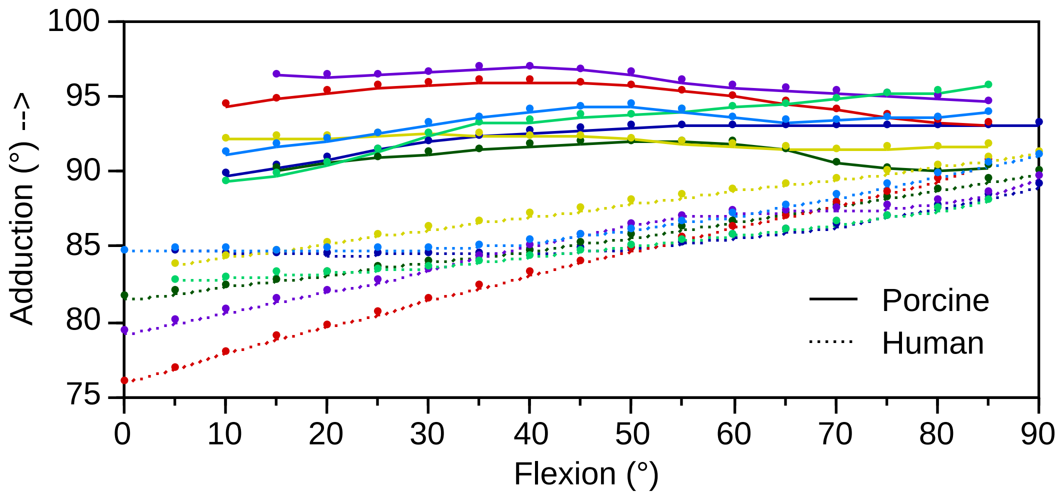

Figure 3.32 Abduction/Adduction Versus Flexion For Intact Human and Porcine Knees

the femur and tibia are in parallel planes. As the knee was extended, the joint became more abducted, until reaching a maximum level at terminal extension. The amount of abduction/adduction in porcine specimens, on the other hand, remained fairly constant throughout the range of motion.

Differences between the species in I/E rotation were also remarkable. Human specimens rotated internally as the knee moved from 90° flexion towards greater extension, until approximately 15-30° flexion (Figure 3.33). At this point, the knee externally rotated as terminal extension was approached, following a pattern commonly known as the "screw home mechanism." Porcine specimens exhibited an almost mirror image pattern (Figure 3.34). Moving from higher degrees of flexion to lower degrees, the knee rotated externally. Then, during the last 15-25° of flexion, the knee rotated internally.

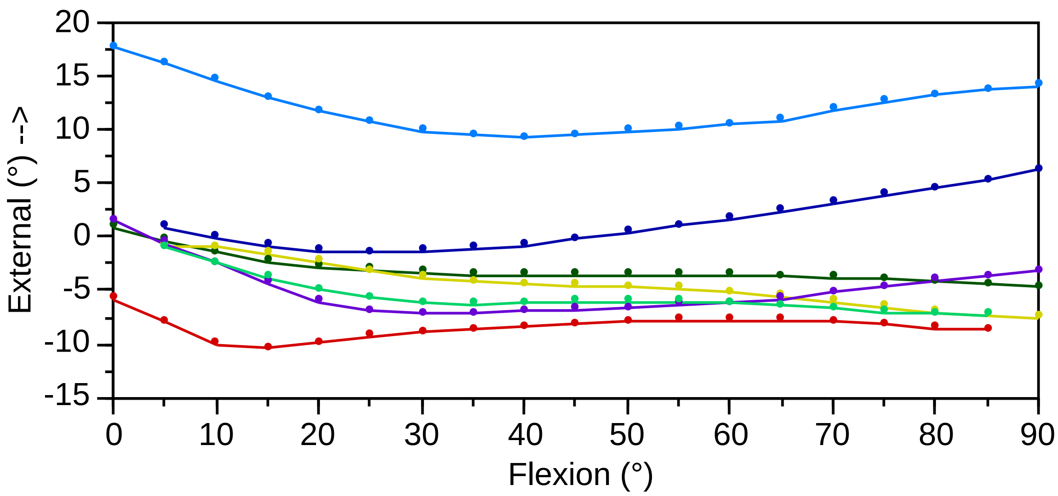

Figure 3.33 Internal/External Rotation Versus Flexion For Intact Human Knees

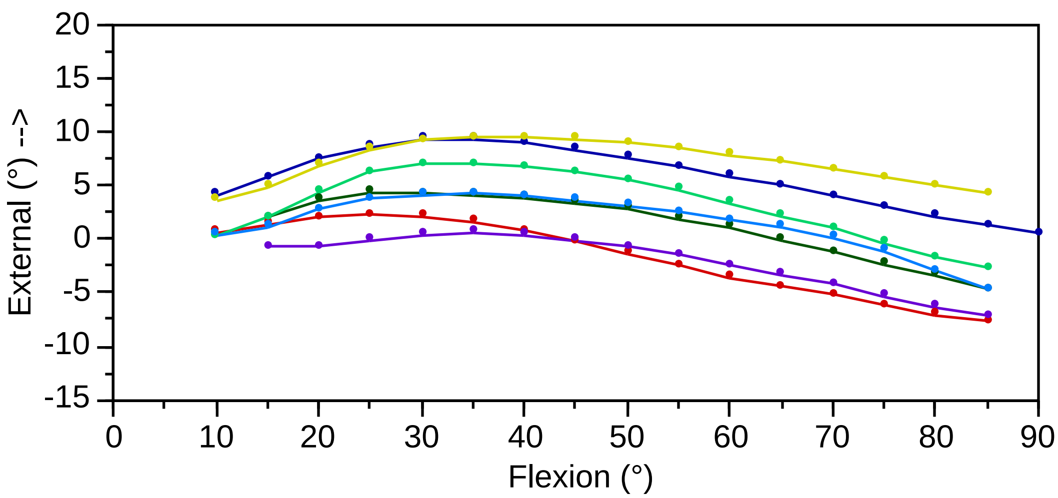

Figure 3.34 Internal/External Rotation Versus Flexion For Intact Porcine Knees

The final difference in kinematic behavior between the species was seen in A/P translation. When a typical human specimen was extended from approximately 60° to 30° flexion, the tibia translated anteriorly with respect to the femur (Figure 3.35). As more extension was then reached, the tibia translated posteriorly. In a typical porcine specimen, little A/P translation was seen until the final 15° near terminal extension, when the tibia translated anteriorly with respect to the femur (Figure 3.36).

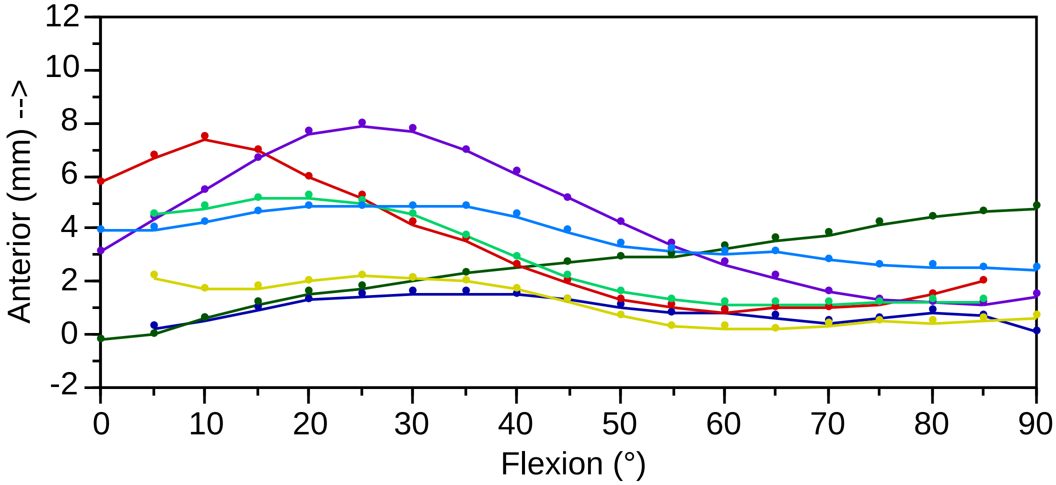

Figure 3.35 Anterior/Posterior Translation Versus Flexion For Intact Human Knees

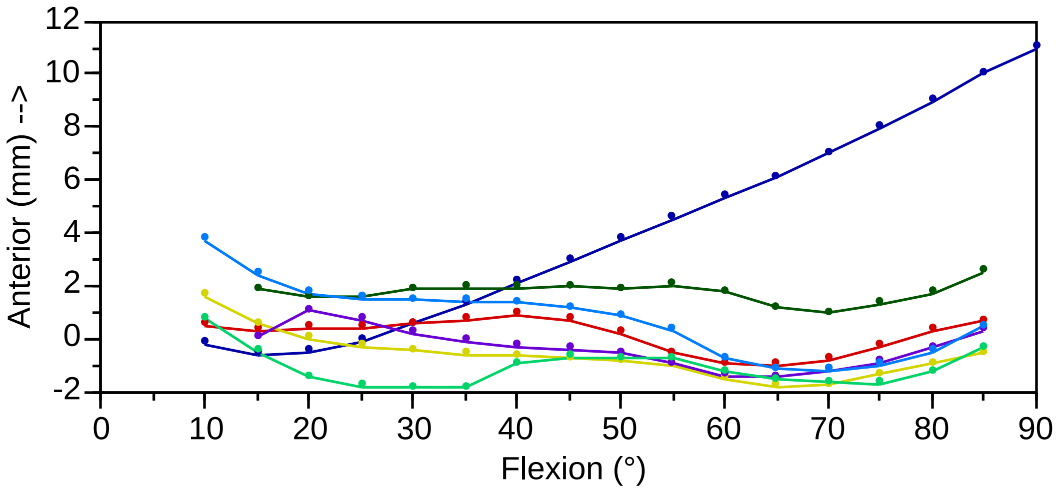

Figure 3.36 Anterior/Posterior Translation Versus Flexion For Intact Porcine Knees

3.3.3.2.2 Intact Versus ACL-Deficient Human Knees

Human knees exhibited no change in abduction/adduction or compression/distraction behavior following transection of the ACL. However, changes did occur in A/P translation, I/E rotation, and medial/lateral translation. Values of translation and rotation on the following graphs are depicted relative to the intact specimen (i.e. a value of 0 corresponds to the value of the intact specimen at that flexion angle).

Following transection of the ACL, all human specimens displayed more anterior translation at terminal extension when compared to intact values (Figure 3.37). Most specimens exhibited this shift throughout the range of flexion. However, some were shifted in a posterior direction at higher degrees of flexion.

All human specimens showed greater internal rotation near terminal extension following transection of the ACL (Figure 3.38). At higher degrees of flexion, some specimens were more externally

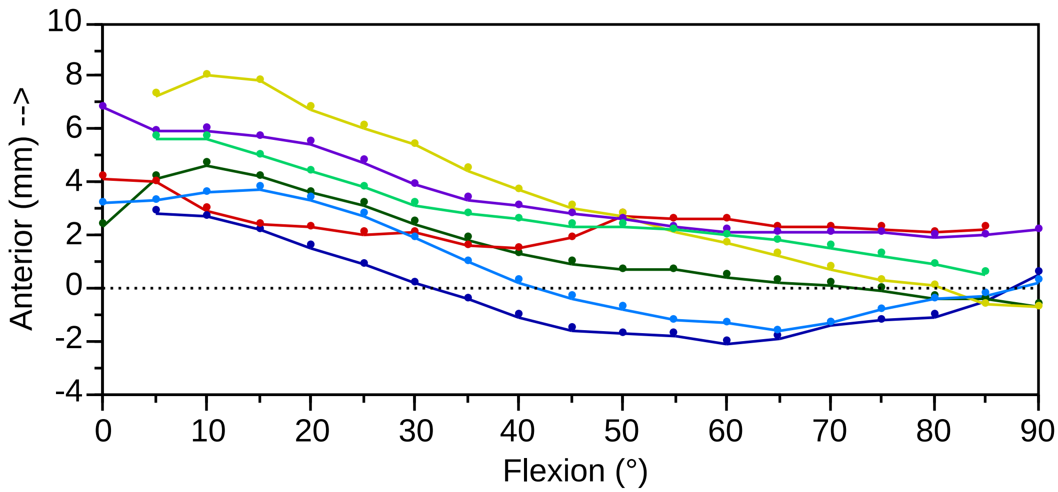

Figure 3.37 Relative Anterior/Posterior Translation Following Transection of the ACL in Human Knees

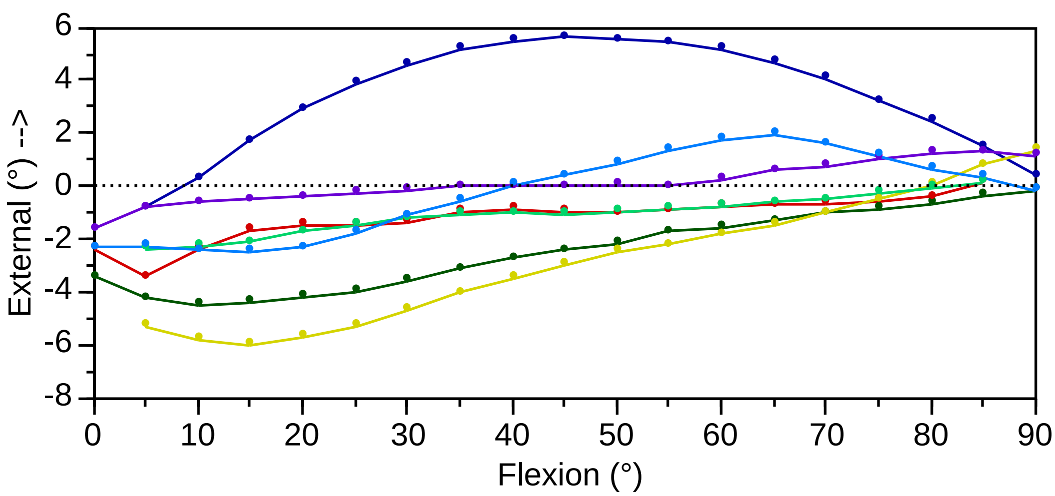

Figure 3.38 Relative Internal/External Rotation Following Transection of the ACL in Human Knees

rotated, and some were more internally rotated. One specimen in particular remained significantly more externally rotated at all degrees of flexion except near terminal extension.

M/L translation behavior of human specimens following transection of the ACL diverged to one of two distinct patterns (Figure 3.39). Four specimens were medially translated throughout the range of flexion, while three were laterally translated. Following detailed examination of each specimen, a plausible explanation is not apparent for this separation.

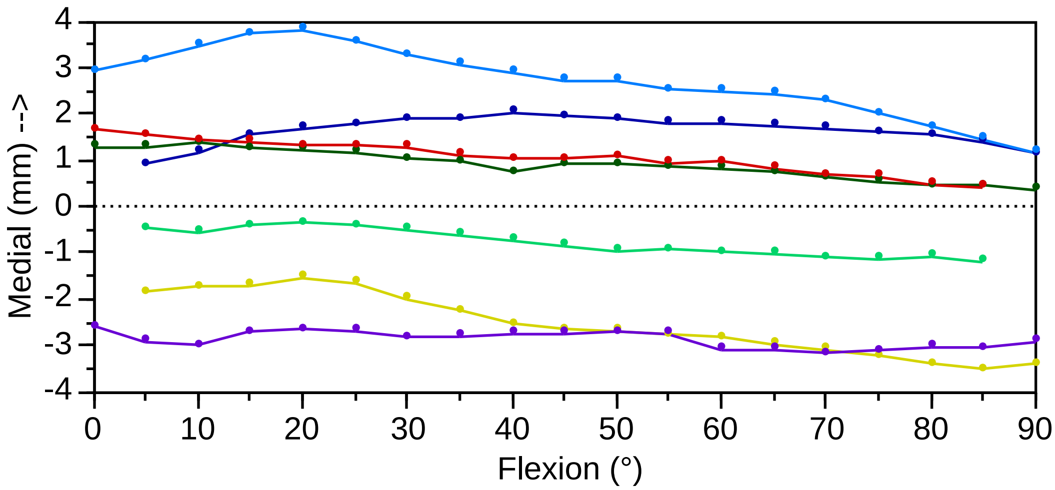

Figure 3.39 Relative Medial/Lateral Translation Following Transection of the ACL in Human Knees

3.3.3.2.3 Intact Versus ACL-Deficient Porcine Knees

Porcine knees exhibited no change in M/L translation, abduction/adduction, or compression/distraction behavior following transection of the ACL. Changes did occur in A/P translation and I/E rotation. Values of translation and rotation on the following graphs are again depicted relative to the intact specimen.

Following transection of the ACL, all porcine specimens exhibited a posterior translation at higher degrees of flexion (Figure 3.40). At approximately 50°-65°, the translation became anterior. Differences from the intact specimens were greatest between 30° and 50° flexion. As greater amounts of extension were reached beyond this zone, the divergence from the intact specimens became less.

At greater extension, ACL-deficient porcine specimens were more externally rotated than were intact specimens (Figure 3.41). At higher degrees of flexion, a consistent pattern was not seen. Specimens were either internally or externally rotated with respect to intact values.

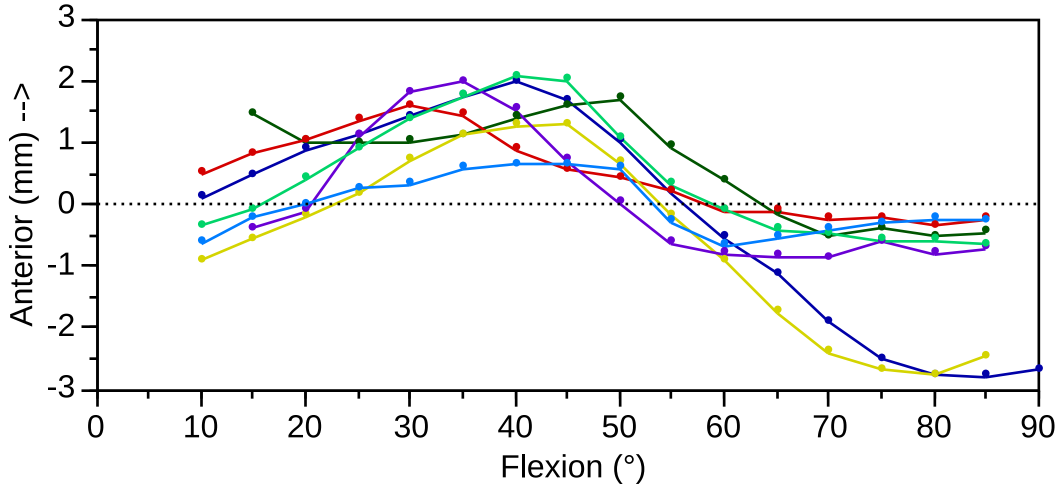

Figure 3.40 Relative Anterior/Posterior Translation Following Transection of the ACL in Porcine Knees

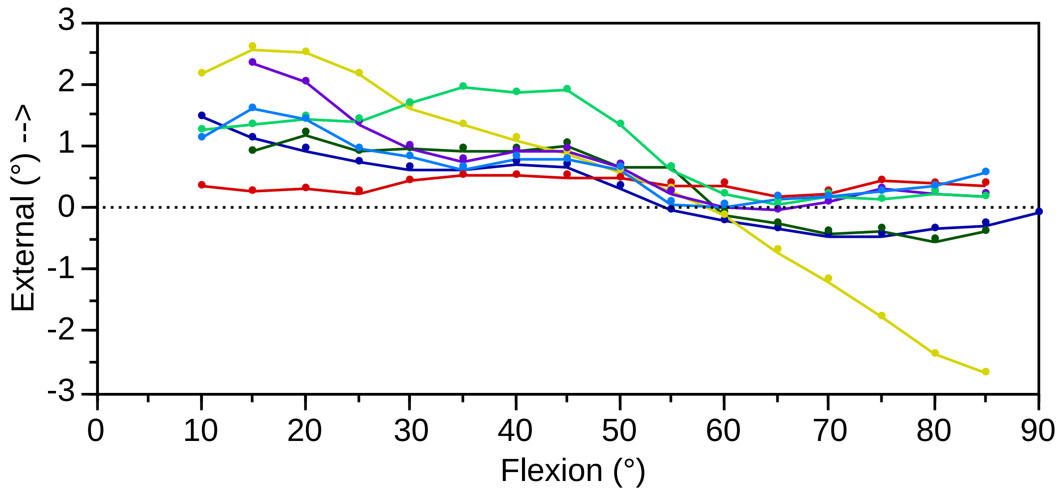

Figure 3.41 Relative Internal/External Rotation Following Transection of the ACL in Porcine Knees

3.3.4 Discussion

The most apparent kinematic difference between human and porcine knees is the flexion angle at terminal extension. According to Fuss [15], porcine knees reach terminal extension at a flexion angle of about 30°, whereas humans reach that point at about 0° flexion. This study found that terminal extension for a typical porcine knee was approximately 10°-15° flexion. The discrepancy between this finding and that of Fuss is possibly due to the surrounding tissue structure. The earlier study investigated porcine knees with all of the surrounding tissue intact, while the current study used specimens with the tissue removed. Because this tissue is quite bulky in the pig, its presence may limit extension of the joint. Since the animals function with the tissue in place, it may be reasonable to consider terminal extension of the porcine knee to occur at about 30° flexion, and to discount kinematic findings at greater extension.

Two noticeable anatomic differences between human and porcine knees are the shapes of the patella and the tibial tubercle. Whereas the human patella is relatively flat, the porcine patella is more like a pyramid in shape, and the porcine tibial tubercle extends further anteriorly than the human one. These differences in anatomy likely play a role in the differing quadriceps load patterns noted between the species. For instance, the moment arm of the quadriceps tendon during extension, and hence the load required to develop a given extension moment, is directly influenced by tubercle size and patellar position.

Anatomic variances also affect the direction of the force applied to the tibia by the quadriceps tendon, which may explain the different A/P translation behavior noted between intact knees of the species. At higher degrees of flexion in human knees, the patella and tubercle are positioned such that the quadriceps force is applied to the tibia with an anterior component. Thus, the tibia is translated anteriorly in this range of flexion. Below approximately 30° flexion, the quadriceps force is directed to the tibia with a posterior component, and the tibia translates posteriorly as terminal extension is approached. A different situation arises in the porcine knee. In this case, due to the size of the patella and the limited extension, the vector of the quadriceps force includes an anterior, not a posterior, component near terminal extension.

The difference in abduction/adduction behavior between human and porcine knees is likely due to two-legged versus four-legged gait and stance requirements, with respect to the angle the femur makes with the pelvis in the frontal plane. In the human, to more easily balance on one leg during locomotion, the distal end of the femur is close to the midline of the body. However, the proximal end is farther from the midline to provide a stable base for the body. Hence, with the knee at full extension, the tibia must be abducted to be nearly vertical. At higher degrees of flexion, the tibia is more in line with the femur (i.e. less abducted than at full extension). Because the pig does not use a two-legged gait pattern, the porcine femur is not inclined to the midline as in the human. As a consequence, the tibia does not need to be abducted. Instead, abduction of the tibia remains near 90°-95° throughout the range of flexion in the pig.

Differences in I/E rotation behavior come about from separate needs of the species. In the human, the “screw home mechanism” is thought to contribute to stability of the joint at full extension. As the tibia externally rotates, the collateral ligaments are tightened and provide stability without the need for excessive muscular force. On the other hand, when a pig is in a standing position, the knee is flexed at about 30°. The tibia is externally rotated at this point, providing stability in the same way as in the human knee. However, when the pig reclines, and the knee is brought to higher flexion angles, the tibia may be internally rotated to fold the lower leg protectively under the animal’s body. The human does not have this requirement.

The ACL plays a role in limiting anterior translation in both species, as evidenced by changes seen in kinematic patterns following transection of the ligament. Transecting the human ACL led to greater anterior translations, particularly near terminal extension, as previously reported [5,8,13,17,30,39,41]. A similar result was seen in the porcine knee.

The ACL also plays a role in limiting I/E rotation of the tibia, although in a different manner in each species. In humans, the ACL limited internal rotation near terminal extension, as previously reported [14,17,30,42,44]. In pigs, the ACL limited external rotation near terminal extension. This finding suggests that the line of action of force applied by the ACL may be different in each species. In particular, the porcine ACL may be aligned such that forces developed in the ligament are applied to the lateral part of the tibia, while in the human, the forces may be more medial.

None of the findings in this study preclude the use of the porcine model to investigate ACL reconstruction in humans. To the contrary, following transection of the ACL, specimens from both species displayed similar changes in A/P translation, the kinematic variable most affected by the ACL. Although I/E rotation patterns were different between the species, it is unlikely that any other four-legged model would be better than the pig in this regard.