ESTIMATION OF ACL FORCES UTILIZING A NOVEL NON-INVASIVE METHODOLOGY THAT REPRODUCES KNEE KINEMATICS BETWEEN SETS OF KNEES

by

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ABSTRACT

ESTIMATION OF ACL FORCES UTILIZING A NOVEL NON-INVASIVE METHODOLOGY THAT REPRODUCES KNEE KINEMATICS BETWEEN SETS OF KNEES

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The force of the anterior cruciate ligament (ACL) in response to simple loads designed to simulate clinical exams, such as a pivot shift test or an anterior drawer test, have been quantified *in vitro*. In addition, force transducers have also been used to estimate the forces in the ACL *in vivo* in animal models, but to date there are few non-invasive non-contact methods for use in human subjects. The specific aim of this study is to evaluate the feasibility of a non-invasive, non-contact methodology for estimating force in the ACL by reproducing average kinematics in 6-degrees of freedom (DOF) from one set of porcine knees (source) onto a separate set of porcine knees (target). A non-invasive, non-contact methodology to estimate forces in the ACL in response to *in vivo* knee kinematics will allow surgical procedures and rehabilitation protocols to be improved.

Source kinematics were collected in response to an anterior load of 100 N and a valgus load of 5 Nm at 30°, 60°, and 90° of knee flexion. After kinematics were collected for eight source knees, the average of these kinematics were calculated. The *in situ* force in the ACL of the target knees in response to reproducing the average kinematics was compared to the *in situ* force in the ACL of the source knees from the applied loads.

A significant difference in the *in situ* force in the ACL between the source knees and the target knees was found for all flexion angles in response to an anterior load and at 60° of knee flexion for valgus loading. These differences can be attributed to the variations in the coupled motions of individual knees due to the applied loads compared to the average kinematics of the source knees. In other words, when average kinematics are computed, variations in knee laxity cause coupled motions to be eliminated or reduced, artificially constraining the motion of the knee. However, knees with similar anterior knee laxity (length of toe region from load-displacement curve) had similar coupled internal-external rotations. Therefore, this study provides promising evidence that this innovative methodology can be extended to account for knee laxity and coupled motions by matching cadaveric knees to groups of subjects with similar anterior and internal-external knee laxity throughout the range of flexion-extension. Thus, an accurate understanding of the forces in the ACL and ACL graft during *in vivo* activities may be obtained using this methodology.

FOREWORD

First and foremost, I would like to thank my wife for being so supportive and longsuffering with me while I was finishing my master's degree. I would like to dedicate this thesis to her.

Second, I am thankful to the members of my thesis committee, Dr. Richard Debski, Dr. Freddie H. Fu, Dr. Rakié Cham, and Dr. Savio L-Y. Woo for reading my thesis and for their advice and suggestions. I would also like to recognize Dr. Richard E. Debski not only for his part in my thesis, but also for his patience with me while I made mistakes. I learned from him not only how to be a successful scientist and engineer but how to work with people and balance home life and work. Seldom have I seen a man with such dedication and a belief that yes, the lab, and his job are important but people are more important.

Third, I would like to thank Jorge Gil for teaching me about the robot and for being such a patient and gracious teacher. Jorge Gil went the extra mile coming in on weekends and early in the morning to teach me about the robot and to finish the program he had created.

I would also like to thank Dr. Woo for his advice. I have learned from Dr. Woo and Dr. Debski to be a critic first and foremost of my own work, and second to evaluate the scientific work of others with a critical eye. This is one of the greatest skills a scientist can develop and I thank them for teaching me these skills and for encouraging me to use them in a tactful and gracious manner.

I would also like to thank the whole MSRC, the ACL group and my family. In particular, I would like to thank my father, brother, and sister for their financial and emotional support.

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ACRONYMS

ACL	Anterior Cruciate Ligament
AM	Anteromedial
DOF	Degrees of Freedom
PL	Posterolateral
UFS	Universal Force Moment Sensor

1.0 MOTIVATION

The ACL is the most frequently injured ligament within the knee. This injury affects roughly one in 3,000 people each year in the United States [1-3]. Despite the large number of ACL reconstructions that are performed every year around the world, these procedures are neither able to fully restore intact knee kinematics nor fully reproduce the force in the intact ACL [4-13]. The force of the ACL in response to simple loads designed to simulate clinical exams, such as a pivot shift test or an anterior drawer test, have been quantified *in vitro* [14-18]. However, the literature provides contradictory and, in many cases, uncertain conclusions as to the force and function of the ACL *in vivo* [19-28]. In particular, there is little scientific data available to guide postoperative rehabilitation, which can range from conservative to aggressive. There is also a lack of clarity as to which rehabilitation exercises and activities of daily living, i.e. closed chain vs. open chain [22, 28], squat vs. knee extension [23, 24, 28], and stair climbing vs. walking [25-27], best restore the function of an intact knee postoperatively.

The force data obtained in future studies from reproducing average kinematics will elucidate how well the ACL graft is performing during various rehabilitation regimens and activities of daily living. The force data obtained from the ACL reconstructed knee will be compared to that of the intact ACL – the "true gold standard of ACL reconstruction." This information will be used to suggest directions for improvements of ACL reconstruction, not only in terms of graft fixation and placement but also in terms of postoperative rehabilitation regimens that are scientifically based. Finally, from reproducing average kinematics, it will be possible to provide force validation for a realistic computational model of the knee. Once the model is validated with the experimental data obtained from a high-payload robotic/UFS testing system, it can be used to determine the forces and force distribution in the ACL in response to more complex dynamic *in vivo* activities.

2.0 BACKGROUND AND SIGNIFICANCE

First, a brief description of the anatomy and function of the knee and ACL will be described. Second, biomechanics of the ACL will be discussed. Third, previous methodologies used to measure *in vivo* force in the ACL will presented. Next, the robotic/UFS testing system will be introduced. Finally, preliminary studies, which lead us to adopt the current methodology for reproducing in vivo kinematics, will be described.

2.1 Anatomy

An understanding of the anatomy of the knee and the ACL are important for understanding the function of the ACL and for ultimately improving reconstruction procedures and rehabilitation protocols. This section of the thesis will describe the anatomy of the knee and ACL. Finally, a review of the most recent findings of the function of the ACL and its bundles will be discussed.

2.1.1 Anatomy of the Knee

The bones of the knee consist of the femur, tibia, fibula, and patella. The femur is the longest bone of the body and extends from the hip joint to the knee joint. At the hip joint, the femoral head fits within the acetabulum, while at the knee joint, the femoral condyles glide upon the plateau of the tibia. The tibia extends from the knee joint to the talus or ankle. The fibula runs parallel to the tibia on the lateral side.

The patella, a sesamoid bone within the quadriceps and the patellar tendon, glides within the trochlear groove of the femur and comprises the patellar-femoral joint [29].

The lateral and medial menisci of the knee conform to the surface of the tibial plateau. The lateral meniscus is circular, while the medial meniscus is more oval. There are four main ligaments of the knee (**Figure 1**): the anterior cruciate ligament (ACL), the posterior cruciate ligament (PCL), the medial collateral ligament (MCL) and the lateral collateral ligament (LCL). The ACL and the PCL cross over one another inside the joint to connect the femur to the tibia.

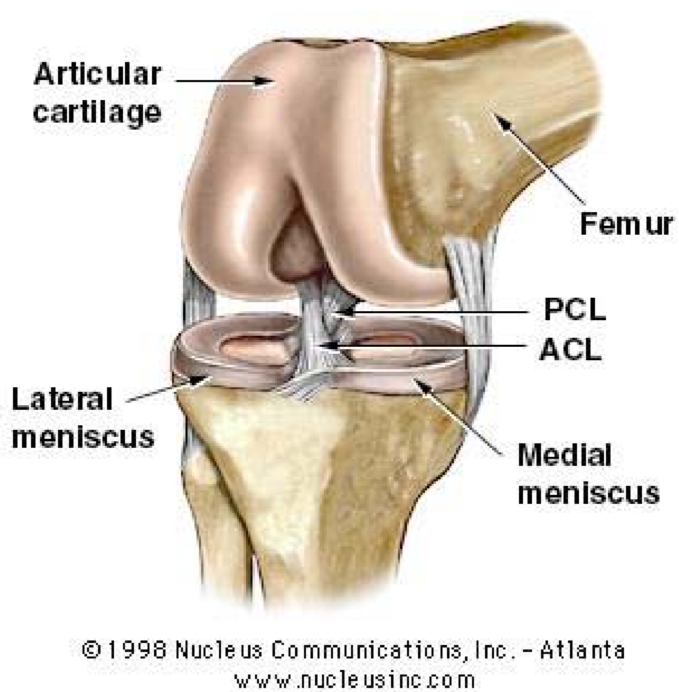

Figure 1. Schematic of the soft tissues of the knee joint including the ligaments and meniscus

The ACL is the primary restraint to anterior translation of the tibia with respect to the femur, while the PCL is the primary restraint to posterior tibial translation. The MCL is located on the medial aspect of the knee, connecting the femur to the tibia; it primarily restrains valgus rotation of the tibia. The lateral collateral ligament connects the lateral aspect of the femur to the head of the fibula and is the primary restraint to varus rotation of the tibia. The many muscles of the knee provide dynamic stabilization to the knee joint. The main muscle groups can be divided into those that produce flexion or extension of the knee. The extensor muscles are known as the quadriceps, which include the rectus femoris, the vastus medialis, the vastus lateralis, and the vastus intermedius. The flexor muscles are known as the hamstrings and include the gracilis, semimembranosus, semitendinosus, and sartorius.

2.1.2 Anatomy of the Anterior Cruciate Ligament

The anterior cruciate ligament is an intra-articular ligament, which means it is located inside the articular surface of the knee joint. It is also extrasynovial, and therefore located outside of the synovium, a surrounding protective layer of synovial fluid.

The 3-4 cm long ACL has a cross-sectional area of roughly 37 mm^2 at the mid-substance that varies little with knee flexion [30] The insertion of the ACL expands to 3.5 times its cross-sectional area at the mid-substance as it directly inserts into the posteromedial aspect of the lateral femoral condyle and the anteromedial tibial plateau. It is angulated in the sagittal plane to allow the ACL to function to restrict anterior tibial translation, internal tibial rotation, and valgus tibial rotation. The semi-circular femoral insertion, longer in the superior inferior direction than in the anteroposterior direction, has an average cross sectional area of 113 mm^2, while the average cross sectional area of the oval tibial insertion, longer in the anteroposterior than mediolateral direction, is 136 mm^2 [31]. Blood supplied to the ACL is shared with the PCL through the middle genicular artery. The nerve supply is also shared with the PCL using the popliteus plexus of nerves.

2.2 Function of the Anterior Cruciate Ligament

The function of diarthrodial joints is mediated by the complex interactions of bones, ligaments and capsule, articular cartilage, and muscle. The interdependence of these structures is such that severe injury or failure of any one of them can lead to deterioration of the others and then the disruption of overall joint function. Ligaments are particularly vulnerable with estimates of the annual rate of ligament injury in North America ranging from 5-10% of all people up to the age of 65 years [32]. Injuries to these soft tissues include frequent sprains as well as complete rupture.

The ACL has been shown to resist excessive anterior tibial translation (ATT), as well as internal and valgus rotation [14, 33-37]. The force distribution in the ACL can be defined as the percentage of load sharing between the two functional bundles of the ACL. The ACL consists of two main bundles, the anteromedial (AM) bundle, and the posterolateral (PL) bundle. The posterior lateral bundle has higher forces with knee extension, while the anterior medial bundle will carry a higher percentage of the forces with knee flexion. The larger, shorter AM bundle is generally taut during passive knee flexion, while the PL bundle is relatively taut in passive knee extension [14, 38]. Both bundles of the ACL carry load during valgus rotation at 15° and 30° of knee flexion [38].

The anterior tibial translation and *in situ* force in the ACL were also examined following the application of a 200 N compressive as well as a 100 N AP load [39]. These loading conditions resulted in a decrease in the total anterior-posterior tibial translation and a significant increase in the anterior tibial translation coupled with an even smaller decrease in the posterior tibial translation. High compressive loads in the knee cause elevated *in situ* forces in the ACL, suggesting that high axial compressive loads to the knee without muscle contraction should be avoided during rehabilitation.

2.3 Biomechanics of the ACL

Ligaments are highly specialized connective tissues that connect bones and transfer forces to mediate smooth movement of diarthrodial joints during normal activities. They also limit excessive displacements between the bones at high external loads. Ruptures of ligaments due to excessively high loads experienced during sports and accidents can upset the dynamic balance between the mobility and stability of a joint and result in abnormal kinematics. This can potentially cause damage to other soft tissues of the joint and eventually lead to pain, morbidity, and osteoarthritis.

For young and active individuals participating in sports activities, the anterior cruciate ligament (ACL) of the knee is especially susceptible to frequent injury. Most ACL tears do not heal and require surgical reconstruction [40, 41]. The results of surgical reconstructions are generally successful and most patients can resume normal activities and return to participating in sports [42, 43]. However, 15-25% of patients have experienced less than satisfactory results at both short and long-term follow-ups [8, 44-47]. While there are numerous factors that contribute to these failures, recent biomechanical studies have helped to gain a better understanding of the complex function of the ACL as well as the function of its replacement grafts. There are also ongoing studies such as this thesis that aims to further enhance this knowledge in order to improve outcomes for patients.

2.3.1 ACL Reconstruction

In spite of the large number of ACL reconstructions that are performed each year around the world (estimated between 75,000 to 100,000 cases in the United States alone), these procedures continue to encounter post-operative problems [48]. Recent literature on long-term follow-up between five and ten years has revealed that 15 – 25% of patients have unsatisfactory results [8, 11, 13, 35]. In terms of ACL surgery, there is much debate over the use of different autographs [49-54], the selection of fixation devices [8, 55-61], and tunnel placement[42, 62, 63].

Our research has found that current ACL reconstruction procedures are insufficient in resisting rotatory loads applied to the knee and that the medial meniscus plays an important role in ACL reconstructions.

A 134 N anterior tibial load applied to an ACL-reconstructed knee with a quadruple semitendinosus/gracilis tendon (QSTG) graft and a bone-patellar tendon-bone (BPTB) graft showed anterior tibial translation (ATT) of 166±33% and 140±32% of the intact knee, respectively. When the same grafts were subjected to a combined internal tibial torque of 10 N-m and valgus torque of 10 N-m, the ATT was 192±52% and 171±48% of the intact knee, respectively. The results demonstrated that both reconstructions were successful in limiting anterior tibial translation under anterior tibial loads. However, in response to a combined rotational load, neither of the two reconstructions were effective in reducing anterior tibial translation. The placement of the QSTG and BPTB graft between the AM and PL bundles made them incapable of providing rotatory stability because they were close to the rotational center of the knee [4]. Furthermore, the mean in situ forces in the grafts under a 134-N anterior tibial load were restored to within 78% to 100% of that in the intact knee.

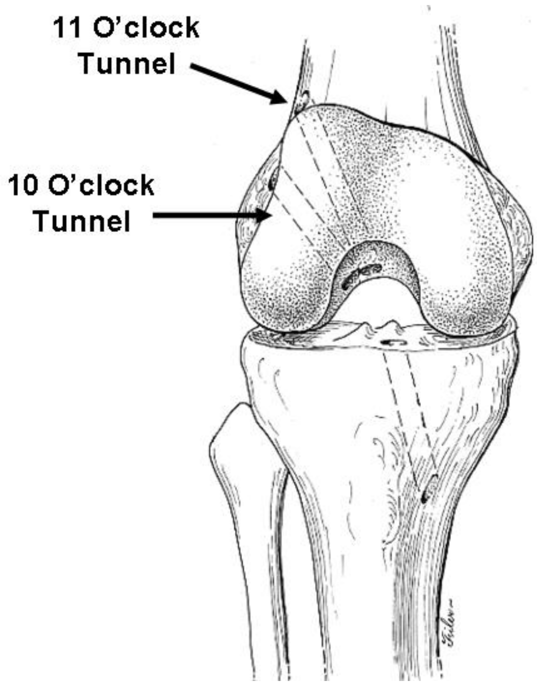

Figure 2. Two possible tunnel locations for ACL reconstruction are shown on a right femur. The ten o'clock tunnel position is lateral to the eleven o'clock position

Thus, a study of the effectiveness of moving the tunnel laterally and away from the rotational center of the knee to improve the function of the ACL graft was performed [64]. A BPTB graft was placed in the femoral tunnel at the 10 and 11 o'clock positions (**Figure 2**). The 10 o'clock position approximates the femoral insertion site of the PL bundle of the ACL while the 11 o'clock position is close to the insertion site of the AM bundle. In response to an anterior tibial load, there were no significant differences in ATT or *in situ* forces between the intact and reconstructed knee for both the 10 and 11 o'clock positions. However, under rotatory loads, there was significantly higher ATT in the reconstructed knees. The *in situ* force in the replacement graft fixed at the 10 o'clock position was significantly higher than the *in situ* force in the graft fixed at the 11 o'clock position. These results indicate that, under rotatory loads, the 10 o'clock position restores function more effectively than the 11 o'clock position, but neither position has been shown to restore the knee to its intact state.

Therefore, it is believed that in order to restore the function of both bundles of the ACL, it may be necessary to consider a more anatomical reconstruction. Using a QSTG graft, both the traditional and anatomical reconstructions were studied [65]. For the anatomic reconstructions, one tibial tunnel and two femoral tunnels, based on the insertion sites of the ACL bundles, were used. It was found that the anatomically reconstructed knee had significantly lower ATT than the single bundle reconstruction (with femoral tunnel at 11 o'clock). However, both the traditionally and anatomically reconstructed knees experienced significantly higher ATT than the intact knee. The anatomically reconstructed knee better restored the *in situ* force of the intact ligament than the traditionally reconstructed knee (**Figure 3**).

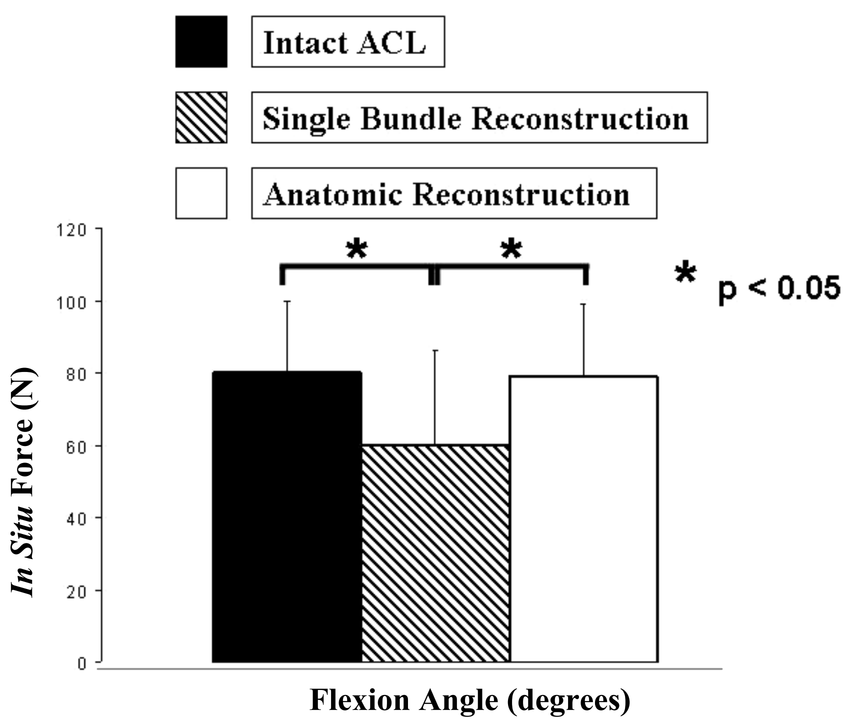

Figure 3. In situ force (mean±SD) in the intact ACL and the single bundle and anatomical reconstructions in response to a combined rotatory load at 15° of knee flexion [65]. (Adapted from Yagi et al. 2002, permission requested)

Thus, it has been shown that an anatomic ACL reconstruction replacing both the anteromedial (AM) and posterolateral (PL) bundles could restore normal knee kinematics and *in situ* force of the ACL more closely than the popular single-bundle reconstruction procedures when the graft is placed at the 11 o'clock femoral tunnel position[65]. In addition, it was demonstrated that using a single ACL graft placed at the 10 o'clock femoral tunnel position is more effective in resisting rotatory loads than those at the 11 o'clock position [66]. Therefore, the objective of the following study was to compare two reconstruction procedures, i.e. anatomic vs. a more lateral graft placement at 10 o'clock femoral tunnel position, which is close to the insertion site of the PL bundle. A 134 N anterior tibial load at knee flexion angles of full extension (FE), 15°, 30°, 60° and 90° and a combined rotatory load of 10 N-m valgus and 5 N-m internal tibial torque at knee flexion angles of 15° and 30° were applied to knees with both the single-bundle reconstruction placed at 10 o'clock position on the femoral side and the double-bundle anatomical reconstruction. Results demonstrated that under anterior loading the anatomic double-bundle reconstruction restored intact knee kinematics and *in situ* force more closely than 10 o'clock single-bundle reconstruction, especially at high flexion angles. However, in response to rotatory loads, 10 o'clock single-bundle reconstruction was as effective at restoring intact knee kinematics and *in situ* force as the anatomical reconstruction. Because the 10 o'clock femoral tunnel position is close to the PL bundle insertion and is located more laterally than the 11 o'clock position, it contributes to the rotatory stability of the knee. Since the PL bundle is maximally taut at low flexion angles, the 10 o'clock single-bundle reconstruction could not restrain the ATT at high flexion angles. The anatomic double-bundle reconstruction more successfully restored intact knee kinematics and *in situ* force than 10 o'clock single-bundle reconstruction at high flexion angles, but these differences may not be clinically significant [67].

The resultant force in the meniscus was found to be significantly elevated in ACL deficient knees in response to a 134 N anterior tibial load with 200 N of axial compression. After ACL reconstruction, these forces returned to intact levels. Conversely, with medial meniscectomy, the *in situ* force in the ACL graft increased by approximately 50%. Thus, the ACL and medial meniscus are interdependent. The medial meniscectomy may cause the ACL graft to fail because of excessive force in the graft [68].

Another study focused on the role of the posterior horn of the medial meniscus (PMM) by removing portions of the PMM and its effect on knee kinematics in response to a 10 N-m varus torque. The results suggest that removing one-third of the PMM will have little effect on knee kinematics while removing more than two-thirds of the posterior horn of the medial meniscus will have a significant effect on the kinematics of the knee. Based on these results, removal of one-third of the posterior horn of the medial meniscus may present little risk to the ACL. However, one needs to give serious consideration to the potential risk to other knee structures[69]. These studies indicate the importance of the medial meniscus on restraining knee motion, which will minimize loads transferred to the ACL and other structures of the knee.

The biomechanical analyses of the ACL-reconstructed knees based on cadaveric studies have helped us understand the complexity of the ACL. However, from this series of studies, it is clear that there are many complex issues involved with ACL reconstructions. As the loading conditions applied in these studies are relatively simple and only designed to mimic clinical exams, more realistic *in vivo* loading conditions will need to be added in order to better understand both the function of the intact ACL and the effectiveness of ACL reconstruction. The new scientific data obtained from *in vivo* loading will enable a more thorough evaluation of the function of ACL replacement grafts, improve reconstruction procedures, and guide rehabilitation protocols [5, 22, 25, 70-77].

2.3.2 Rehabilitation Protocols

The quadriceps muscle of patients who undergo ACL reconstruction becomes significantly weakened following surgery [78, 79]. Various postoperative rehabilitation exercises are used to restore quadriceps strength in order for patients to return safely to their previous activities. Some researchers have suggested that closed kinetic chain exercises (closed-chain) rather than open kinetic chain exercises (open-chain) are better at protecting the ACL replacement graft from excessive force [80, 81].

However, the University of Vermont group [77] recently reported that there were only minor differences in the strain values in the ACL replacement graft obtained during closed- and open-chain exercises. Furthermore, in a prospective randomized clinical study, a combination of closed-chain and open-chain training after ACL reconstruction was found to improve patients' quadriceps strength significantly over closed-chain exercises alone [6]. In a more recent study utilizing subjects who frequently lifted weights, an analytical model found that neither closed nor open chain exercises created forces high enough to predispose the ACL graft to premature failure with the exception of the last 25° of extension for open-chain exercises [28]. Nevertheless, the force distribution of the ACL replacement graft during these exercises is still unclear [23], and the need for further data on rehabilitation exercises that involves tibial rotatory motion has been suggested [82, 83].

The University of Vermont group has ranked various rehabilitation exercise protocols based on *in vivo* strains in the AM bundle of the intact ACL from patients [25, 77, 84, 85]. A similar study whereby the forces in the ACL replacement graft are obtained will further the understanding of ACL replacement graft function in response to these rehabilitation exercises. Whether the ACL replacement graft experiences excessive forces during these exercises, which could predispose it to failure during the early healing process, remains unknown [9, 23, 85-90]. Therefore, it is critical to quantify the level of forces in the ACL replacement graft during these activities. The methodology described in this thesis will enable identification of the difference between the *in situ* force and force distribution of the ACL and ACL graft during these rehabilitation exercises. This knowledge will help surgeons and physical therapists select appropriate post-operative rehabilitation protocols on a scientific basis. Moreover, rehabilitation exercises could be designed to strengthen the quadriceps muscles as well as to maintain adequate *in situ* forces in the ACL graft for healing and remodeling that could lead to improved knee function after ACL reconstruction.

2.4 Previous Attempts to Estimate *In Vivo* Force

During the last decade, intensive efforts have been made to quantify the forces in the ACL *in vivo*[19-21, 91, 92]. As a result, various devices and methods have been developed to measure the force in the ACL, including buckle, femoral fixture and pressure transducers, analytical models and ligament scaling methods. With the use of these devices, valuable information on forces experienced by the ligaments has been obtained. However, each of these devices has limitations. Many of these methods have been contact methodologies that may affect the force measured in the ACL by altering the ligament's structure and environment.

2.4.1 Invasive – Contact Methodologies

Contact methodologies such as fixation, buckle, and pressure transducers have been used to measure force in the ACL (**Figure 4**). The fixation transducer relates tension in the ACL graft to strain caused by the deflection of a cantilever beam within the transducer[91, 92]. Buckle transducers measure the force in the ACL by relating strain caused by the deflection of the transducer to tension in the ACL[19, 20]. Pressure transducers are surgically implanted within one of the functional bundles of the ACL relating pressure to force[21]. These devises are invasive and requiring surgical installation. In addition, impingement and decreases in force due to healing often occur when the force transducers are placed within the joint[20, 92]. Finally, force transducers are impractical for estimating force *in vivo* in human subjects because most devices must be calibrated *in vitro*[19, 21].

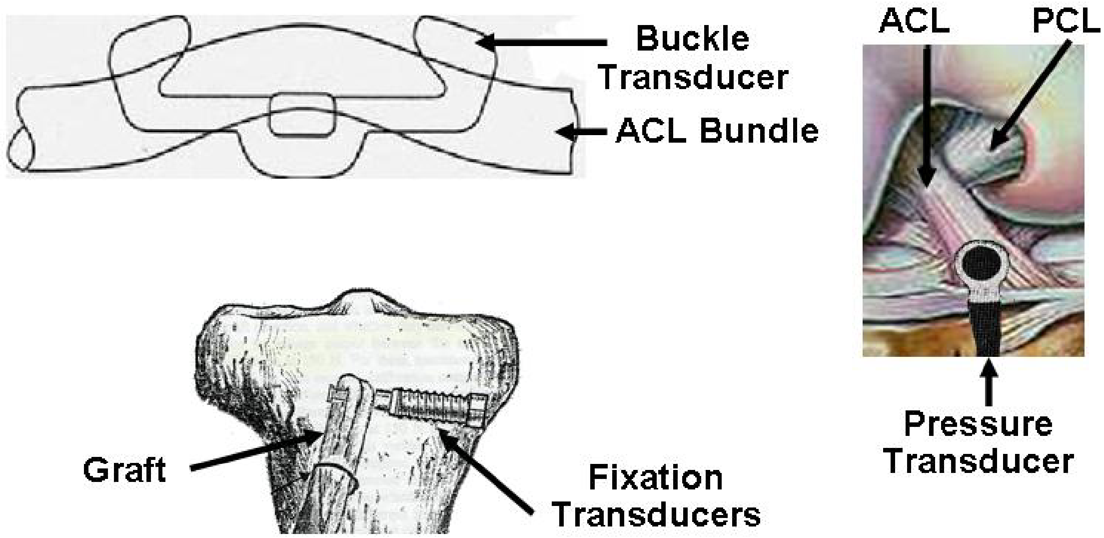

Figure 4. Contact Methodologies for estimating the force in the ACL *in vivo* - Fixation, Buckle, and Pressure Transducers

2.4.2 Noninvasive – Non-Contact Methodologies

Although few in number, there have been previous attempts to develop noninvasive, non-contact methodologies to estimate the force in the ACL. Analytical models have been used to estimate force in the ACL (**Figure 5**). The major limitation of these analytical models is the difficulty of validating these models experimentally[23, 24, 26, 28, 93]. Previous researchers have also attempted to estimate forces in the ACL through a ligament scaling method. The location of the ACL origin and insertion were predicted by mathematically transforming their location from a dissected cadaveric specimen to their respective position in human subjects [94]. These transformations between the ACL origin and insertion were based on anatomical landmarks, which were located by palpation on human subjects. This methodology, although non-invasive and non-contact, was not effective in human subjects due to the difficulty in interpretation and precision of defining anatomical landmarks *in vivo*.

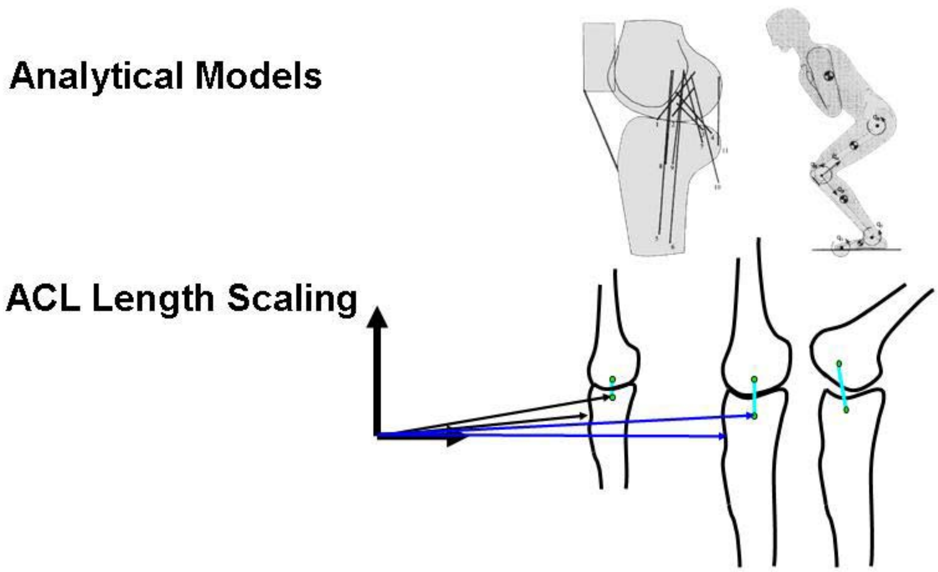

Figure 5. Non-contact methods for estimating the force in the ACL *in vivo* - Analytical Models and ACL Length Scaling Method

2.5 Robotic/UFS Testing System

In our research center, we have successfully used a 6-DOF robotic manipulator in combination with a 6-DOF UFS to measure the *in situ* force in ligaments [95-100]. The advantages of this innovative method are based on the robotic manipulator reproducing positions of the path of knee motion with high fidelity such that the principle of superposition can be employed to calculate changes in force before and after a ligament is transected [95, 98]. Further, the *in situ* force in the ACL is determined without dependence on the specimen geometry, the location of the ligament, or muscle forces and without having a device physically in contact with the ligament. This system is capable of measuring the force and force distribution in the ACL and other soft tissue structures in the same knee specimen, thus eliminating inter-specimen variability and increasing statistical power.

Our research has provided quantitative data on forces and force distribution in both the AM and the PL bundles of the ACL as well as ACL replacement grafts during the anterior drawer test, Lachman test, and simulated pivot shift test using human cadaveric knee specimens [38, 98, 101-103]. In the future, this testing system will be used to reproduce knee kinematics. This thesis is devoted to the partial development of a methodology to estimate the *in situ* force in the ACL by reproducing average kinematics.

2.5.1 Experimental Evaluation of Joint Function

In our research center, two testing systems have been developed that consist of a robotic manipulator combined with a universal force-moment sensor [95, 97, 98, 104]. The low-payload robot (Mitsubishi Electric Corporation, RV-MIS-P2, Nagoya, Aichi, Japan UNIMATE Puma 762) is capable of applying loads up to 300 N while the high-payload manipulator (S-900W, FANUC Robotics North America, Inc, Auburn Hills, MI) can exert loads of 3500 N or more (**Figure 6**). These testing systems allow for multiple axial force and position controls that can be used to apply loads and produce motions of diarthrodial joints in multiple degrees-of-freedom (DOF). The robotic/UFS testing system defines a passive path of flexion extension of the joint. The passive path is the joint motion resulting in zero forces and moments; it serves as a reference position for other loading conditions. In addition, each testing system can record the resulting 6-DOF motions and then reproduce the identical path of motion in a dissected specimen, allowing for the use of the principle of superposition. Since identical loading conditions are applied to the intact and dissected states of the same joint specimen, inter-specimen variability can be minimized and the statistical power increased in each study.

In addition to a brief review of our current development of knee kinematics, the description and control of joint motion and forces/moments with respect to an anatomical reference system for clinical relevance will be described. This includes the determination of the transformations between the axes of all coordinate systems as well as forces in the ligaments of the knee using a non-contact methodology.

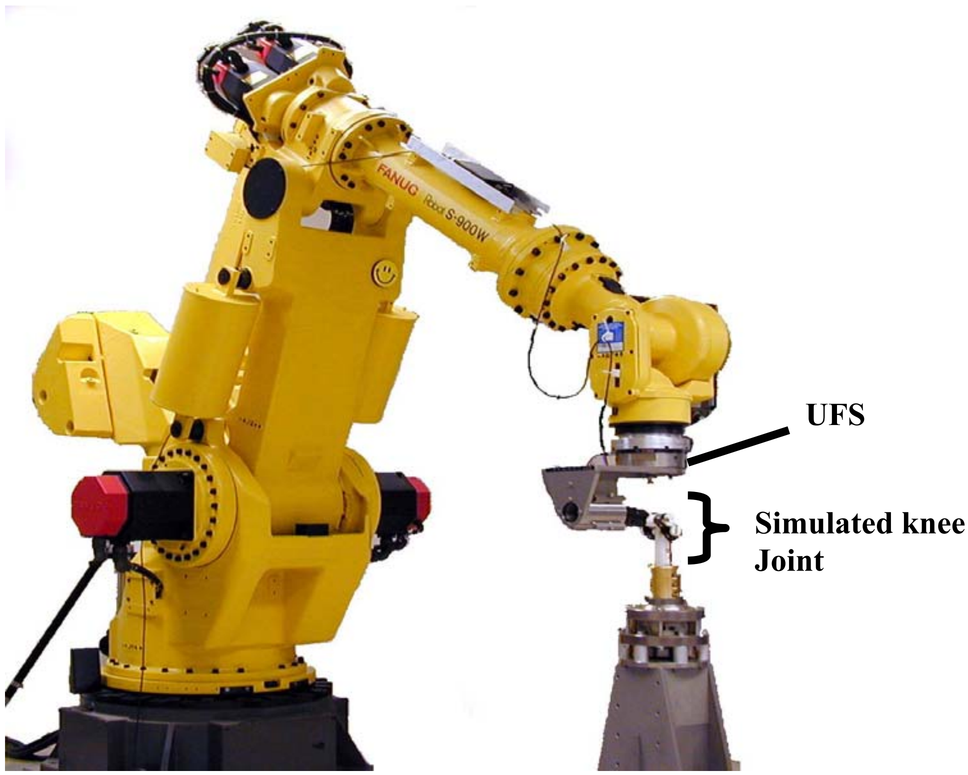

Figure 6. Photograph of the high-payload robotic/UFS testing system showing a medial view of a simulated knee joint mounted between the end-effector and base of the manipulator for testing

2.5.2 Kinematics of the Robotic Manipulator

A robotic manipulator is a tool that can control the location and orientation of its end effector relative to its base (**Figure 6**). This enables the robot to accurately control the location and orientations of an object attached to its end effector and record each location and orientation throughout a path of motion [95]. This can be accomplished using a 4x4-transformation matrix (T) that is composed of rotation (R) and translation components (U), such as the two matrices below:

$$U_0^1 = \begin{bmatrix} 1 & 0 & 0 & \Delta x \\ 0 & 1 & 0 & \Delta y \\ 0 & 0 & 1 & \Delta z \\ 0 & 0 & 0 & 1 \end{bmatrix} \qquad R_0^1 = \begin{bmatrix} \cos\theta & -\sin\theta & 0 & 0 \\ \sin\theta & \cos\theta & 0 & 0 \\ 0 & 0 & 1 & 0 \\ 0 & 0 & 0 & 1 \end{bmatrix} \tag{1}$$

Therefore, the transformation from one link of a multi-link system to another for a serial robot is a combination of these two components. Where the order of operation is translation (U) followed by rotation (R)

$$T_0^1 = \begin{bmatrix} \cos\theta & -\sin\theta & 0 & \Delta x \\ \sin\theta & \cos\theta & 0 & \Delta y \\ 0 & 0 & 1 & \Delta z \\ 0 & 0 & 0 & 1 \end{bmatrix} \tag{2}$$

and

$$P_o = T_o^1 P_1 \tag{3}$$

The robot can then determine the location (P_o) of its end effector or initiate a desired motion using a series of transformations between the end effector and the global coordinate systems. P_o can be described as follows for a six link system [105, 106].

$$P_o = T_o^1 T_1^2 T_2^3 T_3^4 T_4^5 T_5^6 P_6 \tag{4}$$

2.5.3 Application of External Loads to Joint (Force Control)

Once the transformations between the sensor and axes of the joint coordinate system have been established, the application of specified external loads to the knee joint is straightforward using the robotic/UFS testing system. Force (or load) control is used to apply a given external load, or a set of desired forces and moments to the joint. This mode of control is similar to the flexibility method commonly described in the literature [107]. The desired movement of the robotic manipulator is determined by comparing the current forces and moments measured by the UFS to the specified, or target, forces and moments. The robot is then instructed to perform a movement in order to achieve the target forces and moments and the new forces and moments are recorded.

Based on the differences between the target forces and moments and the new forces and moments measured by the UFS, a new movement is calculated and the robot is instructed to move the joint accordingly. This iterative process allows the testing system to move the knee through the appropriate motions such that the specified loads are developed within the joint.

The testing system can apply an identical external force to a specimen in both the intact and ligament-deficient states in force control mode. The difference in the kinematics between the intact and ligament-deficient states can then be determined. These tests are similar to clinical examinations used to diagnose ligament deficiency where the clinician applies a similar load to the uninjured and injured knee to compare the differences in resulting kinematics for diagnostic purposes.

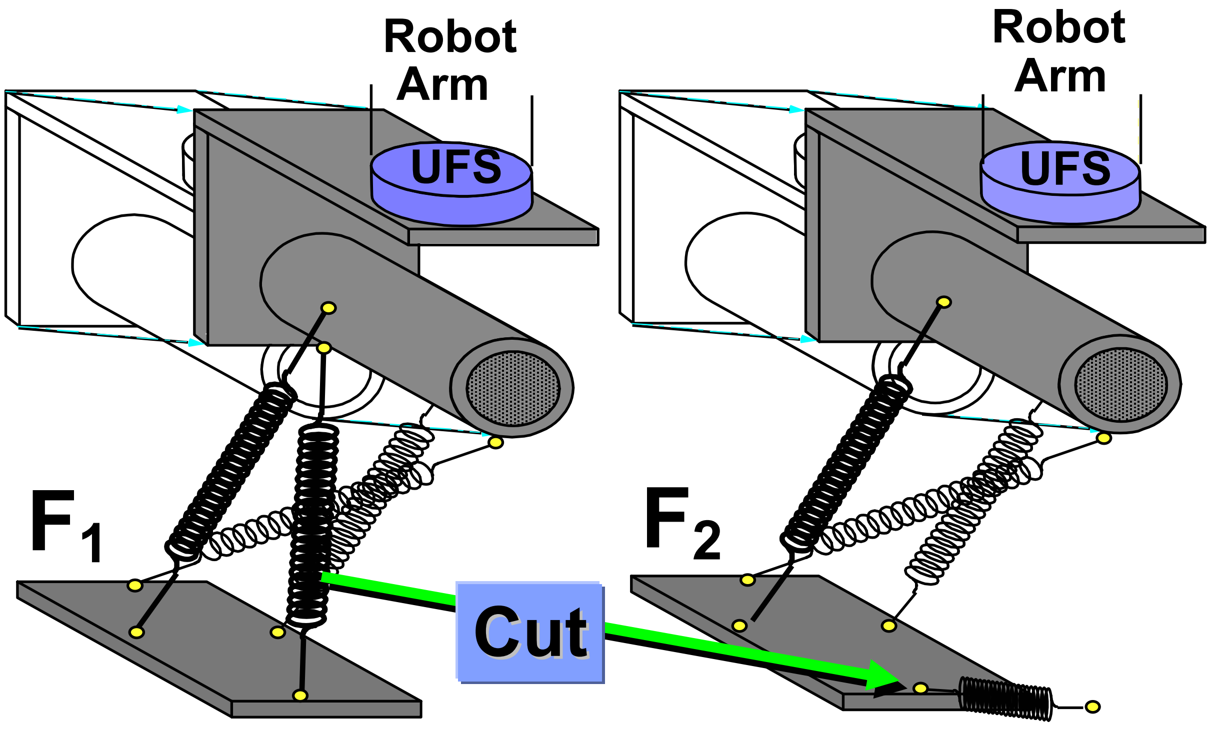

Figure 7. Drawing representing the application of the principle of superposition to determine the *in situ* force in a ligament (represented by the cut spring)

2.5.4 *In Situ* Forces in Ligaments (Position Control)

A detailed understanding of a ligament's function and contribution to overall joint kinematics is dependent on an accurate determination of the *in situ* forces developed in the ligament in response to motion or external loading of the intact joint (**Figure 7**). To accomplish this task, the robotic/UFS testing system is asked to accurately reproduce joint positions (position control). This testing system is capable of not only reproducing the joint positions at the maximum loads but at each intermediate position used to reach the extreme position as well. Therefore, any sequence of joint positions that represents the path of motion of a diarthrodial joint can be reproduced. To determine the *in situ* force in a ligament (e.g. ACL) a known external load (F_1) is applied to an intact knee and the resulting motion is recorded. The ACL is subsequently transected (as represented by the cut spring in (**Figure 7**) and the robot reproduces the previously recorded joint motion and a new set of force data (F_2) is obtained; because the path of motion for both test conditions is identical, the principle of superposition can be applied and the *in situ* force of the ACL is the vector difference in recorded forces, i.e. ($F_1 - F_2$)**.**

The principle of superposition states that "the combined effect of a number of forces acting on a structure is equal to the sum of the effects of each force applied separately" [97]. Three assumptions are required to apply the principle of superposition. First, the bones must be effectively rigid when compared to the soft tissues at the joint. This condition stipulates that the joint is free from any disease such as osteoporosis, which might significantly compromise rigidity. Second, no interactions can exist between the surrounding tissues and the bones. Third, the position of the bones before and after transecting a joint structure must also be repeated exactly (the position of the tibia must be exactly the same with respect to the femur). Soft tissue stress strain curves are nonlinear as a result of their viscoelastic behavior. Even though the stress-strain curve of soft tissues is nonlinear, the principle of superposition still holds when there is no interaction between the surrounding tissues and the bones as stated above. Nonlinearity is not a condition for exemption although it could indicate interaction.

As stated before, the UFS is capable of measuring three forces and three moments along and about a Cartesian coordinate system fixed with respect to the sensor.

$$^{s}F = (f_x, f_y, f_z, m_x, m_y, m_z)^T \tag{15}$$

The magnitude of the external forces can be determined by:

$$Magnitude = \sqrt{f_x^2 + f_y^2 + f_z^2} \tag{16}$$

The direction of the external forces can also be determined as follows:

$$a_x = \frac{f_x}{\sqrt{f_x^2 + f_y^2 + f_z^2}}, a_y = \frac{f_y}{\sqrt{f_x^2 + f_y^2 + f_z^2}}, a_z = \frac{f_z}{\sqrt{f_x^2 + f_y^2 + f_z^2}} \tag{17}$$

where a_x, a_y and a_z represent components of the direction vector, a, of the external force.

2.6 Previous Studies at MSRC/Preliminary Studies

Several preliminary studies were performed in the process of developing our current methodology to estimate ligament force. These studies laid the groundwork for the methodology presented in this paper.

2.6.1 Computational Models

A 3-D computational model has been used to investigate the effect of variation in geometric, mechanical, and structural properties between cadaveric knees on the determination of force in the human ACL during the reproduction of knee kinematics. Ten human cadaveric knee specimens were tested using the robotic/UFS testing system to determine the knee kinematics and *in situ* force in the ACL under an anterior tibial load.

The kinematics of each knee was used as input into a validated 3-D finite element model of the ACL, and forces in the ACL for each knee were calculated. Experimental and calculated ACL forces were then compared.

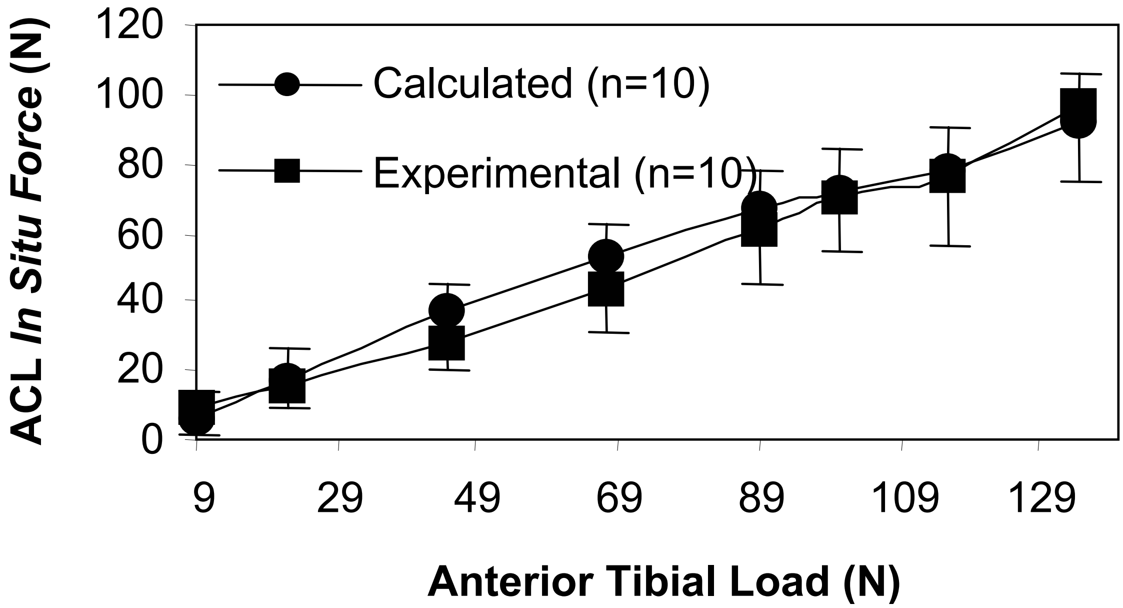

Figure 8. Experimental and calculated force in the ACL obtained by the reproducing knee kinematics obtained in response to anterior tibial loads of 134 N at full extension

Calculated forces in the ACL varied from the experimental *in situ* force of the ACL by 2.6±5.2 N (**Figure 8**). In response to the kinematics obtained under a 134 N anterior tibial load, the maximum force difference was 3.5%. This data showed that differences in kinematics due to variation in geometric, mechanical and structural properties between knees does not have a significant effect on the estimation of force in the human ACL in response to anterior tibial loads at full extension[108].

This model did not include joint contact. Preliminary studies at the musculoskeletal research center have shown that variation in geometric, mechanical and structural properties does effect the joint compressive force when knee kinematics are reproduced.

2.6.2 Reproducing Kinematics

Preliminary studies have shown that the biological variability between knees creates high joint contact forces in the knee making it impossible to simply reproduce absolute motion from one knee onto another. In order to reproduce knee kinematics between different knees, we needed to choose a common reference position to apply kinematics. The passive path of flexion extension can be easily recorded from patients and is a safe reference position from which to apply load. Kinematics were then calculated from the one knee (source) as the difference between 6-DOF kinematics from the applied loads and the passive path of flexion-extension[109]. Kinematics from one cadaveric porcine knee was then added to the specimen specific reference position of another cadaveric porcine knee. This method proved unsuccessful due to variation in coupled motions between knees. These differences in coupled motion damaged the cadaveric porcine knee on which the kinematics were reproduced.

2.6.3 Reproducing Average Kinematics

In a preliminary study, average kinematics from three cadaveric porcine knees (source) were reproduced onto another cadaveric porcine knee (target) at 30° of knee flexion in response to a 100 N anterior load **(Figure 9).** The *in situ* forces in the ACL were not statistically different ($p<0.05$), which indicates that it is possible to reproduce kinematics of one set of knees to obtain *in situ* forces in another.

Since this study was limited to one flexion angle, one specimen, and one loading condition, the current study will more rigorously investigate this methodology using a larger number of knees, more complex loading conditions and multiple flexion angles.

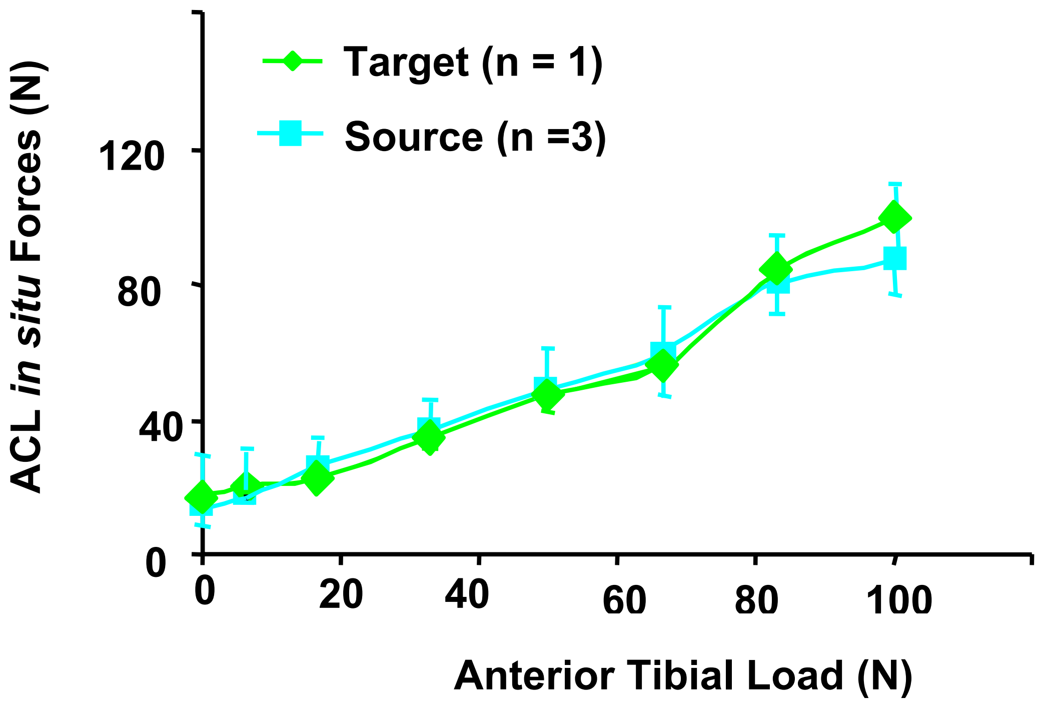

Figure 9. ACL *in situ* force vs. anterior tibial load (mean±SD) of 100 N at 30° of flexion

3.0 OBJECTIVES

In this section, the overall goal and clinical relevance of this study will be presented. The specific aims for reproducing *in vivo* kinematics will also be explained. In section 4.0 the development of a methodology for estimating the force in the ACL in a non-contact and non-invasive manner by reproducing average kinematics will be presented.

3.1 Broad Goal

It continues to be the belief of our research center that, for a successful short and long-term outcome, the ACL replacement graft must restore, as closely as possible, the intact knee kinematics, and reproduce the force of the intact ACL in response to in vivo activities. Thus, the broad goal of this work is to improve ACL reconstruction procedures by restoring the *in situ* force of the ACL graft to that of the intact ACL during in vivo activities this will allow rehabilitation protocols to be designed to replicate the knee kinematics and in situ forces of the intact ACL *in vivo*.

3.2 Specific Aim

The specific aim of this study was to evaluate the feasibility of a non-invasive, non-contact methodology for estimating force in the ACL by reproducing average kinematics in 6-DOF degrees of freedom from one set of porcine knees (source) onto a separate set of porcine knees (target). To do this we will:

a) Record Input kinematics from a set of source knees
b) Compute the average kinematics
c) Reproduce these kinematics on a set of target knees
d) The *in situ* force in the ACL between the applied loads in the source knees and reproduced kinematics in the target knees will then be compared

4.0 METHODS

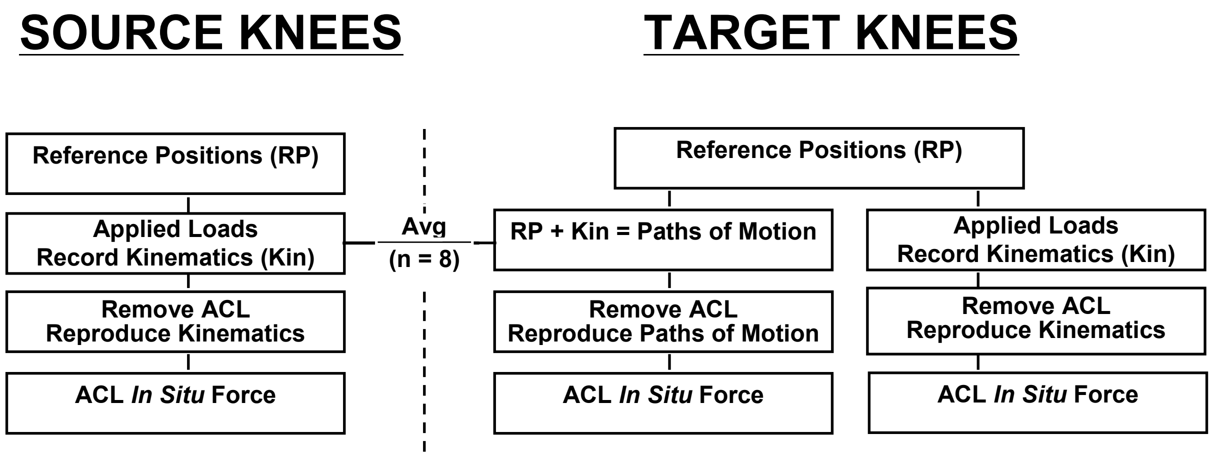

Figure 10. Flow chart representing the experimental protocol for reproducing average kinematics from one set of porcine knees (source) onto a separate set of porcine knees (target)

4.1 Overview of Methods

A robotic/universal force-moment sensor (UFS) testing system was utilized to determine the passive path of flexion-extension of the source knees (n=8) (**Figure 10**). These joint positions served as a reference positions for the application of external loading conditions to the joint and resulted from minimizing the forces and moments during flexion-extension of the knee. Source kinematics were then collected in response to an anterior load of 100 N and a valgus load of 5 Nm at 30°, 60°, and 90° of knee flexion.

After the kinematics were collected for eight source knees, the average kinematics were calculated (**Table 1**). Next, the ACL was removed and the kinematics from the applied loads were repeated to determine the *in situ* force in the ACL for the source knees.

Table 1. Calculation of kinematics in 6-DOF, medial-lateral (ML), anterior-posterior (AP), and proximal-distal (PD) translations are in millimeters, flexion-extension (FE), varus-valgus (VV) and internal-external (IE) rotations are in degrees

	ML	AP	PD	FE	VV	IE
Passive Path (Source)	**-0.5**	**1.2**	**-0.3**	**30.0**	**0.5**	**0.1**
Source Kinematics – A or B	**2.4**	**7.4**	**-0.4**	**30.0**	**-0.3**	**10.1**
Average Kinematics	**2.9**	**6.2**	**-0.1**	**0.0**	**-0.8**	**10.0**
		mm			**degrees**	

The passive path of flexion-extension (reference positions) was subsequently determined for the target knees. Target kinematics were then collected in response to the same loading conditions. The reference positions of the target knee and the average kinematics from the source knees for each loading condition was defined as the paths of motion to be reproduced on the target knees. After removing the ACL, the calculated paths of motion and target kinematics were repeated providing the in situ force in the ACL from the average kinematics of the source knees and the applied loads.

4.2 Collection of Source Kinematics

The robotic/UFS testing system is designed to test human knees, thus, several changes to the robot control algorithms were required in order to adapt it to testing porcine knees. Porcine knees are very lax in internal external rotation. On our first preliminary test, in which IE loading of 10 Nm was attempted, the robot control algorithm diverged at 90° of knee flexions breaking the knee; as a result, the IE loading condition was discarded. On the second preliminary test, the robot control algorithm again diverged at 90° of knee flexion during passive path. To remedy this problem, the IE moment targets for minimizing IE rotations during passive path was decreased by an order of magnitude. This change to the control algorithm minimized the range of IE rotation over which the control algorithm searches for equilibrium. These changes demonstrated very repeatable passive paths and prevent further specimen damage. IE laxity continued to pose a problem during AP loading; IE rotations differed from extreme internal (45°) to extreme external (30°). To prevent this, IE rotations were limited to ±10° of the reference position at each flexion angle were loads were applied. Thus, altered passive path (Flex17p.v2) and loading (Load19p.v2) control algorithm were created. Source kinematics were collected with the successful completion of these control algorithms.

Six-DOF source kinematics were collected from eight cadaveric porcine knees using the robotic/UFS testing system. The porcine model was selected because of the similarity between the human and porcine ACL force magnitude and direction in response to an anterior load[110]. Source kinematics were then collected in response to an anterior load of 100 N and a valgus load of 5 Nm at 30°, 60°, and 90° of knee flexion. (**Table 2**). Only the knee flexion angle was constrained during anterior loading; hence, this loading condition was a 5-DOF test. Knee flexion angle and internal-external tibial rotation were constrained during valgus loading; thus, this loading condition was a 4-DOF test. The robotic/UFS testing system applies loads in incremental steps, where each incremental step is a percentage of the target load. There are nine loading steps, which ramp up to meet the force targets in the anterior and valgus direction (100 N, 5 Nm).

Therefore, the load and displacement is recorded for nine positions between the reference position and the position of maximum target load. The loading conditions resulted in Kinematics – 1A for anterior loads and Kinematics – 1B for valgus loads. The ACL was then transected and Kinematics 1A & 1B were repeated on the ACL deficient knee. The difference between the knee forces in the intact and the ACL deficient knee is attributed to the ligament, giving the *in situ* force in the ACL by the principle of superposition.

Table 2. Experimental protocol for collecting source kinematics

Protocol	**Data Obtained**
Intact knee	
Passive path of flexion-extension	Reference positions
Applied loads at 30°, 60°, and 90°	
Anterior load of 100 N	Kinematics - 1A
Valgus load of 5 Nm (4-DOF)	Kinematics - 1B
ACL –deficient knee	
Reproducing kinematics – 1A, 1B	ACL *in situ* force

4.3 Calculation of Average Kinematics – Source Knees

The 6-DOF kinematics for an applied load were calculated by taking the difference between Kinematics 1A & 1B and the reference positions in each of the 6-DOF. The kinematics in 6-DOF were then calculated by taking the average of the kinematics of the eight source knees tested (**Table 1**). One set of average kinematics was calculated for each loading condition (Anterior load of 100 N, valgus load of 5 Nm) at 30°, 60°, and 90° of flexion, giving a total of six paths of motion to be reproduced in the target knee.

4.4 Reproducing 6-DOF Average Kinematics – Target Knees

The low-payload robotic/UFS testing system moves the tibial with respect to its global coordinate system in response to the force feedback during a force control. The robot records the location of the tibia in Cartesian coordinates, x, y, z, translations and nautical angles yaw, pitch, and role. The robot does not record the relationship between the tibia and femur but does not utilize this relationship to carry out joint motion. Thus, the user can request the Cartesian Coordinates of the tibia with respect to the global coordinate system and the joint motion (tibia with respect to femur). Average kinematics are calculated based on the joint motion but the robot cannot utilize the joint motion because the position and orientation of the tibia is unique for each test. Therefore, given the average joint motion and the position and orientation of the femur with respect to robot global, the motion of the tibia with respect to the robot's global coordinate system can than be calculated (T_{G_T}) (**Figure 11**).

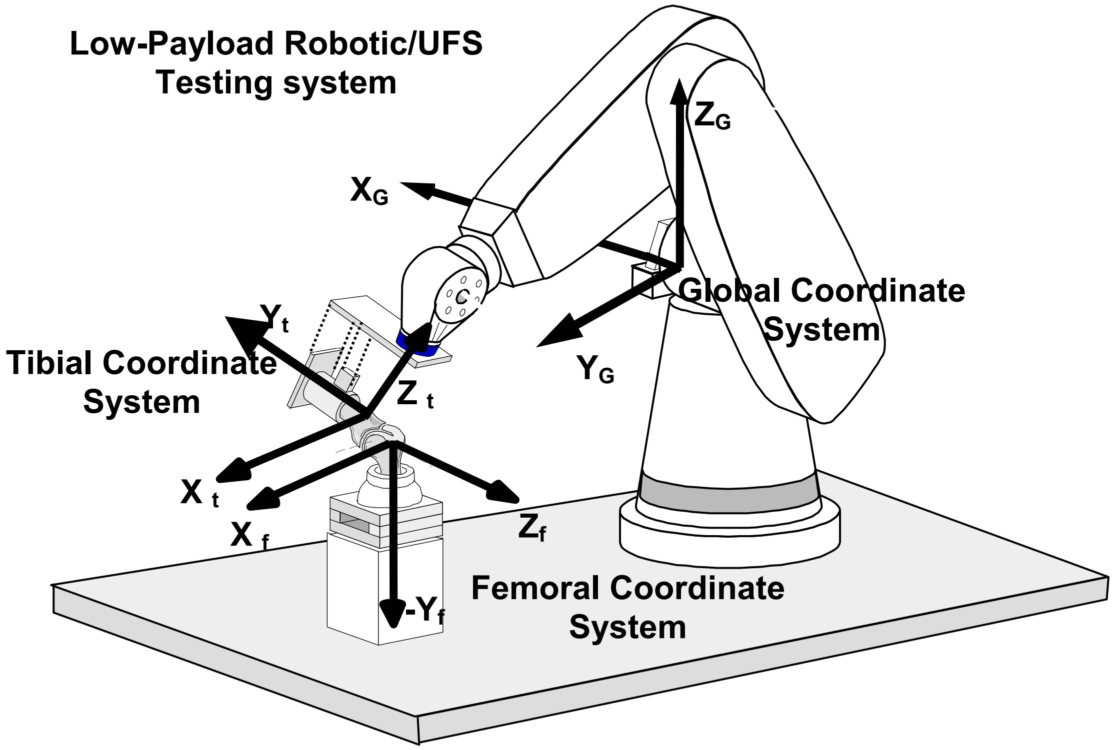

Figure 11. Illustration of the coordinate systems involved in reproducing knee motion

This was accomplished by recording the tibias motion with respect to femur (T_{F_T}) (the knee's path of motion) and the relationship between the robot's global coordinate system and the tibia (T_{G_T}) at the reference position. Both of these relationships are obtained from the data output of the robot control algorithm. Next, the constant relationship between the robot's global coordinate system and the femur (T_{G_F}) can be calculated from tibia with respect to global times the inverse of tibia with respect to femur ($T_{G\_F} = T_{G\_T} * T_{F\_T}^{-1}$) (**Figure 12**).

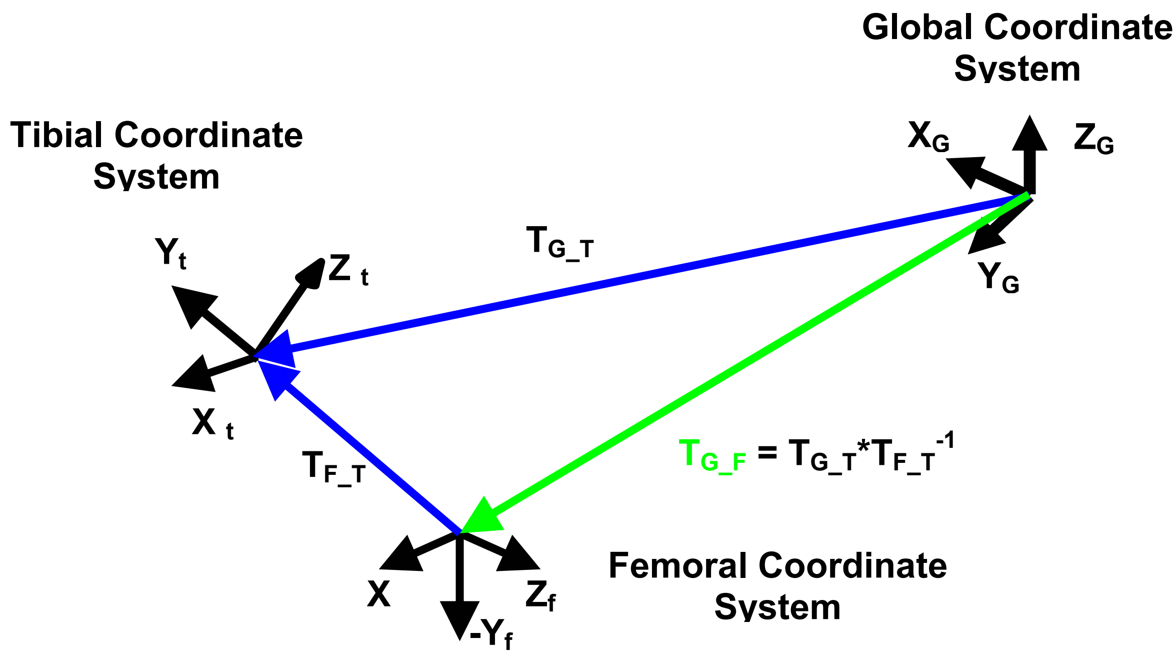

Figure 12. Diagram of the method used to calculate the constant relationship of the femoral coordinate system with respect to the robot's global coordinate system

Once the constant relationship (T_{G_F}) is known, the relationship of the tibia with respect to the robot's global coordinate system (T_{G_T}) can be calculated ($T_{G\_F} * T_{F\_T} = T_{G\_T}$) for each loading condition and flexion angle.

A reference position for each target knees was first defined at 30°, 60°, and 90° of flexion (**Table 3**). The average kinematics from the source knees were added to the reference positions to determine the paths of motion to be reproduced on the target knees for each of the applied loads. Although the same average kinematics were reproduced on each target knee, the reference positions were unique for each target knee.

Once the reference positions were determined for the target knees, the same loading conditions applied to the source knees were applied to the target knees – 100 N anterior load (Kinematics 2A) and 5 Nm valgus load (Kinematics 2B) at 30°, 60°, and 90° of knee flexion. Next, the paths of motion to be reproduced on the target knees were calculated using the reference positions of the target knees and average kinematics from the source knees.

Reproducing these paths of motion in the intact target knee provided the knee force due to reproducing average kinematics. After removing the ACL, the kinematics from the applied loads (Kinematics 2A and 2B) and the calculated paths of motion were reproduced in the ligament deficient target knee. The difference between the knee force in the intact knee and the knee force in the ligament deficient knee are the *in situ* forces in the ACL from the applied loads and average kinematics.

Table 3. Experimental protocol for reproducing average kinematics in the target knees

Protocol	Data Obtained
Intact knee	
Passive path of flexion-extension	Reference Positions
Applied loads at 30°, 60°, and 90°	
Anterior load of 100 N	Kinematics 2A
Valgus load of 5 Nm (4-DOF)	Kinematics 2B
Reproducing average kinematics	Knee force
ACL – deficient knee	
Reproducing kinematics – 2A, 2B	ACL *in situ* force
Reproducing average kinematics	ACL *in situ* force

4.5 Summary of Data Obtained

The kinematics for each of the loading conditions in the target (n = 8) and source (n = 8) knees were obtained from the robotic /UFS testing system in experimental protocols (**Table 2 & Table 3)**. The *in situ* force in the ACL, in response to each of the applied loads, was recorded for both the target and source specimens. In the target specimens, the *in situ* force in the ACL was also recorded from reproducing the average kinematics from the source knees. Tables of the components of the force vector in the ACL and 6-DOF kinematics for anterior tibial loads and valgus loads in the source and target knees are also listed at the end of the results section.

4.6 Statistics

Results of a power analysis (2-way ANOVA) showed that a sample size of 3 to 5 porcine knees was sufficient to prove a significant difference in force with 80% power for all loading conditions. The dependent variable was flexion angle and the independent variable was the *in situ* force in the ACL due to reproducing average kinematics and applied loads. Statistical differences between the anterior tibial translation and valgus rotations between the target and source knees were evaluated at each flexion angle using a two-sample t-test. A two-sample t-test was also used to determine statistical differences between the *in situ* force in the ACL due to applied loads in the source and the target knee. A paired t-test was performed at 30°, 60°, and 90° of flexion to determine if there was a difference in the *in situ* force in the ACL resulting from the applied loads and reproducing average kinematics in the target knee. Finally, a two-sample t-test assuming unequal variance was used to determine if there were significant differences between *in situ* forces in the ACL due to applied loads in the source knees and reproducing average kinematics in the target knees.

5.0 RESULTS

The anterior tibial translations (ATT) for the source knees in response to an applied anterior load of 100 N were 4.5±1.5 mm, 5.8±1.3 mm, and 5.1±1.3 mm at 30°, 60°, and 90° of flexion, respectively (**Figure 13**).

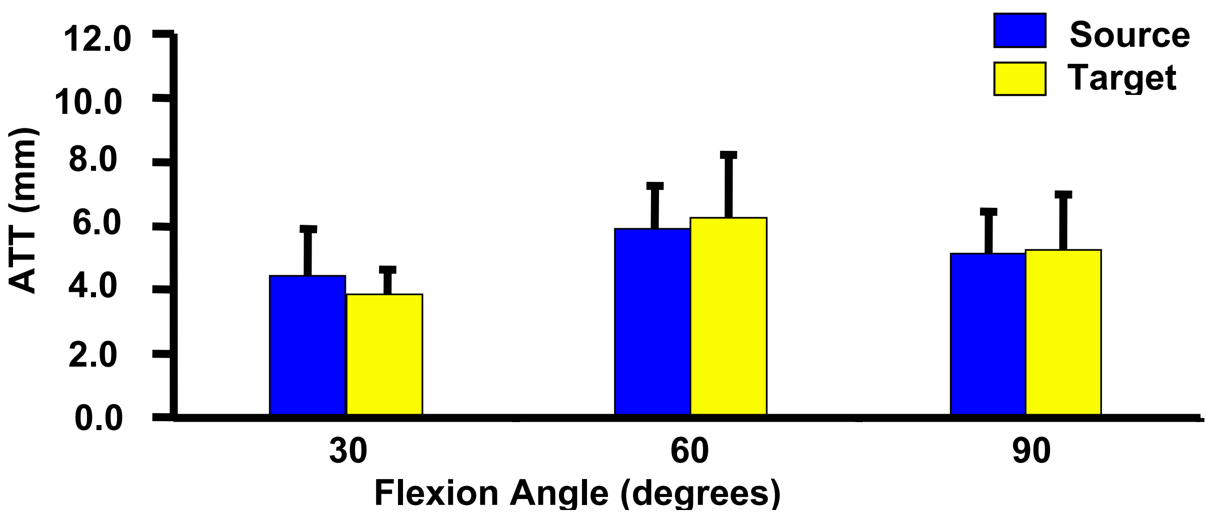

Figure 13**.** Anterior tibial translation-ATT (mean±SD) in response to an anterior load of 100 N, n = 8 – (*, $p<0.05$)

For the target knees, ATT at the same flexion angles were 3.9±0.7 mm, 6.2±2.0 mm, and 5.3±1.7 mm, respectively, for an applied anterior load of 100 N. The ATT of the target and source knees from applied loads were not statistically different at 30° (p=0.5178), 60° (p=0.6881), or 90° (p=0.8736) of flexion**.** The valgus rotations for the source knees in response to an applied valgus load of 5 Nm were 3.2±0.7°, 4.6±1.2°, and 5.4±0.8°, at 30°, 60°, and 90° of flexion, respectively (**Figure 14**).

For the target knees, valgus rotations at these same flexion angles were 2.6±0.5°, 4.0±0.8°, and 5.2±0.7°, respectively, for an applied valgus load of 5 Nm. The valgus rotations of the target and source knees from applied loads were not statistically different at 30° (*p*= 0.1094), 60° (*p*= 0.2531), or 90° (*p*= 0.7299) of flexion.

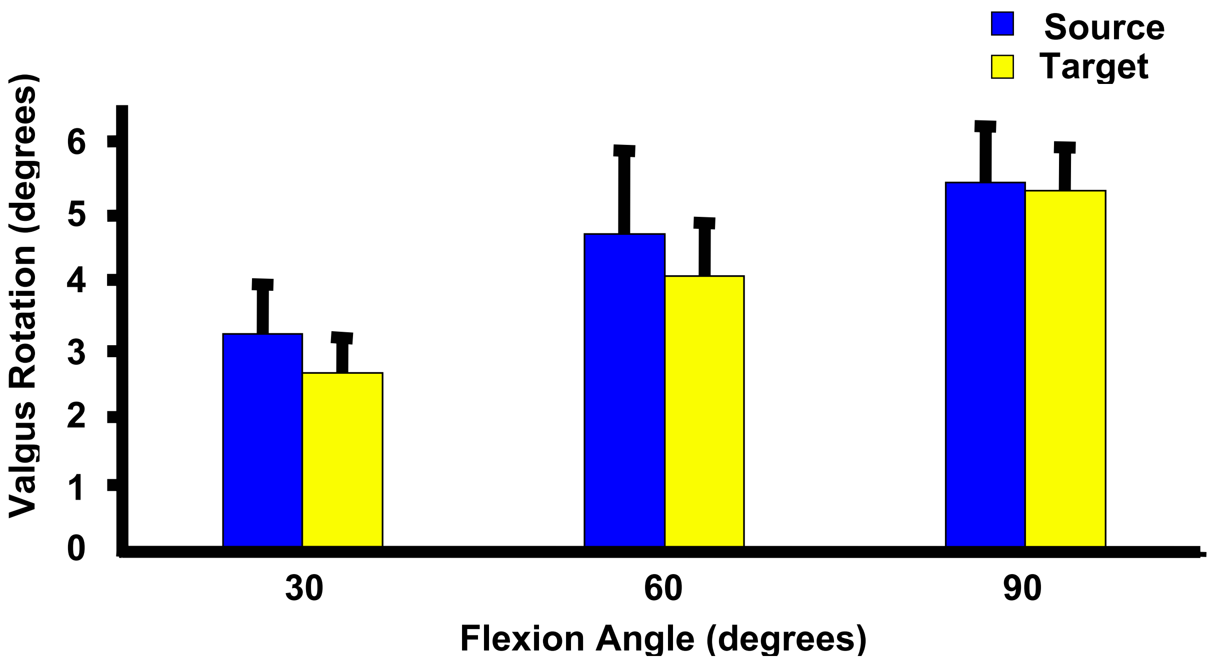

Figure 14. Valgus rotation (mean±SD) in response to a valgus load of 5 Nm, n = 8 – (*, p<0.05)

The *in situ* forces in the ACL for the source knees in response to an anterior tibial load of 100 N were 79.3±11.1 N, 89.3±6.6 N, and 80.7±4.0 N, at 30°, 60°, and 90° of flexion, respectively. For the target knees, *in situ* forces in the ACL at these same flexion angles were 73.7±9.3 N, 83.9±4.0 N, and 76.3±4.4 N, respectively. There was not a significant difference in the *in situ* force in the ACL between the applied anterior loads in the source and target knees at 30°, 60°, and 90° of flexion. The *in situ* forces in the ACL for the source knees in response to a valgus load of 5 Nm were 21.3±8.0 Nm, 18.7±6.4 Nm, and 20.1±9.4 Nm, at 30°, 60°, and 90° of flexion, respectively.

For the target knees, *in situ* forces at these same flexion angles were 24.7±10.1 Nm, 21.8±9.5 Nm, and 16.7±8.4 Nm, respectively. There was not a significant difference in the *in situ* force in the ACL between the valgus loads in the source and target knees at 30°, 60°, and 90° of flexion. The *in situ* forces in the ACL of the target knees from reproducing average kinematics from an anterior load of 100 N were 174.1±27.6 N, 138.1±109.9 N, and 127.4±102.1 N at 30°, 60°, and 90° of flexion, respectively **(Figure 15)**.

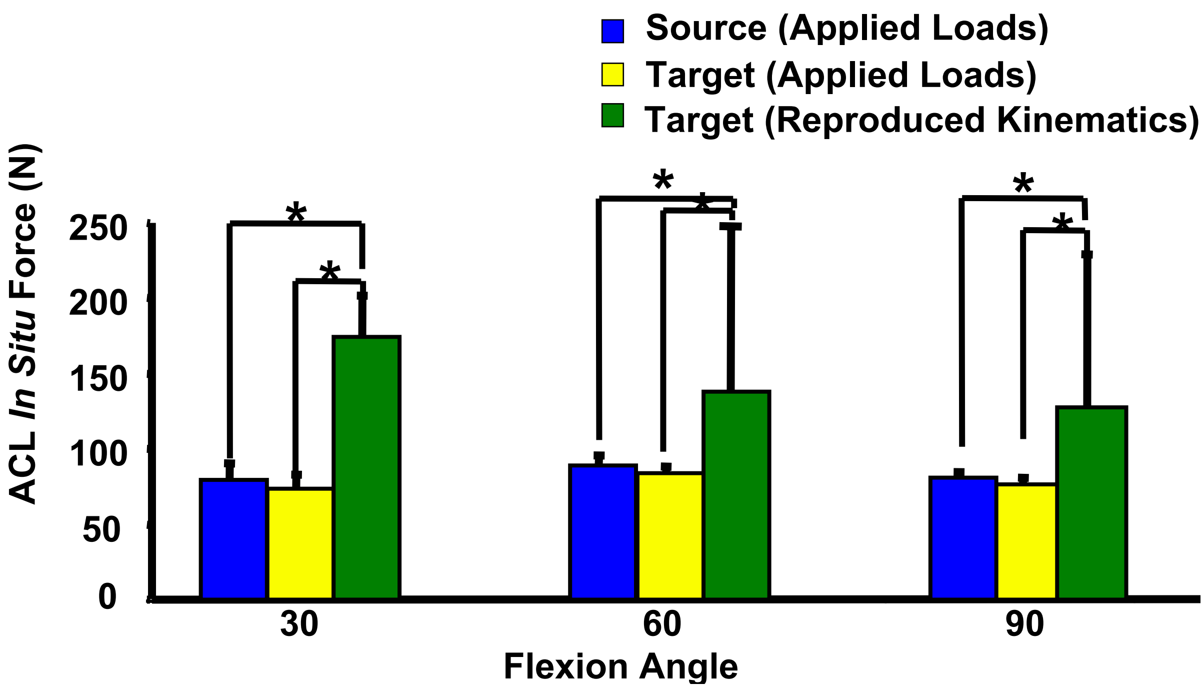

Figure 15. *In situ* force (mean±SD) in the ACL in response to an anterior load of 100 N and average kinematics. (*, p<0.05)

These *in situ* forces in the ACL in response to reproducing average kinematics from an anterior load were significantly different from those of the anterior tibial loads in the source and target knees at 30°, 60°, and 90° of flexion. The *in situ* forces in the ACL due to reproducing average kinematics from a valgus load of 5 Nm were 24.8±21.7 N, 36.9±47.2 N, and 23.9±30.8 N at 30°, 60°, and 90° of flexion, respectively **(Figure 16)**.

These *in situ* forces in the ACL due to reproducing average kinematics from a valgus load of 5 Nm were not statistically different from those of the valgus loads in the source and target knees at 30° (p=0.4914) and 90° (p=0.6632) although force differences were significantly different at 60° of flexion.

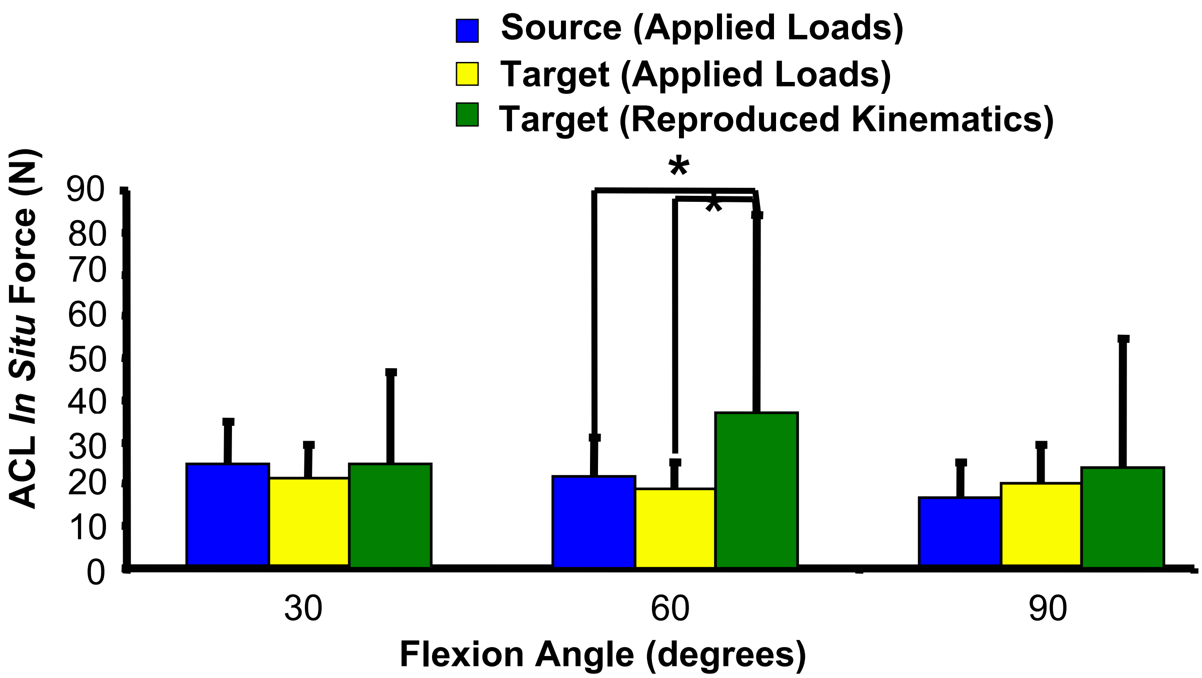

Figure 16. *In situ* force (mean±SD) in the ACL in response to a valgus load of 5 Nm and average kinematics. (*, p<0.05)

Tables of the components of the force vector in the ACL and 6-DOF kinematics of the target knees due to anterior tibial and valgus loads at 30°, 60°, and 90° of flexion are listed on the left hand side of **Table 4-9.** Reproduced average kinematics in response to these loads at 30°, 60°, and 90° of flexion are listed in on the right had side of these tables**.** The applied load in the target knees were identical to those applied in the source knees. Thus, these tables will allow the readers to evaluate the effect of the 6-DOF kinematics from anterior tibial load and the average kinematics on the force components in the ACL.

For example in response to an anterior load at 30° of flexion, the ACL force in the anterior direction of knees 6 and 8 were notably different, although the ATT of both knees were identical. The coupled internal-external rotation of these knees differed by 8° (**Table 4)**. Knee 3 was very lax at 60° and 90° of flexion the reproduced kinematics did not reach the linear region of the load displacement curve thus the ACL force component in the anterior direction were very low **(Tables 5 & 6**). The ACL force in the anterior direction of knee 6 in response to reproducing average kinematics at 90° of flexion was not significantly different from the applied anterior tibial load **(Table 6)**. The resultant force in the ACL in response to reproduced average kinematics and the applied valgus load were identical at 30° of flexion for knee 3. In response to a valgus load at 60° and 90°of flexion, the *in situ* force in the ACL of knee 1 and knee 5 were notably different, although the valgus rotation only varied by 0.1° for both knees **(Tables 8 & 9)**. The coupled anterior-posterior and proximal-distal translations differed most in these knees.

Table 4. Components of the force vector in the ACL and 6-DOF kinematics in response to a anterior tibial loading of 100 N at 30°of flexion, [medial-lateral (ML), anterior-posterior (AP), proximal-distal (PD) translation are in mm, flexion-extension (FE), varus-valgus (VV) and internal-external (IE) rotation are in degrees]

	Applied Anterior Load (Target)						Reproduced Average Kinematics (Target)					
Knee #1	ML	AP	PD	FE	VV	IE	ML	AP	PD	FE	VV	IE
ACL Force (N)	-16.6	56.6	5.0				7.6	154.8	-32.8			
Kinematics	1.5	4.9	-0.6	0.0	0.1	4.9	1.5	4.4	-0.2	0.0	-0.6	1.8
Knee #2												
ACL Force (N)	-15.3	76.3	14.1				-16.5	165.8	-12.8			
Kinematics	2.0	3.8	0.4	0.0	-0.6	1.9	1.5	4.4	-0.5	0.0	-0.7	2.0
Knee #3												
ACL Force (N)	-15.8	59.9	13.9				5.0	215.6	-7.8			
Kinematics	0.9	4.0	0.8	0.0	-1.0	10.0	1.4	4.5	-0.3	0.0	-0.7	1.9
Knee #4												
ACL Force (N)	-21.9	62.1	3.1				-39.8	186.1	18.3			
Kinematics	0.6	3.6	-0.2	0.0	-0.2	5.1	1.3	4.4	-0.3	0.0	-0.7	2.0
Knee #5												
ACL Force (N)	-16.8	75.5	3.7				-20.4	154.1	11.6			
Kinematics	1.7	3.4	-0.3	0.0	-1.0	3.4	1.4	4.3	-0.3	0.0	-0.7	2.0
Knee #6												
ACL Force (N)	-5.5	80.0	9.4				-21.0	181.7	0.9			
Kinematics	2.9	4.4	0.2	0.0	0.6	10.0	1.8	4.4	-0.3	0.0	-0.6	2.0
Knee #7												
ACL Force (N)	-9.7	80.4	19.3				-20.1	188.6	44.4			
Kinematics	0.8	2.5	-0.3	0.0	-0.3	-1.2	1.4	4.3	-0.2	0.0	-0.7	2.0
Knee #8												
ACL Force (N)	-9.2	79.3	12.6				-15.9	125.5	12.6			
Kinematics	2.2	4.4	0.5	0.0	-0.4	10.0	1.7	4.4	-0.4	0.0	-0.7	2.0

Table 5. Components of the force vector in the ACL and 6-DOF kinematics in response to a anterior tibial loading of 100 N at 60° of flexion, [medial-lateral (ML), anterior-posterior (AP), proximal-distal (PD) translation are in mm, flexion-extension (FE), varus-valgus (VV) and internal-external (IE) rotation are in degrees]

	Applied Anterior Load (Target)						Reproduced Kinematics (Target)					
Knee #1	ML	AP	PD	FE	VV	IE	ML	AP	PD	FE	VV	IE
ACL Force (N)	-9.2	89.1	6.1				-14.1	26.2	-12.0			
Kinematics	2.5	8.8	-1.8	0.0	2.1	10.0	1.9	5.8	-1.0	0.0	-0.3	2.1
Knee #2												
ACL Force (N)	1.7	83.9	16.2				-2.8	62.7	-2.4			
Kinematics	2.8	6.9	-0.9	0.0	0.1	6.1	2.1	5.9	-1.0	0.0	-0.3	2.2
Knee #3												
ACL Force (N)	6.4	83.2	6.8				2.1	1.2	-2.9			
Kinematics	3.4	8.5	-0.6	0.0	-0.4	10.0	1.9	5.8	-1.0	0.0	-0.4	2.2
Knee #4												
ACL Force (N)	-4.3	75.5	3.7				-43.0	295.4	23.8			
Kinematics	2.0	5.0	-0.6	-0.1	1.5	9.4	2.0	5.9	-1.1	0.0	-0.3	2.2
Knee #5												
ACL Force (N)	-8.1	83.2	2.3				-14.4	147.5	2.5			
Kinematics	2.2	4.9	-0.4	0.0	-2.3	-0.9	2.0	5.9	-1.0	0.0	-0.3	2.2
Knee #6												
ACL Force (N)	0.3	84.2	-1.9				-13.6	109.4	-14.7			
Kinematics	5.4	7.2	-0.3	0.0	0.2	10.0	1.9	5.8	-1.2	0.0	-0.3	2.2
Knee #7												
ACL Force (N)	-5.6	81.4	11.7				-10.4	281.4	56.8			
Kinematics	0.8	2.9	-0.8	0.0	0.1	0.3	2.1	5.9	-1.1	0.0	-0.3	2.2
Knee #8												
ACL Force (N)	-5.1	86.6	2.0				-17.9	159.1	-5.0			
Kinematics	4.0	5.7	-0.1	0.0	1.0	10.0	2.2	5.9	-1.1	0.0	-0.3	2.2

Table 6. Components of the force vector in the ACL and 6-DOF kinematics in response to a anterior tibial loading of 100 N at 90° of flexion, [medial-lateral (ML), anterior-posterior (AP), proximal-distal (PD) translation are in mm, flexion-extension (FE), varus-valgus (VV) and internal-external (IE) rotation are in degrees]

	Applied Anterior Load (Target)						Reproduced Kinematics (Target)					
Knee #1	ML	AP	PD	FE	VV	IE	ML	AP	PD	FE	VV	IE
ACL Force (N)	-0.5	70.2	2.6				-1.1	17.5	11.5			
Kinematics	4.8	7.8	-2.0	0.0	2.9	10.0	3.2	5.0	-0.9	0.0	0.7	3.5
Knee #2												
ACL Force (N)	1.3	83.9	11.1				1.1	50.1	8.1			
Kinematics	4.5	5.8	-0.5	0.0	0.1	3.6	3.6	5.2	-1.0	-0.1	0.8	3.8
Knee #3												
ACL Force (N)	11.0	74.9	5.0				1.7	9.4	-8.9			
Kinematics	5.3	6.6	-0.4	0.0	0.8	10.0	3.2	5.0	-0.8	-0.1	0.7	3.7
Knee #4												
ACL Force (N)	-2.6	72.4	-1.6				-21.2	261.9	2.0			
Kinematics	2.7	3.6	0.1	-0.1	1.2	5.5	3.3	5.1	-0.8	-0.1	0.8	3.7
Knee #5												
ACL Force (N)	-4.6	75.2	2.2				-9.6	207.8	-5.3			
Kinematics	4.2	3.6	0.5	0.0	-1.1	1.2	3.3	5.0	-0.9	-0.1	0.8	3.7
Knee #6												
ACL Force (N)	6.5	75.2	3.0				-2.5	94.0	-7.6			
Kinematics	6.1	6.1	0.1	0.0	2.1	10.0	3.4	5.1	-0.8	0.0	0.8	3.6
Knee #7												
ACL Force (N)	-7.0	76.1	10.6				10.2	254.9	23.8			
Kinematics	1.2	2.8	-0.8	0.0	0.0	0.0	3.2	5.2	-0.8	-0.1	0.8	3.7
Knee #8												
ACL Force (N)	-3.8	79.2	-3.3				-17.4	111.9	-3.0			
Kinematics	5.2	5.7	-0.4	-0.1	2.7	10.0	3.6	5.0	-0.9	-0.1	0.8	3.7

Table 7. Components of the force vector in the ACL and 6-DOF kinematics in response to a valgus load of 5 N m at 30° of flexion, [medial-lateral (ML), anterior-posterior (AP), proximal-distal (PD) translation are in mm, flexion-extension (FE), varus-valgus (VV) and internal-external (IE) rotation are in degrees]

	Applied Anterior Load (Target)						Reproduced Kinematics (Target)					
Knee #1	ML	AP	PD	FE	VV	IE	ML	AP	PD	FE	VV	IE
ACL Force (N)	-4.3	6.6	5.0				-9.1	16.0	1.0			
Kinematics	0.9	0.1	0.6	0.0	-2.3	1.0	1.2	1.2	0.6	0.0	-3.1	-0.2
Knee #2												
ACL Force (N)	-4.2	14.8	10.1				0.7	1.8	7.4			
Kinematics	1.9	1.2	0.6	0.0	-2.6	-0.7	1.2	1.1	0.4	0.0	-3.2	0.0
Knee #3												
ACL Force (N)	-0.5	15.0	-2.3				-7.3	12.6	4.3			
Kinematics	2.3	1.9	0.6	0.0	-2.7	1.1	0.9	1.0	0.5	0.0	-3.2	0.0
Knee #4												
ACL Force (N)	-12.8	28.6	4.7				-16.1	35.9	-3.5			
Kinematics	0.8	0.6	0.7	0.0	-2.2	-0.6	1.2	1.2	0.6	0.0	-3.1	0.0
Knee #5												
ACL Force (N)	-7.6	29.2	3.8				-4.2	9.5	2.6			
Kinematics	2.0	1.5	0.2	0.0	-3.4	-0.8	1.2	1.0	0.6	0.0	-3.1	0.0
Knee #6												
ACL Force (N)	-8.5	14.8	0.8				-13.5	1.4	0.5			
Kinematics	2.7	1.8	0.5	0.0	-3.3	0.8	1.4	1.1	0.5	0.0	-3.1	0.0
Knee #7												
ACL Force (N)	-6.4	16.5	10.2				-13.1	67.4	13.9			
Kinematics	0.9	-0.3	0.5	0.0	-2.0	-1.1	1.4	1.2	0.6	0.0	-3.2	0.0
Knee #8												
ACL Force (N)	-11.5	25.0	-6.9				-15.2	29.7	1.2			
Kinematics	1.1	1.2	0.4	0.0	-2.5	0.4	1.3	1.2	0.5	0.0	-3.2	0.0

Table 8. Components of the force vector in the ACL and 6-DOF kinematics in response to a valgus load of 5 N m at 60° of flexion, [medial-lateral (ML), anterior-posterior (AP), proximal-distal (PD) translation are in mm, flexion-extension (FE), varus-valgus (VV) and internal-external (IE) rotation are in degrees]

	Applied Anterior Load (Target)						Reproduced Kinematics (Target)					
Knee #1	ML	AP	PD	FE	VV	IE	ML	AP	PD	FE	VV	IE
ACL Force (N)	-0.5	14.7	8.4				-2.0	3.7	2.5			
Kinematics	1.2	1.0	1.3	0.0	-3.6	1.0	1.4	2.1	0.6	0.0	-4.6	-0.1
Knee #2												
ACL Force (N)	0.1	14.3	0.1				0.7	1.8	7.4			
Kinematics	1.6	0.8	1.0	0.0	-3.8	-0.5	1.5	2.1	0.5	0.0	-4.6	0.1
Knee #3												
ACL Force (N)	1.3	4.5	-4.6				2.7	2.6	-1.3			
Kinematics	2.0	3.2	0.7	0.0	-4.2	0.7	1.5	2.0	0.6	0.0	-4.7	0.1
Knee #4												
ACL Force (N)	-10.1	26.0	-5.1				-28.1	97.2	11.6			
Kinematics	0.9	0.5	1.1	-0.1	-3.3	-0.2	1.5	2.1	0.7	0.0	-4.6	0.1
Knee #5												
ACL Force (N)	-0.5	19.2	5.8				-0.1	1.6	1.4			
Kinematics	1.5	2.9	0.5	0.0	-4.7	-0.7	1.4	2.0	0.6	0.0	-4.6	0.1
Knee #6												
ACL Force (N)	-7.3	18.9	-3.2				-1.4	-6.0	-5.7			
Kinematics	2.2	3.5	0.9	0.0	-5.5	0.8	1.3	2.2	0.7	0.0	-4.6	0.0
Knee #7												
ACL Force (N)	-6.7	19.2	2.4				-20.6	110.9	26.7			
Kinematics	1.1	0.5	0.5	0.0	-3.1	-0.9	1.4	2.2	0.6	0.0	-4.6	0.1
Knee #8												
ACL Force (N)	-10.0	19.6	-6.3				-18.6	46.4	-7.8			
Kinematics	1.3	1.3	1.1	0.1	-3.9	0.7	1.6	2.1	0.6	0.0	-4.6	0.1

Table 9. Components of the force vector in the ACL and 6-DOF kinematics in response to a valgus load of 5 N m at 90° of flexion, [medial-lateral (ML), anterior-posterior (AP), proximal-distal (PD) translation are in mm, flexion-extension (FE), varus-valgus (VV) and internal-external (IE) rotation are in degrees]

	Applied Anterior Load (Target)						Reproduced Kinematics (Target)					
Knee #1	ML	AP	PD	FE	VV	IE	ML	AP	PD	FE	VV	IE
ACL Force (N)	-0.9	18.2	1.2				-1.8	1.3	-0.8			
Kinematics	2.1	1.3	2.1	0.0	-5.6	0.4	1.6	1.2	1.1	-0.1	-5.5	-0.2
Knee #2												
ACL Force (N)	-4.5	17.9	1.1				-1.2	1.3	5.0			
Kinematics	3.5	0.4	1.7	0.0	-4.9	-0.8	2.0	1.5	1.1	-0.1	-5.4	0.1
Knee #3												
ACL Force (N)	4.2	4.6	0.3				-1.5	1.3	-0.3			
Kinematics	3.5	1.6	1.5	0.0	-5.7	0.7	1.6	1.1	1.1	-0.1	-5.4	0.0
Knee #4												
ACL Force (N)	-6.8	34.9	-3.2				-34.2	78.7	3.4			
Kinematics	1.9	0.4	1.8	-0.1	-4.6	-1.0	1.8	1.3	1.0	0.0	-5.4	0.0
Knee #5												
ACL Force (N)	-3.0	13.4	2.6				-2.7	6.9	3.1			
Kinematics	2.3	0.7	1.7	0.0	-5.2	-0.8	1.8	1.3	1.0	0.0	-5.4	0.0
Knee #6												
ACL Force (N)	-0.8	11.7	-10.0				0.4	3.8	-2.6			
Kinematics	1.9	1.6	2.0	0.0	-6.4	0.0	1.8	1.1	1.0	0.0	-5.3	-0.1
Knee #7												
ACL Force (N)	-2.3	23.0	1.1				-13.9	50.8	6.5			
Kinematics	1.8	0.7	0.5	0.0	-4.3	-0.1	1.8	1.3	0.9	-0.1	-5.4	0.0
Knee #8												
ACL Force (N)	-7.7	26.5	-12.6				-16.8	24.1	-8.6			
Kinematics	2.4	1.5	1.7	0.0	-5.3	0.4	1.9	1.2	1.1	0.2	-5.4	0.0

6.0 DISCUSSION

This study evaluated the feasibility of estimating force in the ACL by reproducing average kinematics in 6-DOF from one set of porcine knees onto a separate set of porcine knees. Kinematics from a set of source cadaveric porcine knees were recorded in response to an anterior tibial load of 100 N and a valgus load of 5 Nm at 30°, 60°, and 90° of flexion. The same loading conditions applied to the source knees were applied to the target knees. In addition to the applied loads, the average kinematics from the source knees was reproduced on the target knees. To validate this methodology, the *in situ* forces in the ACL of the source knees in response to applied loads were compared to the *in situ* forces in the ACL from reproducing average kinematics in the target knees.

There was not a significant difference in ATT for anterior loads or valgus rotations for valgus loads between the source and target knees for all flexion angles. In addition, neither applied loads in the anterior, nor the valgus directions demonstrated a significant difference in the *in situ* force in the ACL between the source and target knees. Reproducing kinematics was used successfully to estimate the force in the ACL for valgus loads at 30° and 90° of knee flexion. However, a significant difference in the *in situ* force in the ACL between the source and target knees was found for all flexion angles in response to an anterior load and at 60° of knee flexion for valgus loading. These differences can be attributed to the variability of the coupled motions that resulted in response to the applied loads. For example, in response to an anterior tibial load, the *in situ* force in the ACL of two target knees were notably different from that obtained by reproducing the average kinematics from the source knees, although the anterior tibial translations of both knees were identical (**Table 4**). Upon further examination, the coupled internal-external rotations differed by 8°.

Even though forces between the source and target knees were significantly different under anterior loads, the similarities in force between source and target knees under valgus loading provides promising evidence that this methodology can be utilized when the source and target knees are matched according to knee laxity. There were notable force difference when coupled motions differed significantly indicating that all 6-DOF of knee motion are important. When average kinematics are utilized, variations in knee laxity cause coupled motions to be eliminated or reduced, artificially constraining the motion of the knee. An example of this has been illustrated in **Table 10** were in the last row of the table, calculation of the average kinematics eliminated internal-external rotations. Constraining knee motion produced high variability in the *in situ* force in the ACL of most knees [111, 112], this re-emphasizes the importance of the ACL in resisting coupled motions [16, 65, 101, 113]. The large variability in ACL *in situ* force can be illustrated by the load displacement cures of **Figure 17**. For example reproducing the ATT from the average curve (5 mm of ATT) on the knee with the maximum toe region would result in an anterior knee force of 20 N yielding even a smaller force in the ACL. Likewise reproducing the ATT form the average curve onto the curve with the minimum toe region would result in an anterior knee force 2 to 3 times the applied anterior load, which would yield a very large force in the ACL as well. Thus, coupled motions and knee laxity affect the force variability in the ACL.

Table 10. Six DOF kinematics in response to an anterior load for 4 porcine knees. [medial-lateral (ML), anterior-posterior (AP), proximal-distal (PD) translation, flexion-extension (FE), varus-valgus (VV) and internal-external (IE) rotation]

Knee	ML (mm)	AP (mm)	PD (mm)	FE (°)	VV (°)	IE (°)
#1	1.5	4.6	-1.1	0.0	-0.4	-7.4
#2	3.1	5.0	-0.4	0.0	-0.3	6.8
#3	1.4	5.0	-0.4	0.0	-0.6	-3.4
#4	1.7	3.4	-0.3	0.0	-1.0	3.4
Avg	1.9	4.5	-0.6	0.0	-0.6	-0.1

Thus, one of the biggest challenges in this study was caused by variation in knee laxity between specimens. Knee laxity is clinically defined as the length of the toe region of the load-displacement curve. **Figure 17** shows load-displacement curves from this study for an anterior load of 100 N at 90° of flexion, and demonstrates the large variability in anterior knee laxities. The stiffness of the average curve was very similar to the stiffness of the knees with the minimum and maximum toe regions. Therefore, differences in the toe region of the curve and not the stiffness of the knee appear to contribute the most to variability in knee laxity.

Anterior knee laxity was quantified by calculating the length of the toe region of the anterior load-displacement curve. Variability in anterior knee laxity was quantified as the range of anterior laxity divided by the average anterior laxity for the eight source and target knees times 100. The average lengths of the toe region for source and target knees (n = 16) were 2.0 mm, 3.8 mm, and 3.3 mm at 30°, 60°, and 90° of flexion, respectively. The length of the toe region for all source and target knees ranged from 0.6 to 4.3 mm, 1.3 to 7.0 mm, and 1.5 to 6.3 mm at 30°, 60°, and 90° of flexion, respectively. Anterior knee laxity showed variability of 183%, 153%, and 146% at these same flexion angles, respectively.

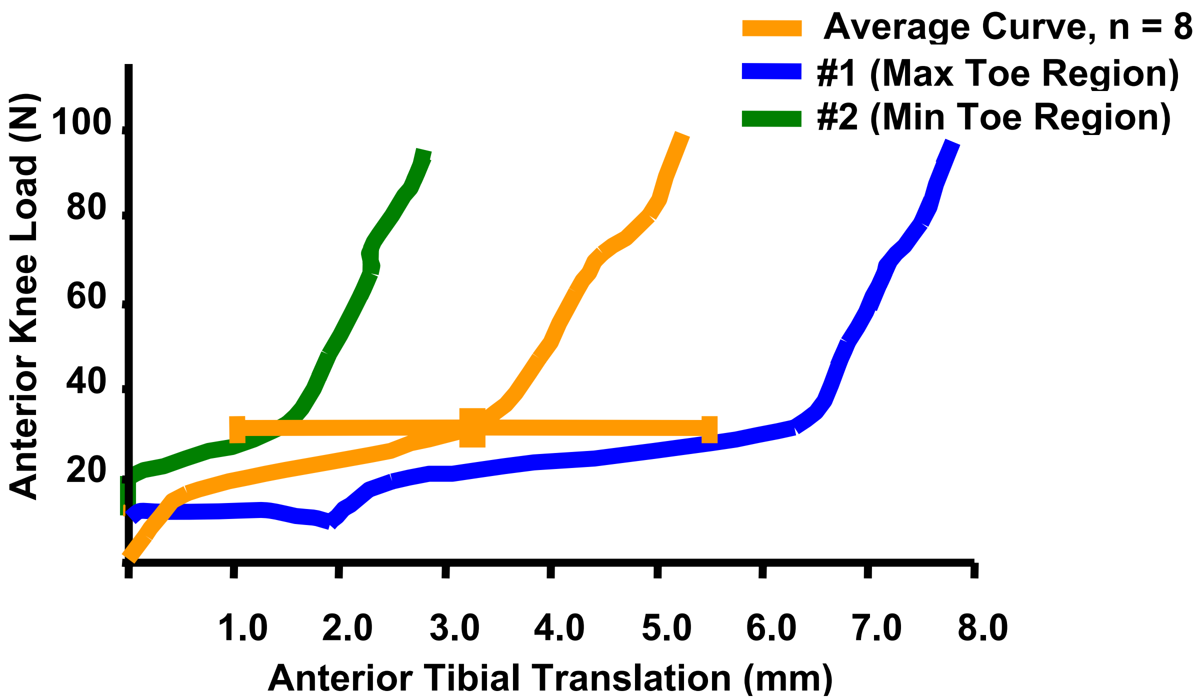

Figure 17. Load-displacement curve in response to an applied load of 100 N at 90°of flexion – The standard deviation of the toe region is represented on the average curve

Multiple measures of knee laxity should to be performed throughout the range of knee flexion-extension because of the nonlinear behavior of the load-displacement curve in biological specimens. For example, the lengths of the toe region in one knee tested were 3.3, 3.0, and 2.5 mm, while another knee had toe region lengths of 3.3, 7.1, and 6.5 mm at 30°, 60°, and 90° of knee flexion. In this study, knee kinematics with similar anterior knee laxity had similar coupled internal-external rotations. Two knees with anterior knee laxity which differed by only 0.0, 0.2, and 1.1 mm at 30°, 60°, and 90° of flexion showed differences in coupled internal-external rotation of 4.6°, 1.2°, and 1.2° at these same flexion angles, respectively. Two knees with anterior laxity which varied by 3.3, 3.6, and 2.7 mm at 30°, 60°, and 90° of knee flexion showed differences in coupled internal-external rotation of 11.2°, 9.8° and 10.0°at these same flexion angles, respectively.

Knee laxity can be quantified in all 6-DOF, but several of these degrees of freedom are either suppressed or coupled because of the constraints provided by the ligamentous and bony structures. For example, axial rotations of the tibia causes coupled anterior-posterior translations and varus-valgus rotations [114, 115]. Most medial-lateral translations of the knee are coupled with rotations of the knee, and are affected by the bony geometry of the tibia and femur[116]. Consequently, medial-lateral translations are a poor predictor of knee laxity. Proximal-distal translations do not pose a problem when reproducing average kinematics, since the average kinematics are applied to the knee specific reference positions. Anterior tibial translation has been shown to be a good indicator of knee laxity in this study and in many previous studies[117-122]. Anterior knee laxity (length of the toe region of the load-displacement curve) was not only variable for the porcine model in this study, but has also been reported to vary in cadaveric human knees by as much as 200%, 116%, 89%, and 84% at 0°, 30°, 60°, and 90° of knee flexion, respectively, in response to an anterior load of 225 N [122].

In addition to translational laxity of the knee, rotational laxity must also be considered. Varus-valgus rotations have been shown to be very repeatable between subjects in previous studies[120, 123]. Therefore, varus valgus rotations are a poor predictor of knee laxity. A constant 3 Nm internal and external rotational load throughout knee flexion-extension has been shown to result in differences in internal-external laxity in human cadaver knees of up to 200% [123] (**Figure 18)**. Furthermore, this type of test for knee laxity has been shown to be consistent and reproducible within the same cadaveric knee specimens [123]. Hence, anterior translation and internal-external rotations are the most useful knee motions for determining laxity. The average kinematics from subjects with similar anterior and internal-external knee laxity throughout the range of passive flexion-extension could be grouped, and later matched to a comparable cadaveric specimen.

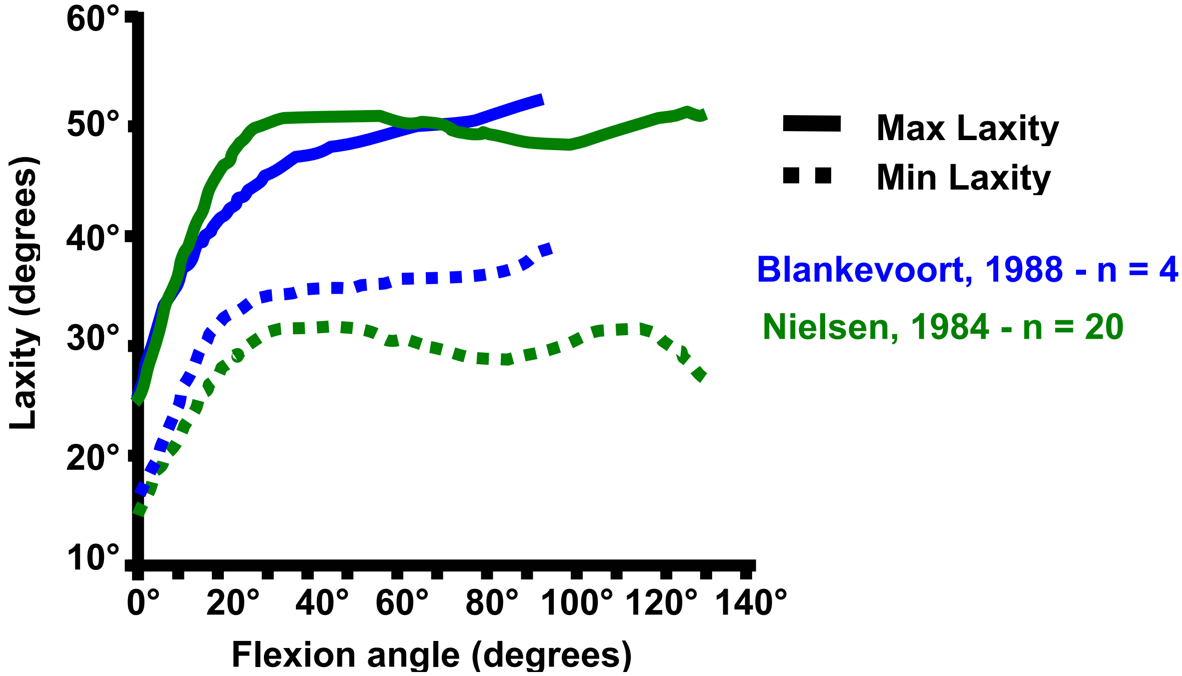

Figure 18. Differences in internal-external knee laxity in response to a 3 Nm internal and external load throughout knee flexion-extension

In previous studies performed on the robotic/UFS testing system [9], anterior tibial translations were slightly higher than those found in this study. This is likely due to the differences in the experimental protocol. This study limited internal-external rotation to ±10° from the reference position for loading due to the internal-external laxity of porcine knees. In addition, this study applied loads at the clinical center of the knee instead of at the geometric center[9, 114]. The clinical center is a point lateral to the geometric center used by clinicians to apply loads during clinical exams. On the robotic/UFS testing system, the clinical center is determined based on the natural valgus alignment of each tibia with respect to its femur[114]. Nevertheless, differences in the *in situ* force in the ACL between this study and previous studies[9] were not significant. An additional limitation of this study was posed by the IE laxity of the porcine model. Human cadaveric knees are stiffer than porcine knees during internal external rotation thus; some of the challenges caused by the rotational laxity of the porcine knee will not affect studies performed in human cadaveric knees.

One of the reasons why porcine knees are very lax is due to the absence of the iliotibial band, which plays a major role in maintaining rotatory stability *in vivo.*

Experimental data utilizing averaged kinematics will be valuable for advancing the understanding of ACL forces during *in vivo* activities. Evaluating the forces of different ACL reconstructions during *in vivo* activities will allow graft function to be compared under clinically relevant loading conditions. Results from this research will also elucidate the forces in the ACL during various rehabilitation exercises. This will allow rehabilitation protocols to be optimized, avoiding excessive forces in the ACL that could damage the healing graft. However, average kinematics must account for coupled motions and knee laxity by grouping knees with similar laxity.

7.0 FUTURE DIRECTIONS

Two extension of the proposed methodology of reproducing average kinematics will be presented. Method 1 would extend the current methodology by accounting for knee laxity in multiple degrees of freedom by grouping sets of knees with similar laxity. Method 2 would account for differences in knee laxity by using the end of the toe region of the load displacement curve as a reference instead of the passive path of flexion extension. Method 1 will require clinicians to apply internal, external, and anterior loads though out the range of flexion extension until a firm stopping point is felt. This will provide the translational and rotational laxity of each subject tested. Load displacement curves for internal, external, and anterior loads would be compared between subjects. Subjects with similar correlation coefficient for both anterior and internal-external laxity would be grouped and the average kinematics calculated. Once a database of subject kinematics has been established, cadaveric knee specimens would be tested in a similar manner to human subjects. An anterior load of 100 N, and an internal, and an external load of 3 Nm would be applied to the cadaveric knee at several flexion angles throughout the range of flexion-extension. Next, the set of average subject kinematics that correlates best to the knee laxity of the cadaveric knee would be reproduced.

Method 2 would require clinicians to apply anterior loads though out the range of flexion extension until a firm stopping point is felt. Recorded subject kinematics would use the end of the toe region instead of the passive as a reference. A 100 N anterior load would also be applied with the robotic/UFS testing system to the cadaveric knee to establish a reference position from which the recorded subject kinematics could be added. Method 1 should also account for coupled motions in some way; this could be accomplished by grouping knees with similar laxities as in Method 1.

In spite of the challenges posed by this research project, several alternative solutions were established. This study provides promising evidence that this innovative methodology can be extended to account for knee laxity and coupled motions. Thus, an accurate understanding of the forces in the ACL and ACL graft during *in vivo* activities may be obtained using an extension of this methodology.

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