Biomechanics of the ACL: Measurements of in situ force in the ACL and knee kinematics

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Abstract

Injury of the anterior cruciate ligament (ACL) can lead to knee instability associated with damage to other knee structures and the increased risk of degenerative joint disease. This has led to the frequent use of intra-articular tissue grafts for ACL reconstruction in an attempt to restore normal knee function. Despite moderate success, the continued failure of ACL reconstruction to restore normal knee kinematics has led many investigators to study the role played by the ACL in normal knee motion. At our research center, we have focused on the development of a new and innovative approach to measure multiple degree of freedom (DOF) knee kinematics and to determine the in situ forces within the ACL. A unique testing system utilizing a 6-DOF robotic manipulator and universal force–moment sensor (UFS) has been developed such that these measurements can be made in a non-contact fashion while allowing a series of experiments to be performed on the same knee. In this manuscript, we will describe the functional and mathematical development of the robotic/UFS system and its use in a series of studies designed to give insight into the function of the ACL and ACL grafts. Our first study investigated the effect of constrained vs. unconstrained knee motion on anterior tibial translation and on the in situ force in the ACL. We found that unconstrained multiple-DOF knee motion significantly increased anterior tibial translation. While the magnitude of the in situ force remained similar to the more constrained condition, its direction, point of application and distribution between the anteromedial (AM) and posterolateral (PL) bundles were found to significantly change. These findings led us to investigate the effect of knee flexion angle and magnitude of anterior applied tibial load on the in situ force in the ACL and its bundles during unconstrained knee motion. We found the PL bundle of the human ACL to carry a greater proportion of the in situ force than the AM bundle near knee extension. Also, the change in magnitude of the in situ force in the PL bundle with changing knee flexion angle was similar to that of the entire ACL. This led us to conclude that the PL bundle must play a significant role in ACL function and in resisting anterior tibial load and that it should receive more serious consideration during ACL reconstruction. Lastly, we used our new testing system to examine two popular ACL reconstruction techniques: bone–patellar tendon–bone (BPTB) and quadruple semitendinosus/gracilis hamstring (QST/G) grafts. We compared them in terms of restoration of anterior tibial translation and reproduction of the in situ force in the intact ACL. Each reconstruction was performed on the same knee, allowing us to minimize interspecimen variability and take advantage of paired statistical analysis. We found that while both reconstructions effectively reduced anterior tibial translation secondary to anterior tibial loading to a level not significantly different from the ACL intact knee, use of a QST/G graft may be advantageous, as it reproduced the in situ forces of the intact ACL more closely. This series of three studies has garnered quantitative data to further our understanding of how the ACL functions and has yielded new concepts to help improve ACL reconstruction. Through the use of a robotic manipulator, our testing system has the capability to reproduce complex physiologic loading conditions that allow us to evaluate the efficacy of various ACL reconstruction techniques and

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1. Introduction

The function of the knee joint is mediated by a complex interaction of its component parts. These component parts include the patella, femur, and tibia; the ligaments and capsule; the articular cartilage and menisci; and the muscles. The interdependence of these structures is such that injury or failure of any one of them leads to deterioration of the overall function of the joint. Ligaments are particularly vulnerable as they are subject to low-degree sprains in virtually every injury of the knee [1,2], and in some cases, are ruptured completely. Complete ligament ruptures, especially of the anterior cruciate ligament (ACL), can lead to the clinical symptom of knee instability [3–7], which may be associated with tears and degeneration of the menisci and articular cartilage, as well as increased laxity of other knee ligaments [8–14].

Because the ACL also fails to heal with a quality that restores its function [15], numerous reconstructive procedures have been developed for patients demanding high levels of knee performance. The most common reconstructive techniques utilize intra-articular autografts [16–36]. While reconstruction has vastly improved patient outcome [34,37–42], allowing sufficient restoration of stability to permit some high-demand athletes to return to their sport of choice, it fails to completely restore normal knee kinematics or to prevent the associated onset of joint degeneration [43,44]. This failure to achieve an adequate restoration of normal knee kinematics has resulted in the need to improve upon our current understanding of ACL function. By learning more about the role of the ACL in the intact knee, it is hoped that improved reconstruction and rehabilitation protocols can be developed on a more scientific basis. This has motivated many studies specifically aimed at the measurement of knee motions (kinematics) and of forces in the ACL in response to various external loads applied to the knee joint.

Initially, studies focused on the measurement of tibial translation in response to anterior tibial loading and the concept of primary and secondary restraints was developed [45–49]. The ACL was shown to be the primary restraint to anterior tibial translation [47]. More recently, efforts were focused on determining the forces in the ACL. These studies, primarily performed under A-P loading, revealed that the anteromedial (AM) and posterolateral (PL) bundles carry unequal proportions of the ACL force [50–63]. This observation, besides providing insight into the normal function of the ACL, can serve as a basis for evaluating reconstructive procedures. Specifically, by comparing this data to that achieved by ACL grafts, variables such as how, where, and under what tension the ACL graft should be placed intra-operatively can be evaluated. Furthermore, the results of such comparisons can also help in the determination of how to position the knee to achieve the safest, most effective post-operative rehabilitation protocol.

In our research center, we have focused on the effect which different testing conditions can have on the measurement of the in situ forces in the ACL. Specifically, we have found that varying knee flexion angle, knee constraint (or degree-of-freedom, DOF), and the magnitude and direction of the applied tibial load can result in different in situ forces in the ACL. Because the techniques we use to reconstruct the ACL clinically are often based on the quantitative data gathered in the laboratory, it is imperative that the data be gathered accurately and in an appropriate physiologic setting.

This manuscript will briefly summarize what we have accomplished in assessing the in situ forces in the ACL in response to multiple-DOF knee motion, specifically during knee flexion/extension with anterior tibial loading, and the clinical application of these results. To achieve a more realistic experimental setting, we developed a new testing system — a robotic manipulator in combination with a universal force–moment sensor (UFS) which can directly measure the force in the ACL with simultaneous measurement of multiple-DOF knee motion. We used this technology to perform a series of studies examining the effect of knee constraints, knee flexion angle, and magnitude of applied anterior tibial load on the in situ forces in the ACL. This approach, also applied to objectively evaluating current bone–patellar tendon–bone and semitendinosus– gracilis hamstring ACL reconstruction procedures, has enhanced our understanding of the function of the ACL and ACL grafts. It is hoped that with these and future studies, our methodology will lead to new and improved con-
cepts of ACL reconstruction and post-operative re-
habilitation.

2. What is the in situ force?

The orthopaedic literature has provided a detailed description of the complex anatomic nature of the ACL [64–74], theorizing that different fibers of the ACL are recruited at different flexion angles. As it is impossible to examine each fiber in the ACL individually, the ‘bundle theory’, based on an anatomic division of the ACL in the pig, goat, sheep and cow [67] and on a functional division in the human [68] has established itself as a good compromise. This theory provides a method of consistently dividing the ACL into its component AM and PL bundles, allowing comparisons designed to give insight into its biomechanical nature, especially regarding the force distribution within the ACL. Knowing this information gives the surgeon an important goal to achieve when selecting such variables as graft type, placement, and tension for ACL reconstruction; i.e. to reproduce the force present in the intact ACL. This force is called the in situ force.

By definition, the in situ force in a knee ligament is the force carried by the ligament in response to a given load applied to the knee while the knee (and ligament) remains intact. The in situ force within a ligament characterizes the role that the ligament is playing in resisting an external load applied to the intact knee. Because force, by definition, is a vector, a complete measurement of in situ force must include its magnitude, direction, and point of application. In theory, the force vector originates from a single point within the ligament insertion sites, and may be oriented in different directions within the substance of the ligament with changing externally applied load and knee flexion angle. This is especially important in the ACL, where different fibers are loaded at different knee positions, thus causing the net in situ force vector to change its direction and point of application. While the literature has traditionally reported only the magnitude of the force, we contend that the direction and point of application of the in situ force are equally important and must also be considered.

3. Brief summary of past testing systems

Several techniques have been developed to determine the in situ force in the ACL which can be roughly classified by whether contact is or is not made with the ligament, the DOF through which the knee is allowed to move during testing, and the ease with which the flexion angle and applied external load can be changed. Contact methods involve making direct physical contact with the ligament midsubstance by a force measuring device. These methods include the use of buckle transducers [60,63,75–78], implantable transducers [79–86] and the implantation of a steel prosthetic cable in place of the ligament [87,88]. While these methods can all measure the in situ force in the ACL, they require instrumentation and contact with the ACL which may alter the in situ length of the ACL and thus the in situ force measured.

For this reason, many investigators developed non-contact approaches to determine the forces in the ACL. These methods include the placement of strain gauges near the ligament insertion site [57], the placement of in-line external force transducers [61,89] and the use of X-rays to make kinematic calculations [58]. These methods avoid the problems of the contact measurements, but are limited in their ability to easily vary flexion angle and applied load.

Whether a contact or non-contact method is used, the system design must also consider the number of DOF through which the knee may move. Historically, studies on knee mechanics were performed allowing the knee to move in only a single DOF, that is, anterior translation of the tibia with respect to the femur. In reality, the knee has the potential to move in 6 DOF: three translations — proximal–distal (PD), anterior–posterior (AP) and medial–lateral (ML); and three rotations — internal–external (IE), varus–valgus (VV) and flexion–extension (FE). The anterior drawer and Lachman tests done at a given knee flexion angle allow 5-DOF tibial motion, as demonstrated by studies that showed the tibia not only to displace anteriorly but also to rotate internally [45,90,91]. Therefore modeling the anterior drawer in the laboratory using a setup allowing only AP translation (1 DOF) may be too restrictive. We hypothesized that the in situ forces in the ACL due to anterior tibial loading measured at a chosen knee flexion angle under conditions of 1-DOF knee motion might not represent the physiological situation of 5-DOF knee motion. While other investigators made similar hypotheses, developing testing systems to measure the ACL in situ force under multiple-DOF knee motion [53,55,59,60,92], their systems often made indirect measurements or lacked the ability to measure the in situ force at multiple flexion angles, under multiple applied loads, or with complete determination of magnitude, direction and point of application (Fig. 1).

4. Development of the robotic/UFS system

Hence, we set out to develop a new technique with the explicit goal of finding a way to measure the in situ force (including magnitude, direction and point of application) in the ACL and its bundles in response to external loading. Our goal was to directly measure these forces without making ligament contact, while...
maintaining multiple-DOF knee motion. Furthermore, the system had to have the ability to repeat the multiple-DOF knee motion after initial testing such that numerous testing conditions could be evaluated within the same knee and thus interspecimen variability eliminated.

With these aforementioned goals, we designed a testing system which incorporated a robotic manipulator and a UFS [93]. The robot manipulator is a six-joint series-articulated manipulator (Unimate, PUMA model 762). It is a position-controlled device with repeatability for position and orientation of less than 0.02 mm and 0.02°, respectively. It has a force capacity of 450 N (newtons) with the arm fully extended. In addition to position-control, the capability of force-control is added by using the force–moment data provided by the UFS (JR3, model 4015) to guide the robotic motion (Fig. 2). The full-force scales of the UFS are ±900 N in its z direction, ±450 N for its other two directions, and ±50 N m (newton-meters) for moments. Repeatability is within the range of 0.2 N in forces and 0.01 N m in moments. Thus, the robotic manipulator and UFS testing system are adequate to produce loads similar to those used during clinical evaluation of the patient [46].

The robotic manipulator is used to learn the complex motion of the knee specimen in response to external loads and then to reproduce this motion in subsequent tests after the specimen has been modified (e.g. removal of the ACL from the knee). The UFS is used to record three forces and three moments in both cases. The ability to duplicate the motion exactly before and after modifying the specimen enables us to apply the principle of superposition. That is, the vector difference of the force as measured by the UFS before and after modifying the specimen equals the in situ force in the structure removed from the specimen. In this manner, we can determine the magnitude, direction and point of application of the in situ forces of the ACL [50,93–99]. The following section will describe the control theory of our testing system and the mathematical arguments on which it is based.

4.1. The joint coordinate description

The testing of cadaveric knee specimens is done with the tibia mounted to the UFS, which in turn is mounted on the end effector of the robotic manipula-
tor (Fig. 2). The femur is mounted rigidly to the robot base. By ensuring a rigid body connection between the UFS and tibia, and the robot base and femur, the UFS is able to record forces and moments applied to the specimen.

To characterize the complex 6-DOF motions of the knee joint in a clinically significant way, we use a non-orthogonal joint coordinate description originally developed by Grood and Suntay [100] and Chao [101] (Fig. 3). This system allows for the rigorous description of the 6-DOF motion of the human knee and can be represented as a sequence of six serial links between the femur and tibia [95].

While the motion of the human knee is completely described in this 6-DOF description, the associated forces and moments are measured with respect to a Cartesian frame fixed with the UFS [95]. Therefore the three forces and three moments measured by the UFS must be mathematically transformed into forces and moments along and about the non-orthogonal joint coordinate description. The details of this mathematical approach can be found in published works from our research center [96] and also in Paul et al. [102].

Briefly, we define a $6 \times 1$ force–moment vector, $^s F$, consisting of the three forces ($f_x, f_y, f_z$) and three moments ($m_x, m_y, m_z$) measured by the UFS (Fig. 4),

$$ ^s F = (f_x, f_y, f_z, m_x, m_y, m_z)^T. \tag{1} $$

We then similarly define another $6 \times 1$ force–moment vector, $^j F$, with respect to the 6-DOF joint coordinate description as defined above, i.e. three forces ($f_{ML}, f_{AP}, f_{PD}$) and three moments ($m_{FE}, m_{VV}, m_{IE}$) such that

$$ ^j F = (m_{FE}, f_{ML}, m_{VV}, f_{AP}, m_{IE}, f_{PD})^T. \tag{2} $$

Through the application of the principle of virtual work, a $6 \times 6$ Jacobian matrix, $J$, can be obtained as shown below [103]:

$$ J = J^T \begin{bmatrix} d_{AP}S_{VV}S_{IE} & -d_{AP}S_{VV}S_{IE} & -d_{AP}S_{VV}C_{IE} & -d_{AP}S_{VV}C_{IE} & 0 & 0 \\ d_{AP}C_{VV} & S_{VV} & 0 & 0 & 0 & 1 \\ -d_{AP}S_{VV}C_{IE} & -d_{AP}C_{VV}S_{IE} & C_{VV}C_{IE} & d_{AP}C_{VV}C_{IE} & 0 & 0 \\ -C_{VV}S_{IE} & 0 & C_{IE} & 0 & 0 & 0 \\ S_{VV} & 0 & 0 & 0 & 1 & 0 \\ C_{VV}C_{IE} & 0 & S_{IE} & 0 & 0 & 0 \end{bmatrix} \tag{3} $$

where the $6 \times 6$ matrix $J^T$ depends upon the relative positions of the sensor and tibia. The terms in this Jacobian matrix represent the displacements (e.g. $d_{PD} =$ proximal–distal displacement) and the sines and cosines of the angles (e.g. $S_{VV} =$ the sine of the $VV$ angle) in the joint coordinate description, as shown in Fig. 3.

Because the sensor coordinate system is rigidly fixed to the tibial coordinate system, the Jacobian $J^T$ is constant and is determined by measuring the relative positions and orientations of the two coordinate systems [103].

This transformation relates the two force–moment vectors, $^j F$ and $^s F$, such that

$$ ^j F = J^T ^s F. \tag{4} $$

Thus, all forces and moments measured with respect to the sensor coordinate system can be expressed with respect to the joint coordinate description (Fig. 3).

### 4.2. Combination of position- and force-controlled modes

The following iteration procedure was developed to operate the robotic manipulator in a force-controlled mode...
mode. The controller reads the position and orientation from the robotic manipulator and reads the forces and moments from the UFS. After the effects of tares and the fixture gravitational force are removed, the results are translated to the specific movement center location and the resultant forces and moments are compared to the target values. Utilizing the instantaneous load-deflection information, the movements needed to reduce the errors are calculated, initially in the tibial coordinate system, transformed into the robotic movement system, and finally, the motion increment is commanded.

In passive knee flexion and extension, the target forces and moments are specified to be zero residual force and moment magnitudes less than 2 N and 0.2 N m, respectively. Once equilibrium is reached through the iterative process, the position is recorded as a control point and an increment of rotation is effected, in order to cause a change in flexion angle, and this process is repeated. The path of flexion–extension is thus a series of the final positions, which are stored for later use to duplicate the positioning of the joint. In practice, a preliminary flexion–extension path is determined, and the range of flexion is repeated 10 times to precondition the knee, at which point the neutral path positions are refined to their final values. To apply a loaded condition (AP, VV, etc.), the same process is used with target forces and/or moments being non-zero, while the flexion angle is held at one of the selection alternatives.

The general procedure to test knees consists of the following steps:

1. Under force–moment control, the robotic manipulator learns the 6-DOF path of the knee joint between the selected end points of flexion, e.g. 0–90° of knee flexion.
2. The knee is then tested in 5 DOF at a fixed knee flexion angle while the controller, subjecting the specimen to a specified external load at a specified rate, records the positions via the robotic manipulator and records forces and moments via the UFS. In practice, this process is repeated 10 times per test to allow pre-conditioning of the knee and thus minimize ligament viscoelastic effects.
3. A ligament or other soft tissue is surgically cut.
4. After the knee specimen is thus modified, the robotic manipulator repeats the identical 5-DOF knee position while the UFS is used to record the forces and moments.
5. The in situ force within the removed ligament or soft tissue is then calculated (superposition) as the difference between the UFS measurements taken before and after removal of the ligament. Although the UFS is part of the control system when a path of motion is being learned under force–moment control, it is used only for data acquisition when the path of motion is repro-
4.3. The in situ force vector

4.3.1. Magnitude and direction

The magnitude of the in situ force, \( F \), in the transected ligament is determined as follows:

\[
|F| = \sqrt{f_x^2 + f_y^2 + f_z^2}.
\]  

(5)

The direction cosine of the in situ force is determined as follows:

\[
a_x = \frac{f_x}{|F|}, \quad a_y = \frac{f_y}{|F|}, \quad \text{and} \quad a_z = \frac{f_z}{|F|}.
\]  

(6)

For ease of description, the direction of the force is transformed from a representation using Cartesian coordinates to one using spherical coordinates. In this representation, elevation, termed \( \alpha \), physically represents the angle the force makes with the tibial plateau; and deviation, termed \( \beta \), represents the angle between the force and the sagittal plane as viewed from the posterior aspect of the tibia (Fig. 5). Both \( \alpha \) and \( \beta \) are determined as follows:

\[
\alpha = \sin^{-1} \left( \frac{f_z}{|F|} \right), \quad \beta = \tan^{-1} \left( \frac{f_x}{f_y} \right).
\]  

(7)

4.4. Point of application

The point of application of the in situ force in the ACL is defined as the point where the line of action of the in situ force passes through the tibial insertion site of the ACL. This is determined in a two-step process. First, we determine the line of action of the in situ force in the ligament, and then we determine the location of the tibial insertion site in space. The point where these two intersect is the point of application of the in situ force in the ligament.

To locate the line of action of the in situ force, we first determine the point \( p = (p_x, p_y, p_z)^T \) where the in situ force vector intersects the surface of the sensor (i.e., the \( x-y \) plane of the sensor coordinate system). This requires us to divide the moments by \( f_z \), and

\[
p_x = \frac{-m_x}{f_z}, \quad p_y = \frac{m_y}{f_z}, \quad \text{and} \quad p_z = 0.
\]  

(8)

If the direction of the force is nearly parallel to the surface of the UFS, and thus \( f_z \) is very small, then the determination of its line of action will diminish in accuracy. A slight measurement error will result in a large error in the location of the point \( p \).

To resolve this difficulty, we use a force transformation scheme by rotating the sensor coordinate system mathematically such that the \( z \)-axis of the new Cartesian coordinate system \((z')\) is parallel to the direction of the measured in situ force (Fig. 4). As described by Fujie et al. [96], this results in a point \( p' = (p'_x, p'_y, p'_z)^T \) along the line of action of the in situ force described with respect to the rotated coordinate system as

\[
p'_x = -\frac{m'_x}{f'_z}, \quad p'_y = \frac{m'_y}{f'_z}, \quad \text{and} \quad p'_z = \text{arbitrary}.
\]  

(9)

In essence, this is a parametric equation of the line of action, which is parallel to the \( z' \)-axis. That is, choosing any two arbitrary values for coordinate \( p'_x \) will result in two points, \( p' \), such that a line is formed between them. Once the line of action is determined with respect to the new, primed coordinate system, it can be mathematically rotated back to its original orientation using a transformation matrix [96]. The parametric equation of the line of action of the in situ force with respect to the sensor coordinate system \((x, y, z)\) is then determined as

\[
p_x = \frac{-f_x m_y + f_y m_z + f_z \sqrt{f_x^2 + f_y^2 + f_z^2}}{f_x^2 + f_y^2 + f_z^2}, \quad p_y = \frac{-f_y m_x + f_x m_z + f_z \sqrt{f_x^2 + f_y^2 + f_z^2}}{f_x^2 + f_y^2 + f_z^2}, \quad p_z = \frac{-f_z m_x + f_x m_y + f_y \sqrt{f_x^2 + f_y^2 + f_z^2}}{f_x^2 + f_y^2 + f_z^2},
\]  

(10)

where \( s \) is a parameter that indicates a location along the line of action of the in situ force with respect to the sensor coordinate system [96,104].

Because the parametric equations that describe the line of action of the in situ force vector are symmetric, there is no dependence upon the direction of the in situ force. The error remains quite small even when the measured force is parallel to the surface of the UFS. This mathematical manipulation enables accurate determination of the line of the in situ force.

Next, we determine the location of the tibial insertion site of the ACL with respect to the sensor coordinate system. To do this, elastin stain is used to mark six locations around the tibial insertion of the ACL. A compressive force of approx. 15 N is applied to the first marker with a 1-mm diameter pin while the
forces and moments developed in the UFS are recorded. A second compressive force of approximately the same magnitude is then applied to the same marker from a different direction. As for the in situ force described above, the line of action of each compressive force is determined. The point where these two lines intersect with respect to the sensor coordinate system is the elastin stain marker’s position in 3-D space. By repeating this process for each of the six elastin stain markers, a plane describing the ACL tibial insertion site in 3-D space is defined relative to the sensor coordinate system.

With this geometrical definition of the ACL tibial insertion and the line of action of the in situ force in the ACL with respect to the sensor coordinate system, the point of intersection between the two can be determined. That is, the plane representing the tibial insertion site of the ACL is generically represented by:

$$ax + by + cz = d,$$

where $x$ can be substituted for with $p_x(s)$, $y$ substituted for with $p_y(s)$, and $z$ substituted for with $p_z(s)$. Solving for $s$ gives the unique point which satisfies the geometric plane and the line of action (i.e. their intersection). Then solving $p_x(s)$, $p_y(s)$ and $p_z(s)$ with the aforementioned $s$ gives the point in the sensor coordinate system which represents the point of application of the ACL in situ force on the tibial plateau [96,104].

4.5. Knee kinematics

If one considers the determination of the in situ force in a ligament to be the engineer’s approach to evaluating the ligament, then the determination of tibial translation following ligament injury would be the clinician’s approach [105,106]. The robotic/UFS system has the ability to evaluate the knee specimen in this fashion also. Following knee specimen alteration (e.g. cutting the ACL), the knee is subjected to the same external load as before specimen alteration and the new position recorded. By comparing this knee position with the position recorded prior to specimen alteration, the change in tibial translation due to the specimen’s alteration can be determined. For the case of analyzing ACL transection, only the anterior tibial translation is considered from the 5-DOF tibial motion. In this way, measurements can be considered to be equivalent to those made in the clinic using measuring devices such as the KT-1000.

5. Experimental studies

Utilizing this technology, we have performed a series of three experimental studies on porcine and human cadaveric knees. The first study was designed to verify the validity of the robotic/UFS system and to evaluate the effect of varying joint constraint on the measurement of in situ forces. The second experiment was designed to evaluate the effect of knee flexion angle and applied anterior tibial load on the in situ force in the human ACL and its distribution between the AM and PL bundles. The third evaluated the efficacy of bone–patellar tendon–bone and semitendinosus and gracilis ACL reconstruction procedures.

In each experiment, comparison of the magnitude of the in situ force between groups was made using analysis of variance (ANOVA) followed by Fisher’s PLSD post-hoc tests. The direction and point of application data were compared through the construction of multivariate confidence intervals determined using the means and standard deviations of the normalized data [52]. Lastly, to evaluate the effect of varying knee flexion angle and applied anterior tibial load on the magnitude of the ACL in situ force, the data were compared using general linear modeling techniques with Student’s $t$-test evaluation. In all cases, significant differences were set at $P < 0.05$.

5.1. Study 1: In situ forces and knee kinematics under multiple-DOF knee motion

5.1.1. Goals and hypotheses

We hypothesized that there was a significant dif-
ference between the in situ forces in the ACL and between kinematic knee motions measured with different constraint conditions placed on the knee, e.g. 5-DOF vs. 1-DOF. To do this, we chose to study porcine knees because of their uniform size, age and bone quality. Specifically, we compared the in situ forces within the ACL, distribution of this force within the two bundles of the ACL, and the anterior tibial translation secondary to applied anterior tibial loads. By using the robotic/UFS system, we were able to perform both the 1-DOF and 5-DOF tests (Fig. 6) on each knee, allowing us to eliminate inherent interspecimen variability and take advantage of paired statistical analysis techniques.

5.1.2. Methods and results

Eight porcine knee joints were dissected free of musculature, leaving the joint capsule and knee ligaments intact. The tibia and femur were cut to a length of 20 cm from the joint line and secured within thick-walled aluminum cylinders and rigidly fixed to the robot as described in Section 4 (Fig. 2).

Using force–moment control, the robotic manipulator was used to first determine and record the 6-DOF path of intact joint motion between 30° and 90° of passive knee flexion [93,95]. This was done to determine the path of passive flexion of a particular knee specimen and to allow subsequent repositioning of the joint at any desired flexion angle by the robot. Specific flexion angles along this path later provided the starting point (zero reference) for externally applied tibial load tests. The knee was placed in 30° flexion and preconditioned with anterior–posterior (AP) loading to 100 N at a rate of 50 mm/min for five cycles to minimize ligamentous viscoelastic effects.

The robot then applied AP loading to ±100 N at a rate of 20 mm/min to the joint at 30°, 60° and 90° while allowing unconstrained, 5-DOF knee motion. Joint kinematics (i.e. position and orientation) during each loading were recorded for later use (Fig. 7).

The test was then repeated with the robot constrained to moving the joint in only 1-DOF (i.e. only AP) motion. AP loading to ±100 N at a rate of 20 mm/min was applied to the joint at 30°, 60° and 90°. As for the 5-DOF case, the joint positions and orientations under load were recorded.

Next, all tissues around the knee were removed except the ACL, leaving a femur–ACL–tibia complex (FATO). The specimen was checked to ensure that only the ACL provided resistance to anterior tibial loading. The 5-DOF path of tibial motion for the intact knee under anterior tibial loading was then reproduced by the robot while the forces and moments applied to the tibia were recorded by the UFS. This was done with the knee at 30°, 60° and 90° of flexion. Given the identical path of motion for the intact knee and the ACL-only knee, the forces and moments measured during the second test represent those in the ACL in the intact knee under 5-DOF anterior tibial loading [52,96]. The magnitude, direction and point of application of the in situ force in the ACL were then determined, as detailed in Section 4.

To determine the in situ forces in the ACL under anterior tibial loading during 1-DOF knee motion, the previously recorded 1-DOF tibial motion path of the intact knee was repeated on the ACL-only knee. Again, as the path of 1-DOF motion was identical for the intact and ACL-only knee, the in situ forces developed in the ACL were determined and the magnitude, direction and point of application calculated.

A similar procedure was used to determine the distribution of in situ forces between the AM and PL bundles of the ACL. To examine the 5-DOF in situ force in the PL bundle, the AM portion of the ACL was bluntly transected, and the 5-DOF path of intact knee motion under AP tibial loading was repeated at 30°, 60° and 90° of knee flexion. The UFS outputs were measured and the in situ forces in the PL bundle determined. The in situ forces in the AM portion of the ACL were then calculated using vector algebra. Similarly, the test was repeated with the knee restricted to 1-DOF anterior tibial motion (Fig. 7).

As documented by previous investigators, unconstrained, 5-DOF anterior tibial loading resulted in significantly greater anterior tibial translation in the intact knee when compared to the 1-DOF case (1.4 times, \( P < 0.05 \)). Associated with this translation was a notable total tibial rotation of \( 13 \pm 4° \) (mean ± S.D.).

However, the magnitudes of the in situ forces developed in the ACL were not significantly different; under 100 N anterior load, the in situ forces were

Fig. 6. Diagrams detailing the joint motions allowed under (a) 1-DOF and (b) 5-DOF anterior tibial load.
Fig. 7. Flowchart detailing the sequence of testing and data obtained from Study 1: 1 DOF vs. 5 DOF.

98.4 ± 7.8 N and 105.3 ± 9.2 N, for 5-DOF and 1-DOF, respectively, at 30° flexion. This trend was also true at 60° and 90° of flexion, although the magnitudes of in situ force were slightly reduced.

On the other hand, the constraint condition was found to significantly affect the direction of the in situ force in the ACL at all knee flexion angles. The direction of the in situ force in the ACL corresponded approximately with the direction in which the AM portion of the ACL inserted in the tibia. Specifically, under 5-DOF anterior tibial loading of 100 N at 30° knee flexion, the elevation (α) and deviation (β) of the in situ force in the ACL were 15.7 ± 4.4° (α) and 4.9 ± 2.1° (β). This direction changed with 1-DOF anterior tibial loading of 100 N, to 26.8 ± 4.1° (α) and 11.0 ± 3.2° (β) (Fig. 8). As the applied load increased, α was noted to decrease slightly in both cases, while β remained almost constant.

Under 5-DOF anterior tibial loading, the point of application of the in situ force in the entire ACL was located near the center of the ACL insertion site. This is shown on a representative specimen (Fig. 9), with the point of application of ligament force (representing the centroid of ligament loading). The more constrained, 1-DOF testing case resulted in the point of application of the in situ force in the entire ACL to be located more anteriorly in the posterior region of AM portion of the tibial insertion site (Fig. 9).

The distribution of force within the ACL was also found to be significantly affected by the constraint condition. For the 5-DOF case, the distribution of forces between the AM and PL bundles of the ACL...
was relatively uniform at all flexion angles (Fig. 10). But in the 1-DOF case, the in situ forces in the AM and PL bundles were roughly equal at only 30° of knee flexion, with the AM bundle supporting significantly greater forces at 60° and 90° flexion ($P < 0.05$). Under an applied anterior load of 100 N, the AM supported 80% of the total ligament force at 90° flexion in the 1-DOF case, but this was reduced to 58% in the 5-DOF case.

5.1.3. Summary of findings and clinical relevance

From this study, we have found that there is a distinct difference between the direction, point of application and force distribution of the in situ force in the ACL as they pertain to the unconstrained (5-DOF) and constrained (1-DOF) conditions. We also observed a significant difference in the anterior tibial translation of the intact knee between 5-DOF and 1-DOF, demonstrating that axial tibial rotation normally occurs with an anterior tibial displacement. This restriction of motion allows secondary restraints [47] to be subjected to a higher load at a smaller tibial translation than might otherwise occur. This, in turn, means that the anterior tibial load to which the tibia was to be loaded is reached at a smaller tibial translation, and this smaller tibial translation results in changes of the direction and point of application of the in situ force in the ACL and its bundles.

Clinically, these observations imply that ligament injuries may or may not occur with a given applied anterior tibial load, depending on the constraints to knee motion. In other words, a force applied to the knee when the knee is free to move in multiple DOF may not cause injury. However, the same force applied to a more firmly constrained knee, such when a football player is cutting to one side, may change the direction, point of application, and distribution of the resulting in situ force in the ACL such that injury may occur. Similar reasoning may also be applied to the physical examination. Classical teaching states that the Lachman test be performed from the same side of the examination table as the knee being examined [107]. By so doing, the temptation to internally rotate the tibia with respect to the femur is avoided. This is
important because applying an internal rotation will limit the DOF through which the knee can move and thus reduce anterior tibial translation.

5.2. Study 2: In situ force and force distribution within the human ACL

5.2.1. Goals and hypothesis

Having displayed the importance of multiple-DOF motion on the in situ forces in the ACL, we hypothesized that measuring the in situ forces in the ACL and its bundles in human cadaveric specimens might present different results than previously published data in which the knees were tested in a more constrained condition. Therefore we performed this study to evaluate the in situ forces in the ACL and its AM and PL bundles at various knee flexion angles when the knee is subjected to various applied anterior tibial loads while allowing unconstrained 5-DOF motion. It was felt that this new and more realistic data on the in situ forces in the ACL might give insight into the function of the ACL and serve as a basis for designing ACL reconstructive procedures using tissue grafts which better restore normal knee kinematics.

5.2.2. Methods and results

Nine human cadaver knee joints (age 44–81 years, mean 71) were dissected free of all musculature except the popliteus muscle, leaving the joint capsule intact. Examination confirmed that these specimens displayed no previous ligamentous injury or signs of degenerative joint disease. The tibia and femur were cut approx. 20 cm from the joint line and secured to the testing system as described in Section 4 (Fig. 2).

Following mounting, the robot was used to learn the path of knee joint kinematics during passive flexion–extension from 0° to 90°. This multiple-DOF path of knee motion was recorded for later reproduction. The robot then applied anterior tibial loading up to 110 N at a rate of 20 mm/min to the specimen at 0°, 15°, 30°, 60° and 90° of knee flexion. The position, orientation and joint forces (i.e. UFS readings) were recorded when the anterior tibial load reached 22 N, 44 N, 66 N, 88 N and 110 N.

Through a medial parapatellar incision, the AM bundle was identified as the portion of the ACL under tension during passive flexion of the knee to 90° [68], and was sharply transected. The robot then repeated the previously recorded 5-DOF knee kinematics at 0°, 15°, 30°, 60° and 90° of flexion, and the new force data registered at the UFS at the tibial displacements which previously corresponded to anterior tibial loads of 22 N, 44 N, 66 N, 88 N and 110 N recorded. Since the joint was following the identical path of motion before and after cutting of the AM bundle, the principle of superposition was applied. That is, the vector difference of the force data recorded in the intact knee at a given applied load and flexion angle and in the AM bundle deficient knee at the same position represented the in situ force in the AM bundle for that applied load and flexion angle [99]. Thus, the in situ force in the AM bundle was determined for five successively increasing applied anterior tibial loads at five knee flexion angles. The magnitude and direction of these in situ forces were determined as described previously in Section 4.

To determine the in situ force in the PL bundle, the remainder of the ACL was transected and the identical testing procedure repeated. Lastly, applying the principle of superposition, the in situ force in the whole ACL could be calculated by analyzing the vector difference of the force data recorded in the intact knee (prior to transection of the AM bundle) and the ACL deficient knee (after transection of the PL bundle).

Under 110 N of applied anterior tibial load the magnitude of the in situ force in the whole ACL varied from a high of 110.6 ± 14.8 N at 15° of knee flexion to a low of 71.1 ± 29.5 at 90° of knee flexion. For 22 N of applied anterior tibial load the magnitude of the in situ force of the ACL varied from a high of 25.7 ± 3.7 N at 15° of knee flexion to a low of 12.8 ± 7.3 N at 90° of knee flexion (Fig. 10a). Statistical analysis revealed the magnitude of the in situ force to be significantly different with respect to applied anterior tibial load (P < 0.05) and between all flexion angles above 30° (P < 0.05) (Fig. 10b).

For the AM bundle, the force magnitude varied from a high of 47.4 ± 34.2 N at 60° of knee flexion to a minimum of 32.6 ± 13.3 N at 0° of knee flexion with 110 N anteriorly applied tibial load. When 22 N anterior tibial load was applied, the magnitude of the in situ force varied from 13.5 ± 5.2 N at 15° of knee flexion to 9.9 ± 7.7 N at 90° of knee flexion (Fig. 11a). Thus, 25–30% of the maximum in situ force can be reached at as little as 22 N of applied anterior tibial load. Changing knee flexion angle did not significantly affect the magnitude of the in situ force in the AM bundle, and only large changes (> 44 N) in applied load caused statistically significant changes in magnitude.

Contrary to these findings, significant changes in the magnitude of the in situ force in the PL bundle with changing knee flexion angle were observed to be similar to those found for the whole ACL. Under an anterior tibial load of 22 N the magnitude of the in situ force in the PL bundle varied from a maximum of 13.7 ± 8.1 N at 0° of knee flexion to a minimum of 4.6 ± 2.5 N at 90° of knee flexion. With 110 N anterior tibial load, the magnitude of the in situ force in the PL bundle of the ACL varied from a maximum of
75.2 ± 18.3 N at 15° of knee flexion to a minimum of 26.2 ± 14.4 N at 90° of knee flexion (Fig. 11b).

The direction of the in situ force in the whole ACL, as described by angles α and β, varied with both knee flexion angle and applied tibial load. α decreased (i.e. became more parallel with the tibial plateau) with increasing flexion angle, a significant difference being noted with a flexion angle of 60° or higher. However, there was no significant change with applied load. β, on the other hand, decreased (i.e. became more parallel with the sagittal plane) with increasing applied load, while showing an insignificant increase with increasing flexion angle (Fig. 12).

The AM bundle showed similar trends of changing α and β with changing flexion angle and changing applied load. Again, α showed significant changes only with changing flexion angle, and again primarily with changes of 60° or greater in knee flexion angle. The PL bundle, in contrast, showed no apparent change with changing flexion angle or changing applied load (Fig. 12).

5.2.3. Summary of findings and clinical relevance

Our results showed the magnitude of the in situ force in the whole ACL to be maximum with anterior loads applied near 15° of knee flexion (Fig. 10b). These data are in general agreement with previous findings in the literature with the knee tested in similar or more constrained conditions [53,54,56,58]. However, the magnitude of the in situ force in the AM and PL bundles revealed some interesting findings. The magnitude of the in situ force in the PL bundle changes with flexion angle and applied load. These changes were remarkably similar to those for the whole ACL (Fig. 13). The AM bundle, on the other hand, remains relatively constant throughout the range of motion and applied load (Fig. 11a). Our findings on the AM bundle are in agreement with
previous investigations performed using testing systems allowing unconstrained motion [53,60] and supports the contention of Fuss et al. [66,67] that the AM bundle contains ‘guiding fibers’, which are always under tension. However, we have found the PL bundle to have a larger magnitude of the in situ force than does the AM bundle at knee flexion angles below 45° (Fig. 13). This is in contrast with previous studies performed in which the human cadaveric knee was constrained to 1 DOF [51,54] and the AM bundle was shown to carry the majority of the force at all flexion angles.

The direction of the in situ force of the ACL becomes more horizontal and slightly more lateral with increasing knee flexion. The force vector for the AM bundle showed similar directional changes. The direction of the PL bundle, however, is more variable. With increasing applied anterior tibial loads, the direction of the in situ force in the ACL approaches parallel with the mid-sagittal plane, while that of the AM and PL bundle follow this trend, though to not as strong a degree. These observations of changing in situ force direction can be explained purely by anatomy. As the knee flexes, the femoral origin of the ACL moves posteriorly and inferiorly [68] such that the anatomic angle made by the ACL with the tibial plateau decreases. Thus, it is only natural that the direction of the in situ force, angle \( \alpha \), should follow. The internal tibial rotation with respect to the femur (i.e. the coupled motion) decreases the anatomic angle that the ACL makes with the sagittal plane, and thus, the angle \( \beta \) likewise follows.

Our findings suggest that when human cadaveric knees are unconstrained, the PL bundle’s role in response to anterior tibial loading may be more significant than previously thought. Restoration of knee function with ACL reconstruction, may hence need to account for the contribution of the PL bundle. Many authors have recommended anterior placement of the tibial tunnel to reproduce the AM bundle primarily because of the belief that the AM bundle carried higher forces than the PL bundle and because it was thought to restore ‘isometricity’ [29,34,108–110]. Recent investigations have recommended progressively greater posterior placement of the tibial tunnel for reasons concerned with avoiding anterior impingement [111], as well as in an attempt to reproduce the good functional outcome of patients with AM-bundle-only tears [112]. Our data offers one additional biomechanical substantiation of the soundness of the latter approach. Furthermore, the fact that the magnitude of the in situ force in the PL bundle follows the general characteristics of the magnitude of the in situ force in the ACL may explain why patients with AM-bundle-only tears are reported to be doing well [112]. Whether one can expect better outcomes from reconstruction of the ACL with graft placement to mimic the PL bundle rather than the AM bundle now deserves debate. Furthermore, our data suggest that a double-bundle reconstruction for the purpose of reproducing both the AM and PL bundle characteristics of the intact ACL may be worth further consideration.

Another clinically significant application of our data relates to the selection of knee flexion angle under which the ACL graft should be tightened intra-operatively. If emphasis of reconstruction is on the AM bundle, then graft tensioning should be performed at 60° of knee flexion where the intact AM bundle is under the greatest force. However, if emphasis is on the PL bundle, then graft tensioning should be carried out with the knee more extended. For the double-bundle reconstruction, the AM-mimicking bundle should be tightened near 60°, while the PL-mimicking bundle should be tightened near extension.

Likewise, the in situ force data can also affect post-operative rehabilitation protocols. Should the stated goal be to keep the forces in the reconstructed graft relatively low, then the safest angle of knee flexion for rehabilitation would vary depending on which bundle was emphasized in the reconstruction. If the emphasis was on the AM bundle, our data suggest that keeping the knee near extension would produce the lowest in situ graft forces. If the emphasis was on the PL bundle, then the lowest in situ graft forces would occur near 90° of knee flexion.
5.3. Study 3: In situ forces in tissue grafts for ACL reconstruction

5.3.1. Goals and hypotheses

Intra-articular reconstruction using tissue grafts to restore ACL function has gained widespread popularity [30,31,33–36,42,113,114]. It is also recognized that no one graft material or method of fixation has been found to completely reproduce the complex function of the ACL. Factors including intra-articular positioning, pre-tensioning of the graft, and initial graft tension can all contribute to the outcome of ACL reconstruction. Thus, the possibility of multiple confounding variables makes the evaluation of different types of grafts and different methods of fixation difficult.

The robotic UFS system, however, offers an opportunity to evaluate and compare different ACL reconstruction techniques by using the same knee to perform multiple reconstructions without having to disrupt the remaining anatomic structures. These measurements can be made because the robot can repeat appropriate knee positions after each graft fixation with high fidelity, a property which can be used to eliminate interspecimen variability. Also, by utilizing the robot to position the knee during graft fixation, the confounding effect of knee position on graft fixation can be minimized.

With this background knowledge, we set our goals to conduct a study to compare two popular reconstruction procedures: bone-patellar tendon–bone with interference screw fixation vs. quadruple semitendinosus and gracilis tendon with endobutton and suture-post fixation (Fig. 14) [115–118]. We hypothesized that the in situ forces in these grafts in response to anterior tibial loads would be different from each other and would also fail to reproduce the in situ forces of the intact ACL.

5.3.2. Methods and results

Human cadaveric knees (ages 70–79, all male) were evaluated radiographically to determine the status of the intra-articular cartilage, the intercondylar roof angle, and patellar tendon length. Specimens with a high degree of degenerative joint disease or other abnormal finding were not included.

A medial incision centered over the pes anserinus insertion was made to allow harvesting of the semitendinosus and gracilis tendons. Both grafts were cut to approx. 180 mm in length and doubled over to form a quadruple tendon graft (QST/G). This was wrapped in normal-saline soaked gauze and placed aside for later use.

The tibia and femur of each specimen were then cut approx. 20 cm from the joint line and secured to the robotic/UFS system as previously described in Section 4. The multiple-DOF path of passive flexion–extension for each intact knee, as well as the kinematics during 110 N of anterior tibial loading at 0°, 15°, 30°, 60° and 90° of knee flexion, were determined with the forces and moments recorded by the UFS.

The ACL of each knee was transected arthroscopically (Linvatec, Largo, FL), causing minimal disturbance to the remaining joint anatomy. The 5-DOF knee kinematics previously recorded at knee flexion angles of 0°, 15°, 30°, 60° and 90° were then repeated and the resulting forces and moments recorded by the UFS. The principles of superposition were then used to determine the in situ forces in the ACL of the intact knee (see Section 4). In addition, a 110 N anterior tibial load was applied to the ACL deficient knee at 0°, 15°, 30°, 60° and 90° of knee flexion and a set of ACL-deficient kinematics determined. From this, an ACL-deficient anterior tibial displacement

Fig. 14. Schematic diagram showing semitendinosus/gracilis ACL reconstruction and bone–patellar tendon–bone reconstructions.
was determined for a 110 N anterior tibial load applied at each flexion angle.

The ACL was then reconstructed using the previously harvested QST/G autograft via the inside-out technique with endobutton (Acufex, Mansfield, MA) femoral fixation and suture-post tibial fixation [21,119]. Prior to placement, the QST/G autograft was preconditioned using a Graftmaster (Acufex, Mansfield, MA) tissue preparation system for 5 min. A notch-plasty was performed, and the tibial tunnel placed using a tibial drill guide (Acufex, Mansfield, MA) positioned 4 mm anterior to the posterior cruciate ligament. This produced a graft which best simulated the PL bundle of the ACL to avoid the anterior impingement as described by Howell et al. [111]. The femoral tunnel was drilled at the center of the femoral insertion site using a 7-mm femoral tunnel guide (Anthrex, Naples, FL). To avoid graft-tunnel mismatch, the sagittal angle of the tibial hole was placed at 55° and the depth of the femoral bony socket drilled to 35 mm. Following femoral endobutton fixation, the robot was utilized to passively flex the knee from 90° to 15° while a graft tension of 45 N was applied using a spring scale. With the knee positioned at 15° and the graft tensioned to 45 N, the suture-post fixation was performed. At this point, the entire testing procedure was repeated on the QST/G graft reconstructed knee, and the post-reconstruction anterior tibial displacements and the in situ forces in the QST/G graft were determined.

After removing the QST/G graft, a 10-mm wide bone–patellar tendon–bone (BPTB) graft was harvested and used to reconstruct the ACL-deficient knee by securing the graft in the femoral and tibial tunnels using interference screw (Acufex, Mansfield, MA) fixation. Prior to tibial fixation, the robot was utilized to move the knee from 90° to 15° through its passive range of flexion–extension while the graft was placed under 45 N of tension. By performing these steps as they were performed in the QST/G reconstruction, the confounding factor of knee positioning during fixation was eliminated and only the reconstructive technique evaluated. Again, the identical testing procedure was performed, and the in situ forces in the BPTB graft and post-reconstruction anterior tibial displacements determined.

Anterior displacements of the intact, ACL-deficient and ACL reconstructed knees are shown in Table 1. After transecting the ACL, anterior tibial displacements were increased by a maximum of 102 ± 13.8% at 15° of knee flexion and a minimum of 50.0 ± 13.6% at 60° of knee flexion. Both reconstruction procedures restored the anterior tibial displacements to those of the intact knees (P < 0.05). The only exception to this is the QST/G reconstruction at 15° of knee flexion (Fig. 15). This suggests the adequacy of both procedures to restore a Lachman or anterior drawer test to normal. There was no significant difference between the post-reconstruction anterior tibial displacements of the QST/G and BPTB grafts.

The magnitude of the in situ force in the intact ACL in response to a 110 N anterior tibial load varied from 50.5 ± 12.8 N at 90° of knee flexion to 86.1 ± 18.3 N at 15° of knee flexion (Fig. 16). Both the QST/G and BPTB reconstructions had similar trends as their maximums and minimums also occurred at the same knee flexion angles. However, the magnitude of the in situ force in the QST/G reconstruction varied from 33.6 ± 25.1 N to 68.6 ± 15.2 N and the BPTB reconstruction varied from 28.3 ± 22.6 N to 71.4 ± 21.1 N (Fig. 16). The difference between each reconstruction and the normal ACL was statistically significant (P < 0.05) at 15° of knee flexion for the QST/G reconstruction, and at 15°, 60° and 90° for the BPTB reconstruction. We found no significant difference between the magnitudes of the in situ force carried by the two reconstructions. In general, both reconstruction procedures carried more than 80% of the in situ forces of the intact ACL near knee extension less than 30° of knee flexion, but the graft force reduced with respect to intact at higher knee flexion angles (Fig. 17).

5.3.3. Summary of findings and clinical relevance

We found that both the QST/G graft and BPTB graft reconstructions can restore anterior tibial displacement to the level of the intact knee. This displays our technical ability to perform ACL reconstruction in the laboratory in a fashion that achieves results that would normally be considered acceptable in a clinical setting. Based on the data on anterior tibial translation alone, it would appear that both

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**Table 1**

Anterior displacements of the ACL intact, ACL-deficient and ACL reconstructed knees (mean ± S.D.) under 110 N anterior tibial load vs. knee flexion angle (n = 4)

<table>
<thead>
<tr>
<th></th>
<th>0°</th>
<th>15°</th>
<th>30°</th>
<th>60°</th>
<th>90°</th>
</tr>
</thead>
<tbody>
<tr>
<td>Intact ACL (mm)</td>
<td>4.9 ± 0.9</td>
<td>6.9 ± 0.6</td>
<td>9.6 ± 1.3</td>
<td>9.4 ± 2.3</td>
<td>7.7 ± 1.4</td>
</tr>
<tr>
<td>ACL-deficient (mm)</td>
<td>9.6 ± 0.8</td>
<td>14.1 ± 2.1</td>
<td>11.3 ± 1.5</td>
<td>14.3 ± 5.0</td>
<td>11.9 ± 3.8</td>
</tr>
<tr>
<td>QST/G (mm)</td>
<td>5.4 ± 1.4</td>
<td>8.9 ± 1.1</td>
<td>11.4 ± 1.2</td>
<td>10.5 ± 2.1</td>
<td>8.0 ± 1.8</td>
</tr>
<tr>
<td>BPTB (mm)</td>
<td>5.7 ± 2.1</td>
<td>8.6 ± 1.6</td>
<td>11.3 ± 1.5</td>
<td>11.1 ± 2.8</td>
<td>8.5 ± 1.2</td>
</tr>
</tbody>
</table>
reconstructive techniques are equally effective. However, examining the magnitude of the in situ forces suggests otherwise.

Both reconstructions had in situ force magnitudes similar to those in the intact ACL at full extension to 30° of knee flexion. However, only the QST/G graft reconstruction was able to restore the intact ACL in situ force to a significant degree at higher knee flexion angles. These findings indicate that the QST/G reconstruction may actually have an advantage over BPTB reconstruction because its in situ forces secondary to anterior tibial loading more closely mimic those normally carried by the intact ACL at a wider range of knee flexion angles.

This result, however, is limited to the external loading conditions studied. Future evaluation of the QST/G and BPTB reconstructions will need to include other combined loading modes (e.g., anterior loading and internal rotation). Nevertheless, it is clear that our new testing system does offer the opportunity to explore these possibilities and eventually can yield a data base to discern the advantages of one reconstruction procedure over another in the hope of optimizing our current reconstruction techniques.

6. Summary and closure

Understanding the role of the ACL in knee function has been the focus of many investigations. Many methods of analysis, including anatomical and biomechanical evaluation, have been used to arrive at our present level of understanding that the ACL indeed plays an important role in limiting excessive anterior tibial translation, axial tibial rotation and varus–valgus angulation. The ACL has the necessary geometric complexity to perform these functions. Anatomically and functionally, it is divided into two bundles: the AM and PL. Historically, studies have emphasized the importance of the AM bundle over the PL bundle. Our studies, however, have yielded new insight into the important role played by the PL bundle in knee kinematics in response to anterior tibial loading. This new insight has significant implications for future ACL reconstruction.

Contributing to our new understanding of the ACL, our research center has relied on a newly developed testing system. Through the use of a robotic manipulator in concert with a UFS, the in situ force vector in the ACL in response to external loads can be measured directly without contact being made with the ligament. This testing system allows the knee to move without constraint at any preselected knee flexion angle.

Our new testing system was first verified to be...
capable of reproducing the results obtained by an established experimental methodology. We then moved on to evaluate the in situ forces in the ACL and its component bundles with the joint unconstrained. Using porcine knees, we obtained data on the in situ force in the ACL, its distribution between the AM and PL bundles and tibial displacement during applied anterior tibial loading. We found significant increases in the anterior tibial displacement in the normal unconstrained knee as compared to a knee that is constrained to A-P motion only. Also, there was a change in the direction and point of application of the in situ forces in the whole ACL. The magnitude of the in situ forces in the AM and PL bundles also changed, with the PL bundle carrying greater in situ forces with less constraints. These findings have clinical relevance in terms of injury mechanism and diagnostic techniques. For injury mechanism, we theorize that a reduction in the DOF, such as during a football player’s cutting maneuver, may cause the knee to displace a smaller amount than it otherwise might, and result in a change in the in situ force and force distribution in the ACL possibly leading to injury.

In terms of diagnosis of ACL injury, the constraints applied to the knee can be critical. If a physical exam maneuver such as the Lachman test is performed while imparting an internal or external rotation to the knee (thus decreasing the typical DOF of the Lachman test from 5 to 4), the resulting anterior tibial translation might be less than if the knee were allowed to move in 5 DOF. This could result in a small anterior translation in the face of an ACL injury, possibly masking the pathology.

Using human knees, we have been able to demonstrate the importance of the PL bundle in response to anterior tibial loading. The higher magnitude of the in situ force in the PL bundle as compared with the AM bundle near extension together with its reproduction of the trend of changing magnitude of the in situ force in the whole ACL with changing knee flexion angle and changing applied anterior tibial load leads us to emphasize the importance of this bundle and the notion that it should not be forsaken in ACL reconstruction.

In terms of current reconstructive techniques, our findings of which flexion angle resulted in the highest in situ forces in the bundles should help guide surgeons’ decisions intraoperatively. Attention must be given to whether an AM or PL bundle is being mimicked by the graft. Specifically, our data imply that for a reconstruction emphasizing reproduction of the function of the AM bundle, that fixation should be done with the knee flexed to near 60°. However, for reconstructions emphasizing the PL bundle, graft fixation might be more appropriately performed near 15°. These same findings also apply to post-operative rehabilitation protocols, i.e. if the goal is to keep graft forces at a minimum, then the appropriate knee position will also depend on the graft placement.

The debate over selection of the most appropriate graft for ACL reconstruction led us to evaluate the QST/G and BPTB grafts and their ability to reduce anterior tibial translation as well as to reproduce the in situ force profile of the intact ACL. We have found that both grafts quite effectively reduce the anterior tibial translation secondary to anterior tibial loading to a level not significantly different from the intact ACL knee. This data confirmed the reasoning behind their high acceptance by the clinical community. However, the QST/G graft reconstruction was superior to the BPTB graft reconstruction in that its in situ force more closely mimicked that of the intact ACL. The differences between the two reconstruction procedures suggest that, for this loading case, the QST/G graft reconstructions may be advantageous but that much investigation remains to be done to delineate superiority.

7. Future directions

From the above discussion, it is clear that the robotic/UFS system is a powerful new methodology for musculoskeletal joint experimentation. It has the advantage of being able to determine the ligament in situ force vector without making mechanical contact with the tissue, and the data can be obtained over the whole range of knee motions. Its application to the knee under anterior tibial loading has demonstrated its potential to investigate knee ligament forces and joint kinematics under complex loading conditions. Using this force–moment control scheme applied to combinations of external loads including anterior–posterior, medial–lateral and proximal–distal forces and flexion–extension, varus–valgus and internal–external moments, complex physiologic loading conditions such as stair climbing and squatting can be modeled. We intend to use these loading modes to examine the in situ forces in the ACL and other knee ligaments as well. Similarly, in situ forces in ACL grafts and the effect of various ACL grafts on knee kinematics can be evaluated. Other factors in ACL reconstructions, such as the initial graft tension, graft type and graft placement and their effect on knee kinematics, will be studied in the future.

As we have seen from this discussion, the robotic/UFS system can apply a known set of joint kinematics to the knee repeatedly. This offers the possibility of investigating knee behavior and in situ ligament forces during in vivo activities. Recently, the point cluster technique developed by Andriacchi et al. [120,121] has been utilized to collect in vivo knee
motion data during normal gait and stair climbing. The point cluster system is a non-invasive method of measuring 6-DOF movements (three angular displacements and three translational displacements) of a patient’s knee joint. Thus, we can collect knee kinematic data during daily functional activities and post-operative rehabilitation protocols. We then can reproduce this motion on cadaveric specimens using our testing system. By doing so, we will be able to indirectly determine the in vivo ligament and/or graft forces. The results will be valuable and can serve as guidelines for optimizing ACL reconstructions as well as for designing the most suitable post-ACL reconstruction rehabilitation regimen. By evaluating changes of knee kinematics in the same patient over time following ACL reconstruction, the effect of time on the restoration of knee kinematics by various reconstructive techniques can also be evaluated. We believe that such investigations will greatly improve our understanding of the ACL and ACL graft, and help to improve ACL reconstruction procedures and rehabilitation protocols.

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