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In Situ Forces in the Anterior Cruciate Ligament and Its Bundles in Response to Anterior Tibial Loads

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Summary: The anterior cruciate ligament has a complex fiber anatomy and is not considered to be a uniform structure. Current anterior cruciate ligament reconstructions succeed in stabilizing the knee, but they neither fully restore normal knee kinematics nor reproduce normal ligament function. To improve the outcome of the reconstruction, it may be necessary to reproduce the complex function of the intact anterior cruciate ligament in the replacement graft. We examined the *in situ* forces in nine human anterior cruciate ligaments as well as the force distribution between the anteromedial and posterolateral bundles of the ligament in response to applied anterior tibial loads ranging from 22 to 110 N at knee flexion angles of 0-90°. The analysis was performed using a robotic manipulator in conjunction with a universal force-moment sensor. The *in situ* forces were determined with no device attached to the ligament, while the knee was permitted to move freely in response to the applied loads. We found that the *in situ* forces in the anterior cruciate ligament ranged from 12.8 ± 7.3 N under 22 N of anterior tibial load applied at 90° of knee flexion to 110.6 ± 14.8 N under 110 N of applied load at 15° of flexion. The magnitude of the *in situ* force in the posterolateral bundle was larger than that in the anteromedial bundle at knee flexion angles between 0 and 45°, reaching a maximum of 75.2 ± 18.3 N at 15° of knee flexion under an anterior tibial load of 110 N. The magnitude of the *in situ* force in the posterolateral bundle was significantly affected by knee flexion angle and anterior tibial load in a fashion remarkably similar to that seen in the anterior cruciate ligament. The magnitude of the *in situ* force in the anteromedial bundle, in contrast, remained relatively constant, not changing with flexion angle. Significant differences in the direction of the *in situ* force between the anteromedial bundle and the posterolateral bundle were found only at flexion angles of 0 and 60° and only under applied anterior tibial loads greater than 66 N. We have demonstrated the nonuniformity of the anterior cruciate ligament under unconstrained anterior tibial loads. Our data further suggest that in order for the anterior cruciate ligament replacement graft to reproduce the *in situ* forces of the normal anterior cruciate ligament, reconstruction techniques should take into account the role of the posterolateral bundle in addition to that of the anteromedial bundle.

The anterior cruciate ligament is one of the most frequently injured structures within the knee joint (39). Because this ligament fails to heal in a manner that would restore its function, reconstructive techniques have been developed for patients demanding high levels of knee performance (3,8,34,41,42,47). These reconstructions, while generally considered to be clinically successful, neither fully restore normal knee kinematics nor reproduce the function of the intact anterior cruciate ligament (12,47). Furthermore, a reconstruction may not prevent secondary degenerative changes (12).

For a successful long-term outcome, the anterior cruciate ligament replacement graft must restore normal knee kinematics and reproduce the complex forces and force distribution of the intact anterior cruciate ligament (29,51). Thus, we proposed to determine the *in situ* forces in the intact anterior cruciate ligament and its bundles — anteromedial and posterolateral — resulting from external loads applied to the knee and to use this information as a guide for planning and evaluating reconstruction techniques and replacement grafts for the anterior cruciate ligament.

The literature has provided a detailed description of the complex anatomic nature of the anterior cruciate ligament (2,9,19-21,40,43), theorizing that different fibers of the anterior cruciate ligament are recruited to resist tibial loads at different flexion angles. As it is impossible to examine each fiber in the ligament individually, studies generally divide this ligament into two components, the anteromedial and posterolateral bundles. This division, based on an ob-

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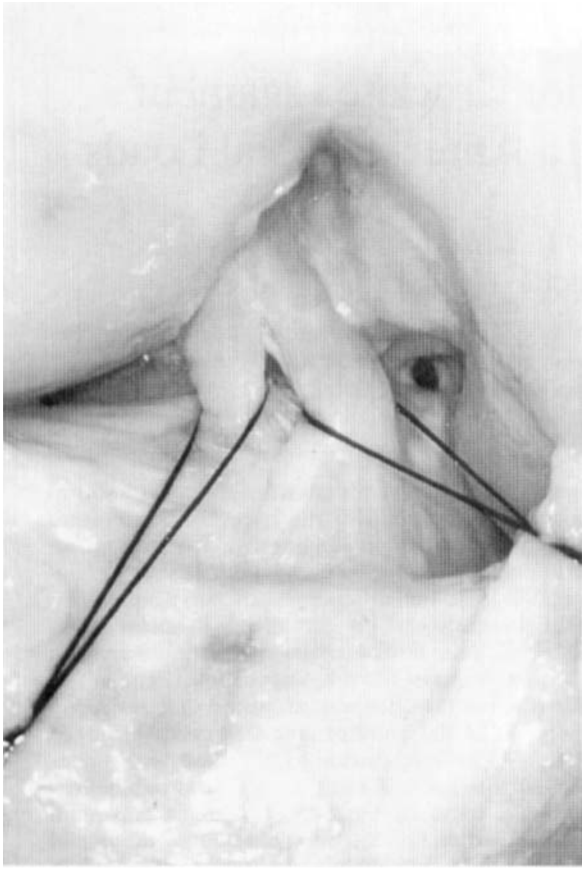

FIG. 1. Photograph of the anterior cruciate ligament viewed from the anterior aspect of the knee following separation of the anteromedial and posterolateral bundles.

served reciprocal tensioning pattern of the bundles during passive flexion-extension in the human knee (21), provides a method of consistently dividing the anterior cruciate ligament into its components. With the ligament thus divided, comparisons designed to give insight into its biomechanical nature can be made.

Initial biomechanical studies on the anterior cruciate ligament focused on its function of resisting anterior tibial translation. These studies applied constant forces to the tibia and measured its anterior translation with respect to the femur before and after transection of the anterior cruciate ligament (6,11,18, 35,36). This method provided an excellent conceptual understanding of the function of the ligament and its bundles. More recently, quantitative data on the force being carried by the anterior cruciate ligament in response to applied loads have gained significant attention (26,30,37,54). Several approaches have been developed to examine the forces in the ligament. Methods requiring ligament contact using buckle transducers mounted on the ligament (4,27,31) and implantable transducers placed within the midsubstance of the ligament (5,10,24), as well as noncontact methods using strain gauges placed near the ligament insertion site (13), a kinematic linkage system (25,49), in-line external force transducers (37), and x-rays to make kinematic calculations (52), are some of the techniques utilized. Although these studies significantly improved our understanding of the function of the anterior cruciate ligament, only a minority focused on the distribution of the *in situ* force between the bundles (23,32,49,54).

We utilized a robotic manipulator in combination with a universal force-moment sensor (UFS) (14-16,45) to study the effects of applied anterior tibial loads on the *in situ* forces in the anterior cruciate ligament and its anteromedial and posterolateral bundles as a function of angle of knee flexion. The advantage of this testing system is that it offers the ability to directly measure the magnitude and direction of the *in situ* forces in the anterior cruciate ligament and its bundles without contacting the ligament, while simultaneously permitting the knee to move in an unconstrained manner (45).

We believe the use of this technique to measure the *in situ* forces in the intact anterior cruciate ligament and its bundles in response to applied anterior tibial loads will yield results that are closely representative of the force distribution in the anterior cruciate ligament and that the data will provide new insight into the complex function of the anterior cruciate ligament and into how to achieve an effective anterior cruciate ligament reconstruction.

MATERIALS AND METHODS

Nine human cadaveric knee joints (mean donor age: 71 years, range: 44-80 years) were dissected free of all musculature except the popliteus muscle, leaving the joint capsule intact. Examination confirmed that these specimens displayed no evidence of previous ligamentous injury or signs of degenerative joint disease. The tibia and femur were cut 20 cm from the joint line and secured within thick-walled aluminum cylinders using Bond Tite body filler (Bond Tite Products, Cleveland, OH, U.S.A.). To maintain its anatomical position, the fibula was fixed to the tibia using a cortical screw and was cut to a length of 5 cm proximally. The femoral cylinder was then secured to the base of the robotic manipulator (Puma model 762; Unimate, Pittsburgh, PA, U.S.A.) through a rigid clamp, while the tibial cylinder was attached to the UFS (model 4015; JR^3, Woodland, CA, U.S.A.), which was in turn fixed to the end-effector of the six-joint robotic manipulator (45). The robot is a position-controlled device with repeatability for position and orientation of less than 0.02 mm and 0.02°, respectively (45). The UFS can measure three forces and three moments along and about a Cartesian axis system and has a repeatability within the range of 0.2 N for forces and 0.01 Nm for moments.

First, the tibial coordinate system was defined with the distal axis parallel to the longitudinal axis of the bone (i.e., medial edge of the tibial tubercle). This was accomplished by aligning the long axis of the tibia with the axis of the cylindrical clamp, which in turn is parallel to the local y-axis of the UFS. The anterior axis was then defined as perpendicular to both the distal and medial axes, thus forming a Cartesian coordinate system to be fixed in the tibia. The origin of the system was located in the center of the tibial plateau, and this local system remained fixed with the tibia. The femoral coordinate system was defined as being coincident with the tibial

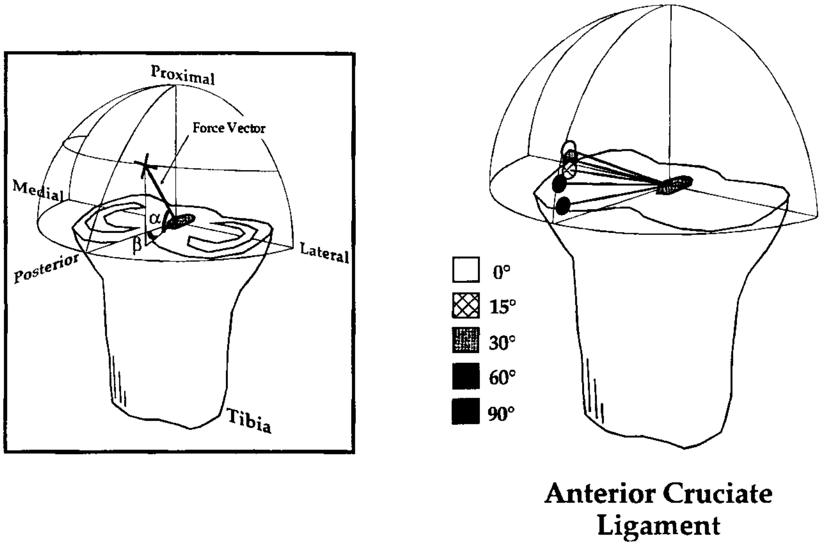

FIG. 2. The averaged direction of the *in situ* forces in the anterior cruciate ligament as viewed from the posterolateral corner at varying knee flexion angles in response to 110 N of anterior tibial loading. The insert shows the angles α and β describing the direction of the anterior cruciate ligament *in situ* force.

system at 0° of knee flexion, such that they shared identical origins and axes at 0° of knee flexion. This system was fixed within the femur. Finally, the orientation of the tibial coordinate system relative to the femoral system can be expressed using the joint coordinate system for the knee, as described first by Chao in 1980 (7) and then by Grood and Suntay in 1983 (22), to obtain the angle of flexion, varus-valgus, and internal-external rotation (16,17).

After the specimen had been prepared and mounted, the robot "learned" the six-degree-of-freedom path of motion of the intact knee joint during passive flexion-extension from 0 to 90°. To accomplish this, the robot moved the tibia in 1° increments of flexion while monitoring force-moment data from the UFS. If the sensor measured forces greater than 2.0 N (or moments greater than 0.2 Nm) along (or about) any axis, the knee position was corrected to eliminate them. In this manner, a path of knee motion from 0 to 90° that satisfied the condition of passive flexion-extension was determined (45). The robot then moved the knee to full extension and applied external anterior tibial loads, while permitting knee motion in the remaining five degrees of freedom. Loads were applied until the UFS registered 22, 44, 66, 88, and 110 N at a rate of 1 mm/sec and, as with the determination of the passive flexion-extension path, forces and moments measured by the UFS were used to guide the robot to ensure that only the desired anterior tibial load (F_0) was applied. Prior to recording data, the robot repeated the first path of 110 N load kinematics five times to minimize the effect of viscoelasticity. Following the application of each load, the position and orientation of the tibia were recorded by the robot, and then the test was repeated at 15, 30, 60, and 90°. These identical knee positions could be subsequently reproduced by the robot following specimen modification (e.g., ligament cutting).

After testing the intact knee, the anteromedial bundle of the anterior cruciate ligament was transected through an anterior arthrotomy. The anteromedial and posterolateral bundles were defined according to their "functional" anatomy, as described by Girgis et al. (21). That is, the anteromedial bundle was identified as the portion of the anterior cruciate ligament under visible tension during passive flexion of the knee to 90° (Fig. 1). Before cutting the bundle, the robot resumed the previously recorded position of the tibia with respect to the femur for the full extension, 22 N loading condition to evaluate the effect of the anterior arthrotomy. The new force measured by the UFS (F_1) was recorded. Following random transection of either bundle (e.g., anteromedial bundle), the robot resumed the same position and the sensor recorded another force (F_2). Since the relative positions of the tibia and femur before and after cutting of the anteromedial bundle were identical at the time when F_1 and F_2 were measured, the principle of superposition could be applied to determine the *in situ* force in the anteromedial bundle. That is, the difference between the force-moment data recorded after the anterior arthrotomy (F_1) and after the anteromedial bundle transection (F_2) equals the *in situ* force in the anteromedial bundle (16). To apply the principle of superposition, some assumptions are required, such as (a) the tibia and femur are rigid, (b) no physical interaction exists between the structures of interest, and (c) exact position control (45). The effects of blunt isolation of the bundles of the anterior cruciate ligament on the changes in the forces were less than 5 N under an anterior tibial load of 110 N during our preliminary experiment. In this study, the anteromedial bundle was cut first for five specimens and the posterolateral bundle was cut first for the remaining four specimens to minimize the effect of physical interaction between the bundles.

The entire procedure was then repeated for each applied anterior tibial load (22-110 N) at each flexion angle (0-90°). The mathematics for determining the magnitude and direction of the *in situ* force are detailed in the Appendix. Briefly, the direction of the *in situ* force is represented by the elevation angle (α), which physically represents the angle that the force makes with the tibial plateau, and the deviation angle (β), which physically represents the angle in the plane of the tibial plateau between the midsagittal plane and the component of the force in the tibial plateau (Fig. 2, insert).

The posterolateral bundle of the anterior cruciate ligament was then transected, and the identical testing procedure was repeated to determine the *in situ* force. That is, the robot resumed the previously determined intact knee position of the tibia with respect to the femur for each applied anterior tibial load at each flexion angle and recorded the new force (F_3) measured by the UFS. The difference between the new force (F_3) and the force measured prior to transection of the posterolateral bundle (F_2) equals the *in situ* force in the posterolateral bundle. Lastly, the difference between the force data recorded prior to transection of the anteromedial bundle (F_1) and after transection of the posterolateral bundle (F_3) was used to determine the *in situ* force

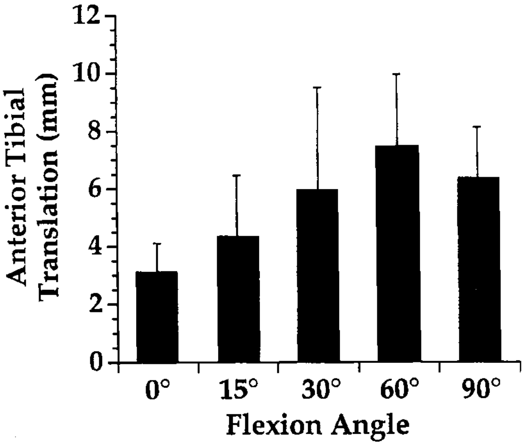

FIG. 3. Anterior tibial translation of the intact knee in response to 110 N of applied anterior tibial load. Mean ± SD and n = 9.

in the whole ligament at each flexion angle under each applied anterior tibial load.

For comparison, the magnitude of *in situ* force in each bundle was defined by projecting its vector onto the direction of the *in situ* force in the whole anterior cruciate ligament. Statistical analysis was performed using general linear modeling techniques followed by Student's *t* test comparisons (46) to examine the effects of varying knee flexion angle and applied anterior tibial load on the magnitude and direction of the *in situ* forces in the anterior cruciate ligament and the anteromedial and posterolateral bundles. These analyses were performed using SAS computer software (Cary, NC, U.S.A.). Comparisons of the magnitudes of *in situ* forces in the anterior cruciate ligament and in the anteromedial and posterolateral bundles at a given flexion angle and applied tibial load were performed using one-way analysis of variance followed by Fisher's protected least significant difference *post hoc* testing. Comparisons of the directions of the *in situ* forces in the anterior cruciate ligament and in the anteromedial and posterolateral bundles were performed using a multivariate extension of a T^2 statistic (28). A probability level of $p < 0.05$ was used as an indicator of statistically significant results in these analyses.

RESULTS

For the intact knee, anterior tibial translation in response to a 110 N anterior tibial load ranged from 3.1 ± 1.0 mm at full extension to 7.4 ± 2.5 mm at 60° of knee flexion (Fig. 3). These measured values are similar to those reported in the literature (18). Under 110 N of applied anterior tibial load, the magnitude of the *in situ* force in the anterior cruciate ligament ranged from a minimum of 71.1 ± 29.5 N at 90° of knee flexion to a maximum of 110.6 ± 14.8 N at 15° of knee flexion (Fig. 4). Lower applied anterior tibial loads resulted in correspondingly lower magnitudes of the *in situ* force in the anterior cruciate ligament. At 22 N of applied anterior tibial load, the values varied from a minimum of 12.8 ± 7.3 N at 90° of knee flexion to a maximum of 25.7 ± 3.7 N at 15° of knee flexion (Fig. 4). Statistical analysis revealed that the magnitude of the *in situ* force changed significantly with respect to applied anterior tibial load ($p < 0.05$) and as a function of flexion angle greater than 30° ($p < 0.05$), except between 60 and 90°.

For the anteromedial bundle, the magnitude of the *in situ* force ranged from a minimum of 32.6 ± 13.3 N at 0° of knee flexion to a maximum of 47.4 ± 34.2 N at 60° of knee flexion under 110 N of applied anterior tibial load. When a 22 N anterior tibial load was applied, the magnitude of the *in situ* force ranged from 9.9 ± 7.7 N at 90° of knee flexion to 13.5 ± 5.2 N at 15° of knee flexion (Fig. 5A). Statistically significant changes in the magnitude of the *in situ* force with respect to anterior tibial loading were seen only for load changes of more than 44 N, except for a significant change noted between 22 and 44 N ($p < 0.05$). Changing knee flexion angle did not significantly affect the magnitude of the *in situ* force in the anteromedial bundle (i.e., p = 0.548 for between 15 and 90°, and p = 0.105 for between 0 and 60°, respectively).

In contrast to the results seen in the anteromedial bundle, significant changes were observed in the magnitude of the *in situ* force in the posterolateral bundle with respect to knee flexion angle and applied anterior tibial load. Under an anterior tibial load of 22 N, the magnitude of the *in situ* force ranged from a minimum of 4.6 ± 2.5 N at 90° of knee flexion to a maximum of 13.7 ± 8.1 N at 0° of knee flexion. Under 110 N of anterior tibial load, the magnitude of the *in situ* force ranged from a minimum of 26.2 ± 14.4 N at 90° of knee flexion to a maximum of 75.2 ± 18.3 N at 15° of knee flexion (Fig. 5B). 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 greater than 15° ($p < 0.05$), except between 60 and 90°.

Comparison of the magnitudes of the *in situ* forces in the anterior cruciate ligament and the anteromedial and posterolateral bundles revealed that the anterior cruciate ligament differed significantly from the anteromedial bundle at all applied loads for flexion angles

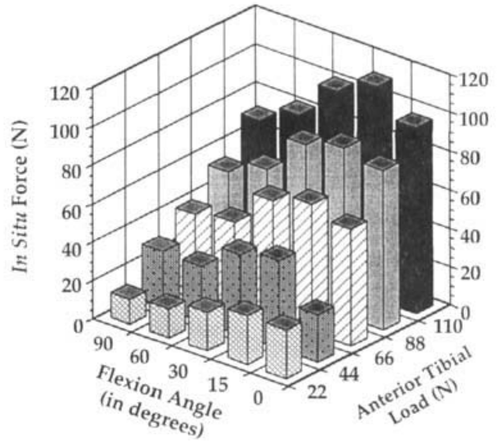

FIG. 4. Variation of the *in situ* force in the anterior cruciate ligament with respect to knee flexion angle and anterior tibial load. Shading indicates common anterior tibial load to highlight the effect of changing flexion angle.

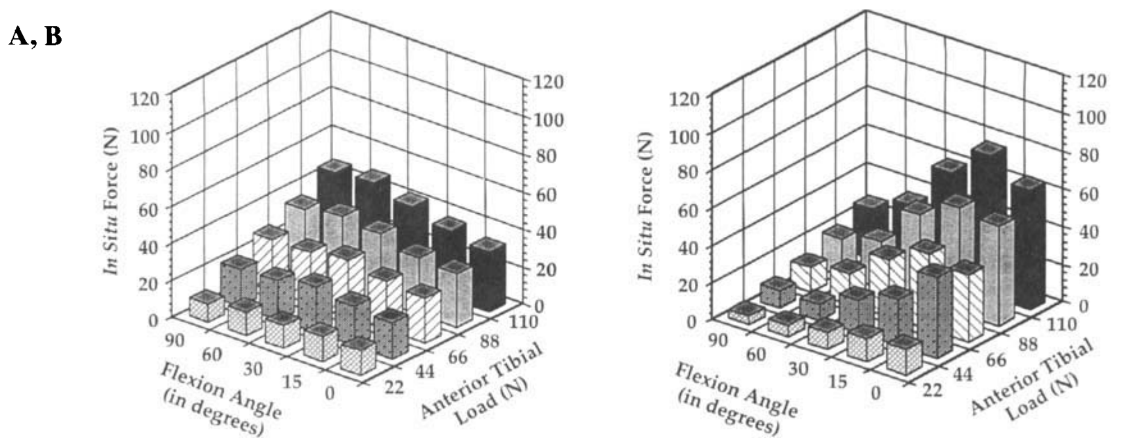

FIG. 5. Variation of the *in situ* force in **A:** the anteromedial bundle and **B:** the posterolateral bundle with respect to changes in knee flexion angle and anterior tibial load. Shading indicates common anterior tibial load to highlight the effect of changing flexion angle.

less than 60°. At 60°, the magnitudes differed significantly only under applied loads of 88 and 110 N, while at 90° no significant difference was found. The magnitudes of the *in situ* forces in the posterolateral bundle and the anterior cruciate ligament differed significantly at all flexion angles under each applied load examined. Between the anteromedial and posterolateral bundles, the magnitude of the *in situ* forces differed significantly only at flexion angles less than 30° under applied loads of more than 66 N. The magnitude of the *in situ* force in the posterolateral bundle in response to anterior tibial loading of 110 N was significantly higher than that in the anteromedial bundle at 0 and 15° of knee flexion (Fig. 6).

The direction of the *in situ* force in the anterior cruciate ligament, as described by angles α and β, varied with both knee flexion angle and applied tibial load. Angle α decreased (i.e., the *in situ* force became more parallel with the tibial plateau) as the flexion angle increased. A significant difference was noted when the flexion angle changed by at least 60°. However, there was no significant change in conjunction with changes in applied load. Angle β, on the other hand, did not change significantly with increasing flexion angle (Fig. 2 and Table 1) and decreased (i.e., the *in situ* force became more parallel with the sagittal plane) with increasing applied load.

The changes in angles α and β with varying flexion angle and applied load for the anteromedial bundle showed trends similar to those of the anterior cruciate ligament. Again, α was altered significantly only by changes in flexion angle of more than 60° and not by changes in applied anterior tibial load. Angles α and β for the posterolateral bundle, in contrast, showed no statistical change with flexion angle (Fig. 7 and Table 1) or applied anterior tibial load.

In comparing the directions of the *in situ* forces between the anteromedial and posterolateral bundles, significant differences were generally found only at 0 and 60°. Specifically, under applied anterior tibial loads of 110 N, the *in situ* force in the anteromedial bundle was found to be directed significantly more proximally and laterally at both 0 and 60° of knee flexion (Table 1).

DISCUSSION

Through the use of a robotic manipulator and a UFS, we measured the *in situ* force, including magnitude and direction, in the anterior cruciate ligament and its component bundles in response to anterior tibial loading. The *in situ* force was measured at five knee flexion angles under five applied loads while maintaining unconstrained knee motion.

Our results showed the magnitude of the *in situ* force in the anterior cruciate ligament to be maximal with anterior tibial loads applied at 15° of knee flexion and minimal when applied at 90° of knee flexion (Fig. 4). These trends confirm previously reported findings (27,49,52,54) but differ slightly from those reported by Markolf et al., who noted that the highest *in situ* forces

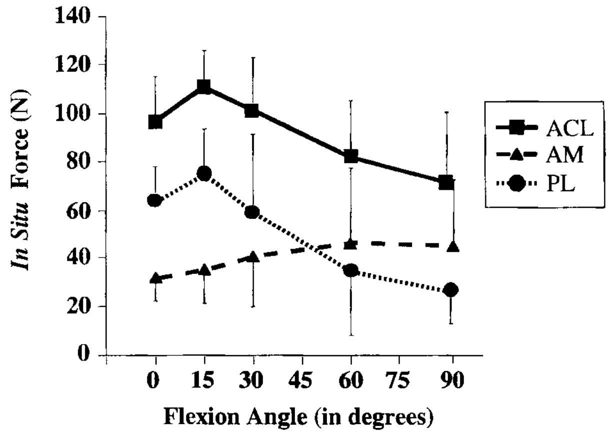

FIG. 6. Magnitude of the *in situ* force in the intact anterior cruciate ligament (ACL), anteromedial bundle (AM), and posterolateral bundle (PL) under 110 N of applied anterior tibial load. Mean ± SD and n = 9.

TABLE 1. *Direction of the* in situ *forces in the anterior cruciate ligament and the anteromedial and posterolateral bundles under 110 N of applied anterior tibial load (mean ± SD, n = 9)*

	Anterior cruciate ligament		Anteromedial bundle		Posterolateral bundle	
Flexion angle	α	β	α	β	α	β
0°	20.0 ± 4.9	1.1 ± 3.2	22.8 ± 6.6	−0.4 ± 6.9	17.7 ± 4.3	2.6 ± 3.7
15°	15.7 ± 6.9	2.1 ± 2.8	18.1 ± 11.1	1.0 ± 3.9	14.9 ± 9.5	5.7 ± 5.9
30°	18.0 ± 8.8	1.8 ± 3.2	16.5 ± 10.4	1.7 ± 4.2	22.7 ± 7.5	4.4 ± 6.5
60°	12.8 ± 10.4	−0.2 ± 4.2	14.0 ± 11.6	−2.2 ± 4.9	7.2 ± 19.0	9.9 ± 8.5
90°	8.4 ± 13.4	2.6 ± 11.6	7.4 ± 14.6	−1.3 ± 6.9	9.6 ± 12.1	8.0 ± 15.6

occurred with 5° of hyperextension (37,38). In our testing protocol, the path of passive flexion-extension was determined with a very small applied force of 2.0 N and moment of 0.2 Nm. As a result, the path of knee motion ends before hyperextension occurs, and thus this region of knee flexion was not evaluated.

The distribution of the *in situ* force between the anteromedial and posterolateral bundles, in contrast, revealed new and interesting findings. The magnitude of the *in situ* force in the posterolateral bundle in response to anterior tibial loading was greater than in the anteromedial bundle, especially when the knee was near extension. Furthermore, changes in the *in situ* force in the posterolateral bundle in relation to changes in flexion angle revealed trends remarkably similar to those for the whole anterior cruciate ligament. By comparison, the *in situ* forces in the anteromedial bundle remained relatively constant, being unaffected by changes in flexion angle and only minimally affected by changes in applied anterior tibial load (Fig. 7A). Ahmed et al. (1) and our laboratory (49), using testing systems allowing unconstrained motion, found similar results regarding the change in the *in situ* force in the anteromedial bundle between 30 and 90° of knee flexion. The observation that the *in situ* force in the bundle remains relatively constant with respect to flexion angle also supports the contention made by Fuss (20) that the anteromedial bundle contains "guiding fibers," which are always under tension. It also correlates with the statement of Sidles et al. (48), who contended that the anteromedial bundle contains isometric fibers that are active throughout the range of knee flexion.

It is interesting to note, however, that the magnitude of the *in situ* force in the posterolateral bundle in response to anterior tibial loads becomes higher than the magnitude of the *in situ* force in the anteromedial bundle when the knee is extended beyond 45° (Fig. 6). This contrasts with some previously published data on the human cadaveric knee (23,54). This difference is likely due to the fact that in the present study there were no constraint conditions placed on the knee, as constraining the knee to limited motion (i.e., anterior-posterior motion only) would have a significant effect on the *in situ* force distribution in the anterior cruciate ligament, as well as on the amount of anterior tibial translation (33). Thus, our findings suggest that when human cadaveric knees are allowed unconstrained movement in five degrees of freedom, the role of the posterolateral bundle in response to anterior tibial

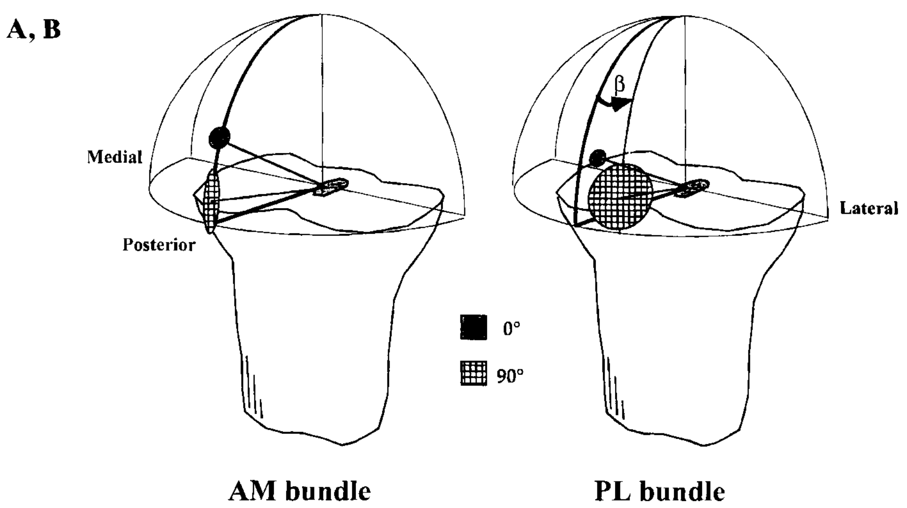

FIG. 7. The average direction of the *in situ* forces in **A:** the anteromedial (AM) bundle and **B:** the posterolateral (PL) bundle at varying knee flexion angles in response to 110 N of applied anterior tibial load at 0 and 90° of knee flexion. Each ellipse represents 1 SD of data.

loading may be more significant than was previously thought.

For a given applied anterior tibial load, the direction of the *in situ* force in the anterior cruciate ligament was found to become more horizontal with respect to the tibial plateau and slightly more lateral to the sagittal plane with increasing knee flexion. The force in the anteromedial bundle showed similar directional changes, while that in the posterolateral bundle was more variable. With increasing applied anterior tibial loads, the direction of the *in situ* force in the anterior cruciate ligament became more parallel with the midsagittal plane. These observations can be explained by the complex anatomy of the anterior cruciate ligament. As the knee flexes, the femoral origin of the anterior cruciate ligament moves posteriorly and inferiorly (21), such that the anatomic angle made by the anterior cruciate ligament with the tibial plateau decreases. Thus, it is reasonable that the angle α decreases as the flexion angle increases. Likewise, the coupled internal tibial rotation decreases the anatomic angle that the anterior cruciate ligament makes with the sagittal plane and thus the angle β follows.

The differing trends of the changes in *in situ* force in the anteromedial and posterolateral bundles with respect to knee flexion angle and anterior tibial loading imply that each bundle may play a separate but equally important role in the complex function of the anterior cruciate ligament. Thus, when restoring knee function with anterior cruciate ligament reconstruction, one should account for the contribution of both bundles of the ligament (44,50).

However, it is important to recognize that these data reflect the function of the anterior cruciate ligament and its bundles only under applied anterior tibial loads. The anterior cruciate ligament, in fact, plays many roles in maintaining knee stability, including limiting varus-valgus and internal-external rotations. To fully elucidate the function of the anteromedial and posterolateral bundles, the application of this testing system will need to be extended to include these and other more realistic loading conditions. Also, the cadaveric knees used in this study, while stable and intact, were from elderly donors. They were not of the same quality as the knees of individuals between 15 and 35 years old to which our findings might be applied (53). However, while age might affect the absolute value of the *in situ* forces in the anterior cruciate ligament and its bundles, we believe the trends observed with respect to changes in knee flexion angle and applied anterior tibial load will remain constant and can serve as a basis and guide for anterior cruciate ligament reconstruction.

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APPENDIX

The UFS returns three orthogonal force components, F_x, F_y, and F_z, for each measurement made. To determine the *in situ* force in a given structure under a given loading condition, the tibia is moved with respect to the femur such that the loading condition is applied, and a measurement of the components of the load applied to the tibia is made by the UFS ($F1_x$, $F1_y$, and $F1_z$). The structure of interest is then transected, and the robot repeats the exact position of the tibia with respect to the femur that was previously determined to represent the loading condition of interest. The components of the new force applied to the knee are then recorded by the UFS ($F2_z$, $F2_y$, and $F2_z$). The difference between the force components of the two measurements equals the force components of the *in situ* force in the transected structure under the particular loading condition evaluated. The magnitude of the *in situ* force, $|F|$, is thus determined:

$$|F| = \sqrt{(F1_x - F2_x)^2 + (F1_y - F2_y)^2 + (F1_z - F2_z)^2}$$

The direction cosines of the *in situ* force are determined:

$$a_x = \frac{F1_x - F2_x}{|F|} \qquad a_y = \frac{F1_y - F2_y}{|F|} \qquad a_z = \frac{F1_z - F2_z}{|F|}$$

For ease of description, the direction of the *in situ* force is transformed from a Cartesian representation to one based on spherical coordinates. In this representation, the elevation angle (α) physically represents the angle that the force makes with the tibial plateau, while the deviation angle (β) represents the angle in the plane of the tibial plateau between the midsagittal plane and the component of the force in the tibial plateau. This can be seen from the posterior aspect of the tibia in the insert in Fig. 2. α and β are calculated as

$$\alpha = \sin^{-1}\left(\frac{F1_y - F2_y}{|F|}\right) \qquad \beta = \tan^{-1}\left(\frac{F1_x - F2_x}{F1_z - F2_z}\right)$$

REFERENCES

1. Ahmed AM, Hyder A, Burke DL, Chan KH: In-vitro ligament tension pattern in the flexed knee in passive loading. *J Orthop Res* 5:217-230, 1987
2. Amis AA, Dawkins GPC: Functional anatomy of the anterior cruciate ligament: fibre bundle actions related to ligament replacements and injuries. *J Bone Joint Surg [Br]* 73:260-267, 1991
3. Andersson C, Odensten M, Gillquist J: Knee function after surgical or nonsurgical treatment of acute rupture of the anterior cruciate ligament: a randomized study with a long-term follow-up period. *Clin Orthop* 264:255-263, 1991
4. Barry D, Ahmed AM: Design and performance of a modified buckle transducer for the measurement of ligament tension. *J Biomech Eng* 108:149-152, 1986
5. Beynnon BD, Stankevich CJ, Fleming BC, Pope MH, Johnson

RJ: The development and initial testing of a new sensor to simultaneously measure strain and pressure in tendons and ligaments. In: *Transactions of the Combined Meeting of the Orthopaedic Research Societies of U. S. A., Japan, and Canada*, p 104. Calgary, Alberta, Committee of the Orthopaedic Research Societies, October 21-24, 1991
6. Butler DL, Noyes FR, Grood ES: Ligamentous restraints to anterior-posterior drawer in the human knee: a biomechanical study. *J Bone Joint Surg [Am]* 62:259-270, 1980
7. Chao EY: Justification of triaxial goniometer for the measurement of joint rotation. *J Biomech* 13:989-1006, 1980
8. Clancy WG, Nelson DA, Reider B, Narechania RG: Anterior cruciate ligament reconstruction using one-third of the patellar ligament, augmented by extra-articular tendon transfers. *J Bone Joint Surg [Am]* 64:352-359, 1982
9. Clark JM, Sidles JA: The interrelation of fiber bundles in the anterior cruciate ligament. *J Orthop Res* 8:180-188, 1990
10. Cummings JF, Holden JP, Grood ES, Wroble RR, Butler DL, Schafer JA: In-vivo measurement of patellar tendon forces and joint position in the goat model. *Trans Orthop Res Soc* 16:601, 1991
11. Daniel DM, Stone ML, Sachs R, Malcom L: Instrumented measurement of anterior knee laxity in patients with acute anterior cruciate ligament disruption. *Am J Sports Med* 13:401-407, 1985
12. Daniel DM, Stone ML, Dobson BE, Fithian DC, Rossman DJ, Kaufman KR: Fate of the ACL-injured patient: a prospective outcome study. *Am J Sports Med* 22:632-644, 1994
13. France EP, Daniels AU, Goble EM, Dunn HK: Simultaneous quantitation of knee ligament forces. *J Biomech* 16:553-564, 1983
14. Fujie H, Livesay GA, Kashiwaguchi S, Blomstrom G: Determination of in-situ force in the human anterior cruciate ligament: a new methodology. *ASME Adv Bioeng* BED22:91-94, 1992
15. Fujie H, Mabuchi K, Woo SL-Y, Livesay GA, Arai S, Tsukamoto Y: The use of robotics technology to study human joint kinematics: a new methodology. *J Biomech Eng* 115:211-217, 1993
16. Fujie H, Livesay GA, Woo SL-Y, Kashiwaguchi S, Blomstrom G: The use of a universal force-moment sensor to determine in-situ forces in ligaments: a new methodology. *J Biomech Eng* 117:1-7, 1995
17. Fujie H, Livesay GA, Fujita M, Woo SL-Y: Forces and moments in six-DOF at the human knee joint: mathematical description for control. *J Biomech* 29:1577-1585, 1996
18. Fukubayashi T, Torzilli PA, Sherman MF, Warren RF: An in vitro biomechanical evaluation of anterior-posterior motion of the knee: tibial displacement, rotation, and torque. *J Bone Joint Surg [Am]* 64:258-264, 1982
19. Furman W, Marshall JL, Girgis FG: The anterior cruciate ligament: a functional analysis based on postmortem studies. *J Bone Joint Surg [Am]* 58:179-185, 1976
20. Fuss FK: Anatomy of the cruciate ligaments and their function in extension and flexion of the human knee joint. *Am J Anat* 184:165-176, 1989
21. Girgis FG, Marshall JL, Al Monajem ARS: The cruciate ligaments of the knee joint: anatomical, functional, and experimental analysis. *Clin Orthop* 106:216-231, 1975
22. Grood ES, Suntay WJ: A joint coordinate system for the clinical description of three-dimensional motions: application to the knee. *J Biomech Eng* 105:136-144, 1983
23. Guan Y, Butler DL, Dormer SG, Feder SM: Anterior cruciate subunit response during anterior drawer, pp 201-204. ASME Summer Biomechanics Symposium, Columbus, Ohio, U.S.A., June 16-19, 1991
24. Holden JP, Grood ES, Cummings JF: Factors affecting sensitivity of a transducer for measuring anterior cruciate ligament force. *J Biomech* 28:99-102, 1995
25. Hollis JM: Development and application of a method for determining the in-situ forces in anterior cruciate ligament bundles [Doctoral thesis]. San Diego, University of California, 1988
26. Hollis JM, Marcin JP, Horibe S, Woo SL-Y: Load determination in ACL fiber bundles under knee loading. *Trans Orthop Res Soc* 13:58, 1988
27. Jasty M, Lew WD, Lewis JL: In vitro ligament forces in the normal knee using buckle transducers. *Trans Orthop Res Soc* 7:241, 1982
28. Johnson RA, Wichern DW: Comparison of several multivariate means. In: *Applied Multivariate Statistical Analysis*, pp 210-272. Ed by RA Johnson and DW Wichern. Englewood Cliffs, New Jersey, Prentice-Hall, 1988
29. Keira M, Yasuda K, Kaneda K, Yamamoto N, Hayashi K: Mechanical properties of the anterior cruciate ligament chronically relaxed by elevation of the tibial insertion. *J Orthop Res* 14:157-166, 1996
30. Lewis JL, Lew WD, Schmidt J: A note on the application and evaluation of the buckle transducer for the knee ligament force measurement. *J Biomech Eng* 104:125-128, 1982
31. Lewis JL, Lew WD, Hill JA, Ohland KJ: Knee joint motion and ligament forces before and after ACL reconstruction. *J Biomech Eng* 111:97-106, 1989
32. Livesay GA, Fujie H, Kashiwaguchi S, Morrow DA, Fu FH, Woo SL-Y: Determination of the in situ forces and force distribution within the human anterior cruciate ligament. *Ann Biomed Eng* 23:467-474, 1995
33. Livesay GA, Rudy TW, Woo SL-Y, Runco TJ, Sakane M, Li G, Fu FH: Evaluation of the effect of joint constraints on the in situ force distribution in the anterior cruciate ligament. *J Orthop Res* 15:278-284, 1997
34. Maeda A, Shino K, Horibe S, Nakata K: Anterior cruciate ligament reconstruction with multistranded autogenous semitendinosus tendon. *Am J Sports Med* 24:504-509, 1996
35. Markolf KL, Mensch JS, Amstutz HC: Stiffness and laxity of the knee — the contributions of the supporting structures: a quantitative in vitro study. *J Bone Joint Surg [Am]* 58:583-594, 1976
36. Markolf KL, Graff-Radford A, Amstutz HC: In-vivo knee stability: a quantitative assessment using an instrumented clinical testing apparatus. *J Bone Joint Surg [Am]* 60:664-674, 1978
37. Markolf KL, Gorek JF, Kabo JM, Shapiro MS: Direct measurement of resultant forces in the anterior cruciate ligament: an in vitro study performed with a new experimental technique. *J Bone Joint Surg [Am]* 72:557-567, 1990
38. Markolf KL, Burchfield DM, Shapiro MM, Shepard MF, Finerman GAM, Slauterbeck JL: Combined knee loading states that generate high anterior cruciate ligament forces. *J Orthop Res* 13:930-935, 1995
39. Miyasaka KC, Daniel DM, Stone ML, Hirshman P: The incidence of knee ligament injuries in the general population. *Am J Knee Surg* 4:3-8, 1991
40. Norwood LA, Cross MJ: Anterior cruciate ligament: functional anatomy of its bundles in rotatory instabilities. *Am J Sports Med* 7:23-26, 1979
41. Noyes FR, Barber-Westin SD: Reconstruction of the anterior cruciate ligament with human allograft: comparison of early and later results. *J Bone Joint Surg [Am]* 78:524-537, 1996
42. O'Brien SJ, Warren RF, Wickiewicz TL, Rawlins BA, Allen AA, Panariello R, Kelly AM: The iliotibial band lateral sling procedure and its effect on the results of anterior cruciate ligament reconstruction. *Am J Sports Med* 19:21-25, 1991
43. Odensten M, Gillquist J: Functional anatomy of the anterior cruciate ligament and a rationale for reconstruction. *J Bone Joint Surg [Am]* 67:257-262, 1985
44. Radford WJP, Amis AA: Biomechanics of a double prosthetic ligament in the anterior cruciate deficient knee. *J Bone Joint Surg [Br]* 72:1038-1043, 1990
45. Rudy TW, Livesay GA, Woo SL-Y, Fu FH: A combined robotics/universal force sensor approach to determine in-situ forces of knee ligaments. *J Biomech* 29:1357-1360, 1996
46. *SAS Institute I, SAS/STAT User's Guide*, version 6, 4th ed, vol 2. Cary, North Carolina, SAS Institute, 1989
47. Shelbourne KD, Klootwyk TE, Wilckens JH, De Carlo MS: Ligament stability two to six years after anterior cruciate ligament reconstruction with autogenous patellar tendon graft and participation in accelerated rehabilitation. *Am J Sports Med* 23:575-579, 1995
48. Sidles JA, Larson RV, Garbini JL, Downey DJ, Matsen FA III: Ligament length relationships in the moving knee. *J Orthop Res* 6:593-610, 1988

49. Takai S, Woo SL-Y, Livesay GA, Adams DJ, Fu FH: Determination of the in situ loads on the human anterior cruciate ligament. *J Orthop Res* 11:686-695, 1993
50. Takai S, Watanabe K, Aso S, Hirasawa T: Early clinical results of anatomic anterior cruciate ligament double-bundle reconstruction [Japanese]. *J Jpn Orthop Assoc* 69:S361, 1995
51. Tohyama H, Beynnon BD, Johnson RJ, Renström PA: The effects of anterior cruciate ligament graft elongation at the time of implantation on the biomechanical behavior of the graft and knee. *Am J Sports Med* 24:608-614, 1996
52. Vahey JW, Draganich LF: Tensions in the anterior and posterior cruciate ligaments of the knee during passive loading: predicting the ligament loads from in situ measurements. *J Orthop Res* 9:529-538, 1991
53. Woo SL-Y, Hollis JM, Adams DJ, Lyon RM, Takai S: Tensile properties of the human femur-anterior cruciate ligament-tibia complex: the effects of specimen age and orientation. *Am J Sports Med* 19:217-225, 1991
54. Xerogeanes JW, Takeda Y, Livesay GA, Ishibashi Y, Kim HS, Fu FH, Woo SL-Y: Effect of knee flexion on the in situ force distribution in the human anterior cruciate ligament. *Knee Surg Sports Traumatol Arthroscopy* 3:9-13, 1995