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Effects of Single-Bundle and Double-Bundle ACL Reconstruction on Tibiofemoral Compressive Stresses and Joint Kinematics During Simulated Squatting

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Abstract

The purpose of this study was to compare tibiofemoral (TF) kinematics and TF compressive stresses between single bundle- (SB-) and double bundle-ACL reconstruction (DB-ACLR) during simulated squatting. Twelve matched pairs of fresh frozen cadaver knees were utilized. A simulated squat through 100° of knee flexion was performed in the ACL-intact joint. The ACL was transected and SB- and DB-ACLR procedures were performed in one knee of each pair. The squat was repeated. Knee kinematics were measured using a motion tracking system and the TF compressive forces were measured using thin film pressure sensors. The posterior shifts of the

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CONFLICT OF INTEREST

Mary K. Mulcahey, MD, Keith O. Monchik, MD, Charles Yongpravat, MS, Gary J. Badger, MS, Michael J. Hulstyn, MD, Paul D. Fadale, MD, and Braden D. Fleming, PhD have no conflicts of interest related to this study to disclose.

tibia for SB- and DB-ACLR knees were significantly greater than the ACL-intact condition for knee flexion angles 0° to 40° ($p < .05$). However, there was no difference between the SB- and DB-ACLR knees at any flexion angle (0° to 100°; $p = .37$). SB- and DB-ACLR knees had greater IE rotation than intact knees from 90° through 50° of flexion ($p < .05$), but not between 40° and full extension. There was no difference between SB- and DB-ACLR knees ($p = .68$). The TF compressive stresses of the DB-ACLR were significantly lower than intact for all angles except 10° ($p = .06$), whereas SB-ACLR knees did not differ from intact at flexion angles between 30° and 50° ($p > .32$). There were no significant differences between the two reconstruction conditions ($p = .74$). This study showed that there was no difference in the TF kinematics or compressive stresses between SB- and DB-ACLR, and only minor differences when compared to the intact state.

Keywords

Knee; ACL; Reconstruction; Kinematics; Contact Stress

Introduction

Traditionally, operative treatment for ACL tears has consisted of single-bundle ACL reconstruction (SB-ACLR); however, failure rates have ranged from 11% to 30% as evidenced by persistent instability, especially in internal-external (IE) rotation¹. Over the past several years, interest has shifted toward double bundle ACL reconstruction (DB-ACLR) with the goal of anatomically reconstructing the native ACL to better restore normal kinematics. “Anatomic” reconstruction has been defined by placement of the tunnels in the native footprints of the ACL, which are delineated by the soft tissue remnants of the ACL and bony landmarks of the medial wall of the lateral femoral condyle². It is thought that anatomic reconstruction may restore normal knee kinematics and improve long-term outcomes, including decreased instability and degenerative arthritis³. The goal of DB- and SB-ACLR is to restore anteroposterior (AP) and rotational joint stability, joint contact stresses, and to improve overall knee function. It is estimated that 11% to 100% of ACL reconstructed patients develop osteoarthritis⁴. Although the mechanism behind post-operative osteoarthritis in the ACL injured patient is unclear, it is hypothesized that abnormal joint contact stresses due to altered knee kinematics may be a significant contributing factor^{5–9}. It remains controversial whether either of these two ACLR procedures can fully restore joint motion and contact mechanics.

The ACL consists of two functional bundles, anteromedial (AM) and posterolateral (PL) as defined by their position of insertion on the tibia. The AM bundle contributes primarily to AP stability, while the PL bundle functions mainly to control IE rotational stability². Previous studies described a reciprocal relationship between the AM and PL bundles during passive knee motion, with the PL bundle being tight in extension and the AM bundle being tight in flexion^{10–13}. Recent studies, however, reveal a complementary function of the AM and PL bundles in weight bearing flexion, especially at low flexion angles^{14–15}. Both bundles reach maximum length near full extension and shorten with flexion, indicating that the two bundles may function differently during physiologic loading as compared to passive loading¹⁶.

A review of the literature suggests that DB-ACLR restores both bundles of the ACL within their native locations, but no consensus has been reached on the advantage of DB-ACLR compared to SB-ACLR. Biomechanical studies have shown that SB-ACLR restores passive AP stability, but fails to correct passive rotational stability¹⁷. Replacement of the PL bundle with DB-ACLR has been shown to more accurately restore passive rotational stability¹⁸. Some clinical studies indicate improved results with DB-ACLR compared to SB-ACLR in

regards to passive anterior laxity and pivot-shift exam^{19–24}. Other studies reveal equivalent results^{25–26}. To date, no clinical studies demonstrate improved patient-oriented outcomes following DB-ACLR relative to SB-ACLR^{19–26}.

Direct comparisons between SB- and DB-ACLR studies are difficult, given the variable surgical and experimental methods employed^{2, 27}. Although DB-ACLR seems to more accurately restore passive knee kinematics, a tibiofemoral (TF) compressive load and/or muscle forces are not included during the assessment which could influence kinematics. These loads are present in most activities of daily living and they affect ACL loading^{28–31}. The objective of the current study was to establish a baseline comparison between SB- and DB-ACL reconstructive techniques on joint kinematics and TF compressive stresses after graft fixation during simulated squatting using a cadaver model. We hypothesized that knee kinematics (AP translation, IE rotation) and the TF compressive stresses of the DB-ACLR were closer to those of the ACL-intact knee as compared to SB-ACLR.

Methods

Specimens

Right and left knees with no evidence of pathology were obtained from twelve fresh frozen human cadavers [9 female; 3 male; mean age = 58 (range 16 – 80)]. The specimens were thoroughly evaluated. Any specimens with visual signs of meniscal pathology were excluded. The specimens were stored at -20°C following harvest and consisted of the distal third of the femur and the proximal two thirds of the tibia. Before testing, each knee was thawed to room temperature. The soft tissues were dissected down to the knee capsule, quadriceps tendon, and semimembranosus and biceps femoris tendons. Threaded rods (0.5-inch diameter Stainless Steel, McMaster-Carr, Los Angeles, CA) were cemented into the intramedullary canals of the femur and tibia with an epoxy resin (Smooth-Cast 300, Smooth-On Inc, Easton, PA) to restore the approximate lengths of the tibia and femur. The rods were then attached to a loading fixture that was designed to simulate squatting activity^{32–34}. Additionally, the quadriceps tendon and semimembranosus and biceps femoris tendons were secured via cables to the loading fixture to allow attachment to the test fixture for subsequent loading.

Active Knee Loading Fixture

The test fixture was previously designed at the University of Vermont to simulate a dynamic squatting activity in cadaver knee specimens (Figure 1)^{32–34}. For this study, we defined the squat as knee flexion through 100° ³⁵. The squatting motion was selected for study because it has been shown to produce high strains on the ACL³⁶. Using this test fixture, it is possible to apply external compressive loads to the knee joint (to simulate weight bearing), and forces to the hamstrings, and quadriceps. The rods in the femur and tibia were supported by universal joints, representing the hip and ankle joints. Both connecting points provided all rotational degrees of freedom. The femoral attachment was stationary, while the tibial attachment translated vertically along the rails of the loading fixture to allow the knee to course through flexion-extension. Knee extension was induced by a servomotor that loaded the quadriceps tendon, while a static 70 N TF compressive load and a 22 N hamstring force were applied. Friction along the rails was reduced through air bearings attached to a 70 PSI source. We did not measure the quadriceps forces, however, the quadriceps forces produced following the same loading protocol have been previously measured with a peak force of approximately 800 N occurring with the knee at 100° of flexion³³.

Joint position measurements

An Optotrak Image Analysis System (Northern Digital; Waterloo, Ontario) with Optotrak Digitizer software (NDI 6D Architect) was used to track the position of the tibia with respect to the femur in 6 degrees of freedom. For both the femur and tibia, three infrared light-emitting diodes (IREDs) were used to define the rigid bodies. The IREDs were securely attached with bolt screws rigidly fixed to a metal plate – on the backside of the plate, two threaded rods extruded out and were press fitted into two corresponding holes drilled into the femur and tibia and cross pinned for rigid attachment. The Optotrak digitizer was then used to establish an anatomically-based coordinate system as described by Grood³⁷. The coordinate system was defined with the knee at full extension, so the femoral and tibial origins coincided. The axis of flexion was defined as running through the epicondyles. A straight line was then made connecting the epicondyles. Varus-valgus was then defined as an axis normal to the axis of flexion and passed through a point that intersected the midpoint of the epicondyle line. Internal-external rotation was then calculated as an axis normal to the plane made by the flexion and varus-valgus axes.

Tibiofemoral pressure sensors

Thin-film pressure sensors (K-scan 4000; Maximum Pressure Range: 10,342 KPa, Sensing Area: 27.9 mm × 33 mm, Sensels per square cm: 62; Tekscan Inc, South Boston, MA) were used to simultaneously measure the TF compressive forces in both compartments of the knee (Figure 2). For insertion, the knee capsule was opened, and each sensor was placed into either the lateral or medial compartment underneath the meniscus. The meniscus coronary ligament was partially detached to slide the sensor into place and the sensors were secured by placing sutures through the non-sensing regions and fixing them to surrounding tissue³⁸. The “I-scan” software program (Tekscan, Inc., South Boston, MA) was used to gather sensor data. Calibration for each sensor was performed separately in a servo-hydraulic material tester where known loads could be applied. We followed the manufacturer’s recommendation for calibrating the sensors, which assumes that the output is linear. The sensors were secured to a rigid wood block and a small foam board was placed on top of the sensor. The sensor was preconditioned 5 times to 380N then a tare load of 0N was taken followed by static loads of 45N and 380N. The “I-scan” software then performed a two-point calibration to convert the RAW data measurements into known measurement units. The calibration was then verified by applying known loads to the sensor and matching it to the Tekscan readout. It should be noted, care was taken to ensure all compressive loads were applied to the Tekscan sensing area. For the squat testing, each flexion-extension cycle took approximately 20 seconds. The data capture rate of the tekscan was 10 Hz and the data during the extension portion of the cycle was analyzed. The peak contact forces were measured in the medial and lateral compartments, divided by the contact area in each compartment, and then added together to give the total TF contact stress.

Test Protocol

After establishing the anatomic coordinate systems and inserting the tekscan sensors within the knee, each knee was preconditioned with five flexion-extension cycles. The Optotrak was then used to record the knee joint rotations and translations while the Tekscan simultaneously measured joint compressive forces during each flexion-extension cycle from 0° to 100° and back to 0° with a period of one minute. For every experiment, each knee would undergo a total of 15 flexion-extension cycles. There would be at least three cycles for preconditioning and then the actual recorded run. Control data was first gathered by measuring the dynamic kinematics and compressive forces of an intact knee. The ACL was then transected and a SB-ACLR or DB-ACLR was performed (randomized per knee pair); after which the joint kinematics and compressive force measurements were repeated.

ACL Reconstruction

Matched pairs were used to compare SB- and DB-ACLR. Within each pair, a left or right knee was randomly selected to undergo SB-ACLR, while the other knee was randomized to DB-ACLR. All surgeries were performed by one of the authors (KM) who was a chief resident at the time. He learned the procedure under the guidance of two sports medicine surgeons (PDF, MJH) with more than 20 years of experience with ACL reconstruction.

The SB-ACLR was performed by opening the capsule (medial parapatellar incision) and directly visualizing the femoral over-the-top position. With the knee in hyperflexion, a 2.4-mm guide wire and an Arthrex Transtibial Femoral ACL Drill Guide (AR-1801, Arthrex, Naples, FL), placed in the center of the anatomic ACL foot print, were used to drill the femoral tunnel. No femoral offset guide was used to drill the femoral tunnel, as the native ACL footprint was clearly visible. A 4.5-mm cannulated Endobutton drill bit was advanced over the guide wire and through the outer cortex. A standardized 10-mm tunnel was drilled in the femur. An Arthrex tip aimer (Arthrex, Naples, FL) angled at 45° was used to standardize anatomical tibial graft tunnel placement. The guide was placed in the center of the tibial footprint and a standardized 10-mm tunnel was drilled in the tibia. Two doubled-over semitendinosus grafts were placed within the single tibial and femoral bone tunnels. Once the graft was passed through the joint, the femoral end of the graft was secured via an Endobutton CL 20-mm (7207323, Smith & Nephew, Andover, MA). The tibial side of the graft was positioned within the tibial tunnel, firmly tensioned at 45°, and fixed over a 3.5 mm AO Screw as a suture-post fixation.

The DB-ACLR was also performed with the capsule opened (medial parapatellar incision). The anatomic insertions of the AM and PL bundles of the ACL on the lateral wall of the intercondylar notch were identified and left in place for accurate localization of the two femoral tunnels. With the knee flexed 100°–120°, a 2.4-mm guide wire and a 5-mm Arthrex offset endofemoral aimer were placed at the center of the femoral AM bundle insertion and used to drill the femoral tunnel for the AM bundle. A 4.5-mm cannulated Endobutton drill bit was then advanced over the guide wire and through the outer cortex. A standardized 8-mm tunnel was drilled in the femur. With the knee flexed to 120°, a 2.4-mm guide wire was drilled through the center of the femoral PL bundle insertion site, 2- to 3-mm from the cartilage surface ensuring an adequate bone bridge between the AM and PL tunnels. A 4.5-mm cannulated Endobutton drill bit was then advanced over the guide wire and through the outer cortex. A standardized 7-mm tunnel was drilled in the femur. The PL tibial tunnel was then drilled with the tibial drill guide set at 55°. The PL tunnel has a more medial and distal entry point on the tibial cortex than a standard ACL tibial tunnel. Once acceptable placement of the PL tibial pin was obtained, a 7-mm tibial tunnel was drilled over the guide pin. The AM tibial tunnel was then drilled in a similar fashion. The AM tibial tunnel in the anatomic reconstruction technique is more anteriorly located than in traditional SB-ACLR reconstruction. Two semitendinosus tendons were used for grafts. Each tendon was doubled-over and placed on an Endobutton CL 20-mm. The PL graft was passed first and the Endobutton loop was flipped in the usual manner. The AM graft was then passed in a similar fashion. The PL bundle was firmly tensioned first at 5°. The AM bundle was then firmly tensioned at 45°. Both bundles were fixed to the tibia over a suture-post fixation.

Data/Statistical Analysis

Each knee within a matched pair was randomly assigned to one of the two experimental procedures (SB- or DB-ACLR). Measurements were made in the ACL intact joint and again after the experimental procedure at each of 11 angles (0°, 10°, 20°, 30°, 40°, 50°, 60°, 70°, 80°, 90°, and 100°). For each outcome measure (AP translation, IE rotation, and TF contact stress), which were expressed as the change from the intact state, a two-factor repeated

measures ANOVA was used to test for differences between the two surgical procedures. Within-subject factors were surgical procedure (SB- vs DB-ACLR), knee flexion angle (0° to 100°), and their interaction. One sample t-tests, which utilized the error term derived from the ANOVA, were used to evaluate the significance of observed changes from the intact state for each outcome. These analyses were performed within angle and both within and across experimental procedures. All analyses of variance were performed using PROC MIXED, SAS statistical software Version 9 (SAS Institute, Cary NC). Statistical significance was determined based on $\alpha = 0.05$. Power calculations were performed to determine the detectable difference between the two procedures for each of our primary outcome measures. Power is estimated to be 80% to detect differences of 25%, 40% and 16% corresponding to AP translation, IE rotation and total joint compressive stress, respectively. These percentages represent the mean difference between procedures being expressed relative to the mean of intact condition.

Results

Kinematics

The differences between the experimental and ACL-intact conditions for AP translation of the tibia relative to the femur were dependent on the knee flexion angle ($p=.001$, Figure 3). The posterior shift of the tibia of both SB- and DB-ACLR knees were significantly greater than the ACL-intact condition for knee flexion angles of 0° to 40° ($p < .05$), but did not differ from ACL-intact at other angles tested. There was no evidence that these differences varied between SB- and DB-ACLR knees ($p=.37$).

The differences in IE rotation between experimental and ACL-intact knees were dependent on flexion angle ($p=.001$, Figure 4). When compared to the ACL-intact state, SB- and DB-ACLR knees had greater IE rotation than intact knees at flexion angles 50° through 90° ($p < .05$), but did not differ from ACL-intact between 40° and full extension. There were no differences between SB- and DB-ACLR knees ($p=.68$).

Compressive Stresses

For total TF compressive stress, values decreased from full flexion to full extension. This decrease was observed in both intact and experimental conditions with no evidence that the differences between experimental and intact conditions were dependent on knee flexion angle ($p=0.99$; Figure 5). When compared to the ACL-intact condition, the compressive stresses of the DB-ACLR were significantly lower than intact for all angles except 10° ($p=.06$), whereas SB-ACLR knees did not differ from intact knees at flexion angles 30° ($p=.32$), 40° ($p=.44$) or 50° ($p=.54$). There were no significant differences in the total compressive stress between the two experimental conditions ($p=.74$). The values of the tibiofemoral stresses in the lateral and medial compartments were also established. The values for both compartments decreased from full flexion to full extension (Figures 6 & 7). These decreases were observed in both intact and experimental conditions with no evidence that the differences between experimental and intact conditions were dependent on knee flexion angle (lateral compartment $p=0.92$; medial compartment $p=0.25$). When compared to the ACL-intact condition, the compressive stresses in the lateral and medial compartments of the DB-ACLR and SB-ACLR followed similar patterns to that of the total joint compressive stress, and there were no significant differences in either compartment between the two experimental conditions (lateral compartment $p=.53$; medial compartment $p=.78$).

Discussion

The joint kinematics (AP translation, IE rotation) of the SB- and DB-ACLR procedures were similar during squatting. For AP translation there was a similar posterior shift of the tibia relative to the femur as the knee was extended, which was significantly different from that of the ACL-intact condition from 40° of flexion to full extension. Likewise, both ACL reconstructed knees were slightly more externally rotated than the ACL-intact condition when the knee was extended from 90° to 50° of knee flexion. However, the mean differences were relatively small (<1°) and most likely clinically irrelevant. The magnitude of the total joint TF compressive stresses between the two conditions were not significantly different from each other, but were significantly less than that of the ACL-intact condition. It was interesting to note that the DB-ACLR failed to restore normal joint TF contact stresses for virtually all flexion angles while SB-ACLR succeeded for the intermediate angles (40° to 60°), but not near the extension or flexion limits. For all TF compression measurements, reconstructed knees were observed to have less TF compressive stress on average than the intact knee (Figure 5). This data implies that both the SB- and DB-ACLR procedures would produce comparable kinematics and TF contact stresses immediately after surgery during this activity.

Previous clinical^{21–22, 39–40} and biomechanical studies^{18, 41–42} have examined the kinematics of the DB-ACLR by applying external forces and moments. It was found that while the DB-ACLR method resisted a translational force as well as the SB-ACLR, it was superior in resisting an applied internal tibial torque and valgus torque under passive loading conditions. Instead of manually applying passive external forces to the joint, the present study evaluated the procedures during a squatting motion, which incorporates simulated muscle forces and TF compressive forces across the joint, both factors that are important in providing joint stability during dynamic activities^{28–31}. We wanted to examine this specific scenario as opposed to utilizing fixed knee flexion angles because squatting is muscle driven. Unfortunately, determination of shear forces/stresses was not included as part of this study but may be an interesting aspect for future study. Nonetheless, the comparisons were made between techniques within each specimen pair on the same fixture so that the shear forces produced during the simulated activity should be similar. Our findings suggest that the SB- and DB-ACLR grafts were similar under these conditions.

A few studies have performed a detailed examination of the distribution of tension in the AM and PL grafts and showed that the distribution changes depending on knee flexion angle, and moreover, that the variable distribution of tension recreated the distribution of force of an intact ACL better than the SB-ACLR^{42–44}. If DB-ACLR produces an optimal distribution of tension on the graft, that should also lead to a more normal distribution of force between the lateral and medial compartments. For the present study, we firmly tensioned the SB-ACLR grafts with the knee at 45° of flexion, and we firmly tensioned the PL and AM bundles with the knee at 5° and 45°, respectively for the DB-ACLR. Under these tensioning conditions, we measured the TF compressive stress in each compartment (Figures 6&7) and found that there were no significant differences between the SB- and DB-ACLR conditions for either compartment.

In the present study, we visually placed the tunnels within the footprints of the intact ACL since we performed the procedures with the joint capsule opened. One of the proposed advantages of DB-ACLR involves reconstruction of the PL bundle, which is important for knee stability during low flexion angles. Clinical studies support this claim, reporting that under an applied external force, AP laxity is the same between SB- and DB-ACLR, while IE is more accurately restored with the DB technique^{39–40}. Several authors emphasize the importance of anatomic reconstruction of the two bundles of the ACL in order to restore

normal knee kinematics. A study by Zantop et al.⁴⁵ found that a PL bundle placed in its anatomical attachment to the femur is capable of restoring normal knee kinematics under both anterior tibial and rotatory loads. A non-anatomic DB-ACLR, on the other hand, had significantly more rotational instability, indicated by pivot shift test, compared to intact and anatomic DB-ACLR. In 2006, Yasuda et al.⁴⁰ performed a prospective comparative study to examine the outcomes following anatomic DB-ACLR versus SB- and non-anatomic DB-ACLR. Patients who underwent anatomic DB-ACLR were found to have significantly less anterior tibial translation and significantly less pivot shift compared to those who underwent SB-ACLR. The results following non-anatomic DB-ACLR were inferior to those obtained following SB-ACLR⁴⁰. Anatomic DB-ACLR has been shown to reduce the pivot shift in a greater number of patients^{20, 40}. To date, no clinical studies have demonstrated improved patient-oriented outcomes following DB-ACLR relative to SB-ACLR^{19–26}.

Addition of the PL bundle is thought to create more normal knee kinematics, which in turn should better restore TF contact stresses^{39–40}. The improvements observed in the present study between the two reconstruction conditions in the TF contact stresses were small and likely clinically insignificant for this activity. It is strongly believed that altered knee kinematics, and hence altered joint contact conditions, predispose the ACL injured knee to OA, however, there are limited data in the literature on joint contact stresses following DB-ACLR. During a direct axial load applied to the TF joint, Morimoto et al.⁴⁶ found that the TF contact areas were smaller and that the contact pressures were higher following SB-ACLR when compared to DB-ACLR at low flexion angles. It should be noted that these differences were also relatively small. These data are not supported in the present study, which utilizes a loading condition that simulates an activity of daily living in which the joint compressive load and muscle loads are simulated.

There is no general consensus on the optimal angle of knee flexion for DB-ACLR graft fixation. A recent literature search suggested that the PL bundle should be fixed at or near full extension to avoid overloading the graft. Fixation of the AM bundle has varied from 90° to 0° of knee flexion¹⁶. Some studies suggest tensioning the AM bundle between 45° and 90° of flexion and the PL near full extension, with the theory being to tension the bundles at the degree of knee flexion at which they function^{3, 20, 23, 25, 47–48}. Other studies suggest tensioning both bundles between 15° and 20°, thereby avoiding excessive stress in either bundle, which may produce stress shielding in the other^{3, 14, 21–22, 49–52}. The protocol for bundle tensioning may have a significant effect on the bundle loading patterns in DB-ACLR, which in turn could influence TF compressive stresses³⁸. Studies have also found both SB- and DB-ACLR to have significantly higher graft loads when the knee is placed under an externally applied force than that of the intact ACL^{18, 22, 38, 41, 43, 53–55}, which would suggest differences in TF compressive stresses.

Previous studies have shown that lowering graft tension reduces joint contact forces^{38, 55}. A study by Miura et al.⁴⁷ compared the force distribution of the anteromedial and posterolateral grafts to those of the respective bundles of the intact ACL. They determined that the angle at which the grafts are tensioned significantly influences the force distribution within the AM and PL grafts. The study concluded that the AM bundle should be fixed between 60° and 30° of knee flexion while the PL bundle should be fixed between 30° of knee flexion and full extension in order to optimize the force distribution within the two grafts. The inference is that joint contact stresses will be lower in the setting of more ideal force distribution within the AM and PL grafts. The angles of fixation for this experiment were 5° for the PL bundle and 45° for the AM bundle, which were close to those used by Miura. Therefore, it is possible that this combination resulted in a lower graft tension and joint contact stresses over the range of motion. These findings may be important since variable bundle tension patterns

may influence the distribution of forces to the menisci and/or articular cartilage, affecting TF contact stresses, and hence contributing to the development of degenerative arthritis.³

The advantage of the *ex-vivo* squatting fixture used in this study is that it enabled us to evaluate the surgical procedures during simulation of a common weightbearing activity of daily living, and to directly measure the TF compressive stresses, which is currently not possible *in vivo*. The general trends observed with the squatting jig are similar to the data gathered from other studies measuring the kinematics of the squatting model. Using the same system, Coughlin et al.³⁴ tested a method of measuring patellar position and found the knee rotated internally from full flexion to 60°, remained constant until 30°, and then underwent significant external rotation until full extension. Churchill et al.³³ also used the same squatting jig and protocol to evaluate the kinematics of total knee arthroplasty and observed that the load on the quadriceps tendon required to extend the knee was highest at 80° to 90° of flexion and decreased quickly through the range of motion. Since the quadriceps tendon compresses the TF joint, the tension on the tendon should be proportional to the compressive stress in the knee joint. Therefore, Churchill et al also observed that joint compressive stress during extension was highest at full flexion and decreased throughout extension³³, as we found in this study (Figure 5).

There are limitations to this study. Post-surgical healing effects such as graft healing, tunnel expansion, biochemical pathways, and changes in graft tension over time could not be assessed in this cadaver study. Additionally, the muscle function of the knee was greatly simplified in this study because the hamstrings and quadriceps were the only muscles considered and because the body weight bearing down on the joint was under-represented with a compressive force of 70N. However, this compressive load was selected since it was enough to engage the tibiofemoral articulation^{32, 56}, without challenging the physical constraints of the ex vivo specimen and loading fixture interface. In this study, the condyle sizes of each of the donor knees were not measured nor were the sizes of the tibial plateaus or the area not protected by the meniscus. Variations in each of these measurements may influence the TF compressive stresses and it may be interesting to examine this in future studies.

Additional limitations exist regarding measurement of TF compressive forces. First, the TF compressive forces were directly measured using thin-film pressure sensor technology (thickness of 0.1mm). Although the presence of a 0.1 mm thick sensor may affect kinematics by changing knee geometry, many believe this effect is negligible. Second, it was necessary to partially detach the anterior portion of the meniscus coronary ligament to slide the sensors into place. This could influence meniscal function and affect the contact stresses. However, we do not feel this is a major limitation since we have previously shown that the implantation of the sensors did not affect knee laxity⁵⁵. Furthermore, Tekscan pressure sensors have been validated and used for other biomechanical studies examining joint contact forces^{38, 57–59}. We feel that any errors that were produced by introducing the sensor were controlled by the experimental design where the comparisons between the ACL intact and ACL reconstructed conditions were performed within each knee, and the comparisons between the two reconstructed conditions were made within each donor pair. It should also be noted that the sensors were inserted prior to any testing within a joint and that the same sensor remained in the joint during the reconstruction procedures. We felt that removing the initial sensors and placing new sensors may introduce additional error. Care was taken to prevent damage to the sensor throughout the reconstruction surgery and the test protocol. The outputs were also checked to ensure that they remained consistent both before and after testing in each specimen.

We followed the manufacturer's recommendation for calibrating the sensors. It assumes that the output is linear, as we found in one of our previous studies³⁸ where we used the same sensors to measure the TF compressive forces to compare different graft tensioning protocols. A study by Bachus et al⁵⁷ demonstrated that the Tekscan sensor output was linear through 4000 N of applied force and that the sensor reading was within 4% of the true pressure at the high loads.

In this study, the absolute translation of the tibia over the range of flexion-extension motion was on average around 15mm, which is higher than would be expected. This may stem from the assumption that the whole knee joint revolves around the axis established by the epicondyles. The coordinate system dictates the magnitude of AP translation which was consistent across all the knees tested in this study. However, the absolute translation is not as critical as the difference between the translations of the intact, SB-, and DB-ACLRs, which still allow us to accurately assess the difference in joint translations between treatment types.

Surgical techniques including type of fixation, graft twisting, and tunnel placement can significantly impact joint kinematics. For this experiment, a closed loop endobutton was used on the femoral side, while a metal interference screw acting as a rigid aperture fixation was used on the tibial side. It is highly possible that alternate ACL-reconstructive choices could influence data results. The purpose of this experiment, however, was to establish a baseline relationship between SB- and DB-ACLR. Future studies should investigate specific surgical parameters to shift the joint kinematics of the DB-ACLR technique closer to the normal knee.

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Figure 1.
Schematic of test fixture for simulated squatting motion. (adapted from Churchill et al²⁷).

Figure 2.
Sensor Map K-Scan 4000. [<http://www.tekscan.com/medical/catalog/4000.html>]

Figure 3. AP translation versus knee flexion angle. Significant differences were detected, albeit small, between ACL intact and experimental knees at angles 0° through 40° independent of treatment condition. There was no evidence of difference between the two procedures.

Figure 4.

IE rotation versus knee flexion angle. External rotation is positive while internal rotation is negative. Significant differences were detected between ACL-intact and both experimental ACLR knees at angles 50° through 90° independent of experimental condition. There was no evidence of difference between the two procedures.

Figure 5.
Average total compressive stress versus knee flexion angle.

Figure 6.
Average TF compressive stress of the medial compartment versus knee flexion angle.

Figure 7.
Average TF compressive stress of the lateral compartment versus knee flexion angle.