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Winner of the 2010 ESSKA Best Paper in Ligaments and Biomechanics

Kinematic Impact of Anteromedial and Posterolateral Bundle Graft Fixation Angles on Double-Bundle Anterior Cruciate Ligament Reconstructions

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Background: Currently in double-bundle anterior cruciate ligament (ACL) reconstructions, the range of knee flexion angles that surgeons use for anteromedial (AM) and posterolateral (PL) bundle graft fixation spans from 0° to 90° for both bundle grafts. Despite the recent popularity of this procedure, no consensus exists on an optimal set of AM and PL graft fixation angles.

Hypothesis: Graft fixation angles that simulate the native tensioning relationship of the AM and PL bundles will produce kinematic results similar to the intact knee, while graft fixation angles that do not simulate this relationship will under- or overconstrain the knee.

Study design: Controlled laboratory study.

Methods: Twelve cadaveric knees were biomechanically tested in the intact state, ACL-sectioned state, and a randomized order of 7 double-bundle ACL reconstructed states at multiple graft fixation angle combinations. For each test state, data were collected for 88 N anterior tibial loads, 10 N·m valgus torques, 5 N·m internal rotation torques, and 2 simulated pivot shift loads consisting of a 5 N·m internal rotation torque coupled with either a 10 N·m valgus torque or an 88 N anterior tibial load at 0°, 20°, 30°, 60°, and 90° of knee flexion.

Results: The AM and PL graft fixation angle combinations of 0°/0° (AM graft fixation angle/PL graft fixation angle), 60°/0°, 45°/15°, and 75°/15° restored normal laxity to the reconstructed knee in all of the biomechanical tests. The 30°/30°, 60°/60°, and 90°/90° graft fixation angle combinations significantly restricted knee laxity compared with the intact state in various biomechanical tests.

Conclusion: We found that as long as the PL bundle graft was fixed between 0° and 15°, the AM graft could be fixed up to 75° without restricting knee laxity. However, fixation of the PL graft at 30° of knee flexion and above significantly overconstrained the knee.

Clinical Relevance: This study provides a range of angles that can be used in double-bundle ACL reconstructions to restore normal knee stability without causing overconstraint.

Keywords: anterior cruciate ligament (ACL); double-bundle ACL reconstruction; anteromedial and posterolateral graft fixation angles; joint overconstraint

The ACL consists of 2 functional bundles named for the relative locations of their tibial insertions: the anteromedial (AM) and posterolateral (PL) bundles.[13,17,45] Previous studies have reported that the AM bundle reaches peak tension at 60° of knee flexion and remains relatively taut throughout the full range of knee motion, whereas the PL bundle is tightest with the knee in full extension and slackens as the knee flexes past 30°.[4,5,16,32,41] Stabilization against anteroposterior loads has been reported to coincide with the natural tensioning pattern of the bundles; the AM bundle bears a majority of the anterior tibial loads at higher knee flexion angles, while the PL bundle bears

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a majority of these loads near extension.[2,16,32,45] Some authors have concluded that the PL bundle is primarily responsible for rotational stability near extension, while others have reported that both bundles synergistically contribute to rotational stability.[5,8,16,41,45,47]

An anatomical double-bundle ACL reconstruction (DB-ACLR) separately reconstructs both the AM and PL bundles at their precise tibial and femoral attachments to theoretically improve knee kinematics compared with a single graft, single-bundle ACL reconstruction.[5,40,42,43] Despite the recent popularity of DB-ACLRs, no consensus exists on an optimal set of AM and PL graft fixation angles.[8,46] Currently, the reported fixation angles for DB-ACLR surgical procedures range from 0° to 90° for both bundle grafts. A large proportion of biomechanical and clinical DB-ACLR studies report fixation of the AM and PL grafts in the range of 45° to 75° and 0° to 15° for the AM and PL bundle grafts, respectively, which approximates the native peak bundle tensions of 60° for the AM bundle and 0° for the PL bundle.[§] Several biomechanical and clinical DB-ACLR studies also report fixing the AM and PL bundle grafts at the same knee flexion angle simultaneously, usually between 0° and 30° of knee flexion.[3,19,22,24,25,28,34,36,43]

Previous biomechanical studies have demonstrated that graft fixation angles affect knee kinematics after a DB-ACLR.[9,26,38] It was reported that fixation of the AM graft at 60° and the PL graft at 0° overloaded the AM graft, while fixing both the AM and PL grafts at 30° overloaded the PL graft compared with the intact bundles.[26] It has also been reported that fixing the AM graft between 15° and 45° and the PL graft at 15° produced normal tensions in the grafts.[38] An ACL bundle graft fixation protocol of 20°/20° was also reported to be superior to fixation at 90°/20° or 20°/90°.[9] To continue improving surgical techniques for double-bundle ACL reconstructions, the effect of graft fixation angles on knee kinematics must be evaluated further.

The purpose of our study was to evaluate how different AM and PL graft fixation angles affect knee kinematics in DB-ACLRs. Our hypothesis was that graft fixation angles that simulate the native tensioning relationship of the ACL bundles—that is, 60° for the AM bundle and 0° for the PL bundle—would produce kinematic results similar to the intact knee, while graft fixation angles that do not simulate this relationship would underconstrain or overconstrain the knee.

MATERIALS AND METHODS

Specimens

Twelve nonpaired, fresh-frozen human cadaveric knees with no evidence of prior injury or degenerative change and a mean age of 50.9 years (range, 29-66 years) were used in this study. The knees were stored in a −20°C freezer and thawed overnight before testing. Adequate

§ 4, 11, 12, 15, 27, 30, 31, 33, 39, 40, 44

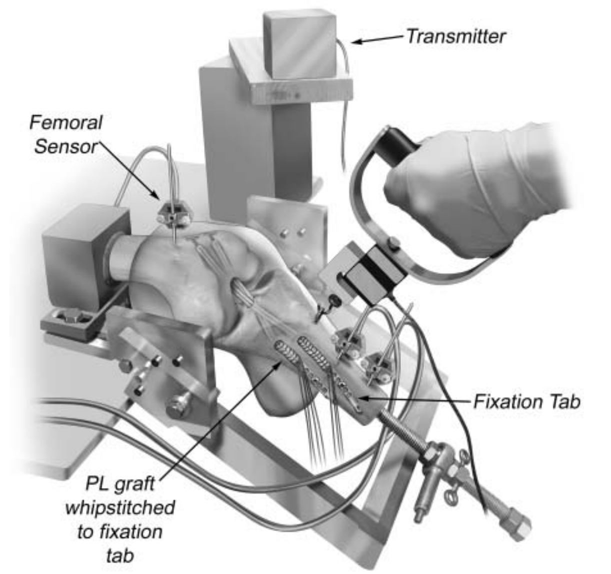

Figure 1. A left knee in the biomechanical testing apparatus during application of an anterior tibial load. The custom fixation tabs were whip-stitched to the distal ends of the grafts left protruding out of the tibial tunnels and secured to the tibia with cortical screws. Cortical buttons secured the grafts proximally to the femur. The electromagnetic transmitter, positioned above the knee, generated an electromagnetic field that the sensors received to determine 3-dimensional positioning.

amounts of soft tissues were removed from the bones to allow for mounting and testing. The femur was cut 20 cm proximal to the joint line and the tibia was sectioned 13 cm distal to the joint line. The femur was placed in a cylindrical mold filled with polymethylmethacrylate (Dentsply, York, Pennsylvania) in preparation for mounting. The intramedullary canal of the tibia was reamed and a threaded fiberglass rod was inserted and fixed with polymethylmethacrylate. A tie-rod was passed onto the fiberglass rod for the application of a valgus moment during the simulated pivot shift test. Subsequently, a custom-made hexagonal nut with attached static eye screws was threaded onto the fiberglass rod for the application of valgus forces. Two additional counter-tightened hexagonal nuts were added to the fiberglass rod for the application of rotational torques. An eye screw was placed into the center of the tibial tuberosity for application of anterior tibial loads (Figure 1).

Fresh bovine extensor tendons (Frontier BioMedical, Logan, Utah), which have been previously shown to be biomechanically similar to human hamstring tendons,[10] were used as the grafts for the ACL reconstructions. The tendons were stored in a 0.9% saline solution at −20°C. After thawing, the native lengths of the tendons, ranging from 240 to 270 mm, were not altered to ensure distal extension of the tendons out of the tibial tunnels, which was a requirement for graft fixation. The ends of the tendons were whip-stitched 25 mm from each end of the tendon

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using FiberLoop suture (Arthrex Inc., Naples, Florida). Prior to graft passage, the grafts were preconditioned with a force of 50 N for 10 minutes. Throughout the experiments, the knees and grafts were kept moist with a 0.9% saline solution.

Data Collection

The Polhemus Liberty system (Polhemus Inc, Colchester, Vermont) was used to quantify motion of the tibia relative to the stationary femur by gathering 6 degrees of freedom positional data from 1 femoral and 2 tibial sensors relative to a low-frequency magnetic field produced by a nearby electromagnetic transmitter device. A calibrated stylus was used to designate anatomical reference points to facilitate the calculation of displacement of the tibia in clinically relevant units of measurement. The MotionMonitor software package (Innovative Sports Training, Chicago, Illinois) collected data from the Polhemus system at 120 samples per second. Previous intraspecimen comparison of repeated measures in our laboratory found this system to be accurate to between 0.56° and 0.92°.[18] Each sensor was secured to the specimen with 2 threaded Kirschner wires. The tibial sensors were positioned to avoid interfering with the ACL reconstruction, the fixation of grafts, and the application of the test forces. The transmitter-to-receiver distance was maintained within the previously recorded optimal range of 22.5 to 64.0 cm to minimize positional error.[1] Data reduction was performed by a custom-written algorithm using MATLAB (The MathWorks, Natick, Massachusetts).

Biomechanical Testing

For each knee the researchers sequentially tested the intact state, performed the DB-ACLR, tested the 7 graft fixation states in a randomized order, retested the first graft fixation state for validation, and tested the sectioned state. The sectioned state was tested last to prevent the increased laxity from damaging the secondary stabilizers of the knee. In addition, it allowed for immediate reaming of the bundles after sectioning to ensure that the tunnels were exactly in the center of the bundle attachments. The 7 predetermined graft fixation states were (AM graft fixation angle/PL graft fixation angle): 0°/0°, 60°/0°, 45°/15°, 75°/15°, 30°/30°, 60°/60°, and 90°/90°. Graft fixation combinations were based on published clinical recommendations as well as discussions in published biomechanical studies. The 60°/0° fixation angle combination was selected because it reportedly simulates the native bundle tensioning relationship[16] and is used clinically. The 45°/15° and 75°/15° combinations were selected based on their approximation of the native bundle tensioning relationship, and because of their clinical use. The 0°/0° and 30°/30° fixation angle combinations represent angles often used when both the AM and PL bundle grafts are fixed at the same angle. The 60°/60° and 90°/90° combinations, although rarely used, were selected to provide information on how graft fixation at increased flexion angles affects knee kinematics.

During the testing of each state, the specimens were secured in a previously described testing apparatus that fixed the femur in the horizontal plane while allowing free movement of the tibia on an adjustable support bar that set the knee flexion angle[6,7,18,35] (Figure 1). At every testing angle for each testing state, each knee underwent an 88 N anterior tibial load, a 10 N·m valgus torque, a 5 N·m internal rotation torque, and 2 simulated "pivot shift" loads consisting of a 5 N·m internal rotation torque coupled with either a 10 N·m valgus torque[35] or an 88-N anterior tibial load at 0°, 20°, 30°, 60°, and 90° of knee flexion. A 100-N force model SM S-type load cell (Interface, Scottsdale, Arizona), with a manufacturer reported nonrepeatability error of ±0.01%, was used to apply anterior tibial loads and valgus forces. A 15 N·m capacity model TS12 shaft-style reaction torque transducer (Interface) with a manufacturer reported nonrepeatability error of ±0.02% was used to apply the internal rotation torques. Data collection was initiated with the knee in a neutral position. Then the test force was applied for several seconds, withdrawn, and repeated for a total of 3 force applications per test force per knee flexion angle and the results were averaged. For all displacement measurements, the MATLAB algorithm calculated the difference between 2 vectors, one created with the knee in neutral position and the second made during test force application. The vectors were generated using the designated anatomical reference points on the knee. To measure anterior tibial translation (ATT) during the anterior tibial loads, the difference in displacement was found between neutral and test force vectors that originated at the midpoint of the medial and lateral femoral epicondyles and extended to the midpoint of the maximal coronal diameter of the tibial plateau. In the same manner, the ATT during both of the simulated pivot shifts was measured with vectors that originated at the midpoint of the epicondylar width and extended through a point immediately lateral to the patellar tendon on the anterior aspect of the tibial plateau. To measure tibial rotation during the internal rotation torques and valgus torques, the algorithm calculated the angular difference between vectors parallel to the vectors used in the anterior tibial load tests.

Double-Bundle ACL Reconstruction

All double-bundle reconstructions were performed by the same board-certified orthopaedic surgeon (R.F.L.) following biomechanical testing in the intact state. The joint capsule was opened with a medial parapatellar miniarthrotomy and the AM and PL bundles of the ACL were isolated at 90° of knee flexion based on their near-horizontal femoral origins at this knee flexion angle. The PL bundle was excised from the femur and an eyelet pin was passed through the center of its footprint. The tunnel was made with a 7-mm cannulated reamer, leaving 4 mm of bone between the lateral femoral cortex and the end of the tunnel. Next, the AM femoral tunnel was created in the same

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manner as the PL tunnel. The 7-mm grafts, looped over 12-mm RetroButtons (Arthrex) for femoral fixation, were passed into the tunnels and secured to the lateral femoral cortex by flipping the buttons; the PL graft was passed first followed by the AM graft. For specimens with softer bone, a small nylon washer was placed between the fixation button and the femoral cortex to prevent the button from pulling through the cortex of the femur during repeated testing.

Next, the AM and PL bundles were sharply dissected away from their tibial insertions. With the tibial guide set to 55°, a drill pin was drilled from the outer surface of the tibia, 5 mm lateral to the superficial medial collateral ligament, up proximally through the anatomical PL bundle tibial footprint. The pin for the AM tunnel was then set 5 mm medial to the tibial tubercle and drilled up proximally through the center of the AM bundle tibial footprint. After verification that the pin position was accurate and would result in a 1- to 2-mm bone bridge between the 2 tibial tunnels, each tunnel was reamed with a 7-mm cannulated reamer, the PL tunnel before the AM tunnel. The distal end of each graft was passed through its respective tibial tunnel. The medial arthrotomies were sutured closed to eliminate any kinematic variability during the application of test forces.

Distal Graft Fixation

To allow multiple fixations of each graft to the tibia without damaging the graft, a custom-fabricated stainless steel metal fixation tab was used (Figure 1). The distal end of each graft, left protruding out of the tibia, was whip-stitched with multiple nonabsorbable sutures to a row of holes on the proximal end of the fixation tab. The middle segment of the tab contained 5 additional holes sized to fit 3.5-mm cortical screws that would secure the tab to the tibia for fixation. The distal end of the tab, which was upturned and rotated 90°, contained a final hole that allowed for tensioning of the graft during fixation using the 100-N force model SM-S load cell. Before testing of graft fixation states, the grafts were cyclically loaded by applying distal traction to both fixation tabs simultaneously and manually moving the knee through 20 cycles of flexion and extension through a full range of knee motion. To remain consistent between specimens and graft fixation states, the PL bundle was always fixed before the AM bundle. The custom testing apparatus was adjusted to set the knee at the appropriate angle for each graft fixation. With 20 N of force applied to the graft using the load cell and a posterior tibial force applied, a 3.5-mm cortical screw was secured bicortically in the tibia through the center of one of the holes on the fixation tab to ensure secure fixation of the graft. A total graft tension of approximately 40 N for hamstring tendons has been shown to be sufficient in restoring normal knee kinematics without causing joint overconstraint for both single-bundle and double-bundle ACL reconstructions.[20,37,39,41] Additionally, the application of a posterior tibial force has been reported to better restore knee kinematics when applied during graft fixation.[20]

Pilot Knee Testing

Pilot testing was carried out on 3 knees to determine the biomechanical testing protocol. The tests included 88 N anterior and posterior tibial forces, 5 N·m internal and external rotation torques, 10 N·m varus and valgus torques, and both of the simulated pivot shift loads at 0°, 20°, 30°, 60°, and 90° of knee flexion. Posterior tibial forces, external rotational torques, and varus torques were excluded from the biomechanical testing protocol because they produced differences in translation of less than 1 mm or differences in rotation of less than 1° of rotation across all pilot knee specimens.

Validation

To validate the graft fixation procedure and the testing protocol, the first graft fixation state was reapplied and retested after the completion of the biomechanical testing of the other graft fixation states. The validation testing included an 88 N anterior tibial load, 5 N·m internal rotation torque, and 5 N·m internal rotation torque coupled with a 10 N·m valgus torque at 0°, 30°, and 60° of knee flexion.

Statistical Analysis

Two-way analysis of variance was performed to compare each of the graft fixation states to the intact and sectioned states following a square root transform for data distribution normalization. Tukey's honest significant difference test was used for post hoc comparisons. For all analyses, statistical significance was assumed for $P < .05$.

RESULTS

All data are summarized in Table 1. Results are reported as the mean ± standard error of the mean. One AM bundle graft failure occurred during validation testing, and a combined AM and PL bundle graft failure occurred during testing of the 60°/60° fixation state.

Kinematic Response to Anterior Tibial Load

In response to an 88 N anterior tibial load, the mean ATT for the intact state was 4.6 ± 0.6 mm, 6.9 ± 1.4 mm, 7.7 ± 1.7 mm, 5.8 ± 1.0 mm, 4.7 ± 0.8 mm at 0°, 20°, 30°, 60°, and 90° of knee flexion, respectively. In the sectioned state, the ATT was significantly increased compared with the intact state at all angles of knee flexion with means of 12.3 ± 1.2 mm ($P < .0001$) at 0°, 15.1 ± 2.1 mm ($P < .0001$) at 20°, 14.1 ± 2.1 mm ($P < .0001$) at 30°, 9.6 ± 1.6 mm ($P < .0005$) at 60°, and 8.0 ± 1.4 mm ($P < .02$) at 90°. No graft fixation states demonstrated significant differences in ATT compared with the intact state.

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TABLE 1
Mean Displacements per Applied Load (Mean ± Standard Error of the Mean)[a]

Applied Load and State		Knee Flexion Angle (deg) 0	20	30	60	90
Anterior tibial translation (mm) during anterior tibial load	Intact	4.6 ± 0.6	6.9 ± 1.4	7.7 ± 1.7	5.8 ± 1.0	4.7 ± 0.8
	0°/0°	4.2 ± 0.7	7.2 ± 1.5	7.4 ± 1.5	7.5 ± 1.1	6.2 ± 0.8
	60°/0°	3.6 ± 0.8[b]	6.2 ± 1.5[b]	7.1 ± 1.7[b]	7.1 ± 1.3[b]	7.0 ± 1.2[b]
	45°/15°	3.6 ± 0.6	6.8 ± 1.4	7.0 ± 1.5	7.3 ± 1.2	7.0 ± 1.1
	75°/15°	4.0 ± 0.7	6.8 ± 1.6[b]	7.5 ± 1.6	7.6 ± 1.4	6.9 ± 1.3
	30°/30°	4.3 ± 0.5	6.4 ± 1.3	7.0 ± 1.4	8.0 ± 1.4	6.7 ± 1.2
	60°/60°	4.2 ± 0.7[b]	6.1 ± 1.3[b]	6.8 ± 1.4[b]	7.5 ± 1.2[b]	7.1 ± 1.3[b]
	90°/90°	3.9 ± 0.5	6.1 ± 1.3	6.6 ± 1.3	7.6 ± 1.4	6.6 ± 1.1
	Sectioned	**12.3 ± 1.2**	**15.1 ± 2.1**	**14.1 ± 2.1**	**9.6 ± 1.6**	**8.0 ± 1.4**
Valgus angulation (deg) during valgus torque	Intact	2.9 ± 0.4	6.3 ± 0.8	8.7 ± 1.2	12.3 ± 1.7	12.6 ± 1.5
	0°/0°	3.6 ± 0.6	6.0 ± 1.0	7.9 ± 1.4	13.0 ± 2.1	12.8 ± 1.8
	60°/0°	3.8 ± 0.8[b]	5.9 ± 0.7[b]	7.4 ± 1.2[b]	10.9 ± 2.1[b]	11.3 ± 1.5[b]
	45°/15°	3.8 ± 0.5	5.4 ± 0.5	7.1 ± 1.0	11.1 ± 1.9	11.9 ± 1.6
	75°/15°	4.0 ± 0.8	5.7 ± 0.6	6.9 ± 1.1	10.5 ± 2.0	11.3 ± 1.4
	30°/30°	4.1 ± 0.8	6.2 ± 1.0	7.0 ± 1.3	9.2 ± 1.9	9.8 ± 1.4
	60°/60°	4.7 ± 1.0[b]	5.5 ± 0.8[b]	6.4 ± 0.6[b]	**8.8 ± 1.6**[b]	**9.0 ± 1.7**[b]
	90°/90°	4.8 ± 0.9	5.4 ± 0.7	6.4 ± 0.4	9.3 ± 1.9	9.8 ± 1.5
	Sectioned	**5.8 ± 0.7**	**11.0 ± 1.9**	**16.1 ± 2.6**	**18.0 ± 2.5**	16.0 ± 1.8
Internal tibial rotation (deg) during internal tibial torque	Intact	12.5 ± 0.9	20.2 ± 1.6	22.2 ± 2.0	19.3 ± 2.4	16.4 ± 1.8
	0°/0°	12.9 ± 10	20.6 ± 1.7	22.6 ± 2.0	21.0 ± 2.2	18.7 ± 1.7
	60°/0°	11.7 ± 1.3[b]	19.7 ± 1.7[b]	21.8 ± 2.0[b]	20.8 ± 2.3[b]	18.3 ± 1.7[b]
	45°/15°	11.0 ± 1.2	19.8 ± 1.8	22.1 ± 2.1	21.2 ± 2.4	18.9 ± 1.8
	75°/15°	10.8 ± 1.1	19.6 ± 1.7	21.7 ± 1.9	21.3 ± 2.2	19.1 ± 1.7
	30°/30°	**9.3 ± 1.2**	18.8 ± 1.6	21.0 ± 1.8	20.8 ± 2.1	18.8 ± 1.7
	60°/60°	**7.9 ± 1.6**[b]	**17.5 ± 1.7**[b]	20.1 ± 1.9[b]	19.1 ± 1.9[c]	18.6 ± 1.8[b]
	90°/90°	**7.3 ± 1.3**	**17.8 ± 1.8**	**20.6 ± 1.9**	21.1 ± 2.2	19.5 ± 1.8
	Sectioned	**17.5 ± 1.1**	**23.3 ± 2.0**	**24.3 ± 2.1**	**21.5 ± 2.4**	18.1 ± 1.9
Anterior tibial translation (mm) during coupled internal rotation and valgus torques	Intact	7.2 ± 0.9	11.8 ± 0.9	12.2 ± 0.8	9.5 ± 0.9	8.6 ± 0.7
	0°/0°	7.6 ± 0.7	12.0 ± 0.9	12.4 ± 1.0	10.1 ± 1.2	8.8 ± 1.0
	60°/0°	6.7 ± 0.8[b]	10.8 ± 0.8[b]	11.2 ± 0.8[b]	9.2 ± 1.1[b]	8.1 ± 0.9[b]
	45°/15°	6.0 ± 0.6	10.7 ± 0.7	11.3 ± 0.8	9.7 ± 1.1	8.7 ± 1.0
	75°/15°	5.9 ± 0.7	10.7 ± 0.8	11.4 ± 0.8	9.9 ± 1.1	8.8 ± 1.0
	30°/30°	**5.3 ± 0.7**	9.9 ± 1.0[b]	10.7 ± 0.9	9.6 ± 1.1	8.7 ± 1.0
	60°/60°	**5.3 ± 0.8**[b]	**9.1 ± 0.8**[b]	**9.8 ± 0.8**[b]	8.9 ± 1.0[b]	8.0 ± 0.9[b]
	90°/90°	**4.9 ± 0.7**	**8.8 ± 0.8**	**9.7 ± 0.8**	9.4 ± 1.1[b]	8.6 ± 0.9
	Sectioned	**13.1 ± 1.3**	**16.1 ± 1.5**	**15.7 ± 1.4**	**11.6 ± 1.5**	9.8 ± 1.2
Anterior tibial translation (mm) during coupled internal rotation torque and anterior tibial load	Intact	9.1 ± 1.1	12.2 ± 1.0	12.5 ± 0.9	8.0 ± 1.0	6.5 ± 0.8
	0°/0°	9.5 ± 1.0	12.1 ± 1.0	12.2 ± 1.1	8.8 ± 1.3	7.1 ± 1.0
	60°/0°	8.7 ± 0.9[b]	11.1 ± 0.9[b]	11.4 ± 1.0[b]	8.0 ± 1.4[b]	6.6 ± 1.0[b]
	45°/15°	7.8 ± 0.8	11.0 ± 0.9	11.0 ± 0.9	8.2 ± 1.4	7.1 ± 1.1
	75°/15°	7.4 ± 0.9	11.3 ± 1.1[b]	11.6 ± 0.9	8.7 ± 1.3[b]	7.1 ± 1.0
	30°/30°	7.0 ± 0.9	10.4 ± 1.2	10.6 ± 1.0	8.3 ± 1.3	7.1 ± 1.0
	60°/60°	**6.4 ± 0.9**[b]	**9.3 ± 1.2**[b]	**10.0 ± 1.1**[b]	7.0 ± 1.2[b]	5.7 ± 0.7[b]
	90°/90°	**6.1 ± 0.7**	**9.2 ± 0.9**	**9.8 ± 1.0**	7.8 ± 1.3	6.8 ± 1.0
	Sectioned	**14.5 ± 1.6**	**15.9 ± 1.5**	15.0 ± 1.5	**10.1 ± 1.8**[b]	**7.9 ± 1.1**

[a]Bold indicates significant difference compared with intact state, $P < .05$.
[b]n = 11.
[c]n = 10.

Kinematic Response to Valgus Torque

In response to a 10 N·m valgus torque, the mean valgus angulation for the intact state was 2.9° ± 0.4°, 6.3° ± 0.8°, 8.7° ± 1.2°, 12.3° ± 1.7°, and 12.6° ± 1.5° at 0°, 20°, 30°, 60°, and 90° of knee flexion, respectively. In the sectioned state, valgus angulation was significantly increased compared with the intact state at 0°, 20°, 30°, and 60° of knee flexion, with means of 5.8° ± 0.7° ($P < .002$), 11.0° ± 1.9° ($P < .02$), 16.1° ± 2.6° ($P < .0001$), 18.0°± 2.5° ($P < .001$), respectively. Significant reductions in valgus angulation compared with the intact state occurred in the 60°/60° fixation state at 60° ($P < .03$) and 90° ($P < .03$) of knee flexion.

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Kinematic Response to Internal Rotation Torque

In response to the application of a 5 N·m internal rotation torque, the mean internal tibial rotation for the intact state was 12.5° ± 0.9°, 20.2° ± 1.6°, 22.2° ± 2.0°, 19.3° ± 2.4°, 16.4° ± 1.8° at 0°, 20°, 30°, 60°, and 90° of knee flexion, respectively. In the sectioned state, the internal tibial rotation was significantly increased compared with the intact state at 0°, 20°, 30°, and 60° of knee flexion with means of 17.5° ± 1.1° ($P < .005$), 23.3° ± 2.0°, ($P < .0005$), 24.3° ± 2.1° ($P < .003$), and 21.5° ± 2.4° ($P < .0005$), respectively. At 0° of knee flexion, significant reductions in internal tibial rotation compared with the intact state occurred in the 30°/30° ($P < .04$), 60°/60° ($P < .0001$), and 90°/90° ($P < .0001$) fixation states. At 20° of knee flexion, significant reductions in internal tibial rotation compared with the intact state occurred in the 60°/60° ($P < .03$) and 90°/90° ($P < .005$) fixation states. At 30° of knee flexion, significant reductions in internal tibial rotation compared with the intact state occurred in the 90°/90° ($P < .04$) fixation state.

Kinematic Response to Simulated Pivot-Shift—Coupled Internal Rotation and Valgus Torques

In response to a 5 N·m internal rotation torque coupled with a 10 N·m valgus torque, the mean ATT in the intact state was 7.2 ± 0.9 mm, 11.8 ± 0.9 mm, 12.2 ± 0.8 mm, 9.5 ± 0.9 mm, and 8.6 ± 0.7 mm at 0°, 20°, 30°, 60°, and 90° of knee flexion, respectively. In the sectioned state, ATT was significantly increased compared with the intact state at 0°, 20°, 30°, and 60° of knee flexion with means of 13.1 ± 1.3 mm ($P < .0001$), 16.1 ± 1.5 mm ($P < .0001$), 15.7 ± 1.4 mm ($P < .0001$), and 11.6 ± 1.5 mm ($P < .0001$), respectively. At 0° of knee flexion, significant reductions in ATT compared with the intact state resulted from the 30°/30° ($P < .05$), 60°/60° ($P < .04$), and 90°/90° ($P < .004$) fixation states. At 20° of knee flexion, significant reductions in ATT compared with the intact state resulted from the 60°/60° ($P < .02$) and 90°/90° ($P < .0003$) fixation states. At 30° of knee flexion, significant reductions in ATT compared with the intact state resulted from the 60°/60° ($P < .02$) and 90°/90° ($P < .0003$) states.

Kinematic Response to Simulated Pivot-Shift—Coupled Internal Rotation Torque and Anterior Tibial Load

In response to a 5 N·m internal rotation torque coupled with an 88 N anterior tibial load, the mean ATT in the intact state was 9.1 ± 1.1 mm, 12.2 ± 1.0 mm, 12.5 ± 0.9, 8.0 ± 1.0, and 6.5 ± 0.8 mm at 0°, 20°, 30°, 60°, and 90° of knee flexion, respectively. In the sectioned state, ATT was significantly increased compared with the intact state at 0°, 20°, 60°, and 90° of knee flexion, with means of 14.5 ± 1.6 mm ($P < .0001$) at 0°, 15.9 ± 1.5 mm ($P < .004$) at 20°, 10.1 ± 1.8 mm ($P < .02$) at 60°, and 7.9 ± 1.1 mm ($P < .007$) at 90°. At 0° of knee flexion, significant reductions in ATT compared with the intact state resulted from the 60°/60° ($P < .008$) and 90°/90° ($P < .002$) fixation states. At 20° of knee flexion, significant reductions in ATT compared with the intact state resulted from the 60°/60° ($P < .01$) and 90°/90° ($P < .004$) fixation states. At 30° of knee flexion, significant reductions in ATT compared with the intact state resulted from the 60°/60° ($P < .03$) and 90°/90° ($P < .002$) fixation states.

Validation

To validate the graft fixation procedure, the first graft fixation state was reapplied and retested following biomechanical testing of the final graft fixation state. The testing found the mean absolute differences between the original test and validation test displacements of 1.1 ± 0.1 mm of ATT for anterior tibial loads, 2.0° ± 0.2° of internal tibial rotation for internal tibial torques, and 1.5 ± 0.2 mm of ATT for the coupled internal rotation and valgus loads.

DISCUSSION

In this study, 7 different combinations of graft fixation angles in DB-ACLRs were biomechanically evaluated. The results validated the first part of our hypothesis in that the 60°/0° fixation angle combination, which simulated the native tensioning relationship between the AM and PL bundles, restored normal stability to the knee joint without causing any underconstraint or overconstraint. Additionally, fixation angle combinations that approximated the anatomical tensioning relationship of the bundles—the 45°/15° and 75°/15° combinations—also restored normal stability to the knee joint without causing any underconstraint or overconstraint. The latter part of our hypothesis was validated as well because the AM and PL graft fixation angle combinations that did not simulate the native tensioning relationship of the intact bundles overconstrained the knee. The 30°/30° fixation angle combination significantly reduced knee laxity with respect to internal rotation and the coupled internal rotation and valgus torques simulated pivot shift. The 60°/60° fixation angle combination significantly reduced knee laxity with respect to valgus angulation, internal rotation, and both of the simulated pivot shift tests. The 90°/90° fixation angle combination significantly reduced knee laxity with respect to internal rotation and both of the simulated pivot shift tests. However, the 0°/0° fixation angle combination differed from our hypothesis because it restored normal knee stability without causing any joint underconstraint or overconstraint to the knee, even though it did not simulate the native tensioning relationship of the intact bundles. We did not find a graft fixation angle combination that underconstrained the knee.

We found that all of the graft fixation states in which the PL bundle was fixed at 30° of knee flexion or higher caused the knee to be significantly overconstrained compared with the normal knee, while all of the fixation states with the PL bundle fixed at 15° or less reproduced normal

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knee kinematics. We also found that the AM bundle graft could be fixed between 0° and 75° of knee flexion without causing joint overconstraint as long as the PL bundle graft was fixed at or below 15° of knee flexion. To avoid abnormal knee kinematics, we conclude that the PL bundle should be fixed at or below 15° of knee flexion and that it is permissible to fix the AM bundle between 0° and 75°.

Our results for tensioning of the ACL bundle grafts are consistent with results found in previous biomechanical studies that evaluated graft fixation angles in DB-ACLRs. The range of graft fixation angles that we concluded does not overconstrain the knee includes the range of fixation angles previously reported to not overload the AM and PL bundle grafts.[38] Our results indicating that the 30°/30° fixation angle combination significantly restricted knee laxity were consistent with reports that the 30°/30° fixation angle combination significantly overloaded the PL bundle graft.[26] However, while we found that the 60°/0° combination did not overconstrain knee laxity, a previous study has reported that the 60°/0° fixation angle combination significantly overloaded the AM bundle graft.[26] Because we did not gather tensioning data on the grafts, we cannot verify if the AM bundle was overloaded in the 60°/0° fixation state, even though knee stability was normal.

It has also been reported that graft bundle fixation at 20°/20° resulted in normal knee anteroposterior laxity, while fixation at both the 90°/20° and 20°/90° angle combinations overconstrained anteroposterior laxity.[9] Because we did not test fixation of the PL bundle between 15° and 30°, we could only confirm that the angle at which the PL bundle begins to overconstrain the knee joint falls somewhere within this range. Because no reduction in anteroposterior laxity was reported in the study by Cuomo et al[9] at the 20°/20° fixation state, these results could be used to expand the range of recommended PL bundle fixation angles from 0° to 15° to a range of 0° to 20°. We did not test the 20°/90° fixation angle combination, but agree with the finding that fixation of the PL bundle graft at 90° would limit knee laxity more than fixation at 20° when the AM bundle is fixed at 20° based on our results.[9] Additionally, we did not find a graft fixation angle combination with the AM bundle graft fixed in flexion that overconstrained the knee. However, we did not test any fixation angle combination with the AM bundle graft fixed above 75°. The fact that fixation at 90°/20° was found to overconstrain anteroposterior laxity in a previous study,[9] but not fixation at 75°/15° in our study, suggests that the angle at which the AM bundle graft becomes overloaded and contributes to joint overconstraint occurs between the angles of 75° and 90°.

Consideration of the native ACL bundle length-tensioning relationship provides a mechanistic framework that may explain why joint overconstraint occurs more with fixation of the PL bundle graft outside the peak bundle tension knee flexion angles than the AM bundle graft. Throughout the range of knee motion, the AM bundle maintains a majority of its tension, while the PL bundle slackens increasingly as the knee is flexed.[16] Likewise, it has been reported that the length of the AM bundle changes relatively little compared with the length of the PL bundle.[21,23] The AM bundle has been reported to shorten by 2.5 mm at knee flexion angles differing from its native peak bundle tension in the range of 0° and 90°, while the PL bundle shortens by 6.7 mm with flexion from 0° to 90°.[23] Fixation of a bundle at a knee flexion angle different than its peak anatomical tension would cause the bundle to experience increased load because of its shorter length at fixation. Because the PL bundle shortens more rapidly with initial knee flexion compared with the AM bundle, the PL bundle graft would theoretically have a more limited range of fixation angles that do not cause the graft to be overloaded than the AM bundle graft.

Fixation that overloads the ACL bundle grafts may increase the risk of graft failure. Recent biomechanical and clinical reports of PL bundle graft failures further support our conclusion that the PL bundle graft has a more restricted range of fixation angles that do not cause overloading of tensions than the AM bundle graft. It has been suggested that the 5 PL bundle ruptures reported in a recent clinical study[3] could be attributed to graft fixation at 30°/30°, which resulted in overloading of the PL bundle graft.[14] Because we found that fixation of the PL graft at 30° or higher overconstrained the knee, our results support the conclusion that the 30°/30° fixation angle combination would predispose the PL bundle graft to an increased risk of failure. Similarly, a recent biomechanical study that included fixation of the PL bundle at 30° during the testing protocol reported 3 PL graft failures out of 14 specimens, all of which occurred during testing at 0° of knee flexion where the PL bundle is tightest.[25] In our study, one AM bundle failure occurred during validation fixation, and a combined AM and PL bundle failure occurred during testing of the 60°/60° graft fixation state, which we theorize occurred because of graft overload.

This study was limited in that it did not evaluate other variables in current DB-ACLR surgical procedures. The placement and number of the femoral and tibial tunnels differ between DB-ACLR studies. It has been reported that nonanatomical placement of the femoral PL bundle tunnel results in rotational instability, while anatomical placement of the femoral PL bundle tunnel restores normal knee stability.[44] Additionally, the use of 2 tibial tunnels was reported to restore knee kinematics better than the use of 1 tibial tunnel.[30] We performed all DB-ACLRs with 2 femoral and 2 tibial tunnels placed at the precise center of their respective bundle footprints to remain as anatomical as possible in our reconstructions.

Furthermore, there is a lack of consensus on the proper amount of graft pretension. The value of 20 N of pretension for each graft, which has been previously reported to sufficiently restore normal stability to the knee,[41] was selected for this study to avoid diminishing the kinematic differences between the graft fixation states due to overtensioning the grafts during fixation. However, some studies have reported that graft pretensions greater than were applied in the present investigation were needed to restore normal stability to the knee.[29,33] Because no graft fixation state resulted in any underconstraint in this study, we conclude that 20 N of pretension on each ACL bundle graft was sufficient for DB-ACLRs.

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CONCLUSION

We found that the fixation angles of the AM and PL ACL bundle grafts significantly affected knee kinematics. Fixation at 0°/0°, 60°/0°, 45°/15°, or 75°/15° restored normal stability to the knee, while fixation at 30°/30°, 60°/60°, and 90°/90° overconstrained the knee. We found that as long as the PL ACL bundle graft was fixed in the range of 0° to 15°, the AM ACL bundle graft could be fixed at up to 75° of knee flexion without restricting normal knee laxity. However, fixation of the PL ACL bundle graft at 30° of knee flexion and above overconstrained the knee and may predispose the PL ACL bundle graft to failure.

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