ACL Deficiency Increases Forces on the Medial Femoral Condyle and the Lateral Meniscus with Applied Rotatory Loads

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Background: The articular surfaces and menisci act with the anterior cruciate ligament (ACL) to stabilize the knee joint. Their role in resisting applied rotatory loads characteristic of instability events is unclear despite commonly observed damage to these intra-articular structures in the acute and chronic ACL injury settings.

Methods: Ten fresh-frozen human cadaveric knees were mounted to a robotic manipulator. Combined valgus and internal rotation torques were applied in the presence and absence of a 300-N compressive load. Forces carried by the individual menisci and via cartilage-to-cartilage contact on each femoral condyle in ACL-intact and ACL-sectioned states were measured using the principle of superposition.

Results: In response to applied valgus and internal rotation torques in the absence of compression, sectioning of the ACL increased the net force carried by the lateral meniscus by at most 65.8 N ($p < 0.001$). Moreover, the anterior shear force carried by the lateral meniscus increased by 25.7 N ($p < 0.001$) and 36.5 N ($p = 0.042$) in the absence and presence of compression, respectively. In response to applied valgus and internal rotation torques, sectioning of the ACL increased the net force carried by cartilage-to-cartilage contact on the medial femoral condyle by at most 38.9 N ($p = 0.006$) and 46.7 N ($p = 0.040$) in the absence and presence of compression, respectively. Additionally, the lateral shear force carried by cartilage-to-cartilage contact on the medial femoral condyle increased by at most 21.0 N ($p = 0.005$) and by 28.0 N ($p = 0.025$) in the absence and presence of compression, respectively. Forces carried by the medial meniscus and by cartilage-to-cartilage contact on the lateral femoral condyle changed by <5 N as a result of ACL sectioning.

Conclusions: ACL sectioning increased the net forces carried by the lateral meniscus and medial femoral condyle—and the anterior shear and lateral shear forces, respectively—in response to multiplanar valgus and internal rotation torque.

Clinical Relevance: These loading patterns provide a biomechanical rationale for clinical patterns of intra-articular derangement such as lateral meniscal injury and osseous remodeling of the medial compartment seen with ACL insufficiency.

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Injury to chondral and meniscal tissues is commonly found with both acute and chronic anterior cruciate ligament (ACL) ruptures. Meniscal injury is observed in up to 81% of individuals undergoing acute ACL reconstruction[1-3] and up to 89% of those undergoing delayed ACL reconstruction[2,3]. Chondral injuries are reported in up to 23% of individuals undergoing acute ACL reconstruction and 54% of those undergoing delayed ACL reconstruction[2].

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TABLE I Forces Carried by the Menisci During Applied Rotatory Loads without Axial Compression*

	5° Flexion			15° Flexion
	ACL Intact†	ACL Deficient†	P Value	ACL Intact†
Lateral meniscus				
Proximal force	−27.7 ± 31.5 (−47.3 to −8.2)	−32.7 ± 38.1 (−56.3 to −9.1)	0.492	−22.9 ± 27.1 (−39.7 to −6.0)
Medial force	−0.8 ± 7.2 (−5.3 to 3.7)	1.2 ± 4.3 (−1.5 to 3.8)	0.315	2.6 ± 8.9 (−3.0 to 8.1)
Anterior force	2.0 ± 4.8 (−0.9 to 5.0)	4.2 ± 5.5 (0.8 to 7.6)	0.260	3.0 ± 6.7 (−1.1 to 7.1)
Net force	30.2 ± 30.2 (11.5 to 49.0)	38.8 ± 32.2 (18.8 to 58.7)	0.212	26.3 ± 26.2 (10.1 to 42.6)
Medial meniscus				
Proximal force	−3.8 ± 7.1 (−8.2 to 0.6)	−3.8 ± 7.5 (−8.5 to 0.8)	0.993	−9.9 ± 6.3 (−13.9 to −6.0)
Medial force	−0.3 ± 1.3 (−1.1 to 0.5)	−0.6 ± 3.0 (−2.4 to 1.2)	0.752	−0.9 ± 2.0 (−2.1 to 0.4)
Anterior force	1.8 ± 2.9 (0.0 to 3.6)	1.1 ± 2.5 (−0.4 to 2.6)	0.155	3.4 ± 5.4 (0.0 to 6.8)
Net force	6.5 ± 5.9 (2.8 to 10.1)	6.8 ± 6.1 (3.1 to 10.6)	0.754	11.4 ± 7.2 (6.9 to 15.9)

*No addition of compression beyond the 10-N axial compression applied at the starting point. †The values are given, in newtons, as the mean and standard deviation with the 95% confidence interval in parentheses.

The medial meniscus acts as a secondary stabilizer against applied anterior tibial forces[4,5]. In vitro data revealed that sectioning the ACL increases load borne by the medial meniscus up to threefold[4,5]. Conversely, medial meniscectomy increases forces carried by the ACL about twofold[4,6].

The lateral meniscus is an important secondary stabilizer during pivoting maneuvers[6,7]. In previous in vitro studies, sectioning the lateral meniscus in ACL-deficient knees produced, on average, an additional 6 mm of anterior subluxation in response to the pivot-shift maneuver compared with the value after sectioning the ACL alone[6,7]. Sectioning of both the lateral meniscus and the ACL increased anterior translation of the lateral compartment during a pivot-shift maneuver compared with isolated ACL sectioning[6]. Moreover, lateral meniscal tears may occur during pivoting events. However, the load borne by the lateral meniscus in response to multiplanar valgus and internal rotation torques that arise during functional pivoting activities remains poorly defined[8,9].

The femoral condyles and the tibial spine likely interact via direct cartilage-to-cartilage contact, thereby resisting rotatory loads including internal rotation torque combined with anterior translation[10]. Moreover, Fairclough et al. reported radiographic findings of osteophyte formation on the medial femoral condyle and medial tibial spine in chronically ACL-deficient knees, which suggests altered loading of the medial compartment[11]. Additionally, Okafor et al. reported that vertical ACL graft placement was followed by abnormal cartilage thinning on the medial femoral condyle, suggesting that the medial condyle is a restraint to abnormal rotations and translations[12]. However, the forces borne by the articular surfaces in the presence and absence of the ACL have not been directly measured[10,13], to our knowledge.

The purpose of this investigation was to determine the effect of ACL sectioning on loading patterns of the medial and lateral tibiofemoral articular surfaces and menisci in response to combined valgus and internal rotation torques that arise during clinical and functional pivoting scenarios. Our research questions were (1) what are the roles of the lateral and medial menisci in ACL-intact and ACL-sectioned knees and (2) does ACL insufficiency increase the forces carried by direct cartilage-to-cartilage contact on the medial and/or lateral femoral condyles? We hypothesized that applying rotatory loads after sectioning the ACL would (1) increase forces carried by the lateral meniscus but not those carried by the medial meniscus and (2) increase forces carried by direct cartilage-to-cartilage contact on the lateral and medial articular surfaces. Tibiofemoral translations were also documented to characterize the amount of subluxation caused by sectioning of the ACL.

Materials and Methods

This institutional review board-approved study was done on 10 fresh-frozen human cadaveric knee specimens (mean age [and standard deviation] of the donors at the time of death, 37.4 ± 12.3 years [range, 20 to 53 years]; 2 female donors; 3 right knees). This was a sample of convenience; a sample size estimate was not possible due to the absence of previous reports on the force carried by the intra-articular stabilizers. Skin and soft tissue >11 cm from the joint line were removed. The skin, fat, and muscles surrounding the joint were removed with the exception of the popliteal muscle-tendon complex, leaving only the ligamentous and capsular structures. Intra-articular structures were visually inspected via a medial arthrotomy. Computed tomography (CT) and magnetic resonance imaging (MRI) were performed on all knees prior to testing. The scans revealed no osteophytes or soft-tissue damage. To maintain the anatomical position of the fibula, it was fixed to the tibia with a wood screw 5 cm distal to the fibular head. Upon preparing the specimens for testing, careful attention was paid to maintaining and subsequently securing the fibula in the anatomic position. This fixation was held constant across all test conditions; therefore, it was controlled for in our experiment and the repeated-measures study design. We do not believe this fixation alters the loading of the articular surfaces. The tibia and femur were potted in bonding cement (Bondo; 3M) using aluminum cylinders with diameters of 5 cm.

Specimens were mounted to a 6-degrees-of-freedom robotic arm (ZX165U; Kawasaki Robotics) with ±0.3-mm repeatability. A universal force/moment sensor (Theta; ATI) was affixed to the end effector. The femur was fixed to the floor via a pedestal. The tibia was mounted to a fixture attached to the force/moment sensor. Anatomical landmarks were defined using a 3-dimensional digitizing arm (G2X;

TABLE I (*continued*)

15° Flexion		30° Flexion		
ACL Deficient†	P Value	ACL Intact†	ACL Deficient†	P Value
−50.6 ± 40.2 (−75.5 to −25.7)	0.013	−25.2 ± 26.1 (−41.4 to −9.0)	−87.7 ± 40.7 (−113.0 to −62.4)	<0.001
3.6 ± 11.1 (−3.2 to 10.5)	0.548	3.4 ± 9.8 (−2.7 to 9.5)	9.3 ± 12.5 (1.6 to 17.0)	0.081
12.0 ± 8.9 (6.5 to 17.5)	0.008	4.2 ± 7.2 (−0.3 to 8.7)	29.9 ± 16.6 (19.6 to 40.2)	<0.001
53.0 ± 41.4 (27.4 to 78.6)	0.014	28.3 ± 26.1 (12.2 to 44.5)	94.0 ± 43.7 (66.9 to 121.1)	<0.001
−8.3 ± 11.6 (−15.5 to −1.1)	0.560	−5.5 ± 2.9 (−7.3 to −3.7)	−5.5 ± 5.4 (−8.9 to −2.2)	0.989
−2.0 ± 3.2 (−3.9 to 0.0)	0.235	0.6 ± 2.1 (−0.7 to 1.8)	−0.5 ± 2.6 (−2.1 to 1.2)	0.177
2.7 ± 2.6 (1.1 to 4.4)	0.623	2.7 ± 5.2 (−0.5 to 6.0)	1.9 ± 1.6 (0.9 to 2.8)	0.521
11.7 ± 9.4 (5.9 to 17.6)	0.859	7.2 ± 4.9 (4.2 to 10.3)	7.4 ± 4.0 (4.9 to 9.9)	0.928

MicroScribe) with an accuracy of ±0.23 mm. A tibiofemoral coordinate system was defined on the basis of the locations of these landmarks[14,15]. The femoral epicondyles defined the axis of flexion, while the long axis of the tibia defined the axis of internal and external tibial rotation with respect to the femur. The common perpendicular of these axes defined the axis of varus-valgus angulation. Relative translations of the tibia and femur were defined along each anatomical axis with respect to a point positioned at the midpoint between the epicondyles. To visualize kinematic changes resulting from sectioning of the ACL, CT scans of the tibial and femoral bones and cartilage of 1 representative specimen were registered to their respective positions in the robot experiment using a validated method[16]. Loads measured by the force/moment sensor were resolved along and about the anatomical axes using a force-moment transformation[17].

A path of passive knee flexion was determined from 0° (full extension) to 90° in 1° increments with a compressive force of 10 N. Knees were not flexed beyond 90° because no rotatory loads were applied beyond this angle. The positions and orientations of the knee along this flexion path served as the starting points for subsequent rotatory loading conditions with and without compression.

Combined torques capturing a subset of loads applied during a clinical pivot-shift examination and generating anterior subluxation of the tibia were applied to the knee[18-21]. The loads were 8 Nm of valgus torque and then, with maintenance of the valgus torque, 4 Nm of internal rotation torque. These combined torques were applied at 5°, 15°, and 30° of flexion since pivoting events typically occur near full extension and the pivot-shift examination begins in full extension and continues through early flexion[22]. Combined torques of 15 Nm of valgus and 6 Nm of internal rotation were applied with 300 N of compression at 15° of flexion in a subset of 5 knees[8,9]. The axial force and combined torques were applied only to a subset of knees because the axial loading protocol was added after several specimens had already been tested and inconsistent rotatory loads were applied across the ACL-intact and ACL-sectioned conditions in some specimens. These multiplanar torques were applied with compression because motion analysis of ACL-deficient individuals during stair descent and crossover cutting had revealed that instability events occurred in the presence of valgus and internal rotation torques[8,9]. The compressive force was nearly half of a typical body weight, which avoided damaging the knee during repeated measurements. The kinematic trajectories of the tibia were recorded in ACL-intact and ACL-sectioned states in response to multiplanar torques in the presence and absence of compression prior to removal of any soft tissues (Fig. 1, Steps 1 and 2). A minimization algorithm was used to determine the positions of the tibia relative to the femur at which the applied loads were within 5 N and 0.4 Nm of the target loads. The convergence tolerances were selected on the basis of a repeatability study, which revealed minimal changes in knee kinematics (≤0.05 mm and ≤0.11°) with application of valgus and internal rotation torques of 8 and 4 Nm, respectively. The uncertainty regarding ACL force was ±4 N given the aforementioned position error. Assuming a similar error in force measurements for the intra-articular tissues, it is unlikely that the load tolerances affected our conclusions.

TABLE II Forces Carried by the Menisci During Applied Rotatory Loads with 300 N of Axial Compression (in 15° of Flexion)

	ACL Intact*	ACL Deficient*	P Value
Lateral meniscus			
Proximal force	−72.4 ± 35.5 (−103.5 to −41.2)	−122.8 ± 76.9 (−190.2 to −55.4)	0.072
Medial force	10.9 ± 5.1 (6.4 to 15.3)	29.4 ± 19.9 (12.0 to 46.9)	0.121
Anterior force	12.1 ± 4.9 (7.8 to 16.4)	48.6 ± 27.4 (24.6 to 72.6)	0.042
Net force	74.5 ± 35.3 (43.5 to 105.5)	136.2 ± 82.1 (64.2 to 208.2)	0.062
Medial meniscus			
Proximal force	−16.0 ± 12.5 (−27.0 to −5.0)	−15.5 ± 9.7 (−24.1 to −7.0)	0.816
Medial force	0.4 ± 4.8 (−3.7 to 4.6)	0.5 ± 3.5 (−2.6 to 3.6)	0.941
Anterior force	3.6 ± 3.7 (0.3 to 6.9)	5.1 ± 4.3 (1.3 to 8.8)	0.066
Net force	17.6 ± 12.0 (7.1 to 28.1)	16.8 ± 10.4 (7.7 to 25.9)	0.663

*The values are given, in newtons, as the mean and standard deviation with the 95% confidence interval in parentheses.

TABLE III Forces Carried by the Femoral Condyles During Applied Rotatory Loads without Compression*

	5° Flexion			15° Flexion
	ACL Intact†	ACL Deficient†	P Value	ACL Intact†
Lateral femoral condyle				
Proximal force	−20.0 ± 26.5 (−36.4 to −3.6)	−21.9 ± 33.3 (−42.6 to −1.3)	0.533	−5.7 ± 9.7 (−11.7 to 0.4)
Medial force	4.2 ± 6.7 (0.0 to 8.4)	5.6 ± 11.8 (−1.8 to 12.9)	0.520	1.0 ± 3.0 (−0.9 to 2.9)
Anterior force	1.1 ± 2.5 (−0.4 to 2.7)	2.5 ± 5.6 (−0.9 to 6.0)	0.232	0.6 ± 0.9 (0.0 to 1.2)
Net force	21.6 ± 26.4 (5.2 to 38.0)	23.8 ± 35.0 (2.1 to 45.5)	0.508	6.9 ± 9.4 (1.1 to 12.7)
Medial femoral condyle				
Proximal force	−14.9 ± 13.0 (−22.9 to −6.9)	−46.6 ± 33.8 (−67.5 to −25.6)	0.007	−12.5 ± 15.6 (−22.2 to −2.8)
Medial force	−8.3 ± 7.0 (−12.6 to −4.0)	−29.3 ± 20.7 (−42.2 to −16.5)	0.005	−6.9 ± 9.6 (−12.9 to −1.0)
Anterior force	1.8 ± 1.7 (0.7 to 2.8)	1.6 ± 2.7 (−0.1 to 3.2)	0.857	1.9 ± 2.9 (0.1 to 3.7)
Net force	17.3 ± 14.6 (8.3 to 26.4)	55.3 ± 39.3 (30.9 to 79.7)	0.006	14.6 ± 18.4 (3.3 to 26.0)

*No addition of compression beyond the 10-N axial compression applied at the starting point. †The values are given, in newtons, as the mean and standard deviation with the 95% confidence interval in parentheses.

After determination of the kinematic trajectories of the ACL-intact and the ACL-sectioned knee in response to the applied multiplanar torques with and without compression, all soft tissues except for the menisci with their anterior and posterior horns and coronary attachments were dissected. Then, each kinematic trajectory was repeated, while recording the reaction forces, before and after removing each meniscus (Fig. 1, Steps 3 and 4). To isolate the force carried by each meniscus, the vector difference of the reaction forces measured before and after meniscal removal was calculated (i.e., the superposition principle was used)[23]. After sectioning of the menisci, only cartilage-to-cartilage contact between the tibia and femur remained. The kinematic trajectories were repeated, while recording the reaction forces, before and after removing each condyle (Fig. 1, Steps 5 and 6). The superposition principle was again employed to isolate the force carried by each femoral condyle via direct cartilage-to-cartilage contact with the tibia (Fig. 2). The condyles were removed in alternating order from specimen to specimen, to account for potential artifacts arising from the sectioning order.

Forces carried by the menisci and by the condyles via direct cartilage-to-cartilage contact were described in terms of the anatomical directions: proximal-distal, medial-lateral shear, and anterior-posterior shear[17]. Specifically, the reported resultant (i.e., net) forces and their components describe the forces acting on the tibia, following the sectioning of each structure, in response to the applied loads. The net force was determined as the square root of the sum of the squares of the individual anterior-posterior, medial-lateral, and proximal-distal forces[23]. Data were summarized using means, standard deviations, and 95% confidence intervals. The data compared between the ACL-intact and ACL-sectioned conditions at each flexion angle tested included the forces carried by the menisci and via cartilage-to-cartilage contact at each femoral condyle. All data were confirmed to be normally distributed using the Kolmogorov-Smirnov test and then were compared between the ACL-intact and ACL-sectioned conditions using paired, two-tailed t tests with $p < 0.05$ as the level of significance.

Results

Sectioning the ACL increased anterior tibial translation up to 6.7 ± 2.0 mm (at 15° of flexion) and medial tibial translation up to 4.2 ± 1.2 mm (at 30° of flexion) (all $p < 0.001$) in

TABLE IV Forces Carried by the Femoral Condyles During Applied Rotatory Loads with 300 N of Axial Compression (in 15° of Flexion)

	ACL Intact*	ACL Deficient*	P Value
Lateral femoral condyle			
Proximal force	−124.3 ± 56.0 (−173.4 to −75.1)	−86.6 ± 54.0 (−134.0 to −39.2)	0.251
Medial force	29.5 ± 23.2 (9.1 to 49.8)	27.2 ± 25.6 (4.7 to 49.7)	0.774
Anterior force	19.1 ± 7.9 (12.1 to 26.0)	31.0 ± 22.1 (11.6 to 50.4)	0.310
Net force	129.7 ± 59.6 (77.4 to 182.0)	96.5 ± 62.6 (41.6 to 151.4)	0.331
Medial femoral condyle			
Proximal force	−89.3 ± 13.3 (−101.0 to −77.7)	−125.0 ± 33.3 (−154.1 to −95.8)	0.064
Medial force	−50.9 ± 12.8 (−62.1 to −39.7)	−78.9 ± 18.1 (−94.8 to −63.0)	0.025
Anterior force	15.2 ± 9.0 (7.2 to 23.1)	27.0 ± 14.8 (14.0 to 40.0)	0.272
Net force	104.5 ± 16.9 (89.7 to 119.3)	151.2 ± 36.0 (119.7 to 182.7)	0.040

*The values are given, in newtons, as the mean and standard deviation with the 95% confidence interval in parentheses.

TABLE III (*continued*)

15° Flexion		30° Flexion		
ACL Deficient†	P Value	ACL Intact†	ACL Deficient†	P Value
−3.6 ± 4.8 (−6.6 to −0.7)	0.286	−7.3 ± 9.4 (−13.2 to −1.5)	−3.0 ± 3.6 (−5.3 to −0.8)	0.112
−0.7 ± 1.5 (−1.6 to 0.3)	0.025	1.0 ± 2.3 (−0.5 to 2.4)	−1.3 ± 1.5 (−2.2 to −0.4)	0.012
0.8 ± 1.1 (0.1 to 1.5)	0.447	1.2 ± 1.6 (0.2 to 2.2)	0.8 ± 1.3 (0.0 to 1.6)	0.190
4.4 ± 4.6 (1.6 to 7.2)	0.184	8.0 ± 9.4 (2.2 to 13.8)	3.9 ± 3.6 (1.6 to 6.1)	0.136
−24.7 ± 23.2 (−39.1 to −10.3)	0.013	−16.7 ± 16.6 (−27.0 to −6.5)	−15.1 ± 17.1 (−25.7 to −4.4)	0.634
−15.5 ± 13.6 (−23.9 to −7.1)	0.006	−10.3 ± 10.3 (−16.6 to −3.9)	−10.1 ± 9.4 (−15.9 to −4.2)	0.922
4.1 ± 5.2 (0.9 to 7.3)	0.116	4.2 ± 4.5 (1.4 to 6.9)	4.5 ± 5.4 (1.2 to 7.9)	0.725
29.7 ± 27.1 (12.9 to 46.5)	0.009	20.2 ± 19.8 (8.0 to 32.5)	19.0 ± 20.0 (6.6 to 31.3)	0.750

response to combined rotatory loads applied without compression (i.e., with only the 10 N of axial compression at the starting point) (Figs. 3 and 4). ACL sectioning also increased anterior and medial translations by 9.0 ± 3.4 mm and 5.3 ± 2.5 mm, respectively (all $p < 0.001$), in response to combined rotatory loads applied with compression (Figs. 3 and 4).

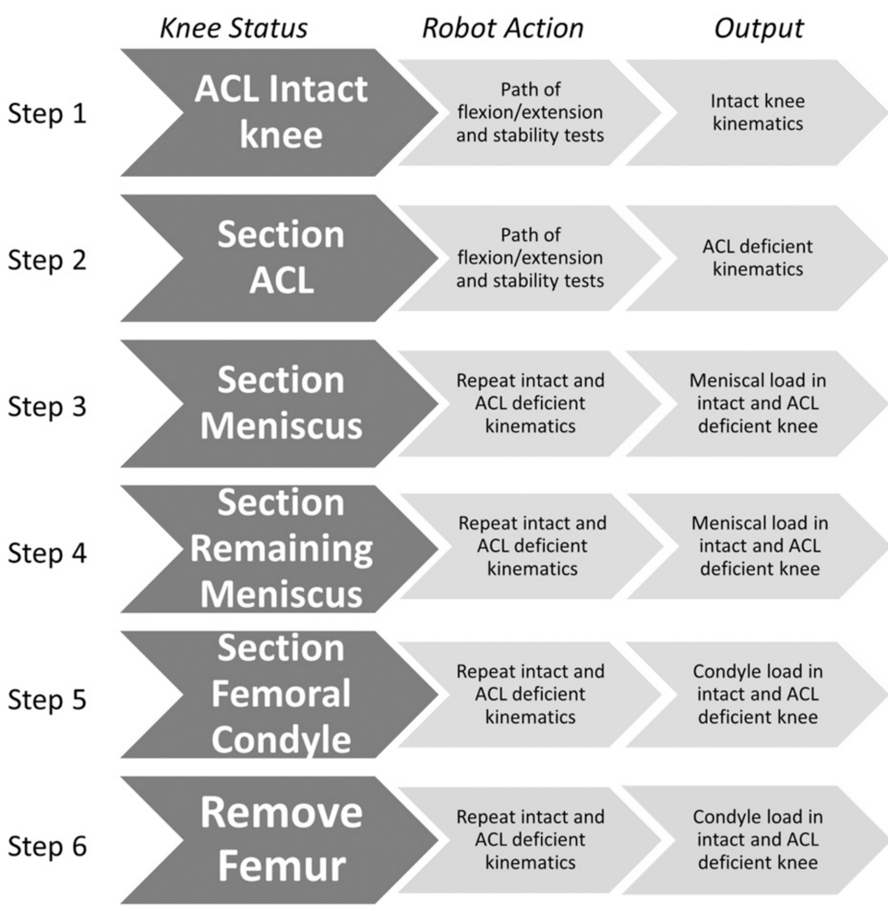

Fig. 1

Flowchart outlining the experimental test protocol.

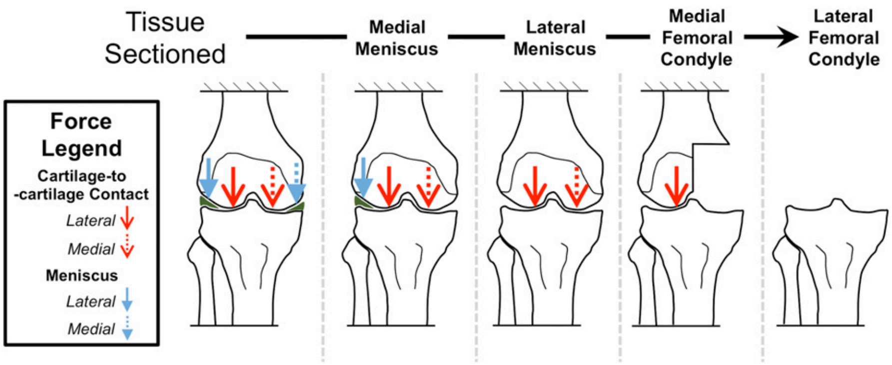

Fig. 2

Schematic diagram detailing the method used to identify forces carried by each meniscus and by each femoral condyle due to cartilage-to-cartilage contact. The force transmitted by the sectioned tissue to the tibia was determined by vector subtraction of the joint reaction force measured before and after sectioning each structure. The force carried by each structure in response to combined valgus and internal rotation torques, in the presence and absence of compression, was determined with the ACL intact and with it sectioned.

ACL sectioning increased the net force carried by the lateral meniscus by 26.7 ± 27.7 N at 15° of flexion ($p = 0.014$) and 65.8 ± 37.5 N at 30° of flexion ($p < 0.001$) in response to combined rotatory torques in the absence of applied compression (Table I). The increase in net force carried by the lateral meniscus was most pronounced at 30° of flexion, at which distal and anterior shear forces carried by the lateral meniscus increased by 62.5 ± 36.0 N ($p < 0.001$) and 25.7 ± 16.4 N ($p < 0.001$), respectively. Sectioning the ACL increased the anterior shear force carried by the lateral meniscus by 36.5 ± 27.6 N ($p = 0.042$) under testing with applied compression (Table II). No changes in the force carried by the medial meniscus were detected at any flexion angle after sectioning the ACL, with net forces averaging <20 N (Tables I and II).

During the application of the combined torques in the absence of applied compression, the knees in which the ACL had been sectioned had a higher net cartilage-to-cartilage contact force on the medial femoral condyle but not on the lateral femoral condyle (Table III). The net cartilage-to-cartilage contact force carried by the medial femoral condyle increased an

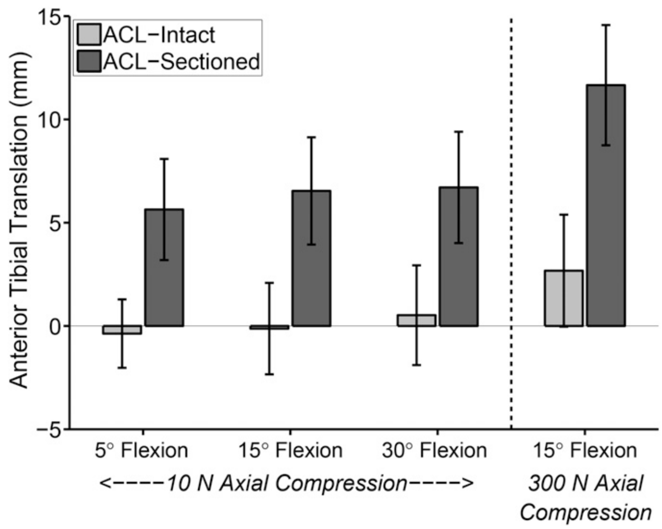

Fig. 3

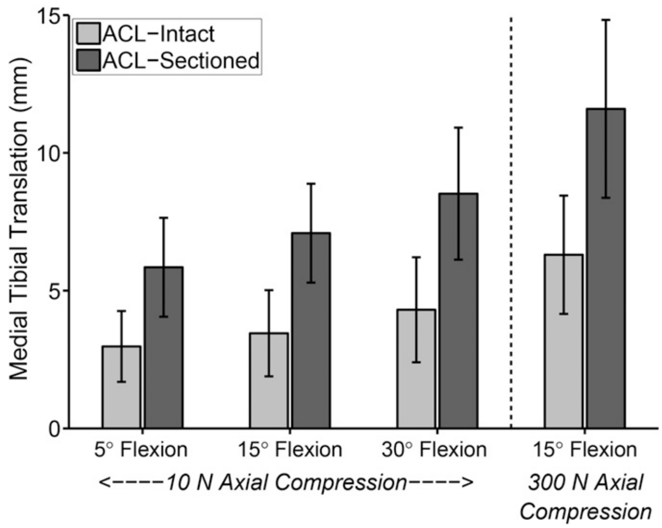

Fig. 4

Fig. 3 Mean anterior tibial translations, measured in millimeters, of ACL-intact and ACL-sectioned knees in response to combined valgus and internal rotation torques without compression (i.e., none added after application of the 10 N at the starting point) and with 300 N of compression. Differences in translation between ACL-intact and ACL-sectioned knees were significant at all reported knee flexion angles ($p < 0.001$). The whiskers represent standard deviations. **Fig. 4** Mean medial tibial translation, measured in millimeters, of ACL-intact and ACL-sectioned knees in response to combined valgus and internal rotation torques without compression (i.e., none added after application of the 10 N at the starting point) and with 300 N of compression. Differences in translation between ACL-intact and ACL-sectioned knees were significant at all reported knee flexion angles ($p < 0.001$). The whiskers represent standard deviations.

average of 38.0 ± 33.6 N (p = 0.006) at 5° of flexion and 15.1 ± 14.3 N (p = 0.009) at 15° of flexion (Table III). At 5° of flexion, distal and lateral cartilage-to-cartilage contact forces carried by the medial femoral condyle in the ACL-sectioned knee increased an average of 31.7 ± 28.5 N (p = 0.007) and 21.0 ± 18.0 N (p = 0.005), respectively (Table III). Changes in cartilage-to-cartilage contact force carried by the lateral femoral condyle after ACL sectioning averaged <5 N (Table III).

Sectioning of the ACL increased net cartilage-to-cartilage contact force carried by the medial femoral condyle by 46.7 ± 34.8 N (p = 0.040) in response to combined rotatory and compressive loads (Table IV). ACL sectioning also increased laterally directed cartilage-to-cartilage contact force carried by the medial femoral condyle by 28.0 ± 17.8 N (p = 0.025) (Table IV). No changes in net cartilage-to-cartilage contact force carried by the lateral femoral condyle were detected as a result of sectioning of the ACL (Table IV).

Discussion

The main findings in this study were that sectioning the ACL increased forces carried by (1) the lateral meniscus and (2) the medial femoral condyle via cartilage-to-cartilage contact in response to multiplanar valgus and internal rotation loads. Independent of applied compression, sectioning of the ACL increased the lateral and anterior forces carried by cartilage-to-cartilage contact on the medial femoral condyle and by the lateral meniscus, respectively.

Our hypothesis that sectioning the ACL would increase forces carried by the lateral meniscus but not the medial meniscus was confirmed. This occurred at 15° and 30° of flexion, which is consistent with the clinical pivoting phenomenon, in which the tibia transitions from an anteriorly subluxated position to a reduced position with knee flexion[6,24-26]. While the knee is subluxated, the lateral meniscus encounters greater forces than encountered in ACL-intact knees without this subluxation[3,27]. Moreover, the increased anterior shear and distal forces carried by the lateral meniscus in response to applied rotatory loads may help explain concomitant lateral meniscal tears during acute ACL rupture and instability episodes[3,13,27]. Specifically, anterior tibial subluxation, particularly at 15° and 30° of flexion, wedges the posterior aspect of the lateral meniscus between the lateral femoral condyle and the posterior portion of the lateral tibial plateau (Fig. 5).

Our hypothesis that applying combined valgus and internal rotation torques would increase the force transmitted across the lateral compartment through direct cartilage-to-cartilage contact was not supported. ACL deficiency only increased the force carried by the lateral meniscus. This finding corroborates previous work demonstrating increased contact stress with combined torques on the posterior aspect of the lateral compartment in the region of meniscus-to-cartilage contact[14,28,29].

ACL sectioning increases the laterally directed cartilage-to-cartilage contact forces that the medial femoral condyle imparts to the tibia in response to rotatory loads. This stabilizing mechanism compensates for the laterally directed force otherwise produced by the ACL[14,30]. Specifically, anteromedial tibial subluxation resulting from ACL sectioning likely causes cartilage-to-cartilage contact between the medial tibial spine and the medial femoral notch (Fig. 6), and this contact likely produces the observed increase in the laterally oriented force carried by cartilage-to-cartilage contact of the medial condyle after ACL sectioning (Fig. 7, Table III). Unfortunately, this stabilizing mechanism increases force on regions of cartilage that would otherwise be minimally loaded by rotatory torques. These abnormal contact forces provide a biomechanical rationale for the common radiographic findings of sequelae of chronic ACL deficiency such as osteophyte formation on the medial femoral condyle and medial tibial spine[11,31-33].

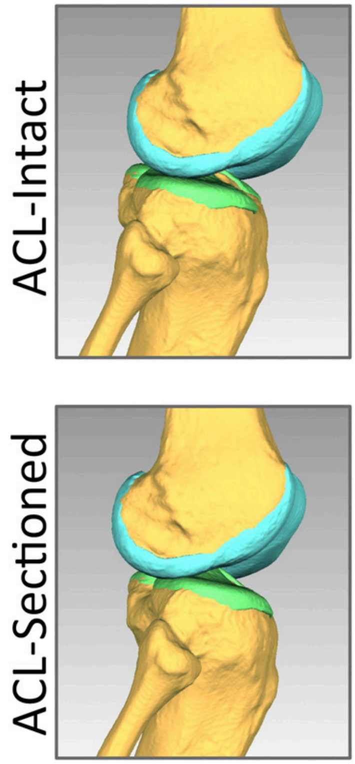

Fig. 5
Sagittal view of the lateral aspect of a representative specimen. The tibia is in the position and orientation resulting from combined valgus and internal rotation torques in the ACL-intact and ACL-sectioned conditions at 30° of flexion. The tibial cartilage is shown in green, and the femoral cartilage is shown in blue.

Our finding that ACL sectioning increased the forces carried by the lateral meniscus and by cartilage-to-cartilage contact on the medial femoral condyle corroborates previous observations in the literature[6,8,9]. Musahl et al. found that deficiency of both the lateral meniscus and the ACL led to more anterior translation of the lateral compartment during a pivot-shift maneuver than did isolated ACL deficiency[6]. Ahmed et al. reported contact between the medial femoral condyle and the tibial spine with applied internal rotation and anterior load[10]. Our data also corroborate the finding by

Fig. 6

Medial view of a representative specimen corresponding to the position and orientation of the tibia in response to combined valgus and internal rotation torques at 5° of flexion. The tibia is subluxated after sectioning of the ACL (semitransparent yellow) compared with the reduced tibial position with the ACL intact (gray).

Li et al. of a lateral shift in contact position on the medial compartment of the tibia toward the medial tibial spine in ACL-sectioned knees[13].

Sectioning the ACL increased cartilage-to-cartilage contact force carried by the medial femoral condyle in extension and by the lateral meniscus in flexion (Tables I and III). The congruency of the femoral notch and tibial spine in extension probably allowed the articular stabilizers to engage, thereby limiting tibial subluxation and preventing the increased lateral meniscus force measured in flexion.

We found that the increase in cartilage-to-cartilage contact force on the medial compartment due to ACL sectioning was 3 to 7 times greater than load differences due to condyle sectioning order. Therefore, the condyle sectioning order did not confound the conclusions of the study. Changes in force carried by cartilage-to-cartilage contact of the lateral femoral condyle were small (<5 N) and clinically irrelevant.

This study had limitations. Loads acting on the knee during high-impact sports are probably greater than those applied in this study. However, the compressive force that we applied approached half of a typical body weight and created direct cartilage-to-cartilage load transfer across the lateral compartment. Moreover, 300 N of compressive force was applied to only 5 of the tested knees. Nevertheless, sectioning of the ACL resulted in statistically significant differences in the forces carried by the lateral meniscus and via cartilage-to-cartilage contact on the medial femoral condyle. Additionally, since the pivot shift is performed in the absence of muscle activation, the role of the patellofemoral joint in stabilizing the knee is minimal; therefore, it was not loaded. Moreover, muscle forces were not included since instability episodes likely occur too quickly (~85 ms) to be resisted by voluntary muscle loading[9,34].

In conclusion, a distinct interplay of intra-articular forces occurs during the multiplanar rotatory loading that is characteristic of clinical and functional pivoting events[8,9]. Specifically, the lateral meniscus and medial femoral condyle impart, respectively, higher anterior and lateral shear forces on the tibia in an ACL-sectioned knee in response to applied torques. These intra-articular forces suggest a mechanism for

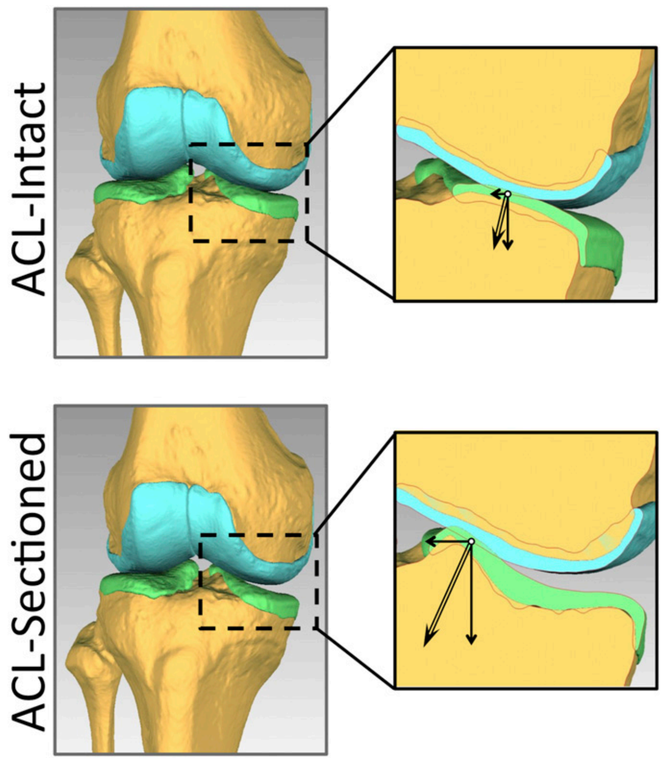

Fig. 7

Coronal view of a representative knee corresponding to the position and orientation resulting from combined valgus and internal torques in the ACL-intact and ACL-sectioned conditions at 5° of flexion. The cross-sections represent the likely location of contact between the medial tibial (green) and femoral (blue) cartilage. The plane identifying the location of contact in the ACL-intact condition is positioned more anteriorly on the tibia than the plane identifying the location of contact in the ACL-deficient condition. The double-lined arrows represent the net coronal plane forces transmitted by cartilage-to-cartilage contact from the medial femoral condyle to the tibia. The single-lined arrows are the lateral and distal components of the net force.

the clinical patterns of intra-articular derangement observed in the setting of acute and chronic ACL insufficiency, including damage of the lateral meniscus and osseous remodeling of the medial compartment. ■

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