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Changes in dynamic medial tibiofemoral contact mechanics and kinematics after injury of the anterior cruciate ligament: A cadaveric model

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Abstract

The effects of tears of the anterior cruciate ligament on knee kinematics and contact mechanics during dynamic everyday activities, such as gait, remains unclear. The objective of this study was to characterize anterior cruciate ligament-deficient knee contact mechanics and kinematics during simulated gait. Nine human cadaveric knees were each augmented with a sensor capable of measuring dynamic normal contact stresses on the tibial plateau, mounted on a load-controlled simulator, and subjected to physiological, multidirectional, dynamic loads to mimic gait. Using a mixed model with random knee identifiers, confidence intervals were constructed for contact stress before and after anterior cruciate ligament transection at two points in the gait cycle at which axial force peaked (14% and 45% of the gait cycle). Kinematic and contact mechanics changes after anterior cruciate ligament transection were highly variable across knees. Nonetheless, a statistically significant increase in contact stress in the posterior-central aspect of the medial tibial plateau at 45% of the gait cycle was identified, the location of which corresponds to the location of degenerative changes that are frequently found in patients with chronic anterior cruciate ligament injury. The variability in the contact stress in other regions of the medial plateau at 45% of the gait cycle was partly explained by the variations in osseous geometry across the nine knees tested. At 14% of gait, there was no significant change in peak contact stress after anterior cruciate ligament transection in any of the four quadrants, and none of the possible explanatory variables showed statistical significance. Understanding the variable effect of anterior cruciate ligament injury on contact mechanics based on geometric differences in osseous anatomy is of paramount clinical importance and may be invaluable to select the best reconstruction techniques and counsel patients on their individual risk of subsequent chondral degeneration.

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Keywords

Knee joint mechanics; anterior cruciate ligament; contact stresses

Introduction

Tears of the anterior cruciate ligament (ACL) are among the most common sporting injuries of the knee, with over 80,000 ACL injuries in the athletic population and 200,000 in the general population reported annually in the United States.^{1–3} These injuries are associated with an approximately US\$1 billion financial burden on the US economy each year, and the long-term effects on the health and function of this can be profound.⁴ Rupture of the ACL is associated with an increased risk of meniscal loss, joint degeneration, and post-traumatic osteoarthritis.^{5–7} While degenerative radiographic changes have been observed in over 80% of patients with long-term follow-up after ACL injury,^{8,9} the etiology of the premature degenerative changes of the knee associated with ACL injury remains unclear and may be multifactorial.

In the vast majority of cases, ACL rupture is associated with a characteristic bone contusion pattern from the inciting traumatic mechanism, with occult microfractures of the subchondral and trabecular bones.^{10–12} The overlying articular cartilage can also be injured as part of the mechanism of ACL injury, resulting in chondral loss or degeneration in up to 48% of patients.^{13,14} The role of this index impact load and secondary chondrocyte necrosis on the natural history of the ACL injury is also supported by the observation that surgical ACL reconstruction does not entirely eliminate the progression of osteoarthritic changes,¹⁵ particularly when accompanied with meniscal resection.¹⁶ The altered kinematics after ACL injury may also be of paramount importance to the development of degenerative changes. For example, while bone bruising associated with ACL rupture is typically found in the lateral compartment, medial compartment degenerative changes are more frequently observed with chronic ACL injury.^{7,17–20} This information suggests that to truly understand the etiology of damage that occurs after ACL rupture, an assessment of the changes in joint kinematics and medial compartment mechanics is required.

Numerous biomechanical studies have confirmed that the ACL is a primary stabilizer of the knee for anterior tibial translation,²¹ while it is a secondary restraint to^{22–25} tibial rotation^{26,27} and varus–valgus rotation.^{28–30} When the ACL is ruptured, the menisci act to limit excessive tibial translation and rotation,^{31,32} which can lead to concomitant damage of these structures. Van de Velde et al.³³ reported variable changes in anterior translation, internal rotation, and medial translation of the tibia in ACL-ruptured patients during the activity of lunging, which resulted in a shift in the location of peak cartilage deformation to an area of thinner cartilage more posterior and lateral on the tibial plateau. But the corresponding change in contact stress on the articular surface has not been directly measured. Without such data, it is impossible to explore the potential relationship between knee joint mechanics changes and the biological response of the articular cartilage to this injury

The objective of this study was to characterize ACL-deficient knee contact mechanics and kinematics during simulated gait.

Methods

Approval to use cadaveric specimens was granted by our Institutional Review Board.

Test apparatus

A four-station, load-controlled Instron–Stanmore KC Knee Joint Simulator (University College London, Middlesex, UK) was used to apply physiological, multi-directional, dynamic loads across cadaveric knees.³⁴ The tibia was free to move in 6 degrees of freedom, which has been shown to allow for physiological movement of cadaveric knees under gait-loading conditions.³⁵ Two parallel springs directed in the anterior–posterior (AP) direction and located on the medial and lateral sides of the tray into which the tibia was potted were used for this experiment.³⁵ The springs (14.5 N/mm) were originally intended to mimic soft tissue constraints in traditional total knee replacement wear tests.³⁵ While the cadaveric knees tested herein do have some soft tissue constraint in the form of the collateral ligaments, a pilot study suggested that the springs were nonetheless needed to avoid uncontrolled translation of the knees, particularly at the extremes of anterior and posterior motions. The base into which the tibia was potted was attached to two potentiometers (Figure 1(a)), the outputs of which were allowed for quantification of the movement of the tibial tray throughout the simulated activity of gait, specifically, the AP translations of the tibia (Figure 1(b)), the internal–external rotations of the tibia, and the location in the transverse plane of the center of rotation of the tibia³⁶ as a function of percent of the gait cycle. During testing, the femoral component rotates about a fixed, predefined flexion–extension axis, while axial force, anterior force, posterior force, and rotational moment (internal–external torque) are applied to the tibia and vary as a function of the flexion–extension angle profile applied to the femur (Figure 2).^{37,38} The inputs are based on the guidelines of the International Standards Organization (ISO #14243-1), which documents the forces required to simulate gait based on data extracted from telemetrized total knee replacement patients.³⁹ The simulator was programmed to apply 20 gait cycles at a frequency of 0.5 Hz, to ensure that steady-state contact stress data that do not vary from cycle to cycle were obtained. Data from each knee for each condition were analyzed at the last loading cycle.

Sample preparation

Using previously developed methods,^{34,40} nine fresh-frozen knees devoid of any ligamentous or meniscal injury were obtained from six cadavers. Skin, subcutaneous fat, muscle, and the patella were removed, taking care to preserve the cruciate ligaments, collateral ligaments, and capsule. The femur and tibia were transected approximately 10 cm above and below the joint line, respectively. Under fluoroscopy, a 2.5-mm Kirschner wire was drilled along the epicondylar axis. AP and lateral fluoroscopic images were obtained to confirm the accuracy of pin placement. With the knee suspended via the transepicondylar wire along the axis of rotation of the simulator, the tibia was centered in the base of the simulator and aligned such that the plateau was parallel to the ground in full extension. The

tibia–fibula complex and femur were potted into custom fixtures using poly(–methyl methacrylate) bone cement.

Contact mechanics

The normal joint contact stresses, hereafter called contact stresses, transmitted to the medial tibial plateau were measured using an array of piezoelectric stress sensing elements contained within a thin sealed sheet of plastic (4010N; Tekscan Inc., South Boston, MA, USA). Each sensor was placed between two layers of Tegaderm™ adhesive dressing (3M, Minneapolis, MN, USA) to avoid any fluid seepage into the sensor.^{34,40} Plastic augment tabs were fixed along the edges of the sensor to allow for suture fixation. Each sensor was conditioned, equilibrated, and calibrated according to manufacturer's instructions. Small, 1-cm incisions were made in the meniscotibial (coronary) ligaments anteriorly and posteriorly in line with their fibers, allowing the sensor to be passed beneath the medial meniscus and placed flush with the tibial plateau without detaching the meniscotibial ligaments, meniscomfemoral ligaments, or their capsular attachments. The sensor was secured using multiple figure-of-eight 3-0 Ethibond sutures (Ethicon, Somerville, NJ, USA) placed through the tibial ACL insertion and posteroinferior knee capsule. Sensor security was tested manually and with pre-cycling on the load-controlled simulator to assure no shift in its position within or between test conditions. The sensor was programmed to record data at 9.5 Hz.

The total force transmitted through the entire sensor for each ACL-transected knee was normalized to that of its intact condition. Using custom-written programs in MATLAB (MathWorks, Natick, MA, USA), the contact stress at each sensing element was averaged across all nine knees at 14% and then at 45% of the gait cycle—when axial forces are at their highest (see Figure 2). The sensor was then virtually divided into quadrants—anteroperipheral, antero-central, postero-peripheral, and postero-central—allowing for peak contact stress for each quadrant to be quantified.^{34,40}

Transection of ACL

Testing was conducted before and after transection of the ACL. The ACL was consistently transected proximally from its femoral footprint to simulate the most clinically observed pattern of injury and avoid any disruption to the secured stress sensor.⁴¹ Knee specimens were moistened with saline for the duration of testing to prevent tissue desiccation. All measures of normal contact stress and kinematics were applied to each knee before and after ACL transection.

Measurement of knee osseous geometry

Measurements of medial and lateral femoral condyle width and length and medial tibial plateau width and depth were made directly using digital calipers (Mitutoyo Inc, Aurora, IL, USA) with an accuracy of ± 0.01 mm. Anteroposterior and lateral knee radiographs with complete condylar overlap were obtained to measure tibial slope and confirm transepicondylar pin position. Measurements of condylar and plateau dimensions were also confirmed with radiographic measurements (Figure 3).

Statistical analysis

Mixed linear regression models with random subject identifier were developed for the mean contact stress before and after ACL transection across the entire medial plateau and for each of the four quadrants at 14% and 45% of the gait cycle. The candidate pool of explanatory variables included demographic, geometric, and kinematic explanatory variables (a total of 22 individual explanatory variables). Because of the small effective sample size, variable screening was conducted to determine the most important explanatory variables by the following two-stage procedure:

Stage 1—Within each of the three groups of explanatory variables (demographic, geometric, and kinematic), stepwise selection was used to identify those variables having greatest predictive power.

Stage 2—A nonparametric Gaussian process model was fit to identify possible interactions from those variables selected in Stage 1 based on two-way total sensitivity indices. A stepwise regression was performed using the Stage 1 candidate variables and potential interactions to identify the important variables and their interactions.

Confidence intervals were constructed for the mean change in contact stress for each of the four quadrants at 14% and 45% of the gait cycle. The statistical analysis was performed using R and JMP statistical software (SAS Institute, Inc., Cary, NC, USA).

Results

Knee kinematics

The magnitude of the change in internal/external rotation after ACL transection was highly variable across specimens, with changes in rotation of the tibia up to 6° in each direction (Figure 4(a)). After ACL transection, the knees tended to rotate about a location that was more medial than when the ACL was intact (Figure 4(b)). The change in AP translation was also highly variable after ACL transection (Figure 4(c)). Due to interspecimen variability, there was no statistically significant effect of ACL transection on knee kinematics at 14% and 45% of the gait cycle (Table 1).

Contact stresses across the medial tibial plateau

Presentation of the data—contact stress data from a sample knee before and after ACL resection—is illustrated in Figure 5. The change in peak contact stress data from each knee after ACL transection at 14% and 45% of the gait cycle is presented in Figure 6, while the results from the statistical analysis are presented in Tables 2 and 3.

At 14% of the gait cycle, there was no significant change in peak contact stress after ACL transection in any of the four quadrants (Figure 6(a)). None of the possible explanatory variables showed statistical significance (see Table 2).

At 45% of the gait cycle, increases in contact stresses in the posterior–central quadrant of the tibial plateau were seen in 7/9 knees (Figure 6(b)). Because the magnitude of the increase varied considerably from knee to knee, the overall change in the *mean* quadrant

contact stress was not statistically significant. However, there was a statistically significant increase in the *peak* contact stress in the postero-central quadrant (Table 3). None of the potential explanatory variables showed statistical significance, suggesting that regardless of the changes in kinematics subsequent to ACL rupture and variations in knee geometry, increased contact stresses, although of a variable magnitude, occur in this area.

At 45% of the gait cycle, there was not a statistically significant difference in the overall change in the mean contact stress for the anterior quadrant of the tibial plateau after ACL rupture (Figure 6(a) and (b)). But the magnitude of change of the mean *peak* contact stress in the anterocentral quadrant showed a strong relationship to several of the geometric predictors (Table 4). In particular, decreased tibial medial concavity, higher tibial slope, and smaller changes in the location of the center of rotation were predictive of higher anterocentral quadrant stresses (Figure 7(a)–(c)). Additionally, at 45% of the gait cycle, decreased lateral condyle lengths are predictive of higher mean *peak* stresses in the anteropipheral quadrant of the tibial plateau after ACL rupture.

Discussion

The objective of this study was to characterize ACL-deficient knee contact mechanics during simulated gait. By way of a novel, dynamic, multidirectional in vitro cadaveric model, we found that the change in rotation and translation after ACL transection at two points in the gait cycle (14% and 45% of gait) was highly variable between knees. While the changes in contact stress were also variable between knees, there was a statistically significant increase in contact stress in the postero-central aspect of the medial tibial plateau at 45% of the gait cycle. Decreased tibial medial concavity, increased tibial slopes, and smaller changes in the location of the center of rotation were predictive of higher stresses in the anterocentral region of the tibial plateau, whereas shorter lateral condylar lengths were also predictive of higher anteropipheral quadrant stresses. This study provides novel information about the changes in contact mechanics and kinematics that occur after ACL rupture in human knees during the activity of walking. Understanding the variable effect of ACL injury on the contact mechanics based on differences in osseous anatomy may be invaluable to counsel patients on their individual risk of subsequent chondral degeneration.

Understanding the functional role of the ACL and the consequences of its rupture on joint contact mechanics is not trivial. In vivo gait analysis studies can allow for joint kinematics to be studied as patients perform everyday activities;^{42–44} using dual fluoroscopic techniques, cartilage deformation during these activities can be measured,⁴⁵ but a direct measurement of contact mechanics before and after ACL rupture in patients is not possible. Experimental models have been developed to study the effect of ACL rupture on joint kinematics under static or quasi-static loads,^{29,46,47} but given the viscoelastic nature of the soft tissue of the knee, such static models provide only limited information about the true effect of ACL rupture. Moreover, the loads applied to these models are often of a lower magnitude than those expected clinically and are often simplified to incorporate only a single loading direction. The human knee experiences large, complex, multidirectional, and dynamically varying forces during everyday activities; therefore, to truly study the effect of ACL rupture on joint contact mechanics requires a model that captures some of this

complexity. The model herein described allows for the application of physiological, dynamically varying axial forces, anteroposterio forces, and internal–external moments, which vary as a function of flexion angle. The model has been previously used to study the effect of meniscal injury and repair on joint contact mechanics;^{34,40} however, its use in the study of ACL injury has not been explored thus far.

An important finding of our multidirectional, physiological model of ACL rupture was an increase in contact stresses in the posterior aspect of the tibial plateau, although the magnitude of that increase varied significantly from knee to knee. This variability is not surprising given the variability seen clinically.³³ Many clinical studies have reported variable functional instability and pivot shift after ACL injury. Injury to the secondary stabilizers of anterior translation of the lateral compartment, including the lateral meniscus, anterolateral capsule, and iliotibial (IT) band, may contribute to the severity of the pivot shift in the ACL-deficient knee. Furthermore, variable morphology of the lateral tibial plateau, including increased posteroinferior tibial slope and small size, can also contribute to magnitude of pivot shift and change in contact mechanics after ACL injury.^{32,34,48} Nonetheless, as the tibial translates more anteriorly after ACL rupture,⁴³ the location of femoral contact along the tibial surface would be expected to shift posteriorly relative to the uninjured knee. While the magnitude of change in contact stress that can lead to cartilage damage in vivo is unknown, it has been hypothesized that abnormal knee kinematics in the setting of ACL insufficiency can lead to a shift in joint contact regions and cause damage to thinner regions of articular cartilage that are not accustomed to higher or frequent loading,⁴⁹ thereby contributing to an accelerated progression to osteoarthritis. The change in contact stresses on articular cartilage before and after ACL rupture during the activity of walking has not been quantified, which means that the connection between the progression of osteoarthritis and the stresses seen by the articular cartilage cannot be made. The data generated in this study indicate that while the magnitude of change in contact stress that occurs after ACL rupture is highly variable, it was nonetheless significantly higher in the posterior–central area of the tibial plateau. Medial compartment degenerative changes that are frequently observed with chronic ACL often manifest as damage to the posterior aspect of the meniscus,^{7,17–20} corresponding to the area in which increased stresses were quantified in this study (Figure 5).

Changes in the contact mechanics across other areas of the tibia were more variable (Figure 8). The contact stress in the anterior region of the medial plateau, for example, either increased or decreased after ACL transection, depending on the knee tested. In an attempt to understand the source of the variability, a mixed model with random knee identifiers was used to assess the relative influence of geometric, kinematic, and demographic variables. The osseous morphologies of the tibial plateau and femoral condyles were identified as variables that influenced the post-ACL rupture contact mechanics in this zone. Specifically, decreased concavity of the medial tibial plateau and increased tibial slope were predictive of higher antero-central quadrant stresses. Knee stability is controlled by osseous geometry and ligamentous stiffness; decreased tibial concavity and increased tibial slope would be expected to decrease the contribution of the osseous geometry to stability, thereby placing more of a reliance on the soft tissue envelope and ligaments.⁵⁰ In our study, decreased tibial concavity and increased tibial slope combined to result in ACL rupture having a more

diffuse effect on contact mechanics, resulting in knees that had an increase in contact stress not only in the posterior aspect of the tibia but also in the anterior aspect. Indeed, while increased tibial slope has been identified as a predictor of predisposition to ACL rupture,⁵¹ and a factor that increased the post-ACL pivot-shift test results,⁵² our study is the first to suggest that patients may be variably predisposed to an unfavorable knee mechanics after ACL injury manifested by a more diffuse increase in contact stress across the tibial plateau.

Herfat et al.⁵³ reported increased anterior translation, internal rotation, and medial translation of the tibia after ACL rupture during gait.³³ This study not only supports these findings but also identifies significant knee-to-knee variability with regard to the impact of ACL injury on joint kinematics and peak medial compartment contact stresses; this result suggests that some knees are inherently stable, even without an ACL (consider knees 8 and 9 in Figure 4).

This study is not without limitations. Contact mechanics was assessed only on the medial tibial plateau, and the analysis was limited to two parts of the gait cycle (14% and 45% of gait cycle) during which axial force is at a peak. While the tibia is free to move in the medial–lateral direction and in varus–valgus orientation, the forces in these directions are not controlled. Furthermore, while gait was modeled in this study, other functional activities—stair climbing, running, and pivoting activities—may offer greater insight into altered contact mechanics after ACL injury. A prospective approach with groups stratified by specific geometric variable is needed to further characterize preliminary findings in this study. Finally, while a statistically significant increase in contact stress was seen in the posterior–central aspect of the tibial plateau at 45% of the gait cycle, the magnitude of the increases was highly variable between knees, the clinical significance of which is as yet unclear.

In conclusion, our study suggests that increased stresses in the posterior aspect of the medial tibial plateau at 45% of the gait cycle are a commonality among all ACL-ruptured knees, but to understand the variable effect of ACL injury on the contact mechanics of the anterior portion of the plateau requires an assessment of osseous anatomy. ACL reconstruction may be more selectively indicated for those patients with functional knee instability and an “at-risk” knee with osseous geometry that is predisposed to progressive joint injury and degeneration in the setting of ACL insufficiency. Future studies that define the variable effect of ACL injury on the contact mechanics-based specific differences in osseous anatomy will be invaluable to counsel patients on their individual risk of subsequent chondral degeneration.

Conclusion

After ACL transection, increased contact stresses at 45% of the gait cycle were found in the posterior aspect of the medial tibial plateau, which corresponds to the location of degenerative changes that are frequently found in patients with chronic ACL injury. The magnitude of changes in contact stress in other regions of the knee is associated with specific geometric features of the individual knee: tibial concavity, tibial slope, and condylar length.

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Figure 1.

(a) Human knee as mounted on the simulator. The epicondylar axis is aligned with the flexion/extension axis of rotation of the simulator. Two potentiometers are attached to the tibial tray to allow for the kinematics of the tibia to be quantified throughout gait. (b) Sample output as calculated from the potentiometers for tibial translation as a function of percent of gait for an intact and ACL-transected knee. The change in translation is calculated at 14% and 45% of gait cycles.

ACL: anterior cruciate ligament.

Figure 2.

Inputs to the simulator for a complete gait cycle. Flexion angle is controlled by flexing the femur; axial force, internal/external torque, and anterior–posterior force are applied through the tibia.

Figure 3.

The following measurements were made on each knee using digital calipers: the medial and lateral femoral width and length, and the medial tibial plateau width and depth. Using anteroposterior and lateral knee radiographs, tibial slope was quantified. Courtesy of Hongsheng Wang PHD.

Figure 4.

(a) Change in internal/external rotation of the tibial plateau after ACL transection at 14% and 45% of gait. All knees demonstrated an increase in rotation after ACL transection, but the magnitude of increase was variable. (b) Change in location of the center of rotation on the tibial tray after ACL transection at 14% and 45% of gait, with more knees rotating about a location that was more medial than before ACL transection. (c) Change in the anterior-posterior translation of the tibial tray at 14% and 45% of gait.

ACL: anterior cruciate ligament.

Figure 5.

Contact stress plot from a representative knee demonstrating an increase in contact stress in the postero-central aspect of the medial tibial plateau.

ACL: anterior cruciate ligament.

Figure 6.

Change in contact stress as a function of ACL rupture for each knee tested at (a) 14% of gait and (b) 45% of gait.

ACL: anterior cruciate ligament.

Figure 7.

A graphical representation of the effect of osseous geometry on the change in contact stresses in the antero-central quadrant after anterior cruciate ligament transection at 45% of the gait cycle. Change in peak contact stress (Figure 6(b)) is plotted against tibial slope and medial tibial concavity in all graphs, but the center of rotation is fixed at one of the three values: (a) median value for change in center of rotation, (b) 25th percentile value for change in center of rotation, and (c) 75th percentile value for change in center of rotation.

Figure 8.

Axial view of tibial plateau demonstrating regions of increased contact stresses observed after anterior cruciate ligament (ACL) rupture. While there is an increase in postero-central contact stresses after ACL injury independent of osseous geometry, relative increases in anterior and central plateau stresses were observed with increased slope and decreased concavity.

Table 1

Estimated mean differences and 95% confidence intervals for change in internal–external rotation, location in center of rotation, and anterior–posterior translation after ACL transection.

Variable	Estimated mean	95% confidence interval
Internal–external rotation		
14% of gait	−0.288	(−3.70, 3.13)
45% of gait	−0.556	(−1.68, 0.57)
Location of center of rotation		
14% of gait	21.896	(−21.90, 65.69)
45% of gait	0.26	(−2.31, 2.83)
Anterior–posterior translation		
14% of gait	1.001	(−0.38, 2.38)
45% of gait	0.132	(−0.74, 1.01)

All confidence intervals span zero, thus differences are not significant.

Table 2

Estimated mean differences, estimated standard errors, and 95% confidence intervals for quadrant peak contact stresses at 14% of gait (constant mean model).

Quadrant	Estimated mean	Estimated standard error	95% confidence interval
Anterior–central	−0.194	0.187	(−0.652, 0.264)
Anterior–peripheral	0.198	0.265	(−0.449, 0.846)
Posterior–central	1.314	0.636	(−0.242, 2.870)
Posterior–peripheral	0.409	0.254	(−0.211, 1.030)

All confidence intervals span zero, thus differences are not significant.

Table 3

Estimated mean differences, estimated standard errors, and 95% confidence intervals for posterior quadrant pressures at 45% of gait (constant mean model).

Quadrant	Estimated mean	Estimated standard error	95% confidence interval
Posterior–central	0.677	0.244	(0.082, 1.271)
Posterior–peripheral	0.643	0.512	(–0.609, 1.896)

The confidence interval for the posterior–central quadrant does not span zero indicating that the differences in peak contact stress after anterior cruciate ligament transection were significantly higher than that of the intact condition.

Table 4

The change in contact stress in the peak contact stress in the anterior quadrants at 45% of the gait cycle showed a strong relationship with several geometric variables, as shown below.

	Estimated effect	Estimated standard error	<i>p</i> value
Anterior–central			
(Intercept)	−0.1893	0.242	0.4697
Tibial medial concavity	−0.0488	0.0005	0.007
Tibial slope	0.1261	0.00014	<0.005
Change in center of rotation (45°)	−0.0616	0.00031	<0.005
Anterior–peripheral			
(Intercept)	385.225	12.8629	<0.005
Condyle lateral length	−6.073	0.1852	0.0194