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Loading of the Medial Meniscus in the ACL deficient knee: a Multibody Computational Study

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Abstract

The menisci of the knee reduce tibiofemoral contact pressures and aid in knee lubrication and nourishment. Meniscal injury occurs in half of knees sustaining anterior cruciate ligament injury and the vast majority of tears in the medial meniscus transpire in the posterior horn region. In this study, computational multibody models of the knee were derived from medical images and passive leg motion for two female subjects. The models were validated against experimental measures available in the literature and then used to evaluate medial meniscus contact force and internal hoop tension. The models predicted that the loss of anterior cruciate ligament (ACL) constraint increased contact and hoop forces in the medial menisci by a factor of 4 when a 100 N anterior tibial force was applied. Contact forces were concentrated in the posterior horn and hoop forces were also greater in this region. No differences were found in contact or hoop tension between the intact and ACL deficient (ACLd) knees when only a 5 Nm external tibial torque was applied about the long axis of the tibia. Combining a 100 N anterior tibial force and a 5 Nm external tibial torque increased posterior horn contact and hoop forces, even in the intact knee. The results of this study show that the posterior horn region of the medial meniscus experiences higher contact forces and hoop tension, making this region more susceptible to injury, especially with the loss of anterior tibia motion constraint provided by the ACL. The contribution of the dMCL in constraining posterior medial meniscus motion, at the cost of higher posterior horn hoop tension, is also demonstrated.

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Conflict of Interest

The authors have no conflict of interest to report.

Ethical Approval

Both subjects involved in this study provided written informed consent approved by the University of Missouri - Kansas City's Institutional Review Board (IRB #SS11-27e).

Keywords

menisci; anterior cruciate ligament; medial meniscus; medial collateral ligament; computational biomechanics

1.0 Introduction

The menisci of the knee are critical to knee function and health. The menisci redistribute contact forces, reducing tibiofemoral contact pressures, and aid in knee lubrication and nourishment. Injury to the anterior cruciate ligament (ACL) of the knee is common [1] and it has been estimated that up to 50% of ACL injuries include meniscal tearing [2]. In the medial meniscus, over 99% of these tears occur in the posterior region [3]. Tearing of the menisci compromises its mechanical integrity, reducing the ability of the tissue to redistribute mechanical loading [4–6]. Meniscectomy [7] and meniscal pathology [8] following ACL injury have been linked to cartilage degeneration and the development of post-traumatic osteoarthritis.

The peripheral tibiomeniscal attachments, specifically the attachments of the deep medial collateral ligament (dMCL), influence medial meniscus biomechanics, but the extent of this influence on meniscus mobility and cartilage contact has not been adequately addressed in the literature [9]. The superficial and deep layers of the medial collateral ligament are separated by a bursa [10], biomechanically separating the layers. Anatomical studies of the dMCL have separated it into two functional units, the meniscomfemoral and meniscotibial divisions, with stout attachments to the meniscus described [11]. But, in a recent study, Stein et al. did not measure any differences in tibial plateau stress after transecting the dMCL meniscal attachments on two cadaver knees loaded with a 500 N axial force, concluding that the dMCL did not affect the stability of the medial meniscus [12].

In their cadaver study, Markolf et al. found that anterior tibial force and external tibial rotation torque about the long axis of the tibia were loading modes that produced higher forces on the posterior horn attachments of the medial meniscus [13]. Markolf et al. presumed that these higher horn attachment forces were the result of the medial femoral condyle pushing against the posterior horn of the meniscus. Despite the known association between ACL injury, posterior medial meniscal tears, and development of post-traumatic osteoarthritis, information on the mechanisms of medial menisci loading under combined translational and rotational knee forces in the normal and ACL deficient knee is lacking in the literature. Greater understanding of the biomechanics of the medial meniscus and its attachments under these dynamic loading conditions may assist in meniscal injury prevention, rehabilitation, and surgical repair.

The aim of this computational study was to assess loading on the medial meniscus resulting from tibial anterior forces and external rotation torques in the intact and ACL deficient (ACLd) knee. Specifically, alterations in meniscotibio contact forces and internal meniscus hoop forces for intact and ACLd knees during combined loading scenarios were quantified via computational models derived from medical images and passive leg motion acquired for two female subjects. The computational models were used to evaluate the hypothesis that

loss of ACL constraint will increase loading on the posterior region of the medial meniscus and that in the ACLd knee, the dMCL peripheral meniscus attachments help restrain medial meniscus motion, which in turn limits motion of the femur relative to the tibia.

2.0 Methods

2.1 Subjects

Subject specific computational knee models were created from magnetic resonance images (MRI) and passive leg motion measurements for two adult females (Subject 1: age 20 years, height 159.5 cm, mass 59.0 kg, Subject 2: age 29 years, height 170 cm, mass 70.3 kg). Both subjects provided written informed consent approved by the institution's human subjects review board (IRB #SS11-27e).

2.2 Experimental Measurements

Prior to data collection, the subjects were fitted with custom shells that included two orthogonal mustard filled tubes visible during MRI. MRIs were acquired for each subject's right leg using a 3 Tesla machine with 0.5 mm spacing and 512×512 resolution. The scans were converted to geometries of bone, cartilage and menisci using the software program 3D-Slicer (www.slicer.org) and processed using Geomagic Studio (Geomagic, Inc., Research Triangle Park, NC, USA). Processing included smoothing and decimation and each processed geometry was overlaid on the original MRI scans to ensure that articulating surfaces were not altered. Reflective markers were placed on the localizing shells and shin and leg motion during passive movement of the knee was recorded using a motion capture system (Vicon Motion Systems Ltd., UK). Measured marker motion was filtered using a 6 Hz low-pass filter. During the passive motion measurements, an orthopaedic surgeon gripped the distal posterior femur and manipulated the lower leg of each subject through the knee's passive flexion-extension, passive abduction-adduction, and passive internal-external range of motion. The surgeon applied as little force as possible to the lower leg throughout the procedure and the subjects laid on their backs on an examination table and were instructed to relax their leg muscles.

2.3 Computational Model

Rigid body dynamics (ADAMS, MSC Software Corporation, Santa Ana, CA) were used to model the right knee of each subject with deformable contacts defined between articulating geometries (Fig. 1). Contact force was a function of penetration between contacting geometries and was defined as:

$$F_c = k\delta^n + B(\delta)\dot{\delta} \quad (1)$$

where F_c is the contact force, δ is the interpenetration of the contacting geometries, $\dot{\delta}$ is the velocity of interpenetration, k is the contact stiffness, n is the nonlinear power exponent, and $B(\delta)$ is a damping coefficient. Tibiofemoral contact values were derived from a previous study that compared contact predictions to an identically loaded finite element analysis [14] and were: $k = 327 \text{ N/mm}^n$, $n = 2.07$, and $B = 5 \text{ Ns/mm}$.

The ligaments were attached to the bones according to attachment sites identified in the MRI and in the literature. The force-length relationship of each ligament bundle was modeled as a tension only nonlinear spring using a piecewise function [15] defined as:

$$\varepsilon = (l - l_0) / l_0 \quad (2)$$

$$f = \begin{cases} \frac{1}{4} k \varepsilon^2 / \varepsilon_l & 0 \leq \varepsilon \leq 2\varepsilon_l \\ k(\varepsilon - \varepsilon_l) & \varepsilon > 2\varepsilon_l \\ 0 & \varepsilon < 0 \end{cases} \quad (3)$$

where the ligament strain ε of Eq. 2 is a function of ligament length l and the zero-load length l_0 (length at which a ligament bundle begins to carry force). To model the toe region, the ligament force-length relationship is quadratic for strains less than $2\varepsilon_l$, where ε_l was assumed to be 0.03, and linear for strain values above $2\varepsilon_l$ (Eq. 3). Ligament stiffness k was unique to each bundle (Table 1) and obtained from the literature [15–20].

The anterior cruciate ligament (ACL) and posterior cruciate ligament (PCL) were both separated into two bundles [21, 22]. The lateral collateral ligament was separated into three bundles [23] and the medial collateral ligament was separated into five bundles, three superficial (sMCL) and two deep bundles (dMCL) [24–26]. The sMCL has been described as having one femoral and two tibial attachments, dividing the ligament into proximal and distal sections [27]. The proximal tibial attachment is connected to soft tissue and the distal tibial attachment is connected directly to bone. In the model, the proximal sMCL tibial attachment was achieved via three-axis springs that were stiffer along an axis directed towards the bone (500 N/mm), preventing penetration of the ligament into the bone, and compliant (2 N/mm) along the bone surface allowing the attachment site to move relative to the tibia. The sMCL proximal tibial attachment stiffness values were determined by comparing the force ratio between the proximal and distal divisions of the sMCL to experimental force measurements [28] for similar motion and loading. The central (tibial) arm of the posterior oblique ligament (POL) was modeled as one bundle [27] as previous studies have described the central arm as the primary component of the POL [29]. Stiffness values of the POL central arm were derived from a cadaver based study by Wijdicks et al. [20]. The anterolateral ligament (ALL) was present in 97% of cadaver knees examined by Claes et al. [30] and in 100% of cadaver knees examined by Kennedy et al. [31]. The ALL was modeled as one bundle with femoral and tibial attachment sites from Claes et al. [30] and stiffness values from the cadaver study of Kennedy et al. [31].

The menisci were modeled as previously described [14, 32] and were comprised of multiple rigid bodies connected by stiffness matrices. The menisci geometries were sectioned into multiple wedge shaped rigid bodies connected by 6×6 stiffness matrices producing forces and torques that are a function of relative displacement and rotation between adjacent meniscus bodies. The medial and lateral menisci were sectioned at 10° intervals radiating from the geometric center of each meniscus. For subject 1, the medial meniscus was divided

into 19 rigid bodies and the lateral meniscus was divided into 25 rigid bodies. For subject 2, the medial meniscus had 20 rigid bodies and the lateral meniscus 25 bodies. Parameters for the stiffness matrices were modified from previous work [14] for the meniscus geometries and sectioning used in the current study. Mass properties were assigned to each meniscus element based on its volume and a density of $1,100 \text{ kg/m}^3$. Deformable contacts were defined between each meniscus element and tibial cartilage and femoral cartilage geometries using Eq. 1 and the values $k = 19 \text{ N/mm}^n$, $n = 3.37$, and $B = 0.1 \text{ Ns/mm}$ [14]. Values for the stiffness matrices connecting the meniscus elements as well as contact model parameters were derived through optimizations that matched displacement and forces from a finite element solution [14]. The hoop force between elements was calculated using the spring force orientated in the circumferential direction of the medial menisci.

The menisci were connected to the tibia via four horn attachments modeled as non-linear elastic elements (Eqs. 2 and 3). The origin and insertion sites for the horns were derived from MRI and from the literature [33–35]. The zero-load length of each horn attachment was assumed to be 96% of the tibial origin to meniscus insertion length in the MRI position. The stiffness of each meniscal tibial attachment was derived from Hauch et al. [36] (Table 1). The anterior intermeniscal ligament (AIML) was included and assumed to have attachments at the anterior horn of the medial meniscus and anterior margin of the lateral meniscus [37]. A stiffness of 750 N was used with a zero-load length derived from the MRI position. A sensitivity analysis of AIML parameters revealed that neither the stiffness nor the zero-load length affected the conclusions of this study.

The two bundles of the dMCL were connected to the medial menisci via three-axis springs, each with a stiffness of 40 N/mm acting in three orthogonal axes. This stiffness was chosen as it is equivalent to the stiffness of dMCL bundles in the linear region (i.e. 1500 N divided by the average dMCL zero-load length). The springs attached at a node on either the anterior bundle of the dMCL or posterior dMCL and the closest medial meniscus element in the MRI position. Simulations were run with attachment spring stiffness parameters set at 40 N/mm and 400 N/mm with no noticeable change in meniscus motion, internal forces, or contact force.

The localizing shells worn by each subject were used to register the femur and tibia to the motion capture coordinate system. Motion constraints driven by motion capture marker trajectories from the passive leg motion measurements were used in initial simulations that determined the zero-load lengths of each ligament bundle. Both knee models were moved through their measured passive motion while the zero-load lengths of each ligament bundle were systematically adjusted based on the following assumptions: 1) each ligament bundle produced force somewhere within the passive motion measurements, and 2) the force produced by any ligament bundle should be less than 50 N. The 50 N threshold for passive ligament bundle force was chosen because it is within the nonlinear “toe region” [17] for all ligaments and because experimentally measured passive flexion forces for cruciate ligament bundles was under 50 N [38, 39]. Initial ligament zero-load lengths came from MRI measurements where the knee was slightly flexed. An iterative bi-section root finding optimization method was then used to iteratively adjust the zero-load lengths of each ligament bundle. The passive motion simulations were 115 seconds long and included

multiple flexion-extension cycles that ranged from 2° to 130° for subject 1 and 8° to 148° for subject 2.

After determination of ligament zero-load lengths, the models were modified such that the femur was held fixed and flexion-extension motion was applied to the tibia. The tibia was unconstrained in internal-external rotation, varus-valgus, compression-distraction, anterior-posterior translation, and medial-lateral translation. In subsequent simulations, axial force, anterior-posterior forces, and an external torque were applied along or about their respective tibial axes of an embedded local coordinate system defined using anatomical features (Grood and Suntay [40]). The compressive axial force was applied along the Grood and Suntay local tibial z-axis and moved with the tibia. The external torque was applied about the tibial z-axis, defined as the mechanical axis of the tibia that passes between the intercondylar eminences and through the ankle center [40]. Anterior and posterior tibial forces were applied along the local tibial Grood and Suntay y-axis and moved with the tibia.

2.4 Model Validation

For the purpose of model validation, model predictions were compared to experimental values available in the literature for the main knee structures evaluated in this study. Compared were cruciate ligament forces, dMCL length patterns, and medial meniscus excursion. Model predicted anterior cruciate ligament forces during passive flexion-extension and with a 100 N anterior force applied (Fig. 2a) and posterior cruciate ligament forces during passive flexion-extension and with a 100 N posterior force applied (Fig. 2b) were compared to experimental measurements of Markolf et al. [38, 39]. In Markolf's experimental cadaver studies, the tibial (ACL) or femoral (PCL) attachment sites were mechanically isolated and "a cap of bone containing the ligament fibers was attached to a load cell" [38], allowing direct measurement of ligament force. Model predicted ligament forces followed the average ($n = 12$, ACL, and $n = 16$, PCL) experimental values over the flexion-extension range for passive motion and with applied anterior or posterior tibial forces. Experimentally measured forces for the deep bundles of the MCL are not available in the literature, but elongation patterns of the dMCL from extension to deep flexion were measured by Hosseini et al. [41] using a dual fluoroscopic imaging system. The model predicted percent changes in length, relative to terminal extension length, for the posterior and anterior bundles of the dMCL generally follow the average ($n = 8$) experimentally measured elongation patterns during passive flexion-extension (Fig. 3). Finally, the anterior-posterior excursions of the lateral and medial menisci were compared to *in vivo* meniscus excursion values measured by Yao et al. [42] (Table 2). For both subjects, model predicted motion of the middle meniscus element during flexion-extension was within 1 standard deviation of MRI measured centroid motions from extension to deep flexion.

2.5 Simulations

Following validation of the models through comparison of model predictions to experimental values, the models were used to evaluate medial menisci loading resulting from application of translational and rotational tibial forces. For these evaluations, the following simulations were used.

1. **Ext** – Intact knee model using passive flexion-extension motion to 90° + a 300 N compressive force + a 5 Nm external tibial torque
2. **ACLd Ext** – Ext simulation with both bundles of the ACL disabled
3. **Ant** – Intact knee model using passive flexion-extension motion to 90° + a 300 N compressive force + a 100 N anterior tibial force
4. **ACLd Ant** – Ant simulation with both bundles of the ACL disabled
5. **Ext Ant** – Intact knee model using passive flexion-extension motion to 90° + a 300 N compressive force + a 5 Nm external tibial torque + a 100 N anterior tibial force
6. **ACLd Ext Ant** – Ext Ant with both bundles of the ACL disabled

3.0 Results

Total contact force between the medial menisci and tibial cartilage typically increased with flexion angle from extension to approximately 30° knee flexion angle for both subjects (Fig. 4). Contact force typically decreased with flexion angle after approximately 40° flexion. Tibiomeniscal contact force increased fourfold in the ACLd knees compared to intact knees when a 100 N anterior force was applied to the tibia. Application of a 5 Nm external torque increased tibiomeniscal contact force compared to 100 N anterior force for the intact knee. There was no difference in contact force between the intact and ACLd knee for external rotation torque. Combined loading of anterior force and external torque provided the greatest tibiomeniscal contact forces for the intact knee. These forces increased in the ACLd knees. For subject 1, contact forces were similar in the ACLd knee for anterior force only and combined anterior and external rotation loading.

Hoop forces between meniscus elements followed the trends of tibiomeniscal contact forces (Figs. 5 and 6). Hoop force was greater between elements posterior to the dMCL meniscal attachments for all simulations with ACLd. In the intact knee, hoop forces posterior to the dMCL attachments were greater than anterior elements for external loading and combined anterior and exterior rotation loading. There was no difference in hoop forces between the anterior and posterior horn regions when only an anterior force was applied for the intact knee.

For the three simulations that created the greatest tibiomeniscal contact forces (ACLd Ant, ext ant, and ACLd ext ant), contact occurred predominately on the posterior horn of the medial meniscus with little or no contact occurring on the anterior horn (Fig. 7). The location of meniscus elements experiencing high contact forces shifted to elements closer to the posterior horn attachments (element 19 for subject 1, and element 20 for subject 2) when external torque was added to the simulation, particularly for subject 1 (Fig. 7a).

4.0 Discussion

Computational multibody models of the knee were derived from medical images and passive leg motion for two female subjects. The models were validated against experimental

measures available in the literature and then used to evaluate medial meniscus contact and internal hoop forces. The computational models of both subjects predicted that the loss of ACL constraint increased contact and internal hoop forces in the medial menisci by a factor of 4 when a 100 N anterior tibial force was applied. Contact forces were concentrated in the posterior horn and hoop forces were greater in the posterior horn region between the dMCL meniscal attachments and the posterior horn attachments. No differences were found in contact or hoop forces between the intact and ACLd knees when only a 5 Nm external tibial torque was applied. Combining the 100 N anterior tibial force and the 5 Nm external tibial torque generated high contact and hoop forces in the posterior horn, even in the intact knee. Removing the constraint of the ACL increased posterior horn forces for combined anterior and external loading. In the ACLd knee, the location of meniscotibial contact changed when external loading was applied, moving to elements closer to the posterior horn attachment.

Anterior translation and external rotation of the tibia pulls the posterior horn region of the medial meniscus into the posterior medial femoral condyle. In the intact knee, the ACL constrains anterior motion of the tibia, preventing the posterior horn from being “pulled” into the femur. Loss of this constraint allows the tibia to translate anteriorly relative to the femur, causing the posterior horn of the medial meniscus to act as a barrier to anterior tibial motion. The ACL does not constrain external tibial rotation, and hence no difference was seen in meniscal loading between the intact and ACLd knees for the external torque only loading case. In the intact knee, applying external torque doubled meniscal loading on the posterior horn and combined anterior force and external torque tripled meniscal loading compared to anterior force loading. Application of the external torque externally rotated the tibia, decreasing ACL length and allowing greater anterior translation of the tibia when the anterior load was applied. This increased anterior translation of the tibia for the combined loading case accounted for the increase in meniscal loading.

This study found that the posterior horn of the medial meniscus is the region that experiences the greatest contact forces. This finding is in agreement with the location of meniscal tearing, which overwhelmingly occurs in the posterior medial meniscus. Based on analysis of 1,485 meniscal tears in individuals with stable knees, Metcalf and Barrett [43] found that 98% of tears on the medial side involved the posterior horn. Similarly, Terzidis et al. [44] found that 93.1% of medial meniscus tears involved the posterior horn in the uninjured knee. Medial meniscus tearing is associated with ACL injury. In an ACLd population, Smith and Barrett (Smith and Barrett, 2001) found 99.4% of tears on the medial meniscus were in the posterior region.

The results of this study are also in agreement with a cadaver study that measured loading on the meniscal horn attachments. Markolf et al. found that anterior tibial force and external tibial torque were loading modes that produced higher forces on the posterior horn attachments of the medial meniscus and that anterior tibial force in the ACLd was the worst case [13]. Also, similar to our study, Markolf et al. reported that when an external tibial torque was applied, poster horn attachment force remained unchanged after removal of the ACL [13].

Studies specific to the influence of the dMCL peripheral attachments on meniscus biomechanics are rare. In their experimental study on two cadavers, Stein et al. [12] concluded that there was “no relevant influence of the medial collateral ligament on the stability of the medial meniscus”. The Stein study applied a 500 N compressive load to intact knees and after detaching the medial collateral ligament from the femur. No difference in meniscus deformation or movement was found for the two cases. The results of the current computational study refute the claim of Stein et al. that the meniscal attachments of the dMCL are not significant to meniscal biomechanics. The Stein study only applied a compressive load with no torsional or anterior forces and the meniscotibial division was left intact. The computational models presented in the current study found that the dMCL meniscal attachments do play a significant role in tibiomeniscomfemoral biomechanics when translational and rotational forces are applied to the tibia.

The tibial compressive force used in this study, 300N, is approximately the average of one half body weight for the two subjects. The main focus of the study was on the effect of external tibia rotation and anterior tibia translation on the medial meniscus in the ACL deficient knee. Activities of daily living, such as walking, would have higher compressive forces which would alter the values of this study, but would not alter the relationships and conclusions of the study. The external torque of 5 Nm was chosen as it is a common value used in experimental cadaver studies examining tibiomeniscomfemoral biomechanics [13, 45–47] with similar reasons for the selection of the anterior tibia force of 100 N [39].

For this study bone motion was not directly measured; only the motion of surface markers was tracked. The experimental motion markers were located on rigid shells tightly attached to the leg and directly on the anteromedial surface of the shin. But, it is expected that the motion markers may move relative to the femur and tibia bone during the passive motion measurements. The passive tibia and femur motion was acquired using in vivo motion capture and MRI, equipment that is more broadly available than the fluoroscopy systems typically required for direct measurement of bone motion. For this study, forces applied to the knee by the examiner were not measured. The surgeon was experienced in knee manipulation, gripped the distal posterior femur while moving the tibia, and was given instructions to apply as little force as possible to the tibia during the movement. But, bone motion was not directly measured and this may affect estimates of zero-load lengths for each individual knee, which will in turn affect ligament force predictions. Previous studies have determined that the most significant factor in ligament biomechanical function is the reference strain, or zero-load length [48–50].

The models presented here used bone, cartilage, and menisci geometries obtained from individual MRIs, but ligament bundles were represented as single line elements and ligament stiffnesses came from the literature. Wrapping of the superficial MCL around the tibia was modeled; wrapping of other ligaments and meniscal horn attachments was not modeled. In addition, passive muscle forces and the joint capsule were not included in this study.

The menisci are mobile and deformable structures which are typically modeled using the finite element method. In our models, the menisci are represented as multiple rigid bodies connected by 6×6 stiffness matrices. Deformation of the meniscus body is only allowed

between individual meniscus elements. For identical loading, the deformations of the multibody models mimic that of linearly elastic transversely isotropic finite element models [14]. However, representation of deformable structures using multiple rigid bodies is an approximation that does not allow prediction of stress and strain internal to the meniscus tissue. The hoop tension force presented here is the total force acting between individual meniscus elements in the circumferential direction. The multibody method is computationally efficient and works well for dynamic studies, such as this one, exploring the interaction of multiple knee structures during movement.

In conclusion, computational knee models were created using medical images and passive leg motion and validated against relevant experimental studies. The models predicted that in the ACLD knee, the posterior horn of the medial meniscus provides a physical barrier that limits anterior motion of the tibia relative to the femur with application of anterior tibia force. The models also demonstrated that the peripheral meniscal attachments of the dMCL help constrain posterior motion of the medial meniscus, relative to the tibia, resulting from contact with the posterior medial femoral condyle. This dMCL constraint results in higher internal hoop tension in the posterior horn than in the anterior horn. In the intact knee, applying only an anterior force produces the smallest medial meniscus loads while combined anterior tibial force plus external tibial torque produced the greatest meniscal forces. The results of this computational study provide data on medial meniscus biomechanics, information which contributes to understanding of medial meniscus injury mechanisms and may aid meniscal repair and rehabilitation decisions.

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Highlights

- Loss of anterior crucial ligament (ACL) constraint increased medial meniscus posterior horn loading four fold with application of anterior tibial force.
- For the intact knee, applying a combined anterior tibial force plus external tibial torque produced the greatest loading on the medial meniscus.
- The peripheral meniscal attachments of the deep medial collateral ligament help constrain posterior medial meniscal motion, increasing hoop tension in the posterior horn.

Figure 1. Multibody knee model of subject 2 (a). Simulation, looking through the femur onto the tibial plateau, with applied tibia anterior force (b) and anterior tibia force plus external tibia rotation (c).

Figure 2. Predicted anterior cruciate ligament and posterior cruciate ligament force for subject 1 and subject 2 during passive flexion-extension simulations and simulations with 100 N anterior tibia or 100 N posterior tibia force applied. Model predictions are compared to experimentally measured forces of Markolf et al. [38, 39].

Figure 3. Percent change in length, relative to terminal extension length, of the anterior and posterior bundles of the deep medial collateral ligament (dMCL) during passive motion. Model predictions are compared to experimentally measured forces of Hosseini et al. [41].

Figure 4. Predicted total contact force, as a function of knee flexion angle, between the medial meniscus and tibial plateau cartilage for subjects 1 and 2.

Figure 5. Medial meniscus internal hoop force for subject 1. Anterior forces are for meniscus elements in the anterior horn and posterior forces are for meniscus elements in the posterior horn.

Figure 6. Medial meniscus internal hoop force for subject 2. Anterior forces are for meniscus elements in the anterior horn and posterior forces are for meniscus elements in the posterior horn.

Subject 1

Subject 2

Figure 7.

Model predicted contact force between the medial tibial plateau cartilage and medial meniscus elements for subject 1 (a) and subject 2 (b). Meniscus element 1 connects to the anterior horn attachments and meniscus element 19 (subject 1) or element 20 (subject 2) connects to the posterior horn attachments. Shown are element contact forces at the flexion angle with the greatest total contact force for the ACLd Ant (35.9° subject 1, 15.9° subject 2), Ext Ant (27.5° subject 1, 31.3° subject 2), and ACLd Ext Ant (27.4° subject 1, 30.3° subject 2) simulations.

Table 1

Stiffness values for the tibiomensicofemoral ligament bundles.

Ligament Bundle	Acronym	Stiffness Parameter (N)
ACL - anteromedial	aACL	6200
ACL - posterolateral	pACL	3400
PCL - anterolateral	aPCL	12500
PCL - posteromedial	pPCL	1500
LCL - anterior	aLCL	2000
LCL - middle	mLCL	2000
LCL - posterior	pLCL	2000
MCL - anterior superficial	asMCL	2500
MCL - middle superficial	msMCL	2600
MCL - posterior superficial	psMCL	2700
MCL - anterior deep	adMCL	1500
MCL - posterior deep	pdMCL	1500
POL	POL	1600
ALL	ALL	750
anterior intermeniscal	AIML	750
meniscus horn attachments:		
lateral anterior		2810
lateral posterior		1269
medial anterior		2349
medial posterior		1486

Table 2

Anterior-posterior excursion of the middle meniscus element during passive motion from extension to deep flexion. Experimental excursion values come from Yao et. al. [42] where the knees of 10 subjects were passively flexed to an average of $139^\circ \pm 3^\circ$.

	Lateral	Medial
Subject 1	7.5 mm	4.3 mm
Subject 2	9.6 mm	3.3 mm
Experimental	8.2 ± 3.2 mm (n=10)	3.3 ± 1.5 mm (n=10)