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Combined Measurement and Modeling of Specimen-Specific Knee Mechanics for Healthy and ACL-deficient Conditions

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Abstract

Quantifying the mechanical environment at the knee is crucial for developing successful rehabilitation and surgical protocols. Computational models have been developed to complement in-vitro studies, but are typically created to represent healthy conditions, and may not be useful in modeling pathology and repair. Thus, the objective of this study was to create finite element (FE) models of the natural knee, including specimen-specific tibiofemoral (TF) and patellofemoral (PF) soft tissue structures, and to evaluate joint mechanics in intact and ACL-deficient conditions. Simulated gait in a whole joint knee simulator was performed on two cadaveric specimens in an intact state and subsequently repeated following ACL resection. Simulated gait was performed using motor-actuated quadriceps, and loads at the hip and ankle. Specimen-specific FE models of these experiments were developed in both intact and ACL-deficient states. Model simulations compared kinematics and loading of the experimental TF and PF joints, with average RMS differences [max] of 3.0°[8.2°] and 2.1°[8.4°] in rotations, and 1.7[3.0] and 2.5[5.1] mm in translations, for intact and ACL-deficient states, respectively. The timing of peak quadriceps force during stance and swing phase of gait was accurately replicated within 2° of knee flexion and with an average error of 16.7% across specimens and pathology. Ligament recruitment patterns were unique in each specimen; recruitment variability was likely influenced by variations in ligament attachment locations. ACL resections demonstrated contrasting joint mechanics in the two specimens with altered knee motion shown in one specimen (up to 5 mm anterior tibial translation) while increased TF joint loading was shown in the other (up to 400 N).

Keywords

ligament; patella; finite element; cruciate deficient; walking

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Conflict of Interest Statement

The authors have no conflicts of interest to report related to this study.

Introduction

When healthy knee mechanics are compromised by injury or disease, load distribution through the joint is altered, which can lead to pain, additional injury, and long-term disability (Fulkerson, 2002; Nebelung and Wuschech, 2005). In particular, long-term studies of anterior cruciate ligament (ACL) injury have associated increased joint loading and altered knee kinematics with a high prevalence of osteoarthritis, knee pain, and instability (Lohmander et al., 2007; Nebelung and Wuschech, 2005). The ACL acts as the primary restraint to anterior translation of the tibia with respect to the femur, and a secondary restraint to internal-external and varus-valgus rotation (Girgis et al., 1975), and is the most frequently disrupted ligament in the knee (Beynnon et al., 2005).

Quantifying the mechanical environment at the knee is crucial for developing successful rehabilitation and surgical protocols following ACL injury. Since joint contact, soft tissue and muscle forces are difficult to quantify in-vivo, researchers have developed in-vitro cadaveric tests to evaluate natural knee mechanics. By simulating everyday activities, in-vitro measurements can be used to compare joint motions and tissue forces in healthy, pathological, and repaired specimens (Maletsky and Hillberry, 2005). Experimental testing provides a repeatable controlled environment for evaluation of joint mechanics, but can be costly and time-intensive when considering multiple design iterations and large numbers of specimens.

Hence, computational models have been developed to complement in-vitro studies (Bendjaballah et al., 1995; Blankevoort and Huiskes, 1996; Godest et al., 2000; Guess and Stylianou, 2012; Pena et al., 2006), and enable prediction of internal joint and soft tissue stresses/strains for efficient evaluations of knee mechanics. Computational knee models are typically built from digital representations of cadaver specimens from imaging, and tissue properties are calibrated using experimental measurements of tissue and whole-joint mechanics. Decisions on model complexity and the ability to calibrate model estimates are influenced by the available experimental data. For example, experimental joint laxity tests have been performed to develop load-displacement curves for calibration of computational representations of the passive soft-tissues of the knee (Godest et al., 2000; Kiapour et al., 2014; Mootanah et al., 2014). While passive experiments are important for quantifying joint stiffness and identifying ligament properties, these data do not necessarily represent the performance of the knee in activities. To that end, researchers have developed muscle-loaded experiments to simulate quadriceps (Ahmad et al., 1998; Baldwin et al., 2009) and hamstrings (Kwak et al., 2000) forces during dynamic tasks, and utilized these data in predictive musculoskeletal simulations (Adouni et al., 2012; Piazza and Delp, 2001; Shelburne et al., 2004) to estimate knee mechanics under conditions challenging to reproduce with in-vitro experiments. Taking a further step, studies have used computational models to simulate injury and degenerated conditions, such as rupture of the ACL and meniscectomy (Halonen et al., 2016; Li et al., 2002; Mesfar and Shirazi-Adl, 2006a; Moglo and Shirazi-Adl, 2003; Shelburne et al., 2004; Tanska et al., 2015), however, computational models are typically not compared to specimen-specific experimental data under both healthy and pathological conditions.

Our prior work focused on the development of computational models of the implanted knee during dynamic activity (Baldwin et al., 2012). Computational predictions were compared to experimental data from the six-degree-of-freedom electro-hydraulic Kansas knee simulator (KKS). More recently, models have been developed of the natural knee. Calibration of specimen-specific PF mechanics was performed in a muscle-loaded rig (MLR) designed to isolate the quadriceps mechanism during knee flexion (Ali et al. 2016). Joint laxity tests were performed on the same specimens to quantify joint constraint and derive optimized TF ligament material properties (Harris et al., 2016). However, these models of the PF and TF articulations of the knee were not combined into a dynamic representation of the natural knee.

The objective of this study was to create specimen-specific finite element (FE) models of the natural knee, including specimen-specific TF and PF soft tissue structures supported through kinematic comparisons to cadaveric experiments, and to evaluate joint mechanics for intact and ACL-deficient conditions. A muscle-loaded in-vitro simulation of gait using motor-actuated quadriceps forces, and loads at the hip and ankle was used to measure the dynamic motion of knee specimens. FE modeling replicated experimental loading conditions and model accuracy was evaluated through direct comparisons to the experimental TF and PF kinematics, and quadriceps forces in intact and ACL-deficient conditions. FE models included predictions of joint contact forces and ligament tensile and shear forces with respect to the tibia.

Methods

Summary

The current work was the third step in a three-step combined measurement and modeling approach to develop FE models of the natural knee for two specimens. In the first step, in-vitro testing replicated a deep knee bend using motor-actuated quadriceps force to calibrate PF mechanics in specimen-specific FE models of the experiment (Ali et al., 2016). In the second step, laxity experiments were performed in the same knees to capture passive constraint of the TF joint (Harris et al., 2016). FE modeling of the laxity experiments allowed calibration of TF soft tissue material properties and attachment locations for intact and ACL-deficient conditions. In the third and final step, the current study integrated TF and PF soft tissue representations developed in the previous two steps to evaluate subject-specific knee mechanics of the same specimens during dynamic activity replicated using the KKS.

Experimental setup

Dynamic in-vitro tests were conducted on two fresh-frozen cadavers (2 male; age: 50, 72 years; height: 175, 183 cm; weight: 127, 77 kg). Knees were thawed at room temperature and computed tomography (CT, 0.39x0.39x0.6mm, resolution:512x512) and magnetic resonance (MR, 0.53x0.53x0.6mm, resolution:320x320, sequence:T2 trufi3d_we_SAG) images were captured. Next, femur and tibia bones were sectioned approximately 20 cm from the joint line, cemented into aluminum fixtures, and all soft tissue beyond 10 cm of the joint was removed except quadriceps muscles. Each knee was subjected to three experiments

in intact and ACL-deficient conditions. First, passive TF laxity was measured by manually applying ± 8 Nm internal-external (I-E) torques, ± 10 Nm varus-valgus (V-V) torques, and ± 80 N anterior-posterior (A-P) loads ~ 300 mm below the joint line at 0–60° knee flexion (Harris et al., 2016). A load cell attached to the proximal end of the tibia recorded 6 DOF loads from each laxity test and provided real-time user feedback via LabView (National Instruments, Austin, TX). Second, PF mechanics were measured by placing the specimens in a test fixture that applied quadriceps force to extend the knee (Ali et al., 2016). Finally, specimens were mounted in the KKS to simulate the stance and swing phase of gait using load-controlled actuators (Figure 1). The KKS is a five-axis simulator designed to replicate knee joint loading during dynamic activity (Maletsky and Hillberry, 2005). Loads applied to the KKS actuators included a vertical hip load, quadriceps load, ankle flexion and I-E torque, and ankle medial-lateral (M-L) load. Quadriceps force was applied through the combined tendons of the rectus femoris and vastus intermedius using a proportional-integral-derivative (PID) controlled actuator tuned to match hip and ankle motions. Three-dimensional kinematic data were collected with an Optotrak motion capture system (Northern Digital Inc., Waterloo, CA). Simulated gait in the KKS was repeated following ACL resection. Anatomical landmarks on the femur, tibia, and patella, cruciate and collateral ligament attachments, articulating geometry (bone and cartilage surfaces), and KKS assembly components were digitized for constructing FE models of the experimental setup.

Computational Modeling

Specimen-specific FE models were developed in Abaqus/Explicit (Simulia, Providence, RI) to recreate the loading and boundary conditions for intact and ACL-resected conditions (Figure 1). Bone and cartilage geometry were manually reconstructed from CT and MR imaging, respectively, using ScanIP (Simpleware, Exeter, UK). Post-processing of geometric reconstructions and mesh refinement was performed in Hypermesh (v11.0, Altair, Troy, MI). Bones were represented using rigid triangular shell elements (R3D3), and cartilage was represented using hexahedral continuum elements (C3D8). The cartilage FE mesh was formed using a semi-automated morphing technique to match the surface geometry reconstructed from MRI to a hexahedral template (Baldwin et al., 2010). Although articular cartilage consists of several fibrous layers and viscoelastic properties (Halonen et al., 2013), cartilage was modeled using rigid pressure-overclosure behavior to minimize computational cost. Penalty-based contact (weight = 0.5, friction = 0.01) was defined between articulating cartilage using a calibrated surface pressure-overclosure relationship (Fitzpatrick et al., 2010); bone and soft tissue contact was defined using a zero surface penetration constraint (Halloran et al., 2005).

Tibiofemoral ligament structures were represented using non-linear tension-only springs (CONN3D2) and included the anteromedial-ACL bundle (ACLam), posterolateral-ACL bundle (ACLpl), anterolateral-PCL bundle (PCLal), posteromedial-PCL bundle (PCLpm), lateral collateral ligament (LCL), popliteofibular ligament (PFL), medial collateral ligament (MCL), deep medial collateral ligament (dMCL), posterior oblique ligament (POL), anterolateral structure (ALS), and medial and lateral posterior capsule (PCAPm, PCAPl). As described by Harris et al. (2016), TF ligament attachment sites, stiffness, and reference strain were optimized using an adaptive simulated annealing algorithm in Isight (Simulia,

Providence, RI) to match specimen-specific laxity measurements. In brief, specimen-specific optimizations were performed across multiple flexion states, multiple laxity tests, and multiple resection levels to provide a wide-ranging representation of joint constraint (Harris et al. 2016). Menisci geometry were developed from MR reconstructions and modeled using hexahedral continuum elements (C3D8) with 1D linear springs (CONN3D2) attaching the horns (N=37) and periphery of the geometry (medial N=16; lateral N=8) to the tibia bone. Menisci geometry were manually meshed and morphed based on reconstructions in Hypermesh (v11.0, Altair, Troy, MI).

Material properties for the menisci utilized Fung orthotropic hyperelastic material models (Erdemir, 2016; Sibole et al., 2010; Yao et al., 2006); material constants for Young's moduli (E, MPa), poisson's ratio (ν), and shear moduli (G, MPa) were $E_x=E_y=27.5$, $E_z=125$, $\nu_{xy}=0.33$, $\nu_{xz}=\nu_{yz}=0.1$, $G_{xy}=12.5$, $G_{xz}=G_{yz}=2$ (Figure 1). Spring stiffness of the horn attachments was computed as a function of literature-reported Young's modulus ($E=600\text{MPa}$) (Hauch et al., 2009), cross-sectional area of digitized attachment locations ($A\sim 30\text{mm}^2$), number of springs ($N=37$), and length of the spring ($L\sim 10\text{--}15\text{mm}$); $k=EA/NL$. Rigid-deformable frictionless contact was defined between the meniscus and articulating cartilage.

Patellofemoral soft tissue structures were represented by 2D fiber-enforced membrane elements and included the rectus-femoris tendon, patellar tendon, and medial and lateral patellofemoral ligaments. Ligament and tendon material properties and soft-tissue attachments of the patellar mechanism were adopted from our prior computational studies (Ali et al., 2016; Baldwin et al., 2009). KKS actuator loads at the hip and ankle joint were applied to the computational model to simulate dynamic activity performed in the experiment (Figure 1). KKS actuator components were represented using point-to-point connectors (CONN3D2) for computational efficiency.

KKS assembly components were aligned using digitized points from the motion tracking system. Experimental actuator loads were applied to the modeled KKS components (vertical hip load, ankle flexion torque, ankle I-E torque, and ankle medial-lateral load) using connector load definitions. Quadriceps excursion drove knee flexion and matched the experimentally prescribed hip flexion profile. By prescribing quadriceps excursion, the resulting connector load was used to predict model quadriceps force. Model setup and dynamic simulation was repeated for all specimens and their ACL-resected conditions.

In summary, experimental measurements consisted of TF and PF kinematics, and quadriceps forces from the KKS for intact and ACL-deficient conditions. Model accuracy was assessed using root-mean-square (RMS) differences between model and experimental TF and PF kinematics computed over the entire range of the gait cycle. Also, peak quadriceps forces during stance and swing phase of gait were compared in the model and experiment for intact and ACL-deficient conditions. Additionally, outputs from FE simulations included TF and PF contact forces, ligament tensile forces, and ligament A-P shear forces with respect to the tibia to describe changes in knee mechanics associated with ACL removal.

Results

TF Kinematics

Experimental TF kinematics were similar in both specimens (Figure 2) with two flexion peaks for the stance and swing phases of gait. TF kinematics were characterized by internal tibial rotation and posterior femoral rollback as a function of knee flexion. Following ACL resection, Specimen 1 showed notable increases in anterior tibial translation during swing phase of gait (+4.0 mm); Specimen 2 showed an overall shift in anterior position of the tibia (avg. +3.5 mm) and an average 4° increase in tibial external rotation.

Model-predicted TF kinematics agreed with the experiment in trend and magnitude. In Specimen 1, RMS and range [max,min] of differences between model and experiment were 3.1°[6.5,-2.7] and 3.5°[7.4,1.9] in flexion-extension (F-E), 1.0°[0.5,-2.2] and 1.6°[0.1,-3.3] in V-V, 5.4°[8.2,0.1] and 6.1°[8.4,0.2] in I-E rotation, and 0.9[2.1,-1.2] mm and 2.4[3.7,-5.1] mm in A-P translation, in the intact and ACL-resected condition respectively. In Specimen 2, RMS and range [max,min] of differences were 2.2°[4.9,-6.1] and 2.9°[4.1,-6.1] in F-E, 0.9°[2.1,-1.1] and 2.4°[1.5,-4.3] in V-V, 2.0°[3.8,-4.1] and 3.9°[5.8,-3.9] in I-E, and 1.6[3.0,-2.5] mm and 2.7[3.5,-5.0] mm in A-P, in the intact and ACL-resected condition respectively.

PF Kinematics

Experimental PF kinematics followed similar trends in both specimens, except in PF tilt, where Specimen 1 rotated internally and Specimen 2 rotated externally through the gait cycle (Figure 3). ACL resection produced minor changes in PF kinematics. Specimen 2 presented a 2–4 mm medial shift in patellar alignment.

Model-predicted PF kinematics agreed with the experiment. In Specimen 1, RMS and range [max,min] of differences between model and experiment were 2.2°[3.3,-4.2] and 1.9°[5.0,-2.6] in F-E, 2.2°[4.5,-3.0] and 2.9°[5.5,-3.9] in I-E, and 1.6[4.5,-4.0] mm and 2.5[5.0,-3.1] mm in M-L, for the intact and ACL-resected condition, respectively. In Specimen 2, RMS and range [max,min] of differences between model and experiment in the intact and ACL-resected condition were 4.2°[0.5,-7.0] and 1.8°[2.5,-4.9] in F-E, 1.0°[2.3,-1.8] and 3.2°[4.4,0.0] in I-E, and 2.2[0.2,-3.5] mm and 1.5[1.6,-3.9] mm in M-L, respectively.

Quadriceps Force

Comparing model quadriceps force to PID-controlled actuator load in the KKS, peak quadriceps forces (during stance and swing phase of gait) in the intact and ACL-deficient trials had differences of 21.1% and 22.1% for Specimen 1, and 9.7% and 7.6% for Specimen 2 (Figure 4). Differences in quadriceps force from intact to the ACL-deficient condition were small with negligible change in Specimen 1 (experimental RMS < 50 N) and a small decrease in peak quadriceps force during swing (328 N) in Specimen 2.

Contact Force

Total TF contact forces demonstrated decreasing trends from 0–45° in both specimens, but diverging trends from mid-to-deep flexion with decreasing forces as a function of knee flexion in Specimen 1 and increasing forces in Specimen 2 (Figure 5). In contrast, total PF contact forces were consistently increasing as a function of knee flexion in both specimens. TF center of pressure travelled posteriorly on the tibia and rotated internally, consistent with experimental TF kinematics; PF center of pressure and contact force travelled distal to proximal and increased in magnitude as knee flexion increased (Besier et al., 2005).

Ligament Forces

Ligament recruitment patterns were unique in each subject (Figure 6). In both specimens, ligament forces decreased as the knee flexed up until approximately 30°, after which ligament forces in Specimen 1 continued to decrease as a function of flexion, while ligament forces in Specimen 2 increased (Figure 6). In Specimen 1, primary contributors to joint constraint were the MCL, LCL, and the ACLam. In Specimen 2, the MCL, POL, DMCL, ACLam, and PCL were primarily active. Following ACL resection, Specimen 1 demonstrated an increase in total ligament force, with the MCL accounting for a majority of the constraint lost by ACL resection. In Specimen 2, POL force increased to compensate for the loss of the ACL. Following ACL resection, Specimen 1 showed little to no changes in ligament A-P shear force, but Specimen 2 demonstrated significantly lower anterior shear force, approximately equal to load carried by the ACL in the intact condition.

Discussion

FE modeling predicted TF kinematics, PF kinematics, and quadriceps force in intact and ACL-deficient specimens for an in-vitro simulation of gait. While prior computational studies have evaluated healthy and ACL-deficient knee mechanics, they have not verified predictions in both states during dynamic activity (Guess and Stylianou, 2012; Li et al., 2002; Mesfar and Shirazi-Adl, 2006a; Moglo and Shirazi-Adl, 2003; Shelburne et al., 2004). The current study provided a specimen-specific representation of the TF and PF joints by incorporating material properties and geometric alignment from previous modeling of the same specimens (Ali et al., 2016; Harris et al., 2016).

Model simulations captured experimental kinematics and loading of the TF and PF joints. In the intact condition, RMS differences between model and experiment TF kinematics were F-E<3.1°, V-V<1.0°, I-E<5.4°, and A-P<2mm. Removing the ACL in the model produced modest increases in RMS of <2° across all rotations and <1.5 mm in A-P. RMS differences in PF rotations were similarly low across all rotations (<4.2°). Even so, portions of the gait cycle were difficult to match to the experiment. For example, TF I-E rotations were the most challenging DOF to match computationally with differences of up to 8° during swing (50–90% in Figure 2) when compared to the experiment. While TF and PF kinematic predictions were similar to differences reported in the literature (Baldwin et al., 2009; Blankevoort and Huiskes, 1996; Guess et al., 2010), large differences highlight challenges in replicating specimen-specific passive constraint during dynamic activity. The largest RMS differences between model and experiment occurred at flexion angles beyond which laxity calibration

was performed ($>60^\circ$). Dynamic modeling suggests the need for additional evaluations of knee laxity at higher flexion angles, and potentially more sophisticated geometric (e.g. ligament wrapping) and material representations.

The computational model was also compared to experimentally-measured quadriceps force. Peak quadriceps force during stance and swing phase of gait had an average error of 16.7% across specimens and pathology. Force predictions in both specimens followed the experimental trend in quadriceps force and matched the F-E angle at which peak quadriceps force occurred within 2° . Since the experimental setup was an in-vitro representation of gait, quadriceps forces changed little following ACL resection.

TF and PF contact forces and center of pressure were consistent with previous reports. Total PF contact forces increased as a function of flexion, similar to the findings of (Besier et al., 2005; Fernandez et al., 2008). While Specimen 1 presented decreasing TF contact forces as the knee flexed, Specimen 2 demonstrated increased TF contact forces in swing phase at deeper flexion angles. This was likely influenced by contrasting ligament recruitment patterns in each specimen (Figure 6).

The modeling and comparison of two specimens revealed important individual differences that can be lost in generic models of the knee calibrated to average behavior (Mesfar and Shirazi-Adl, 2006a; Mootanah et al., 2014; Pandey and Shelburne, 1997). Our previous work, evaluating the joint laxity response of the two specimens, demonstrated significant intersubject variability in both ligament attachment locations and TF load response (Harris et al., 2016). Additionally, PF joint modeling demonstrated specimen-specific load transfer with either increased PF contact forces or increased patellar tendon loads following cruciate resection (Ali et al., 2016). Our recent and current studies successfully capture unique differences in joint mechanics between specimens, and emphasize the need for specimen-specific evaluations in computational modeling.

Although increasing joint contact forces corresponded to increasing ligament loads, each specimen displayed unique patterns of recruitment, especially at higher knee flexion angles (Figure 6). Contrasting ligament recruitment patterns and TF contact trends could stem from variability in knee anatomy (size, tibial slope), alignment (TF position, ligament attachments), and material properties (reference strain, stiffness) (Harris et al., 2016). The current specimens shared a similar size and shape (tibial slope $\approx 7^\circ$), but there were important differences in ligament attachment locations. Ligament engagement was particularly sensitive to the location of soft tissue attachments on the femur. In Specimen 2, the MCL and DMCL femoral attachments were located anterior to the TF center of rotation, causing the anterior bundles to generate substantial force in deep flexion; as a result, total ligament force increased as a function of flexion. Unique ligament engagement highlights the importance of specimen-specific representations of soft tissue structures.

ACL resections resulted in contrasting joint mechanics in the two specimens. Specimen 1 showed small changes in A-P position of the tibia, but displayed an increase in total joint forces, specifically in stance. However, TF contact and ligament forces were small during the peak of swing phase and resulted in a 4 mm anterior shift of the tibia. Specimen 2

demonstrated contrasting joint mechanics with increasing TF contact and ligament loads as a function of flexion. Specimen 2 showed a 4–6 mm shift in initial A-P alignment of the tibia. At deeper flexion angles during swing phase, TF contact forces and ligament loads were more active in preventing excessive TF motion. Measurements of ligament and contact forces were not available from the KKS to corroborate these results, but ACL forces were similar in magnitude to forces measured in situ by (Gabriel et al., 2004). The prediction of specimen-specific response to ACL-deficiency warrants further investigation into the structural characteristics of the knee that allow some individuals to cope with ACL-deficiency (Moksnes et al., 2008).

The main limitation of the current study was the in-vitro representation of gait, which did not fully reproduce in vivo conditions. Quadriceps forces in the experiment and simulations were higher during swing phase than forces reported in vivo. Larger quadriceps forces may have resulted in overestimation of contact and ligament forces at deep flexion angles. The current study modeled the resulting load response in the knee joint following ACL resection, but did not account for adaptive behavior that may be present in vivo, such as neuromuscular adaptation to excessive anterior-posterior motion through increased muscle recruitment. Nonetheless, the computational framework may be used to simulate soft-tissue injury and in-vivo correction by altering the tibial constraint through changes in muscle force (Mesfar and Shirazi-Adl, 2006b).

A second limitation was that the study was limited to two specimens due to challenges in cost and labor of collecting data for passive and dynamic tests, and calibrating specimen-specific FE models. The current work demonstrated the variability of ligament recruitment across two specimens and its impact on knee mechanics. However, additional specimens could better characterize ligament variability across the population to better inform engineers and clinicians on the mechanisms surrounding injury.

Furthermore, passive laxity tests and ligament calibrations were performed without the meniscus, thus the experiment and model may have overestimated the role of the ligaments in passive constraint (Harris et al., 2016). The meniscus is important to load distribution in the TF joint, which protects tibial cartilage from excessive loading and wear (Englund et al., 2009). Also, the meniscus may provide secondary constraint and stability of the ligament-deficient knee under joint load (Allen et al., 2000; Levy et al., 1982; Petrigliano et al., 2011). In prior work, knee laxity experiments were performed on specimens in this study using intact and meniscus-resected conditions to isolate the impact of the meniscus on joint constraint, however, likely due to absence of TF compressive load in the experiment, no significant differences were measured (Shoemaker and Markolf, 1986). TF compressive loads in the KKS were much greater and inclusion of the meniscus more accurately modeled joint constraint to reproduce experimental kinematics. Future work may be strengthened through better characterization of meniscus shape and specimen-specific calibration of meniscus properties.

Finally, FE models of the knee included 1D representations that were necessary for efficient model calibration. Ligament 1D elements effectively captured joint stiffness, but were not capable of modeling stress/strain distributions or wrapping contact. Simplified 1D

representations enabled reasonable computational run times in analyses of the dynamic activity, and also the optimizations used to tune ligament properties in our previous study (Harris et al., 2016). Previous modeling of the natural knee has included depth-dependent, collagen fiber cartilage (Halonen et al., 2013; Shirazi et al., 2008), and subject-specific modeling of the menisci (Guess et al., 2010) that might strengthen the accuracy and realism of our model predictions.

In conclusion, the current work expanded an existing FE framework of the KKS to include evaluations of healthy and ACL-deficient knee mechanics. FE models may be used for investigations that inform researchers and clinicians on the mechanisms surrounding injury, and support of surgical and conservative treatments. Recognizing the challenges in cost and labor to produce in-vitro biomechanical data, and develop specimen-specific computational models, the experimental motion and load data, and knee geometry are available for download at www.du.edu/biomechanics.

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Figure 1.

Knee cadaver mounted in the Kansas Knee Simulator (KKS) (left), and its computational representation (middle) with specimen-specific TF and PF soft tissue structures (right): anterior cruciate ligament (**ACLam, ACLpl**), posterior cruciate ligament (**PCLal, PCLpm**), lateral collateral ligament (**LCL**), popliteofibular ligament (**PFL**), medial collateral ligament (**MCL**), superficial medial collateral ligament (**DMCL**), posterior oblique ligament (**POL**), anterolateral structure (**ALS**), posterior capsule (**PCAPM, PCAPL**)

Figure 2.
Comparison of model (dashed) and experimental (solid) TF kinematics in the KKS simulator for intact and ACL-resected conditions in two specimens

Figure 3.

Comparison of model (dashed) and experimental (solid) PF kinematics in the KKS simulator for intact and ACL-resected conditions in two specimens.

Figure 4. Comparison of model (dashed) and experimental (solid) quadriceps force in the KKS simulator for intact and ACL-resected conditions in two specimens.

Figure 5. Total TF and PF contact force (left) and contact center of pressure with force vectors (right) shown for two specimens in intact and ACL-deficient conditions.

Figure 6.

Total (vector sum of all ligament force) and individual ligament recruitment as a function of knee flexion (left), and total ligament shear and tensile forces (right) for intact (solid) and ACL-deficient (dashed) conditions in two specimens.