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Validation of Predicted Patellofemoral Mechanics in a Finite Element Model of the Healthy and Cruciate-Deficient Knee

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Abstract

Healthy patellofemoral (PF) joint mechanics are critical to optimal function of the knee joint. Patellar maltracking may lead to large joint reaction loads and high stresses on the articular cartilage, increasing the risk of cartilage wear and the onset of osteoarthritis. While the mechanical sources of PF joint dysfunction are not well understood, links have been established between PF tracking and abnormal kinematics of the tibiofemoral (TF) joint, specifically following cruciate ligament injury and repair. The objective of this study was to create a validated finite element (FE) representation of the PF joint in order to predict PF kinematics and quadriceps force across healthy and pathological specimens. Measurements from a series of dynamic in-vitro cadaveric experiments were used to develop finite element models of the knee for three specimens. Specimens were loaded under intact, ACL-resected, and both ACL and PCL-resected conditions. Finite element models of each specimen were constructed and calibrated to the outputs of the intact knee condition, and subsequently used to predict PF kinematics, contact mechanics, quadriceps force, patellar tendon moment arm, and patellar tendon angle of the cruciate resected conditions. Model results for the intact and cruciate resected trials successfully matched experimental kinematics (avg. RMSE 4.0°, 3.1 mm) and peak quadriceps forces (avg. difference 5.6%). Cruciate resections demonstrated either increased patellar tendon loads or increased joint reaction forces. The current study advances the standard for evaluation of PF mechanics through direct validation of cruciate-resected conditions including specimen-specific representations of PF anatomy.

Keywords

knee; patellofemoral; finite element; cruciate deficient

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Conflict of Interest Statement

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

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Introduction

Healthy patellofemoral (PF) joint mechanics are critical to optimal function of the knee joint. The main function of the patella is to distribute quadriceps load to efficiently extend the knee (Buff et al., 1988; Huberti et al., 1984). Patellar maltracking creates increased PF ligament strains, soft tissue injury and/or knee pain (Fulkerson, 2002; Post et al., 2002). In addition, maltracking may lead to large joint reaction loads and high stresses on the articular cartilage; these factors increase the risk of cartilage wear and development of bone abnormalities which ultimately contribute to osteoarthritis (Han et al., 2005; Wu et al., 2000; Zhang et al., 2007). While the mechanical sources of PF joint dysfunction are not well understood, links have been established between PF tracking and soft-tissue pathologies and abnormal kinematics of the tibiofemoral (TF) joint (Li et al., 2004; Mizuno et al., 2001; Powers, 2003). Due to the interaction between TF and PF mechanics, PF dysfunction is prevalent following cruciate ligament injury and repair. The anterior cruciate ligament (ACL) of the knee is the primary restraint to anterior translation of the tibia relative to the femur, a secondary restraint to varus/valgus and internal/external rotations of the tibia, and a key guide to the screw-home mechanism at full extension (Girgis et al., 1975). Follow-up studies of ACL-deficient patients have found altered patellar tracking and PF contact mechanics (Van de Velde et al., 2008), including signs of knee instability, pain, and patellar dislocation (Nebelung and Wuschech, 2005). The posterior cruciate ligament (PCL) of the knee is the primary restraint to posterior translation of the tibia relative to the femur, and contributes more generally to tibiofemoral stability at higher flexion angles. Like those with ACL-deficiency, subjects with posterior cruciate ligament (PCL) deficiency exhibit altered patellar mechanics, particularly in deep knee flexion (von Eisenhart-Rothe et al., 2012). Understanding the interaction between cruciate injury and PF mechanics is important in determining optimal treatment pathways to better restore extensor mechanism function.

While in vivo and in vitro experiments may be used to compare patellar kinematics and quadriceps extensor function between healthy control subjects and cruciate-deficient patients, there are some limitations associated with these studies. Joint loads are impractical to quantify in-vivo and in-vitro studies may allow measurement of contact pressure and joint contact (Elias et al., 2004), but are typically costly and time-intensive so that only small numbers of specimens can be evaluated. Due to the challenges of quantifying patellar function using in vitro and in vivo experiments, computational models of PF mechanics have been developed to understand patellar function and treatment (Barry et al., 2010; Halonen et al., 2015; Mesfar and Shirazi-Adl, 2005, 2006a, b). Prior models incorporated muscle and ligament forces, and the interaction of PF and TF mechanics including contact stresses in cartilage and the menisci. These models were used to study the kinematics and kinetics of the PF joint through simulation of gait (Barry et al., 2010; Halonen et al., 2015) and knee flexion (Mesfar and Shirazi-Adl, 2005, 2006a, b). Furthermore, probabilistic analyses were used to simulate PF pathology due to variability in ligament material properties (Barry et al., 2010; Dhaher et al., 2010; Mesfar and Shirazi-Adl, 2006a), muscle loading (Mesfar and Shirazi-Adl, 2005), and kinematics (Mesfar and Shirazi-Adl, 2006b). While most prior models were based on specimen or subject specific geometry, they may be

limited because they were not calibrated or validated with combined experimental measurements of PF kinematics of the same subject or specimen.

Computational models are an ideal complement to experimental simulations (Beillas et al., 2004; Blankevoort and Huiskes, 1996). Sophisticated PF computational models can be validated using six degree of freedom (DOF) PF motion from identically loaded cadaveric tests (Baldwin et al., 2012; Guess et al., 2010; Heegard et al., 1995). Validated computational models can be used to overcome some of the limitations of in vivo and in vitro experiments; multiple procedures can be virtually performed on the same knee and compared under repeated loading conditions. Similar models have been used to evaluate cartilage damage in osteoarthritic patients (Cohen et al., 2003b), and simulate PF joint surgery (Cohen et al., 2003a), but typically are not validated under both healthy and altered conditions.

Restoration of normal patellar function is difficult to achieve once it has been compromised by injury or disease. To support clinicians and engineers, a reliable model for evaluation of PF joint mechanics is crucial to understanding patellar function and testing conservative and surgical therapies. The objective of this study was to create a validated finite element (FE) representation of the PF joint in order to predict PF kinematics and quadriceps force across healthy and pathological specimens. Specifically, given the relationship between PF dysfunction and cruciate ligament injury, intact, anterior cruciate ligament (ACL)-deficient and both ACL and posterior cruciate ligament (PCL)-deficient models were developed. While prior computational studies have modeled healthy PF mechanics and simulated injured/altered knee conditions, the current study advances the state of the art by recreating specimen-specific PF mechanics in healthy knees and directly validating cruciate-deficient conditions. A secondary objective was to assess the variability of PF mechanics to uncertainty in experimental measurement accuracy. The PF model was calibrated and validated through comparison to measured kinematics and quadriceps loads obtained from in-vitro simulations. Model calibrations were performed on the intact knee, while the subsequent ACL-deficient and PCL-deficient models predicted kinematics, quadriceps forces, and extensor function. Validated FE models may be used for the evaluation of cruciate injury and repair through parametric analyses assessing the variability in ligament/tendon stiffness, geometric shape and alignment, kinematics and muscle forces.

Methods

Experimental Testing

Three fresh frozen cadavers (all male, mean age of 55.3 years (range 44–72), mean height of 180.3 cm (range 175–183), mean weight of 91.5 kg (range 70–127)) were thawed at room temperature and, femur and tibia bones were sectioned approximately 20 cm from the knee joint line. All soft tissue beyond 10 cm of the joint was removed from the bones except quadriceps and hamstring muscles. Knees were examined and found to have no visible signs of injury. Following computed tomography (CT) and magnetic resonance (MR) imaging, a series of dynamic in-vitro tests were performed on the cadavers in the MLR as described by (Shalhoub and Maletsky, 2014). The MLR mounted the knee joint in an inverted position, such that the femur was rigidly attached and the tibia was allowed to move freely (Figure 1).

Quadriceps and hamstring tendons were clamped and passed through a series of pulleys to maintain a physiological orientation to the joint. A stepper motor (Nema 34, Danahar Automation, Wood Dale, IL) and a 1300 N load cell (Transducer Technique, Temecula, CA) were connected in-line with the quadriceps clamp to produce deep knee flexion to approximately 120 degrees and to measure the resulting quadriceps load. The quadriceps line of action was applied through the combined tendons of the rectus femoris and vastus intermedius. In addition, a static weight of 89 N was applied to the semimembranosus and biceps femoris hamstring muscles. An Optotrak 3020 motion capture system (accuracy within 0.04 deg. and 0.03 mm) was used to record tibiofemoral and patellofemoral kinematics (Maletsky et al., 2007).

Each knee specimen cycled through a deep knee bend in the MLR under three conditions: intact, ACL-resected, and both ACL and PCL-resected (referred to subsequently as PCL resected). Dynamic knee flexion tasks were repeated 5 times in intact and cruciate-resected conditions. Following testing, the specimen was removed from the MLR. Anatomical landmarks on the femur, tibia, and patella were digitized to establish bone fixed coordinate systems and track relative kinematics of the bones. In addition, position of soft tissue attachments and MLR components were digitized for constructing a finite element model of the experimental setup (quadriceps and hamstrings muscle line of action, patellar tendon attachment sites, rectus-femoris patellar attachment, biceps femoris and semimembranosus tibial attachments, point clouds of bone and articular geometry).

Computational Modeling

Specimen-specific finite element (FE) models of the MLR experiment were developed in Abaqus/Explicit (Simulia, Providence, RI) to recreate the loading and boundary conditions for the intact and cruciate-resected conditions (Figure 1). Given the complex, changing contact conditions and large deformations of soft tissue structures, explicit analyses were chosen for computational efficiency (Abaqus 6.11 Analysis Users Manual 2011). Bone and cartilage geometry were segmented and reconstructed from CT and MR imaging (1x0.35x0.35mm), respectively, using ScanIP (Simpleware, Exeter, UK). Bones were represented using rigid triangular shell elements (R3D3), and cartilage was represented using hexahedral continuum elements (C3D8R). 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). Patellofemoral soft tissue structures including the rectus femoris tendon, patellar tendon, medial patellofemoral ligament, and lateral patellofemoral ligament were modeled by 2D fiber reinforced membrane (M3D4R) and non-linear spring elements (CONN3D2). Ligament/tendon material properties were established in separate planar analyses to match published experimental uniaxial force-displacement data (Baldwin et al., 2009). A mesh convergence study was performed on the cartilage and PF soft tissue to ensure sufficient accuracy (<5% difference in kinematics and joint loads). Quadriceps and hamstring muscles were represented using point-to-point slot connections (CONN3D2): a single connector element for quadriceps, two connectors for the biceps femoris and semimembranosus.

Bone and cartilage were aligned to MLR test space using a semi-automated procedure that minimized the distance between points digitized on the specimen during testing and reconstructed geometry surfaces. Location and orientation of PF soft tissue structures, quadriceps and hamstring muscle line of actions were defined using digitized points from the experiment. The influence of patellar parameters on PF mechanics was isolated by kinematically prescribing TF motions. TF flexion-extension was driven using quadriceps tendon excursion measured from the experiment. The resulting reaction load in the quadriceps tendon was then compared to the experimental value as part of the calibration and validation of PF mechanics. All other TF DOFs were kinematically prescribed to match the experiment. TF kinematics were applied using tabular amplitude cards to reproduce the motions in the MLR experiment; motions were discretized at 0.1 second intervals over the 8 second analysis. For computational efficiency, bone and cartilage geometry were defined as rigid bodies, with appropriate mass and rotational inertia properties, in the FE simulation. Penalty-based rigid-body contact was defined between the articular cartilage of the patella and femur using a previously calibrated surface pressure-overclosure relationship (Fitzpatrick et al., 2010), and a hard pressure-overclosure relationship (zero surface penetration) was defined between bone and soft tissue (Halloran et al., 2005). A previous study compared deformable and rigid body contact in eight test specimens, and found similar contact pressures and area with kinematic differences less than 0.5° and 0.2 mm (Fitzpatrick et al., 2010). Model setup was repeated for all intact knees and their corresponding cruciate-resected cases.

In order to calibrate the intact models to the experimental kinematic data, soft tissue attachment locations and orientations were perturbed within measurement error of digitized points. Rectus femoris, patellar tendon, and hamstrings attachment locations were adjusted to calibrate model PF flexion-extension, internal-external rotation, and medial-lateral translation in intact knees to the experimental measurements. Additionally, pre-strain in medial and lateral PF ligaments was calibrated for each specimen in the intact condition. The pre-tension was manually adjusted for each specimen to minimize differences in initial experimental and model patellar alignment after quadriceps loading and prior to flexion (Baldwin et al., 2009). To validate performance of the model knees, soft tissue attachment sites and ligament pre-strain were not changed in the following ACL and PCL resected simulations. Outputs from specimen-specific models of the intact and resected knees included PF kinematics, joint contact mechanics, quadriceps force, patellar tendon moment arm, patellar tendon angle, and patellar force ratio.

Results

PF Kinematics

Experimental TF kinematics showed notable increases in internal rotation and posterior translation of the tibia after PCL resections, but presented only small changes in ACL resected specimens (Figure 2). There were consistent trends in all other DOF between the three specimens that remained following cruciate ligament resection.

Model predicted PF kinematics showed agreement with experimental data in both trend and magnitude. The model matched experimental data with average root-mean-square (RMS)

differences of 3.6° in rotations and 2.5 mm in translations for the intact trials, 3.9° and 3.1 mm in the ACL-deficient trials, and 4.6° and 3.7 mm in the PCL-deficient trials (Figure 3a). Specifically, PF flexion-extension and medial-lateral translation matched experimental kinematics with average RMS differences less than 5° and 3 mm. The model predicted PF tilt to the experiment in deep flexion ($> 50^\circ$) with average RMS difference less than 2°.

Measurement uncertainty in soft tissue attachments influenced model predicted PF kinematics (Figure 3b). Accounting for measurement error, the bounds of uncertainty in PF kinematics captured specimen-specific behavior.

Quadriceps Forces

When comparing model quadriceps forces to experimental load cell data, peak forces in the intact, ACL-deficient, and PCL-deficient trials had average errors of 4.3%, 7.2%, and 5.3% averaged across all specimens, respectively (Figure 4, Table 1). The modeled extensor mechanism was able to distinguish peak quadriceps loads between each of the specimens. Specimen 2 demonstrated notable increases in peak quadriceps force with subsequent cruciate resections (~24% from intact to PCL-resected) consistent with trends reported in literature (Ostermeier et al., 2004), however Specimen 1 showed a slight decrease (~8% from intact to PCL-resected) and Specimen 3 showed no difference in quadriceps load.

Patellar Tendon Moment Arm

Patellar tendon moment arm was calculated as the perpendicular distance between the patellar tendon and the helical axis center. The center of rotation in the tibiofemoral joint was determined using equations described by (Spoor and Veldpaus, 1980). In specimens 1 and 2, cruciate resections presented increases in patellar tendon moment arm in deep flexion, while Specimen 3 remained consistent (Figure 5a).

Patellar Tendon Angle

Patellar tendon angle was calculated as the angle between the tendon line of action and the mechanical axis of the tibia. In general, trends showed a positive angle in knee extension, which decreased with knee flexion; a negative angle accompanied wrapping of the patellar tendon around the anterior face of the tibia (Buff et al., 1988; Price et al., 2004; Yamaguchi and Zajac, 1989) (Figure 5b). While the magnitude of patellar tendon angle remained consistent from the healthy to the ACL-deficient condition, in the PCL-deficient condition, the patellar tendon angle was significantly larger through the flexion-extension cycle.

Patellar Force Ratio

Patellar force ratio was defined as the patellar tendon force (F_{pt}) divided by quadriceps force (F_q) (Figure 5c). An increase in force ratio following cruciate resection indicated either higher strain in the patellar tendon or redistribution of load from the quadriceps tendon to joint contact. Specimen 1 showed an increase in patellar force ratio following ACL and PCL resections. In Specimen 2, forces were redistributed from the quadriceps to the patellar tendon only in early flexion, but remained consistent with healthy joint loads at later flexion angles. Specimen 3 demonstrated an increase in patellar force ratio in the ACL deficient knee, but healthy and PCL-deficient trials showed similar magnitudes.

PF Contact

The ratio of PF cartilage contact force to quadriceps force increased with subsequent cruciate resections for all specimens (Figure 5d). PF contact distributions travelled superiorly and contact area increased as a function of knee flexion (Besier et al., 2005) (Figure 6).

Discussion

A detailed representation of the patellofemoral joint and quadriceps mechanism was developed to create a computational tool for the evaluation of PF mechanics in healthy and cruciate-deficient conditions. Prior computational studies of the PF joint have modeled healthy mechanics, and simulated injured and repaired knee conditions (Barry et al., 2010; Mesfar and Shirazi-Adl, 2006a; Salehghaffari and Dhaher, 2014). The current work created specimen-specific PF representations of soft tissue attachments and muscle line of actions, and provided a direct validation of cruciate-deficient mechanics. The FE representation of the healthy PF joint was kinematically calibrated to results from in-vitro testing of three intact specimens. The model was then validated through comparisons to experimental kinematics, quadriceps force, and extensor function in the cruciate resected conditions. The validated computational model of the PF joint effectively captured the overall path and range of motion in the patella (Figure 3a, 3b). Model results for the intact and cruciate resected trials successfully matched experimental kinematics with average RMS differences (4.0°, 3.1 mm) similar to values reported by others (Baldwin et al., 2009; Blankevoort and Huiskes, 1996; Guess et al., 2010), while effectively replicating individual differences between specimens.

Measurement uncertainties in soft tissue attachments and muscle line of action during the experiment affected model accuracy (Figure 3b). AP positions of the rectus femoris and patellar tendon attachments influence the extensor moment arm and quadriceps force. As a result, the orientation and patellar attachment sites of the rectus femoris and patellar tendon had a significant impact on PF flexion range of motion; a 1 mm anterior shift of the rectus femoris attachment site on the patella increased maximum PF flexion by ~3°. Although articulating geometry was the primary determinant of PF medial-lateral translations and internal-external rotations, variation in attachment sites and pre-strain of the PF ligaments influenced initial settling of the patella into the trochlear groove. Medial and lateral PF ligaments provided constraint and improved stability in M-L translations in early flexion.

Across specimens and conditions, peak quadriceps forces matched experimentally measured loads with an average difference of 5.6% (Figure 4). Forces were most accurate in mid flexion, where geometric constraint of the patella was the greatest. Magnitudes of quadriceps forces were consistent with loads described in similar knee-extension experiments (Buff et al., 1988; Grood et al., 1984). Perturbation of parameters with experimental measurement uncertainty (TF kinematics, soft-tissue attachments) demonstrated substantial influence in altering quadriceps load; in particular, uncertainty in the tibial attachment location of the hamstrings altered the flexion angle at which peak quadriceps force occurred, while measurement error in TF anterior-posterior kinematics influenced predicted force at deepest flexion.

Patellar tendon moment arm and angle were consistent with previous findings (Figure 5a, 5b). Moment arm and patellar tendon angle matched values reported in the literature in both trend and magnitude (Ahmed et al., 1987; Grood et al., 1984; Krevolin et al., 2004). When comparing healthy and cruciate-deficient conditions, patellar tendon moment arm increased in deep flexion consistent with the variations presented in patellar tendon angle. The ACL-deficient experimental kinematics did not produce a substantial change in TF anterior tibial translation, and so, patella tendon angle remained relatively consistent when compared to the healthy knee. Greater changes in patella tendon angle were observed in the PCL-resected condition due to the posterior shift of the tibia during knee flexion.

Cruciate resections demonstrated either increased patellar tendon loads or increased joint reaction forces (Figure 5c, 5d). Each specimen was unique in balancing load through the PF joint, highlighting the need for specimen-specific analyses of the quadriceps mechanism. While specimen 1 distributed forces from the quadriceps to both patellar tendon and cartilage surfaces, specimen 2 primarily transferred load to joint contact. Specimen 3 alternated load distribution from the patellar tendon to the joint reaction in subsequent ACL and PCL resections, with an increase in patellar force ratio after removing the ACL and an increase in joint contact force after removing the PCL. PF contact distributions travelled superiorly and contact area increased as a function of knee flexion, consistent with patterns reported by others (Besier et al., 2005; Huberti et al., 1984) (Figure 6). Specimens 1 and 3 showed only small changes to PF contact center of pressure following ACL and PCL resections, however, specimen 2 demonstrated a lateral shift in the PCL-deficient trial (shown in Figure 6). Even though contact patterns were similar in healthy and pathological conditions, peak contact pressures increased (2–12%) with cruciate resections.

The main limitation of the current study is the relatively small differences in TF and PF kinematics between healthy and ACL-deficient conditions. The inverted position of the experimental setup did not promote anterior translation of the tibia from an ACL resection during knee extension as the weight of the remaining lower limb tended to place a posterior load on the tibia. Also, the application of hamstrings forces in extension may have contributed to reducing the anterior motion of the tibia. Because TF A-P differences were small, the ACL may have been lightly loaded in the fully-intact trials as well. However, the PCL-deficient condition, the inverted experimental setup and hamstrings loads caused large posterior tibial translations. Therefore, PCL-deficient trials presented significant TF and PF kinematic variability when compared to the healthy specimen. A second limitation was that thigh/calf contact was not modeled and may have influenced quadriceps force in deep flexion. Model predicted quadriceps forces were in reasonable agreement to the experiment, however, the flexion angle at which peak loads occurred was difficult to match computationally with an average error of 14.8%. In PCL-deficient trials, the experiment reported an average 8° reduction in peak flexion angle. As joint stability decreased with PCL resections, each specimen demonstrated increased posterior translation of the tibia in deep flexion. As a results, the knee joint may have partially been supported in deep flexion by the soft-tissue mass (muscle, fat, posterior capsule) posterior to the joint. The FE representations did not include modeling this soft-tissue, and so quadriceps force predictions did not account for off-loading from thigh-calf reaction force.

Additionally, finite element analyses relied on PF ligament material properties from literature (Baldwin et al., 2009). To evaluate the sensitivity of model kinematics on these properties, ligament stiffness parameters were doubled; differences in PF kinematic predictions remained less than 1° and 1 mm.

Furthermore, the experiment and associated computational model produced loads less than in vivo weight bearing conditions. The purpose of this work was to demonstrate that the computational model could appropriately evaluate different normal and pathological states produced by a repeatable experiment. Considering the role of cruciate ligaments in anterior-posterior constraint at the knee and overall joint stability, cruciate injuries have been identified as a key factor in altered neuromuscular function at the knee (Devita et al., 1997; Patel et al., 2003), which will typically present altered muscle activations thought to compensate for the partial or complete absence of the cruciates. Future application of the validated PF model could be used to simulate in vivo correction of tibial anterior-posterior constraint through changes in muscle force (Mesfar and Shirazi-Adl, 2006b).

Finally, TF motions were kinematically prescribed in the model. A resultant quadriceps line of action was used to describe TF flexion-extension, but may have limited the contribution of individual vastus muscles on PF mechanics, especially PF kinematics in off-axis DOF. Prescribing TF kinematics was necessary to focus on patellar calibration and isolation of the effects of patellar parameters on the extensor mechanism. Future models will include musculoskeletal loads and specimen-specific representations of TF soft tissue structures.

The validated computational model predicts PF joint mechanics in the intact and cruciate-deficient knee. The PF model developed in this study will be used for subject-specific predictions of PF joint kinematics and quadriceps force, integrated into a multi-scale musculoskeletal model of the lower extremity for investigation of normal, pathologic, and repaired function. These models may be used to simulate soft-tissue injury and repair and quantitatively assess the effect of surgical decisions during ACL or PCL reconstruction on PF mechanics and extensor mechanism efficiency and function.

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Figure 1.
Knee cadaver mounted in muscle loading rig (MLR) (right) and its computational representation (left)

Figure 2.

Experimental TF kinematics for the intact, ACL-deficient, and PCL deficient conditions (VV: varus(+)/valgus(-), IE: internal(+)/external(-), ML: medial(-)/lateral(+), AP: anterior(+)/posterior(-), SI: superior(+)/inferior(-))

Figure 3.

a) Comparison of experimental and model predicted PF kinematics in the intact (left), ACL-deficient (middle), and PCL-deficient (right) conditions averaged across specimens. b) Uncertainty in model PF kinematics (F-E, I-E, and M-L) shown for 3 intact specimens with experimental (solid line), model (dashed line), and bounds of uncertainty (shaded region). (FE: flexion(+)/extension(-), VV: varus(+)/valgus(-), IE: internal(+)/external(-), ML: medial(-)/lateral(+), AP: anterior(+)/posterior(-), SI: superior(+)/inferior(-))

Figure 4. Model predicted quadriceps forces in the intact (left), ACL-deficient (middle), PCL-deficient (right) conditions

Figure 5.

a) Patellar tendon moment arm, b) patellar tendon angle, c) patellar force ratio, and d) contact to quadriceps force ratio presented for intact and cruciate-deficient conditions. Shaded regions represent the span of experimental data from literature sources (Ahmed et al., 1987; Buff et al., 1988; Grood et al., 1984; Yamaguchi and Zajac, 1989)

Figure 6.

a) PF contact pressure distributions shown in a representative specimen at knee flexion angles of 0°, 30°, 60°, and 90°. b) PF contact center of pressure through the flexion activity and c) contact distribution at ~90° is shown for a representative specimen in intact and cruciate-deficient conditions.

Table 1

Experiment and model quadriceps forces and corresponding flexion angles shown for intact and cruciate-deficient conditions. Peak forces are averaged over the flexion and extension portions of the activity

Specimen	Maximum Quad Force									
	Intact			ACL			PCL			
	(N)	% Error	(°)	(N)	% Error	(°)	(N)	% Error	(°)	
DU01	Exp	536.0	5.6	103.7	26.8	512.0	0.2	105.0	90.5	18.4
	Model	566.1		76.0		512.9		79.5	73.9	
DU02	Exp	333.1	1.1	84.8	8.9	332.7	11.7	91.2	76.8	25.6
	Model	329.5		92.4		371.6		90.9	96.5	
DU03	Exp	272.7	6.3	77.0	11.4	266.5	9.6	80.1	73.7	13.5
	Model	289.9		85.8		292.1		77.2	83.6	
Average			4.3		15.7		7.2		9.4	19.1