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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).

1. 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

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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 menisectomy (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.

2. Methods

2.1. 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.

2.2. 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, $0.39 \times 0.39 \times 0.6$ mm, resolution: 512×512) and magnetic resonance (MR, $0.53 \times 0.53 \times 0.6$ mm, resolution: 320×320, 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 $\sim$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 (Fig. 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.

2.3. 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 (Fig. 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 (v), and shear moduli (G, MPa) were $E_x = E_y = 27.5$, $E_z = 125$, $v_{xy} = 0.33$, $v_{xz} = v_{yz} = 0.1$, $G_{xy} = 12.5$, $G_{xz} = G_{yz} = 2$ (Fig. 1). Spring stiffness of the horn attachments was computed as a function of literature-reported Young's modulus ($E = 600$ MPa) (Hauch et al., 2009), cross-sectional area of digitized attachment