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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.

1. 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 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

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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, ACL-deficient and both ACL and 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 though parametric analyses assessing the variability in ligament/tendon stiffness, geometric shape and alignment, kinematics and muscle forces.

2. Methods

2.1. 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 a muscle loading rig (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 (Fig. 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° and to measure the resulting quadriceps load. The quadriceps line of action was applied through the combined tendons of the rectus femoris and vastus intermedialis. 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° 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,

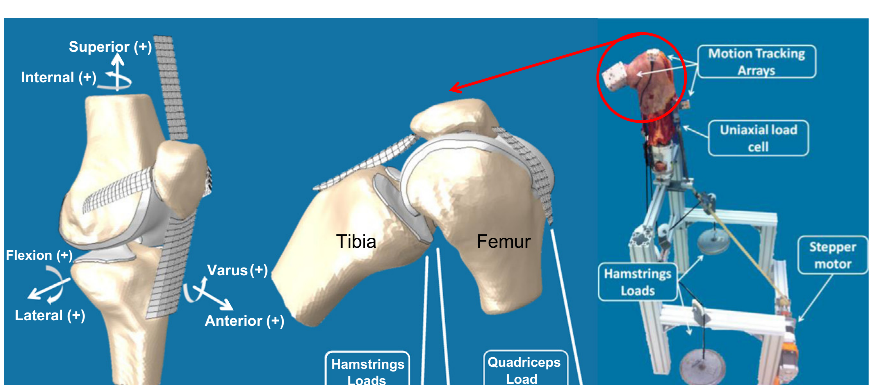

Fig. 1. Knee cadaver mounted in muscle loading rig (MLR) (right) and its computational representation (left).