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SIMULATION BASED PREDICTION OF THE EFFECT OF MPFL RECONSTRUCTION ON PATELLOFEMORAL MECHANICS

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INTRODUCTION

Patellofemoral complications remain the single largest reason for knee related clinical visits. Yet, robust clinical treatment remains a challenge [1]. To establish causal relationships and understand joint behavior, a complimentary approach utilizing simulation and experimentation may offer valuable insight. Simulation can be confirmed with experimental data and can also be exploited in a predictive capacity. For example, the medial patellofemoral ligament (MPFL) is a clinically relevant structure due to its role in patellofemoral stabilization [2]. MPFL reconstruction, which can be explored in a simulation framework, often utilizes a relatively stiff semitendinosus or gracilis tendon autograft [3]. The procedure is accepted to address patients with chronic patellar instability [4]. While joint stability may be achieved with such an approach, the underlying cartilage loading, and potential long term effects, are unknown. Previous simulation results found sensitivity in cartilage pressures during MPFL reconstruction [4], and these findings may be corroborated using a higher fidelity evaluation of clinically relevant factors. In the context of developing a general patellofemoral simulation framework, the goal of this study was to evaluate the effects of reconstructed MPFL zero force reference ("slack") length on predicted joint mechanics across a range of potential values. To support the predictive simulation results, a preliminary model validation was also performed against specimen-specific *in vitro* joint mechanics.

METHODS

Experiments were performed using a non-pathological cadaver knee of a 75 years old, 90 kg male donor. The procedures included magnetic resonance (MR) imaging of the knee followed by mechanical testing of the patellofemoral joint (Rotopod R2000, PRS Corp., Hampton, NH). To match image and mechanical testing coordinate systems, MR opaque registration marker sets were placed on the femur, tibia, and patella (Fig. 1b, three 10 mm radius for both the tibia and femur and three 5 mm radius spheres). For 10 mm radius spheres, this registration setup was shown to have a relative accuracy within 1% for measuring the distance between two markers. A pressure sensor (K-Scan sensor 5051, Tekscan Inc., MA) was placed between the cartilage surfaces of the patella and femur to measure the contact pressure distribution. The femur was fixed to a stationary frame and the tibia to a moving platform. The quadriceps tendon was attached to a linear actuator using a freeze clamp. For image to experiment registration, as well as to quantify joint kinematics, Optotrak (NDI Corp., Ontario, Canada) infrared emitting diode (IRED) marker clusters were rigidly attached to each bone (Fig. 1b). An IRED digitizer was used to record points on each spherical marker to identify marker centers.

To provide an approximate range of expected tibiofemoral flexion angles during gait, tests were performed at 0°, 15°, 30°, 45°, and 60°. Each tibiofemoral joint configuration was achieved through passive flexion. At each angle, a nominal quadriceps load of 20 N was applied before loading from 100 N up to 600 N, in 100 N increments. At each increment, bony positions and contact pressure distributions were recorded.

An explicit **finite element model** of the patellofemoral joint was developed from the MR images and solved using Abaqus/Explicit v6.9 (Simulia, Providence, RI). The model included the tibia, femur, femoral cartilage, patella, patellar cartilage, patellar tendon and single line elements for the lateral retinaculum and MPFL (Fig. 1). Bones were modeled as rigid shells, cartilage was linear elastic [5], and the patellar ligament and quadriceps tendon included a hyperelastic ground substance and embedded nonlinear springs along the fiber direction, with properties obtained from [6]. Patellar to femoral cartilage and patellar to quadriceps tendon contact included a friction coefficient of 0.04 [7]. The lateral retinaculum stiffness was 2 N/mm and the zero force slack length was defined at 0° knee flexion [4]. To simulate an MPFL reconstruction, **"native"** (12 N/mm) and **"reconstructed"** (100 N/mm) stiffness values were adopted [4]. MPFL slack lengths were defined relative to 0° knee flexion across a range of -4 to +4 mm, in 2 mm increments (5 total "reconstructed" simulations). All simulations included prescription of experimentally determined tibiofemoral kinematics at 600 N quadriceps load. For comparison with experimental results, patellofemoral kinematics [8], load distribution (contact force and area) and peak pressure were extracted from the simulations.

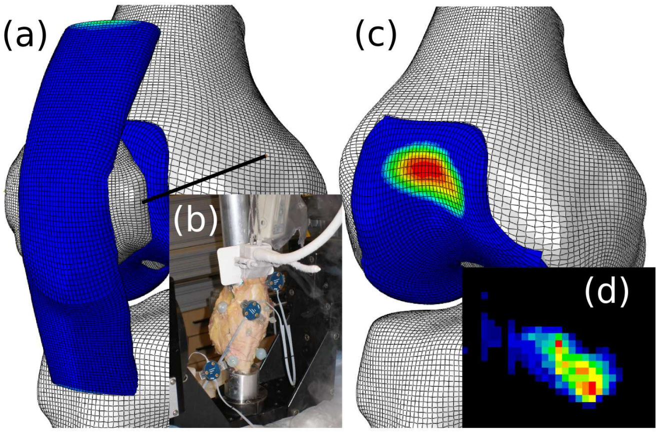

Figure 1. (a) Specimen-specific finite element model and (b) experimental setup. (c) Corresponding contact pressure distribution found at 30° knee flexion and 600 N quadriceps load for both simulation and (d) experimental results.

RESULTS

For the native (non reconstructed MPFL) model, kinematics agreed in overall trend and magnitude with experimental data, realizing a root mean square error of 5.7 mm and 5.3° for displacements and rotations, respectively. Of note, kinematic errors were largely systematic and trends were faithfully reproduced. Pressure predictions displayed similar behavior when compared to experimental measurements but relative trends were generally less pronounced (Fig. 2). For the relatively stiff reconstructed MPFL, zero force slack lengths had an effect on contact mechanics (Fig. 2). As slack lengths shortened, relative peak pressure (based on maximum of experiment or simulation) increased from 0.8 up to 1.45 at a flexion angle of 60° (Fig. 2). Patellofemoral contact force increased systematically for all flexion angles and patella internal rotation and lateral displacement also increased with decreasing slack length.

DISCUSSION

The model achieved adequate initial agreement with experimental data while also being used in a predictive capacity. MPFL slack length for the "reconstructed" state showed a pronounced effect on contact mechanics, as well as relative bony kinematics. The findings may point towards the importance of controlling MPFL slack length during reconstruction, where shorter lengths led to increased patellofemoral force transmission and resultant contact pressures (average and peak, for higher flexion angles).

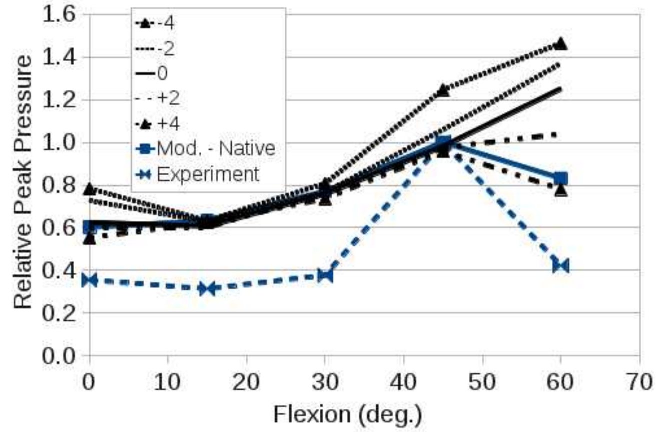

Figure 2. Relative peak pressure results as a function of knee flexion. Results for the MPFL reconstruction are delineated by their slack length values, ranging from -4 to +4 mm, in 2 mm increments. Experiment results ("Experiment") are included, as are simulation results for the native MPFL stiffness ("Mod. – Native").

Before conclusions can be drawn, a number of modeling assumptions should be addressed. Single, linear line elements were used to simulate the MPFL and lateral retinaculum. This may not be an adequate representation of these structures. Tissue attachment sites for the MPFL were approximated by a "non-isometric" configuration [9], though it is recognized this variable may play an important role in joint function [4]. Relatively soft linear behavior was assigned to the cartilage and is likely one factor leading to discrepancies in contact mechanics (Fig. 2). Uncertainty estimation, possibly achieved using a probabilistic approach, will offer much insight into these and other assumptions, and will be a focus of future studies. In spite of the untested assumptions, this exploratory study compares well with previous work, highlighting the importance of MPFL properties in patellofemoral mechanics [4].

Overall, the role experimentation can play in simulation was demonstrated through this synergistic study. Validation data was acquired using a controlled experiment, explicitly developed for modeling purposes, and simulation offered predictive capability beyond the experimental results. Simulation of clinically relevant procedures offers obvious value, but if uncertainties can be properly addressed, pre-operative planning may also become a possibility.

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