CREATION AND VALIDATION OF DISCRETE ELEMENT MODELS OF THE KNEE

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INTRODUCTION

Patellofemoral Pain Syndrome (PFPS) accounts for 25-40% of knee related injuries presented in sports medicine centers. The site of pain is the patellofemoral joint, where the patella meets the femur creating a small groove along the femoral end of the knee having a cartilage lined surface. PFPS is associated with maltracking of the patella within this groove and over activity of the joint during knee motion [1,2]. Pain is most prevalent during or after physical activity in exercises such as walking, stair-climbing, squatting, and sitting. The cause of pain is not well defined, but it has been associated with increased joint pressure at the cartilage surface of the patellofemoral joint. Further understanding of the location and cause of pain will expedite intervention and mitigate the risks for early-onset osteoarthritis due to cartilage wear on both the patella and femoral surfaces [3]. Evaluation of joint pain in vivo in the knee is limited to computational modeling for studying joint pressures due to the inability to physically measure these joint pressures. However, in vitro data is capable of capturing joint pressures generated during motion consistent with various physical tasks. An in vitro study was designed and implemented to generate discrete element analysis (DEA) computational models of the knee joint during normal physiological tasks. Models were produced using kinematic data to drive motion. During experimental data collection, in vivo muscle loading conditions were used to generate data as similar to in vivo joint pressures as possible. Previous in vitro studies including computational modeling lacked physiological knee loading [4]. Before implementation in vivo, models required validation. Experimentally collected pressure data acted as validation between model-generated pressures and experimental measurements in the patellofemoral and tibiofemoral joints. Therefore, experimental methodology for a cadaveric study was generated with in vivo conditions and tested for creation and validation of DEA models.

OBJECTIVE

Our objective was to develop and validate DEA models of three cadaveric knees. Models were developed to accurately predict average contact pressures and contact area both quantitatively and qualitatively within the patellofemoral and tibiofemoral joints. Additionally, our goal was to generate data as similar to in vivo data as physiologically possible using an in vitro model.

HYPOTHESIS/SUCCESS CRITERIA

We hypothesize contact area and average contact pressure will have an average of less than 20% error between the predicted model output and the experimentally collected data. Still further, we expect model predicted data to resemble previous literature values. We expect increased contact pressure due to greater flexion of the knee as well as increased contact pressure due to femoral rotation at the joint.

METHOD

JOINT GEOMETRY

Three fresh, frozen cadaveric knees were dissected removing skin, excess soft tissue, and quadriceps muscles. The ligaments and soft tissue inside the joint space were kept intact to retain normal physiological knee motion. Registration cubes were fixed to the center of the patella and the tibia and femur on the medial side closest to the joint line. Cubes served as bone tracking via digitization and were fixed using cyanoacrylate and baking soda. High definition MRIs were collected for each knee in an extended unloaded position. Knee geometry was computationally created using imaging software (Mimics, Plymouth, MI). 3D knee structures were generated using frame by frame segmentation of each MRI in sagittal, coronal, and axial plane slices for sub-millimeter accurate 3D computational geometric surface models of the bones, cartilage, and meniscus. Knee surface geometry was imported into meshing software (Hypermesh, Altair Engineering Inc., Troy, MI) for surface meshing and smoothing. Bone, cartilage, and meniscus geometry were generated as tetrahedral shell elements with an average element size of 2 mm.

KINEMATIC DATA COLLECTION

Figure 1. An anterior view of the knee within the STAR-IV system with registration blocks attached and quadriceps tendons clamped.

The quadriceps tendons were isolated for application of simulated loading of the quadriceps muscles. The femur, tibia, and fibula were cut 20 cm from the joint space. The fibula was fixed to the tibia for normal knee geometry and motion. The femur and tibia/fibula were potted with epoxy-putty for grip inside of the STAR-IV jig-apparatus. The quadriceps tendons were individually clamped and loaded at angles representative of in vivo muscles with 51 N applied to the vastus medialis, 87 N to the vastus intermedius, and 77 N to the vastus lateralis. [5] A 35 kg compression load was applied to the top of the femur to simulate half of body weight through one leg. These loads were applied to give accurate joint pressures as representative...
of in vivo as possible. For model development, the knee was load in 14 quasistatic conditions, which included 0, 15, 30, 45, 60, and 75 degrees of flexion at neutral rotation as well as 5 degrees internal, external, varus, and valgus rotation of the femur at 0 and 15 degrees of flexion. Kinematic motion of the bones was tracked via digitization of cube surfaces using a 3D digitizer (FaroArm®, Lake Mary, FL). Local coordinate systems at each cube were generated for creation of transformation matrices between patella/femur and tibia/femur within both experimental and MRI system to drive the model.

**EXPERIMENTAL CONTACT MECHANICS**

Contact stresses in the tibiofemoral and patellofemoral were measured using 4000 and 5051 TekScan pressure sensors, respectively. Pressure sensors were conditioned, equilibrated, and calibrated using a two-point calibration according to TekScan operation requirements (Iscan, TekScan Inc., South Boston, MA). The 5051 pressure sensor was sutured to the soft tissue above and below the patella on the patellar tendon, with sensing surface covering the entire patella. The 4000 pressure sensor was inserted under the meniscus after the capsule was cut and sutured to the soft tissue on the anterior tibia as seen in Figure 1. During kinematic testing, stress distribution maps were recorded for both joints at all positions, which served as validation for model generated stresses.

**RESULTS**

![Figure 2](image)

Qualitatively (Figure 2), the contact patterns between the computational model and experimental data were similar and changed with femoral rotation. The largest difference in contact area between the DEA analysis and experimental data occurred on the lateral facet in 5 degrees of internal rotation (Error = 25%). The largest error in average contact pressure occurred on the medial facet in neutral (Error = 46%) (Table 1). A strong correlation of contact pressure with femoral rotation existed that showed an increase in average contact pressure on the lateral facet from an externally rotated femur to an internally rotated femur (Experimental: $r = 0.99$; Computational: $r = 0.93$). There were also strong correlations of a decrease in average contact pressure on the medial facet from an externally rotated femur to an internally rotated femur (Experimental: $r = -0.95$). The contact pressure at 5 degrees of internal rotation was below the minimum threshold set by the computational model (10 KPa) and was not calculated.

**DISCUSSION**

The preliminary data for two knees shows strong similarity between stress distributions both qualitatively and quantitatively. There is clear correlation between increased flexion angle and increased joint pressure in the patellofemoral joint. In addition, internal rotation of the femur is accompanied by greater medial compartment pressure in the tibiofemoral joint and lateral pressure in the patellofemoral joint. The opposite is true for external rotation of the femur similar to previous literature [6]. Differences between the computational model and experimental data could be due to the number and size of the springs utilized for the DEA analysis, material properties of the springs, as well as a minimum threshold of 10 KPa used to determine contact that may have missed regions of contact. Model error was not quite below the 20% error threshold for validation of model outputs for 15 degrees of flexion, but data for a third knee and at all other positions has yet to be computed. Limitations of software and hardware malfunctions during tested prevented data at greater flexion angles. Additionally, maintaining biomechanical properties of cadaveric tissue during testing limited the time and speed of testing. In the future, validated computational models will be used to quantify contact stresses of dynamic motion in vivo.

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**REFERENCES**