Published in final edited form as:

Comput Methods Biomech Biomed Engin. 2015 November ; 18(14): 1509–1515. doi:
10.1080/10255842.2014.921814.

Finite Element Analysis to Characterize How Varying Patellar Loading Influences Pressure Applied to Cartilage: Model Evaluation

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Abstract

A finite element analysis (FEA) modeling technique has been developed to characterize how varying the orientation of the patellar tendon influences the patellofemoral pressure distribution. To evaluate the accuracy of the technique, models were created from MRI images to represent five knees that were previously tested in vitro to determine the influence of hamstrings loading on patellofemoral contact pressures. Hamstrings loading increased the lateral and posterior orientation of the patellar tendon. Each model was loaded at 40°, 60° and 80° of flexion with quadriceps force vectors representing the experimental loading conditions. The orientation of the patellar tendon was represented for the loaded and unloaded hamstrings conditions based on experimental measures of tibiofemoral alignment. Similar to the experimental data, simulated loading of the hamstrings within the FEA models shifted the center of pressure laterally and increased the maximum lateral pressure. Significant ($p < 0.05$) differences were identified for the center of pressure and maximum lateral pressure from paired t-tests performed at the individual flexion angles. The ability to replicate experimental trends indicates that the FEA models can be used for future studies focused on determining how variations in the orientation of the patellar tendon related to anatomical or loading variations or surgical procedures influence the patellofemoral pressure distribution.

Keywords

patellofemoral joint; cartilage pressure; patellar tendon; finite element analysis

INTRODUCTION

For patients with patellofemoral pain, a common focus of treatment is reducing the pressure applied to lateral patellofemoral cartilage. Lateral cartilage can be overloaded due to anatomical factors such as a lateralized tibial tuberosity (Balcerek et al. 2011) or a weak vastus medialis obliquus (Makhsous et al. 2004) elevating the resultant lateral force applied

to the patella by the quadriceps muscles and the patellar tendon. Repetitive overloading can lead to degradation of cartilage, and eventually arthrosis (Fulkerson 2004). Overloading the subchondral bone, activating subchondral nociceptive fibers, is noted as a source of pain (Biedert and Sanchis-Alfonso 2002).

Several finite element analysis (FEA) studies have characterized how anatomical parameters or surgical treatment options influence patellofemoral pressure levels. Some studies focused on the contact pressure distribution (Benvenuti et al. 1997; Fernandez and Hunter, 2005; Fitzpatrick et al. 2011; Shirazi-Adl and Mesfar 2007), while others quantified hydrostatic pressures and octahedral shear stresses within the patellofemoral cartilage (Besier et al. 2008; Farrokhi et al. 2011). Due to the assumptions and mathematical approximations included in FEA, models should be validated against accepted measurements before addressing clinical concerns. Ideally, validations should be based on parametric variations and output parameters similar to those that will be studied. Validation studies have been performed for several models used to characterize the patellofemoral pressure distribution. These studies compared FEA output to measures such as contact area from in vivo studies (Besier et al. 2005; Besier et al. 2008; Farrokhi et al. 2011), in vitro and in vivo measures of kinematics (Baldwin et al. 2009; Farrokhi et al. 2011; Heegaard et al. 1995) and kinematics and pressure values from other finite element models (Fitzpatrick et al. 2010), but have not incorporated parametric variations associated with symptoms or treatment methods.

An FEA modeling technique has been developed to characterize how varying the loading condition applied to the patella influences the patellofemoral pressure distribution. A particular focus of the modeling technique is addressing the influence of anatomical variations, loading variations, and surgical treatment methods that alter the orientation of the patellar tendon on the patellofemoral pressure distribution for symptomatic knees. Before application to symptomatic knees, the current study was performed to determine the ability of FEA models to accurately identify variations in patellofemoral contact pressures related to alterations in the orientation of the patellar tendon induced by varying hamstrings loading.

METHODS

FEA models were created to represent five cadaveric knees that were previously tested in vitro. Each knee was MRI scanned (MAGNETOM Symphony 1.5T, Siemens) in the sagittal plane with an extremity coil (CP Extremity Coil, Siemens) using a three-dimensional T2-weighted scan with fat saturation (repetition time = 38 ms, echo time = 5 ms, flip angle = 25°, field of view = 16 cm, slice thickness = 1.0 mm). The DICOM images were utilized to create triangulated surface meshes representing the femur, tibia and patella and the cartilage on the femur and patella (3D-Doctor, Able Software, Lexington, MA). The solid models of the bones and cartilage were meshed (TrueGrid, XYZ Scientific Applications, Livermore, CA) for FEA (Fig. 1). The bones were treated as rigid structures (Fitzpatrick et al. 2010), with the surfaces modeled with four-node quadrilateral elements. Cartilage at the bone-cartilage interface was rigidly fixed to the underlying bone. The cartilage was modeled with three layers of deformable eight-node hexahedral elements. At the surface, the elements were approximately 0.5 mm along each side. A convergence analysis showed that increasing

the mesh density by approximately 25% increased the maximum cartilage pressure by less than 0.2%.

The scanned knees were previously tested in vitro, for a study focused on the influence of hamstrings loading on patellofemoral contact pressures (Elias et al. 2011). Hamstrings loading increased tibial external rotation and posterior translation (Mani et al. 2011), which influenced patellofemoral contact pressures by altering the orientation of the patellar tendon. Each knee was tested with the femur fixed horizontally in isometric knee extension at 40°, 60° and 80° of flexion. The tibial tuberosity of each knee was osteotomized and shifted laterally by approximately 5 mm to represent an elevated lateral force acting on the patella. Forces were applied to the patella through the vastus medialis obliquus (27 N), the vastus lateralis (127 N) and the combination of the vastus intermedius/vastus medialis longus/rectus femoris (432 N), producing a loading pattern with a weak vastus medialis obliquus representative of patients with patellofemoral pain (Elias et al. 2009). When loaded, hamstrings forces were applied through the semimembranosus (100 N) and the biceps femoris (100 N) tendons. All forces were applied through loading cables oriented to represent anatomic loading directions and run over pulleys (Elias et al. 2011; Mani et al. 2011).

For the in vitro study, patellofemoral pressures were measured using thin (0.1 mm thick) film sensors (I-Scan 5051, Tekscan, Boston, MA) inserted into the joint (Elias et al. 2011). The sensors were calibrated on a material testing machine while sandwiched between two sheets of neoprene rubber. For each experimental test, the position of the patellar ridge was palpated on the sensor. Patellofemoral and tibiofemoral kinematics were also quantified. Sensors from a pulsed DC magnetic tracking system (trakSTAR, Ascension Technology, Burlington, VT) were fixed to the femur, patella and tibia, with a separate sensor used to digitize landmarks on the bones. The landmarks determined the orientations of anatomical reference axes that were tracked based on the motion of the sensors fixed to the bones. The digitized landmarks were the most medial and lateral points on the femoral epicondyles and two points along the long axis of the femur, the most medial, lateral and distal points on the patella, and the most medial and lateral points on the tibial epicondyles and a central point at the distal cut of the tibia. Tibiofemoral (Mani et al. 2011) and patellofemoral (Elias et al. 2011) translations and rotations were quantified based on the floating axis convention (Grood and Suntay 1983).

The experimental testing configuration was reproduced within the FEA software (Abaqus/Standard, Simulia, Providence, RI). The meshed bones of each knee were oriented to represent each experimental test (Fig. 1). The landmarks used to create the anatomical reference axes were identified on the models of the femur and tibia, so that tibiofemoral alignment could be reproduced from the measured kinematics (Grood and Suntay 1983). The patella was initially aligned with a mating surface within the trochlear groove (Elias et al. 2010, Elias and Saranathan 2013). At 60° of flexion, a point of contact was identified on the patellar ridge, halfway along the length of the articular surface. To incorporate proximal translation of the contact area on the patella with knee flexion, at 40° and 80°, the contact point was shifted by one-eighth of the length of the articular surface distally and proximally, respectively. The experimentally determined patellar flexion was applied, and the patella

was translated along the trochlear groove until a tangent to the point of contact was aligned with a parallel surface.

The patella was allowed six degrees of freedom, while the positions of the femur and tibia were fixed in space. The conditions with the hamstrings loaded and unloaded were replicated by altering the orientation of the tibia based on the in vitro kinematics data, with the position of the tibial tuberosity determining the orientation of the patellar tendon (Elias and Saranathan 2013). Hamstrings forces were not applied to the model. For comparison to the experimental data, an FEA model with the patellar tendon oriented based on the hamstrings loaded in vitro kinematics data is referred to as the hamstrings loaded condition. For the three flexion angles, the tibial tuberosity was on average 3 to 4 mm more lateral with the hamstrings loaded than unloaded and 2 to 5 mm more posterior, with the posterior change increasing with the flexion angle. The articular cartilage was modeled as isotropic and linearly elastic, as done for previous studies (Farrokhi et al. 2011; Fitzpatrick et al. 2010; Kwak et al. 2000), with an elastic modulus of 5 MPa and Poisson's ratio of 0.45 (Blankevoort and Huiskes 1991; Kwak et al. 2000). The coefficient of friction between cartilage surfaces was set to 0.02 (Besier et al. 2008).

Quadriceps forces were applied to the patella to match the forces applied experimentally. The site of the pulley for each quadriceps loading cable was replicated in the FEA environment to apply forces along the experimental orientation. When a loading cable wrapped around the distal femur, the wrapping points were identified on the model to alter the orientation of the applied force. Using measurements from the tested knees and anatomical landmarks on the patella, a line of attachment on the meshed patella was identified for each of the loaded quadriceps muscles. The orientations of the forces were determined by vectors from the attachment points on the patella to points representing the connection at the pulley or the wrapping point on the femur. Quadriceps forces were applied through tension-only springs representing the quadriceps tendon. Each quadriceps force was applied to a rectangular 2-D rigid shell connected to the patella through ten springs with a total stiffness of 1350 N/mm (Stäubli et al. 1996). The quadriceps forces were ramped to the values applied experimentally.

Tension-only spring elements were used to represent the patellar tendon, medial patellofemoral ligament (MPFL) and medial patellomeniscal ligament (MPML). The patellar tendon, MPFL and MPML were assigned stiffness values of 2000 N/mm, 12 N/mm and 5 N/mm, respectively (Elias and Cosgarea 2006). The sets of four and three springs representing the MPFL and MPML, respectively, were oriented from the patella to the femur and tibia, respectively, based on anatomical landmarks (Elias and Cosgarea 2006). Wrapping points on the femur were identified when needed to adjust the orientation of the MPFL. The five springs representing the patellar tendon were oriented from attachment points on the distal patella to the tibial tuberosity. The attachment points of the patellar tendon on the distal patella and the tibial tuberosity were determined from anatomical landmarks and measurements on the cadaveric knees.

The comparison between the FEA and experimental results focused on the maximum lateral and medial contact pressure, contact area, and the center of pressure on the surface of the

patellar cartilage. Pressure output was divided between the lateral and medial facets based on identification of the patellar ridge on the model and the experimentally tested patella. The position of the center of pressure was quantified as the distance along the lateral facet to the patellar ridge. A two level repeated measures ANOVA was performed at each flexion angle to determine if there was a significant ($p < 0.05$) difference between the computational and experimental data and if hamstrings loading influenced the output. When hamstrings loading significantly influenced the output, paired t-tests were performed for both the experimental data and the FEA results at each flexion angle to examine for a difference due to the loading condition. In addition, the maximum lateral pressure was compared to the maximum medial pressure for the computational and experimental data at each flexion angle using a paired t-test.

RESULTS

For both the FEA and experimental data, pressure tended to be concentrated on the lateral facet of the patella, with the lateral concentration larger for the loading condition including the hamstrings (Fig. 2). For all the experimental tests and FEA simulations, the average center of pressure varied from 4 to 9 mm lateral to the patellar ridge (Fig. 3). The center of pressure was significantly more lateral with the hamstrings loaded for the experimental data at all flexion angles ($p < 0.02$) and for the FEA data at 80° ($p = 0.03$). For the computational data, loading the hamstrings increased the average lateral position of the center of pressure by 0.5 to 0.8 mm at the individual flexion angles, compared to 0.9 to 1.1 mm for the experimental data. The center of pressure from FEA was not significantly different from the experimental data at any flexion angle ($p > 0.09$).

For both the FEA and experimental data, the maximum pressure applied to cartilage tended to be larger with the hamstrings loaded. For both the experimental tests and FEA simulations, the maximum lateral pressure (Fig. 4) tended to be larger than the maximum medial pressure (Fig. 5), with the difference significant ($p < 0.01$) at 60° and 80° . For both the experimental tests and FEA simulations, the maximum lateral pressure was significantly ($p < 0.05$) larger with the hamstrings loaded than unloaded at 60° . Loading the hamstrings increased the average maximum lateral pressure from FEA by 0.1 to 0.2 MPa at the individual flexion angles, compared to 0.5 to 0.7 MPa for the experimental data. The FEA maximum lateral pressure was significantly larger than the experimental values at 40° ($p = 0.01$), but not at 60° or 80° ($p > 0.1$). At 80° , the experimental maximum medial pressure decreased significantly when the hamstrings were loaded, but the decrease in the maximum medial pressure from FEA was not significant. The maximum medial pressure did not show a tendency to decrease when the hamstrings were loaded at 40° and 60° . The maximum medial pressure from FEA was not significantly different from the experimental data at any flexion angle ($p > 0.09$). Loading the hamstrings did not significantly ($p > 0.07$) influence the contact area (Fig. 6), and no significant differences were identified between the experimental contact area measurements and the FEA values ($p > 0.06$).

DISCUSSION

The FEA model was developed to characterize the influence of variations in the orientation of the patellar tendon on the pressure applied to patellofemoral cartilage. The pressure applied to lateral cartilage increases with the lateral position of the patellar tendon attachment at the tibial tuberosity (Ramappa et al. 2006; Saranathan et al. 2012), and medialization of the tibial tuberosity is a common approach for surgical patellar realignment. Multiple computational models have characterized the influence of the orientation of the patellar tendon on the patellofemoral pressure distribution without specifically validating the models for variations in the orientation of the patellar tendon (Benvenuti et al. 1997; Cohen et al. 2003; Fernandez and Hunter, 2005; Shirazi-Adl and Mesfar 2007). A modeling system that represented cartilage with elastic springs was shown to accurately characterize how varying the orientation of the patellar tendon influences patellofemoral contact pressures (Elias and Saranathan 2013), but that system cannot characterize how pressure varies throughout the thickness of cartilage or produce other relevant pressure output, such as hydrostatic pressure and octahedral shear stress. The current model is not validated for these additional parameters due to the limitations of in vitro measurements, but can be used to quantify additional measures of cartilage pressure in future studies. Other FEA models that have been developed to characterize patellofemoral pressures have been validated against in vivo and in vitro measures of contact area and kinematics (Baldwin et al. 2009; Besier et al. 2005; Besier et al. 2008; Farrokhi et al. 2011; Heegaard et al. 1995) and kinematics and pressure values from other finite element models (Fitzpatrick et al. 2010).

The ability of FEA to characterize the influence of the orientation of the patellar tendon on patellofemoral contact pressures was assessed using data from an in vitro study focused on the influence of hamstrings loading on the pressure distribution. For the in vitro study, hamstrings loading directly influenced the orientation of the patellar tendon. With the femur fixed horizontally, hamstrings forces were added to forces applied through the quadriceps muscles. Because no other changes were made to the experimental set-up when the hamstrings were loaded, the variations in the orientation of the tibia and the patellofemoral contact pressures were recorded with minimal experimental error (Elias et al. 2011). For the FEA model, the only change applied to represent application of hamstrings forces was the altered position of the tibial tuberosity changing the orientation of the patellar tendon. Future implementations of the model should include tibiofemoral degrees of freedom to allow variations in tibiofemoral kinematics that determine the orientation of the patellar tendon.

Similar to the experimental results, FEA showed that increasing the lateral and posterior orientation of the patellar tendon shifts pressure toward the lateral facet of the patella. For both the FEA models and the experimental data, the altered orientation of the patellar tendon due to loading the hamstrings tended to increase the maximum lateral pressure and shift the center of pressure laterally. For both the FEA models and the experimental data, the maximum lateral pressure tended to be larger than the maximum medial pressure for both loading conditions, which was expected due to the lateral shift of the tuberosity and the reduced force applied by the vastus medialis obliquus to simulate symptomatic loading conditions.

The output values from FEA were generally similar to the experimental data, with the exception of the maximum lateral pressure being significantly larger for FEA than the experimental data at 40°. Although the change was not statistically significant, the FEA-determined contact area tended to be lower than the experimental measurements at 40° as well, likely contributing to over-estimation of the peak pressure. Increased constraint as the patella moves deeper into the trochlear groove likely contributes to improved agreement between the experimental and computational data as the flexion angle increases. Flexion angles less than 40° were not tested in vitro to avoid patellar instability that could be caused by the simulated lateral malalignment, vastus medialis obliquus weakness and large quadriceps forces (Saranathan et al 2012). The FEA models include several approximations that could influence the pressure output, such as the material properties assigned to the cartilage, patellar tendon and ligaments and initial patellofemoral alignment. Error could also influence the experimental pressure output. Contributors to experimental error include the sensor influencing contact mechanics (Wu et al. 1998), wrapping the sensor around the patella (Wilson et al. 2003), sensitivity of the sensor to shear loads (Ramappa et al. 2006), and representing cartilage with rubber during calibration (Wilson et al. 2003).

FEA is a powerful tool for evaluating how anatomical variations and surgical treatment methods influence the patellofemoral pressure distribution. Ideally, the models should be evaluated with respect to parametric variations and output similar to those that will be studied. The FEA modeling technique produces a reliable characterization of the influence of alterations in the orientation of the patellar tendon on the patellofemoral pressure distribution. The modeling technique can be expanded to include representation of symptomatic knees and evaluation of specific surgical procedures, although tibiofemoral degrees of freedom are needed to fully represent surgical realignment of the tibial tuberosity. The current model represents an initial step in a FEA-based methodology for evaluating how variations in the orientation of the patellar tendon produced by loading conditions, anatomy and surgical treatment influence the patellofemoral pressure distribution in symptomatic knees.

ACKNOWLEDGEMENT

The study was supported by National Institute of Arthritis and Musculoskeletal and Skin Diseases/National Institutes of Health grant R03AR054910.

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Figure 1.

Finite element model representing one knee showing the springs representing the patellar tendon, medial patellofemoral ligament (MPFL) and medial patellomeniscal ligament (MPML). The forces applied by the vastus medialis obliquus (VMO), vastus lateralis (VL), and combination of the vastus intermedius/rectus femoris/vastus medialis longus (VI) were applied to the rigid rectangles connected to the patella through the springs shown. The knee is also shown in profile with patella flexed to 40°, 60° and 80°. Initial alignment of the patella in the trochlear groove is also shown.

Figure 2.

Contact pressure patterns from the FEA model and the in vitro experimental measurements superimposed over the patella for two knees at 40°, 60° and 80° of flexion. Areas of highest pressure (red) shift laterally when loading the hamstrings increases the lateral and posterior orientation of the patellar tendon. The pressure scale varies between flexion angles and between the FEA model and in vitro data to maximize the color distribution but is constant for each set of the two loading conditions.

Figure 3.

Average (+ standard deviation) distance from the center of pressure to the patellar ridge from the in vitro experimental data and FEA models with the hamstrings unloaded and loaded. Significant differences due to loading the hamstrings are marked (*).

Figure 4.

Average (+ standard deviation) maximum lateral pressure from the in vitro experimental data and FEA models with the hamstrings unloaded and loaded. Significant differences due to loading the hamstrings are marked (*). A significant difference between the FEA data and the experimental data is also marked (A > B).

Figure 5.

Average (+ standard deviation) maximum medial pressure from the in vitro experimental data and FEA models with the hamstrings unloaded and loaded. A significant difference due to loading the hamstrings is marked (*).

Figure 6.

Average (+ standard deviation) contact area from the in vitro experimental data and FEA models with the hamstrings unloaded and loaded. No significant differences were identified due to loading the hamstrings.