THE INFLUENCE OF HAMSTRINGS LOADING ON PATELLOFEMORAL BIOMECHANICS: A FINITE ELEMENT STUDY

A Thesis

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ABSTRACT

Patellofemoral pain can be caused due to overloading, overuse, patellar malalignment or trauma. Lateral malalignment of patella is observed commonly to cause PF pain. There are various non-operative treatments employed before exploring surgical options. Muscle strengthening is one of the non-operative treatments for relieving patellofemoral pain. However, strengthening of the wrong muscle group could further the lateral malalignment exacerbate the situation. In previous in-vitro studies it has been observed that hamstrings activation causes posterior shift and external rotation of the tibia which could lead to lateral orientation of the patellar tendon resulting in the lateral shift of the patella furthering malalignment in symptomatic knee. Due to the nature of in-vitro studies it limits the physiological conditions and surgical procedures that can be evaluated over the time of study. Thus, the main aim of the current study was to create a validated finite element model to characterize the influence of hamstrings loading on patellofemoral pressures and kinematics. The model was validated by a correlation and regression analysis taking into account the variations due to anatomy, loading conditions and flexion angle between the finite element model and the experimental results. The relationship between the two methods was significant for predicting maximum contact pressure, mean contact pressure and patellar flexion. There was a positive correlation for predicting patellar tilt between the two methods. Hydrostatic pressures, contact shear
forces and octahedral shear stresses were the additional parameters obtained from the finite element model which helped to characterize the 3D state of stress in the cartilage. In the finite element model, paired t-tests were performed to test the variations on hamstrings loading on pressure and kinematic parameters and the additional parameters computed to characterize the 3D stress-state. The current model can be used further to simulate surgeries like tibial tuberosity transfers which affect the patellofemoral pressures and kinematics.
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4.27 Femur cartilage octahedral stresses averaged over three knees at each flexion angle for quadriceps and hamstrings loading and quadriceps-only loading obtained from the computational model........................................................................... 88
1.1 Patellofemoral Pain Syndrome (PFPS)

Patellofemoral pain syndrome (PFPS) is a common knee problem. PFPS is commonly referred to as 'anterior knee pain' or 'runner's knee'. Specific populations at high risk of anterior knee pain are young athletes and very active patients. 16 to 25 % of all injuries in runners is due to PFPS while 11 % of musculoskeletal complaints in an office setting are caused due to PFPS [1, 2].

1.2 Cause and Effect of PFPS

PFPS can be caused due to various reasons like muscle dysfunction, anatomic anomalies, patellar hyper mobility, tight lateral structures, overuse or trauma. The risk factors that could cause PFPS alter the force distribution and resultant moment in the quadriceps muscles and patellar tendon which results in abnormal patellar tracking. Thus, patellar malalignment is believed to contribute to PFPS commonly [3]. Due to valgus orientation of the normal knee, the resultant forces
and moment in the quadriceps muscles and patellar tendon tend to shift and tilt the patella laterally. An excessive lateral force and excessive lateral tilt could cause an abnormal pressure distribution at the articular cartilage which could lead to cartilage degradation and eventually osteoarthritis [4]. Cartilage damage can lead to increased pressure on the subchondral bone causing knee pain [3, 5].

1.3 Non-operative Treatment for PFPS

Patients with PFPS are initially treated non-operatively. Standard physical therapy for patients with patellofemoral pain usually consists of muscle strengthening, anti-inflammatory medication, orthotic devices, stretching the lateral retinaculum, prone quadriceps muscle stretching, aerobic conditioning, taping and bracing [3, 6]. Non-operative treatments like physical therapy and anti-inflammatory medication work for most patients [6, 7]. During non-operative treatments for PFPS, strengthening of the inappropriate muscle could also worsen the pain. For instance, in-vitro studies have shown that activation of hamstring muscles moves the tibia more posterior and rotates it externally during flexion movements [8, 9, 10]. This leads to a lateral orientation of the patellar tendon and consequently to lateral malalignment of the patella increasing lateral facet contact pressures. Excessive pressures can lead to cartilage degradation resulting in pain.

1.4 Purpose of Study

In previous in-vitro studies it has been observed that hamstrings strengthening can contribute to lateral malalignment of patella shifting pressure on the lateral
facet [13]. Although due to the nature of in-vitro studies the pressures characterized are in 2D. The purpose of the current study was to develop three validated FE models by replicating a previous in-vitro study to predict the variations in patellofemoral pressures and kinematics with and without hamstrings loaded. The FE model gave 3D stress-state information of the cartilage which gave better insight about cartilage stress-state.

1.5 Specific Aim of Study

1. Develop three finite element knee models by replicating a previous in-vitro study.

2. Validate the FE models by performing a correlation and regression analysis between the finite element and experimental method. Statistically understand the effect of hamstrings loading on patellofemoral pressures and kinematics.

3. Obtain additional parameters from the computational study to characterize the 3D stress-state of the articular cartilages.

The contact pressures and patellofemoral kinematics from the computational study were compared to the experimental study for validation of the finite element model. The contact parameters compared were maximum contact pressures, mean contact pressures and lateral force ratio. The patellar ridge is the partition that divides the patellar cartilage into medial and lateral sides. The kinematic parameters compared were patellar flexion, lateral tilt, lateral rotation and lateral shift. The additional parameters computed were hydrostatic pressure, contact shear force and octahedral stresses. These parameters help to evaluate the
condition of the cartilage in 3D compared to experimental testing where only contact pressures could be evaluated.

1.6 Statements of Hypothesis

Finite element knee models were developed for three specimens. For validation of trends on hamstrings loading a correlation and regression analysis was performed to compare the experimental and computational data.

The mean and standard deviation of additional parameters, obtained from the computational study, were computed for the three knees for each flexion angle to understand the effect of hamstrings loading and compare the variations with experimental results.

1.6.1 Null Hypotheses:

H_{01}: A validated finite element model can be created which has a significant (p< 0.05) positive correlation with the experimental method. Maximum contact pressure, mean contact pressure and lateral force percentage were the pressure parameters and patellar flexion, lateral tilt, lateral rotation and lateral shift were the kinematic parameters compared between the experimental and computational methods. Differences in anatomy, loading conditions and flexion were accounted for while comparing the two methods.

H_{01a}: There is a significant positive correlation for maximum contact pressure between computational and experimental method

H_{01b}: There is a significant positive correlation for mean contact pressure between computational and experimental method
H_{01c}: There is a significant positive correlation for lateral force percentage between computational and experimental method

H_{01d}: There is a significant positive correlation for patellar flexion between computational and experimental method

H_{01e}: There is a significant positive correlation for patellar tilt/external rotation between computational and experimental method

H_{01f}: There is a significant positive correlation for lateral/valgus rotation between computational and experimental method

H_{01g}: There is a significant positive correlation for lateral shift between computational and experimental method

For the finite element model, variations in pressure and kinematic parameters with and without hamstrings loaded were compared at all flexion angles using paired t-tests (p<0.05). Also, the trends with and without hamstrings loading were compared between the two methods.

H_{02a}: Hamstring loads do not affect the maximum contact pressures of the patellar cartilage at 40°, 60° and 80° degrees of flexion in the computational study.

H_{02b}: Hamstring loads do not affect the mean contact pressures of the patellar cartilage at 40°, 60° and 80° degrees of flexion in the computational study.

H_{02c}: Hamstring loads do not affect the lateral force ratio of the patellar cartilage at 40°, 60° and 80° degrees of flexion in the computational study.

H_{02d}: Hamstring loads do not affect the patellar flexion at 40°, 60° and 80° degrees of flexion in the computational study.
\( H_{02e} \): Hamstring loads do not affect the patellar tilt at 40°, 60° and 80° degrees of flexion in the computational study.

\( H_{02f} \): Hamstring loads do not affect the lateral rotation at 40°, 60° and 80° degrees of flexion in the computational study.

\( H_{02g} \): Hamstring loads do not affect the lateral shift at 40°, 60° and 80° degrees of flexion in the computational study.

For the finite element model, variations in the additional parameters with and without hamstrings loaded were compared at all flexion angles using paired t-tests.

\( H_{03a} \): Hamstring loads do not affect the hydrostatic pressures of the patellar cartilage at 40°, 60° and 80° degrees of flexion in the computational study.

\( H_{03b} \): Hamstring loads do not affect the contact shear forces of the patellar cartilage at 40°, 60° and 80° degrees of flexion in the computational study.

\( H_{03c} \): Hamstring loads do not affect the octahedral stresses of the patellar cartilage at 40°, 60° and 80° degrees of flexion in the computational study.

1.6.2 Summary of null hypotheses

\( H_{01} \): \( \mu_{\text{comp}} = \mu_{\text{exp}} \) for all pressure and kinematic parameters between computational and experimental methods.

\( H_{02} \): \( \mu_{\text{hams}} = \mu_{\text{nohams}} \) for all pressure and kinematic parameters using the computational method at 40°, 60° and 80° degrees of flexion in the computational study.

\( H_{03} \): \( \mu_{\text{hams}} = \mu_{\text{nohams}} \) for all additional parameters using the computational method at 40°, 60° and 80° degrees of flexion in the computational study.
1.6.3 Alternate Hypotheses:

H$_{11a}$: There is no significant positive correlation for maximum contact pressure between computational and experimental method

H$_{11b}$: There is no significant positive correlation for mean contact pressure between computational and experimental method

H$_{11c}$: There is no significant positive correlation for lateral force percentage between computational and experimental method

H$_{11d}$: There is no significant positive correlation for patellar flexion between computational and experimental method

H$_{11e}$: There is no significant positive correlation for patellar tilt/ external rotation between computational and experimental method

H$_{11f}$: There is no significant positive correlation for lateral / valgus rotation between computational and experimental method

H$_{11g}$: There is no significant positive correlation for lateral shift between computational and experimental methods

For the finite element model, variations in pressure and kinematic parameters with and without hamstrings loaded were compared at all flexion angles using paired t-tests. Also, the trends with and without hamstrings loading were compared between the two methods.

H$_{12a}$: Hamstring loads affect the maximum contact pressures of the patellar cartilage at 40°, 60° and 80° degrees of flexion in the computational study.
H_{12b}: Hamstring loads affect the mean contact pressures of the patellar cartilage at 40°, 60° and 80° degrees of flexion in the computational study.

H_{12c}: Hamstring loads affect the lateral force ratio of the patellar cartilage at 40°, 60° and 80° degrees of flexion in the computational study.

H_{12d}: Hamstring loads affect the patellar flexion at 40°, 60° and 80° degrees of flexion in the computational study.

H_{12e}: Hamstring loads affect the patellar tilt at 40°, 60° and 80° degrees of flexion in the computational study.

H_{12f}: Hamstring loads affect the lateral rotation at 40°, 60° and 80° degrees of flexion in the computational study.

H_{12g}: Hamstring loads affect the lateral shift at 40°, 60° and 80° degrees of flexion in the computational study.

For the finite element model, variations in the additional parameters with and without hamstrings loaded were compared at all flexion angles using paired t-tests.

H_{13a}: Hamstring loads affect the hydrostatic pressures of the patellar cartilage at 40°, 60° and 80° degrees of flexion in the computational study.

H_{13b}: Hamstring loads affect the contact shear forces of the patellar cartilage at 40°, 60° and 80° degrees of flexion in the computational study.

H_{13c}: Hamstring loads affect the octahedral stresses of the patellar cartilage at 40°, 60° and 80° degrees of flexion in the computational study.
1.6.4 Summary of alternate hypotheses

\( H_{11} \): \( \mu_{\text{comp}} \neq \mu_{\text{exp}} \) for all pressure and kinematic parameters between computational and experimental methods.

\( H_{12} \): \( \mu_{\text{hams}} \neq \mu_{\text{nohams}} \) for all pressure and kinematic parameters using the computational method at 40°, 60° and 80° degrees of flexion in the computational study.

\( H_{13} \): \( \mu_{\text{hams}} \neq \mu_{\text{nohams}} \) for all additional parameters using the computational method at 40°, 60° and 80° degrees of flexion in the computational study.
Chapter II

BACKGROUND

2.1 Anatomy of the Knee

Figure 2.1 Anatomy of knee,

Courtesy: Brukner and Khan, Clinical Sports Medicine, 3E,
The main parts of the knee joint are bones, ligaments, tendons, cartilage and the joint capsule. The four bones that make up the knee joint are femur, patella, tibia and fibula. The ligaments give stability to the knee. The medial and lateral collateral ligaments attach the femur to tibia and provide stability to the knee by limiting medial-lateral and varus-valgus motion. The anterior cruciate ligament attaches the femur to tibia to limit its internal rotation and anterior motion. The posterior cruciate ligament limits posterior motion of the tibia. Patellar tendon and the quadriceps tendons are the major tendons in the knee joint. The patellar tendon connects the patella to the tibia. The quadriceps tendon connects the quadriceps muscle to the patella. The two main muscle groups in the knee are the quadriceps and hamstrings.

2.2 Mechanical Role of Patella

![Diagram showing moment arm of the quadriceps (d2>d1).](image-url)
The patella acts as a fulcrum which balances the moments at the patellar tendon and quadriceps's tendon. Patella’s main function is that it increases the moment arm for the extension moment applied by the quadriceps. The moment arm increases with extension of the leg (Fig 2.2). Patella is the site of attachment for the four quadriceps muscles. It integrates the divergent forces of the quadriceps and transfers it to the patellar tendon.

Figure 2.3 Resultant forces on the patella.

The resultant force acting on the patella is the combination of forces in the quadriceps tendon and patellar tendon (Figure 2.3). This resultant force helps to keep the patella in the trochlear groove and it varies with the flexion angle. The resultant patellofemoral force is not distributed uniformly over the lateral and medial facets of the patella. Patella's morphology shows that the subchondral
bone on the lateral patellar facet to have more density when compared to the medial facet suggesting that the lateral facet is subjected to higher forces (Fig 2.4). The articular cartilage at the patellofemoral joint is adapted with hyaline cartilage to bear the high compressive forces with minimal friction. In case there is any damage or loss to the cartilage of the patella, it can result in excessive forces on the patella causing pain.

Figure 2.4 Anterior and posterior view of the patella.

2.3 Function of Quadriceps and Hamstrings

Quadriceps and hamstrings muscle groups help to keep the knee stable and well aligned during motion. The quadriceps muscle group consists of vastus intermedius (VI), vastus lateralis (VL), rectus femoris (RF) and vastus medialis, which can be broken up into vastus medialis obliquus (VMO) and vastus medialis longus (Fig 2.5). The quadriceps primarily are activated during leg's extension movements. This includes activities like walking, running, jumping and squatting. Vastus medialis plays an important role of stabilizing the patella in the trochlear
groove against the lateral forces applied by the other components of the quadriceps and the patellar tendon.

The hamstrings run on the posterior side of the femur. Hamstrings comprise of three muscle groups i.e. semitendinosus, semimembranosus and the biceps femoris (Fig 2.6). Hamstrings act on two joints, hips and knee. They assist in knee flexion and hip extension. Semitendinosus and semimembranosus assist during knee flexion and simultaneously rotate the tibia internally. The biceps femoris has a long and short head which assist during flexion and rotate the tibia externally. The hamstrings play a vital role in all daily activities like walking, running, jumping and squatting. Their main function is to act antagonistically to the quadriceps muscle group during extension.

Figure 2.5 Quadriceps muscle group, courtesy - FCIT
2.4 Patellofemoral Pain and Effect of Hamstring Loading

One of the common causes of patellofemoral pain is lateral malalignment. Quadriceps muscles and the patellar tendon apply a laterally oriented force on the patella which tilts and shifts the patella laterally. Excessive lateral shift and/or lateral tilt of the patella can lead to excessive pressures on the lateral facet of the cartilage which could lead to cartilage degradation. Pain can develop due to overloading of the subchondral bone [3,5]. Alteration in tibiofemoral kinematics can alter the orientation of the patellar tendon and this could lead to patellofemoral disorders. Increase in posterior translation and external rotation of the tibia can contribute to patellofemoral disorders [11, 12]. In previous in vitro studies it has been observed that hamstrings loading causes posterior translation
and external rotation of tibia [13]. This can contribute to lateral malalignment of patella leading to patellofemoral pain.

2.5 Previous Studies

It has been observed that posterior translation of tibia causes the patellar tendon to have a more posterior orientation which has been related to increased patellar flexion and patellofemoral compression [8]. External rotation of the tibia increases lateral orientation of the patellar tendon which has been related to lateral translation of the patella causing elevated patellofemoral pressures [9, 10]. In vitro studies have shown that hamstrings loading causes posterior translation and external rotation of the tibia [14, 15, 16, 17]. This increases patellar flexion and lateral translation of the patella [14]. In another study, hamstring loading increased lateral shift and tilt of the patella which can contribute to lateral malalignment and shift pressures on the lateral facet of the patella, which could lead to overloading of the lateral cartilage [13].

The available biomechanical results indicate that hamstring activation could exacerbate patellofemoral disorders. Thus, characterizing the influence of hamstrings on patellofemoral biomechanics can help improve the treatment techniques for patients with patellofemoral pain due to malalignment.

2.6 Advantages of Finite Element Technique

The in vitro studies involves various issues like tissue expense, condition of specimen before testing and degradation during testing which limits the number of physiological conditions that can be simulated over the length of the study.
Moreover, the knee specimens obtained may not have lateral malalignment. Thus computational modeling helps characterizing multiple physiological conditions and evaluating various surgical procedures with a validated model. It also can quantify additional parameters, which is a key to our new data.

2.7 In- vitro Study

A primary objective of the computational study was to validate the finite element model by replicating the experimental study. The experimental set-up of the in-vitro study defined the modeling parameters of the computational study (Fig. 2.7). The finite element model is validated by comparing the contact parameters obtained from the in-vitro and finite element studies. Lateral pathology was simulated by moving the tibial tuberosity laterally [13] (Fig 2.8). For the lateral position, the average tibial tuberosity-trochlear groove (TT-TG) distance was 19 ± 1 mm, which was approximately 5 mm lateral to the average value in non-symptomatic knees [18, 19].
Figure 2.7 Diagram of the in-vitro experimental setup.

Figure 2.8 In-vitro simulation of lateral pathology
In the in-vitro study, biomechanical testing was carried out to characterize influence of hamstring loading on patellofemoral biomechanics [13]. Ten knees were tested at 40, 60 and 80 degrees of flexion. At each flexion angle, two different loading scenarios were investigated. The two loading conditions were:

1. Loading only the quadriceps muscle group.
2. Loading both the quadriceps and hamstring muscles.

2.7.1 Experimental Setup

![Experimental set-up](image)

Figure 2.9 Experimental set-up to flex the knee to 40, 60 and 80 degrees

The knee was kept fixed on the loading frame which consisted of two hinged plates which helped to set a particular flexion angle (Fig 2.9). The femur was secured on the frame while the tibial extension was controlled by passing a rod through a fixture into the tibia (Fig 2.10 and 2.11). The patella had no motion.
constraints on it. The patella was sufficiently constrained in the trochlear groove for the given range of flexion angles.

Figure 2.10 Experimental set-up to restrict femur translation and rotation

Figure 2.11 Experimental setup to support tibia
2.7.2 Muscle Loading

The knees were first subjected to quadriceps loading only, and then to a combination of quadriceps plus hamstring loading. The straps sutured to the isolated muscle tendons were secured to loading cables that passed over pulleys and were connected to weights (Fig 2.12). The loading cables representing the vastus intermedius (VI)/ rectus femoris (RF) / vastus medialis longus (VML), the vastus lateralis (VL) and the vastus medialis obliqus (VMO) were loaded with dead weights. Based on a combination of EMG determined contributions of each component of the quadriceps to the quadriceps extension moment [20, 21] and computational modeling [22], force distributions representing a weak VMO were determined.

Figure 2.12 Experimental set-up to simulate physiological loading of quadriceps and hamstrings

For a weak VMO, forces were distributed as 27N, 127N and 432N for VMO, VL and VI respectively. A weak VMO was represented due to the prevalence of
VMO weakness for patients with pain [23]. Once the readings were recorded for quadriceps loading, 200N of hamstrings load was applied.

2.7.3 Kinematic and Pressure Measurement

For kinematic measurements, sensors from a pulsed DC magnetic tracking system (trakSTAR, Ascension Technology, Burlington, VT) were used to get the position and orientation of the patella with respect to the femur (Fig 2.13).

![Electromagnetic Sensors](image1.png)

**Figure 2.13 Electromagnetic sensors to track patellofemoral kinematics**

Thin film sensors were used to measure patellofemoral forces and pressures (I-Scan 5051, Tekscan, Boston, MA). The pressure sensor was placed such that it covered the entire patellofemoral contact area (Fig 2.14). The patellar ridge was identified by loading the VMO and VL and recording the pressure pattern while palpating the portion of the ridge that was accessible, with the sensor between the palpating finger and the patella. Thus while recording force and pressure
data, a 5mm wide band was used to characterize the ridge which helped dividing the forces and pressures on the medial and lateral side.

Figure 2.14 Pressure sensors to measure contact pressures at the patellofemoral articulation

2.8 Patellofemoral Joint Coordinate System:

To calculate the patellofemoral rotations and translations before and after hamstring loading, the joint-coordinate system proposed by E S Grood and W J Suntay was used [27]. The advantage of using this joint coordinate system is that clinicians can readily understand it and can directly relate experimental data towards clinical diagnosis. Anatomical landmarks were digitized on the knee in
order to establish patella and femur reference axis with the knee in full extension [28].

The three axes about which rotational motions occurred are used to specify the joint coordinate system (JCS). Figure 2.15 shows JCS applied to two bodies A and B. The fixed axes for bodies A and B along their longitudinal directions are e1 and e3 respectively. An orthogonal axis perpendicular to both e1 and e3 is defined as the floating axis e2. Rotational parameters are quantified by the rotation of the body about these three axes. A reference point PA is defined on body A and a reference point PB is defined on body B. The vector that connects these two points is defined as the translational vector H. The translational parameters are calculated as a dot product of the translational vector H and the three axes of the joint coordinate system.

Figure 2.15 Joint coordinate system applied to two bodies.
2.8.1 Femur Reference Axes:

The transepicondylar axis was established along the most medial and lateral digitized points on the femoral epicondyles as x-axis, positive towards the right. The knee center was positioned midway between the two digitized points. Two points were digitized along the midline of the posterior surface of the femoral diaphysis in order to identify the long axis along the femur. The mutual perpendicular of the transepicondylar axis (x-axis) and the long axis along the femur was the anterior-posterior axis. The anterior-posterior axis was the y-axis with anterior direction positive. The cross product of the transepicondylar axis and anterior-posterior axis, i.e. x and y axis, gave the proximal-distal z-axis, with proximal direction positive (Fig 2.16).

Figure 2.16 Joint coordinate system for measuring patellofemoral kinematics.
2.8.2 Patella Reference Axes:

A similar procedure was used to establish the patellar reference axis (Fig 2.16). One of the reference axes was along the most medial and lateral points on the patella while the second was from the mid-point of the medial-lateral axis to the most distal point on the patella. By characterizing the position and orientation of the patellar and femoral reference axes with respect to the sensors fixed to the respective bones, the position and orientation of the axes were quantified throughout testing.

![Joint coordinate system established for the patellofemoral joint](image)

Figure 2.17 Joint coordinate system established for the patellofemoral joint

The patellofemoral kinematics were determined by the translations and rotations of the patellar reference axis with respect to the femoral axes. Patellar flexion,
lateral rotation, lateral tilt and lateral shift were the rotational and translational parameters obtained using a floating coordinate system [27] (Fig 2.17). The analysis focused on variations in patellofemoral kinematics caused by applying hamstrings forces to a knee already loaded through the quadriceps muscles. Statistical comparisons were performed with paired t-tests at each flexion angle.

2.9 Results from In-vitro Study [13]

Patellar flexion, lateral tilt, lateral rotation and lateral shift of the patella with respect to the femur increased with hamstrings loading (Fig 2.17 and 2.18). The average patellar flexion increased by a minimum of $1.0^\circ$, with the increase statistically significant at each flexion angle (Table 1). The average lateral rotation increased significantly at 80°. Tilt increased significantly at 40° and 60° by approximately $0.5^\circ$. The increase in the average lateral shift was 0.2 mm at 60° and 80°, with the change significant at both flexion angles.

Loading the hamstrings shifted force and pressure towards patella’s lateral facet (Fig. 2.18). Loading the hamstrings increased the average lateral force percentage by at least 5% at each flexion angle, with each change statistically significant. The average maximum lateral pressure increased significantly by 0.3MPa at 40° and 60°.
Table 2.1

Average patellofemoral kinematics for loading with and without hamstrings (hams)

<table>
<thead>
<tr>
<th>Flexion Angle</th>
<th>Flexion (Deg)</th>
<th>Lateral Rotation (deg)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>No Hams</td>
<td>With Hams</td>
</tr>
<tr>
<td>40°</td>
<td>30.1</td>
<td>31.1</td>
</tr>
<tr>
<td>60°</td>
<td>44.5</td>
<td>46.1</td>
</tr>
<tr>
<td>80°</td>
<td>59.7</td>
<td>61.2</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Flexion Angle</th>
<th>Lateral Tilt (Deg)</th>
<th>Lateral Shift (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>No Hams</td>
<td>With Hams</td>
</tr>
<tr>
<td>40°</td>
<td>5.3</td>
<td>5.8</td>
</tr>
<tr>
<td>60°</td>
<td>7.6</td>
<td>8.1</td>
</tr>
<tr>
<td>80°</td>
<td>10.0</td>
<td>10.2</td>
</tr>
</tbody>
</table>

SE of diff: standard error of the mean difference between loading conditions.
Figure 2.18. Patellofemoral pressure distributions for two knees at all three flexion angles with and without hamstrings loading [13].

Table 2.2

Average force and pressure output for loading with and without hamstrings (hams)

<table>
<thead>
<tr>
<th>Flexion Angle</th>
<th>Lateral Force Percentage</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>No Hams</td>
</tr>
<tr>
<td>40</td>
<td>67</td>
</tr>
<tr>
<td>60</td>
<td>71</td>
</tr>
<tr>
<td>80</td>
<td>76</td>
</tr>
</tbody>
</table>

SE of diff: standard error of the mean difference between loading conditions

Thus, the in vitro study was followed by the finite element evaluation of patellofemoral pressures and kinematics. The major stages involved for the FEA
study were 3D reconstruction of knee geometry from MRI images, meshing, creating the FEA model and validation. The validated model will be used for future variations in patellofemoral loading related to abnormal anatomy, injury, physical therapy and surgical treatment not studied experimentally. FEA is also used to compute additional parameters related to hamstrings loading that could not be quantified experimentally.
CHAPTER III

MATERIALS AND METHODS

3.1 Overview

One of the main objectives of the computational study was to replicate the experimental study and validate the FEA model. Subject-specific knee geometry was obtained by 3D reconstruction of MRI scans. The knees were then aligned at 40, 60 and 80 degrees of flexion based on the kinematic data obtained from the in vitro study. Alignment was followed by meshing of the 3D surfaces. The meshed surfaces were used to create finite element models simulating the experimental study. Finite element models were built for three knees. For each flexion angle, the physiological conditions evaluated were:

1. Quadriceps loading only.

2. Quadriceps and Hamstrings loading
Figure 3.1 Flowchart for the study
For validation, the contact parameters were the primary comparison between the experimental and computational study. A correlation analysis was performed to compare the experimental and computational contact data. Additional computational parameters were obtained to characterize the 3D state of stress of the cartilage. Additional parameters include hydrostatic pressures, contact shear forces and octahedral stresses. The flow chart explains the testing procedure (Fig 3.1).

3.2 3D Reconstruction from MRI Scans

The subject-specific knee geometry was obtained from MRI scans (Symphony 1.5T, Siemens, New York, NY). To capture the anatomy the scans were taken in the sagittal plane using a three-dimensional spoiled gradient echo sequence with fat suppression (TR = 55 ms, TE = 5 ms, flip angle = 40°, field of view = 14 cm), with a slice thickness of 1.5 mm [24, 25]. The knee was imaged while fully extended using an extremity coil. Surface models of the bones and cartilage were created from these images from the scan with the leg extended. Triangulated surface geometry representing each bone and cartilage surface was created using medical image reconstruction software (3D-DOCTOR), giving a representation of the extended knee (Fig 3.2 and 3.3). For reconstruction, each part was segmented by tracing the boundary of each part using the boundary tool.
Figure 3.2 3D Reconstruction from MRI scans

Figure 3.3 Extended knee after MRI reconstruction.
3.3 Alignment of Surface Geometries

Initially the knee was in full extension, it had to be aligned at the 40, 60 and 80 degrees of flexion (Fig 3.3). The reconstructed knee was imported in AC3D, Invis where anatomical points were digitized on the femur, patella and tibia (Fig 3.4). These digitized surfaces geometries where exported into MATLAB where they were aligned at different flexion angles by applying inverse kinematics. The tibia was aligned by applying inverse kinematics calculated from the experimental study. The femur was positioned at the origin of virtual space reproducing its experimental alignment on the testing frame [29]. Patella was positioned by flexing and aligning parallel surfaces on patella and trochlear groove. Once the different parts were aligned, they were ready for meshing followed by creating the FE model.

Figure 3.4 Anatomical points digitized on the knee in AC3D
3.4 Meshing

Mesh generation software, TrueGrid (XYZ Scientific Applications), was used to create meshes. Eight-node hexahedral elements and 4-noded quadrilateral elements were used to create the cartilage and bone meshes respectively. Closest Point projection method was used to create the mesh [26]. In this method, a cubic block structure was modified to surround the surface being meshed. The block was then projected on the surface to obtain a mesh. Coupled with the multi-block approach, this method allowed to build quality meshes for complex geometries. A diagnostic check was performed to check the quality of the mesh.

3.4.1 Basic Terminologies

TrueGrid is comprised of physical space and an abstract computational space. The surface to be meshed was imported in the physical space. The final mesh was generated in the physical space. The physical space also contained the physical mesh whose vertices and edges could be manipulated (Fig 3.5). It was the physical mesh that was projected on the surface to obtain the mesh.

The computational space was an abstract space with only integer valued points. It contained a computational mesh which was used to control the physical mesh in the physical space (Fig 3.5).
Figure 3.5 Physical mesh and computational mesh in physical and computational space respectively.

Figure 3.6 Physical mesh blocked to fit the patellar cartilage
3.4.2 Blocking

There was a correspondence between every node of the computational mesh and the corners of the physical mesh, usually referred to as vertices or control points, of the mesh. The physical mesh in the physical space was moved to fit the surface being meshed (Fig 3.6). The computational mesh did not move in the computational space but it served as a convenient way of identifying regions of the physical mesh which were the object of various functions. Surfaces in the physical space had three coordinates named $x$, $y$, and $z$. Computational mesh in the computational space had three coordinates or indices named $i$, $j$, and $k$. Mathematically, a mesh was a vector function that the computational mesh mapped to the physical mesh in physical space. The physical mesh for cartilages had three divisions along its thickness to have three layer of elements along the thickness. While for the bones, the internal blocks were deleted.

3.4.3 Closest-Point Projection Technique

Figure 3.7 Projection of computational block on the patellar cartilage surface in physical space.
Blocking simplified the mesh generation because it automatically filled each block with the required elements. The input to be specified for generating a mesh were the pair of opposite exterior faces of the physical mesh that were to be projected, and the surface being meshed (Fig 3.7). Thus, the physical mesh block was deformed to the required shape i.e. shape of the surface. However, this was a crude mesh. In the projection method, nodes of the physical mesh were placed on the surfaces. When a node of the physical mesh was projected onto a surface, it was moved to the point on the surface which was closest to the initial position of the node. The line connecting the original node with its projection onto the surface was orthogonal to the tangent plane of the surface at the point of projection.

3.4.4 Linearizing along Thickness

In the crude mesh, edges of the block along the thickness were divided with considerable inconsistency. The blocks were linearized so the edges of the elements of the three layers along thickness of the cartilage were similar in length along the thickness (Fig 3.8).

![Figure 3.8 Patellar cartilage block edges, before and after linearization.](image-url)
3.4.5 Seeding

The crude mesh was seeded by specifying the direction, i.e. i, j or k, and the number of divisions in which each block was to be divided. Seeding decided the mesh density in each block. For seeding, 'mesq' command was used which was followed by the direction, i, j or k. The direction was followed by the seeding information (Fig 3.9).

![Figure 3.9 Seeding of the patellar cartilage physical mesh block.](image)

3.4.6 Uniform

The mesh was uniformed to have smooth distribution and transition between elements. The final mesh was obtained after uniform distribution of the mesh elements (Fig 3.10).
3.4.7 Different Blocking Techniques

The computational mesh used can be a cube in structure or have a 'butterfly' shape (Fig. 3.11). For femur cartilage a cube was used while for femur and patella the internal blocks of the cube were deleted. Patellar cartilage was
blocked using the butterfly technique. In this approach the edge row elements were deleted and the adjacent surface were projected on each other (Fig 3.11).

Three layers of eight-node linear brick elements were used for the deformable cartilage meshes. Four-node 3-D bilinear rigid quadrilateral element were used for the rigid bone meshes (Fig 3.12).

Figure 3.12 Deformable hexahedral elements for the cartilage and rigid quadrilateral elements for the bones.

The orthogonality and volume of the mesh was checked to obtain a good quality mesh. The volume of the elements should always be positive and greater than zero and the angle between the adjacent faces should be close to 90 degrees. A '.inp' file was created which contained the node and element information of the
mesh. The '.inp' files were exported to ABAQUS/Std, Simulia to create the model.

3.5 Finite Element Modeling

The meshes were exported from the mesh generation software into ABAQUS. These input files contained node and element information. The node information comprised of the coordinates while the element information described the connectivity of the nodes. Once the mesh files were obtained, each part-mesh was aligned to the flexion angle being analyzed using the iterative closest point (ICP) algorithm\[30\]. The ICP algorithm performs aligning by shape matching the mesh to the surface which was aligned after applying inverse kinematics.

The three flexion angles at which the knees were analyzed were 40, 60 and 80 degrees. All finite element analysis were non-linear static analysis. Since each surface geometry had a separate input file, an input script was written to include '.inp' files of each part and import all the meshes into one model. The input script was imported into ABAQUS where the model was built further.
3.5.1 Modeling - Individual Parts

3.5.1.1 Rectangular Rigid Planar Shell part

An additional 3-D discrete rigid rectangular planar shell part was created which provided attachment points for the quadriceps muscle group (Fig 3.14). The dimensions of the part were 20 x 10 mm with planar thickness 1 mm. Partitions were created on this rectangular shell part so spring attachment points are distributed evenly. The partitions were created on the sketch face. The part was meshed in ABAQUS using four-node 3D bilinear rigid quadrilateral elements.
The concentrated loads were applied on these shell rigid bodies which were connected to the patella via tension only springs. These springs represent different muscle bundles of the quadriceps.

3.5.1.2 Rigid-body Reference Points

Patella, femur and the shell rigid body were the only 3-D rigid body parts of the model. A rigid body reference point was assigned to each rigid body part (Fig 3.14 and Fig 3.15). The boundary and loading conditions were applied at the rigid body reference points for the rigid body parts.
Figure 3.15 Rigid body reference points on femur and patella.

3.5.1.3 Node and Surface Element Sets

Figure 3.16 Node sets on patellar and femur cartilage to apply constraints.
Node sets were assigned on the cartilages for applying constraints (Fig 3.16), while for applying interaction properties surface element sets were created (Fig 3.17).

![Figure 3.17 Surface elements set for applying interaction properties.](image)

### 3.5.2 Cartilage Properties

The cartilage properties were chosen to represent its nearly steady state response to isometric loading recorded experimentally, rather than the instantaneous response. A material property for the cartilages had a Young's Modulus 5MPa and Poisson's ratio 0.45[31, 32]. The cartilage material was treated as linear, elastic and isotropic.
3.5.3 Creating an Assembly

In the Assembly module, we instance each part as either dependent or independent. By default, Abaqus/CAE creates a dependent instance of a part (Fig 3.18).

3.5.3.1 Instances of a Part

![Figure 3.18](image)

Figure 3.18 A basic assembly of instances of all parts

A dependent instance is only a pointer to the original part. The changes that are made to the part are reflected on the part instance. Certain changes cannot be made to the dependent instance which would affect its geometry. However, boundary conditions, loading conditions and interaction properties can be applied to the instances. The advantages of dependent part instances were that they consumed fewer memory resources and the part needed to be meshed only once.
A mesh which is imported into ABAQUS from a pre-processor and does not have surface geometry for reference under part module is called an orphan mesh. An orphan mesh is always instanced as dependent. All the orphan meshes in the current model were instanced as dependent.

3.5.3.2 Creating Muscle Attachment and Origin Points

Once the parts were instanced, origin and attachment points were created for the patellar tendon, medial patellomeniscal ligament and the quadriceps muscle group, i.e. vastus intermedius (VI), vastus medialis obliquus (VMO) and vastus lateralis (VL). These points were obtained based on anatomical features of the knee and attachment maps created manually for the three knees that were imaged to create the FEA models (Fig 3.19, 3.20, 3.21).

Figure 3.19 VI and VMO attachment points on the patella
Figure 3.20 VI and VL attachment points on the patella

Figure 3.21 Attachment points for the patellar tendon and MPML
3.5.3.3 Springs to Represent Muscle, Tendon and Ligaments

The three shell rigid body parts were positioned at the reference points representing the origin of each muscle force at the pulley (Fig 3.18). The position of the pulleys from the experimental set-up was determined from the 3-D reconstruction of the experimental setup. The three sets of 10 linear, tension-only springs were created to represent Vastus Intermedius, Vastus Medialis Obliquus and Vastus Lateralis.

Figure 3.22 Springs representing VI, VMO and VL
These muscle-springs were attached from the patella to the three shell rigid body parts (Fig 3.22). The springs representing the quadriceps tendon were given a stiffness of 1350 N/mm [33].

A set of five, four and three springs each was created to represent the patellar tendon, medial patellofemoral ligament (MPFL) and medial patellomeniscal ligament (MPML) (Fig 3.23). The springs used were of type Spring-A. Spring-A were connected between two nodes where line of action was the line joining the two nodes. The patellar tendon, MPFL and MPML were also modeled as linear tension-only springs with stiffness values of 2000 N/mm, 12 N/mm and 5N/mm respectively[34].

Figure 3.23 Springs representing patellar tendon, MPML and MPFL
3.5.3.4 Local Coordinate Systems for Muscles

Three local coordinate systems were created with the x-axis oriented from the muscle insertion point on the patella to the origin point on the pulley, for the three muscle groups. These were rectangular coordinate systems created using the three point technique. They were named VI_csys, VMO_csys and VL_csys (Fig 3.25). These were used for applying loads along the line joining from the insertion point to the origin for the muscles.
3.5.4 Modeling Contact Interaction Properties

To define contact interaction, surface element sets on the patellar and femur cartilage were picked. The coefficient of friction at contact was 0.02. The normal behavior on contact was defined as 'hard contact'. Finite-sliding formulation was used with surface to surface discretization.

3.5.5 Modeling - Constraints

In the model, the cartilages and bones had different surface definitions. To allow the motion of bones and their respective cartilage together, the back surface of the cartilage was given a MPC (Multi Point Constraint) constraint.
Figure 3.26 Contact Interaction properties applied to the surface elements of the cartilages.

The MPC was of type 'Beam'. The constraint provides a rigid beam between two nodes to constrain the displacement and rotation at the first node to the displacement and rotation at the second node, corresponding to the presence of a rigid beam between the two nodes [36]. The first node was the Master/Control point and the node selected was the rigid body reference point of the bones (Fig 3.27 and Fig 3.29). The second set of nodes were the slave nodes which comprised of the nodes on the back surface of the cartilages (Fig 3.28 and Fig 3.30). Thus the cartilage moved with the bone i.e. they had same degrees of freedom.
Figure 3.27 Control point for the MPC beam constraint on the patella

Figure 3.28 Slave node set for the MPC beam constraint on the patellar cartilage
Figure 3.29 Control point for the MPC beam constraint on the femur

Figure 3.30 Slave node set for the MPC beam constraint on the femur cartilage
3.5.6 Boundary and Loading Conditions

Boundary conditions were applied to the model to replicate the experimental study. All the boundary conditions applied were of 'displacement/rotation' type. The femur was kept fixed on the loading frame throughout the in-vitro study while the patella had no restriction of motion. Therefore, a boundary condition was applied to the femur's rigid body reference point, which restricted its motion in all six degrees of freedom (Fig 3.31).

The patella was given a boundary condition to be displaced during the first step of the analysis to bring the two cartilages into contact. The patella was translated during the displacement step by the same magnitude as it was
translated away to create a clearance between the two cartilages (Fig. 3.34). When loads were applied on the knee, the patella was free to move in all degrees of freedom (Fig 3.32).

Figure 3.32 Boundary condition applied to patella's rigid body reference point.

A boundary condition was applied to the tibia-end of patellar tendon and MPML attachment points to restrict motion through the analysis (Fig 3.33). All the above boundary conditions were applied in the global coordinate system.
Figure 3.33 Boundary condition applied to tibial end of patellar tendon and MPML

In order to allow loads to be applied along the line of muscle insertion on the patella to the origin point on the pulley, a boundary condition was applied on each shell rigid body part in their respective rectangular local coordinate system i.e. VL_csys, VMO_csys and VL_csys (Fig 3.25 and 3.34). The x-axis of each local coordinate system was oriented from the muscle insertion point on the patella to the origin point on the pulley. The shell rigid bodies were fixed in space during the first displacement step while during loading they were restricted to move only in the x-direction of the local coordinate system. The muscle loads were applied on the shell rigid part instances in their respective local coordinate system (Fig 3.30). In order to represent a weak VMO force, a 27N concentrated
force was used for VMO while 432 N and 127 N force was used for VI and VL respectively. The loads applied were same in the in-vitro study.

Figure 3.34 Boundary conditions and loads applied at the origin of each muscle group.

3.5.7 Output Requests from Model after Analysis

Field and history output requests were made to extract data from the model. Field output requests were created to obtain stresses, displacements, rotations, strains, contact pressure, contact force and concentrated forces throughout the model. History output requests were made to obtain the total
contact area, forces in the tendon and ligaments and center of total force due to contact pressure.

3.5.8 Steps of Analysis

The entire analysis was divided into three general static stress-analysis steps. The first step was the displacement step and the concentrated loads were applied in the second and the third step.

A clearance was created between the cartilages in cases where excessive over-closure was observed because it prevented the convergence of the analysis (Fig 3.35). Thus a displacement step was required before applying the loads.

The time period for each step was unity and no automatic stabilizations were used. Non-linearity of the model was accounted for in each step. The first
step involved displacing the patella and bringing it in contact with the femur cartilage. The application of force was distributed over the next two steps due to convergence reasons (Fig 3.36).

![Step Loading](image)

Figure 3.36 Step-loading through the analysis

3.6 Convergence Analysis

A convergence analysis was performed to confirm that a fine enough element discretization was used. The results for the analysis were recorded for a particular mesh discretization. The same analysis was performed using finer mesh, i.e. more number of elements, and the results were compared to the previous analysis. Mesh discretization was changed for the cartilages. Maximum
contact pressure was the parameter compared for the convergence analysis since this was the parameter that was compared to the experimental data. Based on the convergence analysis, approximately 21,000 cartilage elements were used for every analysis (Fig 3.33).

![Convergence Analysis Graph]

Figure 3.37 Convergence graph

3.7 Statistics

A correlation and regression analysis was performed accounting for variations due to anatomy of knees, flexion angle and loading conditions to understand the relationship between the finite element and experimental technique. The correlation and regression was tested for the pressure parameters, maximum contact pressure, mean contact pressure and lateral force ratio, and kinematic
parameters, patellar flexion, patellar tilt, lateral rotation and lateral shift. The means and standard deviations were computed for both pressure and kinematic parameters averaged across each flexion angle to test the ability of the model to differentiate between knee anatomies. The means and standard deviation for each knee were compared between the computational and experimental method to observe the variations between knees for the two testing methods. Two-tailed paired t-tests with 95% significance were used to study the variations in pressures and kinematics due to hamstrings loading in the finite element technique. The influence of hamstrings on pressure parameters; maximum contact pressure, mean contact pressure and lateral force ratio, and kinematic parameters; patellar flexion, patellar tilt, lateral rotation and lateral shift were statistically compared with t-tests for both experimental and computational technique. Statistical results from paired t-tests for hamstrings loading were discussed for the finite element method. The trends for the pressure and kinematic parameters due to hamstrings loading were observed between the finite element and experimental technique. The focus was on the variation due to hamstrings loading and not the absolute values obtained. The influence of hamstrings loading on the additional parameters, hydrostatic pressures, octahedral stresses and contact shear forces was statistically tested using paired t-test with significance of 95 %.
CHAPTER IV

RESULTS

4.1 Overview of Results

This chapter presents the validation of the trends predicted by the finite element model by comparison of the patellofemoral pressures and kinematics obtained computationally and experimentally. Additional parameters were obtained from the FE model to characterize the 3D stress-state of the cartilage. Mean contact pressure, maximum contact pressure and lateral force ratios were characterized to study the influence of hamstrings loading on cartilage pressures. The alignment of patella with respect to femur was presented as patellar flexion, lateral tilt, lateral rotation and lateral shift. The additional parameters were hydrostatic pressures, octahedral stresses and contact shear force. Finite element models were created for three knees. A correlation and regression analysis was performed to establish the relation between the two methods. The comparison between the experimental and computational results for variations
with hamstrings loaded showed a significant relationship for maximum contact pressure, mean contact pressure and patellar flexion. For patellar cartilage, though the hydrostatic pressures increased with hamstrings loading (fig) but variations were not significant (Table 4.7). Contact Shear force increased significantly with hamstrings loading at 40 degrees of flexion (Table 4.7). Octahedral stresses did not show significant increase with hamstrings loading (table 4.7). For femur cartilage variations due to hamstrings loading were not significant for additional parameters.

4.2 Results for Pressure Parameters

The FE model was validated by comparing variations in pressures on hamstrings loading between the computational and experimental study. The focus will be on variations with hamstrings loading predicted by FE model and not on the absolute values.

4.2.1 Correlation and Regression Statistics for Pressure Parameters to Validate the Trends Predicted by the Model with Hamstrings Loading.

This analysis was performed to test the ability of the FE model to account for variations in anatomy, loading conditions and flexion angles while predicting pressures.
Figure 4.1 The scatter plot indicating the correlation for mean contact pressure between the FEA and experimental study, accounting for variations due to loading conditions, flexion angle and anatomical variations between knees.

Figure 4.2 The scatter plot indicating the correlation for maximum contact pressure between the FEA and experimental study accounting for variations due to loading conditions, flexion angle and anatomical variations between knees.
Figure 4.3 The scatter plot indicating the correlation for lateral force ratio between the FEA and experimental study accounting for variations due to loading conditions, flexion angle and anatomical variations between knees.

Table 4.1. Correlation and regression statistics for pressure parameters between the experimental and computational method.

<table>
<thead>
<tr>
<th>Pressure Parameters</th>
<th>R Square</th>
<th>F value</th>
<th>Significance F</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean Contact Pressure</td>
<td>0.63</td>
<td>27.40</td>
<td>0.0001</td>
</tr>
<tr>
<td>Maximum Contact Pressure</td>
<td>0.42</td>
<td>11.52</td>
<td>0.0037</td>
</tr>
<tr>
<td>Lateral Force Ratio</td>
<td>0.0012</td>
<td>0.019</td>
<td>0.8920</td>
</tr>
</tbody>
</table>

The coefficient of determination for mean contact pressure and maximum contact pressure were 0.63 and 0.42 respectively (Table 4.1 and Fig 4.1 & 4.2). The relationship between the FE model and the experimental study was significant for
mean contact pressure and maximum contact pressure. The relationship was not significant for lateral force ratio (Table 4.1 and Fig 4.3).

4.2.2 The pressure parameters were averaged across the three flexion angles with hamstrings loaded to examine how accurately the finite element model distinguishes differences between knee anatomy (Table 4.2 and Fig 4.1, 4.2, 4.3). The mean contact pressure and maximum contact pressure averaged over three flexion angles was highest for knee 1 and lowest for knee 3 for both experimental and finite element study (Table 4.2 and Fig 4.1, 4.2). The table 4.2 and figure 4.3 shows lateral force ratio for finite element model and experimental study with absolute values in the same range but less variation between testing conditions.

Table 4.2 Pressure parameters averaged across 40 60 and 80 degrees of flexion for hamstrings loaded condition.

<table>
<thead>
<tr>
<th>(Mean ± Std Dev)</th>
<th>Knee 1</th>
<th>Knee 2</th>
<th>Knee 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean Contact Press</td>
<td>Exp</td>
<td>1.07±0.05</td>
<td>0.89±0.16</td>
</tr>
<tr>
<td></td>
<td>FEA</td>
<td>2.02±0.14</td>
<td>1.49±0.07</td>
</tr>
<tr>
<td>Max Contact Press</td>
<td>Exp</td>
<td>3.3±0.85</td>
<td>3.11±0.67</td>
</tr>
<tr>
<td></td>
<td>FEA</td>
<td>3.95±0.59</td>
<td>2.91±0.19</td>
</tr>
<tr>
<td>Lat force ratio</td>
<td>Exp</td>
<td>0.76±0.05</td>
<td>0.73±0.06</td>
</tr>
<tr>
<td></td>
<td>FEA</td>
<td>0.64±0.03</td>
<td>0.70±0.07</td>
</tr>
</tbody>
</table>
Figure 4.4 Comparing trends between experimental and finite element model for each knee for maximum contact pressures averaged over 40, 60 and 80 degrees of flexion with hamstrings loaded.

Figure 4.5 Comparing trends between experimental and finite element model for each knee for mean contact pressures averaged over 40, 60 and 80 degrees of flexion with hamstrings loaded.
Figure 4.6 Comparing trends between experimental and finite element model for each knee for lateral force ratio averaged over 40, 60 and 80 degrees of flexion with hamstrings loaded.

Figure 4.7 Comparing the trends at 40, 60 and 80 degrees of flexion between loading conditions, quadriceps and hamstrings loading and quadriceps-only loading, using experimental and computational methods for maximum contact pressure averaged across all three knees.
4.2.3 The pressure parameters were averaged over all three knees at each flexion angle to understand the effect of quadriceps and hamstrings loading versus quadriceps-only loading for experimental and computational methods (Fig 4.4, 4.5, 4.6 and 4.7).

The magnitude of mean contact pressure is obtained as the ratio of total contact force to total contact area. The magnitude of mean contact pressures are higher for the finite element model because the total contact area is lower and total contact force is higher in the finite element model compared to the experimental results. The difference in terms of percentage is higher for the total contact area

![Figure 4.8 Comparing the trends at 40, 60 and 80 degrees of flexion between both loading conditions, quadriceps and hamstrings loading and quadriceps-only loading, using experimental and computational methods for mean contact pressure averaged across all three knees.](image)

Mean Contact Pressure
Avg hams vs nohams for 3 Knees

- hams exp
- no hams exp
- hams fea
- no hams fea

Mean Contact Pressure (MPa)
Flexion Angle

0 0.5 1 1.5 2 2.5

40 60 80
Figure 4.9 Comparing the trends at 40, 60 and 80 degrees of flexion between both loading conditions, quadriceps and hamstrings loading and quadriceps-only loading, using experimental and computational methods for lateral force ratio averaged across all three knees.

Table 4.3 Paired t-tests for hams vs. no hams between FEA and In-vitro methods (p<0.05)

<table>
<thead>
<tr>
<th></th>
<th>40</th>
<th>60</th>
<th>80</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Mean Contact Press</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Exp</td>
<td>0.191</td>
<td>0.006</td>
<td>0.001</td>
</tr>
<tr>
<td>FEA</td>
<td>0.059</td>
<td>0.0102</td>
<td>0.293</td>
</tr>
<tr>
<td><strong>Max Contact Press</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Exp</td>
<td>0.881</td>
<td>0.026</td>
<td>0.309</td>
</tr>
<tr>
<td>FEA</td>
<td>0.048</td>
<td>0.221</td>
<td>0.299</td>
</tr>
<tr>
<td><strong>Lateral Force Ratio</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Exp</td>
<td>0.056</td>
<td>0.001</td>
<td>0.001</td>
</tr>
<tr>
<td>FEA</td>
<td>0.897</td>
<td>0.247</td>
<td>0.214</td>
</tr>
</tbody>
</table>

The table 4.3 shows the p-values for comparison between the two loading conditions for both experimental and finite element method. The results were
significant with hamstrings loading for finite element technique for maximum contact pressure at 40 degrees of flexion and for mean contact pressure at 60 degrees of flexion.

4.3 Results for Kinematic Parameters:

The FE model was validated by comparing variations in kinematics on hamstrings loading between the computational and experimental study. The focus will be on variations with hamstrings loading predicted by FE model and not on the absolute values.

4.3.1 Correlation and regression statistics for kinematic parameters to validate the trends predicted by the model with hamstrings loading.

![Figure 4.10 The scatter plot indicating the correlation for patellar flexion between the FEA and experimental study accounting for variations due to loading conditions, flexion angle and anatomical variations between knees.](image)

\[ y = 0.9741x - 1.3699 \]
\[ R^2 = 0.9597 \]
Figure 4.11 The scatter plot indicating the correlation for patellar tilt between the FEA and experimental study accounting for variations due to loading conditions, flexion angle and anatomical variations between knees.

Figure 4.12 The scatter plot indicating the correlation for lateral shift between the FEA and experimental study accounting for variations due to loading conditions, flexion angle and anatomical variations between knees.
Figure 4.13 The scatter plot indicating the correlation for valgus/lateral rotation of patella between the FEA and experimental study accounting for variations due to loading conditions, flexion angle and anatomical variations between knees.

Table 4.4 Correlation and regression statistics for pressure parameters between the experimental and computational method.

<table>
<thead>
<tr>
<th>Kinematic Parameters</th>
<th>R square</th>
<th>F value</th>
<th>Significance F</th>
</tr>
</thead>
<tbody>
<tr>
<td>Patellar Flexion</td>
<td>0.96</td>
<td>380.5775</td>
<td>0.0001E-8</td>
</tr>
<tr>
<td>Valgus/ Lateral Rotation</td>
<td>0.72</td>
<td>40.50376</td>
<td>0.0009E-02</td>
</tr>
<tr>
<td>Lateral Shift</td>
<td>0.29</td>
<td>6.493661</td>
<td>0.0215</td>
</tr>
<tr>
<td>External Rotation</td>
<td>0.04</td>
<td>0.6961</td>
<td>0.4164</td>
</tr>
</tbody>
</table>

There was a positive correlation for patellar flexion and external rotation. The regression was significant for patellar flexion. Lateral shift and lateral rotation were not predicted as accurately as the other kinematic parameters, in part due to the error associated with reproducing the coordinate systems digitized on the
knees experimentally on the finite element models. A significant negative correlation for lateral shift and valgus rotation indicates that the model does not accurately predict changes in lateral shift and rotation due to variations in flexion angle, anatomy and hamstrings and loading conditions. There is a possibility that the FE model is not accurately predicting the change in kinematics caused by the flexion angle.

4.3.2 The kinematic parameters were averaged across the three flexion angles with hamstrings loaded to examine how accurately the finite element model distinguishes differences between knee anatomy (Table 4.2 and Fig 4.1, 4.2, 4.3)

The patellar flexion averaged over three flexion angles was highest for knee 2 for both experimental and finite element study (Table 4.5 and Fig 4). The table 4.5 and figure 4 shows external rotation for finite element model and experimental study with knee3 minimum for both methods.

Lateral shift and lateral rotation were not predicted as accurately as the other kinematic parameters, in part due to the error associated with reproducing the coordinate systems digitized on the knees experimentally on the finite element models (Fig 4.10 and 4.11).
Table 4.5 Kinematic parameters averaged across 40, 60 and 80 degrees of flexion for hamstrings loaded condition.

<table>
<thead>
<tr>
<th>(Mean ± Std Dev)</th>
<th>Knee 1</th>
<th>Knee 2</th>
<th>Knee 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Patellar Flexion</td>
<td>Exp</td>
<td>41.19±8.66</td>
<td>60.5±8.61</td>
</tr>
<tr>
<td></td>
<td>FEA</td>
<td>40.71±7.6</td>
<td>58.54±7.51</td>
</tr>
<tr>
<td>Patellar Tilt</td>
<td>Exp</td>
<td>8.77±0.48</td>
<td>5.23±1.3</td>
</tr>
<tr>
<td></td>
<td>FEA</td>
<td>6.17±1.91</td>
<td>11.61±0.44</td>
</tr>
<tr>
<td>Lateral Shift</td>
<td>Exp</td>
<td>0.59±0.7</td>
<td>6.97±2.18</td>
</tr>
<tr>
<td></td>
<td>FEA</td>
<td>13.26±1.29</td>
<td>5.03±1.01</td>
</tr>
<tr>
<td>Lateral Rotation</td>
<td>Exp</td>
<td>13.19±0.31</td>
<td>-8.19±1.29</td>
</tr>
<tr>
<td></td>
<td>FEA</td>
<td>-1.89±0.13</td>
<td>-0.24±0.31</td>
</tr>
</tbody>
</table>

Figure 4.14 Comparing trends between experimental and finite element model for each knee for patellar flexion averaged over 40, 60 and 80 degrees of flexion with hamstrings loaded.
Figure 4.15 Comparing trends between experimental and finite element model for each knee for patellar tilt averaged over 40, 60 and 80 degrees of flexion with hamstrings loaded.

Figure 4.16 Comparing trends between experimental and finite element model for each knee for lateral shift averaged over 40, 60 and 80 degrees of flexion with hamstrings loaded.
Figure 4.17 Comparing trends between experimental and finite element model for each knee for lateral rotation averaged over 40, 60 and 80 degrees of flexion with hamstrings loaded.

4.3.3 The kinematic parameters were averaged over all three knees at each flexion angle to understand the effect of quadriceps and hamstrings loading versus quadriceps-only loading for experimental and computational methods (Fig 4.4, 4.5, 4.6 and 4.7)
Figure 4.18 Patellar flexion averaged over all three knees at each flexion angle for both loading conditions, quadriceps and hamstrings loading and quadriceps-only loading, using experimental and computational method.

Figure 4.19 Lateral tilt or external rotation averaged over all three knees at each flexion angle for both loading conditions, quadriceps and hamstrings loading and quadriceps-only loading, using experimental and computational method.
Figure 4.20 Lateral or valgus rotation averaged over all three knees at each flexion angle for both loading conditions, quadriceps and hamstrings loading and quadriceps-only loading, using experimental and computational methods.

Figure 4.21 Lateral shift averaged over all three knees at each flexion angle for both loading conditions, quadriceps and hamstrings loading and quadriceps-only loading, using experimental and computational method.
Table 4.6 Paired t-tests for hams vs. no hams between FEA and In-vitro for kinematic parameters (p<0.05).

<table>
<thead>
<tr>
<th></th>
<th>40</th>
<th>60</th>
<th>80</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Patellar Flexion</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Exp</td>
<td>0.452</td>
<td>0.014</td>
<td>0.013</td>
</tr>
<tr>
<td>FEA</td>
<td>0.045</td>
<td>0.016</td>
<td>0.141</td>
</tr>
<tr>
<td><strong>Lateral Tilt</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Exp</td>
<td>0.858</td>
<td>0.103</td>
<td>0.027</td>
</tr>
<tr>
<td>FEA</td>
<td>0.221</td>
<td>0.011</td>
<td>0.360</td>
</tr>
<tr>
<td><strong>Lateral Rotation</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Exp</td>
<td>0.229</td>
<td>0.335</td>
<td>0.004</td>
</tr>
<tr>
<td>FEA</td>
<td>0.423</td>
<td>0.530</td>
<td>0.481</td>
</tr>
<tr>
<td><strong>Lateral shift</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Exp</td>
<td>0.536</td>
<td>0.01</td>
<td>0.076</td>
</tr>
<tr>
<td>FEA</td>
<td>0.562</td>
<td>0.189</td>
<td>0.246</td>
</tr>
</tbody>
</table>

Variation with hamstrings loading was significant for patellar flexion with the FE method at 40 and 60 degrees of flexion. Variation for lateral tilt was significant at 60 degrees of flexion (Table 4.6).

For kinematic parameters, the absolute values are not predicted with the same accuracy as pressure values, in part due to the error associated with reproducing the coordinate systems digitized on the knees experimentally on the finite element models.

For kinematic parameters, the finite element model accurately accounts for the variations due to hamstrings loading but not the variations due to anatomy and flexion angle.
4.4 Additional pressure parameters computed for patellar cartilage

Figure 4.22 Patellar cartilage hydrostatic pressures averaged for three knees at each flexion angle, for quadriceps and hamstrings loading and quadriceps-only loading obtained from the computational model.

Figure 4.23 Patellar cartilage contact shear force averaged over three knees at each flexion angle for quadriceps and hamstrings loading and quadriceps-only loading obtained from the computational model.
Figure 4.24 Patellar cartilage octahedral stresses averaged over three knees at each flexion angle for quadriceps and hamstrings loading and quadriceps-only loading obtained from the computational model.

Table 4.7 Paired t-tests between hams vs. no hams for additional parameters obtained for the patellar cartilage using the finite element technique (p<0.05)

<table>
<thead>
<tr>
<th>For Patellar Cartilage</th>
<th>40</th>
<th>60</th>
<th>80</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hydrostatic Pressures</td>
<td>0.36</td>
<td>0.2</td>
<td>0.3</td>
</tr>
<tr>
<td>Contact Shear Forces</td>
<td>0.01</td>
<td>0.91</td>
<td>0.26</td>
</tr>
<tr>
<td>Octahedral Stresses</td>
<td>0.23</td>
<td>0.85</td>
<td>0.57</td>
</tr>
</tbody>
</table>

Although hydrostatic pressures increased with hamstrings loading (fig) but variations were not significant (Table 4.7). Contact Shear force increased significantly with hamstrings loading at 40 degrees of flexion (Table 4.7). Octahedral stresses did not show significant increase with hamstrings loading (table 4.7)
4.5 Additional Parameters Computed for Femur Cartilage.

Figure 4.25 Femur cartilage hydrostatic pressures averaged over three knees at each flexion angle for quadriceps and hamstrings loading and quadriceps-only loading obtained from the computational model.

Figure 4.26 Femur cartilage contact shear force averaged over three knees at each flexion angle for quadriceps and hamstrings loading and quadriceps-only loading obtained from the computational model.
Figure 4.27 Femur cartilage octahedral stresses averaged over three knees at each flexion angle for quadriceps and hamstrings loading and quadriceps-only loading obtained from the computational model.

Table 4.8 Paired t-tests between hams vs. no hams for additional parameters obtained for the femur cartilage using the finite element technique (p<0.05)

<table>
<thead>
<tr>
<th>Parameter</th>
<th>40</th>
<th>60</th>
<th>80</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hydrostatic Pressures</td>
<td>0.08</td>
<td>0.2</td>
<td>0.29</td>
</tr>
<tr>
<td>Contact Shear Forces</td>
<td>0.71</td>
<td>0.19</td>
<td>0.19</td>
</tr>
<tr>
<td>Octahedral Stresses</td>
<td>0.64</td>
<td>0.99</td>
<td>0.2</td>
</tr>
</tbody>
</table>

The hydrostatic pressures increased with hamstrings loaded at 40, 60 and 80 degrees of flexion (fig) but the variations were not significant (Table 4.8). Variations due to hamstrings loading were not significant for additional parameters.
4.6 Summary of Results

The correlation and regression analysis indicate that there is a significant relationship between the finite element and experimental method for predicting mean contact pressure and maximum contact pressure. The average of maximum contact pressure and mean contact pressure over three flexion angles was maximum for knee 1 and minimum for knee 2. The average of pressure parameters across the three knees showed significant (p<0.05) variations with hamstrings loading for the finite element model for maximum contact pressure at 60 degrees of flexion.

For kinematic parameters, patellar flexion and patellar tilt (tilt not significant) had a positive correlation between the experimental and finite element method. The correlation was significant for predicting the patellar flexion. The patellar flexion averaged over three flexion angles was highest for knee 2 for both experimental and finite element study. External rotation for knee 3 was minimum for both finite element model and experimental study. The variations with hamstrings loading using finite element method were significant for patellar flexion at 40 and 60 degrees of flexion. Patellar tilt was significant on hamstrings loading at 60 degrees of flexion using the finite element method. Lateral shift and lateral rotation were not predicted as accurately as the other kinematic parameters, in part due to the error associated with reproducing the coordinate systems digitized on the knees experimentally on the finite element models. For kinematic parameters, the finite element model accurately accounts for the variations due to hamstrings loading but not the variations due to anatomy and flexion angle.
The additional parameters computed were hydrostatic pressures, contact shear forces and octahedral stresses for patellar and femur cartilage. The variation due to hamstrings loading was tested using paired t-test ($p<0.05$) for the additional parameters. For patellar cartilage, contact shear forces increased significantly with hamstrings loading at 40 degrees of flexion. For femur cartilage, there was no significant change in the additional parameters due to hamstrings loading.
CHAPTER V

DISCUSSION

There are no published studies with validated finite element models characterizing the influence of hamstring loading. The model was validated by testing its correlation and regression with the experimental technique while accounting for variations due to anatomy of knees, flexion angle and loading conditions. For pressure parameters, the finite element model accurately accounted for variations due to anatomy, flexion angles and loading conditions but for kinematic parameters, the finite element model accurately accounts for the variations due to hamstrings loading only and not the variations due to anatomy and flexion angle. Additional parameters obtained computationally help to understand the 3D stress state of the cartilage.

There have been other finite element models which have been validated based on kinematic parameters but there are no finite element models comparing experimental and computational results. S. Farrokhi et al. [38] created patient-specific finite element models to compare the patellofemoral joint stresses in
subjects with and without patellofemoral pain. They validated the finite element model by comparing the estimated contact area and final patella position predicted by the finite element model with the actual contact area and final patella position measured from the loaded MRI images [38, 39]. A patellofemoral model was developed by Baldwin et. al where the model was validated by comparing the kinematics from the FE model to the experimental six-degree-of freedom kinematic data from knee cadaver specimens tested using a knee simulator [40].

5.1 Pressure Results

The correlation analysis and regression statistics was performed which took into account variations due to anatomy of each knee, loading conditions and flexion angle. It indicated that there is a significant correlation between the computational and experimental results for mean contact pressure and maximum contact pressure (Fig and Table). The finite element model did not characterize all the variations in the lateral force ratio accurately, partly due to the relatively small variations in this parameter over the various testing conditions.

The mean contact pressure and maximum contact pressure averaged over three flexion angles was highest for knee 1 and lowest for knee 3 for both experimental and finite element study (Table 4.2 and Fig 4.1, 4.2). The table 4.2 and figure 4.3 shows lateral force ratio for finite element model and experimental study with absolute values in the same range but less variation between testing conditions. The maximum contact pressure and mean contact pressure averaged over the
three flexion angles with hamstrings loaded showed similar variations between the computational and experimental methods. It indicates that the finite element model is taking into account the difference in anatomy between each knee.

The table 4.4 shows the p-values for variation due to hamstrings loading for both finite element and experimental method. The variations were significant with hamstrings loading for finite element technique for maximum contact pressure at 40 degrees of flexion. In the finite element model, an increase in maximum contact pressures and mean contact pressures was observed with hamstrings loaded and these variations were similar to the in-vitro study. Lateral force ratio increased with hamstrings loaded at 60 and 80 degrees of flexion in the finite element model similar to the experimental results. This indicates, that the contact forces were shifted on the lateral facet on hamstring loading. This shift of pressure on the lateral facet of patella indicates lateral orientation of the patellar tendon caused due to lateral shift or external rotation of the tibia causing patella to shift laterally. Thus, the finite element model predicts similar variation in pressure parameters as the in-vitro study with hamstrings loaded.

5.2 Kinematic Results

There was a positive correlation for patellar flexion and external rotation. The regression was significant for patellar flexion. Lateral shift and lateral rotation were not predicted as accurately as the other kinematic parameters, in part due to the error associated with reproducing the coordinate systems digitized on the knees experimentally on the finite element models.
The patellar flexion averaged over three flexion angles was highest for knee 2 for both experimental and finite element study (Table 4.5 and Fig 4). The table 4.5 and figure 4. shows external rotation for finite element model and experimental study with knee3 minimum for both methods. The patellar flexion and patellar tilt averaged for each flexion angle with hamstrings loaded followed the same trends for both computational and experimental methods. It indicates that the finite element model is taking into account the difference in anatomy between each knee.

Variation with hamstrings loading was significant for patellar flexion with the FE method at 40 and 60 degrees of flexion. Variation for lateral tilt was significant at 60 degrees of flexion. The variations in patellar flexion predicted by FE model were similar to the in-vitro study. Thus, indicating posterior shift of tibia on hamstring loading which consequently increases the flexion of patella. Valgus rotation or lateral rotation increased with hamstrings loaded in the finite element model, similar to the experimental results. Lateral tilt had absolute values similar to the experimental results but there was no major variation observed with hamstrings loaded. A consistent increase in patellar lateral shift was observed on hamstrings loading for all flexion angles for both methods. The finite element model predicted similar variations in kinematic parameters with hamstrings loading as the experimental study.
5.3 Additional Parameters from Validated Model.

The stress state at a particular location in the cartilage can be described using parameters called stress invariants which are derived from the principal stresses. They are independent or invariant to the choice of coordinate system in which the stress is represented. Hydrostatic and octahedral stresses are stress invariants which represent the state of cartilage. The deformation caused by hydrostatic stresses leads to change in volume but no change in shape even if the material is compressible. The octahedral stresses causes a change in shape but no change in volume. When both octahedral stresses and hydrostatic stresses are present, there will be a local change in shape and, if the material is compressible, a change in volume. Hydrostatic pressures uniformly compress the cartilage, causing a change in volume without change in overall shape [41]. Octahedral shear stresses distort the cartilage causing a change in shape without change in volume. Cartilage damage can be related to the different mechanobiological responses of cartilage to the hydrostatic and octahedral stresses. The stress invariants are often used in the analysis of engineering materials since fatigue damage and material fracture criteria are often related to these quantities. Similarly, these stress invariants can be used to analyze the articular cartilage. The articular cartilage is viscoelastic in nature. The collagen fibers and proteoglycans form the solid phase and water primarily forms the fluid phase of the cartilage. Thus when compressive forces are applied on the cartilage, the fluid in the cartilage moves from region of high hydrostatic pressure to regions of low hydrostatic pressure. The tensile or shear stresses are carried
by the collagen fibers. More increase in shear forces compared to hydrostatic pressures acting on the cartilage due to hamstrings loading can damage the cartilage. Thus, the stress invariants obtained from the computational model helps to better understand the 3D state of stress in the cartilage.

For patellar cartilage, hydrostatic pressures increased with hamstrings loading (fig) but variations were not statistically significant (Table 4.7). Contact Shear forces increased significantly with hamstrings loading at 40 degrees of flexion (Table 4.7). Octahedral stresses did not show significant increase with hamstrings loading (table 4.7). For femur cartilage, the hydrostatic pressures increased with hamstrings loading at 40, 60 and 80 degrees of flexion (fig) but the variations were not statistically significant (Table 4.8). Variations due to hamstrings loading were not significant for either contact shear forces and octahedral stresses.

The average octahedral stresses for three knees remained the same with hamstrings loading whereas averaged hydrostatic pressures for three knees increased. This indicates that there are less chances of cartilage degeneration with hamstrings loading due to octahedral shear stresses. The increase in contact shear forces on hamstrings loading could cause wear of the cartilage over time.

5.4 Limitations

1. The articular cartilage is biphasic in nature. The fluid phase primarily consisting of water and solid phase consisting of collagen fibers and
proteoglycans. In order to accurately represent articular cartilage, it should be modeled as non-linear, inhomogeneous, anisotropic and viscoelastic. In the current model, the cartilage is modeled as single-phase, homogeneous, isotropic and linear-elastic. The cartilage properties were chosen to represent its nearly steady state response to isometric loading recorded experimentally, rather than the instantaneous response. However, there have been previous studies where the cartilage has been modeled as single-phase [42], linear elastic continuum material, this simplification is justified to represent the fundamental behavior of the cartilage [37, 38, 43].

2. Tendons, muscles and ligaments are viscoelastic in nature [44]. Error could be introduced in the finite element model since the muscles, tendons and ligaments were represented using linear tension-only springs.

3. The absolute value for pressures obtained from the finite element model are in the same range as the in-vitro study. The difference in the values can be due to errors in the experimental values as well as computational errors. Experimental errors can be related to calibration of pressure sensors, artifacts during measurement, and wrapping of the sensors. Computational sources of error include tracing error and error in kinematics reconstruction of tibial and medial patellomeniscal ligament attachment points.

4. The difference in kinematic values between the two methods can be due to small errors in the experimental values due to metal interference in the electromagnetic tracking system and less variation in the final position of patellar
tendon attachment points due to medial shift or internal translation of tibia during in-vitro testing. Computational source of error could be in reproducing the digitized coordinate axes of the knee in the experimental study on the finite element knee model.

5. The bones were modeled as rigid surface instead as a deformable surface to improve computational cost and efficiency. In the current study we were modeling the contact between the cartilages. Moreover, it has been observed in previous studies there is no significant difference between the results between bones modeled as rigid and deformable [37, 40].

6. For the current study finite element models for three knees were developed. Developing more FE models will give a better insight in the relationship between the computational and experimental technique.

5.5 Conclusion

The current study indicates that a finite element model can be developed which is validated to predict the variations in pressure and kinematic parameters due to hamstrings loading. The model was validated by testing its correlation and regression with the experimental technique while accounting for variations due to anatomy of knees, flexion angle and loading conditions. For pressure parameters, the finite element model accurately accounted for variations due to anatomy, flexion angles and loading conditions but for kinematic parameters, the finite element model accurately accounts for the variations due to hamstrings loading only and not the variations due to anatomy and flexion angle. The finite
element model had a significant correlation with the experimental method to predict maximum contact pressures, mean contact pressures and patellar flexion. Thus we failed to reject the null hypotheses that there is a significant correlation for maximum contact pressure, mean contact pressure and patellar flexion between computational and experimental method. Using the FE model, the general trend showed increase in cartilage pressures and shift of pressure on the lateral facet with hamstrings loading but the variations were not significant at all flexion angles. The hydrostatic pressures increased with hamstrings loading but the variations were not significant. The chances of cartilage damage due octahedral stresses are less since it did not vary much with hamstrings loading. The contact shear forces increased with hamstrings loading and increased significantly for one of the flexion angles. Thus, over a period of time chances of cartilage damage due to contact shear forces are higher. A validated model can help evaluate multiple physiological conditions and surgical procedures, like tibial tuberosity transfers and MPFL release for elevated medial pressures.
REFERENCES


APPENDIX A

INPUT FILE FOR ABAQUS

**=================================================================================================================
**Heading: Initial Input File before developing the model in GUI
**
**Model name:
**=================================================================================================================
*PART,NAME=Patellar_Cartilage
*INCLUDE,INPUT=trugrdo_pc_knee2
** Section:
*Solid Section, elset=PM1, material=cartilage

*END PART

**=================================================================================================================
*PART,NAME=Femur_Cartilage
*INCLUDE,INPUT=trugrdo_fc_knee2
** Section:
*Solid Section, elset=PM1, material=cartilage*END PART

**=================================================================================================================
**===========================================================
*PART,NAME=Femur
*INCLUDE,INPUT=trugrdo_femur_knee2
** Section:
*SHELL GENERAL
SECTION,MATERIAL=M1,ELSET=SS1M1,ORIENTATION=COR1
0.0
*END PART
**===========================================================

**===========================================================
*PART,NAME=Patella
*INCLUDE,INPUT=trugrdo_patella_knee2
** Section:
*SHELL GENERAL
SECTION,MATERIAL=M1,ELSET=SS1M1,ORIENTATION=COR1
0.0
*END PART
**===========================================================

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10/18/2011
Kushal Shah
University of Akron,
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Thank you

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