FACTORS AFFECTING PATELLAR MECHANICS AND BONE STRAIN IN PATIENTS WITH CROUCH GAIT

by

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The following individuals read and discussed the thesis submitted by student Erika Ramirez, and they evaluated her presentation and response to questions during the final oral examination. They found that the student passed the final oral examination.

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DEDICATION

This work is dedicated to my family for their never-ending actions and words of support throughout my life. I am truly blessed to have all of you.

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First and foremost, I thank my God for being my strength and guiding me in life. I never would have imagined reaching this far. I express my most sincere gratitude to my thesis advisor, Dr. Clare Fitzpatrick, for her constant patience, advice, and guidance. I could not have had a better advisor and mentor. I would also like to thank my committee Dr. Lujan and Dr. Uzer for your questions, insightful comments, and time. Finally, I would like to recognize all the teachers and professors that have ever taught me. I admire the work you do.

ABSTRACT

Crouch gait is a musculoskeletal impairment that results in higher than normal stresses at the patellofemoral (PF) joint that can lead to instances of anterior knee pain and loss of ambulation. The impact of commonly implemented surgical procedures to correct for crouch gait can be quantified by evaluating stresses and underlying patellar bone strain during a gait cycle. The aims of this thesis work were (1) to analyze changes in PF mechanics and patellar bone strain between pre- and postoperative conditions; (2) to quantify the variability of predicted patellar strain due to different kinematic/loading profiles or patellar material properties; and (3) to quantify the impact of varying surgical decisions on predicted patellar strain. To accomplish these aims, five finite element (FE) models of the PF joint were developed with density-mapped material properties of patellar bone and appropriate pre- and postoperative kinematic/loading conditions obtained from experimental gait laboratory data. Patients underwent surgical procedures which included patellar advancement, distal femoral extension osteotomy, or a combination of both. These procedures were virtually simulated in the FE environment to predict stresses and strain across the PF joint. It was found that preoperative patellar bone strain and cartilage stresses were consistently higher with approximately two times the variability than their postoperative counterparts. In addition, the variability associated with different kinematic/loading conditions was higher than that of different PF geometries/material properties. A parametric evaluation of surgical decisions found that patellar position and, to a lesser extent, angle of wedge resection for the distal osteotomy

influence predicted patellar strain values. Understanding baseline changes in PF loads and patellar strain due to surgical intervention may be used to inform a surgeon on treatment pathways which best alleviate strain in the patella thus reducing the risk of anterior knee pain and early onset osteoarthritis on a patient-specific basis.

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CP	Cerebral palsy
DFEO	Distal femoral extension osteotomy
EMG	Electromyography
FE	Finite element
FEM	Finite element methods
HSL	Hamstring lengthening
HST	Hamstring transfer
PA	Patellar advancement
PF	Patellofemoral
ROM	Range of motion
TF	Tibiofemoral

CHAPTER ONE: INTRODUCTION

1.1 Motivation

First described by William Little in the 1840s, cerebral palsy (CP) is a permanent, progressive developmental disability (Sankar & Mundkur, 2005) that affects approximately 17 million children worldwide. Developmental disability can range from mild to severe and can be exacerbated by accompanying impairments such as abnormal gait patterns. Among CP patients, crouch gait is the most commonly observed pathological gait pattern affecting approximately 75% of this population (Wren et al., 2005). Crouch gait is defined by an overly flexed knee during stance-phase of gait that reduces the capacity of muscles to extend the knee and hip leading to higher than normal stresses at the patellofemoral (PF) joint (Steele et al., 2012).

Depending on the severity of crouch, PF contact pressures have been found to increase from 1.0 MPa in a normal gait pattern to 3.3 MPa for severe crouch gait (Brandon et al., 2018). Excessive joint loading from either an acute impact event or repetitive motion (cumulative contact stress) have been linked to cartilage degeneration (Buckwalter et al., 2013) which is one of the primary clinical syndromes of PF osteoarthritis (Culvenor et al., 2013). Although osteoarthritis is classified as a cartilage disorder, this disease is accompanied by defined changes to the subchondral and periarticular bone. It has been hypothesized that increased subchondral bone stiffness limits a joint's ability to distribute generated strain leading to magnified peak dynamic forces in the overlying articular cartilage, which in turn, accelerate the amount of damage experienced (Hayami et al., 2004). Therefore, the functional integrity of cartilage is directly tied to the mechanical properties of the patella. For this reason, quantifying strains in the patella can be used as a direct comparison metric to assess the likelihood of developing early onset PF osteoarthritis.

As both PF osteoarthritis and CP have no existing cure, current medical intervention focuses on symptomatic relief (Powell et al., 2005) since it is known that if left untreated or if inadequately addressed, the rate of progression of crouch increases leading to joint deformity, loss of ambulation, and anterior knee pain (Rethlefsen et al., 2015). As such, within the medical community and researchers alike, high interest is placed on determining the intrinsic and extrinsic biomechanical factors that cause crouch gait. Identifying all underlying factors aids in the development of targeted, cohesive surgical treatment plans that could potentially reduce the number of surgical interventions required.

Current clinical protocol consists of a combination of medical history, physical examinations, functional assessments, imaging, and gait analysis data to prepare a treatment plan; however, physical examinations and imaging data provide limited information seeing as they are based on a static response whereas gait is dynamic (Novacheck et al.,2010). To overcome such limitations, preoperative lower extremity kinematics and dynamic electromyography (EMG) data are collected at a gait analysis laboratory to provide information on muscle activation patterns and joint kinematics (Simon, 2004). Analyzing this data provides quantitative information on the effects of multiple factors to identify gait deviations which can improve upon treatment plans formulated through consultations alone.

A systematic literature review (Galey et al., 2017) investigating gait outcomes of surgical interventions found that 27 out of 30 qualifying studies treated crouch gait with a hamstring lengthening (HSL) or a hamstring transfer (HST) procedure. However, other researchers have demonstrated that not all patients with crouch gait would benefit from HSL or HST (Arnold et al., 2006), shifting treatment focus into augmenting the power of the knee extensors. Distal femoral extension osteotomy (DFEO) and patellar advancement (PA) are two procedures that correct knee flexion deformities and produce results comparable to a hamstring lengthening or transfer (Novacheck et al., 2009). Even though surgical interventions have evolved beyond HSL/HST procedures there is little quantitative data comparing the effects of DFEO and/or PA on PF mechanics or underlying patellar bone strain.

Studies that have assessed postoperative gait outcomes have found that while most patients show positive improvements there are patients whose condition deteriorates despite undergoing invasive surgical procedures. Variability of treatment outcomes observed can be attributed to a lack of standardized protocols that address the multifactorial nature of crouch gait on a patient-specific basis. When accounting for the influence of gait analysis on surgical decision-making, prior studies have found that surgical recommendations were changed in 40-89% of all patient cases considered (DeLuca et al., 1997; Wren et al., 2011; Cook et al., 2003). These studies not only demonstrated a change in the overall treatment plan after performing gait analysis but also an overall reduction in the number of surgical procedures performed (Wren et al., 2011; Cook et al., 2003).

Reducing the number of unnecessary surgical procedures has the potential to improve the quality of life for CP patients by minimizing their overall rehabilitation period as well as minimizing associated hospitalization costs. Figure 1.1 shows the number of procedures along with the associated costs patients with CP underwent in a calendar year (Gillette Children's Specialty Healthcare, 2014). Inpatient admissions and same-day procedures averaged to 2.1 admissions per patient at a median charge of approximately $66,000 per admission. Over a lifetime, it has been estimated that long-term supportive care or services for an individual diagnosed with CP will be in the range of $1 million USD (Honeycutt et al., 2003).

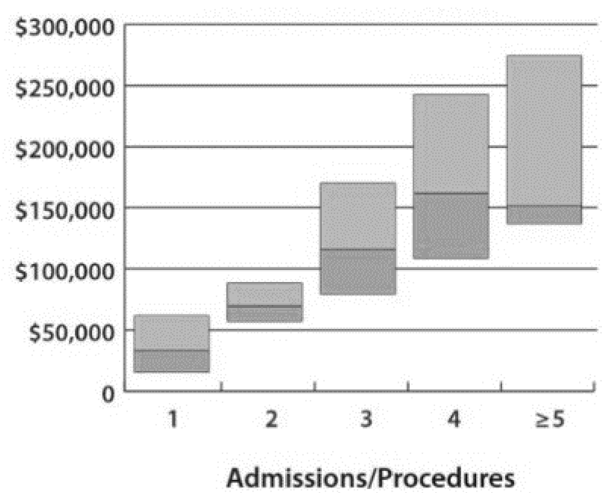

Figure 1.1 Charges per patient by number of admissions/procedures. The 25th to 75th percentile charges incurred per patient for a cohort study (n = 391) during a 39-month period in which "admissions" refer to overnight stays and "procedures" did not require a stay

1.2 Research Goals

Lack of standardized surgical guidelines and/or knowledge of underlying biomechanical factors highlights the need for prospective studies to address the question as to whether surgical decisions should be made with respect to the anatomy and musculature of the patient to improve postoperative outcomes. This thesis work seeks to provide answers to that question by (1) comparing the efficacy of two surgical procedures; (2) determining the effects of two biomechanical factors; and (3) evaluating the impact of varying surgical decisions. To accomplish this, we developed a computational workflow that incorporated anatomical and physiological data, rigid-body musculoskeletal simulations, and finite element methods (FEM). Employing computational tools helped eliminate variability and uncertainty, as consistent boundary conditions were enforced between subjects. Additionally, computational simulations allow for prediction of parameters not otherwise measurable in-vivo such as strains and stresses in soft and bony tissue.

The primary contributions of the research presented in this thesis are:

- **Development of a computational platform to determine an optimal surgical treatment plan on a patient-specific basis**

  For the scope of this work, an optimal surgical treatment was defined as the procedure(s) that restored normal quadriceps efficiency, minimized stresses in articular cartilage and strain in the underlying bone as well as maximized knee extension function. Changes in patellar mechanics and bone strain between pre- to postoperative conditions for two surgical procedures as well as the variability of predicted strain due to different kinematic/loading conditions and inherent bone properties were quantified.

- **Parametrized evaluation of the effects of varying surgical decisions**

  In our initial study, the effects of varying surgical decisions were neglected as virtual surgical procedures were performed in the same manner from patient to patient. In this secondary study, an existing FE model in conjunction with average kinematic/loading conditions was used to quantify changes in patellar mechanics and bone strain with varying levels of patellar advancement and distal osteotomies.

CHAPTER TWO: MANUSCRIPT "FACTORS AFFECTING PATELLAR BONE STRAIN IN PATIENTS WITH CROUCH GAIT"

2.1 Introduction

Cerebral palsy (CP) reportedly occurs in 3 – 4 per 1000 children in the United States making it the most common motor disability in childhood (Christensen et al., 2014). Crouch gait, one of the most predominant musculoskeletal impairments within this population, is characterized by excessive hip and knee flexion with exaggerated dorsiflexion during stance-phase of gait resulting in abnormal stresses at the patellofemoral (PF) joint (Steele et al., 2012). Factors contributing to this gait impairment can include hamstring tightness, lever-arm dysfunction, skeletal deformities, weakness, and/or impaired balance (Stout et al., 2008). The severity of the flexion contracture can worsen over time placing patients with CP at high risk of patella alta with possible patellar stress fractures and anterior knee pain with incidence rates reported at 2.7-4.5% (Mergler et al., 2009) and 39% (Jahnsen et al., 2004) respectively.

Knee pain and loss of articular cartilage are characteristic of PF osteoarthritis whose primarily cited cause is abnormal PF stress (Mills and Hunter, 2014). Higher PF stresses have been correlated to person-based risk factors (i.e. age, gender, obesity) in addition to joint-based risk factors such as increased quadriceps forces (Steele et al., 2012) and/or abnormal patella alignment (Farrokhi et al., 2011). Orthopaedic procedures can address joint-based risk factors by modifying musculoskeletal pathology through tendon lengthening, tendon transfers, or rotational osteotomies (McGinley et al., 2012).

Distal femoral extension osteotomy (DFEO) and patellar advancement (PA) are two surgical procedures aimed at restoring normal quadriceps efficiency (Novacheck et al., 2009) while minimizing stresses in the patellar and femoral trochlear cartilage and strain in the underlying bone. However, while the kinematic outcomes of surgery to correct for crouch are frequently measured and reported, the impact of surgery on stresses and strains at the patella are less well defined with little consensus on a procedure that produces favorable results among all patients.

Determining the appropriate treatment aimed at correcting crouch is difficult given that the factors contributing to hip and knee extension during normal gait are not fully understood (Arnold et al., 2005). Clinical gait analysis provides an assessment of locomotion that can aid in the determination of appropriate surgical or orthotic intervention (Davis et al., 1991). Multiple factors that contribute to a musculoskeletal impairment can be simultaneously analyzed and used to develop a comprehensive treatment plan that could potentially reduce the number of surgical interventions required. However, major limitations of gait analysis are the inability to measure forces generated by individual muscles and joint tissues or fully understand the cause-and-effect relationships in such a complex dynamic system (Delp et al., 2007). As a result, rigid-body muscle-driven dynamic simulations are commonly used to complement experimental and physical approaches by providing additional insights into muscle function not otherwise measurable in vivo (Reinbolt et al., 2011).

Rigid-body simulations can be combined with finite element (FE) modeling to develop comprehensive biomechanical models. Such models provide an effective, non-invasive method that can be used to assess and compare the efficacy of potential surgical procedures (Cohen et al., 2003) by providing a framework for the integration of anatomical and physiological data (Holzbaur et al., 2005). In the current study, clinical gait analysis data from patients with crouch gait were incorporated into a series of detailed 3D dynamic FE models of the PF joint to evaluate the comparative changes in patellar bone strain and cartilage stress between pre- to postoperative conditions as well as between two potential surgical interventions. In addition, the variability of predicted patellar strain due to different kinematic/loading conditions and inherent bone properties was also quantified. We hypothesized that (1) postoperative conditions would show significantly reduced patellar bone strain compared to the preoperative state due to a reduction in quadriceps muscle force, and (2) patellar bone strain is more sensitive to geometry/material properties than kinematic/loading conditions.

2.2 Methods

<u>Clinical Data Collection</u>

A retrospective chart review of patients treated at Children's Hospital Colorado in Aurora, Colorado was performed to select subjects for this study. Eleven patients (M=6, F=5; surgical age = 13.3 ± 1.9 years; BMI = 18.0 ± 5.0) were selected to cover a broad range of crouch severity. Selection criteria included (1) a primary diagnosis of cerebral palsy; (2) treatment with patellar tendon advancement, distal femoral extension osteotomy, or a combination of both procedures; (3) the ability to walk without an assistive device; and (4) availability of pre- and postoperative gait analysis data.

Lower extremity kinematics were collected using a lower-limb marker-based motion capture system at a rate of 120 Hz (Vicon Motion Systems). In-ground force plates were used to record ground reaction forces and moments at 1080 Hz. Random noise in data was filtered using a fourth-order zero-lag low-pass Butterworth filter with a cutoff frequency of 40 Hz (Yu et al., 1999). Retroreflective marker placement followed a protocol set published in the literature (Davis et al., 1991). Following a standing calibration, trial patients were instructed to walk at a self-selected pace. These patients subsequently underwent surgery to correct for crouch and the experimental data collection process was repeated postoperatively. Surgical recommendations were based on instrumented gait laboratory findings and systematic data interpretation as part of a typical clinical referral (Chang et al., 2006).

Musculoskeletal Modeling

As it is not possible to non-invasively measure in vivo muscle forces, muscle forces for pre- and postoperative conditions were predicted using the rigid-body musculoskeletal platform OpenSim (Delp et al., 2007). A previously developed lower limb and torso musculoskeletal model with a total of 23 degrees of freedom, 92 musculotendon units, and a modified knee joint (Navacchia et al., 2016) was used as the baseline model and scaled for each patient according to their anthropometric measurements obtained from their respective standing calibration trials. Collected kinematics and ground reaction forces were then used as inputs into OpenSim to predict muscle forces required for each subject to perform the gait cycle activity (Figure 2.1). To account for the effects of different walking speeds kinematic and force data was time-normalized to a single gait cycle (i.e. 0 – 100%, where 0% is heel strike).

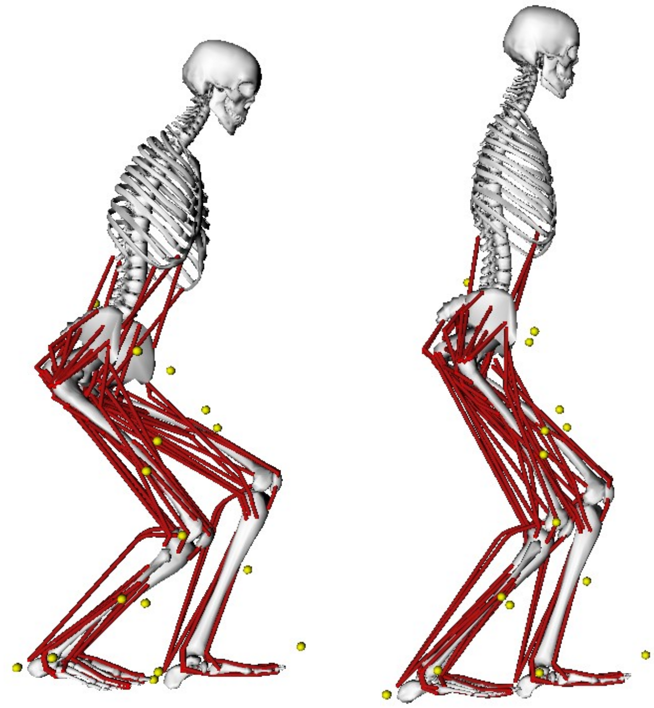

Figure 2.1 Pre- (left) and post- (right) operative rigid body musculoskeletal simulations performed in OpenSim for a patient during a gait cycle

<u>Finite Element Model</u>

Patient-specific 3D knee imaging was not available for the 11 patients, however, a series of five detailed, deformable representations of the PF joint was constructed from a publically available repository (Harris et al., 2016). FE models for healthy male subjects (age: 57.6 ± 9.7 years) were constructed from magnetic resonance (MR) and computed tomography (CT) scans. Femur, tibia, and patella bone geometries were segmented from CT scans and then aligned to the patellar, femoral, and tibial cartilage segmented from MR images using the commercial software Amira (ZIB, Berlin, Germany).

For computational efficiency, femoral and tibial bony geometries were represented using rigid triangular shell elements due to their much greater stiffness when compared to joint soft tissues (Mesfar and Shirazi-Adl, 2005). Bones were meshed with an average element edge length of 1 mm (Hypermesh, Altair Hyperworks, Troy, MI).

Patellar bone was represented in greater detail with an average element set of 30,800 tetrahedral elements to capture subject-specific inhomogeneous bone material properties. Bone properties were extracted from CT data using BoneMat (Istituto Ortopedico Rizzoli, Bologna, Italy) (Figure 2.2). Hounsfield units were correlated to be linearly related to apparent density (Peng et al., 2006) and elastic moduli were then calculated by applying an experimentally derived non-linear density-elasticity relationship taken from the literature (Keller, 1994). A semi-automated mesh-morphing technique was subsequently applied via Matlab (MathWorks Inc., Natick, MA) to represent articular cartilage as hexahedral continuum elements with linear elastic material properties (E=12 MPa, υ=0.45) (Baldwin et al., 2010).

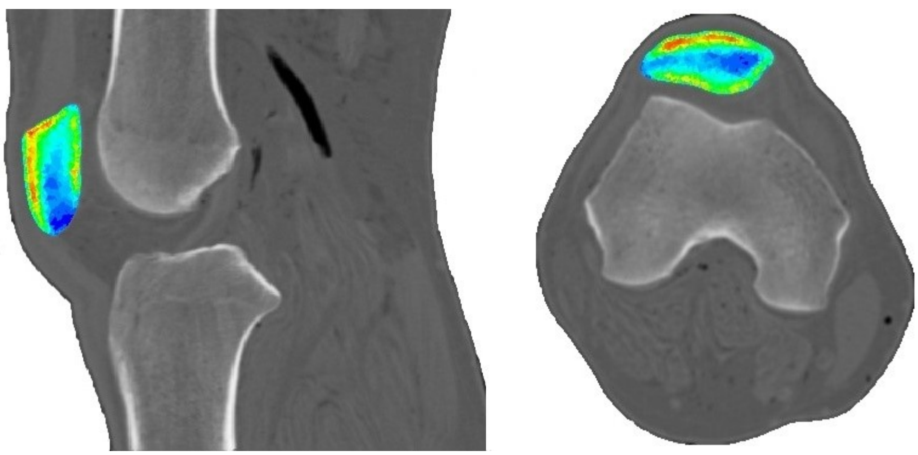

Figure 2.2 CT scan rendering in BoneMat displaying heterogeneous material property distribution within the patella.

Extensor mechanism muscles and tendons (rectus femoris, vasti, patellar ligament) and medial and lateral PF ligaments were represented using 2D fiber-reinforced membranes (Fitzpatrick et al., 2011) with fibers represented as tension-only, non-linear springs consistent with other models in the literature (Holzbaur et al., 2005). Material properties for the composite structure were assigned to match published experimental uniaxial force/displacement data (Atkinson et al., 2000). All ligaments were attached to the bone over physiological insertion/origin sites to provide appropriate transmission of quadriceps load and ligament tension to the patellar bone. Patellar cartilage was attached to the underlying bone through a shared set of equivalenced nodes between structures.

Each detailed FE model was modified to represent the preoperative anatomy of the clinical patients by matching patellar alta-baja and tibial tubercle position to subject-specific lateral radiographs acquired before surgery. Tibiofemoral (TF) joint kinematics and quadriceps forces from the musculoskeletal simulations were applied to the FE models to simulate a gait cycle in the FE framework. Of the 11 clinical patients, ten underwent bilateral procedures while one underwent unilateral surgical intervention. Hence, preoperative simulations included 21 sets of kinematic/loading conditions applied to each of the five different FE models resulting in 105 simulations.

Each FE model was subsequently modified to represent the postoperative anatomy of the clinical patients by virtually simulating the surgical procedures they underwent (Figure 2.3). The PA procedure was simulated by inferiorly advancing the insertion of the patellar tendon by 2 cm thereby correcting patella alta (Arnold et al., 2005). For DFEO, the angle of the wedge resection is equal to the degree of knee contracture and the extension that is desired which varies from patient to patient and is done with a large enough wedge to get approximately 5° of hyperextension which is determined intra-operatively. To normalize the effects of different wedge resections a 30º anterior wedge osteotomy was implemented across all DFEO surgeries (Lenhart et al., 2017). Gait cycle simulations were repeated using modified FE models and kinematic/muscle forces from the postoperative conditions. For each simulation, maximum and minimum principal patellar bone strains were predicted. Because peak strain values can be highly susceptible to localized mesh quality/element skewness effects, an alternative highly strained bone volume threshold was implemented instead. Strains were compared using a threshold level of 5% (i.e. 5% of the patella bone, by volume, was above this strain value) (Fitzpatrick et al., 2011).

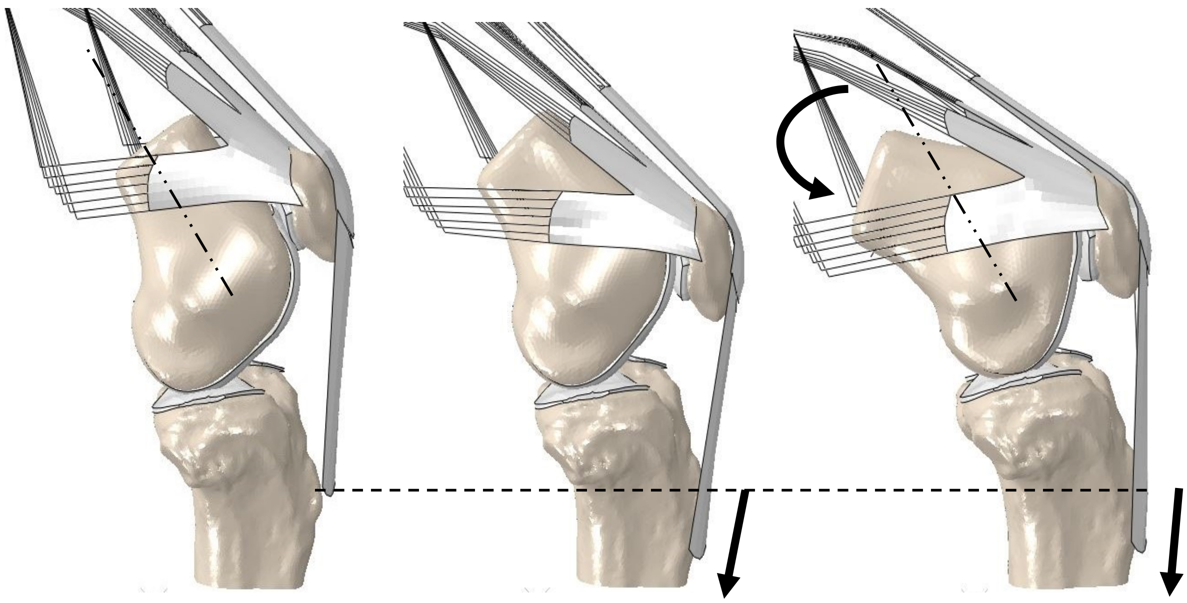

Figure 2.3 Lateral view of a customized finite element model setup for preoperative condition (left), after PA (center), and PA + DFEO (right)

<u>Patient-specific variables affecting patellar bone strain</u>

While kinematics and muscle efficiency may be affected by surgical decisions, other inherent patient-specific factors such as bone quality will also contribute to strains generated in the patellar bone. The inclusion of five FE models with unique geometries and mapped material properties allowed us to compare the contribution of patellar bone anatomy on patellar bone strain to the contributions of different kinematics and quadriceps forces. To quantify the effect of kinematic/force variability (21 sets pre- and postoperatively) on patellar bone strain the 210 simulations were further subdivided to analyze all kinematic/loading conditions within a specific FE model (5 groups x 42 simulations). Similarly, to quantify the effect of geometry/material variability, the variation in bone strain was calculated when a specific kinematic/force condition was implemented across all FE models (21 groups x 10 simulations). For all groups analyzed an average pre- and postoperative standard deviation was calculated.

Statistical Analysis

Descriptive statistics (mean and standard deviation) were performed for maximum/minimum principal strain, patellar cartilage stress, and quadriceps forces. All data were tested for departures from normality through the usage of the Shapiro-Wilk test. Comparisons between pre- and postoperative results, as well as the differences between knees that had undergone PA+DFEO ($n = 13$) or only PA ($n = 8$), were assessed at 10% gait cycle intervals using the Wilcoxon rank sum test. For both tests, the level of significance was set at $p \leq 0.05$.

2.3 Results

Effect of surgery on PF mechanics

Throughout the gait cycle, preoperative patellar bone strain and cartilage stresses were consistently higher than their postoperative counterparts; statistically significant differences ($p < 0.01$) were observed across all intervals in the gait cycle. At peak muscle loads (~55% of gait cycle) a reduction of 25% in mean quadriceps force led to a reduction of 45% in highly strained bone volume between mean pre- and postoperative conditions (Fig. 4). Postoperative maximum patellar bone strain was reasonably consistent across subjects (1 standard deviation (SD) = 9.5 $\mu\varepsilon$), while preoperatively, patellar strain variability was in the order of two times larger (1 SD = 19.3 $\mu\varepsilon$) (Figure

2.4). Average minimum knee flexion angle was 24.6° (ROM: 29.2°) preoperatively, compared to 21.6° (ROM: 38.8°) postoperatively.

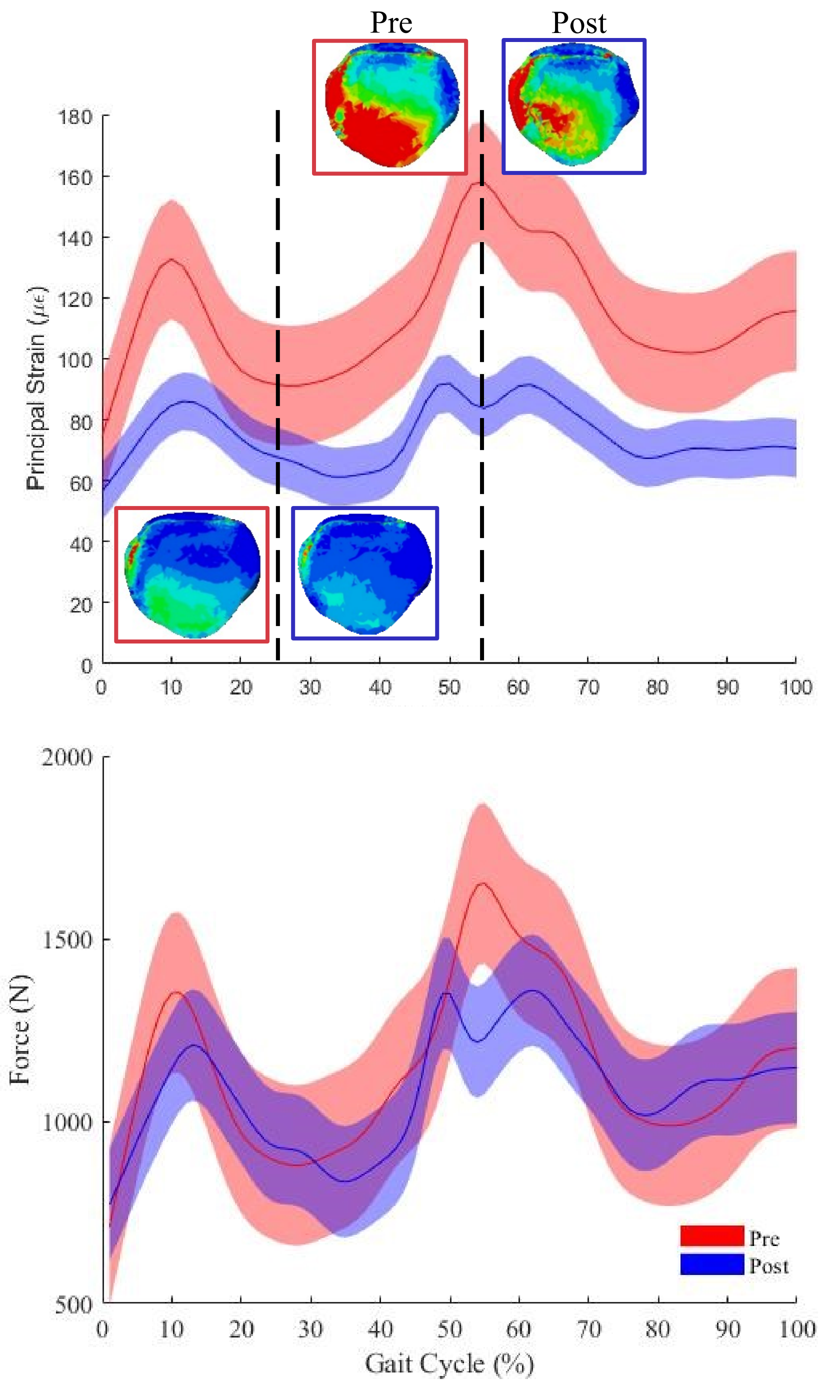

Figure 2.4 Mean (± 1 standard deviation) maximum principal strain (top) and mean muscle forces (bottom) for preoperative and postoperative simulations. Insert: Maximum principal strain distribution for a pre- and postoperative simulation at 25% and 55% of the gait cycle.

When different surgical procedures were compared, statistically significant reductions in bone strain from pre- to postoperative conditions were observed across 90% and 100% of the gait cycle for PA and PA+DFEO respectively (Figure 2.5). Preoperative average minimum knee angle was 7.8° (ROM: 37.1°) for PA patients compared to 34.9° (ROM: 24.4°) for PA+DFEO patients. Despite preoperative differences, both surgeries showed similar improvement in knee flexion – minimum knee angle and ROM improved by an average of 2.5° and 9.4°, respectively, after PA surgery, and 3.3° and 9.7°, respectively, after PA+DFEO surgery.

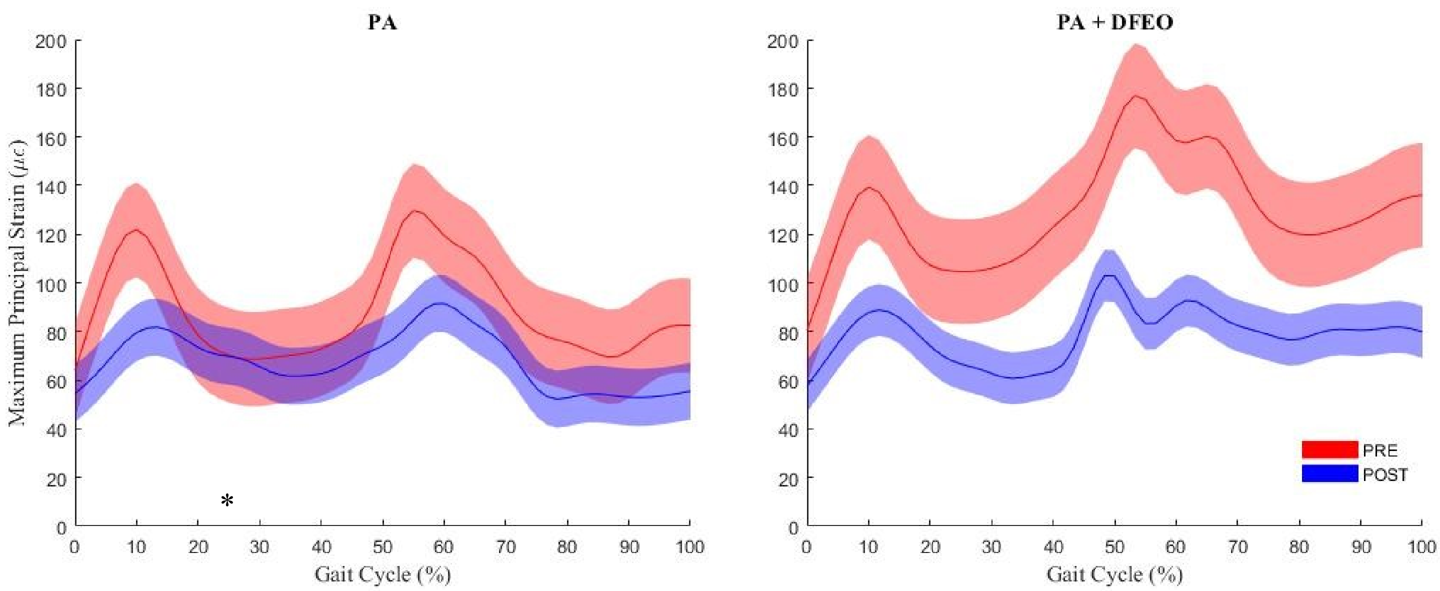

Figure 2.5 Mean (± 1 standard deviation) maximum principal patellar strain for PA (left) and PA + DFEO (right). No statistically significant differences (*p* < 0.05) between pre- and postoperative conditions at each 10% gait cycle intervals is denoted by *

<u>Patient-specific variables affecting patellar bone strain</u>

When all preoperative kinematic/loading conditions were implemented in each unique geometry/material model a linear relationship between quadriceps forces and maximum principal strain with high coefficients of determination ($R^2 \geq 0.96$) were found for all five models (Figure 2.6). Standard deviations of maximum principal strain for all kinematic/loading conditions applied to each geometry/material model ranged from 9.7 – 46.1 με (mean = 19.4 με) preoperatively to 4.8 – 24.8 με (mean = 9.6 με) postoperatively. Conversely, when each specific kinematic/loading condition was implemented across all geometry/material models, the SD of maximum principal strain across models ranged from 4.7 – 174.9 με (mean = 57.1 με) preoperatively to 2.3 – 88.7 με (mean = 33.2 με) postoperatively. Hence, the average variation associated with different material/geometries was approximately 3 times greater preoperatively, and 3.5 times greater postoperatively, than the variation associated with kinematic/loading conditions. Similar trends were observed when minimum principal strain values were compared.

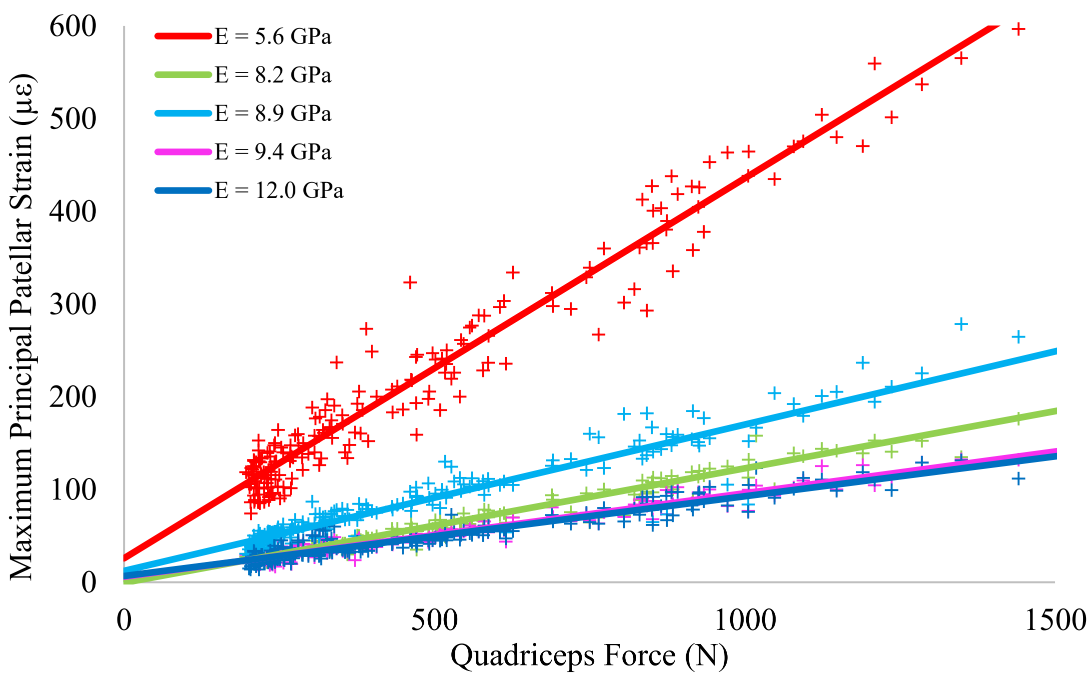

Figure 2.6 Relationship between total quadriceps force and corresponding maximum principal patellar strain at 10% gait cycle intervals for each of the 21 different preoperative kinematic/force loading conditions. Effects of different geometry/material properties are shown by the varying steepness of the linear regression lines. Reported elastic moduli (E) values represent the average modulus for each of the five FE models.

2.4 Discussion

The objectives of this study were to compare patellar mechanics and bone strains between pre- and postoperative conditions in a sample population with crouch gait, compare patellar strain outcomes for two surgical interventions, and quantify the relative impact of patellar geometry/material properties and kinematics/forces on patellar mechanics. Comparing pre- and postoperative strains and stresses, a significant decrease in patellar strain and cartilage stress was observed after surgical intervention. That difference is attributed to a decrease in quadriceps force requirements throughout the gait cycle in the postoperative simulations. In addition, there was also greater consistency in results postoperatively with half the variability compared to preoperative conditions.

Both PA and PA+DFEO surgeries showed similar changes in postoperative minimum knee angle and ROM as well as reductions in bone strain postoperatively; however, reductions between pre- and postoperative patellar strain were of a greater magnitude for PA+DFEO. However, clinical patients selected for PA+DFEO surgery had an average preoperative minimum knee angle of 39.6° (ROM: 26.0°) compared to the PA patients with a minimum knee angle of 20.0° (ROM: 37.6°). Hence, from this retrospective dataset, it is difficult to attribute changes in patellar strain to the DFEO procedure itself or differences in crouch severity classified by minimum knee angle (Steele et al., 2012). Further computational analyses may be applied to quantitatively address this question.

We investigated if the potential impact of surgical changes affecting kinematics and muscle forces would contribute to a substantial change in bone strain from pre- to postoperative conditions, or if bone strain was dominated by other inherent factors (i.e., bone material properties) that are not altered during surgery. Preliminary results indicate that both kinematics/loading conditions and bone geometry/material play a role in determining the change in bone strain from pre- to postoperative conditions; however, geometry/material properties have a greater influence (~3 times more) on bone strain variation. Contrary to expectations, neither average nor maximum patellar elastic modulus values were a clear indicator of expected strain behavior. However, differences in generated strain may be explained by the high degree of anisotropy found in the patellar bone as well as other subject-specific factors including limb alignment, weight, and PF contact area (Sharkey et al., 1997). A larger sample-based study is required to determine if these results are consistent across the population considering that kinematic/loading conditions (from clinical data) and geometry/material (from CT/MR) datasets came from small samples.

A limitation of the dataset used in this study was that CT and/or MR data were not collected for the clinical patients. Ideally, CT and MR scans from the clinical patients would have been used to develop completely subject-specific FE models since this prevented an absolute subject-specific prediction of bone strain in pre- and postoperative conditions. However, the primary focus of this work was the comparative change between these conditions and the comparative evaluation of different surgical procedures. Hence, we evaluated the effect of pre- to postoperative kinematic and muscle force changes on an arbitrary set of FE models customized to represent pre- and postoperative anatomical changes. Additionally, this allowed us to separate the effect of patellar bone material property variability in our simulations.

The same properties were used for articular cartilage and ligament geometries across all models. In actuality, subject-specific variations in soft tissue properties can introduce substantial differences in contact stresses (Li et al., 2001). Decreasing Poisson's ratio or increasing the elastic modulus may lead to increased shear and volumetric deformations within the cartilage causing changes in reported von Mises stress. While these changes would alter the absolute magnitude of cartilage stress, the comparative trends between pre- and postoperative conditions would be minimally affected. Even if such changes were made, validating strain predictions from FE models with experimental internal in vivo strain data is not feasible. However, this model is based on a previously published PF model that includes validation of PF contact mechanics with experimental pressure-sensitive film data and kinematic agreement with published literature (Fitzpatrick et al., 2011). While this is not ideal to validate the magnitude of predicted bone strains, it does support a comparative analysis between different loading and anatomic conditions.

2.5 Conclusion

Determining the optimal surgical technique for an individual is dependent on the ability to predict the loads within the various structures of the joint. Patellar fractures and anterior knee pain are of particular concern for patients suffering from crouch for which reason, patellar mechanics and strain were the focus of the current study. While a greater reduction between pre- and postoperative maximum patellar strain was found with PA+DFEO, when comparing postoperative strain, no statistically significant differences were observed between PA and PA+DFEO until the mid-swing to heel strike portions of the gait cycle (70-100%). This suggests that the inclusion of patellar advancement is necessary to reduce patellar strains and maximize ROM; this finding is consistent with those reported by a clinical study (Stout et al., 2008). Furthermore, results demonstrate that within a specific geometry/material model, patellar strain behavior can be accurately predicted regardless of the kinematic/force loading applied. Work is ongoing to expand this analysis to develop a computational testbed that includes additional CT/MR scans to further quantify the effects of different geometry/materials and to continue the assessment of prospective surgical interventions. Findings from the proposed study may help inform a surgeon on treatment pathways which best alleviate strain in the patella and hence reduce the risk of anterior knee pain and early onset osteoarthritis on a patient-specific basis.

CHAPTER THREE: PARAMETRIC ASSESSMENT OF SURGICAL DECISIONS AND THEIR RELATIVE IMPACT ON PREDICTED PATELLAR MECHANICS AND BONE STRAIN

3.1 Introduction

Patella alta (i.e. a superiorly displaced patella with respect to the femur) is a clinical condition often associated with anterior knee pain, instability, and osteoarthritis (Munch et al., 2016). Patellar position (i.e., alta-baja) can cause patellar maltracking leading to a malalignment of the quadriceps mechanism predisposing an individual to patellar instability (Koeter et al., 2007). Furthermore, incidence rates of patella alta are reported to be as high as 72% (Lotman, 1976) in crouch gait, a pathological gait pattern in which an overly flexed knee during stance phase of gait results in a generation of higher than normal muscle forces (Galey et al., 2017). Higher muscle forces, in turn, lead to an increased energy cost of gait as well as increased magnitudes of stresses and strains across the PF joint (O'Sullivan et al., 2018). For these reasons, addressing patella alta and improving knee extension should be the primary aims of any surgical treatment plan designed to treat crouch gait.

In literature, a hamstring lengthening or transfer is the most commonly implemented procedure to correct for crouch; however, a recent study has demonstrated that not all patients suffer from tight hamstrings (Arnold et al., 2005) shifting treatment focus intro augmenting the power of the knee extensors. Distal femoral extension osteotomy (DFEO) and patellar advancement (PA) are two surgical procedures that correct knee flexion deformities and produce results comparable to a hamstring lengthening/transfer (Novacheck et al., 2010). Briefly, a distal osteotomy consists of resecting a wedge of femoral bone and realigning the knee while a patellar advancement inferiorly relocates the insertion point of the patellar tendon. Furthermore, how the PA procedure is normally performed tends to place the patella in a baja position (Novacheck et al., 2009).

A prior study investigated the effects of posture and patellar position on tibiofemoral and patellofemoral contact pressures and found that crouch severity greatly affected contact pressures at both joints while patellar position affected the location of PF contact pressure (Brandon et al., 2018). However, this type of analysis could be taken a step further by not only quantifying location and magnitude of contact pressure but also evaluating underlying patellar bone strain. To our knowledge, no study has addressed the question as to how different surgical procedures influence predicted patellar bone strain. Hence, the objective of this study was a parametric evaluation of the effects of varying surgical parameters for PA and DFEO (i.e. patellar position or degree of rotation) on predicted patellar strain and PF contact pressures. To accomplish this, retrospective gait analysis data were used to create average kinematic and quadriceps force profiles to integrate into a previously developed 3D dynamic finite element (FE) model of the PF joint.

3.2 Methods

Retrospective experimental gait analysis data were obtained from a database at Children's Hospital Colorado for eleven children with a primary diagnosis of crouch gait (mean ± SD: surgical age = 13.3 ± 1.9 years; BMI = 18.0 ± 5.0). Lower extremity kinematics were collected through a motion capture system (Vicon, Denver, CO) with in-ground reaction force plates to record marker trajectories and ground reaction forces at 120 Hz and 1080Hz, respectively. All patients subsequently underwent surgery to correct for quadriceps inefficiency through a patellar advancement, distal osteotomy, or a combination of both procedures and the experimental data collection process was repeated postoperatively. All participants provided informed consent and data were collected under Institutional Review Board approval.

A previously developed lower-limb musculoskeletal model (Navacchia et al., 2016) was used to predict in vivo muscle forces for pre- and postoperative conditions in OpenSim (Delp et al., 2007). Baseline joint kinematics and inverse dynamics for each patient were obtained from collected marker trajectories and filtered ground reaction force data. Joint moments were further resolved into individual muscle forces through a static optimization technique that solves the equations of motion by minimizing the sum of squared muscle activations at each time step (Anderson and Pandy, 2001).

Kinematic and force data were time normalized to a single gait cycle (i.e. 0 – 100%, where 0% is heel strike) to account for the effects of different walking speeds. Intrasubject variability due to a lower repeatability of gait analysis data in patients with crouch (Steinwender et al., 2000) was accounted by calculating average kinematic and quadriceps-loading profiles. Furthermore, potential influences of side-to-side differences (i.e. muscle volume, loading history, etc.) (Farrokhi et al., 2011) were addressed by comparing the same striking foot regardless of side.

Average postoperative tibiofemoral (TF) joint kinematics and quadriceps forces from the musculoskeletal simulations were applied to a previously described detailed 3D dynamic FE model of the PF joint (Ramirez et al., in review). Briefly, PF joint geometries (femoral, tibial, and patellar bony geometries with corresponding articular cartilage) for a healthy male subject were constructed from medical imaging data and the patella was modeled with heterogeneous elastic moduli properties extracted from CT scans. All extensor mechanism and tendons were modeled as composite structures with material properties derived from published experimental force/displacement data.

The healthy FE model was subsequently modified to represent the expected postoperative anatomy by virtually simulating the two surgical procedures. The PA procedure was simulated by inferiorly advancing the insertion of the patellar tendon by 1, 2, and 3 cm with respect to the tuberosity of the tibia, the original insertion point of the patellar tendon (Figure 3.1). For the osteotomy, the angle of the wedge resection can be virtually simulated by rotating the femur (Lenhart et al., 2017) by an amount equal to the degree of knee contracture, which varies from patient to patient. To account for the varying crouch severities captured in the gait analysis data wedge resections of 15, 30, and 45º were modeled (Figure 3.1). Additionally, combined procedures of the patellar advancement and distal osteotomy for all possible combinations were also simulated. For each simulation (N = 14) maximum and minimum principal patellar bone strains, contact area, and peak pressure were calculated.

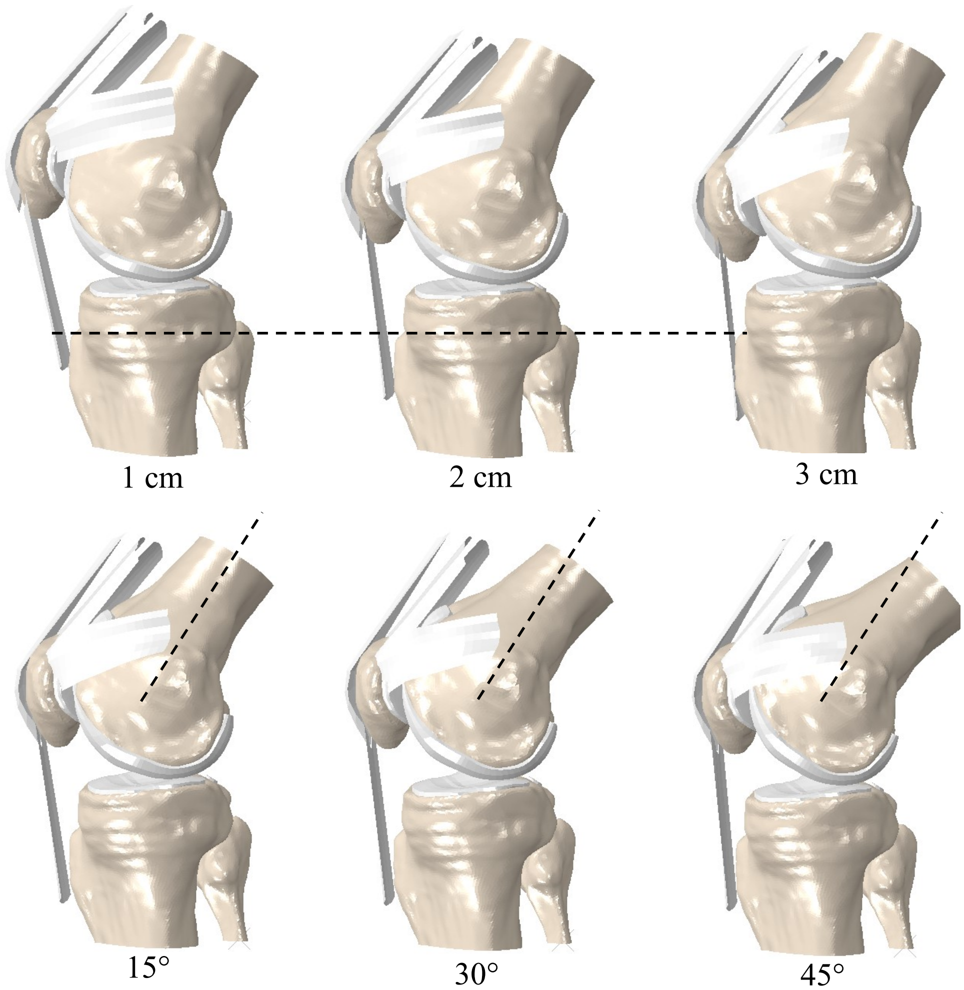

Figure 3.1 Customized finite element model setup for varying levels of patellar advancement (top) and distal osteotomies (bottom)

Statistical Analysis

For all metrics of interest, descriptive statistics (mean and standard deviation) were calculated. The Shapiro-Wilk test was used to test data for departures from normality and the statistical significance between all three levels of advancements and osteotomies were assessed using the Kruskal-Wallis test. Significance for both tests was set at $p < 0.05$. Additionally, changes between paired advancement or osteotomy levels were compared using a Mann-Whitney U test with a Bonferroni adjusted p-value.

3.3 Results

Evaluation of Patellar Advancement Levels

Reductions in maximum/minimum principal strain, peak pressure, and contact area throughout the gait cycle were observed as the patella was inferiorly advanced. The 1 cm advancement had consistently worse results causing statistically significant differences to be observed between all three advancement levels (Table 1). However, when comparing an advancement of 2 cm to 3 cm no statistically significant differences were observed for contact area, peak pressure and maximum principal strain. At peak muscle loads (~ 50% of the gait cycle), a reduction of 46% and 58% in minimum principal strain was seen going from a 1 cm to 2 cm and a 1 cm to 3 cm advancement, respectively (Figure 3.2).

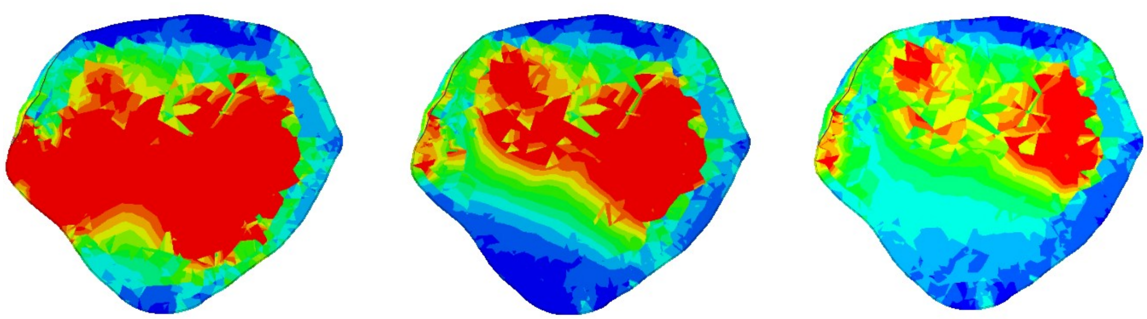

Figure 3.2 Coronal view of minimum principal patellar strain distribution at peak muscle loads for a 1 cm (left), 2 cm (middle) and 3 cm advancement (right)

Evaluation of Varying Osteotomy Angles

Throughout the gait cycle, predicted maximum principal strain values were lower at smaller wedge resections but had higher minimum principal strain values; however, no statistically significant differences were observed. Furthermore, the range of minimum principal bone strain values for all osteotomy angles were comparable to a 2 or 3 cm patellar advancement ($p = 0.46$). As the angle of the wedge resection increased, the effective contact area between the patellar and femoral cartilage decreased leading to higher peak pressures (Table 3.1).

Table 3.1 Summary of descriptive statistics for different surgical interventions

	Amount of Advancement			p-value		
	1 cm	2 cm	3 cm	(1,2)	(2,3)	(1,2,3)
Contact Area (mm^2)	283.9 ± 39.2	257.8 ± 38.6	231.6 ± 55.2	***	ns	***
Peak Pressure (MPa)	5.1 ± 1.6	4.1 ± 0.9	4.3 ± 1.0	***	ns	**
Max Principal Strain (με)	50.6 ± 22.7	41.9 ± 18.6	42.7 ± 15.5	*	ns	*
Min Principal Strain (με)	-4337 ± 2111	-2518 ± 1019	-1945 ± 750	***	***	***

	Angle of Wedge Resection			p-value		
	15°	30°	45°	(15, 30)	(30, 45)	(15,30,45)
Contact Area (mm^2)	240.6 ± 35.6	217.6 ± 26.0	187.3 ± 25.0	**	***	***
Peak Pressure (MPa)	3.9 ± 0.8	4.2 ± 0.8	4.8 ± 1.0	ns	***	**
Max Principal Strain (με)	41.0 ± 17.4	43.0 ± 18.0	46.0 ± 17.9	ns	ns	ns
Min Principal Strain (με)	-2235 ± 864.2	-2203± 802.4	-2263 ± 696.6	ns	ns	ns

ns: $p > 0.05$; *: $p \leq 0.05$; **: $p \leq 0.01$; ***: $p \leq 0.001$

Evaluation of Combined Procedures

Contact area was dependent on both the degree of rotation and patellar position with the latter being a more significant factor. As the patella was inferiorly advanced from 1 cm to 3 cm a reduction of 25% and 48% in contact area at peak quadriceps forces were observed with a 15° and 45° osteotomy, respectively. Conversely, comparing between a 15° to 45° osteotomy a reduction of 11% and 33% in contact area were seen at a 1 cm and 3 cm advancement, respectively. When comparing peak pressure across the gait cycle no significant differences were found; however, if the gait cycle was broken down into its respective stance and swing phase, significant differences ($p < 0.001$) were observed during swing phase (60 – 100 % of gait cycle). Patellar position nor the degree of rotation had any significant effect ($p = 0.407$) on predicted maximum principal bone strain; however, minimum principal bone strain was highly impacted by such factors (Figure 3.3). Regardless of the angle of the osteotomy, a 3 cm advancement had an average two-fold reduction in minimum principal bone strain magnitudes throughout the gait cycle when compared to a 1 cm advancement.

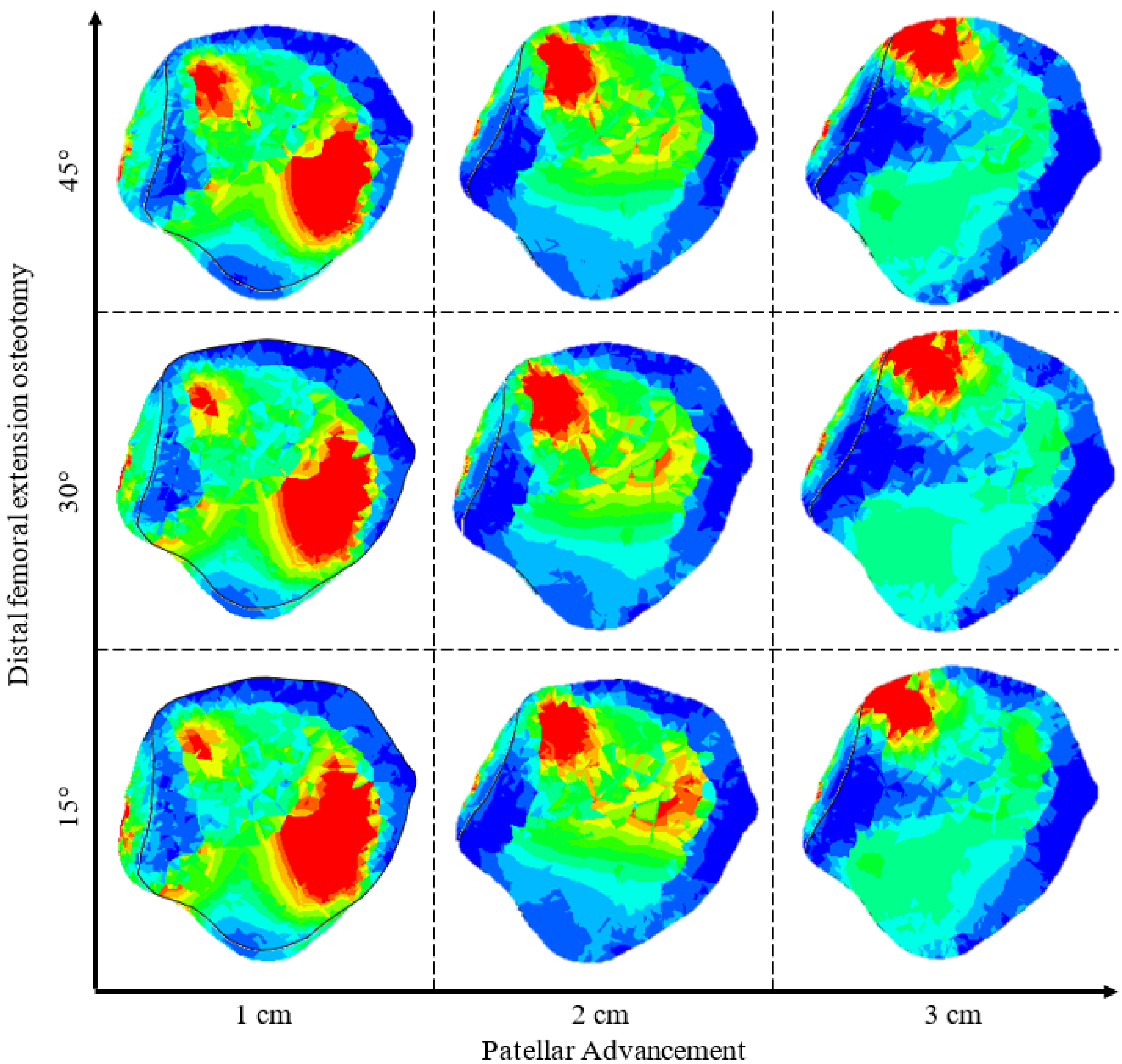

Figure 3.3 Minimum principal strain for all of the combined procedures

3.4 Discussion

The aims of this study were to quantify how varying surgical parameters for PA and DFEO, two surgical procedures used to correct crouch gait, affected predicted patellar strain and PF contact pressures. For PA, the 1 cm advancement procedure had the highest magnitudes across all metrics analyzed. This is consistent with clinical practice, where a typical PA procedure consists of advancing the patellar tendon by a distance equal to the width of the tendon, which ranges from 2 – 2.5 cm (Novacheck et al., 2009). Hence, for an advancement level of 1 cm, an incomplete correction may have taken place leaving the patella in a position that may still be considered to fall under the definition of patella alta. This conclusion is further supported by considering how a comparison between the 2 and 3 cm advancement levels yields no statistically significant differences in contact area, peak pressure, and maximum principal strain.

For all simulated osteotomies at higher flexion angles (i.e., swing phase) contact area was reduced to an average value of 188.3 mm^2 compared to 232.6 mm^2 during stance phase. A prior study reported the same trend; however, they also reported that peak pressure substantially increased during deep flexion to the point of nearly reaching the damage limits of cartilage (Thambya et al., 2005). In our simulations, peak pressure was not maximized during intervals of the gait cycle with deep flexion due to low quadriceps forces during swing phase. Contact area and thus peak pressure are dependent on the degree of rotation from DFEO and the patellar position from PA. For the single PA procedure at peak muscle loads, the location of maximum PF contact pressure on the patellar cartilage occurred laterally and changed from an inferior to a superior position as the patella shifted from alta to baja. In contrast, location of peak pressure for all osteotomies occurred on the superior-medial quadrant. For the combined procedures the same trends were observed however, the advancement level determined the location of maximum contact pressure (Figure 3.4).

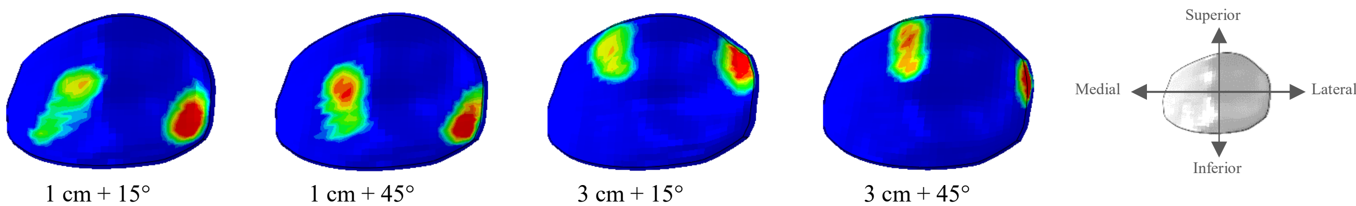

Figure 3.4 Location of peak contact pressure for combined procedures at peak quadriceps forces

This work has several limitations which should be acknowledged. First, the number of simulations for this study was low (N = 10). Hence, magnitudes of metrics analyzed can be highly susceptible to localized mesh quality or to the effects of skewed elements. For this reason, an alternative highly strained bone volume with a threshold of 5% (i.e. 5% of the patella bone, by volume, was above this strain value) (Fitzpatrick et al., 2011) was used to compare maximum and minimum principal bone strains.

Second, patellar cartilage was modeled with a constant thickness when in actuality cartilage is thicker in the middle and thinner along the patellar cartilage perimeter (Cohen et al., 1999). This leads to an over- and underestimation of the contact pressure distribution and overall magnitudes. A prior study demonstrated that variations in cartilage thickness may result in differences of approximately 10% in peak contact stress (Li et al., 2001). However, modeling cartilage with uniform thickness facilitated the creation of a higher quality FE mesh thus minimizing the likelihood of stress singularities. Furthermore, the aims of this study were not to quantify stress magnitudes but rather evaluate the comparative changes occurring between different variations of PA and DFEO. Lastly, although gait cycles were time normalized, kinetic patterns should have also been synchronized to reduce the inter-subject variability that results when calculating average gait profiles (Winter, 1984). However, for this study, average TF quadriceps and kinematics were only used to prescribe motion to the FE model and the patella was free to move in response to the loads applied.

When treating patella alta, one of the primary surgical treatment goals is to increase PF contact area and improve PF articular congruence (Biedert and Tscholl, 2017). This study has shown that the degree of rotation and amount of advancement affects contact area and patellar strain. To minimize patellar strains and peak pressures, patella alta needs to be fully corrected and the location of peak contact pressure should ideally be at the center of the cartilage where it is much thicker. If the goal were to minimize patellar strain and peak pressures while maximizing contact area the inclusion of the distal osteotomy procedure would not be recommended. However, this conclusion fails to account for other underlying factors that may be corrected by performing a distal osteotomy. For this reason, when creating a surgical treatment plan an orthopeadic surgeon should take into consideration whether a distal osteotomy is required on a patient-specific basis.

CHAPTER FOUR: CONCLUSIONS

4.1 Summary

There is a lack of consensus in the literature regarding which surgical procedure(s) should be performed to address the underlying factors that cause crouch gait. To address this gap in knowledge, the research aims of this thesis work were to evaluate the efficacy of two surgical procedures along with potential variations in surgical technique as well as the impact of joint-based risk factors on patellar mechanics and bone strain. Key findings from the work done include:

- Patients that underwent surgery to correct for knee contractures and quadriceps inefficiency consistently reduced patellar bone strain
- Patellar strain variability across patients was halved postoperatively
- Two different surgeries showed similar improvements in knee flexion and range of motion
- Kinematic/force conditions and bone material properties impacted predicted bone strain
- Varying the amount of patellar advancement or the angle of the wedge resection for the osteotomies affected predicted patellar strain
- Patellar position significantly influenced the magnitude of compressive patellar strains

4.2 Limitations

The accuracy of predicted quadriceps and kinematic profiles used to drive the motion of the FE models were dependent on OpenSim as well as on the experimental data collection process. Briefly, a scaled generic musculoskeletal model was used for all rigid-body simulations rather than having a patient-specific model based on clinical patients. In a similar manner, the variation in musculoskeletal geometries and properties from patient to patient were not accounted. From the experimental side, the largest source of error is the uncertainty associated with marker placement during the collection of lower extremity kinematics (Navacchia et al., 2016). However, differences in minimum knee angle and ROM between clinical data and predicted simulation values were on average less than two degrees. To determine the sensitivity of kinematic/loading conditions entered into the FE models a kinematic profile for one knee was modified by ± 3°. Within this range, no statistical significances were observed across predicted patellar mechanics or bone strain.

Similar to other studies, there were limitations associated with the retrospective nature of this study. While both surgeries analyzed demonstrated reductions in bone strain postoperatively, the PA+DFEO procedure displayed a more significant reduction in the magnitude of patellar strain between pre- and postoperative conditions. However, the criteria for patellar advancement is patella alta and quadriceps lag whereas the combined procedure accounts for those factors in addition to any knee flexion contractures. Hence, from this dataset, it is difficult to conclude whether the inclusion of the distal osteotomy is driving the changes in postoperative patellar strain or if the improvements observed are due to worse preoperative conditions.

4.3 Future Work

Moving forward, the framework developed can be further expanded upon experimentally, computationally, and clinically. For this study, the density-elasticity relationship selected was for femoral cortical bone as, to date, no such relationship has been calculated for the patellar bone. As such, one of the first aims would be to run mechanical tests on patellar bone specimens to experimentally derive a new relationship to use for the assignment of elastic modulus values. Preferably the specimen samples would be for an age range comparable to the clinical patients.

One of the foreseeable concerns regarding the feasibility of computational models as clinical tools is the time and knowledge required to both segment the PF geometries as well as to set up the simulation. To mitigate those concerns principal component analysis, radial basis functions, or other mesh morphing techniques could be applied to modify an existing model to match a new patient. For any of these techniques to be implemented a rough reconstruction of articular cartilage and bony geometries are still needed which would require some amount of training. However, prior researchers have developed methods to automate the segmentation of articular cartilage from MRI scans (Dodin et al., 2010; Fripp et al., 2007) which could be applied to bypass training and further reduce the setup time required.

The current work was a collaborative effort with Dr. Jason Rhodes, an orthopeadic surgeon at Children's Hospital Colorado, who is in the position to implement and oversee a larger scale clinical study. The prospective study would take the current setup to determine an optimal treatment plan, implement those surgical decisions, and then follow patients to evaluate their long-term outcomes. However, to be more inclusive of additional underlying factors the computational platform would be modified to include more surgical procedures as well as to analyze other pathological gait patterns.

For a long-term goal, with sufficient gait analysis and medical imaging data, simulation results could be used to create a multi-variable statistical model to sub-sample for age, gender, crouch severity, etc. and predict expected changes in patellar mechanics and bone strain. Results could then be concisely summarized as an intervention algorithm to assist in determining optimal treatment pathways. In this manner, the use of computational models would be used as a subject-specific tool to develop targeted interventions that reduce the risk of anterior knee pain and early onset osteoarthritis on a patient-specific basis.

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