Quantifying the Roles of the Menisci and ACL and the Success of a Cruciate-Substituting Total Knee Replacement System in Constraining the Knee

By

Matthew Hastings Dickinson

Submitted to the graduate degree program in Bioengineering and the Graduate Faculty of the University of Kansas in partial fulfillment of the requirements for the degree of Master of Science.

_____

Dr. Lorin Maletsky, Chairperson

_____

Dr. Lisa Friis, Committee Member

_____

Dr. Carl Luchies, Committee Member

Date Defended: December 15, 2016

The Thesis Committee for Matthew Dickinson certifies that this is the approved version of the following thesis:

Quantifying the Roles of the Menisci and ACL and the Success of a Cruciate-Substituting Total Knee Replacement System in Constraining the Knee

_____

Chairperson Dr. Lorin Maletsky

Date approved: _____

Abstract

Excessive anterior-posterior (A-P) and internal-external (Int-Ext) laxity has been associated with early onset osteoarthritis. The anterior cruciate ligament (ACL) and the menisci are the most commonly injured soft tissue structures in the knee during sporting events and activities of daily living. A-P and Int-Ext laxity are also a primary concern in total knee arthroplasty as they have been tied to surgical outcomes and patient satisfaction. Conflicting reports have indicated greater success in achieving targeted laxity measures and appropriate femoral rollback in cruciate-retaining (CR) and cruciate-substituting (CS) designs. The first aim of this study was to characterize the roles that the ACL and menisci play in A-P translation and Int-Ext rotation. The second aim was to quantify the differences in constraint between a CR and CS design and determine which prosthesis better achieved femoral rollback. A series of manual manipulations were performed on six cadaveric specimens before and after a meniscectomy, ACL-resection, and CR and CS total knee replacements. Kinematics were calculated using the Grood-Suntay coordinate system definition, and A-P translation was assessed by tracking the lowest point on the medial and lateral condyles of a femoral bone model. These kinematic data were used to fit radial basis functions that approximated the passive constraint to serve as consistent measures of laxity across conditions. The primary role of the menisci in joint constraint was in external rotation; a 6° increase was observed at 80° flexion following meniscectomy. Up to 3° more internal rotation was attributed to the resection of the ACL in early flexion (0°-30°), and a maximum of 10 mm more anterior tibial translation was observed at 30°. This work provides an in-depth description of the roles of the menisci and ACL and inform evaluations of reconstruction and replacement procedures. No significant differences in either Int-Ext or A-P laxity were seen between the CR and CS prostheses; however, posterior femoral

rollback was not retained in the CS design. These results indicate that a CS implant can be used to achieve similar joint laxity as the CR design for patients with a deficient PCL, but the effects of reduced femoral rollback must also be considered.

Table of Contents

List of Figures

List of Abbreviations

M-L	Medial-Lateral
A-P	Anterior-Posterior
S-I	Superior-Inferior
F-E	Flexion-Extension
Vr-Vl	Varus-Valgus
Int-Ext	Internal-External
LP	Lowest Point
LLP	Lateral Lowest Point
MLP	Medial Lowest Point
TF	Tibiofemoral
ROM	Range of Motion
RBF	Radial Basis Function
ACL	Anterior Cruciate Ligament
PCL	Posterior Cruciate Ligament
CR	Cruciate-retaining
CS	Cruciate-substituting

CHAPTER 1. Introduction and Significance

Maintaining proper joint constraint is linked to the health of the knee and comfort for patients. Cartilage thinning occurs after changes to internal-external (Int-Ext) rotation [1], and early development of osteoarthritis has been linked to excessive Int-Ext and anterior-posterior (A-P) laxity [2]. The anterior cruciate ligament (ACL) and menisci are soft tissue structures in the knee that govern tibial rotation and A-P translation and play a role in proprioception and stability [3]. Damage to these structures raises the risk of arthritis and other knee injuries [1, 4, 5]. Due to the prevalence of meniscal and ACL tears both in sports [6] and daily living [7], understanding the specific contributions of the menisci and ACL to constraint is necessary so that surgical interventions, reconstructions, and implant designs can be designed and validated.

There are two competing viewpoints on whether the posterior cruciate ligament (PCL) should be retained during in total knee replacements. Surgeons and researchers that advocate for cruciate-retaining (CR) designs advocate that it maintains more natural kinematic patterns, including the posterior femoral rollback observed in flexion [8, 9]. The PCL has been reported as a contributing factor to rollback due to its alignment and increased tension in flexion [10]. However, the literature also has reports of greater rollback in both cruciate-substituting (CS) and posterior-stabilized (PS) TKRs [11-13]. The primary interest in femoral rollback lies in the fact that it can increase the flexion range of motion (ROM) by reducing the forces the quadriceps need to exert as the tibia displaces anteriorly [14]. Most prosthetics that sacrifice the PCL have a cam-post interaction that engages as the knee flexes to prevent posterior tibial translation and often elevates varus-valgus (Vr-Vl) and Int-Ext constraint [15]. Reducing the contribution of soft tissues to joint constraint may sometimes be necessary [16], but this comes at the cost of lowering knee functionality in certain activities such as tennis and golf. A CS design has raised

anterior lip and heightened tibiofemoral conformity and may increase ROM while ensuring A-P and Int-Ext constraint.

This research had two main objectives: 1) quantify the changes to tibiofemoral constraint following meniscectomy and ACL resection and 2) quantify the passive constraint of a cruciate-substituting total knee replacement system relative to the equivalent cruciate-retaining design. This chapter has laid out the importance of the study reported herein. The second chapter is a literature review establishing the background information related to the anatomic structures and prosthetic designs important for this study. The third chapter describes the effects of meniscectomy and ACL-resection on tibiofemoral kinematics and uses this to describe the roles of the involved anatomic structures. Chapter four presents the results comparing a CR and CS design with respect to their constraint of Int-Ext rotation, A-P translation, and femoral rollback. The fifth chapter of this thesis concludes the work and introduces future directions for research to continue.

CHAPTER 2. Literature Review

2.1 Anatomy and Physiology

The motion and stability of the knee are largely influenced by both the soft tissue structures at the joint and the geometry of the articulating bones or implants. This review describes the anatomy and physiology of the menisci and cruciate ligaments. It also addresses how different total knee replacement (TKR) designs that either retain or substitute the posterior cruciate ligament (PCL) affect the kinematics and constraint of the knee.

2.1.1 Anterior Cruciate Ligament

The anterior cruciate ligament (ACL) originates from a posteriomedial fossa, or depression, on the lateral femoral condyle and inserts onto the anterior portion of the intercondylar fossa on the tibia [17]. The ACL is made up of two sets of connective tissue: the anteromedial and posterolateral bundles [18]. The ACL and PCL cross between the femur and the tibia.

The ACL is the primary restraint of anterior tibial translation [19, 20]. The constraint provided by the ACL is not uniform across flexion [21, 22]. The ACL has been shown to shorten as the knee flexes, which reduces the tension in the ligament and diminishes its ability to restrain motion [22]. The maximum strain in the ACL occurs when the knee reaches 30 degrees flexion, and the minimum occurs in full knee flexion (>120 degrees) [21, 23]. It has been shown that upon ACL resection, there are significant increases in anterior movement of the tibia throughout flexion [20, 24, 25].

The ACL has also been reported to have a secondary role in constraining internal-external (Int-Ext) rotation; however, there are inconsistencies in the literature. Gollehon et al. reported that ACL resection alone did not affect internal rotation, and Wroble et al. concluded

that the primary resistance to internal rotation comes from the lateral collateral ligament and antero- and posterolateral joint capsule structures [26, 27]. Contrary to this, Ahn et al. found increased internal rotation upon a simulated pivot shift test [24]. A five degree increase in external rotation has also been observed in ACL deficient subjects during stair ascension [28]. Additionally, the strain in the ACL is increased by application of internal torque [4, 29]. This strain reached a maximum at full extension and hyperflexion exhibiting a different trend than that observed either in passive flexion or applied anterior load, which may be explained by the primary contributors to IE constraint, namely the collateral ligaments, playing a diminished role in the highly flexed knee [29]. Maximum forces through the ACL are achieved by a combination of anterior load and internal torque when the knee is mostly extended [4, 30].

2.1.2 Posterior Cruciate Ligament

The PCL originates within the intracondylar notch on the medial femoral condyle and attaches to the posterior intercondyloid fossa on the tibia [31]. Similar to the ACL, the PCL has an anterolateral and posteromedial bundle [32]. The PCL is the strongest of the knee ligaments, which partially explains why it is injured less frequently than the others [33].

The PCL is the primarily responsible for constraining posterior tibial translation [19, 34]. During activities of daily living, specifically stair ascent, patients with PCL deficiency often experience a posteriorly subluxed tibia, which can lead to discomfort and instability [34]. The PCL has its largest effect on posterior motion around 90°, but it contributes to constraint throughout flexion [26, 35]. The minimum strain in the PCL occurs at full extension where it provides the least restraint to motion [23]. Few studies have examined how PCL deficiency affects the medial and lateral compartments of the knee individually, but Logan et al. showed that the medial compartment was more affected by the PCL [36].

The PCL also acts as a stabilizer of external rotation. Excessive external rotation in conjunction with posterior laxity at 90° has been used to diagnose PCL injury [37]. Additionally, Kennedy et al. reported a slight, but significant, reduction in external constraint following an isolated PCL resection throughout flexion [38]. They also reported that it acted to restrain internal rotation, especially in deep flexion. Gollehon et al. reported that isolated PCL resection did not affect external rotation, but their study was limited to flexion below 90° [26]. The effect of the PCL on IE laxity has been observed to be significantly less than that of the collateral ligaments [39].

2.1.3 Menisci

The menisci are fibrocartilage structures that cover roughly two-thirds of the tibial plateaus [40]. They have a semicircular shape, and the majority of the fibers run circumferentially, which allows axial stress to be converted into a hoop stress that helps the menisci bear compression. The meniscus is separated into three sections: an anterior horn, a meniscal body, and a posterior horn [41]. Two entheses, which are connective tissues that attach a ligament, tendon, or meniscus to bone, connect the menisci to the tibial plateau at the anterior and posterior horns [42]. The posterior horn of the medial meniscus is more pronounced than the anterior horn. The lateral meniscus is smaller and more mobile than the medial meniscus [43]. There is generally more motion in the lateral compartment during normal daily activities [20]. The menisci aid with load distribution, joint stability, proprioception, and improves conformity between the convex femoral condyles and flat tibial plateaus [3, 40, 44, 45].

If the meniscus is damaged to an extent that it is not repairable, a total meniscectomy is often performed to reduce pain. This causes the center of contact to move posteriorly and slightly toward the center of the knee [6, 44, 46]. Contact stresses also increase as the contact area

diminishes [44, 46]. A full radial tear near the posterior horn has been shown to produce similar stress profiles [46] and kinematics [47] as a total meniscectomy, which is likely because the circumferential fibers are severed preventing the development of hoop stress in the meniscus. An incomplete tear, even one that crosses 90% of the meniscus, has significantly reduced stress from a total meniscectomy indicating that meniscal function is somewhat preserved in the cases of severe, but not total, meniscal tears [44].

The roles of the medial and lateral menisci to anterior-posterior (A-P) and Int-Ext constraint are not fully understood. Isolated medial meniscus damage or resection has been showed by some researchers to result in an increase in anterior tibial translation [48-50], but other groups show no change in A-P laxity [42, 51]. It is generally accepted, however, that it plays a substantial secondary role in A-P constraint following ACL injury [20, 24, 25, 52, 53]. A lateral meniscectomy does not significantly affect tibial translation in the absence of other injuries [20], but it does act as a secondary constraint following ACL injury [54]. A total bilateral meniscectomy increases tibial translation to a greater extent than either meniscectomy individually [55]. Both menisci are involved in controlling the Int-Ext rotation of the knee. The prominent posterior horn of the medial meniscus prevents anterior tibial translation of the medial plateau, which explains observations that external rotation is primarily affected following meniscectomy [46]. Increases in external rotation have been observed in both cadaveric experiments and in live patients walking [56, 57] or ascending and descending stairs [6]. Some studies report the largest deviations around 30 degrees knee flexion [42, 46], but others report increasing laxity in deeper flexion [6]. The lateral meniscus has a smaller role in Int-Ext constraint, and these changes are usually only seen in conjunction with an ACL injury [20]. Approximately three degrees more internal rotation has been reported in knees with an ACL

injury and a lateral meniscal tear [28]. Both medial and lateral meniscal tears are more common following ACL injury or reconstruction [58]. The compression of the joint has been implicated in affecting the contribution of the menisci with the menisci having a larger role in constraint as the knee is compressed [55].

2.2 Total Knee Replacement

Patients experiencing chronic knee pain or instability often need to have total knee arthroplasty in order to perform everyday activities and improve their quality of life. There are several different prosthetic designs that differ in the amount of natural tissue is retained. Three common types are cruciate-retaining (CR), cruciate-substituting (CS), and posterior-stabilizing (PS). CR designs sacrifice the ACL but retain the PCL. CS TKRs remove both cruciate ligaments and use a more conforming tibial insert with a raised anterior or posterior lip. PS designs also sacrifice both cruciate ligaments, but the tibial insert contains a post that interacts with a cam in the intercondylar notch of the femoral component to control posterior tibial translation. The decision whether to retain or resect the PCL is generally up to the surgeon's discretion. It is often desirable to keep the PCL in knees that have also had MCL released because it helps stabilize the medial compartment [59]. Proponents of CS and PS implants prefer to remove the PCL to reduce overall variability in femoral rollback, which is the posterior motion of the femur on the tibia that naturally occurs with increasing flexion [60]. Surveys and functional evaluations of these prosthetics have been thoroughly conducted with equivalent performance and patient satisfaction noted between implants [61-66]. In this study, CR and CS TKRs are used to examine the effects of the PCL in TKR systems.

2.2.1 Cruciate-Retaining Design

CR TKRs are preferred by some because it is believed that the PCL maintains joint stability in deeper flexion by securing the relative position of the femur and tibia [67, 68] and potentially improves proprioception [69], although Simmons et al. reported otherwise [70]. Retaining the PCL also diminishes stresses in the fixtures by transmitting some of the load [71]. However, significant problems with instability and pain occur if the PCL deteriorates following surgery, which may happen if the PCL was released too much during surgery or was in poor condition already [14]. If the PCL is too lax, it allows paradoxical anterior motion during mid flexion [72, 73] and instability [74], but this may also be dependent on the rate of change of the radius of curvature on the femoral condyles, which has been shown to cause this motion in prosthetics that have a uniform radius [75].

A more natural range of motion during activities of daily living has been reported for CR designs [69, 76]. Some researchers have reported that the overall ROM of the knee is improved in CR TKRs due to an increased femoral rollback [8, 9], which increases the moment arm of the quadriceps muscles due to the relative anterior translation of its insertion on the tibia [14], but other groups have found insufficient rollback and flexion ROM in these designs in comparison with CS and PS TKRs [11-13]. With appropriate ligament balancing techniques, Heesterbeek et al. showed that CR TKRs can match natural contact mechanics [73].

2.2.2 Cruciate-Substituting Design

In certain cases, however, use of a CR TKR is not possible because the PCL is incompetent or is damaged during the surgery. The choice is then between the PS and CS TKR. CS designs do not have an issue of cam-post impingement, have an easier surgery, and preserve more bone than posterior-stabilizing TKRs because there is no need for the box cut [77] while

achieving the same performance as a PS TKR [63]. They also minimize patellar clunk, which is a syndrome in which scar tissue build up gets lodged in the notch during normal activities, because the femoral notch is smaller in CS TKRs than PS TKRs [65, 78]. CS implants also have the advantage of being able to be used in the aforementioned cases where the PCL is damaged during surgery as the femoral component and tibial tray are the same; only the tibial insert changes between CR and CS designs [65]. Watanabe et al. showed that there was greater knee flexion following PCL sacrifice in a CR TKR [79]. Furthermore, they demonstrated a reduction in external rotation during squatting, lunging, and kneeling. Massin et al. also revealed a decrease in both AP and IE laxity following replacement with an ultracongruent CS TKR relative to the natural knee [80].

CHAPTER 3. Quantifying the Roles of the Anterior Cruciate Ligament and Menisci in the Internal-External and Anterior-Posterior Constraint of the Knee

3.1 Introduction

The menisci and anterior cruciate ligament (ACL) are crucial soft tissue structures that affect load distribution, proprioception, and joint stability. Maintaining proper constraint is essential for the health of the knee joint. Changes to tibial rotation during gait has been shown to cause cartilage thinning due to abnormal shear stress patterns [1]. Reduced internal-external (Int-Ext) and anterior-posterior (A-P) constraint have been correlated to the development of osteoarthritis and further joint deterioration [2]. ACL and meniscal tears are known to increase the risk of arthritis and future injury to other ligaments [1, 4, 5, 81-83]. Meniscal tears are the second most common sports related injury behind only ACL rupture [6, 84, 85]. Furthermore, in people over the age of 65, 67% had a meniscal tear without the presence of osteoarthritis and 91% of those presenting with osteoarthritis had a meniscal injury in a previous study [7]. It is important to understand how the menisci and ACL contribute to joint constraint so that surgical interventions, reconstructions, and implant designs can be evaluated and modified to minimize the risk of future injury or osteoarthritis.

Meniscal tears and meniscectomies produce a shift in the center of contact between the femoral condyles and the tibial plateau, which moves the load onto regions of cartilage that do not support the loads as well [44, 46, 47]. Eliminating the menisci reduces the contact area and increases the stresses on the cartilage. The effects of meniscectomy on A-P motion of the joint are not fully understood. Some researchers reported an increase in anterior tibial translation after medial meniscectomy [48-50], but others show no change [42, 51]. Both menisci are believed to behave as a secondary constraint to anterior motion of the tibia following ACL injury [20, 24,

25, 52-54]. External laxity has been shown to increase in both in vitro [46] and in vivo [6, 56, 57] experiments reaching a maximum in either early flexion around 30° [42, 46] or in deep flexion [6]. Patients demonstrate reduced flexion at the knee while walking and compensate for this at other joints [57] so it is important to study these tissues in cadavers to directly examine the mechanistic changes to joint motion. Anterior tibial translation is primarily restrained by the ACL. Research has shown that the maximum strain through the ligament occurs at 30° [21, 23], and an increase in laxity in early flexion is observed after ACL injury [20, 24, 25]. Internal torque also increases the strain in this ligament [4, 29], and changes to internal rotation have been observed in some studies [24, 86] but not others [26, 27] following isolated ACL injury. Current ACL reconstruction techniques do not always provide sufficient rotary stability [86].

The involvement of the menisci and ACL in constraint of the knee has been examined at specific flexion positions. This study quantifies the contribution of those soft tissue structures by sequentially resecting the tissues in a cadaveric model and evaluating the changes to the passive envelope of constraint, which describes the joint's primary kinematic degrees of freedom in response to specific loads throughout flexion. This provides more detail than isolated force-displacement curves and kinematic measures at discrete flexion positions. By characterizing the knee's full range of Int-Ext and A-P motion, a more complete understanding of meniscal tears, meniscectomies, and ACL ruptures is achieved.

Greater anterior tibial translation was hypothesized to occur following the bilateral meniscectomy and this increase would be greater on the medial plateau because of the more pronounced medial posterior horn and greater inherent mobility of the lateral meniscus. An increase in external rotation is also expected, again due to the removal of the medial posterior horn. Following the ACL resection, significantly more anterior tibial translation primarily in

early flexion was hypothesized because of the orientation and engagement of the ligament. Severing the ACL was expected to reduce constraint in internal rotation.

3.2 Methods

3.2.1 Cadaver Preparation

Six fresh-frozen cadaver specimens (all male, age 66.3 ± 5.1 years, BMI 24.7 ± 2.0) with no history of either degenerative bone disease or lower limb pathologies were obtained for this study. The specimens were thawed at room temperature for 24 hours prior to magnetic resonance imaging. All soft tissues more than 19.0 cm distal to the femoral epicondyle axis on the tibia and proximal to a mark 21.5 cm above the axis on the femur and were removed. Aluminum fixtures were secured to both bone ends with bone cement, and the fibula was secured to the tibia fixture with a hose clamp. A spacer was positioned between the fibula and the fixture as needed to ensure the normal spacing between the fibula and tibia.

3.2.2 Experimental Protocol

The specimens were mounted to a platform in an inverted position, and infrared-emitting rigid bodies were attached to the femur and tibia fixtures for tracking with an Optotrak Certus motion capture system (NDI, Waterloo, Canada) (Figure 1). A six degree-of-freedom force-torque sensor (JR3, Woodland, CA) was attached to the distal tibia with an apparatus that enabled Int-Ext torques, varus-valgus (Vr-Vl), and A-P loads to be applied to the joint.

A series of laxity evaluations was performed on each specimen. External loads were manually applied to the distal tibia to induce Int-Ext torques, Vr-Vl torque, and A-P loads at the knee ranging between ±8 N-m, ±10 N-m, and ±50 N, respectively, while minimizing the applied compressive load. Viscoelastic effects were minimized by slowly applying the external loads and forces to the joint. In addition, the knees were moved through the full range of motion briefly

prior to collection to condition the ligaments. To ensure that the experimenter consistently achieved the targeted loading sets, visual feedback that approximated joint load and flexion angle was provided using a custom LabVIEW application (National Instruments, Austin, TX). Additional manipulations were performed until appropriate coverage was reached. The necessary density of points to achieve an accurate approximation (normalized root mean square errors below 4%) had previously been determined [87].

A surgeon performed a total meniscectomy of both menisci and transection of the ACL with the aid of an arthroscope. Meniscectomies were performed prior to the ACL resection in all knees. The manual laxity evaluations were repeated after each condition. Prior to surgery, the articular cartilage of both the tibia and femur were digitized to align an MRI model.

3.2.3 Data Analysis

The MR images of the femur and tibia were segmented using 3D Slicer to create STL geometries of the bones and cartilage. These files were uploaded into Hypermesh (Altair Engineering, Inc., Troy, MI) and a uniform mesh was created for each bone. Probed points of the articular surfaces were used to register these models into the appropriate rigid body's reference frame. Points from the registered models defined the bone-fixed coordinate systems, which were used to calculate the kinematics of the joint using the 3-cylindrical open-chain system described by Grood and Suntay [88]. However, A-P displacement was calculated and analyzed using the lowest point [89] along the tibial superior-inferior axis of the medial (MLP) and lateral (LLP) femoral condyles. This was done to focus on the changes observed in each compartment of the knee to discern the effects of each menisci. To obtain these points, the registered femur model was divided into a medial and lateral compartment and transformed into the local tibial coordinate system. The MLP and LLP were calculated at each time point.

The external loads and torques measured at the distal tibia during the laxity evaluations were transformed to the tibial coordinate system. The loads were processed with a two-way, low-pass, 6th order Butterworth filter in MATLAB (Mathworks, Natick, MA). To minimize the effects of hysteresis, only the portion of these trials in which the targeted load or torque was increasing was included in subsequent data analysis (Figure 2). Radial basis functions were trained and used to create an envelope of constraint for each knee to facilitate comparison across specimens and between conditions. This methodology has been previously described in the literature [87, 90]. Briefly, the experimental data collected during the manual laxity evaluations served as the training set for the radial basis functions. The flexion angle, A-P force, Int-Ext torque, and Vr-Vl torque were treated as independent variables and the resultant MLP, LLP, Int-Ext rotation, and Vr-Vl rotation were the dependent variables. Since this method has been shown to be resilient to decimation and missing data, the regressed envelopes of motion are consistent representations of the constraint of the knee joint [87].

To quantify the differences following meniscectomy and ACL-resection, both low and high stiffness envelopes were approximated for each condition. These two regions were of interest because activities of daily living would fall in the low stiffness portion while extreme motions like a cutting maneuver would likely push the knees toward the end range of motion. Torque-rotation stiffness plots were used to identify the Int-Ext torque predicted for the two regions: 1.5 N-m for the low stiffness region and 6 N-m for the high stiffness region (Figure 3). Vr-Vl was not a focus of this study so the Vr-Vl torque was selected conservatively such that sufficient experimental points were available for all combined torque sets. Therefore, to examine the Int-Ext changes the radial basis function was used to approximate the kinematic measures for the following set of 1040 flexion-load coordinates. Flexion spanned from 0° to 120°, A-P load

ranged from -10 N to 10 N in 10 N increments, and the Vr-Vl and Int-Ext torques spanned ±4 Nm or ±1 Nm and ±6 Nm or ±1.5 Nm as shown in Figure 4. The resulting kinematics were visualized as two separate envelopes of constraint (Figure 5). The differences between the envelopes of each condition (intact, meniscectomy, ACL-resected) were calculated. Two-tailed paired t-tests ($\alpha<.05$) were used to evaluate the statistical significance of these differences.

The A-P motion of the medial and lateral LP was also analyzed using the radial basis function; however, due to the dependence of A-P translation on flexion angle, these data were better shown by evaluating specific isolines from the approximation. Zero Int-Ext and Vr-Vl torque was chosen and the changes in A-P translation in response to ±20 N and ±40 N A-P load were calculated. Two-tailed paired t-tests ($\alpha<.05$) were again used to evaluate the statistical significance of these differences.

3.3 Results

3.3.1 Meniscectomy

After the meniscectomy, a significant increase in external rotation ranging from 3° – 6° occurred from mid to deep flexion in the low stiffness envelope (Figure 6A). The largest change was seen at 80 degrees. The high stiffness envelope only reached a statistical significance in deep flexion with a maximum increase in external rotation of 5°. When subjected to anterior loads of 20 N and 40 N, there was an increase in anterior tibial translation of 2-3 mm on the medial side in early to mid flexion (Figure 7), but no changes were observed on the lateral side (Figure 8). No differences were observed in the MLP under posterior load (Figure 7), but significantly more posterior tibial translation of around 4 mm was observed between 60° and 90° on the lateral side (Figure 8). Both results indicate that high anterior or posterior loads cause the knee to externally rotate more than the intact knee.

3.3.2 ACL-Resection

The ACL-resection was performed after the meniscectomy so all differences and statistics were run between the ACL-resected case and the meniscectomy case. In both the low stiffness and high stiffness envelopes, a significant increase in internal rotation of up to 3° was observed in early flexion (Figure 9). Internal rotation also increased in mid flexion in the high stiffness envelope. Resecting the ACL led to a large increase in anterior tibial translation on both the medial and lateral sides of up to 10 mm in early to mid flexion (Figures 7 & 8). This corresponds to an increase in overall anterior translation without changing Int-Ext rotation.

3.4 Discussion

3.4.1 Menisci

This study found significantly greater external translation following meniscectomy. These changes occurred primarily in deep knee flexion, which is in agreement with some of the previous research on meniscectomy [6, 56, 57]. It is interesting that in mid flexion significant changes were only observed in the low stiffness envelope, which indicates that the menisci may play an important role in constraining rotation in the transverse plane initially, but the end range constraint is provided by other structures like the collateral and cruciate ligaments. Consistent with our hypotheses, an increase in anterior tibial translation of the medial plateau was also observed; however, the reduction in constraint occurred in early flexion. An increase in posterior tibial translation of the lateral plateau was also observed, which has not been previously reported. The anterior horn of the lateral meniscus likely provides significant constraint in mid flexion. The menisci are responsible for constraining the knee in Int-Ext and A-P motion at opposite times in flexion. It is possible that they become more involved in rotational constraint in deep flexion because the collateral ligaments are more lax in this range so secondary structures take

more of the burden. Contrastingly, the menisci are more prominently involved in constraining A-P translation in early flexion because the anterior and posterior horns are more conforming to the femur.

3.4.2 Anterior Cruciate Ligament

As hypothesized, significantly more anterior tibial translation occurred following the resection of the ACL. The largest change occurred at 30° to both the medial and lateral, which is in line with previous findings that the strain in the ACL was greatest in this region. By minimizing compression of the joint, the constraint offered by the geometric conformity of the femur and tibia is reduced, which may explain why such a large change in anterior translation was observed. The ACL was also responsible for constraining internal rotation near full extension. This is in agreement with previous findings [24], and may occur because the ACL and PCL bind in extension providing increased rotational stability. However, it should be noted that the menisci were also removed in this study, so it is possible that with the menisci intact, these changes would not be observed as was the case in previous works [26, 27]. The difference between the low and high stiffness envelopes of constraint indicated that the ACL governed the end range of motion and may not be as involved during low intensity activites.

3.4.3 Limitations

This study had a few limitations. First, a low number of specimens were used in this study (n = 6) so additional work will be done to expand the number of knees tested to increase confidence that the changes observed are representative of the larger population. A post hoc power analysis indicated that this study was sufficiently powered (power = .8) to detect a 2.8 mm change in A-P motion of the LLP, a 4.0 mm change in A-P motion of the MLP, and a 4.3° change in Int-Ext rotation. Another limitation of this study was that all tests were performed with

low compressive load, which may reduce or alter the effect of a meniscectomy because they play a critical role in transmitting that load across the joint. Additionally, the meniscectomy was always performed prior to the ACL resection so additional work should be done to identify the effects of an isolated ACL resection. Finally, due to the limitations of testing with cadaveric tissue, only double meniscectomies were studied so future research to examine single meniscectomies would be beneficial.

3.5 Conclusion

The findings of this research further our understanding of the consequences of common knee injuries and the underlying soft tissue structures. It is important to recognize that the menisci and ACL are involved in constraint to differing degrees across the flexion range. The ACL is partially responsible for constraining anterior and internal motion near full extension, which is why people are at greater risk of developing osteoarthritis following damage to the ACL. The medial meniscus also acts to constrain anterior tibial translation in early flexion, but meniscectomies have a much smaller effect on this translation than the ACL resection. Bilateral meniscectomy resulted in significantly more external rotation from mid to deep flexion. Lateral meniscectomies were not previously thought to affect knee kinematics (4, 41), but these results imply otherwise and additional investigation is warranted. In order to properly analyze the success of reconstruction techniques or evaluate the risks of meniscectomies, the full range of motion must be addressed and this work provides necessary data to inform these evaluations.

3.6 Figures

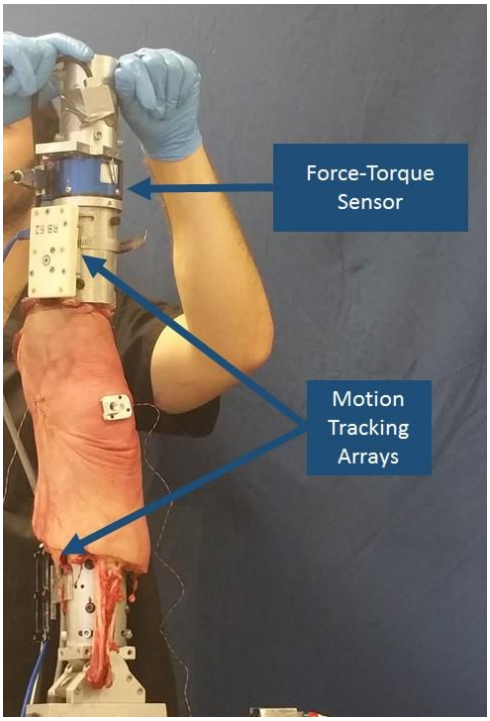

Figure 1. Motion tracking arrays are attached to the tibia and femur, and the force-torque sensor was secured to the distal tibia. The femur is mounted in an inverted position.

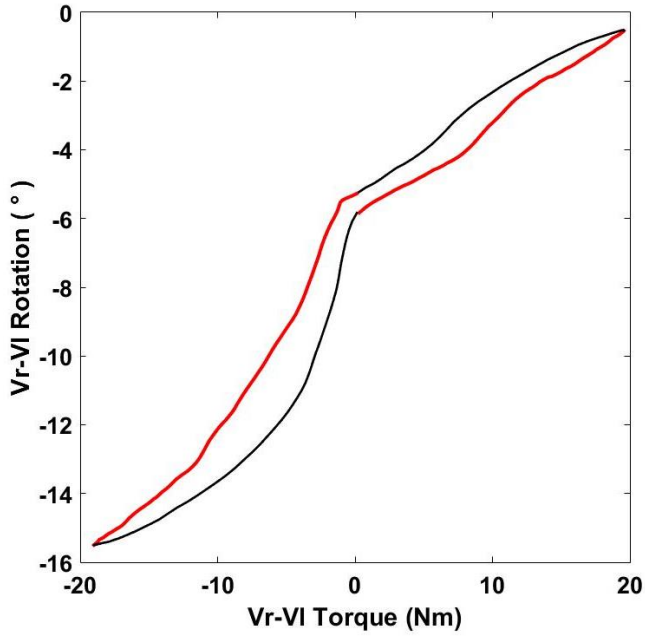

Figure 2. The load cell data was separated into loading (red) and unloading (black) regions. Only the loading regions were used.

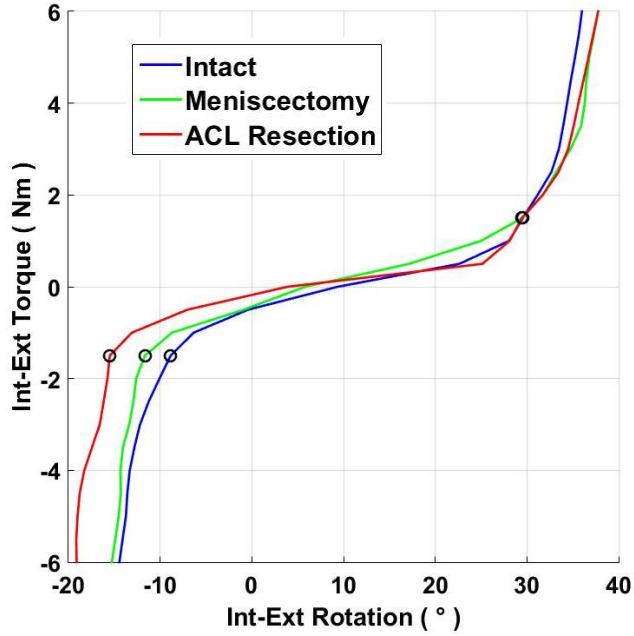

Figure 3. Representative Int-Ext rotational stiffness plot for all three conditions. The low-high stiffness division at 1.5 Nm has been marked.

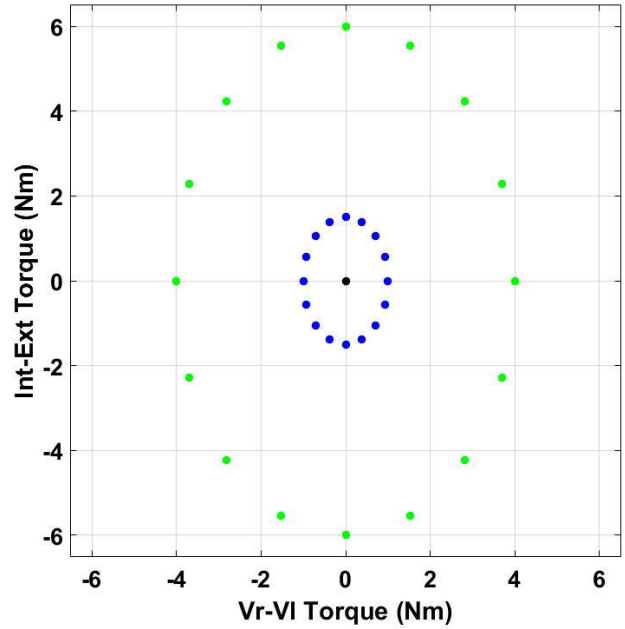

Figure 4. Predicted Vr-Vl and Int-Ext torque combinations for the low (blue) and high (green) load levels.

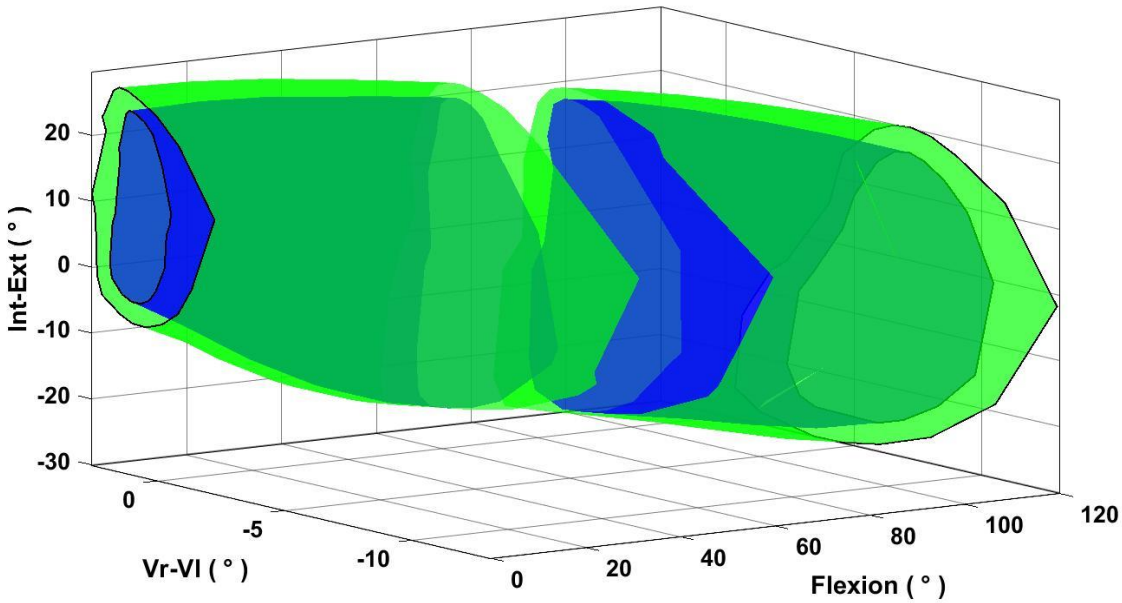

Figure 5. Representative passive envelope of constraint at the low (blue) and high (green) load levels. The region between 50° and 70° has been omitted to improve visualization.

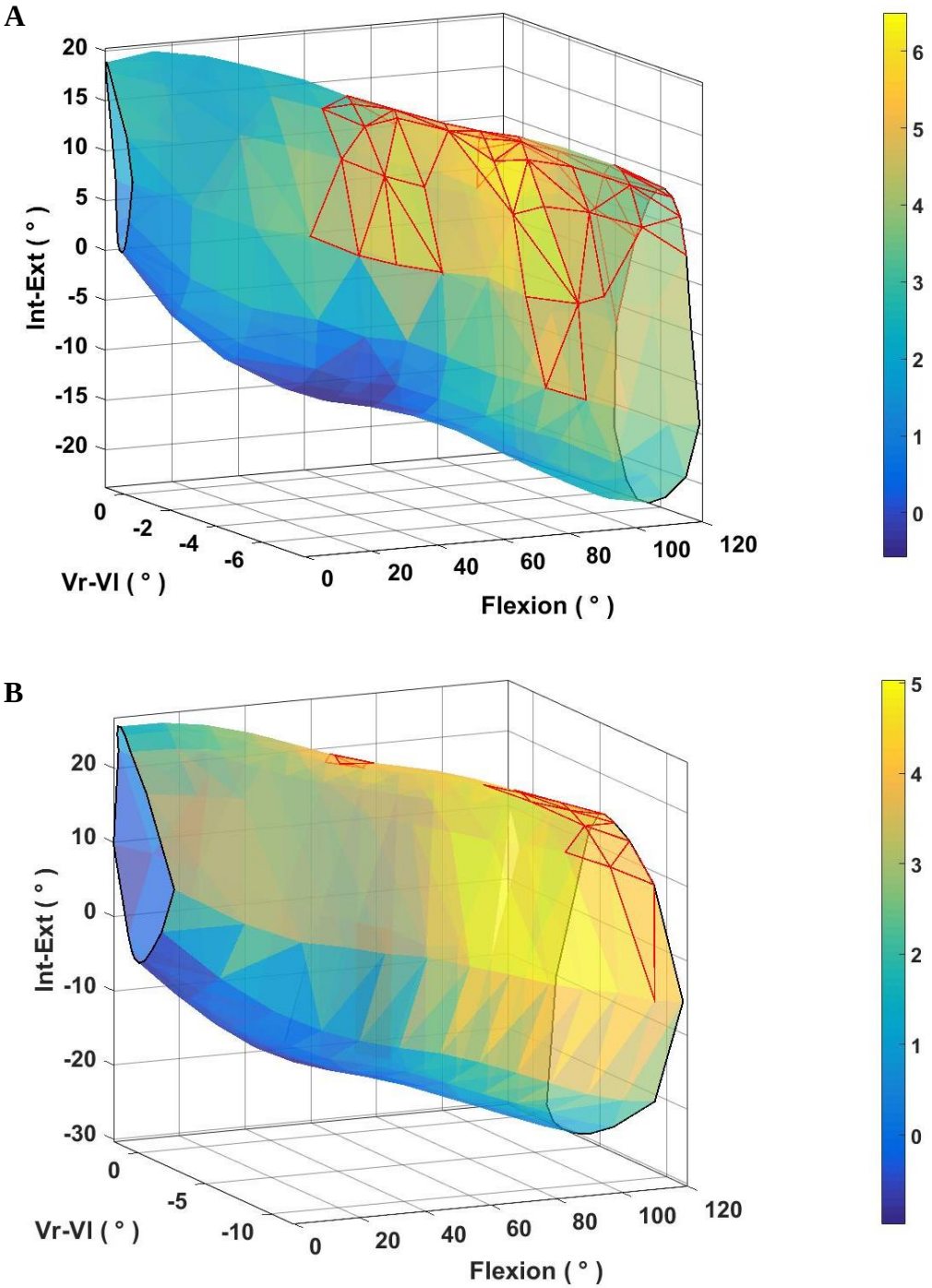

Figure 6. The mean passive envelopes of constraint for the intact condition that correspond to the (A) low stiffness and (B) high stiffness regions have been plotted. The color of each facet indicates the magnitude of change in Int-Ext rotation following meniscectomy. Facets highlighted in red indicate regions where significant differences ($p<.05$ for all vertices).

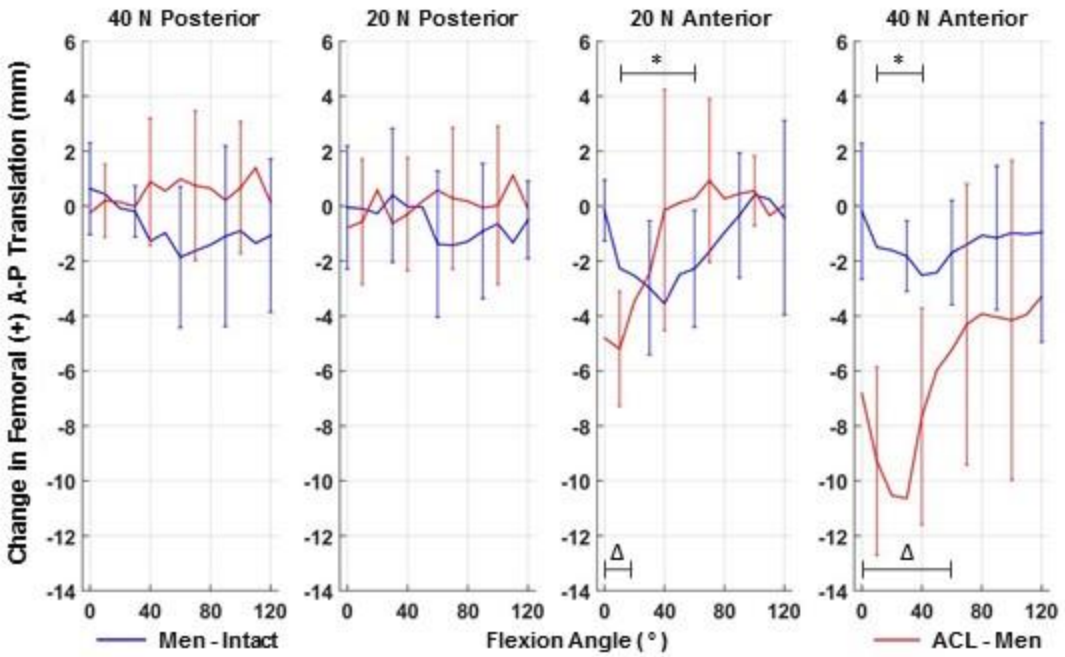

Figure 7. The changes to A-P translation of the MLP in response to 20 and 40 N anterior and posterior loads following meniscectomy and ACL resection. Increases in posterior femoral translation correspond to increases in anterior tibial translation. Significant changes (p<.05) after the meniscectomy (*) and ACL resection (Δ) are shown.

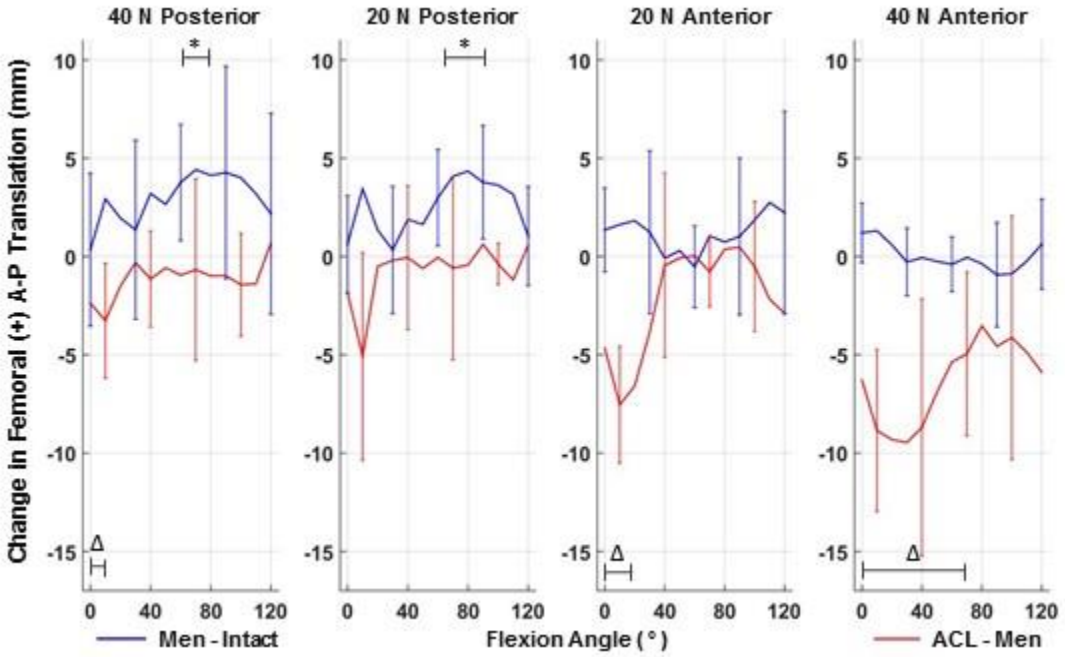

Figure 8. The changes to A-P translation of the LLP in response to 20 N and 40 N anterior and posterior loads following meniscectomy and ACL resection. Increases in posterior femoral translation correspond to increases in anterior tibial translation. Significant changes (p<.05) after the meniscectomy (*) and ACL resection (Δ) are shown.

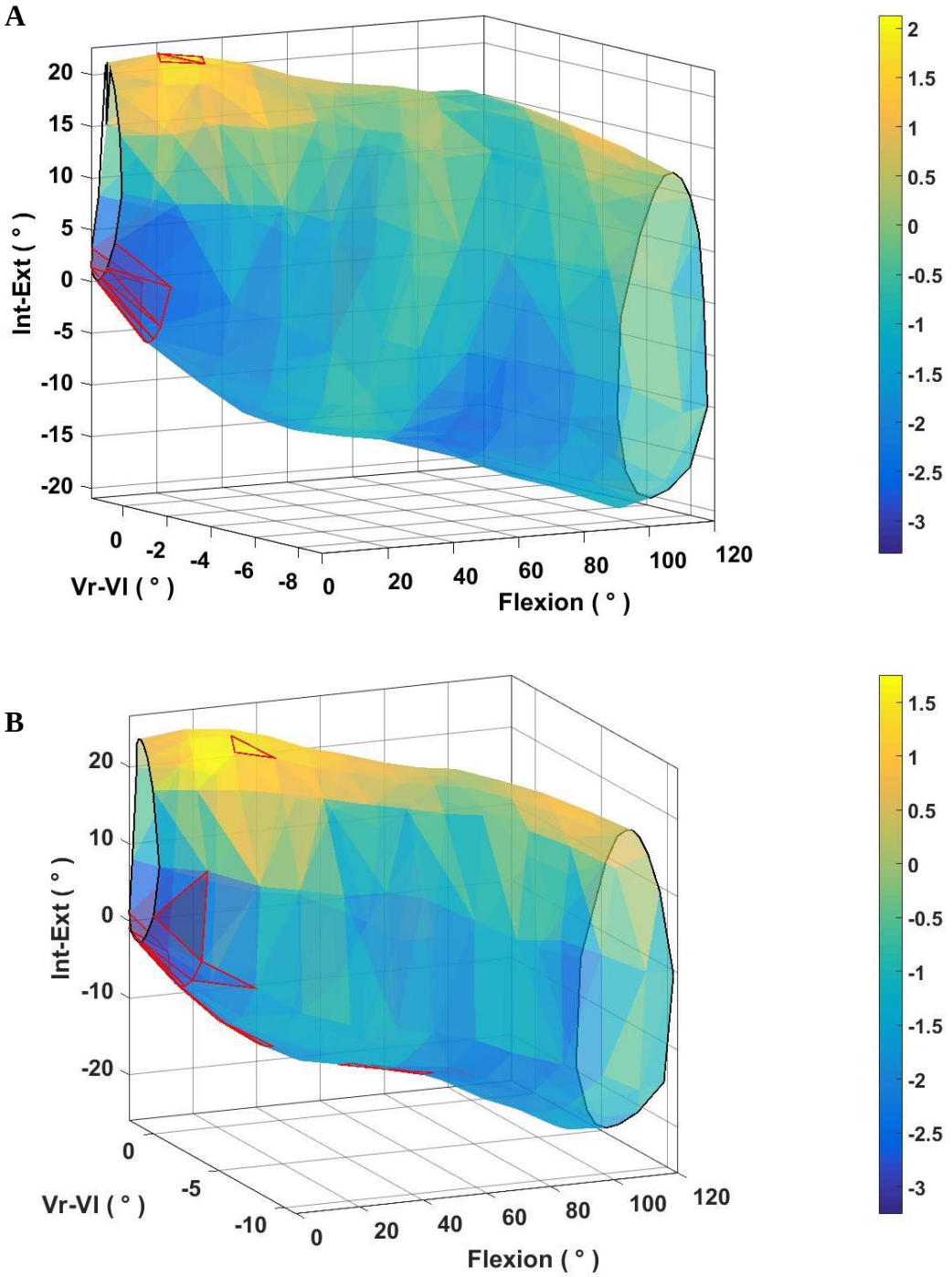

Figure 9. The mean passive envelope of constraint for the meniscectomy condition that correspond to the (A) low stiffness and (B) high stiffness regions have been plotted. The color of each facet indicates the magnitude of change in Int-Ext rotation following meniscectomy. Facets highlighted in red indicate regions where significant differences ($p<.05$ for all vertices).

CHAPTER 4. Quantifying the Performance of a Cruciate-Substituting Total Knee Replacement System Relative to the Cruciate-Retaining Design

4.1 Introduction

Currently, there is a debate about whether the posterior cruciate ligament (PCL) should be retained or substituted in total knee replacements (TKR). There is also a push by some toward bicruciate-retaining designs because it is thought that they will best restore natural knee kinematics and achieve better patient satisfaction. The PCL has a primary role in constraining posterior tibial translation [19, 34], and a secondary role in external constraint [37-39]. Additionally, the PCL is thought to play a role in femoral rollback [10], and more natural rollback is observed in cruciate-retaining (CR) designs [8, 9]. However, other researchers reported greater rollback in both cruciate-substituting (CS) and posterior-stabilized (PS) TKRs [11-13]. Achieving femoral rollback is desirable because it leads to improved flexion ROM due to the increase in quadriceps moment arm that occurs as the tibia displaces anteriorly [14].

When the patient has a lax or damaged PCL, it is common to sacrifice the ligament and use a PS design that has a central tibial post that engages with a femoral cam in flexion to restrain posterior tibial translation. These PS designs often constrain varus-valgus (Vr-Vl) and internal-external (Int-Ext) rotation to a greater extent [15], and have the added risk of cam-post impingement and greater chance of patellar clunking [65, 77, 78]. While increasing the geometric conformity reduces the contribution of soft tissues to joint constraint may sometimes be necessary [16], the reduction of ROM may impede certain activities and reduce patient satisfaction. Therefore, CS designs that utilize a more conforming tibial tray and raised anterior lip may lead to greater range of motion while still maintaining the necessary anterior-posterior (A-P) and Int-Ext constraint. Most research currently focuses on either CR or PS designs, so this

study aimed to characterize the performance of the CS implant relative to its CR alternative. Femoral rollback, A-P translation, and Int-Ext rotation were the focus of this comparison. It was hypothesized that no difference in posterior tibial translation or Int-Ext rotation would be observed due to the elevated anterior lip on the tibial insert. A reduced femoral rollback was expected in the CS design because it lacks a force to drive it posterior that the PCL provides as tension increases in flexion [26].

4.2 Methods

4.2.1 Experimental Protocol

The same cadaveric specimens and protocol were used as in Chapter 3. Following the ACL resection, the surgeon performed a total knee replacement using the OMNI APEX CR Knee system (OMNIlife Science, Inc., East Taunton, MA). The PCL was then removed and an OMNI Ultra Congruent tibial insert was inserted for the CS system (Figure 10). Two different surgeons (S1 and S2) participated in this study. The same series of manual manipulations was performed on each TKR system. Each implant geometry was digitized in the same way as the natural articular surface.

4.2.2 Data Analysis

STL geometries of the femoral component and tibial insert were uploaded into Hypermesh (Altair Engineering, Inc., Troy, MI) and a uniform mesh was created for each surface. The geometries were registered in the appropriate rigid body's reference frame using the probed articular surfaces. Points from the registered meshes defined the bone-fixed coordinate systems, and the same kinematic and lowest point definitions were used as in Chapter 3. The methodology to create the passive envelope of constraint and the analyses performed on them were consistent with Chapter 3. Because femoral rollback was of greater interest in this study,

the motions of the medial and lateral femoral lowest points without any load were examined. The radial basis function was used to predict this zero-load flexion path. Paired student t-tests ($p < .05$) were used to identify significant differences between the insert designs.

4.3 Results

Due to complications during surgery, only the CR implant was tested for one knee. Therefore, for the paired statistics, the remaining five specimens' data were compared. There were no significant differences in Int-Ext rotation between the CR and CS designs at either stiffness level (Figure 11). Slightly more external rotation in deep flexion was observed, but was not statistically relevant.

The MLP and LLP moved approximately 9 mm more posteriorly during unloaded flexion in the CR design than the CS insert, but this difference was not significant (Figure 12). Due to the large variance in this data, the LP motion of the individual knees were examined. Femoral rollback occurred in all CR knees operated on by S1 (Knees 1, 2, 5, 6), but paradoxical anterior femoral translation is observed in the two knees operated on by S2 (Knees 3, 4) (Figure 13). One knee dislocated at 120° under minimal compression. Paradoxical anterior femoral translation with flexion was observed in all CS knees (Figure 14).

Minimal differences were seen in response to posterior load between the CR and CS designs on tibial translation (Figures 15 & 16). No statistically significant changes to anterior tibial translation occurred, but on average the CS insert had 6 mm more anterior translation in deep flexion than the CR (Figures 15 & 16).

4.4 Discussion

The CS had similar laxity in Int-Ext rotation and posterior translation despite the resection of the PCL. This indicates that the heightened anterior lip and more conforming insert

geometry of the CS designs provides sufficient constraint without the need of either the PCL or a tibial post-femoral cam as used in PS implants. The increase in anterior tibial translation may have been a consequence of the loss of femoral rollback. In the CR designs, the femur translated near the posterior edge of the insert leaving little room for additional anterior tibial translation. However, in the CS design, the femur remained centrally or even anterior on the tibia so there was much more room for anterior translation.

In cruciate-retaining procedures, it is not uncommon for the surgeon to resect or release the PCL if excessive femoral rollback is observed. This is done subjectively when the surgeon assesses that the tibia is too far anteriorly displaced in flexion [91]. It is hypothesized that S2 released the PCL too much during the surgeries, which resulted in the loss of femoral rollback observed in the knees on which he operated. This is consistent with observations in previous studies that if the PCL is too lax in CR prostheses a paradoxical anterior motion of the femur on the tibia with flexion occurs [72, 73]. Combined with the limited sample size, these differences between surgeons makes it difficult to state anything conclusive about differences in femoral rollback between the two designs. Nevertheless, the results of this study indicate that the PCL may play a large role in driving femoral rollback as the knee flexes.

The two main limitations of this study are the small sample size (n = 5) and minimal compression. Additional specimens will be tested to determine if these findings are representative of a larger population. The low compressive load may reduce the effect of different geometries have on constraint.

4.5 Conclusion

The CS design tested in this study performs comparably to the CR system in the context of this study. It provides similar constraint in Int-Ext and A-P motion so in circumstances where

the PCL is either deficient or damaged during surgery, a CS insert may be used without additional bone cuts to maintain equivalent laxity. Additionally, it serves as another option for patients when the elevated Vr-Vl constraint seen in PS designs is not desired. The main kinematic difference between the CR and CS designs appears to be in femoral rollback. While the anterior lip on the CS insert sufficiently prevents posterior tibial translation, it does not apply a large enough posterior force on the femur as the knee flexes to retain femoral rollback. This is an important consideration because proper rollback can lead to better ROM and patient satisfaction. Future studies should investigate whether other CS designs behave similarly, and whether changes to the design can be made to achieve rollback. Activities with physiologic loads at the joint should also be simulated to quantify differences between the designs during activities of daily living.

4.6 Figures

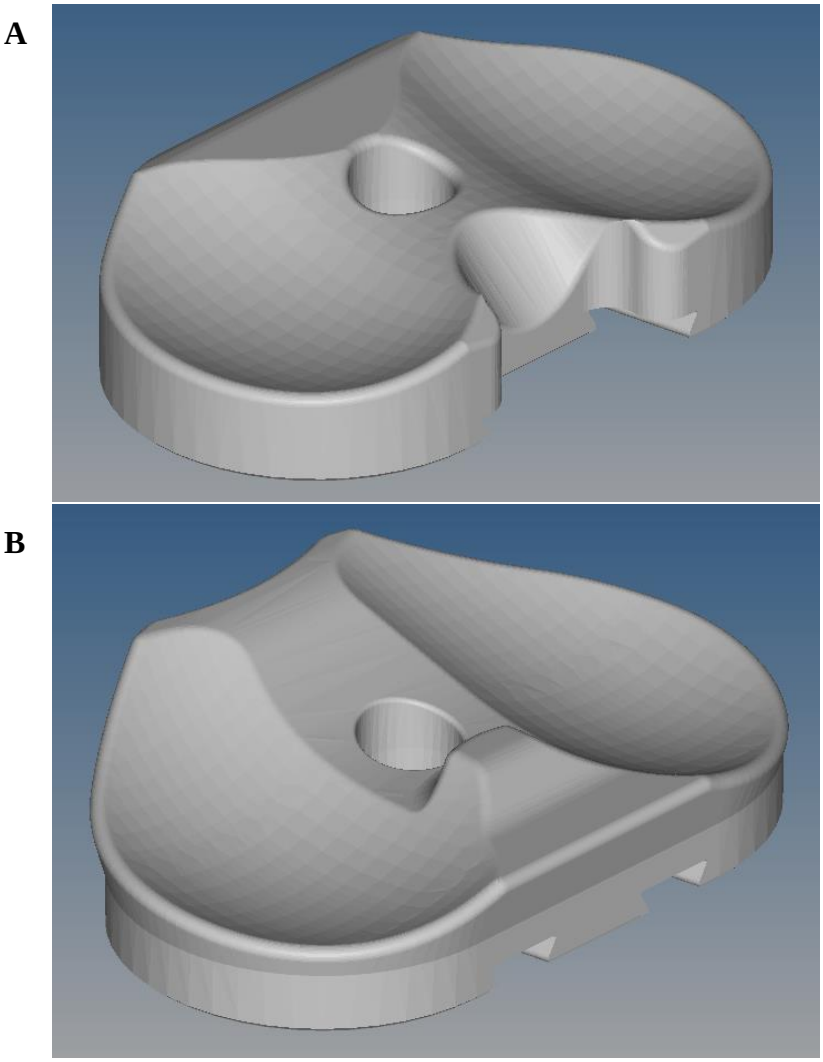

Figure 10. (A) Cruciate-retaining and (B) cruciate-substituting tibial inserts.

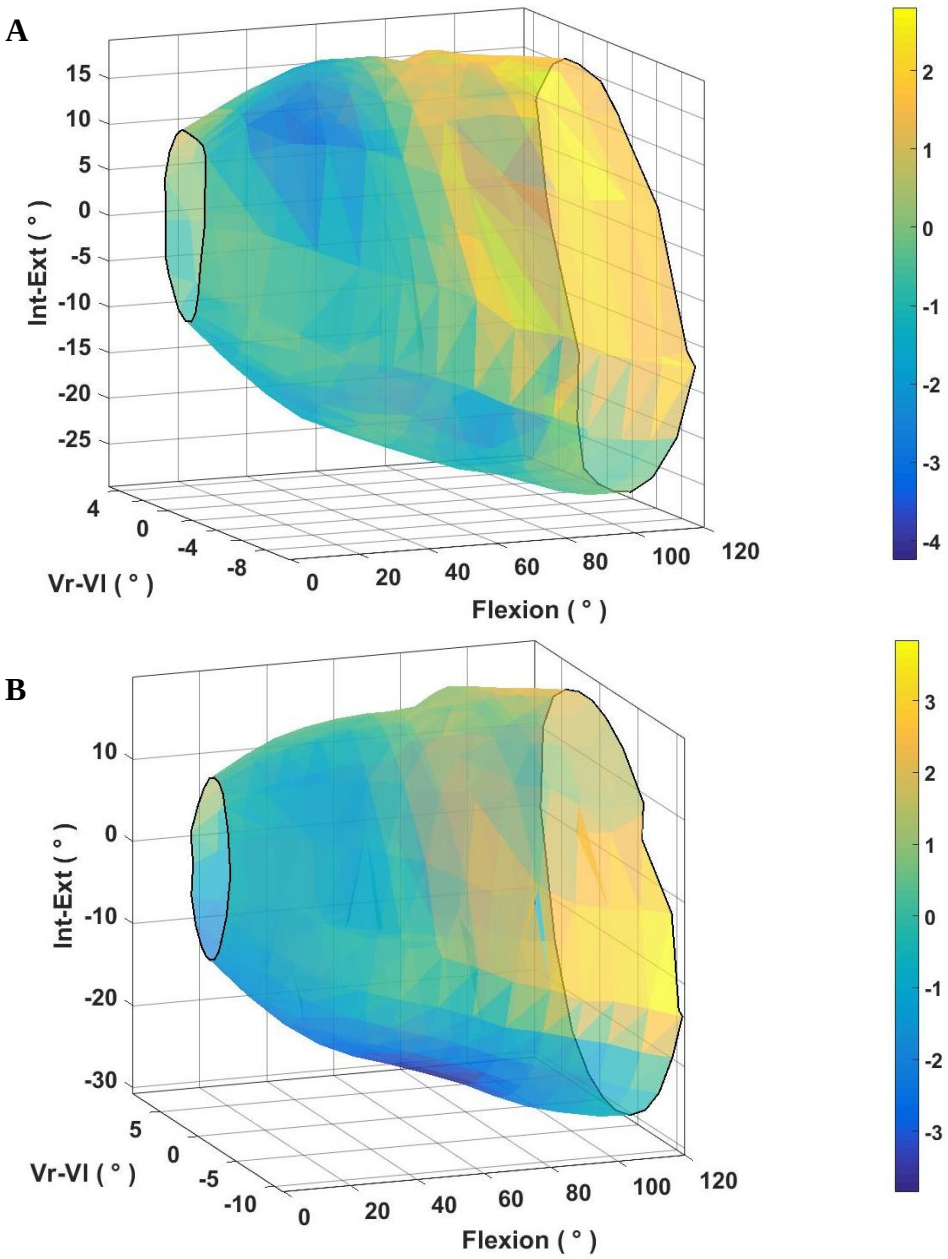

Figure 11. The mean passive envelopes of constraint for the CR prostheses that correspond to the (A) low stiffness and (B) high stiffness regions have been plotted. The color of each facet indicates the magnitude of change in Int-Ext rotation following PCL resection and replacement with CS prostheses. No significant differences ($p<.05$) between the two designs were observed.

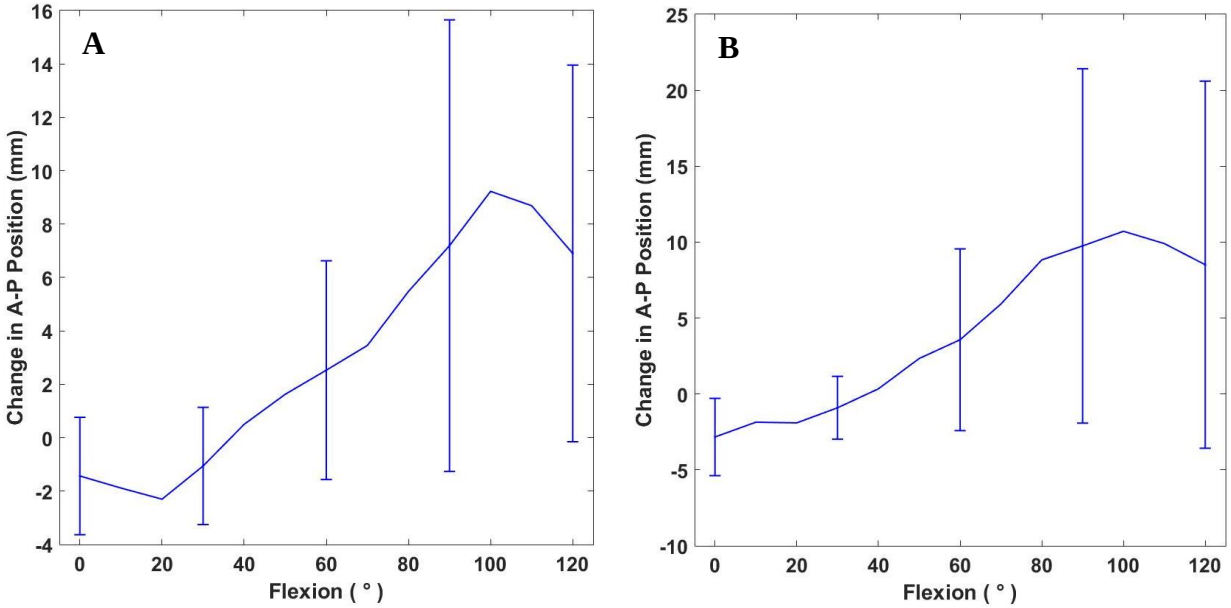

Figure 12. The mean difference in (A) MLP and (B) LLP position between the CR and CS designs during unloaded flexion is shown.. No significant differences were observed between the two designs.

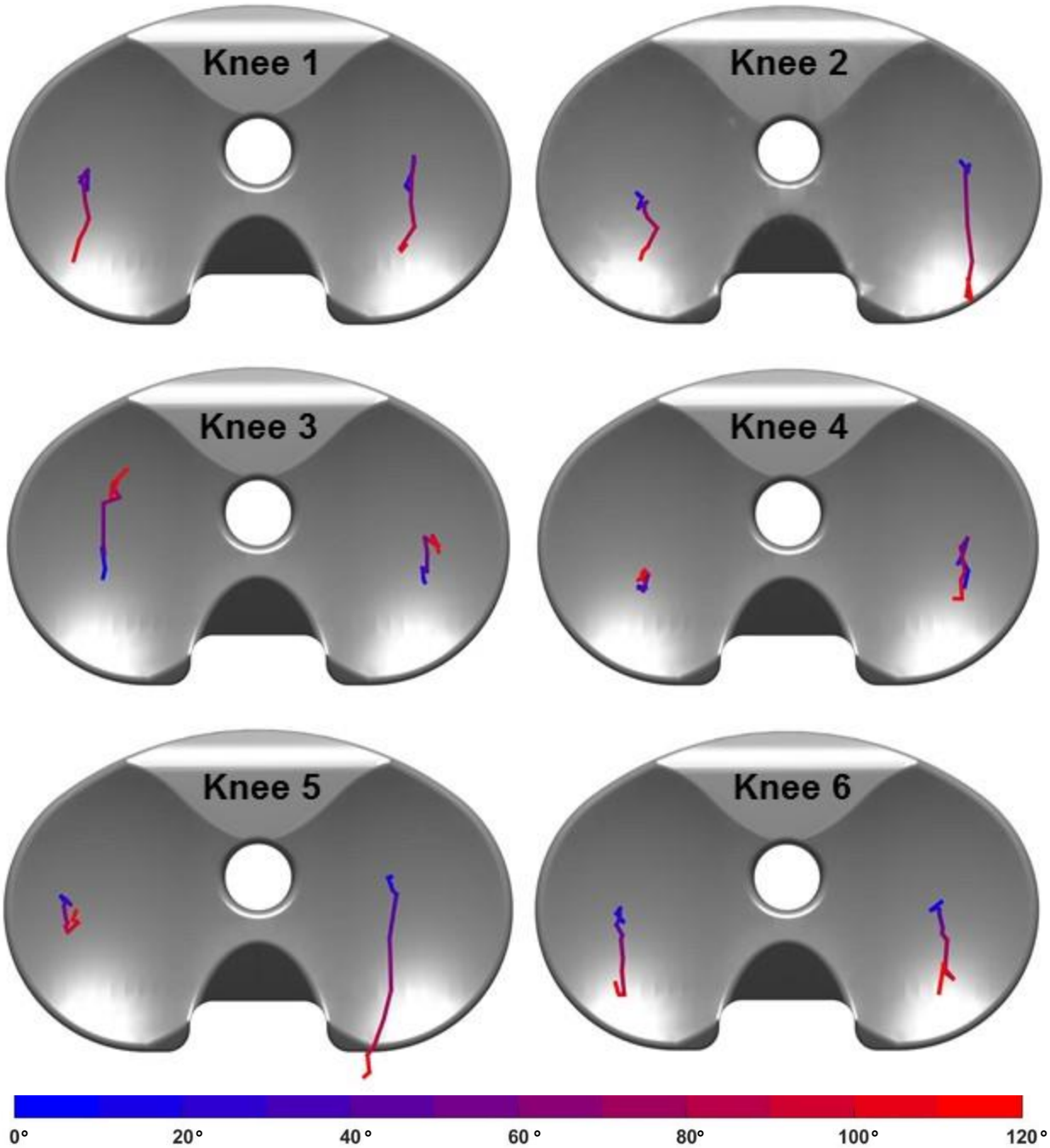

Figure 13. The lowest point motions during passive flexion for each CR are shown. All knees are displayed as right limbs; the left side is the medial compartment of the insert.

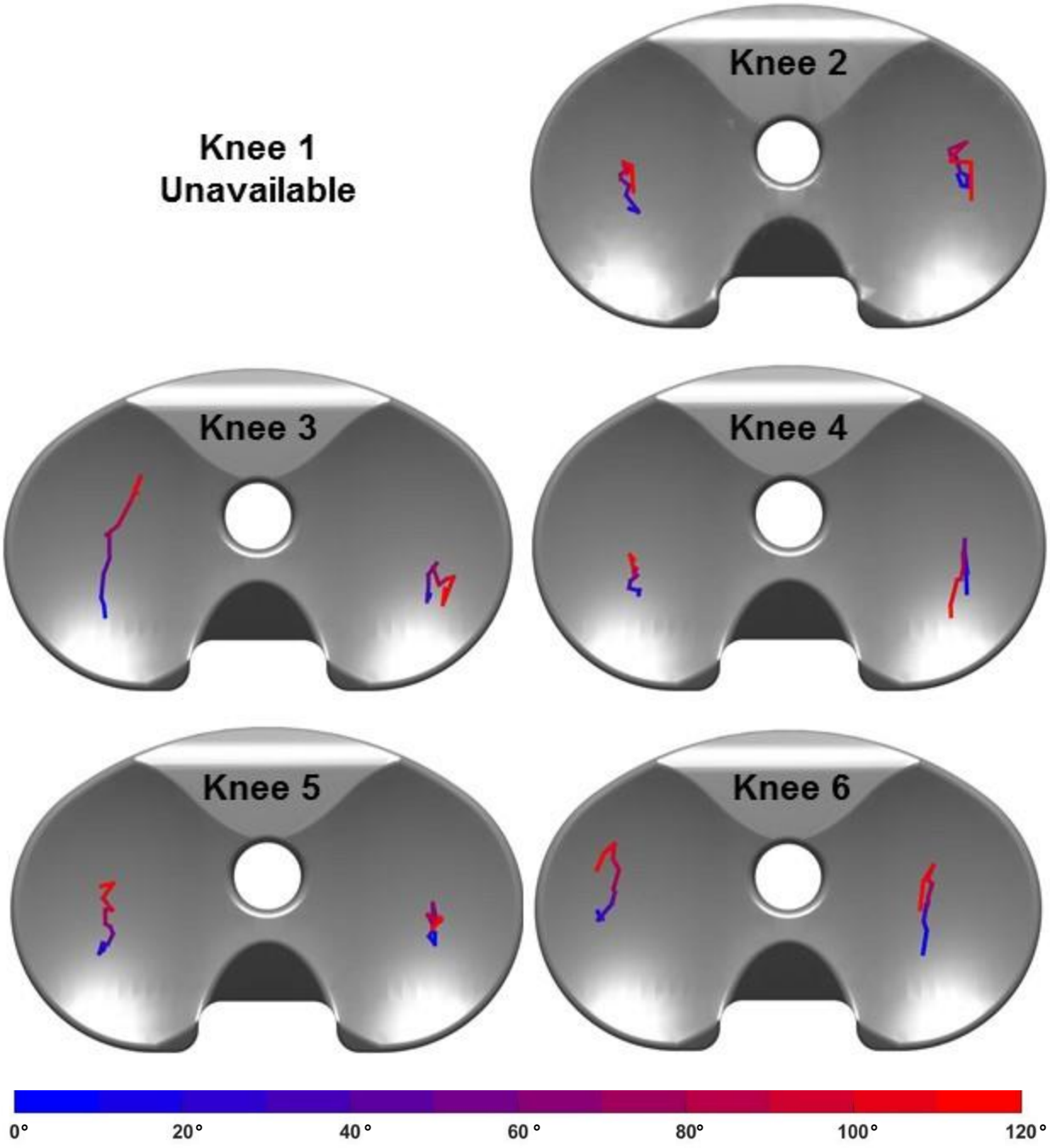

Figure 14. The lowest point motions during passive flexion for each CS are shown. Data was not collected on Knee 1 for the CS condition. All knees are displayed as right limbs; the left side is the medial compartment of the insert. Insert depicted is the CR design to facilitate comparison.

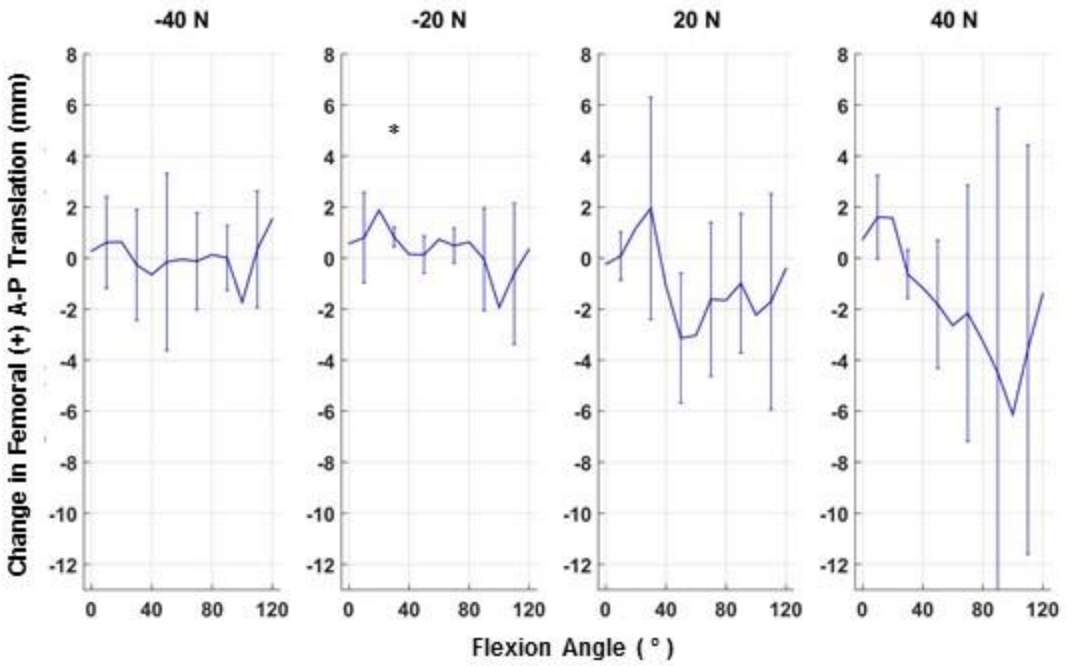

Figure 15. The difference in A-P translation of the MLP in response to 20 and 40 N anterior and posterior loads between CR and CS prostheses. Increases in posterior femoral translation correspond to increases in anterior tibial translation. Significant changes (p<.05) are shown (*).

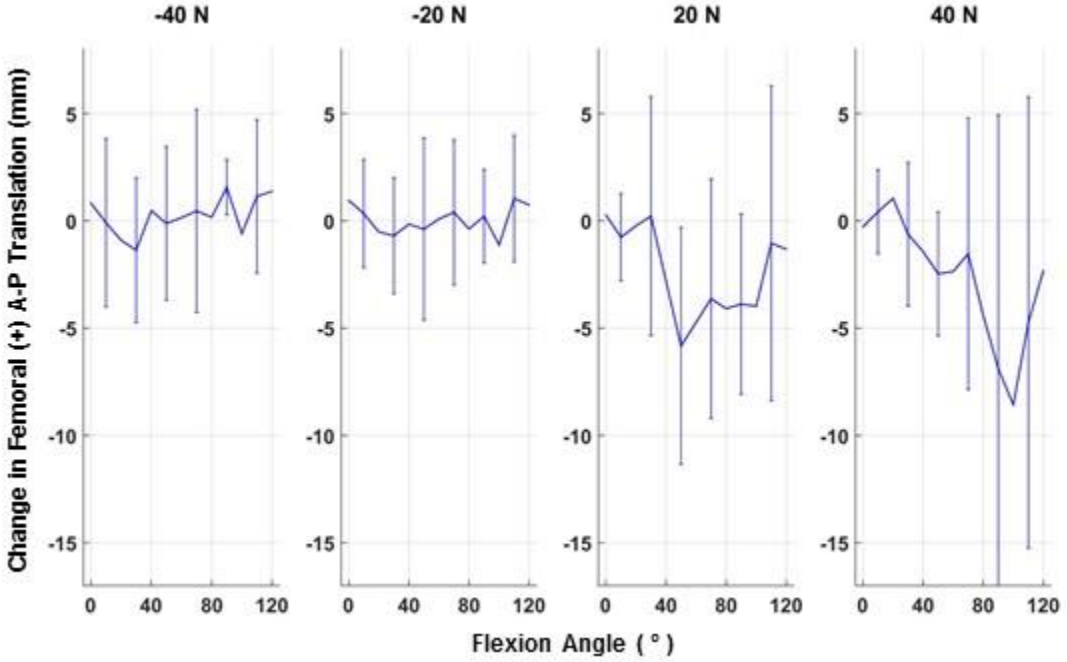

Figure 16. The difference in A-P translation of the LLP in response to 20 and 40 N anterior and posterior loads between CR and CS prostheses. Increases in posterior femoral translation correspond to increases in anterior tibial translation. No significant changes (p<.05) were observed.

CHAPTER 5. Conclusion & Future Work

The first aim of this research was to characterize the roles of the menisci and ACL in joint constraint. These soft tissue structures are frequently injured and these injuries predispose patients to osteoarthritis. The ACL was found to be responsible for constraining anterior translation and internal rotation near full extension. The medial meniscus also acts to constrain anterior tibial translation in early flexion but to a much smaller extent than the ACL. Bilateral meniscectomy resulted in significantly more external rotation from mid to deep flexion. Interestingly, the menisci affected joint constraint more so under low loads whereas the ACL appeared to govern the end range of motion at higher loads. This may explain why meniscal tears are commonly present with ACL tears because the loads that highly engage the ACL have already exceeded the constraint the menisci provide. The data presented in this research can better inform evaluations of reconstruction techniques and improves our understanding of the roles of each tissue. Additional work should be done to expand the data set to increase the power of these studies and ensure that these results are representative of a broad population.

Cruciate-retaining and cruciate-substituting TKR systems were also evaluated in this work on the basis of passive joint constraint and femoral rollback. Providing appropriate joint constraint is essential for patient satisfaction as it influences the perceived stability and functionality of the joint. Achieving femoral rollback is desirable because it leads to improved flexion ROM due to the increase in quadriceps moment arm that occurs as the tibia displaces anteriorly. The CS design tested in this study performed comparably to the CR system in Int-Ext and A-P constraint. However, patterns of femoral rollback greatly differed between designs. While the anterior lip on the CS insert prevented excessive posterior tibial translation, it did not apply a sufficient posterior force to retain femoral rollback. Future studies should investigate

whether other CS designs behave similarly, and whether changes to the design can be made to achieve rollback. Additionally, work should be done to investigate whether the same patterns of constraint and femoral rollback occur during activities of daily living.

References

1. Andriacchi, T.P., et al., *Rotational changes at the knee after ACL injury cause cartilage thinning.* Clinical Orthopaedics and Related Research, 2006(442): p. 39-44.
2. Zantop, T., et al., *The role of the anteromedial and posterolateral bundles of the anterior cruciate ligament in anterior tibial translation and internal rotation.* American Journal of Sports Medicine, 2007. **35**(2): p. 223-227.
3. Renstrom, P. and R.J. Johnson, *Anatomy and Biomechanics of the Menisci.* Clinics in Sports Medicine, 1990. **9**(3): p. 523-538.
4. Markolf, K.L., et al., *Combined knee loading states that generate high anterior cruciate ligament forces.* Journal of Orthopaedic Research, 1995. **13**(6): p. 930-935.
5. Jones, A.O., et al., *Medial meniscus posterior root attachment injury and degeneration: MRI findings.* Australas Radiol, 2006. **50**(4): p. 306-13.
6. Edd, S.N., et al., *Alterations in Knee Kinematics After Partial Medial Meniscectomy Are Activity Dependent.* American Journal of Sports Medicine, 2015. **43**(6): p. 1399-1407.
7. Bhattacharyya, T., et al., *The clinical importance of meniscal tears demonstrated by magnetic resonance imaging in osteoarthritis of the knee.* J Bone Joint Surg Am, 2003. **85-A**(1): p. 4-9.
8. Conditt, M.A., et al., *The PCL significantly affects the functional outcome of total knee arthroplasty.* Journal of Arthroplasty, 2004. **19**(7): p. 107-112.
9. Hirsch, H.S., P.A. Lotke, and L.D. Morrison, *The posterior cruciate ligament in total knee surgery. Save, sacrifice, or substitute?* Clin Orthop Relat Res, 1994(309): p. 64-8.
10. Ishii, Y., et al., *Sagittal laxity in vivo after total knee arthroplasty.* Arch Orthop Trauma Surg, 2005. **125**(4): p. 249-53.
11. Banks, S.A., G.D. Markovich, and W.A. Hodge, *In vivo kinematics of cruciate-retaining and -substituting knee arthroplasties.* J Arthroplasty, 1997. **12**(3): p. 297-304.
12. Dennis, D.A., et al., *A multicenter analysis of axial femorotibial rotation after total knee arthroplasty.* Clin Orthop Relat Res, 2004(428): p. 180-9.
13. Stiehl, J.B., et al., *The cruciate ligaments in total knee arthroplasty: a kinematic analysis of 2 total knee arthroplasties.* J Arthroplasty, 2000. **15**(5): p. 545-50.
14. Waslewski, G.L., B.M. Marson, and J.B. Benjamin, *Early, incapacitating instability of posterior cruciate ligament-retaining total knee arthroplasty.* J Arthroplasty, 1998. **13**(7): p. 763-7.
15. Lombardi, A.V., Jr. and K.R. Berend, *The role of implant constraint in revision TKA: striking the balance.* Orthopedics, 2006. **29**(9): p. 847-9.
16. Luger, E., S. Sathasivam, and P.S. Walker, *Inherent differences in the laxity and stability between the intact knee and total knee replacements.* Knee, 1997. **4**(1): p. 7-14.
17. Bicer, E.K., et al., *Current knowledge in the anatomy of the human anterior cruciate ligament.* Knee Surgery Sports Traumatology Arthroscopy, 2010. **18**(8): p. 1075-1084.
18. Giuliani, J.R., K.G. Kilcoyne, and J.P. Rue, *Anterior cruciate ligament anatomy: a review of the anteromedial and posterolateral bundles.* J Knee Surg, 2009. **22**(2): p. 148-54.
19. Butler, D.L., F.R. Noyes, and E.S. Grood, *Ligamentous Restraints to Anterior-Posterior Drawer in the Human Knee - Biomechanical Study.* Journal of Bone and Joint Surgery-American Volume, 1980. **62**(2): p. 259-270.
20. Musahl, V., et al., *The Effect of Medial Versus Lateral Meniscectomy on the Stability of the Anterior Cruciate Ligament-Deficient Knee.* American Journal of Sports Medicine, 2010. **38**(8): p. 1591-1597.
21. Bach, J.M., M.L. Hull, and H.A. Patterson, *Direct measurement of strain in the posterolateral bundle of the anterior cruciate ligament.* J Biomech, 1997. **30**(3): p. 281-3.

22. Yoo, Y.-S., et al., *Changes in ACL length at different knee flexion angles: an in vivo biomechanical study.* Knee Surgery, Sports Traumatology, Arthroscopy, 2010. **18**(3): p. 292-297.
23. Giffin, J.R., et al., *Effects of increasing tibial slope on the biomechanics of the knee.* Am J Sports Med, 2004. **32**(2): p. 376-82.
24. Ahn, J.H., et al., *Longitudinal Tear of the Medial Meniscus Posterior Horn in the Anterior Cruciate Ligament-Deficient Knee Significantly Influences Anterior Stability.* American Journal of Sports Medicine, 2011. **39**(10): p. 2187-2193.
25. Lorbach, O., et al., *The influence of the medial meniscus in different conditions on anterior tibial translation in the anterior cruciate deficient knee.* Int Orthop, 2015. **39**(4): p. 681-7.
26. Gollehon, D.L., P.A. Torzilli, and R.F. Warren, *The Role of the Posterolateral and Cruciate Ligaments in the Stability of the Human Knee - a Biomechanical Study.* Journal of Bone and Joint Surgery-American Volume, 1987. **69A**(2): p. 233-242.
27. Wroble, R.R., et al., *The Role of the Lateral Extraarticular Restraints in the Anterior Cruciate Ligament-Deficient Knee.* American Journal of Sports Medicine, 1993. **21**(2): p. 257-263.
28. Hosseini, A., et al., *Meniscus Injuries Alter the Kinematics of Knees With Anterior Cruciate Ligament Deficiency.* Orthop J Sports Med, 2014. **2**(8): p. 2325967114547346.
29. Hame, S.L., D.A. Oakes, and K.L. Markolf, *Injury to the anterior cruciate ligament during alpine skiing - A biomechanical analysis of tibial torque and knee flexion angle.* American Journal of Sports Medicine, 2002. **30**(4): p. 537-540.
30. Takeda, Y., et al., *Biomechanical Function of the Human Anterior Cruciate Ligament.* Arthroscopy, 1994. **10**(2): p. 140-147.
31. Barba, D.B., Lloyd; Chhabra, Anikar, *Anatomy and Biomechanics of the Posterior Cruciate Ligament and Posterolateral Corner.* Operative Techniques in Sports Medicine, 2015. **23**(4): p. 256-268.
32. Makris, C.A., et al., *Posterior cruciate ligament architecture: evaluation under microsurgical dissection.* Arthroscopy, 2000. **16**(6): p. 627-32.
33. Kennedy, J.C., et al., *Tension studies of human knee ligaments. Yield point, ultimate failure, and disruption of the cruciate and tibial collateral ligaments.* J Bone Joint Surg Am, 1976. **58**(3): p. 350-5.
34. Goyal, K., et al., *In vivo analysis of the isolated posterior cruciate ligament-deficient knee during functional activities.* Am J Sports Med, 2012. **40**(4): p. 777-85.
35. Li, G., et al., *Effect of the posterior cruciate ligament on posterior stability of the knee in high flexion.* Journal of Biomechanics, 2004. **37**(5): p. 779-783.
36. Logan, M., et al., *The effect of posterior cruciate ligament deficiency on knee kinematics.* Am J Sports Med, 2004. **32**(8): p. 1915-22.
37. Patel, D.V., et al., *Injuries to the posterolateral structures of the knee.* Knee, 1998. **5**(1): p. 21-31.
38. Kennedy, N.I., et al., *Kinematic analysis of the posterior cruciate ligament, part 1: the individual and collective function of the anterolateral and posteromedial bundles.* Am J Sports Med, 2013. **41**(12): p. 2828-38.
39. Grood, E.S., S.F. Stowers, and F.R. Noyes, *Limits of movement in the human knee. Effect of sectioning the posterior cruciate ligament and posterolateral structures.* J Bone Joint Surg Am, 1988. **70**(1): p. 88-97.
40. Englund, M., *The role of the meniscus in osteoarthritis genesis.* Rheum Dis Clin North Am, 2008. **34**(3): p. 573-9.
41. Fox, A.J., et al., *The human meniscus: a review of anatomy, function, injury, and advances in treatment.* Clin Anat, 2015. **28**(2): p. 269-87.
42. Chen, L., et al., *The effect of medial meniscal horn injury on knee stability.* Knee Surg Sports Traumatol Arthrosc, 2015. **23**(1): p. 126-31.

43. Thompson, W.O., et al., *Tibial meniscal dynamics using three-dimensional reconstruction of magnetic resonance images.* Am J Sports Med, 1991. **19**(3): p. 210-5; discussion 215-6.
44. Bedi, A., et al., *Dynamic Contact Mechanics of the Medial Meniscus as a Function of Radial Tear, Repair, and Partial Meniscectomy.* Journal of Bone and Joint Surgery-American Volume, 2010. **92A**(6): p. 1398-1408.
45. Rodkey, W.G., *Basic biology of the meniscus and response to injury.* Instr Course Lect, 2000. **49**: p. 189-93.
46. Allaire, R., et al., *Biomechanical consequences of a tear of the posterior root of the medial meniscus.* Journal of Bone and Joint Surgery-American Volume, 2008. **90A**(9): p. 1922-1931.
47. Messner, K. and J.Z. Gao, *The menisci of the knee joint. Anatomical and functional characteristics, and a rationale for clinical treatment.* Journal of Anatomy, 1998. **193**: p. 161-178.
48. Arno, S., et al., *The effect of arthroscopic partial medial meniscectomy on tibiofemoral stability.* Am J Sports Med, 2013. **41**(1): p. 73-9.
49. Yammine, K., *Effect of partial medial meniscectomy on anterior tibial translation in stable knees: a prospective controlled study on 32 patients.* BMC Sports Sci Med Rehabil, 2013. **5**(1): p. 17.
50. Spang, J.T., et al., *The effect of medial meniscectomy and meniscal allograft transplantation on knee and anterior cruciate ligament biomechanics.* Arthroscopy, 2010. **26**(2): p. 192-201.
51. Levy, I.M., P.A. Torzilli, and R.F. Warren, *The effect of medial meniscectomy on anterior-posterior motion of the knee.* J Bone Joint Surg Am, 1982. **64**(6): p. 883-8.
52. Allen, C.R., et al., *Importance of the medial meniscus in the anterior cruciate ligament-deficient knee.* Journal of Orthopaedic Research, 2000. **18**(1): p. 109-115.
53. Papageorgiou, C.D., et al., *The biomechanical interdependence between the anterior cruciate ligament replacement graft and the medial meniscus.* American Journal of Sports Medicine, 2001. **29**(2): p. 226-231.
54. Levy, I.M., et al., *The effect of lateral meniscectomy on motion of the knee.* J Bone Joint Surg Am, 1989. **71**(3): p. 401-6.
55. Shoemaker, S.C. and K.L. Markolf, *The role of the meniscus in the anterior-posterior stability of the loaded anterior cruciate-deficient knee. Effects of partial versus total excision.* J Bone Joint Surg Am, 1986. **68**(1): p. 71-9.
56. Netravali, N.A., N.J. Giori, and T.P. Andriacchi, *Partial medial meniscectomy and rotational differences at the knee during walking.* Journal of Biomechanics, 2010. **43**(15): p. 2948-2953.
57. Magyar, O.M., et al., *Effect of medial meniscectomy on gait parameters.* Knee Surgery Sports Traumatology Arthroscopy, 2008. **16**(4): p. 427-433.
58. Church, S. and J.F. Keating, *Reconstruction of the anterior cruciate ligament: timing of surgery and the incidence of meniscal tears and degenerative change.* J Bone Joint Surg Br, 2005. **87**(12): p. 1639-42.
59. Whiteside, L.A., *Choosing your implant: cementless, patella sparing, and posterior cruciate ligament retaining.* J Arthroplasty, 2005. **20**(4 Suppl 2): p. 10-1.
60. Laskin, R.S., *Choosing your implant: cemented, tricompartmental, and posterior stabilized.* J Arthroplasty, 2005. **20**(4 Suppl 2): p. 7-9.
61. Seon, J.K., et al., *Comparisons of kinematics and range of motion in high-flexion total knee arthroplasty: cruciate retaining vs. substituting designs.* Knee Surg Sports Traumatol Arthrosc, 2011. **19**(12): p. 2016-22.
62. Sharma, A., et al., *High-flexion TKA designs: what are their in vivo contact mechanics?* Clin Orthop Relat Res, 2007. **464**: p. 117-26.
63. Parsley, B.S., et al., *Posterior cruciate ligament substitution is not essential for excellent postoperative outcomes in total knee arthroplasty.* Journal of Arthroplasty, 2006. **21**(6): p. 127-131.

64. Tanzer, M., K. Smith, and S. Burnett, *Posterior-stabilized versus cruciate-retaining total knee arthroplasty - Balancing the gap.* Journal of Arthroplasty, 2002. **17**(7): p. 813-819.
65. Hofmann, A.A., et al., *Posterior stabilization in total knee arthroplasty with use of an ultracongruent polyethylene insert.* Journal of Arthroplasty, 2000. **15**(5): p. 576-583.
66. Sathappan, S.S., et al., *Midterm results of primary total knee arthroplasty using a dished polyethylene insert with a recessed or resected posterior cruciate ligament.* J Arthroplasty, 2006. **21**(7): p. 1012-6.
67. Worland, R.L., D.E. Jessup, and J. Johnson, *Posterior cruciate recession in total knee arthroplasty.* J Arthroplasty, 1997. **12**(1): p. 70-3.
68. Ritter, M.A., P.M. Faris, and E.M. Keating, *Posterior cruciate ligament balancing during total knee arthroplasty.* J Arthroplasty, 1988. **3**(4): p. 323-6.
69. Andriacchi, T.P., J.O. Galante, and R.W. Fermier, *The influence of total knee-replacement design on walking and stair-climbing.* J Bone Joint Surg Am, 1982. **64**(9): p. 1328-35.
70. Simmons, S., et al., *Proprioception following total knee arthroplasty with and without the posterior cruciate ligament.* J Arthroplasty, 1996. **11**(7): p. 763-8.
71. Andriacchi, T.P. and J.O. Galante, *Retention of the posterior cruciate in total knee arthroplasty.* J Arthroplasty, 1988. **3 Suppl**: p. S13-9.
72. Yoshiya, S., et al., *In vivo kinematic comparison of posterior cruciate-retaining and posterior stabilized total knee arthroplasties under passive and weight-bearing conditions.* Journal of Arthroplasty, 2005. **20**(6): p. 777-783.
73. Heesterbeek, P.J.C., et al., *A new spacer-guided, PCL balancing technique for cruciate-retaining total knee replacement.* Knee Surgery Sports Traumatology Arthroscopy, 2014. **22**(3): p. 650-659.
74. Pagnano, M.T., et al., *Flexion instability after primary posterior cruciate retaining total knee arthroplasty.* Clinical Orthopaedics and Related Research, 1998(356): p. 39-46.
75. Clary, C.W., et al., *The influence of total knee arthroplasty geometry on mid-flexion stability: An experimental and finite element study.* Journal of Biomechanics, 2013. **46**(7): p. 1351-1357.
76. Kelman, G.J., et al., *Gait Laboratory Analysis of a Posterior Cruciate-Sparing Total Knee Arthroplasty in Stair Ascent and Descent.* Clinical Orthopaedics and Related Research, 1989(248): p. 21-26.
77. Peters, C.L., et al., *Comparison of total knee arthroplasty with highly congruent anterior-stabilized bearings versus a cruciate-retaining design.* Clin Orthop Relat Res, 2014. **472**(1): p. 175-80.
78. Hozack, W.J., et al., *The Patellar Clunk Syndrome - a Complication of Posterior Stabilized Total Knee Arthroplasty.* Clinical Orthopaedics and Related Research, 1989(241): p. 203-208.
79. Watanabe, T., et al., *Knee kinematics in anterior cruciate ligament-substituting arthroplasty with or without the posterior cruciate ligament.* J Arthroplasty, 2013. **28**(4): p. 548-52.
80. Massin, P., P. Boyer, and M. Sabourin, *Less femorotibial rotation and AP translation in deep-dished total knee arthroplasty. An intraoperative kinematic study using navigation.* Knee Surg Sports Traumatol Arthrosc, 2012. **20**(9): p. 1714-9.
81. Freutel, M., et al., *Influence of partial meniscectomy on attachment forces, superficial strain and contact mechanics in porcine knee joints.* Knee Surg Sports Traumatol Arthrosc, 2015. **23**(1): p. 74-82.
82. Higuchi, H., et al., *Factors affecting long-term results after arthroscopic partial meniscectomy.* Clin Orthop Relat Res, 2000(377): p. 161-8.
83. Petty, C.A. and J.H. Lubowitz, *Does arthroscopic partial meniscectomy result in knee osteoarthritis? A systematic review with a minimum of 8 years' follow-up.* Arthroscopy, 2011. **27**(3): p. 419-24.

84. Sanders, T.L., et al., *Incidence of Anterior Cruciate Ligament Tears and Reconstruction: A 21-Year Population-Based Study.* Am J Sports Med, 2016. **44**(6): p. 1502-7.
85. Nicolini, A.P., et al., *Common injuries in athletes' knee: experience of a specialized center.* Acta Ortop Bras, 2014. **22**(3): p. 127-31.
86. Ferretti, A., E. Monaco, and A. Vadala, *Rotatory instability of the knee after ACL tear and reconstruction.* J Orthop Traumatol, 2014. **15**(2): p. 75-9.
87. Cyr, A.J. and L.P. Maletsky, *Technical note: a multi-dimensional description of knee laxity using radial basis functions.* Computer Methods in Biomechanics and Biomedical Engineering, 2015. **18**(15): p. 1674-1679.
88. Grood, E.S. and W.J. Suntay, *A Joint Coordinate System for the Clinical Description of 3-Dimensional Motions - Application to the Knee.* Journal of Biomechanical Engineering-Transactions of the Asme, 1983. **105**(2): p. 136-144.
89. Arnold, E.M., et al., *A model of the lower limb for analysis of human movement.* Ann Biomed Eng, 2010. **38**(2): p. 269-79.
90. Cyr, A.J. and L.P. Maletsky, *Unified quantification of variation in passive knee joint constraint.* Proceedings of the Institution of Mechanical Engineers Part H-Journal of Engineering in Medicine, 2014. **228**(5): p. 494-500.
91. Kim, Y.H., J.S. Kim, and S.H. Yoon, *A recession of posterior cruciate ligament in posterior cruciate-retaining total knee arthrosplasty.* J Arthroplasty, 2008. **23**(7): p. 999-1004.