

Investigating the Role of the Femoral Tunnel Position on Knee Biomechanics in ACL Reconstruction

by

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A thesis

presented to the University of Waterloo

in fulfillment of the

thesis requirement for the degree of

Master of Applied Science

in

Mechanical and Mechatronics Engineering

Waterloo, Ontario, Canada, 2025

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Author's Declaration

This thesis consists of material all of which I authored or co-authored: see Statement of Contributions included in the thesis. This is a true copy of the thesis, including any required final revisions, as accepted by my examiners.

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Statement of Contributions

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CHAPTER 2	<i>Sole Author – Teresa Marotta</i>
CHAPTER 3	<i>Study Design – Teresa Marotta, Stewart McLachlin, Ryan Willing, Ryan Martin, Tim Lasswell</i> <i>Specimen Preparation – Christopher Hewison, Matthew Free, Alexandra Rocha, Marc Boyce, Madalyn Richards</i> <i>Data Collection – Teresa Marotta, Marc Boyce, Madalyn Richards</i> <i>Data Analysis – Teresa Marotta</i> <i>Statistical Analysis – Teresa Marotta</i> <i>Data Visualization – Teresa Marotta, Stewart McLachlin, Ryan Willing, Ryan Martin, Tim Lasswell</i> <i>Discussion – Teresa Marotta</i>
CHAPTER 4	<i>System Design & Ideation – Teresa Marotta, Ryan Martin, Tim Lasswell, Stewart McLachlin</i>
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APPENDIX C	<i>Sole Author – Teresa Marotta</i>

Abstract

The femoral tunnel position for the graft plays a critical role in the success of anterior cruciate ligament (ACL) reconstruction surgery. Malposition of the femoral tunnel is one of the top reasons for graft failure in this procedure. Despite the potential benefits of surgical navigation systems for ACL reconstruction to aid in more accurate femoral tunnel positioning, there has been no significant adoption or evidence of improved clinical outcomes with navigated ACL reconstruction procedures. However, this may stem from a lack of understanding of how the femoral tunnel position in ACL reconstruction alters the biomechanics of the knee joint relative to the native knee. To address this knowledge gap, the goals of this work were to (1) investigate the role of the femoral tunnel position in its influence on knee biomechanics, and (2) design and develop a proof-of-concept surgical navigation system to aid in selection and placement of the femoral tunnel. For the first goal, an *in vitro* study investigated the influence of five different femoral tunnel positions on knee kinematics, force carried by the graft, and on graft length change over flexion. Varying femoral tunnel insertion led to statistically significant kinematic differences in anterior tibial translation range at 90 degrees of knee flexion. As well, different femoral tunnel positions led to significant differences in the force carried by the graft compared to the estimated force carried by the intact ACL at all knee flexion angles. Graft length change over knee range of motion showed that an anteriorly placed femoral tunnel is more isometric while a posteriorly placed graft had the greatest change in length. For the second goal of this work, a surgical navigation system workflow was developed based on these insights to allow for the calculation of graft length changes intraoperatively over different knee flexion angles. A proof-of-concept system with this feature was developed in 3D Slicer using the SlicerIGT extension. This work enhances understanding of how the femoral tunnel position affects knee biomechanics and how surgical navigation systems can be designed to provide surgeons with methods to better identify and position the femoral tunnel for ACL reconstruction surgery.

Acknowledgements

Firstly, I would like to express my gratitude to my supervisor, Dr. Stewart McLachlin. Thank you for bringing so many ideas to this work. There are a thousand other research ideas we discussed at length for this work, and I think narrowing all those ideas down that really helped us create an impactful body of work. Additionally, thank you for always being willing to spend time discussing all the little details of this project. We spent many hours looking at 3D Slicer, Excel, and Matlab to figure out the best way to present everything and I'm very grateful for that time.

I would also like to thank Dr. Naveen Chandrashekar and Dr. Stacey Acker for their contributions as readers of this thesis. I appreciate the time and effort you both put into reviewing this work. Thank you for helping me improve the quality and clarity of the final document.

I would also like to thank Dr. Ryan Martin and Dr. Tim Lasswell for their support and contributions to the experimental work and surgical navigation system development. Thank you both for your contributions and ideas. You both spent many early hours in Calgary to discuss our experimental protocol, results, and system ideas. The clinical expertise that you both brought to this project elevated the way this research was able to be conducted, presented, and discussed.

I would also like to express my gratitude to Dr. Ryan Willing. The entire experimental work for this thesis was conducted in his lab and with the help of his students. Dr. Willing was instrumental in allowing this study to succeed through his efforts in developing the user interface that allowed us to implement the virtual ligaments for this study and his insights into the experimental and analytical protocols. I would also like to express my appreciation to Dr. Christopher Hewison, Dr. Matthew Free, and Dr. Alexandra Rocha who were all involved in preparing the specimens for the experimental study. Additionally, thank you to all the members of the Biomechanical Engineering Research Lab who were involved in this study or helped with the work conducted in this study, especially Marc Boyce, Madalyn Richards, and Emma Donnelly. Thank you to Marc and Madalyn for staying for many late nights to help me conduct my testing for this work. This study couldn't have been completed without you.

I would also like to thank all my colleagues in the OrthoTron lab, especially Anandita Gautam, Claire Thompson, Jonah Leinwand, Martine McGregor, Richard Barina, and Sarah Hynes. Thank you all for the support you provided when I was conducting this work and for all your help with all my projects. I also want to thank the undergraduate students who helped support my work, especially Marie DeMey.

You were involved with this research since day one and I'm very grateful for all the time and enthusiasm you brought to this work.

Additionally, I was very fortunate to be involved in many other projects with other labs during my degree that did not make it into this thesis. I would like to thank Dr. Yue Hu and the members of the AIRLab for their support with the KUKA robotics project, especially Jeffrey Lee, Evie Bouganim, and Noam Borenstein. I also wanted to thank Dr. Mihaela Vlasea and the members of the MSAM lab for their support with the auxetic projects, especially Sagar Patel.

Finally, I want to acknowledge my family and friends for their support during my study. I am grateful for their encouragement and interest in my work. To my parents, thank you for being willing to listen to me talk endlessly about my work, your patience, and for letting me take the car to London – often on short notice. To Sebastian, thank you for unintentionally inspiring my work by having ACL surgery not once, but twice during my degree.

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List of Abbreviations

ACL	Anterior Cruciate Ligament
AM	Anteromedial
ATT	Anterior Tibial Translation
BEAR	Bridge-Enhanced ACL Repair
CT	Computed Tomography
DIS	Dynamic Intraligamentary Stabilization
EMM	Estimated Marginal Means
IBLA	Internal Brace Ligament Augmentation
IR	Internal Rotation
IT	Iliotibial
LCL	Lateral Collateral Ligament
LET	Lateral Extra-articular Tenodesis
LMM	Linear Mixed Model
MCL	Medial Collateral Ligament
MRI	Magnetic Resonance Imaging
PCL	Posterior Cruciate Ligament
PL	Posterolateral
SARA	Symmetrical Axis of Rotation Approach

List of Symbols

α	Flexion angle in Grood and Suntay Convention
β	Adduction in Grood and Suntay Convention
e_1	Body Fixed Axis on the Femur in Grood and Suntay Convention
e_1^r	Reference Line on the Femur in Grood and Suntay Convention
e_3	Body Fixed Axis on the tibia in Grood and Suntay Convention
e_3^r	Reference Line on the tibia in Grood and Suntay Convention
e_2	Floating Axis in Grood and Suntay Convention
ε	Strain
ε_r	Reference Stain
ε_1	Spring Parameter
k	Stiffness Parameter
l	Ligament Length
l_r	Reference Length
l_0	Zero Load Length (Slack Length)
mm	Millimeter
N	Newton
Nm	Newton Meters
P_A	Reference Point on the Femur in Grood and Suntay Convention
P_B	Reference Point on the Tibia in Grood and Suntay Convention
γ	External Rotation in Grood and Suntay Convention

Chapter 1

Introduction

1.1 Motivation

Anterior cruciate ligament (ACL) injuries are extremely prevalent with over 120 000 injuries occurring annually in the United States [1], [2]. These injuries are typically caused by non-contact mechanisms, such as poor knee positioning during landing after jumps and abruptly stopping or pivoting [1]. ACL injuries have both serious short and long-term repercussions [1]. In the short term, patients will experience functional instability when walking as well as pain [3]. In the long-term, patients with ACL tears have a higher risk of developing knee osteoarthritis [4]. To treat these injuries, ACL reconstruction surgery is a procedure where the ruptured ACL is replaced with a graft [5]. The aim of this procedure is to restore native joint kinematics, while preventing graft failure [5]. Despite the common occurrence of the procedure, ACL graft failure is a key concern as many patients (3-14%) will experience instability post-surgery [3]. Aside from trauma, femoral tunnel malposition has been shown to be the primary technical cause of failure for ACL reconstruction [3], [6], [7]. In particular, small differences in the location of tunnel placement can cause large differences in the amount of strain that the graft ACL will experience during the physiologic range of motion [8].

To improve femoral tunnel positioning, surgical navigation has been proposed as a means to achieve greater precision during tunnel placement [9]. Surgical navigation systems can allow surgeons to identify idealized femoral tunnel locations, confirm the accurate drilling of the selected tunnel location, and prevent graft impingement [9]. For these reasons, it has been widely believed that navigated ACL reconstruction would lead to better clinical outcomes. However, navigated ACL reconstruction has not yet demonstrated significantly improved clinical outcomes compared to traditional ACL reconstruction [10]. As such, there is a need for further research into the resulting kinematic differences that minor deviations in femoral tunnel positioning creates.

The biomechanical influence of femoral tunnel positioning has previously been investigated in literature [11]–[14]. Within this work, the cadaveric *in vitro* experimental testing typically focuses on capturing the kinematics of the intact knee, cutting the ACL and capturing the kinematics of the ACL-deficient knee, and performing an ACL reconstruction and capturing the kinematics of the knee with the reconstruction [11]–[14]. With the advancements in virtual soft tissue modelling in joint motion simulator controllers, previous literature has shown that specimen-specific virtual ACL grafts can be

implemented to achieve a similar effect of real ACL grafts during *in vitro* biomechanical testing [15]. However, no group has yet examined how virtual ligaments can be used to examine the role of femoral tunnel location variation.

To address this knowledge gap, this thesis sought to investigate the role of the ACL repair femoral tunnel location on knee kinematics and graft strain profiles using *in vitro* experimental testing. By leveraging specimen-specific virtual ligaments in place of real ACL grafts, human cadaveric ACL-deficient knee joints can be stabilized for various femoral tunnel and graft characteristics. The differences in kinematic behaviour between the different femoral tunnel locations can be captured experimentally and analyzed across specimens to determine the effect, if any, of the femoral tunnel location. By creating minor adjustments in the positioning of the femoral tunnel experimentally, a better understanding of the biomechanical effect of the femoral tunnel position can be determined and leveraged to determine how to better intraoperatively record knee biomechanics and to better inform surgeons of how to select the optimal femoral tunnel location. Additionally, this thesis work also proposes an ACL reconstruction surgical navigation system which aims to aid in the selection and positioning of the femoral tunnel. Overall, these methodologies and system development aim to improve knee ligament biomechanics comprehension and enhance femoral tunnel selection to better stabilize the knee post-ACL reconstruction with a goal of improving long-term outcomes for patients.

1.2 Thesis Objectives

The overall goal of this thesis was to investigate the biomechanical role of the femoral tunnel location for ACL reconstruction. Within this context, the following specific objectives were identified:

1. to quantify the influence of femoral tunnel location on knee biomechanics when a constant tibial tunnel location is used; and,
2. to develop an ACL reconstruction surgical navigation workflow to improve the identification and positioning of an optimal femoral tunnel location.

1.3 Thesis Structure

Chapter 1: *Introduction.*

Chapter 2: *Literature Review:* an overview of relevant background and the current state of biomechanical research for ACL reconstruction and surgical navigation.

Chapter 3: *Comparison of 5 Femoral Tunnel Locations in Single-Bundle Virtual Anterior Cruciate Ligament Reconstructions: An In Vitro Biomechanical Study:* investigating five specimen-specific femoral tunnel positions with the same tibial tunnel location. It was hypothesized that the different femoral tunnel positions would create significant differences in kinematics, force carried by the ligament, and graft length change over flexion.

Chapter 4: *Design & Development of a Proof-of-Concept Surgical Navigation System for ACL Reconstruction:* details a clinically informed surgically navigated ACL reconstruction workflow and presents the design of a proof-of-concept surgical navigation system.

Chapter 5: *Thesis Conclusions.*

Chapter 2

Literature Review

2.1 Anatomy of the Knee

2.1.1 Bony Anatomy

The bony anatomy of the knee includes three components: the distal end of the femur, the proximal end of the tibia, and the patella [16], [17]. Figure 1 depicts this anatomy.

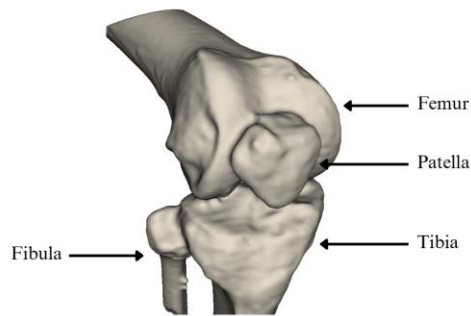


Figure 1: Boney anatomy of the right knee.

Knee anatomy depicted where the femur, patella, tibia and fibula are highlighted.

2.1.1.1 Femur

A key part of the knee joint, the femur is the longest, heaviest, and the strongest bone in the human body [18]. The primary function of the femur is weight bearing and for gait stability [18]. In particular, when a person is standing, the femur transmits weight from the pelvis to the tibia [19]. The femur consists of a shaft and two ends [19]. The proximal end of the femur includes a spherical head which connects to the cylindrical shaft [18]. This spherical head portion of the femur is part of the hip joint [18]. The distal end of the femur is the portion of the femur that makes up part of the knee joint [16], [18]. In this portion, the femur has a medial and lateral condyle, these condyles match the shape of the corresponding medial and lateral tibial plateaus [16]. The corresponding shape is a key component of what allows for the smooth tibial movement on the femur [16]. On the lateral condyle is the lateral epicondyle where the lateral collateral ligament (LCL) inserts on the femur [20]. On the medial condyle is the medial epicondyle, where the medial collateral ligament (MCL) inserts on the femur [20]. The two condyles are separated by the intercondylar notch [20]. The intercondylar notch is the insertion site

of both the ACL and the posterior cruciate ligament (PCL) on the femur [20]. Covering the femoral condyles is cartilage, which helps to reduce friction while the knee is moving [18], [19].

2.1.1.2 Tibia

The tibia is known as the shin bone and is the second largest bone in the body [19]. The tibia is key for weight bearing as weight is transferred from the knee joint to the ankle by the tibia [19]. The tibia consists of a shaft and two ends [19]. The proximal end of the tibia makes up part of the knee joint [19]. Specifically, this portion of the tibia contains two flattened surfaces which are called the tibial plateaus or condyles [20]. The medial tibial plateau is the larger of the two plateaus and presents as a concave surface [20]. The lateral tibial plateau is more narrow than the medial plateau and is more convex [20]. In between the tibial plateaus are the intercondylar eminences [20]. Anterior and posterior to these eminences are the anterior and posterior intercondylar areas where the ACL and the PCL insert on the tibia [19], [20].

2.1.1.3 Patella

The patella is the largest sesamoid bone in the human body [21], [22]. It is shaped like an upside-down triangle and rests on the patellar groove of the femur [21]. The patella has a thin layer of cortical bone and is made of a trabecular core [22]. The posterior surface of the patella is often divided into many facets [22]. There is a clear middle line of the patella, which divides it into medial and lateral halves [22]. This can then be divided further into seven facets [21], [22]. The many facets of the patella allow it to rest in the concave distal surface of the femur and the wider lateral side of the femur helps keep the patella properly aligned [22]. In between the anterior side of the femur and the posterior side of the patella is a thick layer of articular cartilage which dissipates large forces [22]. The average patella has a length of 4-4.5 cm, a width of 5-5.5 cm, and is 2-2.5 cm thick [22]. The Wilberg classification system is used to classify patellar facet sizes [21]. Type I has both concave medial and lateral facets approximately equal in size (nearly symmetrical), Type II has a flat to slightly convex medial facet that is smaller than the lateral facet, and Type III has a much smaller medial facet with a convex form [21], [23].

The patella has a variety of key functions. Firstly, both the quadriceps tendon and the patellar ligament are attached to the patella [23]. The quadriceps muscles converge to the quadriceps tendon which then attach to the patella [23]. The patellar ligament connects the patella to the tibia [23]. The

patella also acts as a lever arm for the quadriceps muscle, allowing it to help with the extension of the knee [21]–[23]. The patella also minimizes the contact between the quadriceps tendon and the femur, which reduces the frictional forces on the quadriceps tendon [23]. Finally, the patella acts as a shield for the rest of the structures in the knee [23].

2.1.2 Knee Joint

The knee is most often defined as a complex modified hinge joint [24]. It is typically viewed as composing of two separate joints: the tibiofemoral joint and the patellofemoral joint [16], [24]. The tibiofemoral joint allows for the transfer of weight from the femur to the tibia [25]. Specifically, the tibiofemoral joint allows for rotation along the sagittal plane (flexion and extension) [25]. The patellofemoral joint is composed of the posterior surface of the patella and the patellar groove of the femur (surface of the distal anterior femur) [21], [22]. The patella acts as a lever arm for the quadriceps, where it's contribution becomes greater as the knee increases in extension [21], [22]. The combination of these two joints allows the knee to have a six degree of freedom range of motion. Figure 2 depicts these ranges of motion which include: flexion/extension, internal/external rotation, varus/valgus, anterior/posterior translation, medial/lateral translation, superior/inferior translation [24]. It's largest range of motion is about the sagittal plane (flexion and extension), but the knee joint also has motion about the frontal plane (varus and valgus rotation) [24].

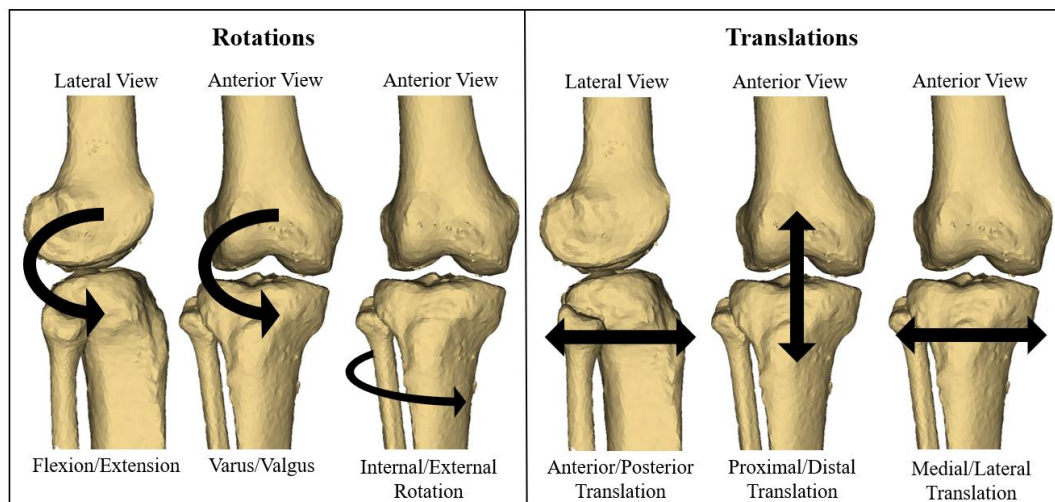


Figure 2: Motions of the knee joint.

The knee joint is depicted in six images where the three clinical rotations (flexion/extension, varus/valgus, internal/external rotation) and three clinical translations (proximal/distal, anterior/posterior, medial/lateral) are labelled.

2.1.3 Ligamentous Anatomy

The knee ligaments are split into two categories: intra-articular ligaments, extra-articular ligaments [26]. The principle intra-articular ligaments are the ACL and the PCL [26]. Key extra-articular ligaments include the MCL and the LCL [26]. An overview of the ligamentous anatomy is presented in Figure 3.

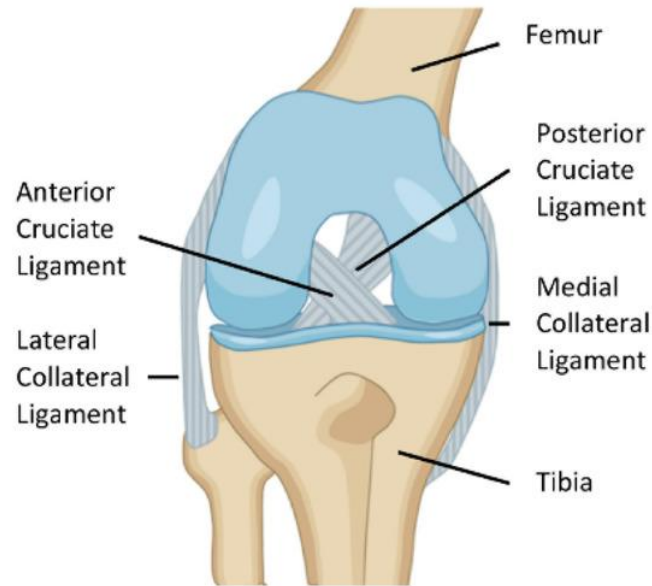


Figure 3: Ligamentous knee anatomy.

Key bony anatomy, including the tibia and femur, is highlighted along with key ligaments including the anterior cruciate ligament (ACL), posterior cruciate ligament (PCL), medial collateral ligament (MCL), and the lateral collateral ligament (LCL). Adapted from [27].

2.1.3.1 Anterior Cruciate Ligament

The ACL is a ligament that connects the posteromedial portion of the lateral femoral condyle to the anterior portion of the tibial plateau [28]. The ligament is primarily made up of type I collagen, with a small portion of type III collagen [28]. As such, the ACL can stretch and return to its original length without sustaining damage under many loading conditions [28]. The ACL acts as a stabilizer in anterior tibial translation and during large tibial rotations [28]. Additionally, it contains two separate bundles: the anteromedial (AM) bundle and the posterolateral (PL) bundle [28]. The footprints of these two bundles on the femur are shown in Figure 4. The average intra-articular length of the ACL is 33mm and the average width is around 11mm [28].

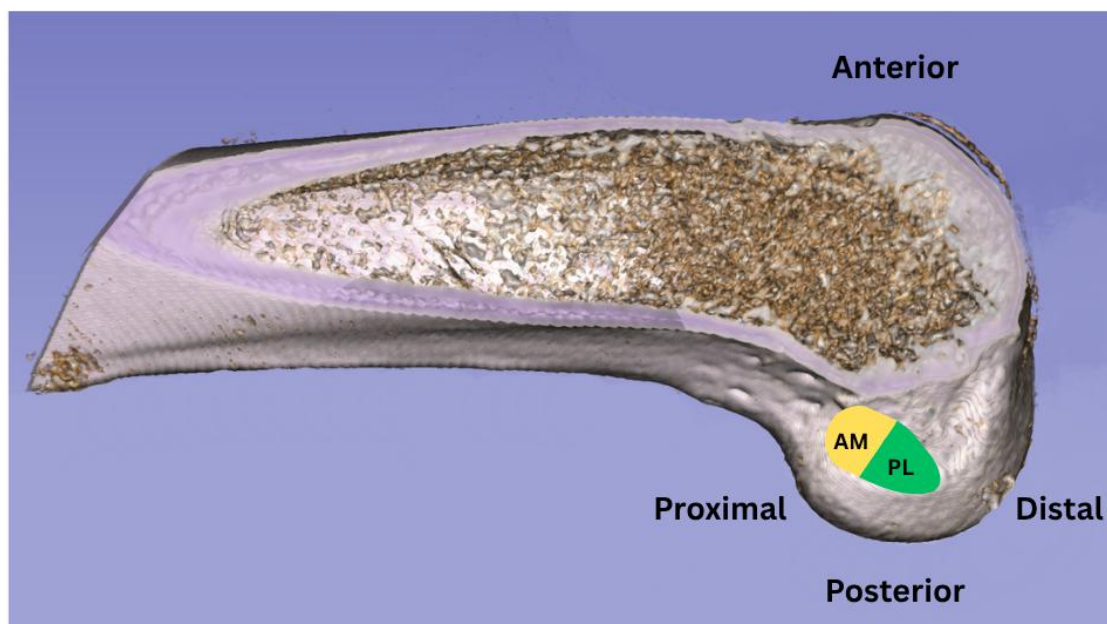


Figure 4: Footprints of the bundles of the anterior cruciate ligament (ACL).

Footprints of the anteromedial (AM) bundle and posterolateral (PL) bundles of the ACL on a sagittal cross-section of the femur.

2.1.3.2 Posterior Cruciate Ligament

The PCL is a ligament that connects the lateral portion of the medial femoral condyle to the posterior portion of the intra-articular tibia [26]. The PCL is a primary stabilizer of the knee and prevents excessive rotation between 90 degrees and 120 degrees of flexion [29]. The PCL has two separate bundles: the anterolateral bundle and the posteromedial bundle [26], [29]. These two bundles work together where in knee flexion the anterolateral bundle is tight and the posteromedial bundle is lax and in knee extension the anterolateral bundle is lax and the posteromedial bundle is tight [26], [29]. The PCL is around 32 to 38 mm in length and has a thickness that is almost twice of the ACL [26], [29]. Due to its thickness, it is harder to tear the PCL than the ACL [26].

2.1.3.3 Medial Collateral Ligament

The MCL is a ligament that connects the medial femoral epicondyle and the medial condyle of the tibia [26]. The MCL is a stabilizer for the medial side of the knee joint and provides support during valgus stress, rotational forces, and anterior translation forces on the tibia [30]. The MCL is around 8 to 10 cm in length [30]. The MCL is a complex layered ligament where it can be separated into a superficial and deep component [26], [30]. The superficial component (tibial collateral ligament) connects the posterior

portion of the medial femoral epicondyle and the medial condyle of the tibia [30]. The deep component (mid-third capsular ligament) is further divided into the meniscomfemoral and meniscotibial components [26], [30]

2.1.3.4 Lateral Collateral Ligament

The LCL is a ligament that connects the lateral epicondyle of the femur to the lateral side of the head of the fibula [31]. The LCL is a stabiliser for the lateral side of the knee joint and provides support during varus stress, external rotation, and posterior translational forces on the tibia [31]. Specifically, during full knee extension, the LCL prevents varus instability and in knee flexion, the LCL is more slack which permits knee rotation [32]. The LCL is around 69.9 mm long and is 2.6 mm thick [31].

2.1.4 Musculature

The muscles that surround the knee joint allow for the production of knee motion in all six degrees of freedom [24]. Additionally, the muscles act as secondary stabilisers for the knee joint [24]. The anterior part of the knee contains the quadriceps muscle group which aid in generating knee extension [24]. This group is comprised of the rectus femoris, vastus lateralis, vastus medialis, and vastus intermedius [24]. The posterior part of the knee contains the hamstring muscle group which control the flexion of the knee [24]. This group is comprised of the biceps femoris, semimembranosus, and semitendinosus [24]. The medial portion of the knee contains the sartorius and gracilize muscles [24]. The medial musculature helps control flexion of the knee and allow for medial rotation [24]. The lateral portion of the knee contains the iliotibial band and the popliteus muscles [24]. These muscles, along with the semimembranosus and semitendinosus of the hamstring muscle group, help control knee flexion and act as hip extensors [24]. The lateral stability of the knee is supported by the iliotibial band while the popliteus muscle controls the external and internal rotation of the knee [24].

2.2 Knee Kinematics

The knee joint has six degrees-of-freedom with the three rotations described as flexion-extension, internal-external rotation, and varus-valgus [24]. Knee flexion-extension motion occurs about the sagittal plane around a medial-lateral axis [24], [33]. Average range of flexion in a normal knee is around 130-140 degrees [33]. Full knee extension is considered 0 degrees of flexion (straight leg), but hyperextension is possible in some individuals and often in children (5 degrees to 10 degrees) [33]. Internal-external rotation occurs about the transverse plane around a vertical axis [33]. When the knee

is flexed (between 30 degrees to 90 degrees), the knee can externally rotate about 45 degrees and internally rotate about 25 degrees [34]. Additionally, as the knee extends, the range of motion for internal-external rotation decreases [34]. Knee varus valgus occurs about the frontal plane [24]. Varus-valgus motion is very minimal in unloaded conditions [35], [36]. When the knee is externally loaded, varus-valgus motion becomes pronounced [35], [36].

2.2.1 Kinematic Analysis of the Knee Joint

To calculate the clinical translations (lateral, anterior, joint distraction) and rotations (flexion, adduction, external rotation), Grood and Suntay's (1983) joint coordinate system is often used in research [37]. In their paper, Grood and Suntay (1983) proposed a three axes system where one axis is fixed to the femur (base vector e_1), one axis is fixed to the tibia (base vector e_3), and the third axis is a floating axis which is perpendicular to both (base vector e_2) [37]. The body fixed axis on the femur passes proximally through the center of the femoral head and passes through the most distal point on the posterior surface of the femur midway between the medial and lateral condyles [37]. The body fixed axis on the tibia passes midway between the two intercondylar eminences proximally and through the center of the ankle distally [37]. There are three relative rotations (α, γ, β) where α (flexion) is the angle between the floating axis and the reference line in femur (e_1^r), γ (external rotation) is the angle between the floating axis and the reference line in the tibia (e_3^r), and β (adduction) is the angle between the two fixed axes [37]. The translations are described using reference points (P_A in the femur and P_B in the tibia) where their relative positions are characterized by vector H , connecting the points [37].

Another common way of analyzing knee joint kinematics is to determine the location of the flexion extension axis (axis of rotation). There are a variety of different methods that were developed to calculate the knee flexion extension axis. In particular, there are two main categories of methods: functional methods and predictive techniques [38], [39]. Functional methods leverage the relative motion between body segments for an identified joint, while predictive techniques leverage regression equations and the specific position of tracked markers used to located boney landmarks [38], [39]. Some common functional methods include: the helical axes methods (finite helical axis technique and instantaneous helical axis technique) and the symmetrical axis of rotation approach (SARA) [38], [40]. The SARA method proposed by Ehrig *et al.* (2007) is a popular method to calculate a joint's axis of rotation as it allows for the rotational movement of two segments to be considered independently, meaning both segments can move [39], [41]. For this method, two tracked markers are used, one on the

first segment and one on the second segment [41]. The transformation matrices of each marker is then determined with respect to the global coordinate system for each time point during the movement of the segments [41]. The SARA method relies on the singular value decomposition of a linear least squares problem which determines a unique representation of the joint axis in the local coordinate system of each segment [41].

2.3 ACL Tears & Clinical Diagnosis

Approximately 1-10 in 1000 people tear their ACL per year [1]. This injury occurs frequently when there is a large anterior or valgus stress on the extended knee [1]. While this is often considered a sports-associated injury, 80% of ACL injuries are caused by noncontact mechanisms (not near other players) where there is no direct blow to the knee [1]. The four sports that have the highest rate of ACL injuries are alpine skiing, soccer, baseball, and football [1]. Some common movements that can potentially lead to ACL injury include: cutting, pivoting, abruptly stopping, and landing with the knee in extension [1], [42]. When these movements occur, they generate excessive tension force on the ACL which causes it to rupture [42]. More specifically, these movements can create a large anterior shear force at the proximal end of the tibia, the primary contributor to ACL loading [43]. Knee valgus, varus, and internal moments can also contribute to ACL loading, but only when the large anterior shear force is already present at the proximal end of the tibia [43]. Finally, there is strong evidence that ACL injury is likely caused by multi-planar mechanisms, not solely as a result of mechanisms on the sagittal, frontal, or transverse plane [44].

When the ACL gets fully ruptured, historically it has been assumed that it does not heal [45], [46]. Newer research has shown that the ACL has limited healing capacity and could spontaneously heal under certain circumstances [46]–[49]. The primary beliefs behind the limited healing capabilities of the ACL are due to limited blood supply and the interference from synovial fluid [46], [48]. When the ACL ruptures fully, the blood vessels within the ACL are torn [48]. Typically, when blood vessels are torn, blood clots at the injury site, creating a hematoma which starts the wound healing process. Since the ACL is surrounded by synovial fluid, though, the blood dissipates and prevents hematoma formation [46], [48]. The two ends of the ruptured ACL are therefore not able to be bridged in a fully torn ACL and thus limits healing [46], [48]. Additionally, as previously discussed, ligaments connect bone to bone and have poor blood supply [50]. In particular, the ACL has limited vascular supply at its

distal stump [46]. This limited blood supply further complicates the healing process of the ligament [46].

Females have a higher risk of ACL injuries than males with some studies showing a 4 to 5 fold greater incidence in female athletes compared to male athletes playing the same sports [1], [51], [52]. While there are many proposed extrinsic and intrinsic risk factors that could be leading to the higher incidence of ACL tears in women, three key factors are: smaller intercondylar notch size, influence of the menstrual cycle on ligament properties, and higher inclination towards knee valgus during landing [1], [51]. Smaller intercondylar notch sizes are associated with smaller ACLs. A narrow notch could create greater elongation of the ACL under higher tension, which leads to a larger risk [51]. As more women have smaller intercondylar notch sizes, this is one factor that could be contributing to a larger incidence of ACL tears [51]. Estrogen and relaxin have been shown in the literature to affect the tensile properties of ligaments [51]. As such, some researchers believe that when estrogen is higher or lower across the menstrual cycle, it could lead to increased or decreased laxity of the ACL and possibly affect the risk of injury rate [51]. There are some studies in this space, but there is contradictory data depicting different effects of estrogen and other hormones on ACL laxity [51]. Finally, studies have shown that female athletes tend to land in a valgus position compared to male athletes who land in a knees apart position [1], [51].

2.3.1 Clinical Diagnosis of ACL Tear

To diagnose an ACL tear, typically clinicians will begin the diagnostic process by perform laxity tests to assess ACL integrity [53]. These laxity tests include the Lachman test, the anterior drawer test, and the pivot-shift test [53]. In the Lachman test, the clinician will place the knee at 15-30 degrees of flexion and pull the tibia forward to see how large the femoro-tibial displacement is between the injured and uninjured knee [53]. A patient with a ruptured ACL will have a greater femoro-tibial displacement than a patient with an intact ACL [53]. In the anterior drawer test, the clinician will position the patient in a supine position and then place patient's knee at 90 degrees [53]. The tibia will then be pulled forward to see how large the femoro-tibial displacement is [53]. Finally, in the pivot shift test, the clinician will position the patient in a supine position, extend the knee, rotate the tibia internally, and then pull forward on the tibia while slowing bringing the knee into extension [53], [54]. If an ACL tear is suspected after the clinician performs the clinical laxity tests, magnetic resonance imaging (MRI) will be used to confirm if the ACL has been torn and if any other injuries were identified [53].

2.4 ACL Reconstruction & Repair

ACL reconstruction is the primary surgical intervention for the treatment of ACL rupture [5]. In this procedure, the aim is to replace the ruptured ACL with a graft that should recreate the native knee biomechanics [5]. The procedure begins with a surgeon cutting portals (openings) around the knee. While the number of portals and their locations can vary depending on the surgeon, there is at least one working portal and one viewing portal [55], [56]. Working portals are the portals that surgeons perform surgical tasks through while viewing portals are used for the arthroscope, which allows surgeons to see into the knee joint space. In a three-portal technique, there is an accessory AM portal which is the working portal, the anterolateral portal which is the viewing portal, and a central AM portal which is used as both a viewing and a working portal [55]. The torn ACL can then be debrided through a working portal using an arthroscopic shaver or ablation device [57].

After the portals are made, the surgeon will assess the torn ACL and check if there are any other injuries. A graft is then needed to replace the ruptured ACL [5]. A variety of grafts have been used over the history of ACL reconstruction including autologous, allogenic, and synthetic grafts [5], [58], [59]. Each type of graft has different advantages and disadvantages where the choice of graft type influences the subsequent kinematics, biological response, knee stability, flexion/extension strength, risk level, and rate of complications [53], [60]. In clinical practice, though, autologous grafts are most commonly used [5]. The three most commonly used autologous grafts are hamstring tendon grafts, quadriceps tendon grafts, and bone-patellar tendon-bone grafts [5]. Once the graft is harvested, it will be cleaned, sized, and prepared for insertion [61]. One tibial tunnel and one femoral tunnel will be drilled for the attachment of the graft [57]. The native ACL contains two bundles (AM and PL) which insert in different points on the femur and the tibia [28]. In this surgical approach, a single tunnel is drilled in both the femur and tibia. As a result, the choice of where to position the tunnels to generate the best post-surgical outcomes is highly debated and often varies depending on surgeon preference. The graft is inserted through the tunnel and then the ends of the graft are fixed to secure the graft in the tunnel [60]. This ACL reconstruction method is the most common version of the procedure and is a single bundle approach.

Another popular method for ACL reconstruction is the double-bundle approach [28]. Since the native ACL contains two bundles, the double-bundle approach aims to replicate this by reconstructing both the AM and PL individually with two distinct grafts [28], [53]. Two guide pins are inserted to

determine the tunnel locations and orientations of the two bundles [53]. The guide pins are carefully placed far enough away from each other to make sure that the tunnels do not merge [53]. The prepared grafts are then inserted through the tunnels and fixed in the tunnels [53]. The double-bundle approach has shown some success in the literature, but it has a significant learning curve for surgeons [58]. Since two additional tunnels must be drilled, this methodology has increased the opportunity for errors [58].

2.4.1 Tunnel Placement

2.4.1.1 Isometric vs. Anatomic

While the aim behind ACL reconstruction is the same for all reconstruction methodologies, there are different beliefs behind why different tunnel placements should be utilized. Primarily, the concept of isometric graft placement compared to anatomical graft placement are two different rationales for the placement of tunnels [28]. Isometry refers to the fact that some ligaments maintain their length during the permitted range of motion of the joint [28], [62]. The benefit of isometric graft placement is that it prevents changes in graft length and therefore tension on the graft during knee flexion range of motion, which can avoid overstretching the graft and possibly graft failure [28], [62]. It is important to note, though, that natively when the knee goes through its full range of motion, there is no long-term deformation of the ACL and no point of the femur maintains a fixed distance from a single point on the tibia [28]. As such, in its native state, the ACL is not isometric [28]. Anatomical tunnels refer to placing the graft tunnels in anatomical attachment positions of the native ACL [28]. The goal behind this concept is that the anatomical attachment sites could potentially better replicate the biomechanics of the native ACL and reduce the risk of re-rupture, ongoing instability and early onset arthritis [28].

An isometric graft position places the femoral tunnel high and posterior in the lateral femoral condyle [28]. An anatomically placed femoral tunnel places the tunnel in the footprint of the native ACL femoral attachment site [28]. An anatomically placed tibial tunnel would place the tibial tunnel at the center of the native ACL tibial attachment site [28]. Biomechanical testing has shown that anatomical tunnel placement can produce more similar kinematics to the intact knee than isometrically-placed tunnels [14], [28], [63], [64].

2.4.1.2 ACL Strain Profile

The stress-strain profile of the ACL is nonlinear in nature [65]. In particular, it has three distinct regions: toe region, linear region (constant slope), and prefailure region (decreasing non constant slope) [65].

This can be idealized by the stress-strain profile and load-deformation profile shown in Figure 5 and Figure 6 [65]. As the mechanical properties of the ACL change with age, the load at failure changes as well [65]. The load at failure is typically between 650N-2160N depending on the person's age and sex [65]. Maximum deformation is typically around 10mm [65]. Once the ACL is stretched around 10mm, it has typically entered into the prefailure region of the stress-strain curve and it cannot return back to its original length [65].

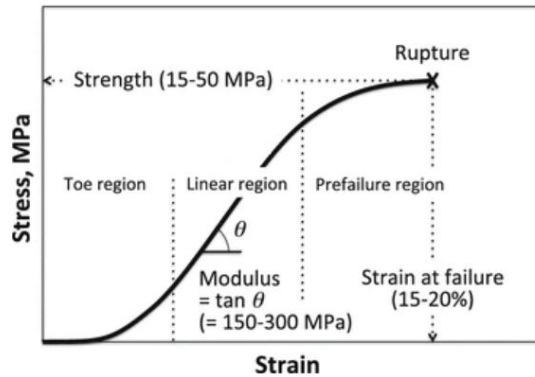


Figure 5: Stress strain relationship of the anterior cruciate ligament (ACL).

Stress in MPa vs. strain plot where the toe region, linear region, and prefailure regions are identified. Adapted from [65].

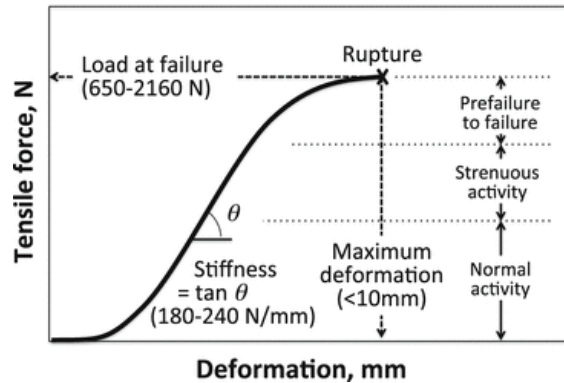


Figure 6: Load deformation relationship of the anterior cruciate ligament (ACL).

Tensile force in N vs. deformation in mm where the curve is separated by three activity levels (normal activity, strenuous activity, and prefailure to failure). Adapted from [65].

As previously discussed, tunnel positioning to achieve graft isometry compared to anatomic tunnel placement leads to different strain profiles of the graft over the knee range of motion. This is why improper tunnel placement can cause graft failure [3], [3], [64]. When discussing the degree of

isometry of a graft, length change in the ACL graft compared to knee flexion angle is an important measurement [66]. In particular, a length change over 4% is associated with overstretching and graft failure [66]. Anatomically placed tunnels results in longer intra-articular distance than isometrically placed grafts [66]. Additionally, when the knee is extending, this is when the most length change in the graft occurs and the graft is the longest [66], [67].

2.4.1.3 History of Tunnel Placement

Before the introduction of the arthroscope, early ACL reconstruction surgeries were open knee surgeries [58], [68]. In the 1960s and 1970s, the “over-the-top” approach was a popular surgical approach, where surgeons harvested a band of fascia lata that was then passed over the top of the lateral femoral condyle and screwed into the anterior tibial plateau [58], [68]. This approach, though, lead to non-anatomical placement of the graft [68].

When the arthroscope became commonly used for ACL reconstruction in the 1980s, it revolutionized the way this procedure was conducted [68]. The arthroscope allowed for visualization of the femoral notch without an open knee procedure [68]. Through this, the outside-in technique was developed, where the tibial tunnel is drilled using guides and the femoral tunnel is drilled into the knee joint space through an incision made on the lateral side of the femur [68]. The use of a second lateral incision allowed for more freedom in the placement of the femoral tunnel, leading to more anatomically-oriented femoral tunnel positioning [68].

In the early 1990s, the transtibial technique was developed [68]. In this technique, the tibial tunnel is drilled first and then the femoral tunnel is drilled with tools passed through the tibial tunnel [68]. This technique grew in popularity as it removed the need for a second lateral incision on the femur [68]. The disadvantage of this methodology, though, is that the femoral tunnel location and orientation is dependent on the location of the tibial tunnel [68]. As such, this technique is associated with the less anatomically-placed femoral tunnels [68]. This method was the most commonly used ACL reconstruction technique for many years [68].

In 1995, the use of a low AM portal was first described [68]. This portal allows for independent AM drilling of the femoral tunnel [68]. In particular, using an AM portal for femoral tunnel drilling is associated with good anatomic tunnel positioning on the lateral femoral condyle [68]. With increasing surgeon preference of anatomic femoral tunnel placement, this technique became the most preferred technique for surgeons as of 2013 [68]. The uses of flexible reamers through the AM portal has further

strengthened this approach by allowing for femoral tunnel drilling that is longer and farther away from the posterior femoral cortex [68].

2.4.1.4 Associated Risk of Tunnel Placement

Incorrect femoral tunnel placement is a primary cause of failure for ACL reconstruction [3], [6], [7], [28], [69]. When the femoral tunnel is placed anteriorly, the resulting graft has a mismatch in graft tension in extension vs. flexion [69]. A graft tensioned in extension can restrict knee flexion and the patient may feel tightness of the graft during flexion [28], [69]. A graft tensioned in flexion can be too loose in extension and increase the laxity of the knee [69]. When the femoral tunnel is placed too high in the notch, the resulting graft is vertically oriented [69], [70]. This type of positioning does not replicate the orientation of the native ACL (which is oblique), which could hinder the ability of the graft to restore normal knee kinematics, especially rotationally [69], [70]. In particular, the location of the femoral tunnel causes changes in the graft length-tension relationship [69]. Finally, the last common position of the femoral tunnel associated with risks is a posterior position [69]. A posteriorly-placed femoral tunnel can lead to blowout of the back wall of the tunnel [69]. This is summarized in Table 1.

When the tibial tunnel is placed anteriorly, graft impingement is a significant risk, which could lead to graft failure [28], [69]. In this situation, the graft gets impinged on the intercondylar roof when the knee is in extension [69]. This can lead to a reduction of the range of extension motion or progressive elongation, which can lead to the failure of the graft [69]. If the tibial tunnel is drilled too medially, there is a risk of graft impingement against the PCL [69]. If the tibial tunnel is drilled too laterally, there is a risk of graft impingement against the medial portion of the lateral femoral condyle [69]. Finally, a tibial tunnel that is drilled too posteriorly can lead to a graft that is too lax in flexion [69]. This is summarized in Table 2.

Table 1. Risk Resulting from Femoral Tunnel Location Based on Common Position

Position	Risks
Anterior	Mismatch in graft tension in extension vs. flexion
Posterior	Blowout of the back wall of a tunnel
Too high in the notch	Vertical graft. Less resistance to rotational forces.

Table 2. Risk Resulting from Tibial Tunnel Location Based on Position

Position	Risks
Anterior	Risk of impingement on the intercondylar roof in extension
Posterior	Too lax in flexion
Medially	Risk of graft impingement against the PCL
Laterally	Risk of graft impingement against the medial aspect of the lateral femoral condyle

2.4.2 ACL Repair

While ACL reconstruction is the current primary surgical intervention for a ruptured ACL, historically this was not always the case [71]. In the 1970s and 1980s, open primary ACL repair was the main surgical intervention for a ruptured ACL [71]. Open primary ACL repair was a highly invasive open knee surgery where the ends of the ACL were stitched back to femur or the tibia using catgut ligatures [71], [72]. Initially, this methodology showed good success in the short-term, but the results began deteriorating around a mid-term follow-up [71], [72]. There was a high reinjury rate associated with this procedure and high rates of pain, stiffness, and instability [72]. Due to these issues, ACL repair shifted out of favour and eventually, with the rise in minimally invasive surgery and significant innovation within arthroscopy, ACL reconstruction became the predominant surgical intervention for treatment of a ruptured ACL [72].

In the last decade, investigations into arthroscopic ACL repair has begun [72]. Recent results have shown that arthroscopic ACL repair, when applied to certain patient demographics (proximal tears) has shown excellent results [72]. Additionally, newer approaches including the Internal Brace Ligament Augmentation (IBLA), Dynamic Intraligamentary Stabilization (DIS), and the Bridge-Enhanced ACL Repair (BEAR) procedures have been developed [71], [73], [74]. These approaches do not rely on grafts and rely on reattachment of the native ACL two ends of the native ACL [74]. These approaches all preserve the native ACL ligament, native insertion sites, are less invasive compared to ACL reconstruction, and require no graft harvesting [74]. There are currently a limited number of large high quality randomized controlled trials to compare between ACL reconstruction and ACL repair [74]. Additionally, more high-quality large randomized controlled trials are needed to compare between the different ACL repair techniques as well [74].

2.5 Experimental Investigations of Knee Biomechanics

To understand and measure knee biomechanics, there are a number of techniques that have been employed in literature. Custom built testing simulators, industrial testing simulators, or 6 degrees of freedom robotic arms are some of the primary methods used in research to prescribe defined motions or loads to the knee joint and record the resulting loads or kinematics.

2.5.1 Custom Built Testing Simulators

Custom-built testing simulators are designed by users to perform desired motions, apply specific forces, or both for conducting and recording biomechanical testing data. There are many examples of custom-built knee biomechanics simulators in literature, some examples include the squatting/lunging simulator by Driscoll *et al.* (2012) and the pivot-shift test apparatus by Markolf *et al.* (2010) [13], [14].

Driscoll *et al.* (2012) used a custom-built simulator to compare two different anatomical femoral tunnel locations using a single-bundle ACL reconstruction approach [14]. Six fresh-frozen cadaveric knees were used for this study where two different ACL reconstructions were done to each specimen using bone-patellar tendon-bone autographs (center of the AM portion of the femoral footprint or centered in the femoral footprint) [14]. This study focused on using two loading conditions (anterior tibial load and a combined valgus load with an internal tibial torque) [14]. The simulator was a squatting/lunging simulator which was designed to replicate the ankle and hip joints during a squat or a lunge and is shown in Figure 7. [14]. The tunnel positioned in the center of the femoral footprint had a greater resistance to internal tibial rotation at all knee flexion angles [14].



Figure 7: Example of a custom-built testing simulator from Driscoll *et al.* (2012).
Custom simulator is shown with a mounted knee specimen. Adapted from [14].

Markolf *et al.* (2010) leveraged a custom-built pivot-shift test apparatus to apply a valgus moment to the tibia through a weight-pulley system [13]. In their experiment, single-bundle ACL reconstruction was performed using two different femoral tunnel positions (standard and oblique) on 17 unpaired cadaveric knees using a patellar tendon autograft [13]. The results showed that moving the femoral tunnel to a more oblique position did not create a significant change in the pivot-shift kinematics [13].

Cassidy *et al.* (2013) developed a custom-built dynamic knee simulator to replicate sagittal plane knee motion and apply dynamic muscle forces [75]. In their experiment, motion capture data of subjects performing a jump landing on a force plate was collected [75]. This data was then inputted into a musculoskeletal model to determine the force-time profiles of each muscle group across the knee and joint motion data of the hip and ankles to input into the simulator [75]. The results showed that the simulator was able to accurately simulate dynamic muscle loading and motion of a jump landing [75].

2.5.2 Industrial Testing Simulators

Industrial testing simulators are systems sold by companies that are used in both research and industry for conducting biomechanical testing on medical implants and on cadaveric specimens. One popular industrial testing simulator is the Advanced Mechanical Technology Inc (AMTI) VIVO six degree of freedom joint motion simulator, designed to test prostheses and biological joint specimens [76].

Sarpong *et al.* (2019) leveraged the AMTI VIVO system to for virtual ligament reconstruction of the PCL for component testing in total knee arthroplasty [77]. They used three human cadaveric knee specimens, recorded the intact knee kinematics, cut the PCL, recorded the PCL-deficient knee kinematics and then added a virtual PCL to attempt to restore the native knee kinematics [77]. A three-bundle virtual PCL was reconstructed which achieved less than 10% random mean square error [77].

Vakili *et al.* (2024) leveraged the AMTI VIVO system to develop a methodology to create specimen-specific virtual ACL reconstructions in cadaver knees, shown in Figure 8 [15]. In this study, twelve cadaver knees were used [15]. Native knee kinematics and ACL deficient knee kinematics were recorded during replicated clinical laxity tests (anterior drawer test, Lachman test, pivot shift test). A multi-objective optimization was used to create specimen-specific virtual ACLs that best replicated the force-elongation behaviour as the intact knee [15]. The results showed that the specimen-specific virtual ACLs could be used instead of the real ACLs [15].

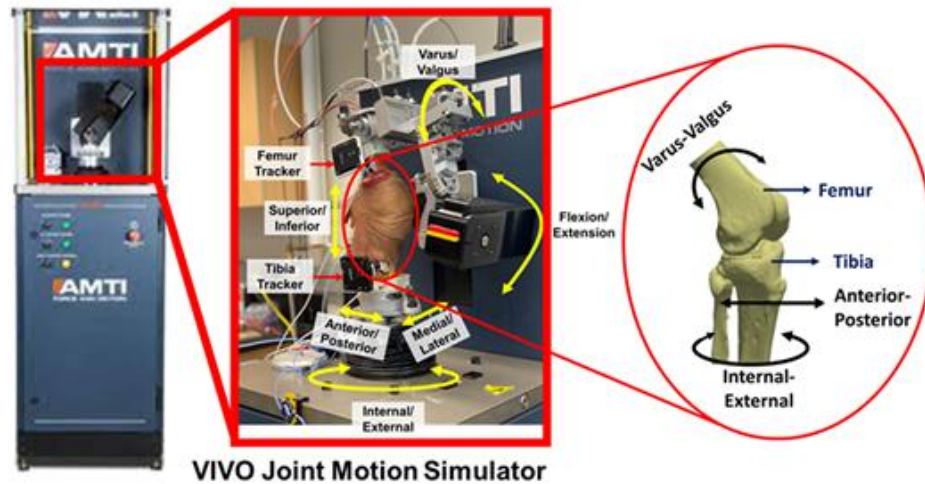


Figure 8: Example of an industrial testing simulator (VIVO) used by Vakili *et al.* (2024). *VIVO joint motion simulator is shown with a mounted knee specimen. Clinical rotations and translations of the positioned knee joint are noted. Adapted from [15].*

2.5.3 Robotic Arms

Robotic arms combined with additional sensors/system (often force-torque sensors and motion capture or tracking system) can also be used for conducting biomechanical experimental testing. Robotic arms can create a faster way to develop testing methodologies compared to custom-built testing simulators and can be more cost-effective than some industrial testing simulators. Some examples of robotic arms

used for biomechanical testing are the KUKA KR 60-3 used by Goldsmith *et al.* (2013), and a robotic universal force-moment sensor testing system used by Musahl *et al.* (2005) [11], [63].

Goldsmith *et al.* (2013) leveraged a KUKA KR 60-3 6 degree of freedom robotic system equipped with a force-torque sensor to compare between the anatomic single- and double-bundle ACL reconstruction approaches, shown in Figure 9 [11]. They used nine pairs of cadaveric knees to evaluate the kinematics of ACL-deficient, anatomic single-bundle, or anatomic double-bundle reconstructed knees using simulated clinical laxity tests (anterior tibial load, simulated pivot-shift, and rotation motion) [11]. The results showed no significant differences between the double-bundle reconstruction and the single-bundle reconstruction [11]. This experimental setup is shown in Figure 9.

Musahl *et al.* (2005) leveraged a robotic universal force-moment sensor testing system to apply different loading conditions (anterior load, combined valgus and internal rotation load) to ten cadaveric knees [63]. In their study, two tunnels were chosen in the femoral notch where one tunnel was within the anatomical footprint of the ACL and the other tunnel was chosen for best graft isometry [63]. A hamstring tendon graft was used for the ACL reconstruction [63]. The tunnel anatomical tunnel position resulted in knee kinematics that were closer to the intact knee, but neither tunnel restored the native knee kinematics [63].

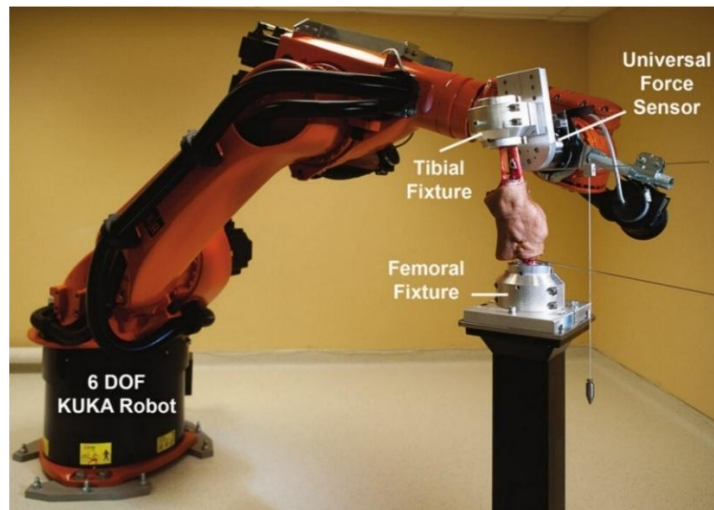


Figure 9: Example of a robot arm used for knee biomechanics testing from Goldsmith *et al.* (2013).

KUKA 6 degree of freedom robot is shown with a labelled femoral fixture, tibial fixture, and universal force sensor. Adapted from [11].

2.6 Surgical Navigation for Knee Reconstruction

Surgical navigation is the combination of imaging and optical tracking methods during surgery which allows for real-time feedback during surgery, increased accuracy, and greater reproducibility of surgical procedures [78], [79]. Surgical navigation systems are key assets to surgeons for certain orthopaedic procedures such as hip and knee replacements, spine surgeries, tumor surgery, and for some trauma surgeries [79].

Optical tracking systems used in the surgical environment allows for real-time continuous localization of medical tools and specific patient anatomy [80]. Additionally, as all tracked objects exist in the same environment, tracked tools can be localized alongside patient anatomy and specific patient anatomy can also be tracked with respect to other patient anatomy (e.g., tracked tibia with respect to the femur) [80]. The use of tracking systems allow for more precise intervention since these systems provide accurate spatial information about patients and allow surgeons to position tracked tools accordingly [80]. Optical tracking systems are very accurate, where individual markers can be tracked with an accuracy of less than 1 mm [81]. The primary disadvantage of optical tracking systems is that they require a direct line-of-sight at all times between the tracker and the markers [80]. As such, surgical workflow planning is necessary for leveraging optical tracking systems to ensure that a direct line-of-sight will be permissible at all points during the procedure [80].

2.6.1 Surgical Navigation Workflow

Surgical navigation procedures typically share a common workflow regardless of surgical discipline [82]. This workflow can be split into two stages: preoperative and intraoperative [82]–[84]. For the preoperative workflow the stages are as follows: image acquisition, segmentation of 3D models, and review of 3D models [82]. The type of medical image acquisition that is needed varies depending on the system, but computed tomography (CT) or fluoroscopy are examples of commonly used medical images for surgical navigation systems [78]. The preoperative workflow process begins with medical images being taken from patients [78]. Key anatomy, dependent on the procedure, would then be segmented from the medical images to get patient-specific 3D models [82]. These models would then be reviewed by clinicians to ensure that they are reflective of the patient medical images [82]. Once all of these stages are complete, the preoperative workflow is complete.

For the intraoperative workflow the stages are: positioning of the patient and markers, registration, and procedure-specific workflow [82]. Once the procedure starts, the optical tracking

system will be set up, the patient will be positioned, and the markers will be positioned on the patient. Then, the relationship between the patient's anatomy and the anatomy in the surgical navigation system user interface needs to be defined. Defining this relationship between patient anatomic landmarks and the same landmarks in patient-specific 3D models is called registration [78]. The registration process typically involves selecting key anatomical landmarks on patient-specific 3D models images prior to surgery and then defining those landmarks on the patient intraoperatively [78]. The mathematical relationship between the pairs of points is then defined, which allows for the position of other anatomy between the points to be defined as well [78]. The process allows the 3D models to move as the patient's anatomy is moved. Finally, procedure-specific workflow can then be followed dependent on the system being used.

2.6.2 Surgical Navigation for ACLR

The use of surgical navigation technology for ACLR has been mentioned in literature since 1995 [9]. Before 2006, though, there were few mentions in literature (9 listed Medline publications) [85]. In the early days of navigated systems for ACLR, most applications focused on improving the accuracy of tunnel placement [9]. Later generations of navigated systems focused more on measuring knee kinematics and knee laxity [9]. Proposed by S. Zaffagnini *et al.* (2016), there are four primary research areas of navigated ACLR: assistance for tunnel placement, kinematic evaluation, comparison of effectiveness of different ACLR procedures, and ability of navigation to improve ACLR outcomes and cost-effectiveness [9].

As previously discussed, poor tunnel placement is one of the key reasons behind ACL graft failure [3], [7], [86], [87]. Many studies have shown that there is a reasonable interobserver and intraobserver variability in ACL tunnel positioning as well as for intraarticular landmark identification [88], [89]. Additionally, the viewing portal has been shown to play a key role in the accuracy with which surgeons can located their intended femoral tunnel position [90]. As such, many navigated ACLR systems aim to better select and confirm the accurate selection of the tibial and femoral tunnels [9]. In particular, many systems focus on femoral tunnel placement and have analysis to prevent graft impingement [91]–[93].

To prevent graft rupture, the graft must stay in the elastic region of the stress-strain curve [8], [94]. Patient-to-patient differences, such as bone morphology, play a key role in kinematic differences across patients as well [94]. Evaluation of static and dynamic laxity as well as knee stability are some

of the key kinematic evaluations that have been researched for use in ACLR to gain a better understanding of the knee kinematics post-reconstruction [9]. As well, there has also been research interest in understanding pre-operative laxity compared to post-operative laxity, where a higher pre-operative laxity is often correlated with a higher laxity post-ACLR [95].

For the area of surgical comparison, most studies are focused on the comparison of kinematic outcomes post-reconstruction [9]. Key metrics included anterior tibial translation and internal external rotation at varying degrees of flexion [9]. Different ACLR methods that have been compared include single-bundle compared to the double-bundle techniques [9]. As well, investigations into the use of lateral extra-articular tenodesis (LET) to reduce the tibia's anterior translation and internal rotation along with the single-bundle technique have also been conducted [9].

Finally, the performance, clinical outcomes, and cost-effectiveness of navigated ACLR systems have been key areas of research in the navigated ACLR field [9]. As previously discussed, navigation systems have been shown to improve tunnel position [9]. More accurate tunnel positioning should theoretically lead to enhanced clinical outcomes (less graft failure, thus fewer revision surgeries), navigated ACLR procedures have not yet shown this to be true though [9]. Investigations into the clinical outcomes of navigated ACLR procedures have shown that navigated ACLR systems do not create significantly better outcomes [10]. As well, navigated ACLR procedures are longer when compared to traditional ACLR procedures and are more costly [10]. Many of the studies investigating this comparison have relatively small sample sizes and focus on short-term clinical outcomes. Graft failure though, is not the only key comparative outcome of ACLR procedures though. Long-term knee stability and prevention of osteoarthritis are the two primary goals post-ACLR [9]. Studies focused on comparing the long-term clinical outcomes of standard ACLR compared to navigated ACLR have not yet been conducted [9].

2.7 Summary of Background Literature

The literature review presented in this chapter provided an overview of the knee anatomy, knee kinematics, ACL injury epidemiology, ACL reconstruction methodologies, *in vitro* biomechanical testing methods, and the current limitations of surgical navigation systems used for ACL reconstruction. *In vitro* biomechanical testing is critical to understanding the effects of ACL reconstruction techniques on the kinematic and clinical outcomes of patients. In particular, this type of testing has provided valuable insight on how surgeons conduct these procedures and on why certain techniques produce

better clinical outcomes than others. While there have been some *in vitro* biomechanical investigations into the influence of femoral tunnel position on ACL biomechanics, there is still limited understanding of exactly how much kinematic variation is produced through deviations in femoral tunnel positioning while keeping the tibial tunnel location constant. Additionally, despite the innovation into surgical navigation systems for ACL reconstruction, no significantly improved clinical outcomes have been associated with these systems. A greater understanding of the influence of tunnel positioning on knee biomechanics is needed in order to better develop newer generations of surgical navigation systems for ACL reconstruction. This thesis work will explore these knowledge gaps by leveraging the virtual ligament capabilities of the AMTI VIVO joint motion simulator to investigate the role of the femoral tunnel position in knee biomechanics and will additionally propose a surgical navigation system for ACL reconstruction.

Chapter 3

Comparison of 5 Femoral Tunnel Locations in Single-Bundle Virtual Anterior Cruciate Ligament Reconstructions: An *In Vitro* Biomechanical Study

3.1 Introduction

The femoral tunnel plays a key role in ensuring successful clinical outcomes following ACL reconstruction. While there have been some investigations into the biomechanical influences of femoral tunnel positioning, it is still not well understood how it affects knee biomechanics during laxity tests, the force the graft carries during these tests, or the graft's isometry profile. Therefore, the overall goal of this *in vitro* biomechanical study was to investigate the influence of femoral tunnel position on knee biomechanics.

3.1.1 Objectives

The specific objectives of this study were as follows:

1. to investigate the influence of femoral tunnel location on knee kinematics, including anterior tibial translation and internal rotation, during simulated clinical laxity tests;
2. to quantify the influence of femoral tunnel location on the force carried by the virtual ACL graft, during simulated clinical laxity tests;
3. to examine the influence of femoral tunnel location on graft length change across knee flexion (0 deg to 90 deg).

3.1.2 Hypotheses

The specific hypotheses of this Chapter, aligned respectively with each objective, are as follows:

1. varying the femoral tunnel insertion location of a virtual ACL graft, while keeping the tibial insertion location constant, will result in significant differences in knee joint kinematics during motion;
2. varying the femoral tunnel insertion site of a virtual ACL graft, while keeping the tibial insertion location constant, will lead to significant changes in the forces experienced by the virtual graft;

3. varying the femoral tunnel insertion location of a virtual ACL graft, with a constant tibial insertion location, will produce significant differences in graft length change throughout the knee flexion range of motion.

3.2 Methods

3.2.1 Experimental Setup

The use of the VIVO joint motion simulator (AMTI, Waltham, MA, USA) for biomechanical testing has been well documented in previous literature [15], [96]–[101]. The VIVO has two actuators, one upper and one lower, where the upper actuator contains the flexion arm and the abduction arm, and the lower actuator is responsible for the translation movements and one rotational movement [76]. The VIVO simulator has six degrees of freedom, allowing for the application of loads, translations, rotations, and combined loading protocols [76]. For this experiment, a custom modification to the VIVO at the Biomechanical Engineering Research Lab at Western University was used, where a stepper motor was attached to the VIVO to separately control the degree of knee flexion. This allowed the lateral side of the specimen to always be facing the optical tracking system. This stepper motor attached to the VIVO is shown alongside a mounted specimen in Figure 10.

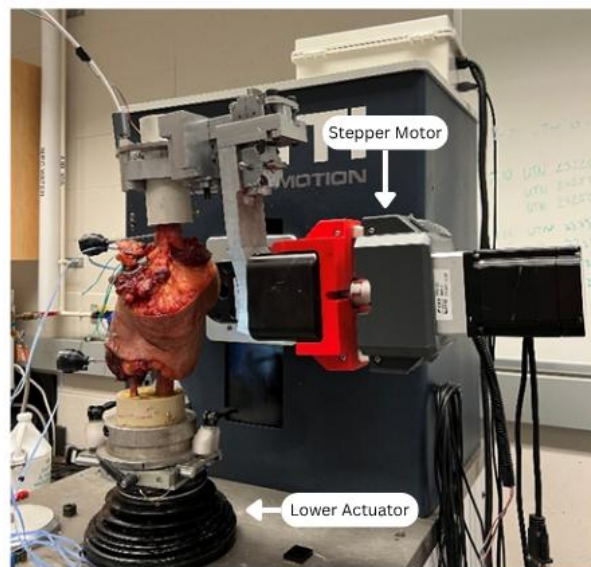


Figure 10. VIVO joint simulator with a mounted cadaveric specimen.

A custom modification to the VIVO joint motion simulator with a stepper motor is shown and labeled, along with the VIVO lower actuator.

The Optotrak Certus system (Northern Digital Inc., Waterloo, Ontario, Canada) leverages active optical technology to measure the position and orientation of markers within a known volume. The markers contain light emitting diodes (LEDs) that are able to be tracked by the position sensor once the markers are active [102]. The Optotrak Certus has a measurement volume of 30 m³ [102]. The Optotrak Certus position sensor has sub-millimetre accuracy and has been previously used for data recording in biomechanical testing [15], [100], [102].

A custom-designed cable system for the VIVO was also used in this experiment, which is shown in Figure 11. The cable system directly interfaces with the custom-designed aluminum fixtures, which are used to mount the proximal femur to the upper actuator of the VIVO. One end of the cable passes through the aluminum fixture and can be attached to tissue on the specimen while the other end attaches to a weight. This allows for specific constant loads to be applied to the attached tissue.



Figure 11. Custom-designed cable system, side view.

The side view of a custom-designed cable system used to apply load to the iliotibial (IT) band.

A custom-built MATLAB app was used for this experiment. This interfaced directly with the Optotrak Certus and the VIVO. The app was used to record the Optotrak Certus data, control the flexion angle of the stepper motor, and to input the parameters for the virtual grafts. As well, the optimization for calculating the reference strain of the virtual grafts was also done in MATLAB directly through the MATLAB app. The complete experimental setup including the VIVO joint motion simulator, Optotrak Certus, and a custom-designed cable system is shown in detail in Figure 12

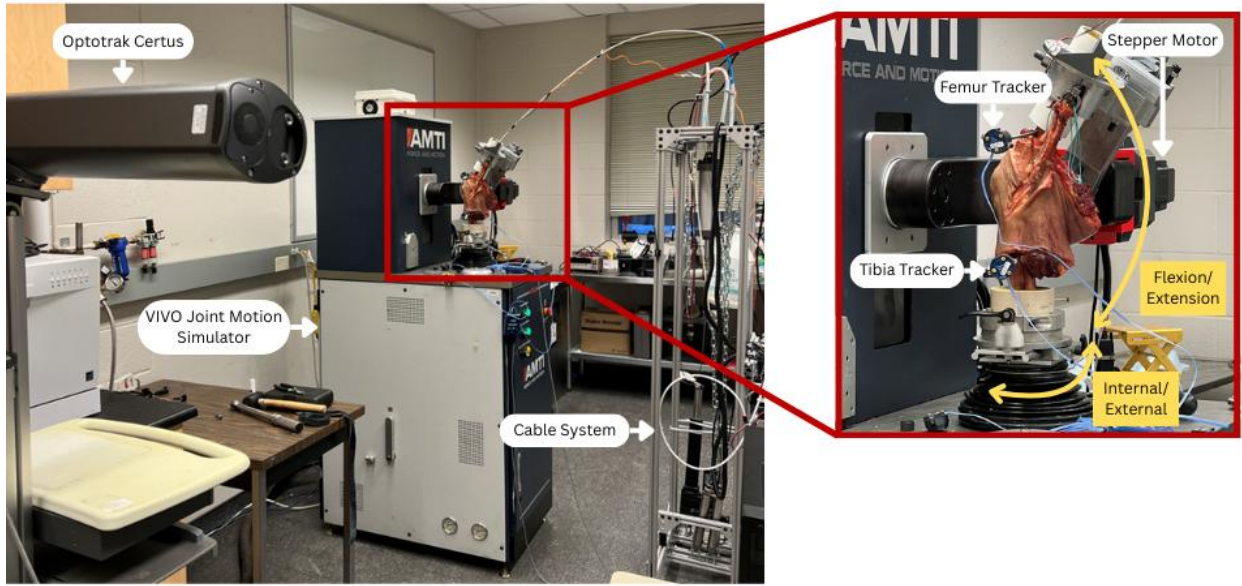


Figure 12. Overview of the experimental setup.

Experimental setup depicted includes the AMTI VIVO joint motion simulator, Northern Digital Inc. Optotrak Certus, cable system to load the iliotibial (IT) band. Knee specimen mounted on the simulator with highlighted femur tracker, tibia tracker, stepper motor, and key clinical rotations labelled.

3.2.2 Specimen Preparation

Nine fresh frozen intact human cadaveric knee specimens (3 male, 6 female, average age 63 ± 11.4 , and average BMI of 27.6 ± 10.8) were used in this study. A computed tomography (CT) scan of each specimen was done prior to the experiment with custom 3D printed plastic fiducials that were fixed to the femur and tibia. Threshold-based segmentation methods in 3D Slicer were used to create models of the femur and tibia bones and to mark the flexion axis of the knee. 24 hours before the experiment, the specimen was taken out and thawed at room temperature. The femur, tibia, and fibula were transected approximately midfemur and midtibia, respectively, leaving all the soft tissues around the knee intact. The proximal femur was potted in a polyvinyl chloride (PVC) pipe using dental cement and mounted

onto the upper actuator of the VIVO simulator using a custom-designed aluminum fixture. The distal tibia and fibula were then potted into a PVC pipe using dental cement and secured to the lower actuator. Two LED rigid body trackers (Optotrak Certus, Northern Digital Inc., Waterloo, Ontario, Canada) were used to track the position of the bones, where one tracker was attached to the femur and the other to the tibia. The tracked position of the bones was used to align the femur and tibia central to the simulator. The reference pose of the knee was then defined with the knee in full extension with 10 N of compression applied. This potting procedure has been described fully in a previous study [103].

3.2.3 Simulated Motions

For each intact knee, three sequential loading conditions were applied: (1) an anterior-directed force to the tibia, (2) an internal torque applied to the tibia, and (3) a combined anterior-directed force with an internal torque applied to the tibia. For the anterior-directed force, 89 N of force was applied. For the internal torque, 5 Nm was applied. For the combined loading protocol, 89 N of anterior-directed force and 5 Nm of internal torque were applied. These three loading scenarios were sequentially applied (anterior-directed force, internal torque, combined loading) and then repeated for two additional cycles (three cycles total applied to each specimen). The complete waveform of the motion sequence is shown in Figure 13. This motion sequence was applied to each specimen at 0 deg, 30 deg, 60 deg, and 90 deg of flexion. The anterior-directed force at 30 deg of flexion represents a simulated Lachman test while the anterior-directed force at 90 deg of flexion represents a simulated Anterior drawer test [104]. The knee flexion angle was manually adjusted between tests with the simulator in displacement-control mode, while the motion sequence was done in force control mode (with the flexion angle fixed).

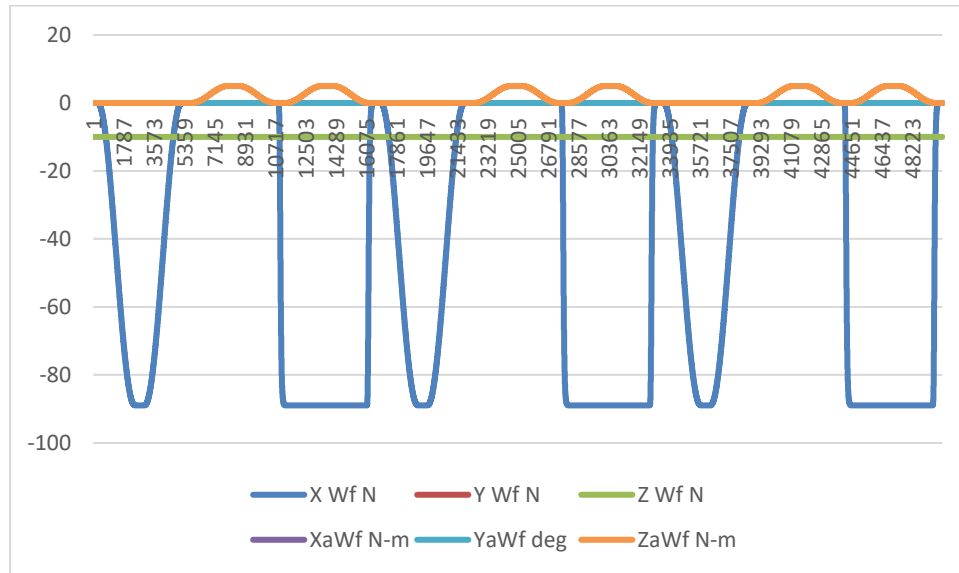


Figure 13. Individual waveforms of the six degrees of motion of the applied motion sequence.

Six degrees motion include the X, Y, and Z force waveforms which are in N and the Xa, Ya, Za moment waveform which are in Nm.

3.2.4 Applying Load to the Iliotibial (IT) Band

The custom-designed cable management system allows a cable to run through the top of the custom-design fixtures that mount the femur to the VIVO, which allow a tension of 30N to be applied to the IT band. This system was primarily used for a secondary study (not part of this thesis) and is shown in Figure 12.

3.2.5 Anterior Cruciate Ligament (ACL) Removal

Once the intact knee kinematics were captured, the specimen was removed from the simulator and the ACL was dissected by a fellowship-trained orthopaedic surgeon with a shaver under arthroscopic visualization. The specimen was then re-mounted by aligning a marked position on the PVC pipe with a marked position on the custom fixture and then tested in two manners. Firstly, the destabilized specimen was tested in displacement control using the recorded displacements of the intact specimen. Secondly, the destabilized specimen was tested in force control using the same previously described motion sequence.

3.2.6 Virtual Anterior Cruciate Ligament Reconstruction

The use of virtual grafts during VIVO simulation testing has been well documented in previous articles, where the VIVO is able to incorporate continuous nonlinear tension-only force contributions of virtual grafts which are represented by one-dimensional point-to-point springs [15], [77], [100], [101]. The force response of the virtual graft is represented by a nonlinear force-elongation relationship shown in Equation 1, where the spring parameter ε_1 is assumed to be 0.03 as in other studies [105]–[108]. To design a virtual ACL, the virtual graft requires the following parameters: graft stiffness parameter (k), reference strain (ε_r), and the attachment points to the femur and tibia. The reference strain (ε_r) is the strain in the virtual graft at the reference pose (at full extension). The reference strain is a function of the zero-load length (l_0) and the reference length (the length of the graft in the reference pose) (l_r) shown in Equation 2 and 3.

$$f = \begin{cases} \frac{1}{4} k \varepsilon^2, & 0 \leq \varepsilon \leq 2\varepsilon_1 \\ \frac{\varepsilon_1}{\varepsilon}, & \varepsilon > 2\varepsilon_1 \\ 0, & \varepsilon < 0 \end{cases} \quad (1)$$

$$\varepsilon_r = \frac{l_r - l_0}{l_0} \quad (2)$$

$$l_0 = \frac{l_r}{\varepsilon_r + 1} \quad (3)$$

3.2.7 Virtual Graft Parameters

To define the attachment coordinates of the virtual ACL, an expert orthopaedic surgeon identified the ideal femoral and tibial attachment points within 3DSlicer using a volume rendering of the specimen anatomy and CT images (anatomic single bundle, center of the ACL footprint on the femur and the tibia). Two additional femoral attachment points were identified 4.6 mm proximal and distal to the ideal femoral attachment point along the long axis of the femur (identified using a line in 3DSlicer) [90]. Two additional femoral attachment points were identified 4.6 mm anterior and posterior to the ideal femoral attachment point perpendicular to the long axis of the femur. 4.6 mm was selected as the distance between the tunnel positions as data in the literature has shown that the average distance between intended and achieved femoral tunnel positions for high-volume ACL surgeons is 4.6 mm

when using a free-hand technique [90]. An example annotation of the femoral attachment points in 3DSlicer is shown in Figure 14.

To define the virtual graft parameters, the femoral and tibial attachment coordinates (of the tracked specimen on the simulator) were inputted along with the graft stiffness and the tensioning force. The virtual graft was meant to represent a quadriceps tendon graft, as such the graft stiffness that was used was 466.2 N/mm [109]. Based on the VIVO system requirements, the stiffness value was inputted as a value in N, where the ACL length was assumed to be 33 mm, giving a stiffness of 15385 N [110]. The tensioning force that was applied was 44 N [111]. The tensioning force was used to define the reference strain value (2), where the tibia was permitted to freely move (relative to the femur) which created adjustments in the zero-load length until the reference strain reached a consistent value which allowed for the chosen tensioning force to be applied.

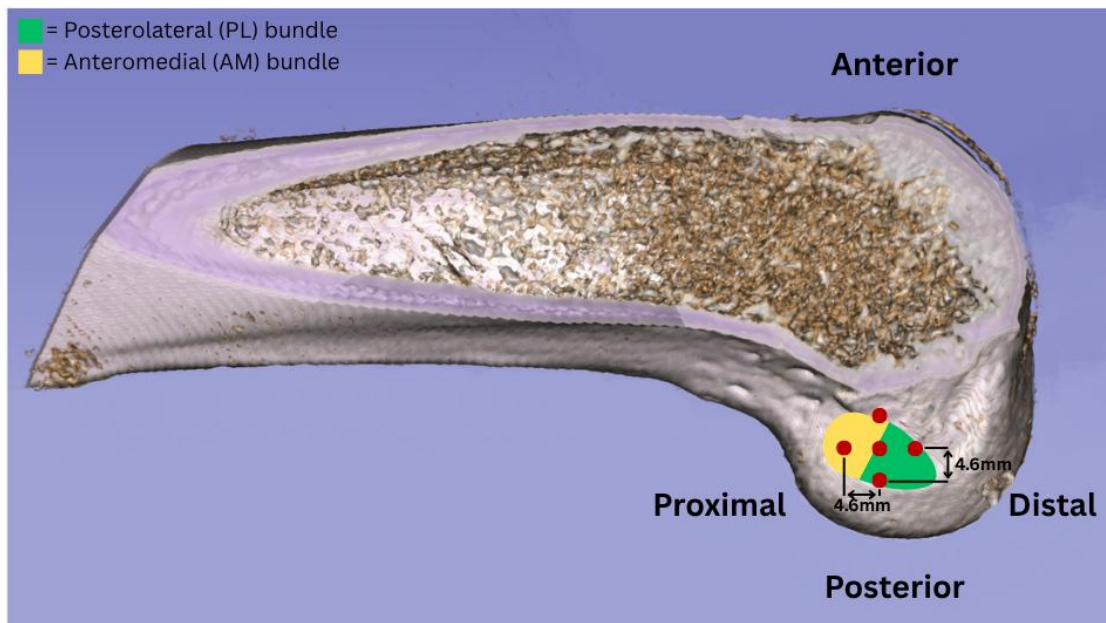


Figure 14. Five femoral attachment conditions.

The annotated five femoral attachment coordinates are named relative to their direction from the ideal (central). Graft conditions included ideal, distal, proximal, anterior, and posterior.

3.2.8 Virtual Graft Motion Capture

Once the virtual graft was successfully implemented in the VIVO simulation environment, the specimen was then tested using the same motion sequence as previously described. The motion sequence and the range of motion capture was then completed at the following flexion angles: 0 deg,

30 deg, 60 deg, 90 deg. Once the motion sequence was completed for each flexion angle, a final test was performed with the simulator in force control mode and the knee flexion angle returned from 90 deg to 0 deg flexion while the tracking data was recorded. This entire protocol was then completed for the remaining femoral attachment locations of the virtual grafts (five total).

3.2.9 Data Analysis

3.2.9.1 Kinematic Analysis

Kinematic data representing the anterior tibial translation (ATT) (VIVO simulator output “X mm”) and internal rotation (IR) (VIVO simulator output “Z deg”) during the intact, virtual ACL, and destabilized conditions were measured using the simulator at 2000 Hz (500 samples/cycle). This data was then smoothed using a low-pass Butterworth filter with a cutoff frequency of 2 Hz. As the ATT and IR kinematic data was collected over the entire duration of the applied waveform, a separate function was written to determine the regions of the dataset corresponding to each motion. As the simulator applied the each of the three motions, three times (sequentially, as previously discussed), there were nine total regions. Regions 1, 4, and 7 were when the 89 N of anterior directed force was applied to the tibia (ATT regions), regions 2, 5, and 8 were when the 5 Nm internal rotation torque was applied to the tibia (IR regions), and regions 3, 6, and 9 were combination of anterior directed force and internal rotation was applied to the tibia (combined regions).

For the IR and Combined regions, due to the positioning of the specimen on the VIVO simulator, chirality determined the version of the algorithm that was used. For right knees, the peaks of the negative signal (filtered data) corresponded to the maximums of the data. These maximums were recorded within both the IR regions and the combined regions. For right knees, the minimums were then calculated by finding the intersection of the start of the IR regions with filtered data. For the combined minimums, these were located by taking the peaks of the positive data in between the recorded negative (maximum) peaks. For the left knees, the peaks of the positive signal corresponded to the maximums of the data. These maximums were recorded within both the IR regions and the combined regions. For left knees, the minimums were then calculated by finding the intersection of the start of the IR regions with filtered data. For the combined minimums, these were located by taking the peaks of the positive data in between the recorded negative (maximum) peaks. This analysis is summarized in Figure 15 and Figure 16, below. Note that in Figure 15 and Figure 16 on the y axis the sign is flipped (negative for right knee, positive for left knee). In both these graphs, an internal rotation

was performed on the specimen, only the chirality of the specimen changed, so the applied Z deg waveform was adjusted.

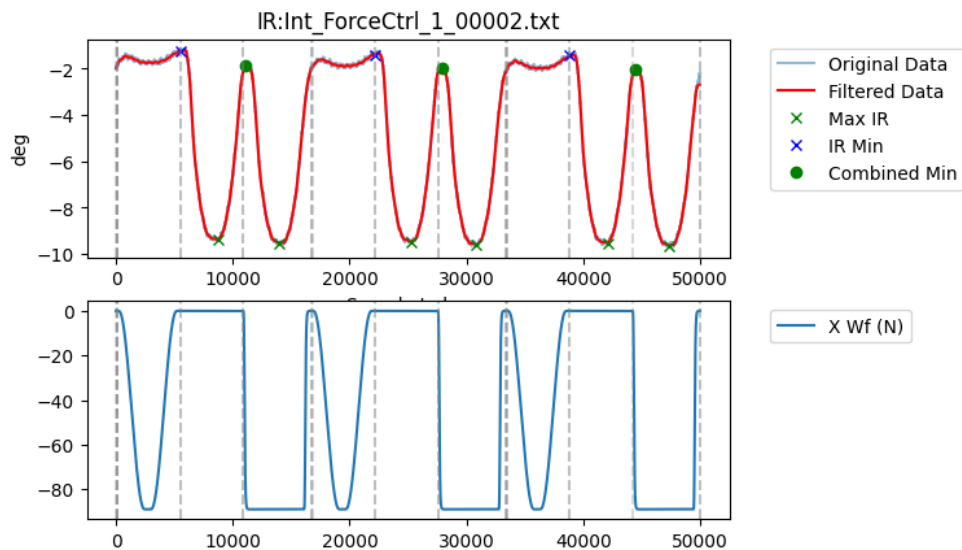


Figure 15. Overview of data recorded for the internal rotation and combined regions, right knee. The top graph is the recorded internal rotation range of motion, and the bottom graph represents one degree of freedom of the input waveform.

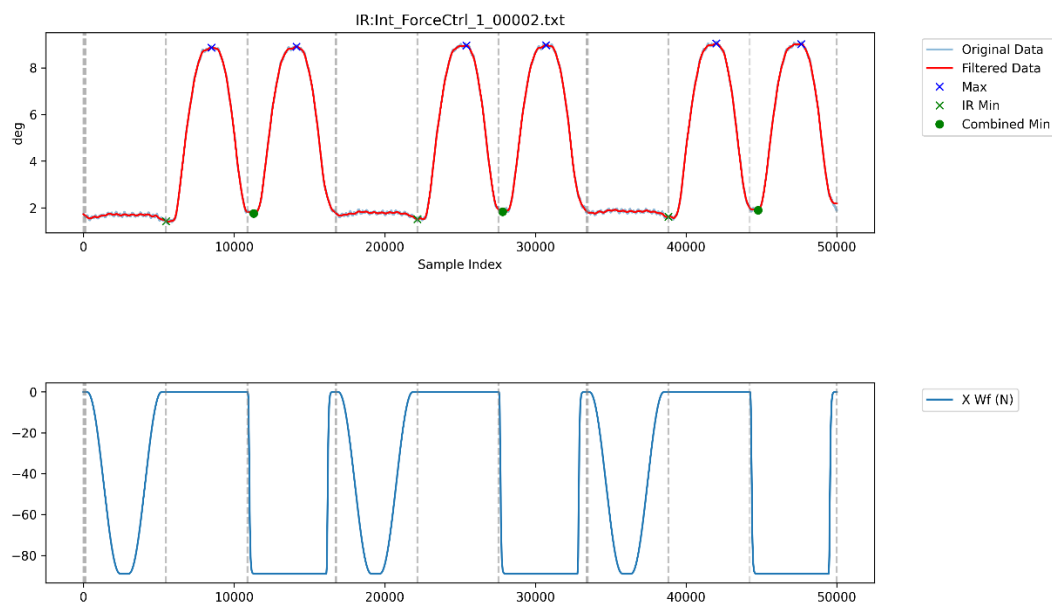


Figure 16. Overview of data recorded for the internal rotation and combined regions, left knee. The top graph is the recorded internal rotation range of motion, and the bottom graph represents one degree of freedom of the input waveform.

For the ATT regions, the peaks of the positive signal (filtered data) were recorded, and the peaks of the negative signal were recorded. The peaks of the positive signal corresponded to the minimums of the data, essentially the start of the ATT region before the maximum anterior directed force was applied. The peaks of the negative signal corresponded to the maximums of the data. Additionally, for the combined motion when the peak internal rotation was applied to the specimen, these indices were used to record the current ATT value at moment. This analysis is summarized, below, in Figure 17. As three maximums were recorded and two or three minimums were recorded (dependent on the specimen data), an average of the maximums and an average of the minimums were taken. From the averaged values, an ATT range (magnitude of the maximum subtracted by the minimum) was calculated to determine the range of anterior tibial translation. A full overview of the kinematic analysis algorithm is also included below, shown in Figure 18.

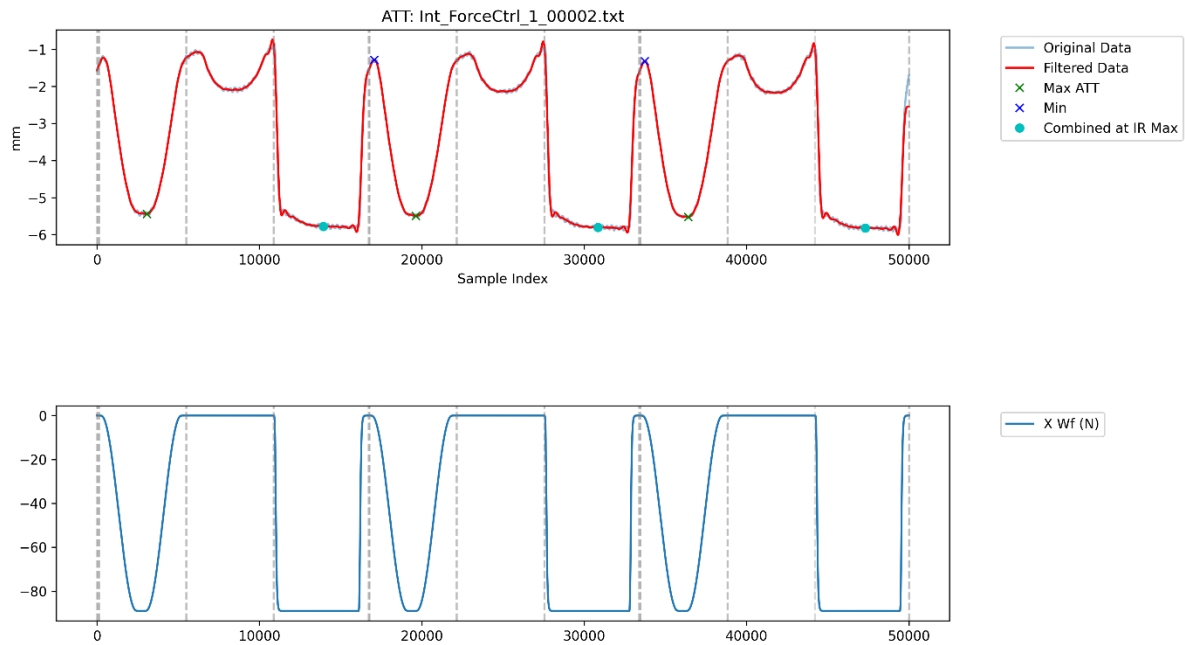


Figure 17. Overview of data recorded for the anterior tibial translation regions.

The top graph is the recorded translation range of motion, and the bottom graph represents one degree of freedom of the input waveform

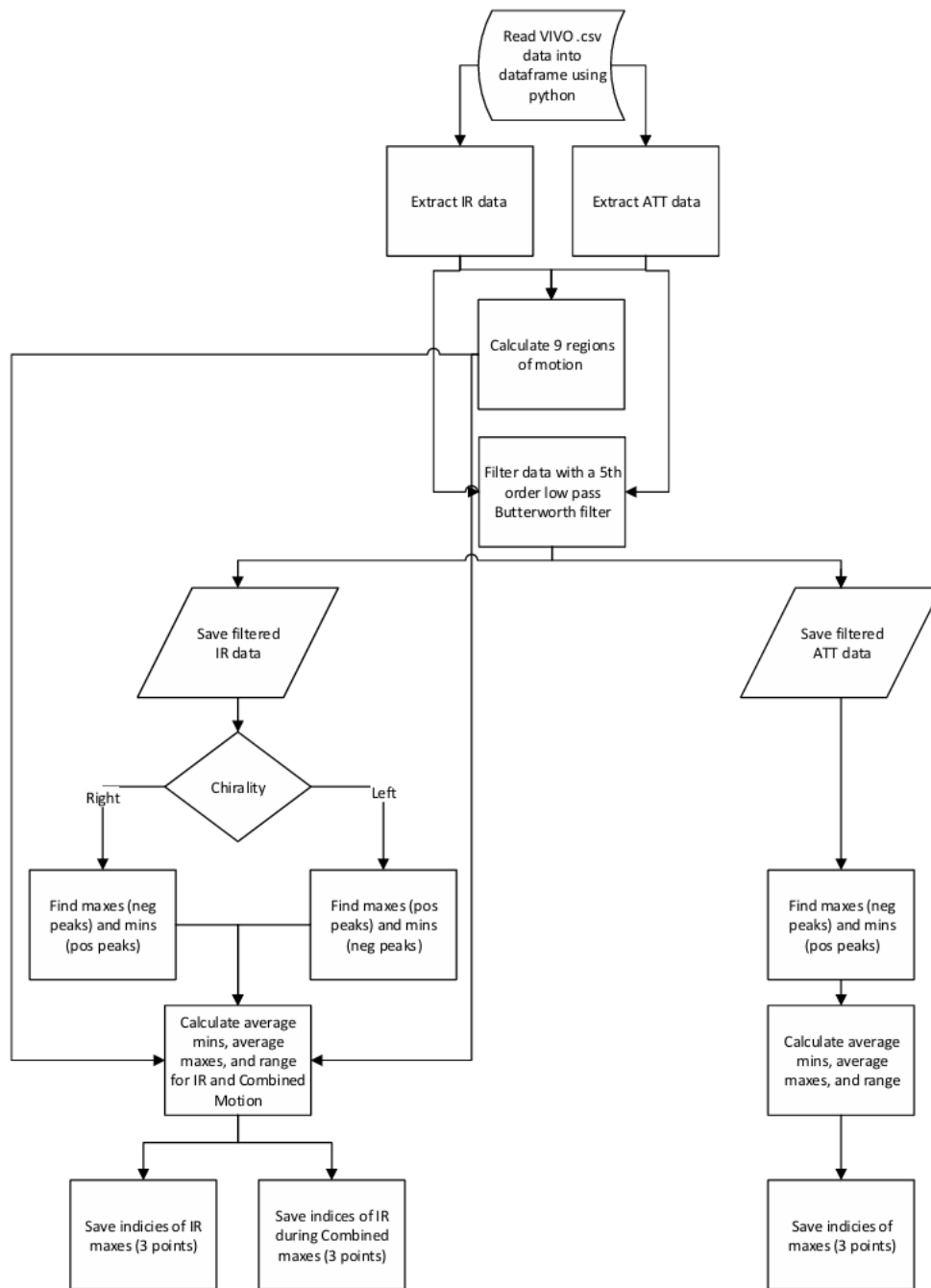


Figure 18. Kinematic analysis algorithm methodology overview.

Flow diagram depicts that the internal rotation and internal rotation during combined kinematic analysis is done together, and the anterior tibial translation analysis is done separately.

3.2.9.2 Force Analysis

Force data representing the F_x , F_y , and F_z during the intact, virtual ACL, and destabilized conditions were measured using the simulator at 2000 Hz (500 samples/cycle). The force analysis varied depending on the condition that was being tested on the specimen. For the virtual graft conditions, the recorded virtual graft force components (VIVO simulator output “ F_x Cs”, “ F_y Cs”, “ F_z Cs”) were all extracted from the dataframes made from the .csv files. To calculate the initial force (approximately when 0N of ATT force was applied), the first five datapoints of each force component was recorded and averaged (F_x Cs average, F_y Cs average, F_z Cs average). These averaged forces were then used to calculate the resultant initial force. After this, the entire data set of each force component was smoothed using a low-pass Butterworth filter with a cutoff frequency of 2 Hz. The indices (sample number) of the previously recorded kinematic maximums (ATT maximums, IR maximums, IR during combined maximums) were then used to find the corresponding force components at these nine indices. This is shown below in Figure 19. The force components corresponding to ATT maximums, IR maximums, and IR maximums during combined were all extracted, averaged (respectively), and then used to calculate each resultant force.

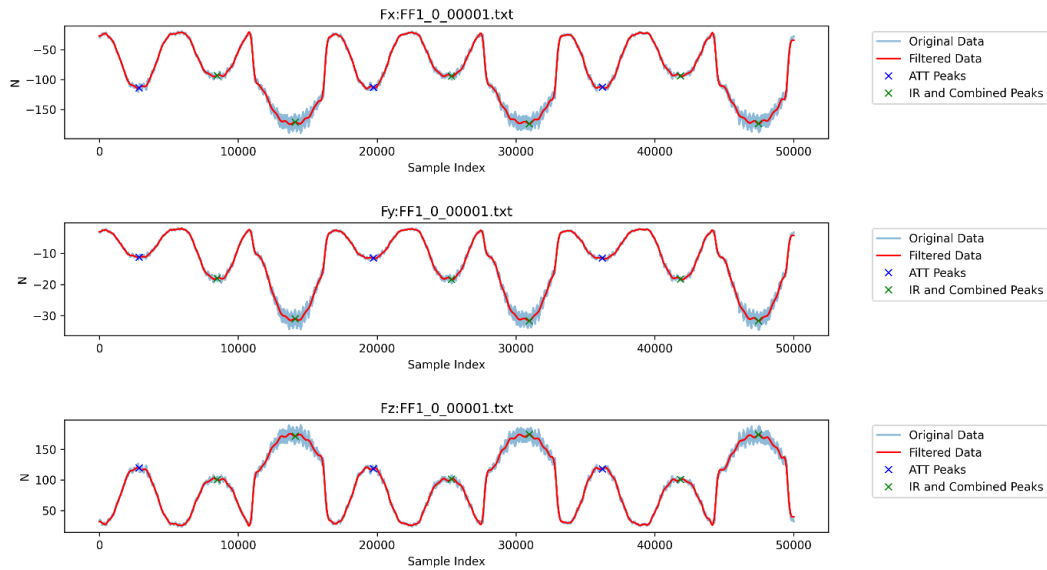


Figure 19. Force analysis process for virtual graft conditions.

The top graph shows the x component of the force, the middle graph shows the y component of the force, and the bottom graph shows the z component of the force. All three graphs depict the raw data and the filtered data. Anterior tibial translation kinematic peaks and internal rotation and internal rotation during combined kinematic peaks are labelled on each graph.

For the intact condition, the force was calculated differently. As the destabilized specimen condition was tested in two ways (displacement control and force control), the displacement control test of the destabilized condition could be leveraged to calculate the force carried by the ACL. Specifically, the difference in the recorded force component (Fx, Fy, and Fz) could be used to calculate the force carried by the ACL. The recorded force components (VIVO simulator output “Fx”, “Fy”, “Fz”) were all extracted from the dataframes made from the .csv files. To calculate the initial force (approximately when 0N of ATT force was applied), the first five datapoints of each force component for both the intact condition and the corresponding displacement control file of the destabilized conditions were recorded and averaged respectively (intact Fx average, intact Fy average, intact Fz average, destabilized Fx average, destabilized Fy average, destabilized Fz average). The average of each component was then calculated, and the intact component averages were then subtracted by the destabilized component average. The remainders of each component were then used to take the resultant, representing the initial force carried by the ACL. After this, the entire data set of each force component was smoothed using a low-pass Butterworth filter with a cutoff frequency of 2 Hz for both the intact and the destabilized components. The indices (sample number) of the previously recorded kinematic maximums (ATT maximums, IR maximums, IR during combined maximums) were then used to find the corresponding force components at these nine indices in the intact files. For the displacement control data files of the destabilized conditions, there was no previously recorded kinematic data. These files were therefore sent through the same analysis methodology as detailed in the previous kinematic analysis section. Once the indices for the kinematic maximums were calculated, the corresponding force components could then be determined for the destabilized condition. Each intact force component was then subtracted by the corresponding destabilized condition force component for each of the motions (ATT, IR, Combined). The resultants were then calculated to represent the force carried by the ACL during the applied anterior force, internal rotation torque, and the combined motion. An example of the recorded intact force component data is shown in Figure 20. An overview of the entire force analysis process is shown in Figure 21.

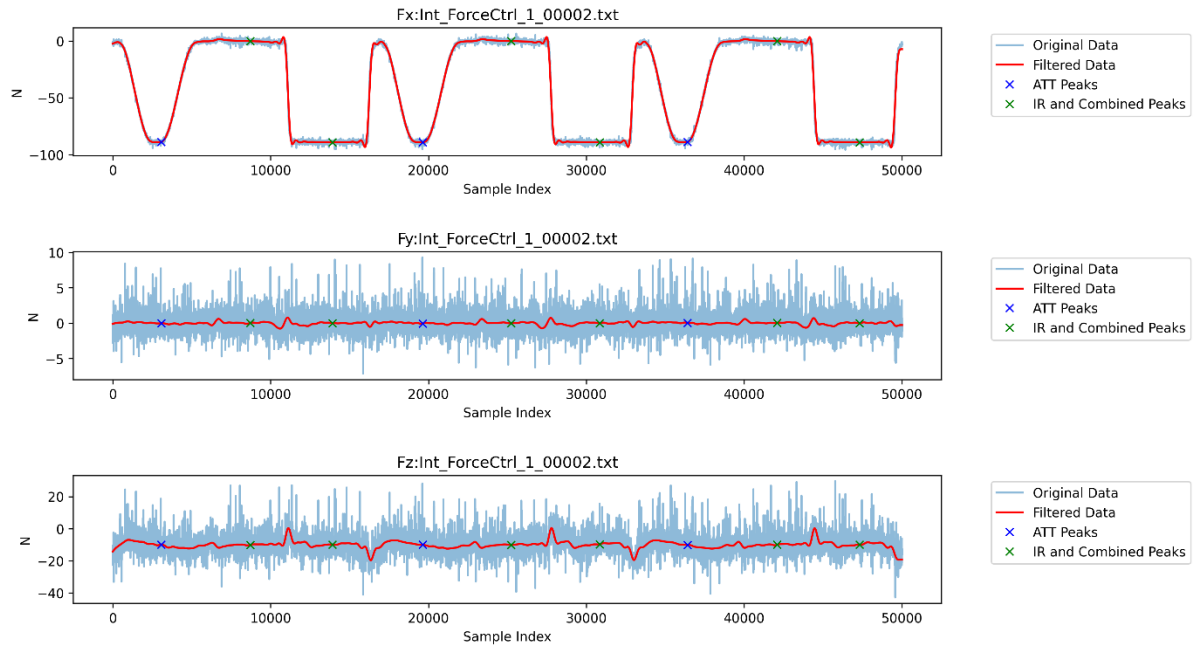


Figure 20. Intact force component data.

The top graph shows the x component of the force, the middle graph shows the y component of the force, and the bottom graph shows the z component of the force. All three graphs depict the raw data and the filtered data. Anterior tibial translation kinematic peaks and internal rotation and internal rotation during combined kinematic peaks are labelled on each graph.

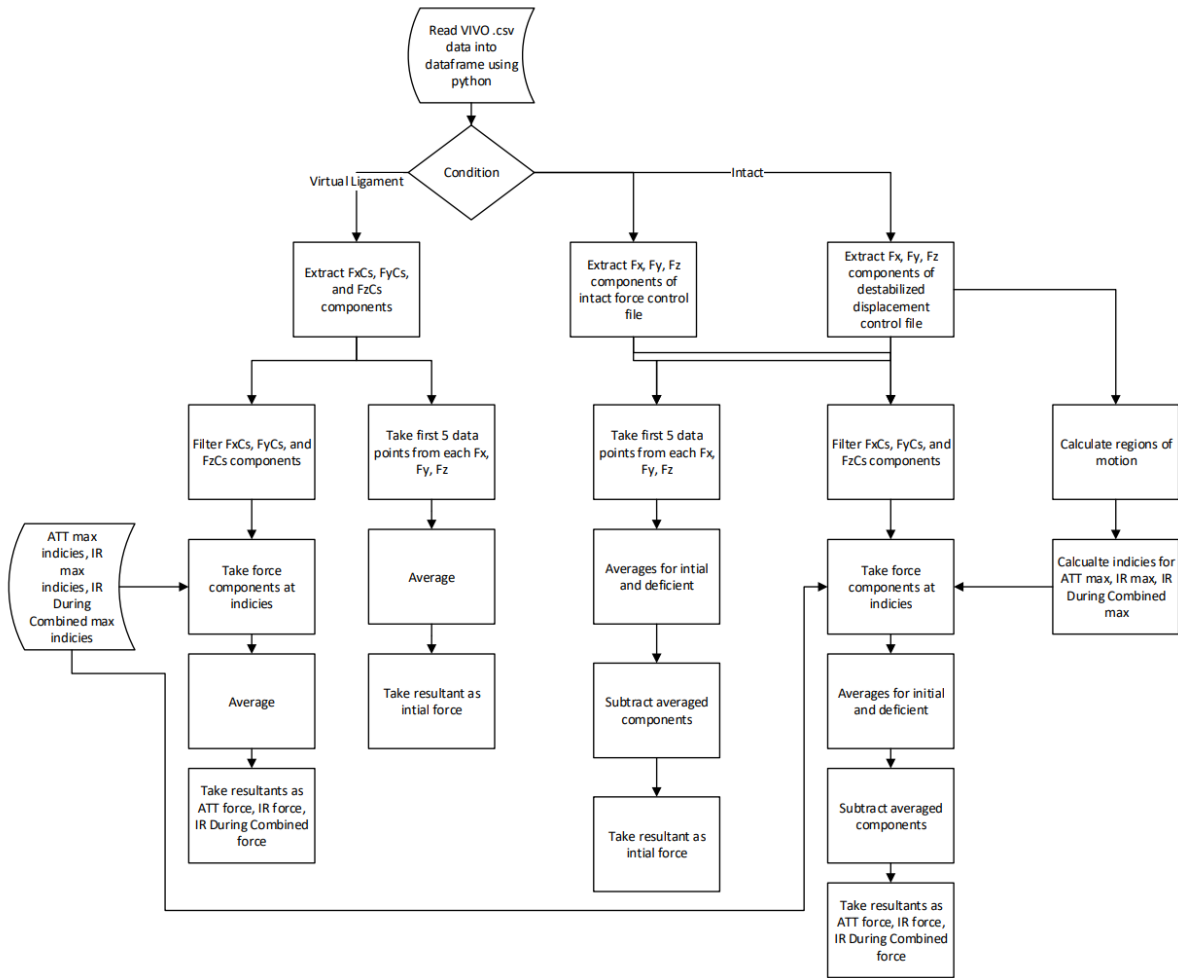


Figure 21. Force analysis algorithm methodology overview.

Flow diagram depicts that the virtual ligament forces are calculated first and then the force data during the intact condition is calculated. The force data for the destabilized condition is calculated last and then used estimate the force carried by the intact anterior cruciate ligament (ACL) by subtracting the force components during the intact condition and during the displacement control destabilized condition.

3.2.9.3 Virtual Graft Length Change Analysis

To calculate the virtual graft length change vs. flexion angle, there were two stages to the analysis: graft length calculation and flexion angle calculation. For the line length calculation, the fcsv files of the femoral tunnel positions and the fcsv file for the tibial tunnel position are used to get the initial line length between each femoral tunnel and the tibia tunnel (each initial graft length). Optotrak Certus tracking data recorded during each test contained the transforms the rigid body trackers attached to the

femur and the tibia. The femoral tunnel points and tibia tunnel point were transformed by the recorded transforms of each tracker. For each new transform, the line length (graft length) for each virtual ACL was calculated. This process was done in Matlab. In addition to the virtual graft length, the zero load length (slack length) for each femoral tunnel location of each specimen was calculated. This was calculated through Equation 3, where the reference length was calculated as the length between the femur and tibia insertion coordinates in the reference pose and reference strain was recorded during the virtual graft insertion process. These values were calculated in Excel.

To calculate the flexion angle, each specimen's femur and tibia bone models were first annotated according to a modified Grood and Suntay convention [37]. On the femur, the most posterior points on the femoral condyles are selected, the most distal point on the posterior surface on the femur between the medial and lateral femoral condyles is selected, and the center of the femoral shaft were selected [37]. On the tibia the most medial and lateral part of tibial plateau, the midpoint of the tibial plateau, and the center of the tibial shaft were selected [37]. These landmarks are used to calculate the origin, i, j, and k axis for both the femur and tibia (unit vectors). The unit vectors are then transformed by the recorded femur and tibia marker transforms. The Grood and Suntay e_1 and e_3 base vectors are then calculated and used to calculate the e_2 base vector. The Grood and Suntay e_1^r and e_2 base vectors are then used to calculate the knee flexion angle, where in Matlab flexion is calculated through Equation 4 and 5.

$$flexion = atan2d(-n * \|(e_1^r \times e_2)\|, (e_1^r \cdot e_2)) \quad (4)$$

$$n = \frac{(e_1^r \times e_2) \cdot e_1}{\|(e_1^r \times e_2) \cdot e_1\|} \quad (5)$$

Once the knee flexion angle was calculated for each transform, this data was saved into a .csv file along with the corresponding graft length values for each transform. This was also calculated in Matlab.

All the .csv files of each virtual graft's length at each flexion angle was read into a dataframe in python. The flexion angle data was first normalized so that the first value of the dataset became zero degrees. This was done as the tracking data was only recorded once the specimen was in the reference pose (0 degrees) and due to the manual nature of the Grood and Suntay landmark annotation process, not all the recorded flexion angles started at 0 degrees, even though through the experimental setup protocol the specimen was at 0 degrees. Once the flexion angle data was normalized, the flexion angle data vs. sample number was plotted, this was used to identify the section of the data corresponding to

90 degrees flexion extending to 0 degrees. This portion of the data was cropped out into a separate data frame and then was downsampled so that only data from 90 degrees to 0 degrees, in 5-degree increments, was saved. The downsampling process involved searching for the closest angle in the overall data to the desired angle increments from the downsampled increments list of angles. If an angle was found within 2.5 degrees of the target angle, that line length was added to the dataframe. If no angle was found within the threshold, then no value was added to the dataframe. If there was a missing value for one of the downsampled increments, then the missing value was interpolated in python (linear interpolation). Each downsampled dataset for each virtual graft condition was then plotted against the knee flexion angle and the respective zero load length (slack length) was plotted on the same graph, for each specimen. An average plot of the graft lengths across all the specimens was then produced. The individual specimen figures with the labelled zero load length are shown in Appendix B.

3.2.10 Statistical Analysis

For each of the three analyses, a linear mixed-effects model (LMM) was used to assess the effects of condition, knee flexion angle, and their interaction on the measurements specific to each analysis. The model included “Condition”, “Angle”, and their interaction (“Condition x Angle”) as fixed effects. In order to account for repeated measures and the partial pairing of knees from the same cadavers, “Specimen” was modeled as a random effect nested within “Cadaver”. However, when the variance component for “Cadaver” was estimated as zero, thus showing no meaningful variability, this random effect was removed from the model to avoid singularity and improve model stability. To evaluate the statistical significance of fixed effects, a Type III ANOVA was performed on the fitted LMM. When a significant main or interaction effect was observed ($\alpha = 0.05$), Tukey-adjusted post hoc comparisons of estimated marginal means (EMMs) were conducted to determine which specific conditions differed.

For the kinematic analysis, the measurements included ATT Range, IR Max, IR Max during Combined. The conditions included intact, destabilized, and the five virtual graft conditions (ideal, distal, proximal, anterior, posterior). For the force analysis, the measurements included the initial force, resultant force during ATT, resultant force during IR, and the resultant force during the combined motion. The conditions included intact and the five virtual graft conditions. For the virtual graft length change analysis, the measurements included the change in graft length at each angle in 5-degree increments from 0 degrees to 90 degrees. The conditions included the five virtual graft conditions.

3.3 Results

3.3.1 Omitted Specimens

Post-hoc analysis of the specimen data revealed some inconsistencies in the knee laxity following ACL removal, leading to some uncertainty regarding the full removal of the ligament under arthroscopic visualization. As a check to ensure that the experimental methodology had been completed properly, each specimen's maximum ATT (mm) during the applied anterior directed load was evaluated in the intact condition and compared to the destabilized condition at 0 degrees, 30 degrees, and 90 degrees. As the ACL plays a significant role in preventing ATT, there should be a large increase in maximum ATT in the destabilized condition when compared to the intact condition. When performing the Lachman test at 30 degrees knee flexion on an intact knee, there should be less than 2 mm of ATT, as over 2mm of ATT is considered positive for the Lachman test [112]. When performing the Lachman test at 30 degrees knee flexion, a Grade 1 ACL tear will have 3-5mm ATT, a Grade 2 ACL tear will have 5-10mm ATT, and a Grade 3 ACL tear will have over 10mm of ATT [112], [113]. If an intact knee had 3mm of ATT, for Grade 1/2, there is a 66% increase (5mm of ATT) and for a Grade 2/3, there is a 233% increase (10 mm of ATT).

Table 3 shows the percentage increase in ATT max in the destabilized condition compared to the intact condition for all the tested specimens. In the table below, any percent increase less than 66% was identified at 30 degrees knee flexion. This corresponded to specimen 1, 2, and 6. As there was only a minor change between the intact and destabilized conditions for these specimens, this showed that an error occurred during the experimental methodology where the ACL likely was not fully disrupted. As such, these three specimens were omitted from the remaining analysis.

Table 3. Percent Increase in ATT Max in the Destabilized Condition from Intact

Specimen	0 Degrees	30 Degrees	90 Degree
1_UTN24-26250L	-3%	9%*	10%
2_UTN24-26250R	12%	25%*	23%
3_UTN24-26236R	142%	124%	36%
4_UTN24-26224L	52%	96%	82%
5_UTN24-26058L	53%	88%	33%
6_UTN24-26224R	-4%	46%*	19%
7_UTN24-26058R	153%	203%	38%
8_UTN24_26257R	89%	197%	106%
9_UTN24_26257L	62%	128%	101%

3.3.2 Kinematic Analysis

For the kinematic analysis, knee laxity metrics were extracted for each specimen. These values included: ATT min, ATT max, ATT range, IR min, IR max, IR range, IR min during combined, IR max during combined, and IR range during combined. From these metrics, this section will include figures for ATT range, IR max, and IR max during combined. Additional figures showing the maximums and minimums are shown in Appendix A. ATT range was chosen as the virtual graft conditions tended to bring the tibia into a posterior position. This limited the maximum anterior tibial translation that was then able to be achieved in the virtual graft conditions. As such, ATT range offered a less skewed way to compare the results across specimens. For IR, the neutral position of the knee posed a strong effect on the IR range and IR range during combined values. In the destabilized condition, the neutral position shifted into an externally rotated state which confounded the range value for some specimens. As such, the IR max and IR max during combined were chosen for this section. Additional figures showing the average ATT min, ATT max, IR min, IR max, IR min during combined, and IR max during combined are shown in Appendix C.

3.3.2.1 Anterior Tibial Translation Range

Both the maximum and minimum ATT values were calculated from the start of the ATT motion and the peak of the ATT motion. These values were then used to calculate the ATT Range. A Type III ANOVA revealed a significant main effect of Condition ($p = 1.83 \times 10^{-44}$). There was also a significant main effect of Angle ($p = 2.00 \times 10^{-11}$), as well as a significant Condition $\times$ Angle interaction ($p = 5.50$

$\times 10^{-10}$). As the primary focus of the present study was on the effect of Condition, post hoc analyses were conducted only for Condition, and the effects of Angle and the interaction were not further explored. Figure 22, Figure 23, and Figure 24 depict the average ATT range (across all the included specimens) by each condition (intact, destabilized, ideal, distal, proximal, anterior, posterior) at 0 degrees, 30 degrees, and 90 degrees. For these figures, the significant differences were calculated and shown. At 0 degrees of knee flexion, the significant differences were between all the virtual grafts and the destabilized condition, the intact condition and some of the virtual grafts (ideal, distal, posterior) conditions, and the intact and the destabilized condition. At 30 degrees of knee flexion, the significant differences were between the intact condition and all the virtual graft conditions, between the destabilized condition and all the virtual graft conditions, and between the intact and destabilized condition. At 90 degrees of knee flexion, there were significant differences between the destabilized condition and all the virtual graft conditions, the intact and some of the virtual graft conditions (distal, proximal, anterior), the intact and the destabilized conditions, between the posterior and anterior virtual graft conditions, and between the posterior and proximal virtual graft conditions.

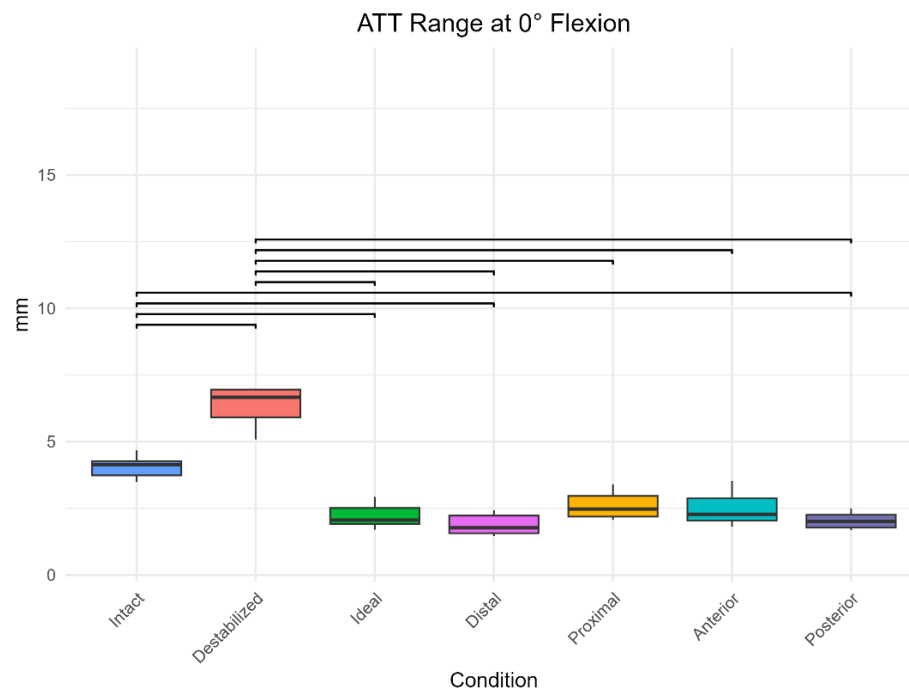


Figure 22. ATT range by condition at 0 degrees knee flexion, with significant differences labelled. The anterior tibial translation (ATT) range was measured under 89N of applied loading for intact, removal of the anterior cruciate ligament (ACL) (destabilized), and the five virtual graft conditions at 0 degrees knee flexion.

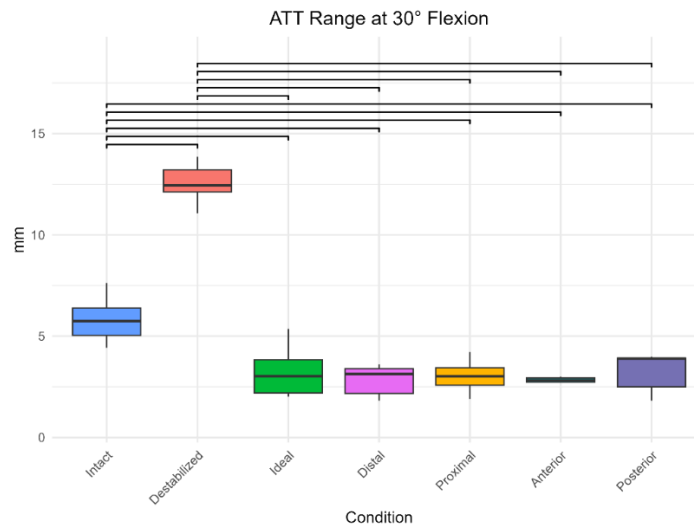


Figure 23. ATT range by condition at 30 degrees knee flexion, with significant differences labelled.

The anterior tibial translation (ATT) range was measured under 89N of applied loading for intact, removal of the anterior cruciate ligament (ACL) (destabilized), and the five virtual graft conditions at 30 degrees knee flexion.

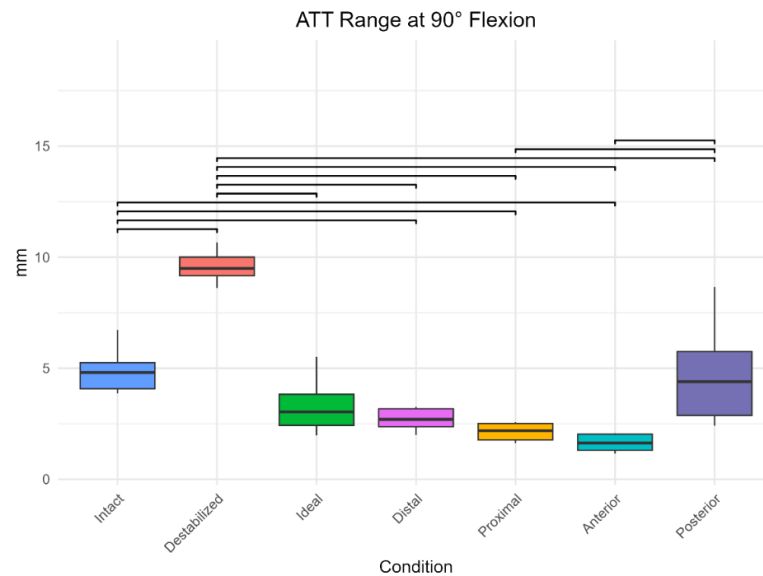


Figure 24. ATT range by condition at 90 degrees knee flexion, with significant differences labelled.

The anterior tibial translation (ATT) range was measured under 89N of applied loading for intact, removal of the anterior cruciate ligament (ACL) (destabilized), and the five virtual graft conditions at 90 degrees knee flexion.

3.3.2.2 Internal Rotation Max

Figure 25, Figure 26, Figure 27 depict the average IR max (across all the included specimens) by each condition (intact, destabilized, ideal, distal, proximal, anterior, posterior) at 0 degrees, 30 degrees, and 90 degrees. The Type III ANOVA conducted on the LMM revealed that the effect of Condition was not statistically significant, suggesting that there were no significant differences in the dependent variable across the different condition levels. Additionally, the effect of Angle and the Condition x Angle interaction were not statistically significant as well. Therefore, no post hoc pairwise comparisons were performed. Table 4 includes the percent increase between the destabilized condition from the intact condition for all included specimens at each knee flexion angle.

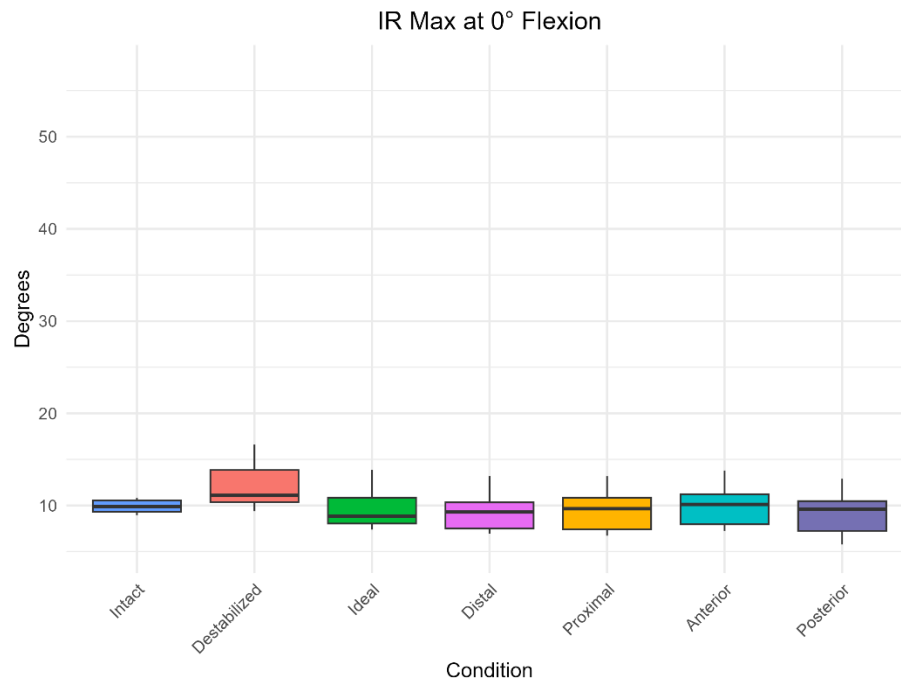


Figure 25. IR max by condition at 0 degrees knee flexion.

The maximum internal rotation (IR) was measured under 5Nm of applied loading for intact, removal of the anterior cruciate ligament (ACL) (destabilized), and the five virtual graft conditions at 0 degrees knee flexion.

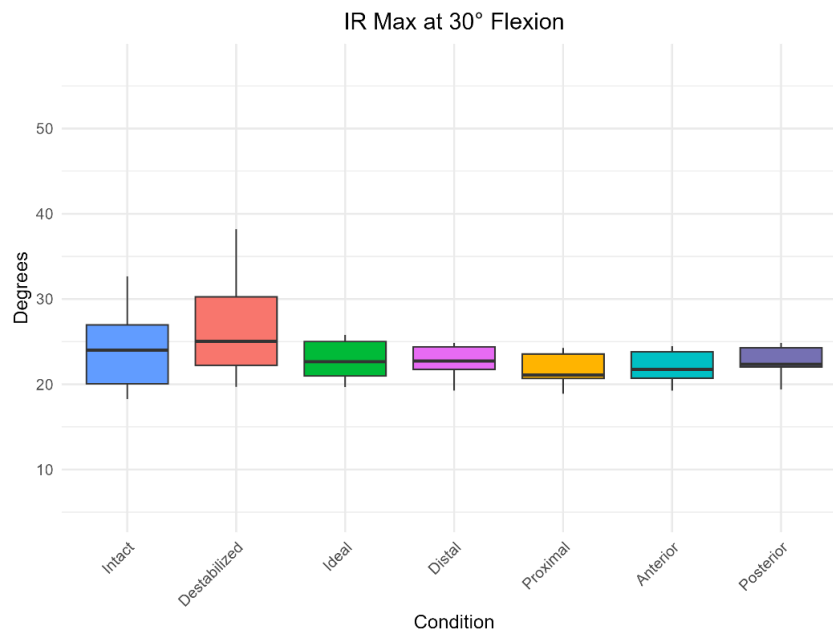


Figure 26. IR max by condition at 30 degrees knee flexion.

The maximum internal rotation (IR) was measured under 5Nm of applied loading for intact, removal of the anterior cruciate ligament (ACL) (destabilized), and the five virtual graft conditions at 30 degrees knee flexion.

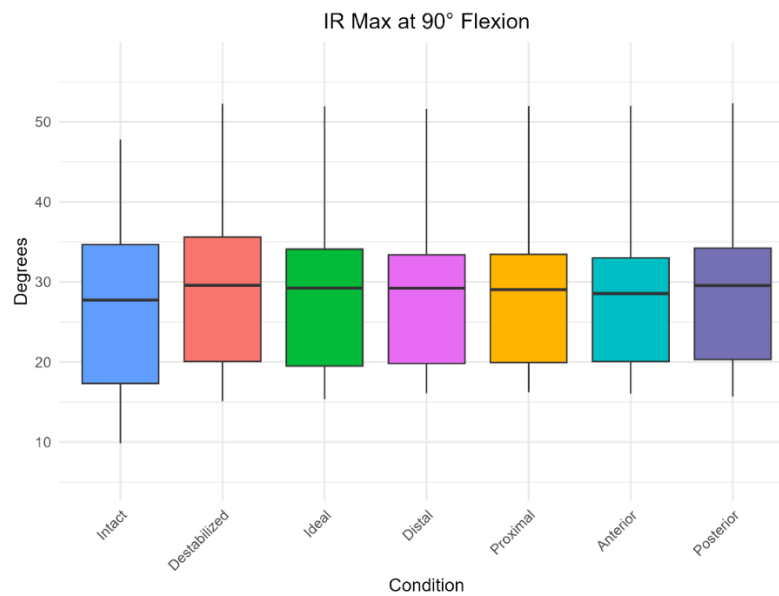


Figure 27. IR max by condition at 90 degrees knee flexion.

The maximum internal rotation (IR) was measured under 5Nm of applied loading for intact, removal of the anterior cruciate ligament (ACL) (destabilized), and the five virtual graft conditions at 90 degrees knee flexion.

Table 4. Percent Increase in IR Max in the Destabilized Condition from Intact

Specimen	0 Degrees	30 Degrees	90 Degree
3_UTN24-26236R	9%	3%	5%
4_UTN24-26224L	16%	6%	9%
5_UTN24-26058L	14%	14%	72%
7_UTN24-26058R	1%	8%	8%
8_UTN24_26257R	37%	14%	2%
9_UTN24_26257L	53%	17%	9%

3.3.2.3 Internal Rotation Range During Combined

Figure 28, Figure 29, and Figure 30 depict the average IR Max during Combined (across all the included specimens) by each condition (intact, destabilized, ideal, distal, proximal, anterior, posterior) at 0 degrees, 30 degrees, and 90 degrees. From the Type III ANOVA conducted on the LMM, it was clear that the effect of Condition was not statistically significant. Additionally, the main effect of Angle and the Condition x Angle interaction were not statistically significant as well. As such, post hoc pairwise comparisons were not performed. Table 5 shows the percentage increase in IR max during combined from the intact state for every included specimen at each knee flexion angle.

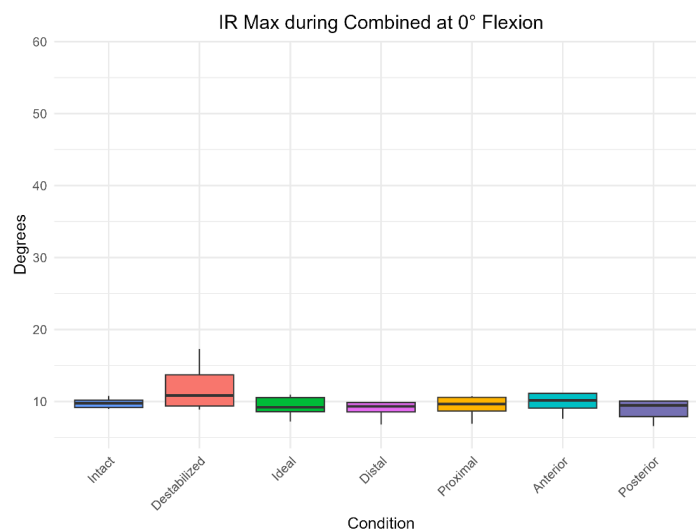


Figure 28. IR Max during Combined by condition at 0 degrees knee flexion, with significant differences labelled.

The maximum internal rotation (IR) during the combined motion was measured under 5Nm of applied loading for intact, removal of the anterior cruciate ligament (ACL) (destabilized), and the five virtual graft conditions at 0 degrees knee flexion.

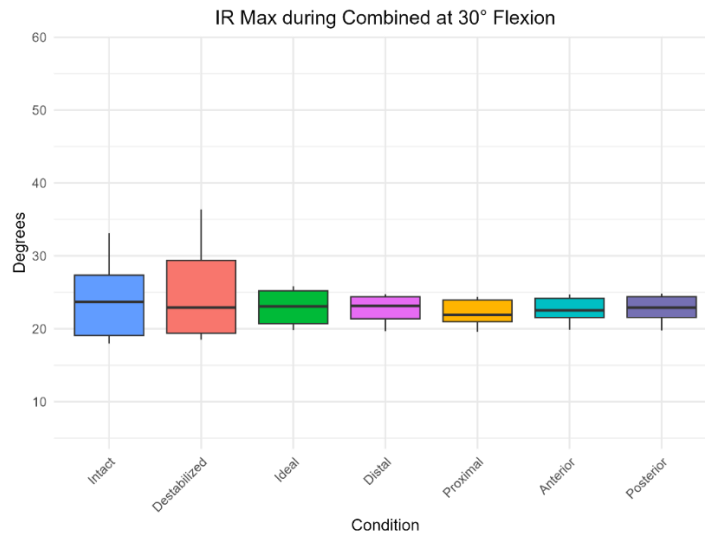


Figure 29. IR Max During Combined by condition at 30 degrees knee flexion, with significant differences labelled.

The maximum internal rotation (IR) during the combined motion was measured under 5Nm of applied loading for intact, removal of the anterior cruciate ligament (ACL) (destabilized), and the five virtual graft conditions at 30 degrees knee flexion.

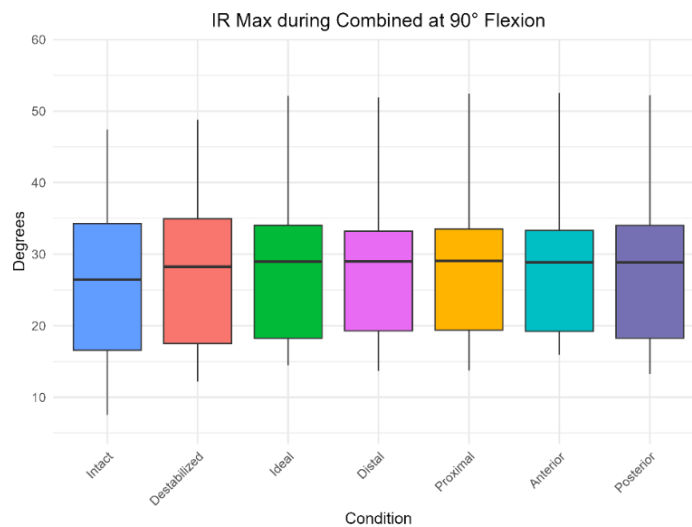


Figure 30. IR Max During Combined by condition at 90 degrees knee flexion, with significant differences labelled.

The maximum internal rotation (IR) during the combined motion was measured under 5Nm of applied loading for intact, removal of the anterior cruciate ligament (ACL) (destabilized), and the five virtual graft conditions at 90 degrees knee flexion.

Table 5. Percent Increase in IR Max During Combined in the Destabilized Condition from Intact

Specimen	0 Degrees	30 Degrees	90 Degree
3_UTN24-26236R	-7%	-10%	3%
4_UTN24-26224L	18%	3%	10%
5_UTN24-26058L	12%	2%	62%
7_UTN24-26058R	-1%	7%	6%
8_UTN24_26257R	42%	9%	0%
9_UTN24_26257L	61%	10%	3%

3.3.3 Force Analysis

For the force analysis, four different linear mixed models were produced for each type of force data (initial force, peak force during ATT, peak force during IR, and peak force during combined) for the virtual grafts as well as an estimate of the force carried by the intact ACL. Three plots were produced, one per knee flexion angle, with all the significant differences labelled. These are shown below in Figure 31, Figure 32, and Figure 33.

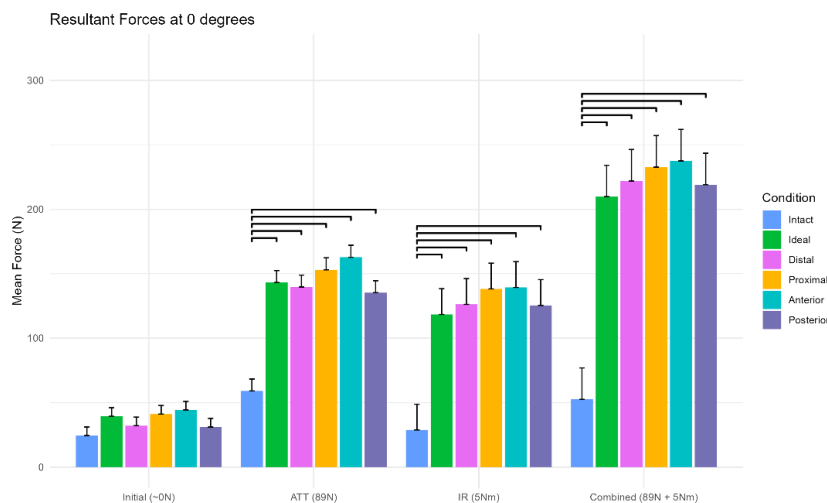


Figure 31. Resultant forces by condition at 0 degrees knee flexion.

Results are grouped by the four force types (initial force, force during peak anterior tibia translation (ATT), force during peak internal rotation (IR), and force during peak internal rotation (IR) during the combined motion) with significant differences labelled.

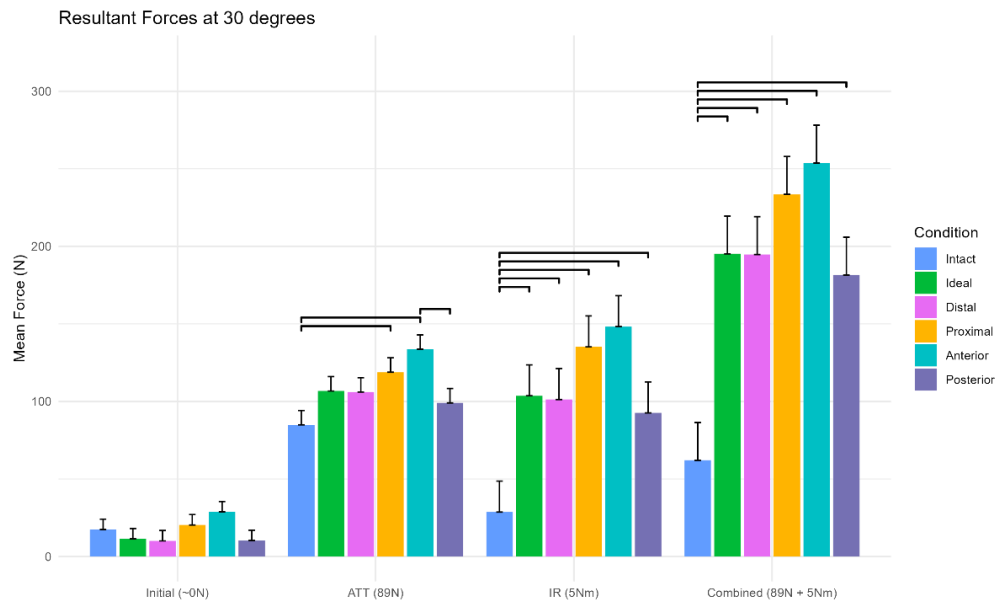


Figure 32. Resultant forces by condition at 30 degrees knee flexion.

Results are grouped by the four force types (initial force, force during peak anterior tibia translation (ATT), force during peak internal rotation (IR), and force during peak internal rotation (IR) during the combined motion) with significant differences labelled.

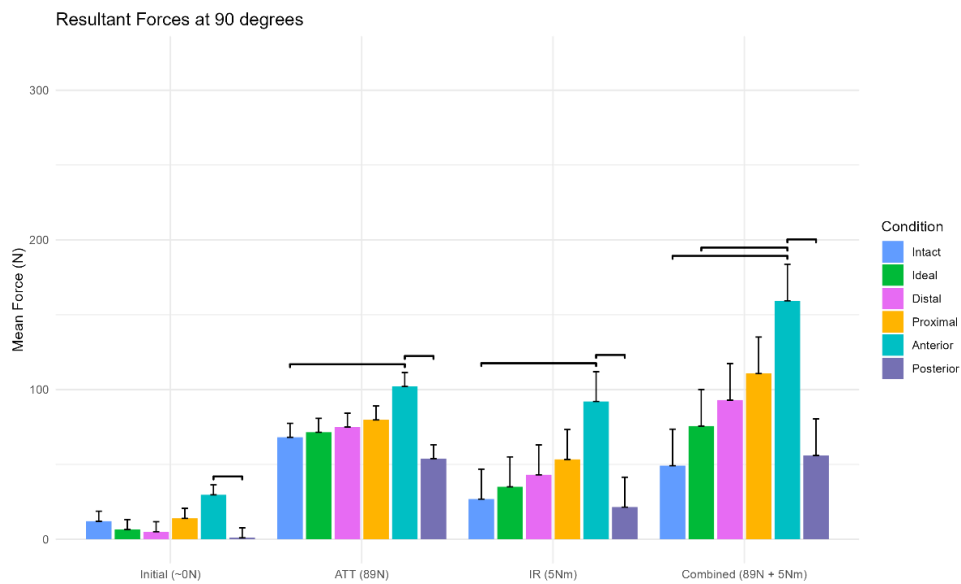


Figure 33. Resultant forces by condition at 90 degrees knee flexion.

Results are grouped by the four force types (initial force, force during peak anterior tibia translation (ATT), force during peak internal rotation (IR), and force during peak internal rotation (IR) during the combined motion) with significant differences labelled.

3.3.3.1 Initial Forces

The initial force is the resultant force of the Fx, Fy, and Fz force components based on an average of the first five data points recorded. When performing the statistical analysis for this portion of the data, “Specimen” was modeled as a random effect nested within “Cadaver”. In this model, though, the variance component for “Cadaver” was estimated as zero. Thus, for the Initial Force analysis, “Specimen” was unnested to avoid singularity in the LMM. From the Type III ANOVA conducted on the fitted model, it was clear that the effect of Condition was statistically significant ($p = 0.0022$). The main effect of Angle and the interaction effect of Condition $\times$ Angle were not statistically significant. Therefore, post hoc pairwise comparisons were only conducted for Condition to investigate which differences were significant. Specifically, there was only one significant difference and that was at 90 degrees knee flexion between the anterior and posterior virtual graft conditions.

3.3.3.2 Force During Anterior Tibial Translation

The peak force during ATT is the resultant force at the same timepoint of the maximum recorded ATT. A Type III ANOVA was conducted on the fitted LMM. The ANOVA revealed a significant main effect of Condition ($p = 2.55 \times 10^{-14}$). There was also a significant main effect of Angle ($p = 5.95 \times 10^{-21}$), as well as a significant Condition $\times$ Angle interaction ($p = 7.47 \times 10^{-6}$). Since the primary focus of the study was on the effect of Condition, post hoc analyses were conducted only for Condition, and the effects of Angle and the interaction were not further explored. At 0 degrees knee flexion, there were significant differences between the intact condition and every virtual graft condition. At 30 degrees knee flexion, there were significant differences between intact and the proximal virtual graft condition, intact and the anterior virtual graft condition, and between the anterior and posterior virtual graft conditions. At 90 degrees knee flexion, there were significant differences between intact and the anterior virtual grafts condition and the anterior and posterior virtual graft conditions.

3.3.3.3 Force During Internal Rotation

The peak force during IR is the resultant force at the same timepoint of the maximum recorded IR. A Type III ANOVA performed on the fitted LMM. The ANOVA revealed a significant main effect of Condition ($p = 3.88 \times 10^{-10}$). There was also a significant main effect of Angle ($p = 1.93 \times 10^{-11}$). There was no significant Condition $\times$ Angle interaction though. Post hoc analyses were conducted only for Condition, and the effect of Angle was not further explored. At 0- and 30-degrees knee flexion, there were significant differences between intact and every virtual graft condition. At 90 degrees knee

flexion, there were significant differences between intact and the anterior virtual graft condition and between the anterior and posterior virtual graft conditions.

3.3.3.4 Force During Combined Motion

The peak force during Combined is the resultant force at the same timepoint of the maximum recorded Combined motion. The ANOVA revealed a significant main effect of Condition ($p = 8.82 \times 10^{-16}$). There was also a significant main effect of Angle ($p = 2.44 \times 10^{-16}$), as well as a significant Condition $\times$ Angle interaction ($p = 0.0098$). As the primary focus of the study was on the effect of Condition, post hoc analyses were conducted only for Condition, and the effects of Angle and the interaction were not further explored. At 0- and 30-degrees knee flexion, there were significant differences between intact and every virtual graft condition. At 90 degrees knee flexion, there were significant differences between intact and the anterior virtual graft condition, between the anterior and ideal virtual graft conditions, and between the anterior and posterior virtual graft conditions.

3.3.4 Virtual Graft Length Change

3.3.4.1 Average Across Specimens

To understand the isometry of five different virtual graft conditions, virtual graft length vs. flexion angle was calculated. Figure 34 shows the average graft length vs. knee flexion angle for each of the conditions. Figure 35 shows the same data, but plotted as a change in graft length in mm from the zero-degree graft length. This graph clearly indicates that the anteriorly placed femoral tunnel created the most isometric graft. Additionally, the posterior graft has the greatest change in length across the recorded knee range of motion. Figure 37 depicts the maximum graft length change in mm for each of the virtual graft conditions. This shows that the anterior femoral tunnel has the smallest maximum graft length change, while the posterior femoral tunnel has the greatest maximum length change. As well, Figure 38 depicts the averaged initial resultant forces by each angle of each virtual graft condition. This figure shows that the posterior and distal grafts carried the least amount of force over the tested knee range of motion.

A LMM was used to investigate the change in graft length by knee flexion angle. A Type III ANOVA was performed on the fitted LMM. The ANOVA revealed a significant main effect of Condition ($p = 4.32 \times 10^{-67}$). There was also a significant main effect of Angle ($p = 4.20 \times 10^{-118}$), as well as a significant Condition $\times$ Angle interaction ($p = 8.64 \times 10^{-12}$). As the primary focus of the study

was on the effect of Condition, post hoc analyses were conducted only for Condition, and the effects of Angle and the interaction were not further explored. Figure 36 depicts the same information as Figure 35, but with 5 separate subplots for each condition with the significant differences labelled. There are 63 significant differences, 41 of which are significant differences between a virtual graft condition and the posterior virtual graft data. The remaining 22 significant differences are between a virtual graft condition and the anterior virtual graft data. All the significant differences occurred after 35 degrees of knee flexion, but most differences (46) occurred between 60 and 90 degrees of flexion.

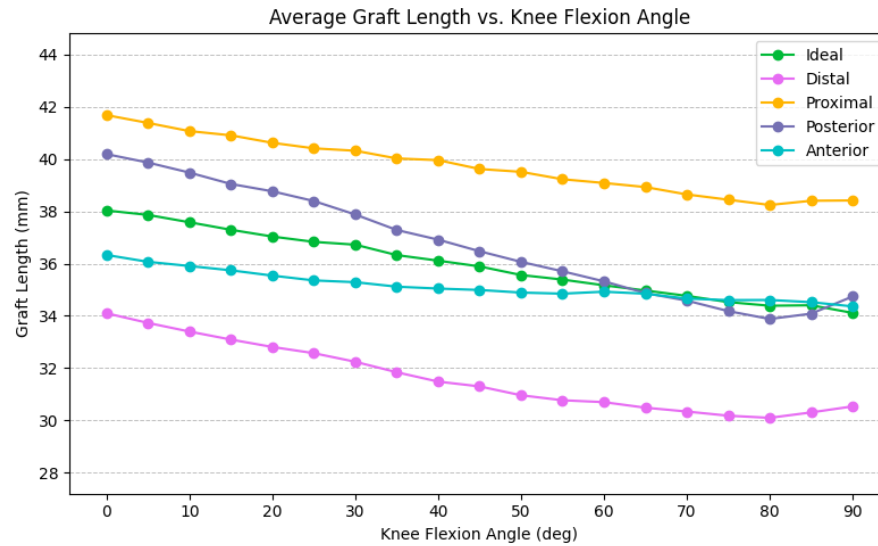


Figure 34. Average graft length vs. knee flexion angle.

The average graft length was measured during knee flexion-extension range of motion for the five virtual graft conditions.

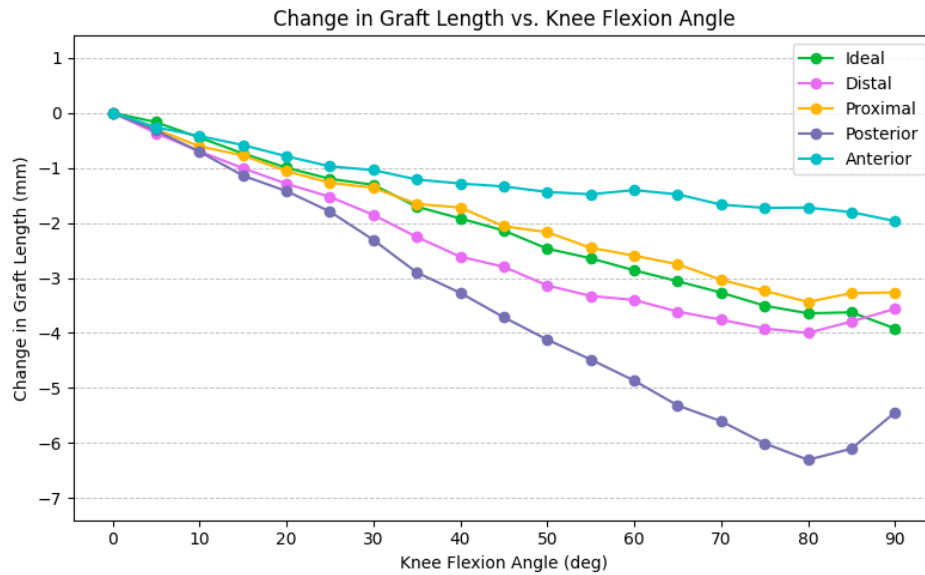


Figure 35. Average change in graft length vs. knee flexion angle.

The average change in graft length was measured during knee flexion-extension range of motion for the five virtual graft conditions.

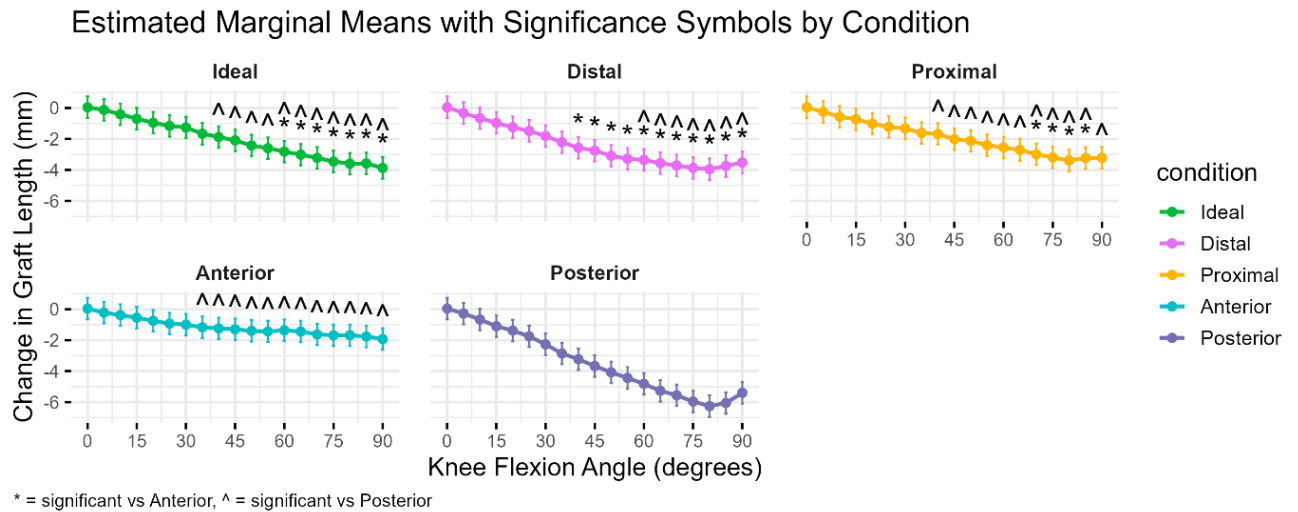


Figure 36. Average change in graft length vs. knee flexion angle.

*Each virtual graft condition (ideal, distal, proximal, anterior, posterior) is shown in its own graph, with significant differences labelled where * signifies a significant difference between the anterior virtual graft condition and ^ signifies a significant difference between the posterior virtual graft condition.*

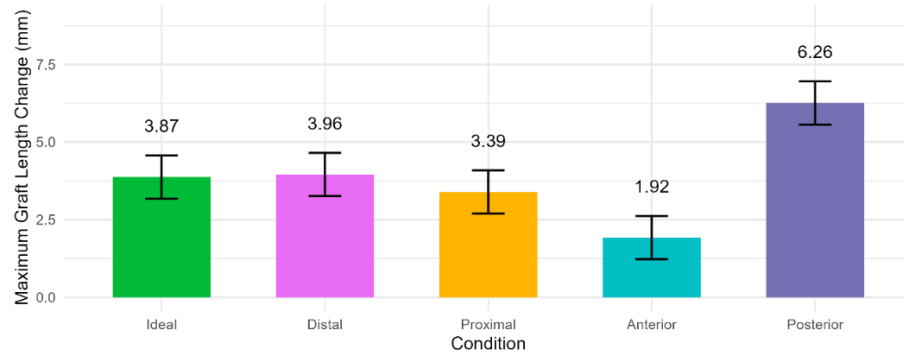


Figure 37. Maximum change in graft length by virtual graft condition.

Each virtual graft condition (ideal, distal, proximal, anterior, posterior) is individually shown where the average maximum graft length change is labelled above each bar.

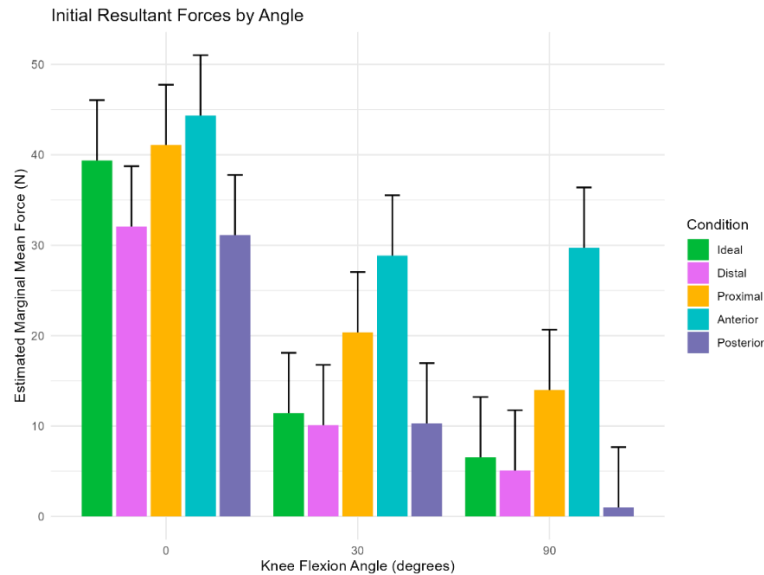


Figure 38. Initial Resultant Forces by Angle.

The initial force (approximately 0N of load) carried by the graft was measured for each virtual graft condition (ideal, distal, proximal, anterior, posterior) and is shown by the colour of the bar.

3.4 Discussion

3.4.1 Kinematics Analysis

3.4.1.1 Anterior Tibial Translation Range

For the ATT range, at 0, 30, and 90 degrees most of the virtual graft conditions have significant differences between the intact and the destabilized conditions indicating a stabilizing and potentially

overstabilizing effect. This may be the result of the virtual graft implementation and the tensioning protocol used in this study to implement the virtual grafts in the neutral 0-degree position, where the tibia was unconstrained and there was no difference between the virtual ACL and remaining intact PCL. At 0 degrees flexion, there was no significant difference between the intact and the posterior virtual ligament condition and the intact and the anterior virtual ligament condition. At 90 degrees flexion, there was no significant difference between the intact condition and the ideal virtual ligament and the intact condition and the posterior virtual ligament condition. As 90 degrees flexion is used for the anterior drawer test, these results are particularly important to highlight [53]. While no differences were identified between virtual ligament conditions at 0 degrees and 30 degrees, at 90 degrees flexion there was a significant difference between the proximal and posterior virtual graft conditions and between the anterior and posterior virtual graft conditions. The posterior virtual grafts, while posing a large standard deviation, do lead to the closest ATT range to the intact condition at 90 degrees flexion. Additionally, from this, it is clear that anterior-posterior translation of femoral tunnel placement had a more significant impact on ATT range than proximal-distal translation. These results indicate there is an effect of the femoral tunnel location in knee biomechanics following ACL reconstruction.

Driscoll *et al.* (2012) selected two different tunnel positions (center-AM and center-center) to perform and test two different ACL reconstruction using bone-patellar tendon-bone autografts [14]. As this study used a virtual quadriceps tendon graft though, which is often stiffer than a bone-patellar tendon-bone graft, the comparisons that will be discussed will not focus on specific magnitudes but rather on trends associated with certain tunnel positions [109]. Driscoll's center-center is most comparable to the ideal femoral tunnel in this study and the center-AM is most comparable to the proximal femoral tunnel in this study. At 90 degrees, both the center-center and center-AM had a greater ATT when compared to intact, with center-AM allowing for slightly more motion [14]. In this study, the ideal and the proximal femoral tunnel locations had a decreased ATT range compared to the intact condition. The center-center femoral tunnel showed more resistance to ATT at 30 degrees in the Driscoll *et al.* study [14]. In this study, at 30 degrees, there was no significant difference between the ideal and the proximal femoral tunnel locations.

Musahl *et al.* (2005) tested two different femoral tunnel locations, position 1 (40% of total notch depth/25% of total anteroposterior condyle diameter) and position 2 (8% of total notch depth/25% of total anteroposterior condyle diameter) using a hamstring tendon graft [12]. As this study used a virtual quadriceps tendon graft though, which is often stiffer than a hamstring tendon graft, the

comparisons that will be discussed will not focus on specific magnitudes but rather on trends associated with certain tunnel positions [114]. Position 1 is most comparable to the posterior tunnel position tested in this study and position 2 is likely somewhere between the anterior and ideal femoral tunnel positions used in this study. In response to a 134 N anterior directed force, the ATT was measured [12]. Musahl *et al.* (2005) showed that an ACL reconstruction done at position 2 had a significantly higher ATT when compared to position 1 for all knee angles except 90 degrees [12]. In this study, at 0 degrees flexion, the anterior femoral tunnel position had a slightly larger ATT range when compared to the posterior tunnel. The ideal tunnel position had approximately the same range as the posterior tunnel position. At 30 degrees, the posterior tunnel position had a higher ATT range when compared to the anterior tunnel position. The ideal tunnel position ATT range was closer to the posterior tunnel position ATT range but was still slightly lower. As previously discussed, there is a significant difference between the posterior and anterior tunnel positions at 90 degrees, where the posterior tunnel position had an increased ATT range compared to the anterior position. This is also seen in Musahl *et al.* (2005) where position 2 had a significantly lower ATT when compared to position 1 [12].

3.4.1.2 Internal Rotation

For the IR max and IR max during combined, the results across the specimens showed no significant difference between any of the seven conditions. For IR max at 0 degrees, 30 degrees, and 90 degrees, the destabilized condition showed slightly more internal rotation than the intact condition. This result is expected as the a secondary function of the ACL is to control internal rotation [115]. For IR max during combined, at 0- and 90-degrees knee flexion, the destabilized conditions show a slight increase in internal rotation when compared to the intact condition. At 30 degrees knee flexion, there is a slight decrease in internal rotation when compared to the intact condition. Table 5 offers additional insight into this as it is clear that Specimen 3 shows a decreased internal rotation in the destabilized condition when compared to the intact condition at both 0- and 90-degrees knee flexion.

As the ACL's secondary function is to resist internal rotation, there other key stabilizers within the knee that contribute to its ability to resist internal rotation. In particular, the anterolateral complex plays a key role in resisting internal rotation [116]. The anterolateral complex consists of the iliotibial band, the anterolateral capsule, the anterolateral ligament, the lateral meniscus, and the lateral collateral ligament [116], [117]. Since the ACL is engaged at lower flexions angles (when no external forces are present), the ACL is a key restraint for internal rotation at lower flexion angles [116]. At higher flexion

angles (50 degrees to 90 degrees), though, the secondary stabilizers play an increasingly larger role in resisting internal rotation and the ACL plays less of a role in IR stability [116]. This can be seen in some of the specimens of this study in Table 4, where at 90 degrees knee flexion, there is a smaller percent increase in IR rotation in the destabilized condition when compared to intact. This is not shown, though in all of the specimens or as a trend across the specimens of this study.

Since there were no significant differences between the virtual grafts or between the virtual graft conditions and intact or destabilized conditions, the virtual graft conditions will not be discussed individually. Instead, the virtual graft conditions will all be compared to results from other ACL reconstruction studies. Noyes *et al.* (2017) investigated the effect of ALL reconstruction on rotational stability in addition to ACL reconstruction [117]. In the study, a 100 N anterior directed load with 5 Nm internal rotation torque and a 7 Nm valgus torque was applied on the specimens [117]. In the ACL reconstruction, the reconstruction was done with a bone-patellar tendon-bone graft [117]. At 25 degrees knee flexion, there was approximately 19 deg of internal rotation with an applied 5 Nm internal rotation torque [117]. At 90 degrees knee flexion, there was approximately 23 degrees of internal rotation with an applied 5 Nm internal rotation torque [117]. Since the virtual graft conditions showed no significant differences, all of these virtual ligament conditions in this study can be generally compared to the reconstructed ACL values from the Noyes *et al.* (2017) work. It is important to note though that this work focused on creating virtual quadriceps tendon grafts as opposed to bone-patellar tendon-bone grafts that were used in the Noyes *et al.* study. At 30 degrees knee flexion, the virtual graft conditions reached 21-23 degrees of internal rotation. At 90 degrees knee flexion, the virtual graft conditions reached 28-29 degrees of internal rotation.

Driscoll *et al.* (2012) investigated a center-AM and center-center femoral tunnel position for ACL reconstruction [14]. IR was measured under a combined 10 Nm valgus and 5 Nm IR torque at 15 degrees, 30 degrees, and 45 degrees of flexion [14]. At 30 degrees of flexion, the center-AM femoral tunnels showed slightly increased IR when compared to intact and center-center [14]. In this study, at 30 degrees of flexion, the proximal tunnel has slightly less IR when compared to the ideal femoral tunnel position.

3.4.2 Force Analysis

3.4.2.1 Initial Force

The initial forces were also reviewed again in the graft length change analysis section. In particular, Figure 38 shows the average initial forces at each recorded angle for each virtual graft condition. The only significant difference occurs at 90 degrees, where the anterior and position conditions have a significant difference in force. The results of the force values at each angle for the virtual graft conditions can be directly compared to Tanabe *et al.* (2019) where five anatomic single bundle ACL reconstructions were done using polyester threads and the length change was reported [8]. The five graft locations were: anatomic AM bundle reconstruction (H-1), anatomic PL bundle reconstruction (E-3), the femoral tunnel at the center of the femoral attachment of the entire ACL midsubstance (F/G-2), the femoral tunnel at the most posterior point of the posterior condyle with two different tibial tunnel location (C-1) and (C-2) [8]. Tanabe *et al.* (2019) used three different tibial tunnels (1, 2, 3), but the femoral tunnels that are most comparable to this study are the following: E is most comparable to the posterior tunnel position, C is most comparable to the proximal tunnel position, F/G is most comparable to the ideal tunnel position, and H is most comparable to the anterior tunnel position. Tanabe *et al.* (2019) reported the load changes among all the grafts, where the load was highest at full extension for each graft (12-15N) and the load reached a plateau (around 12N) for all the grafts between 30 degrees and 120 degrees [8]. In this study, there virtual grafts held more force than the grafts in the Tanabe *et al.* (2019) study. All the virtual grafts held over 30 N of force in extension. Additionally, not all the grafts reached a plateau after 30 degrees like seen in Tanabe *et al.* (2019). The anterior femoral tunnel did reach a plateau (very similar at 30 degrees and 90 degrees), but the other virtual grafts did have a decrease between 30 degrees and 90 degrees (though in most cases, this decrease was about 5 N). If this study had undertaken testing at 120 degrees of flexion, it is possible that this study could have more closely aligned with the results of Tanabe *et al.* (2019).

3.4.2.2 Force During ATT

Musahl *et al.* (2005) tested two different femoral tunnel locations, position 1 (40% of total notch depth/25% of total anteroposterior condyle diameter) and position 2 (8% of total notch depth/25% of total anteroposterior condyle diameter) [12]. Position 1 is most comparable to the posterior tunnel position tested in this study and position 2 is likely somewhere between the anterior and ideal femoral tunnel positions used in this study. In response to a 134 N anterior directed force, graft forces were

measured for the intact knee, graft in position 1, and graft in position 2. At full extension, the intact condition had around 84 N of force, at 30 degrees it had 130 N of force, and at 90 degrees it had 65 N of force [12]. In this study, at 0 degrees the intact condition had 59 N of force, at 30 degrees it had 84 N of force, and at 90 degrees it had 68 N of force. Note that only 89 N of anterior directed force was used in this study, so the magnitude of measured force is not directly comparable. In both studies, 30 degrees had a larger force than 0 degrees and 90 degrees. Position 1 and 2 follow a similar trend where the largest force between 0, 30, and 90 degrees occurs at 30 degrees [12]. In this study, the virtual graft conditions all had the most force at full extension (where they were tensioned). At all angles, except 90 degrees, position 1 carried more force compared to position 2 [12]. In this study the posterior virtual graft condition carried less force at all angles compared to all other virtual graft conditions. Importantly, position 1 and position 2 carried less force than the intact condition at all angles [12]. In this study, all virtual graft conditions carried more force than the intact condition, except at 90 degrees where the posterior virtual graft condition carried less force than the intact condition.

3.4.2.3 Force During IR

The five virtual graft conditions all had forces of 90 N to 160 N at 30 degrees of flexion. At 90 degrees of knee flexion, the virtual graft conditions had forces of 23N to 90N. This can be compared to other work such as Noyes *et al.* (2017) who investigated the effect of ALL reconstruction on rotational stability in addition to ACL reconstruction [117]. The ACL graft force (N) was measured at each condition, where at 25 degrees knee flexion there was approximately 129 N of force and at 90 degrees knee flexion there was approximately 39 N of force [117]. This means that the tested virtual graft conditions in this study have forces present during IR are comparable to the forces seen during IR during ACL reconstruction.

Additionally, Musahl *et al.* (2005) tested two different femoral tunnel locations, position 1 (40% of total notch depth/25% of total anteroposterior condyle diameter) and position 2 (8% of total notch depth/25% of total anteroposterior condyle diameter) [12]. Position 1 is most comparable to the posterior tunnel position tested in this study and position 2 is likely somewhere between the anterior and ideal femoral tunnel positions used in this study. Musahl *et al.* (2005) tested specimens with a combined loading protocol of 5 Nm internal tibial and 10 Nm valgus torque torque at 15 degrees and 30 degrees [12]. While this study did not apply a valgus torque on the specimens, the results of the combined loading protocol by Musahl *et al.* (2005) will be compared to the force measured during IR

results in this study [12]. At 30 degrees of flexion, position 1 had a force of 25.3 ± 15.6 N and position 2 had a force of 19.8 ± 18.3 N [12]. In this study, at 30 degrees of flexion, the virtual graft conditions all had forces over 100 N during IR (except the posterior virtual graft condition which was slightly under 100 N) and therefore are much higher than the forces seen in the study by Musahl *et al.* (2005).

3.4.2.4 Force During Combined

For force during combined, there was no previous literature identified which had both an anterior-directed force and an internal rotation torque applied on the tibia.

3.4.3 Graft Length Change Analysis

3.4.3.1 Average Across Specimens

For the average graft length change across the specimens, Figure 35 shows that the anterior femoral tunnel position is the most isometric while the posterior point has the greatest length change across the knee flexion range. When looking further into the significant differences between the conditions, all the virtual graft conditions (except posterior) have some significant differences with the anterior or posterior femoral tunnel locations in the 30 to 90 degrees of knee flexion range.

These results can be directly compared to the same study done by Tanabe *et al.* (2019) where five anatomic single bundle ACL reconstructions (H-1, E-3, F/G-2, C-1, and C-2) were done and the length change was reported [8]. The femoral tunnels that are most comparable to this study are the following: E is most comparable to the posterior tunnel position, C is most comparable to the proximal tunnel position, F/G is most comparable to the ideal tunnel position, and H is most comparable to the anterior tunnel position. Tanabe *et al.* (2019) report that H-1 is the most isometric graft (has the smallest change in length). This study also reaches the same conclusion, where the anterior femoral tunnel location has a maximum average change in length of 1.92 mm. Tanabe *et al.* (2019) report that C-1 and C-2 have the greatest maximum length change (around 7mm), while in this study the posterior femoral tunnel has the greatest length change (average of 6.26 mm). The proximal point, comparable to C-1/2 only has 3.39 mm length change. The F/G position has about 4mm of maximum length change, which is also seen in this study where the ideal tunnel has 3.87mm of maximum length change. The E graft has around 6 mm of length change, which is close to the 6.26mm seen in the posterior position. Since C-1 and C-2 use different tibial tunnel positions when compared to E-3 (uses position 3), while all the femoral tunnel positions in this study had the same tibial tunnel position, this could explain why Tanabe

et al. (2019) report the C femoral tunnel as having the greatest change in length while this study reports the posterior tunnel as having the greatest change in length.

Yoo *et al.* (2010) used CT scans of 10 subjects with normal knee to measure the length of four virtual ACL bundles (anatomical single bundle (center of the ACL footprint), AM, PL, and over the top [67]. CT images were taken of each subjects' knee at four different flexion angles: 0 degrees, 45 degrees, 90 degrees, and 135 degrees [67]. The tunnel placed at the center of the ACL foot is most comparable to the ideal tunnel location in this study, the center of the PL is most comparable to the distal tunnel location in this study, the center of the AM is most comparable to the proximal tunnel, and the over the top is most comparable to the anterior tunnel in this study (though the over the top is also oriented slightly more distally than the anterior tunnel). In their study, Yoo *et al.* (2010) also used three different tibial points (center of the ACL footprint, center of the AM bundle, and center of the PL bundle) [67]. In this study, only one tibial tunnel position was chosen, at the center of the ACL footprint on the tibia. Yoo *et al.* (2010) have four virtual grafts that were then tested: mid ACL to mid ACL, mid AM to mid AM, mid PL to mid PL, and mid ACL to femoral over the top [67]. Their results show that the AM graft has the largest graft length while the PL graft had the shortest graft length, over the recorded range of motion [67]. This is also seen in this study, where the proximal graft had the largest graft length over the recorded range of motion and the distal graft had the shortest graft length over the recorded range of motion. Yoo *et al.* (2010) showed that the over-the-top single bundle graft had a slightly larger graft length over flexion angle than the anatomic single bundle graft. This is not shown in this study, where the ideal had a larger graft length than the anterior tunnel, but as previously mentioned the anterior tunnel position is not located exactly where the over-the-top single bundle graft was.

3.4.4 Limitations

3.4.4.1 General Limitations

As previously discussed, this study tested nine specimens, but three specimens were omitted due the lack of change in ATT between the intact and destabilized conditions. The minimal change seen between the destabilized and intact conditions means that the ACL was likely not fully disrupted in the study. As this was only discovered through the analysis after the testing, this protocol choice was not ideal to ensure that all specimens tested for this study yielded data that could be utilized. For future studies, a brief period to perform the analysis comparing the maximum ATT between the intact and

destabilized conditions should be undertaken. If there is no change between the maximum ATT in the destabilized condition compared to the intact condition, the specimen should be rechecked arthroscopically to ensure that the ACL is fully ruptured. Additionally, the usage of virtual ligaments can be said to be a general limitation itself. Specifically, since virtual ligaments are one dimensional point-to-point springs, they act in a straight line. The ACL though, can interact with the PCL at deeper knee flexion angles due to their crossing orientation in the intercondylar notch which can change the force carried by the ACL [118]. This effect cannot be replicated, though, when using virtual ligaments.

3.4.4.2 Kinematic Analysis Limitations

The virtual ACL graft tensioning protocol used in this study allowed the tibia to move freely relative to the femur until the prescribed graft tensioning force was reached. This created posterior translation of the tibia relative to the femur and external rotation of the tibia relative to the femur in all the virtual ACL conditions. Brady *et al.* (2009) reported anterior-posterior translation of the tibia and axial rotation of the tibia when different graft tensioning was used [119]. The results showed that an increase in graft tension caused external rotation of the tibia and posterior translation of the tibia at all knee flexion angles [119]. Specifically, at 0 degrees flexion, there was posterior translation of around 1-2 mm and around 1-2 degrees of external rotation when 30 and 60 N of graft tension was applied [119]. At 20 degrees, though there was no posterior translation [119]. At 20 degrees, there was still around 1 degree of external rotation for 60N of graft tension, but internal rotation (around 1mm) for 30N of graft tension [119]. In this study, there was still posterior translation and external rotation for some of the virtual graft conditions, though, at 30 degrees.

Furthermore, a study done by Vakili *et al.* (2024) demonstrated the ability to construct a specimen-specific virtual ACL that restored knee stability to the same extent as real ACL reconstructions [15]. The virtual ACL graft conditions in this study had increased posterior translation and external rotation when compared to the intact ACL. As no real reconstruction was done for this study, the results of the current study can only be compared to the intact ACL. Additionally, the virtual ACL grafts over constrained the motion of the knee joint, where the maximum ATT reached was less than the intact specimens and the ATT range was also lower than the intact specimens. As previously discussed, likely due to the influence of secondary stabilizers, the IR max results were not as affected by this, but also did not show any significant results.

3.4.4.3 Force Analysis Limitations

The first limitation of the force analysis is due to the tensioning protocol. Specifically, as the tensioning protocol over constrained the motion allowable for the virtual graft conditions, it is likely that the forces are higher than expected during the applied motions (ATT, IR, Combined). A secondary limitation for the force analysis is the method chosen to estimate the force carried by the intact ACL. As this was calculated as a subtraction between the intact force control data and the displacement control ACL-destabilized condition, if there was high error in the measurement of the intact force or the forces in the destabilized condition, then the calculated value for the intact force components would have errors. Additionally, if the displacements applied to the destabilized specimens were not exactly the same as what was recorded from the intact specimens, then this could lead to an additional source of error in the calculation of the intact force.

3.4.4.4 Graft Length Change Analysis Limitations

Tanabe *et al.* (2019) discussed a variety of limitations for their analysis that are relevant to this study [8]. In this work, graft length change was only calculated over the flexion-extension range of motion for the specimens. While this is clinically relevant and is a common way to discuss the degree of graft isometry, as the ACL contributes in resisting internal tibial rotation, understanding of how the graft length changes during rotation and pivoting could also provide additional clinical context [116]. Furthermore, while the kinematic analysis was conducted and the tracking data was recorded for these sections, no analysis was done to investigate the graft length change during conditions when the ACL was loaded (ATT, IR, and combined motion) was done. As the loading protocol was designed to simulate relevant clinical laxity exams, understanding how the graft length changes during this testing could offer additional clinical insights. Additionally, the location of the specimen's actual ACL intact was not known. Therefore, each virtual graft length change was not able to be compared to the actual ACL's length change. Finally, using cadaveric specimens for this study creates additional limitations. As the cadaveric specimens used in this study were not from young donors, this data might not be entirely reflective of the demographic often experiencing ACL tears. While the virtual ACL grafts were given a stiffness that matched younger individuals, the cadaveric specimens were not from young donors, so the surrounding tissues and musculature of the cadaveric specimens is likely not equivalent to those of younger individuals. In particular, ligaments and tendons of older individuals are less elastic and have a lower tensile strength than younger individuals [120].

Chapter 4

Design & Development of a Proof-of-Concept Surgical Navigation System for ACL Reconstruction

4.1 Introduction

Based on the experimental study, in the previous chapter, of the virtual ACL reconstruction using different femoral tunnel locations, it is clear that the position of the femoral tunnel has an impact on resulting knee kinematics, the force carried by the graft and graft length compared to flexion angle. While previously investigations of surgical navigation systems for ACL reconstruction have not shown significantly different clinical outcomes compared to traditional ACL reconstruction, with better intraoperative evaluation of possible femoral tunnel locations, it is possible that a surgical navigation system could be designed and used to yield better clinical outcomes. As such, the goal of this Chapter was to develop a proof-of-concept surgical navigation system for arthroscopic ACL reconstruction for intraoperative femoral tunnel location optimization.

4.1.1 Objectives

The primary objective of this chapter was to design and develop a proof-of-concept system for ACL reconstruction surgical navigation for the purpose of better femoral tunnel identification and positioning. The specific objectives are as follow:

1. to design a clinically-informed workflow for surgical navigation for ACL reconstruction to aid surgeons in selecting the optimal femoral tunnel position;
2. to develop a proof-of-concept surgical navigation system to aid surgeons in selecting the optimal femoral tunnel position.

4.2 Design Requirements

The design requirements for the proposed system were developed with clinician input and are summarized below in Table 6.

Table 6. Design Requirements for a Surgical Navigation System for ACL Reconstruction

Requirement	Benchmark
1. Must easily fit into the current ACL reconstruction workflow.	The system should not significantly increase the length of the procedure more than 15 minutes.
2. Must be intuitive for surgeons to use the system.	The system should not require a long training period for surgeons.
3. Must provide users with information regarding different femoral tunnel start points.	The system should provide clear metrics of how different femoral tunnel start points behave differently from other start points.
4. Must have high accuracy of tracked tools.	The system should have sub-millimetre precision since femoral tunnel positioning deviations are within the millimetre range [90]

4.3 System Design Workflow

The proposed system was designed to easily fit into the current ACL reconstruction workflow and to be intuitive for surgeons to use (Requirement 1 and Requirement 2), which would aid in clinical adoption of the system. As previously discussed, standard ACL reconstruction involves medical images being taken of the patient (MRI), the patient is prepared for surgery, arthroscopic portals are made, a graft is harvested, the tibial and femoral tunnels are made, the graft is inserted and tensioned, and then the incisions are closed [53], [56], [58]. This standard workflow is shown below in Figure 39, in yellow. The proposed surgical navigation system workflow steps are shown in the same figure in black.

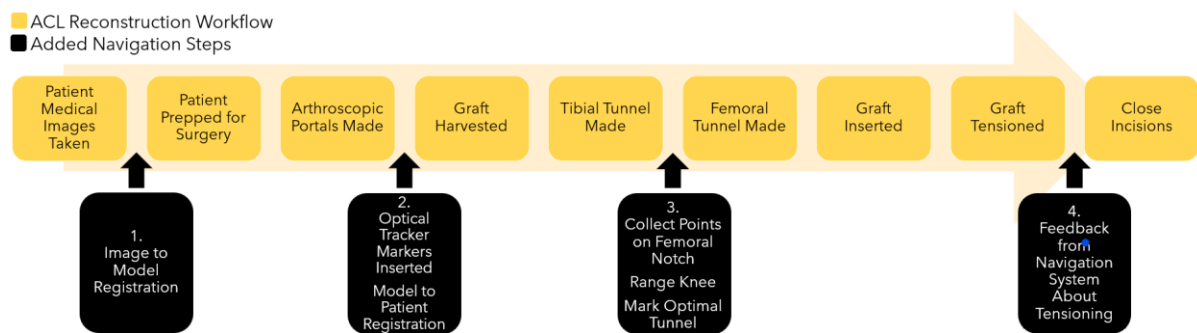


Figure 39. Proposed surgical navigation system for arthroscopic knee ligament reconstruction. The standard anterior cruciate ligament (ACL) reconstruction workflow steps are shown in yellow and the added navigation steps for the proposed system design are shown in black.

The added steps for the proposed surgical navigation system were designed to specifically address the design requirements. There are four primary added steps to the surgery and these steps are detailed below, sequentially:

- 1. Image to Model Registration:** Since ACL tears are often diagnosed with medical images, these images can be leveraged for the surgical navigation system. 3D models of patient specific anatomy (femur and tibia) would be generated ahead of time from these patient medical images (MRI). When this step is done, the 3D models would be aligned with the patient medical images, effectively completing an image to model registration. This step is the only necessary preoperative step, though femoral tunnel planning could also be done preoperatively using the generated femur model (identifying possible femoral tunnel locations on the femur model using landmarks).
- 2. Model to Patient Registration:** At the start of the surgery, the optical trackers need to be inserted into the femur and tibia. This is done by drilling bone pins into the femur and tibia where the optical trackers are attached to the bone pins at the top of the pins. After the optical trackers are attached to the patient, the 3D model to patient registration needs to be done. This registration process is required to allow the bones in the 3D scene to move when the patient's knee moves. For this process, surgeons will be able to leverage the three arthroscopic portals typically made for ACL reconstruction surgeries to gain access into the knee inner joint space to identify key landmarks within the knee where the same landmarks can be identified on the patient-specific 3D models. Once this registration is completed, the surgeon will drill a tibial

tunnel following standard procedure and note where this tunnel was drilled in the user interface by selecting the location using the tracked stylus.

- 3. Collect Points & Range Knee:** For this stage in the process, the surgeon can select possible femoral tunnel locations either on the patient using the tracked stylus or on the 3D model on the femur in the scene. After the points are selected, the surgeon will then range the patient's knee from 0 degrees flexion to 90 degrees flexion. A change in graft length vs. knee flexion angle graph can then be generated, along with other information about knee kinematics, for the proposed femoral tunnel locations along with the identified tibial tunnel location. This plot can then be used by the surgeon to select an optimal femoral tunnel location for the specific patient. Once this point is selected in the user interface, the surgeon will mark this exact location on the patient's femur using tracked stylus. Next, the surgeon will then drill the femoral tunnel and insert the prepared graft.
- 4. Graft Tensioning Feedback:** When the graft is being tensioned, the surgical navigation system can provide feedback to the surgeon regarding the current location of the tibia compared to the femur and if the tibia is in an appropriate (reduced) position. Once the graft is appropriately tensioned, the surgeon can then remove the bone pins and close up all the incisions.

As shown in Figure 39, only four main steps were added to the current ACL reconstruction workflow. These steps were determined with clinician input as to minimize the amount of modification to the procedure, while still getting the core benefits of surgical navigation. While the exact timing cannot be confirmed for how long this would take, the clinician partners on this project strongly believe that these modifications would only take 15-20 additional minutes, thus meeting the first requirement. As much of this proposed workflow is the same as the standard ACL reconstruction steps and the added navigation steps are common among most surgical navigation systems, the proposed system is intuitive for clinicians to use and would not require a long training period. This allows the system to meet the second design requirement. For the third design requirement, this is met in the third added step of the proposed workflow. Specifically, a graft length vs. flexion angle graph will allow surgeons to better understand the differences between their selected potential femoral tunnel start points, as shown by the study in the previous chapter. Finally, the fourth design requirement is met by selecting an appropriate

optical tracking system. For the proof-of-concept development, an NDI Polaris system will be used which has been shown to meet the sub-millimetre accuracy [121].

4.4 Proof-of-Concept Development

From the developed proposed workflow, a proof-of-concept system was developed using 3D Slicer and the SlicerIGT extension along with the Sports Med Knee from Tactile Orthopedics and NDI Polaris optical tracking system. Key developments will be detailed in the subsequent sections.

4.4.1 Image to Model Registration:

For the system development, the first stage was performing a registration and building an appropriate transformation hierarchy. To make the models easier to visualize for the user, the femur was chosen as a reference. As such, the transformation hierarchy was developed so that the tibia will move with respect to the femur and the femur model would be in a fixed position. As such, a registration was done between the femur tracker model in 3D Slicer and the tracker on the femur. As well, the femur bone model was registered to the Tactile femur. The registration was then done between the tibia tracker model in 3D Slicer and the tracker on the tibia. The tibia bone model was also registered to the Tactile tibia. Both of these two transforms were then nested under the tibia tracker transform (brought into 3D Slicer scene). Then the entire tibia transform hierarchy was nested underneath the femur transform of the femur tracker model to allow the tibia to move relative to the femur. Finally, the femur model in the 3D Slicer scene was registered to the femur within CT images taken of the Sports Med Knee. This kept the femur fixed to the images, but allowed the tibia to move freely when the Sports Med Knee was moved. The results of this transformation hierarchy development is shown below in Figure 40, where the femur is horizontal in the 3D Slicer scene and aligned with the femur in the CT images (despite not being horizontal with the Sports Med Knee) and the tibia is positioned on an angle.

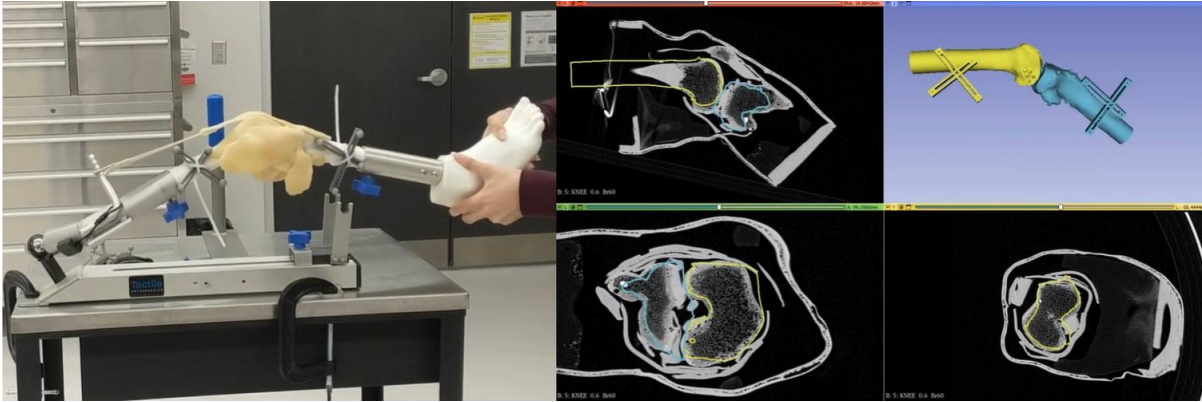


Figure 40. Registered Sports Med Right Knee Model (Tactile Orthopedics) within a 3D Slicer scene.

Tactile Knee Model and trackers on the physical model are shown registered to 3D CAD models in the 3D Slicer scene.

4.4.2 Calculating the Knee Flexion-Extension Axis

To get a better understanding of where the knee flexion-extension axis lies relative to the Sports Med Knee within the 3D Slicer scene, the symmetrical axis of rotation approach (SARA) method created by Ehrig *et al.* (2007) was leveraged. In the SARA method, the joint axis of rotation of two segments can be determined by analyzing the motion of two independent segments [41]. In particular, the transformation matrices of each marker is determined with respect to the global coordinate system at each time point during the flexion-extension motion of the segments [41]. An objective function is then minimized (singular value decomposition) to determine two coordinates on the axis of rotation in the local coordinate system of each segment [41]. To implement this method in 3D Slicer, two initial points are created (one in the Slicer Scene under no transform and the other under the tibia tracker transform, placing one in the femur coordinate space and one relative to the tibia). A Python script was developed to perform the singular value decomposition of the SARA objective function. As this script runs, the user flexes and extends the Sports Med Knee. The axis of rotation then updates live of the screen. The results of this process are shown below in Figure 41, where the two points from each local coordinate system on the axis of rotation are in red and have a line connected between them to depict the axis of rotation.

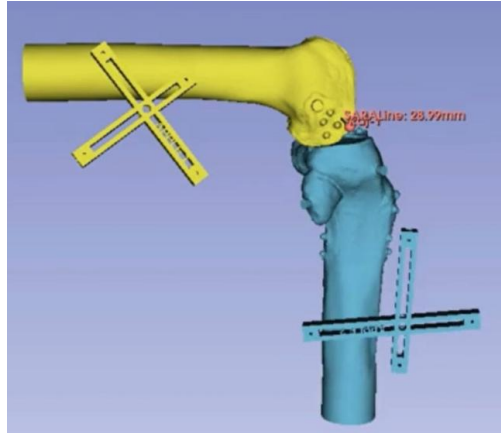


Figure 41. Flexion-extension knee axis identified in a registered 3D Slicer scene.

Depicted axis of rotation shown was calculated through the symmetrical axis of rotation approach (SARA) method.

4.4.3 Generating the Change in Graft Length vs. Knee Flexion Angle Graphs

To select proposed femoral tunnel locations on the registered femur model in the 3D Slicer scene, the Collect Points module from the Slicer IGT extension was utilized. This module allows for the user to add landmarks within the 3D Slicer scene from a tracked calibrated stylus (calibrated through the Slicer IGT Pivot Calibration module). Figure 42 shows six points annotated on the femur model that were selected through the Collect Points module.

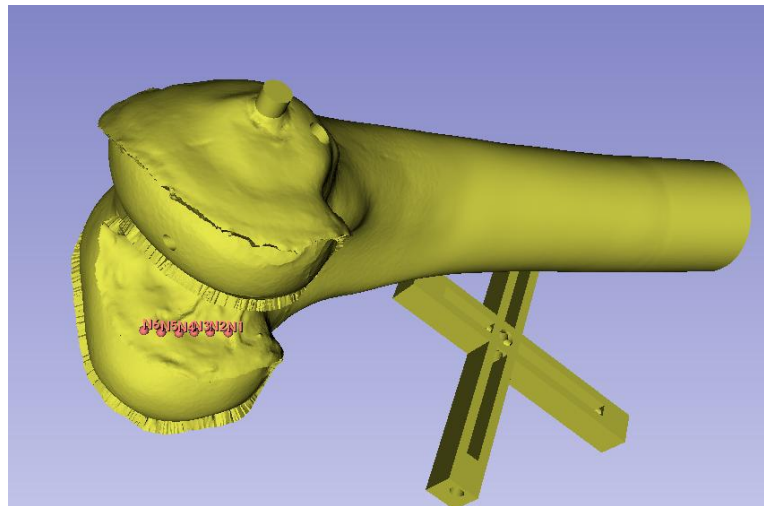


Figure 42. Landmark annotation on femur model within a 3D Slicer scene.

Six landmarks selected on a tracked model of the femur in a 3D Slicer scene.

Once the user has identified the points of interest on the femur (either through the Collect Points module or added manually in the 3D Slicer scene as fiducial points through the Markups module), a line length vs. knee flexion angle algorithm was developed to generate the line length vs. knee flexion angle graph. For calculating the knee flexion angle, the algorithm relies on Grood and Suntay axes definition and landmark identification. The user first begins by identifying four landmarks on the femur and four landmarks on the tibia. On the femur, the landmarks are the most posterior edge of the lateral femoral condyle, the most posterior edge of the medial femoral condyle, the middle, and the center of the femoral shaft. To calculate the center of the femoral shaft, four points are placed at each edge of the femoral shaft. This is shown in Figure 43. On the tibia, the landmarks are the most lateral side of the proximal tibial plateau, the most medial side of the proximal tibial plateau, the center of the proximal tibial plateau, and the center of the tibial shaft. To calculate the center of the tibial shaft, three points are placed at each edge of the tibial shaft. This is shown in Figure 44. These landmarks are identified as separate lists within 3D Slicer Markups module. A Python script is then used to determine the Grood and Suntay axes and then calculates the relative anatomical angles and translations from the calculate axes along with the transformation matrices of the tibia and the femur relative to the global coordinate system. For the larger line length vs. knee flexion angle algorithm, though, only the flexion angle is used. At the same time, lines are created between each potential femoral tunnel location (list of points) and a chosen tibial tunnel location (one point). As the tracked bone models move (transforms change), the lines update and their length parameter updates. For each new tibial tracker transform (movement of the tibia), there is a new set of line lengths, and a new knee flexion angle added to a table within 3D Slicer (3D Slicer Informatics Module: Tables). This table can then be exported as a csv file, which can then be used to graph the line lengths vs. knee flexion angle graph. This is shown in Figure 45.

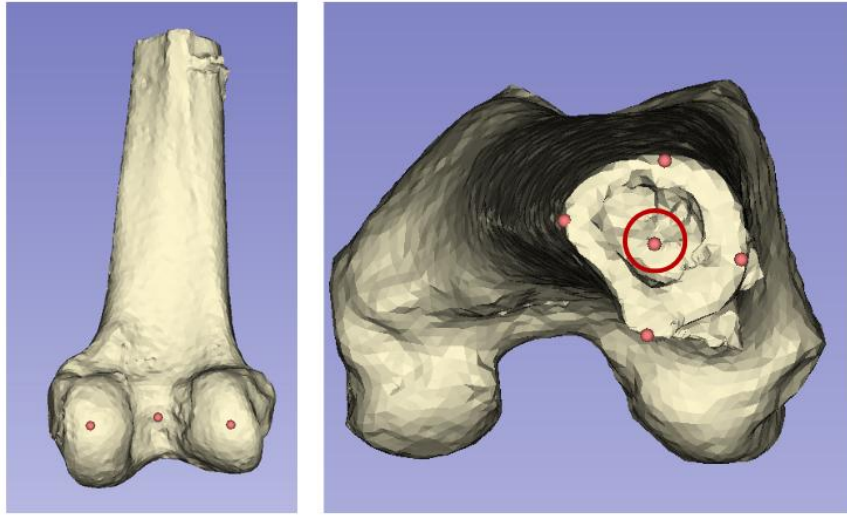


Figure 43. Femur Grood and Suntay landmarks identified in 3D Slicer.

On the left a posterior view of the femur is depicted with three Grood and Suntay landmarks identified, the two posterior edges of the femoral condyle and the center point between them. On the right, a superior view of the femur is depicted with four pointed annotated used to calculate the fourth Grood and Suntay landmark, the center of the femoral shaft.

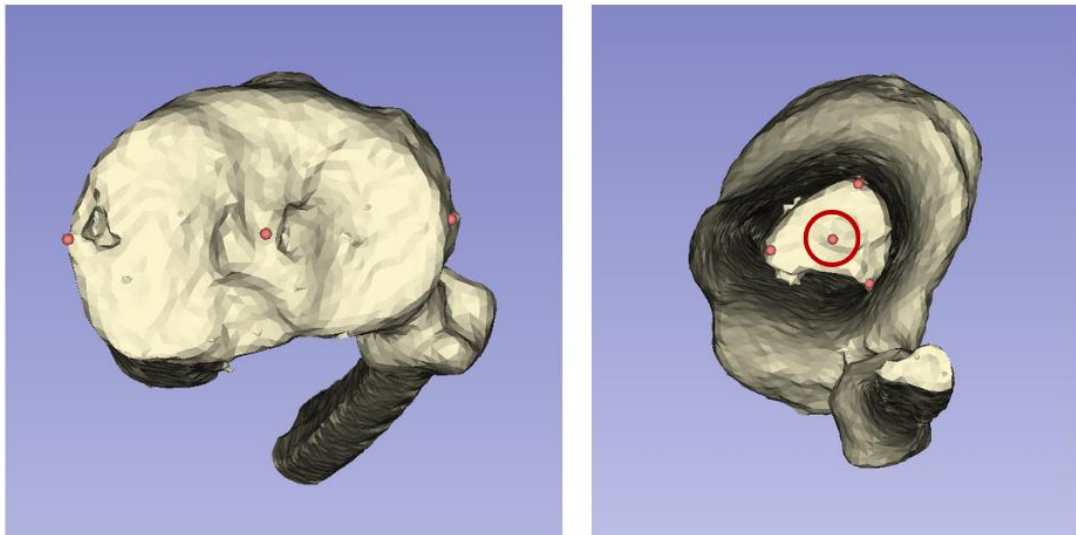


Figure 44. Tibia Grood and Suntay landmarks identified in 3D Slicer.

On the left a superior view of the tibia is depicted with three Grood and Suntay landmarks identified, the two posterior most lateral points on the tibial plateau and the center of the tibial plateau. On the right, an inferior view of the tibia is depicted with three pointed annotated used to calculate the fourth Grood and Suntay landmark, the center of the tibial shaft.

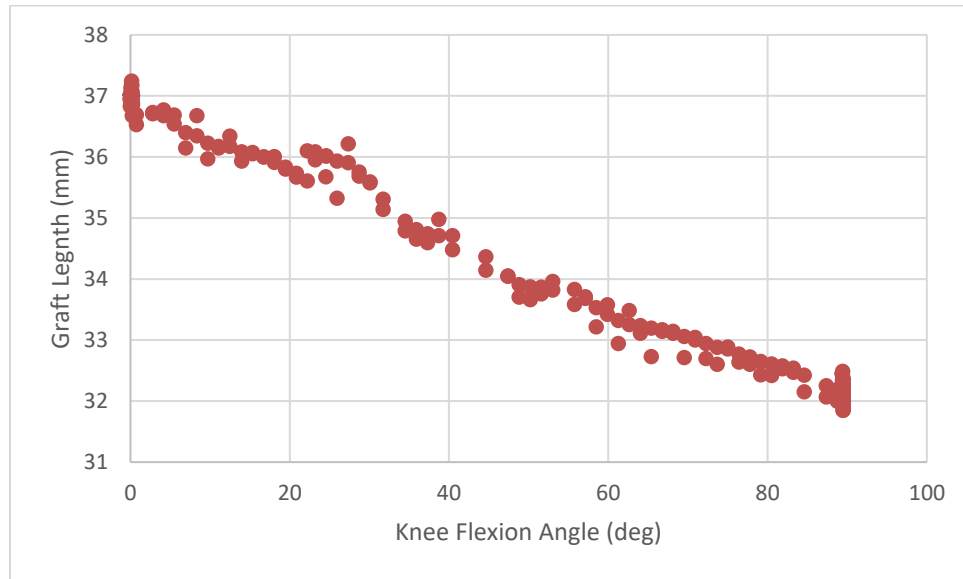


Figure 45. Generated graft length vs. knee flexion angle graph.

Raw data of graft length was measured during knee flexion-extension range of motion for a selected femoral tunnel in 3D Slicer scene using the tracked Tactile knee model.

4.4.4 Ensuring Accurate Femoral Tunnel Drilling

Once the user has decided which proposed femoral tunnel location they want to use for the procedure, the user enters a python script to convert the selected fiducial point into a model. The purpose of this procedure is to be able to use the Breach Warning module through the Slicer IGT extension. This module allows the user to input a model and input a tracked stylus. Once this has been done, the 3D Slicer scene updates to allow the user to see the distance of the stylus tip to the identified model of interest. In the presented use case, the femoral tunnel of interest is turned into a model and then selected as the model input in the Break Warning module. The stylus can then be used to mark the location of the start point of the desired femoral tunnel (indenting the bone with the stylus). This allows the user to be sure that they will drill the femoral tunnel in the location that produces the desired graft length vs. flexion angle curve. The Breach Warning module used in the proof-of-concept system is shown below in Figure 46.

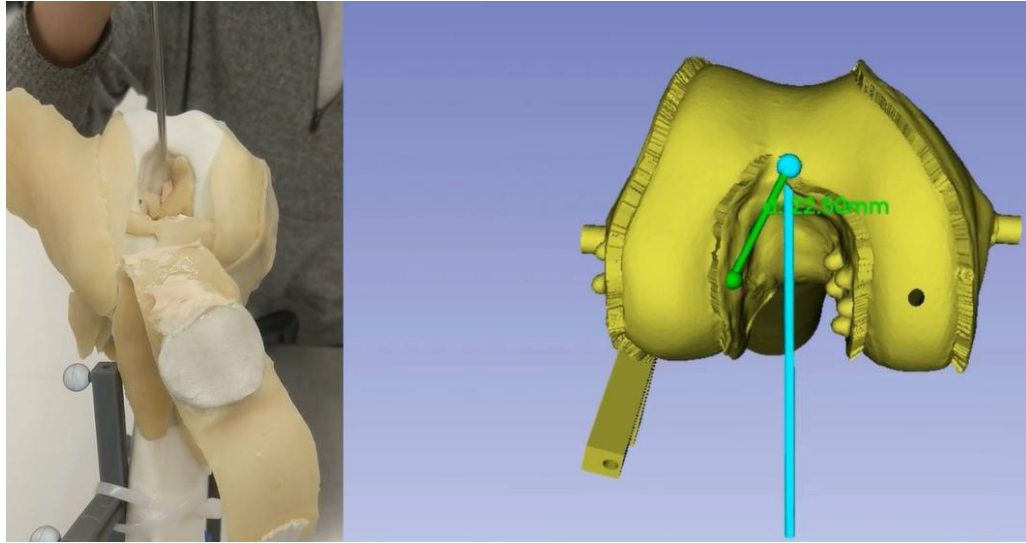


Figure 46. Breach Warning Module used within a registered 3D Slicer scene.

On the left the tracked stylus is shown relative to the Tactile Knee Model and on the right is shown relative to the 3D model of the femur in 3D Slicer.

4.5 Discussion

There are several examples of surgical navigation systems used for ACL reconstruction in literature and in industry. Luites *et al.* (2007) developed a method to describe the attachment geometry of the bundles of the ACL through the arthroscopically visible landmarks [122]. Leveraging their previously developed methods, an implementation of this in a surgical navigation system was used for determining patient-specific anatomical tunnel positions and accurate tunnel placement in Luites *et al.* (2014) [91]. Their morphable template was implemented in the SurgiGATE (Medivision, Switzerland) computer-assisted surgical system and was tested for a double-bundle ACL reconstruction [91]. The results showed the system had a high accuracy for tunnel placement [91]. Margier *et al.* (2015) investigated the use of the Surgetics navigation system (Praxim, France) for navigated ACL reconstruction [10]. The Surgetics system allows surgeons to determine the isometry profile of different tunnel positions over the flexion-extension knee range of motion [85]. The Surgetics system increased the mean operating time by 28 minutes compared to the non-navigated procedures [10]. Overall, the use of the system was associated with a greater cost than the non-navigated procedures due to increased operating time, equipment cost, staff cost, and OR occupancy [10]. Additionally, the Surgetics system did not have any better clinical outcomes compared to the non-navigated procedures [10].

The proof-of-concept system detailed in this chapter has many similarities with the previously described systems. All three systems used methods to allow users to select and compare different femoral tunnel positions [10], [85], [91]. In particular, the proof-of-concept system detailed in this Chapter and the Surgetics systems develop isometry profiles detailing the graft length change over the knee flexion-extension range of motion. All three systems had methods to ensure that the tunnel was placed accurately, through the use of tracking modalities [10], [85], [91]. While the accuracy of the proof-of concept system was not formally quantified, many of the components of the system are similar to other surgical navigation systems and to surgical navigation systems for ACL reconstruction that have well defined techniques for calibrating and measuring system level accuracy.

4.5.1 Limitations

There are obviously limitations when developing a proof-of-concept system. In particular, no testing was undertaken to investigate the accuracy of this system. Due to this, specific comparisons in accuracy between this system and other systems on the market or detailed in literature cannot be made at this time. Another significant limitation of the proof-of-concept system is that the fourth component of the proposed workflow (Graft Tensioning) was not implemented in the system. Due to this, this part of the system was not able to be tested.

Chapter 5

Thesis Conclusions

5.1 Introduction & Summary

This thesis investigated the influence of the femoral tunnel and developed a proof-of-concept surgical navigation system to aid clinicians in better identifying and placing the optimal femoral tunnel starting points. In particular, the cadaveric biomechanical investigation in this thesis analyzed the multifaceted influence of the femoral tunnel on resulting knee kinematics post-graft insertion (ATT and IR), on the force carried by the graft, and on the graft length change over a flexion-extension range of motion of the knee. Additionally, the designed surgical navigation system workflow for ACL reconstruction leveraged clinician input to develop a system that would be easy for surgeons to use and not significantly increase the time of the procedure. As well, the proof-of-concept system that was developed met the key design requirements and demonstrated the feasibility of integrating tracking and guidance for acute femoral tunnel positioning during ACL reconstruction surgery. Overall, this thesis work successfully demonstrated that the femoral tunnel positioning does influence knee biomechanics and that surgical navigation can be used to more accurately identify and position the femoral tunnel start point.

5.2 Research Significance & Potential Impact

5.2.1 *In Vitro* Biomechanical Study

The study presented in this thesis is the first investigation into the influence of the femoral tunnel position using virtual ACL grafts. Specifically, many cadaveric *in vitro* biomechanical investigations on the role of femoral tunnel positioning focus on testing common ACL reconstruction methods which lead to different femoral and tibial tunnel positioning. Due to the dominant role of the femoral tunnel in the failure of ACL reconstruction procedures, this study chose to perform the investigation with a constant tibial tunnel location. This study demonstrated that varying the femoral tunnel insertion can create statistically significant kinematic differences in ATT at some knee flexion angles (90 degrees). Additionally, varying the femoral tunnel location is associated significant differences in forces carried by the graft at all angles during all applied motions. Graft length change throughout the knee range of motion can create different length change profiles, where some femoral tunnel positions lead to more

isometric grafts and others are less isometric. The anterior and posterior positioning of the femoral tunnel, in particular, led to the most differences across the three analyses methods. Overall, the femoral tunnel position plays a key role in knee kinematics, force carried by the graft, and graft length change over knee flexion post-ACL reconstruction surgery, making its precise placement a crucial surgical decision.

5.2.2 Surgical Navigation System Design

The proof-of-concept surgical navigation system presented in this work is the first use of 3D Slicer and the SlicerIGT extension for developing a surgical navigation system for ACL reconstruction. By leveraging the built-in modules of the SlicerIGT extension, the proof-of-concept system required less development time compared to entirely custom-built systems. Additionally, the proof-of-concept system leverages commonly used methods for calculating clinical rotations (Grood and Suntay (1983)) and calculating the knee flexion-extension axis (Ehrig *et al.* (2007)) which can help support the system's reliability. Overall, the proposed surgical navigation system was developed with significant input from clinicians and demonstrates a feasible way of leveraging commonly used surgical navigation techniques applied to the ACL reconstruction procedure.

5.3 Recommendations for Future Work

5.3.1 *In Vitro* Biomechanical Study

5.3.1.1 Overall

One recommendation for future work would be to consider adjusting or varying the tensioning protocol that was designed for this study. Specifically, posterior subluxation of the tibia when the ACL graft is tensioned is not seen clinically to the magnitude that was seen in this study [119]. A better understanding how clinicians constrain the movement of the tibia relative to the femur during graft tensioning or how clinicians position the knee joint while tensioning the graft is critical to designing an updated tensioning protocol to limit posterior subluxation.

5.3.1.2 Kinematic Analysis

While posterior tibial translation and external rotation after tensioning was seen in the tested specimens of this study, the amount of translation and rotation is relative to the reference pose of the specimens when they are mounted in the simulator. As tracking data was recorded for the entire duration of the

testing protocol, clinical translations and rotations can be calculated based on the Grood and Suntay axes definition convention that were calculated based on the specimen-specific Grood and Suntay landmarks. In this manner, clinical translations and rotations can be reported relative to the location of the Grood and Suntay axes, which can aid in better understanding the exact degree of translation/rotation relative to known landmarks. Additionally, this can help better specify the exact positioning of the femur relative to the tibia. For example, the exact distance to the posterior edge of the tibial plateau to the most posterior edge of the femoral condyles can be calculated.

5.3.1.3 Force Analysis

With an improved tensioning protocol, better insight into the forces carried by the virtual graft conditions during each of the applied motions can be understood. As the force to tear the ACL ranges from 650N-2160N depending on age, the exact magnitude of the forces carried by the virtual ACL during the simulated motions has strong clinical significance [65]. Specifically, understanding which femoral tunnel positions correspond with larger or smaller force magnitudes can aid in specifying which femoral tunnel locations have greater risks for rupture [65].

5.3.1.4 Graft Length Analysis

Future work for graft length analysis should add a focus on using the available tracking data to identify the graft length change under applied loads. As ACL tears occur under loaded conditions, understanding how the graft length changes for each femoral tunnel location when an anterior directed force is applied on the tibia, when an internal rotation torque is applied on the tibia, and when a combined load is applied on the tibia can offer a better insight into understanding the difference in femoral tunnel locations. Specifically, when the virtual graft is stretched farther with an applied load, it can be determined which virtual grafts will stretch beyond 10mm (when the ACL is typically considered to enter into the prefailure region of the stress-strain curve) [65].

5.3.2 Surgical Navigation System Design

While the proof-of-concept system successfully met the design requirements and was able to demonstrate how to better inform the selection and positioning of the femoral tunnel, there is a lot of future work that can be done to improve this system. Firstly, as the graft tensioning portion of the workflow was not currently completed for the proof-of-concept, this would be the next to implement in the system. Additionally, the proof-of-concept was built using the SlicerIGT extension modules. As

such, the system does not have its own user interface separate from the standard 3D Slicer user interface. The next stage of this work would be to create a custom-built guidelet within 3D Slicer to link all the steps of the workflow together in one application. Creating a custom-built guidelet would allow the system to be more user-friendly and allow it to be more comparable to surgical navigation systems on the market. A final recommendation is that future work needs to investigate determining how to ensure the range of motion used to generate the graft isometry profiles is physiologic. Since the destabilized knee is the knee being ranged, a physiologic range of motion still needs to be applied to ensure that accurate graft isometry profiles can be calculated for each femoral tunnel position.

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Appendix A

Additional Kinematic Analysis Figures

Figure 47, Figure 48, Figure 49, Figure 50, Figure 51, Figure 52, Figure 53, Figure 54, and Figure 55 all show the min and max average values for each condition at all tested flexion angles.

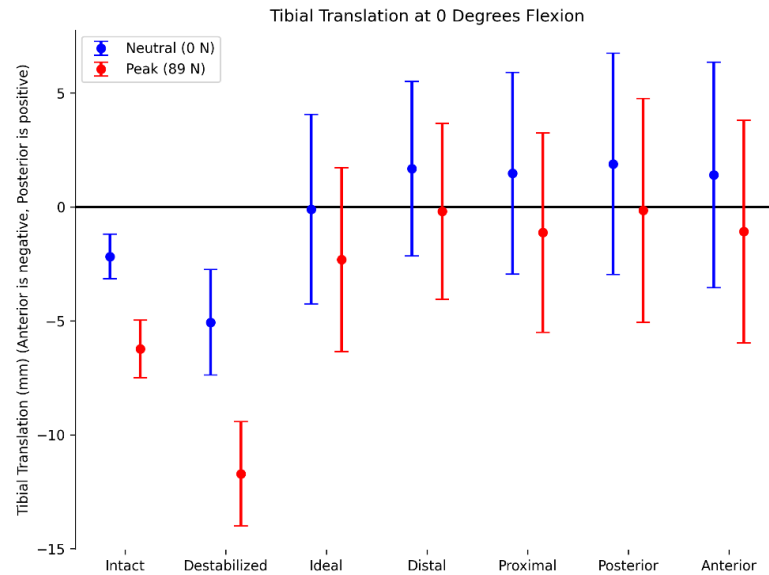


Figure 47. Tibial translation maximums and minimums by condition at 0 degrees flexion.

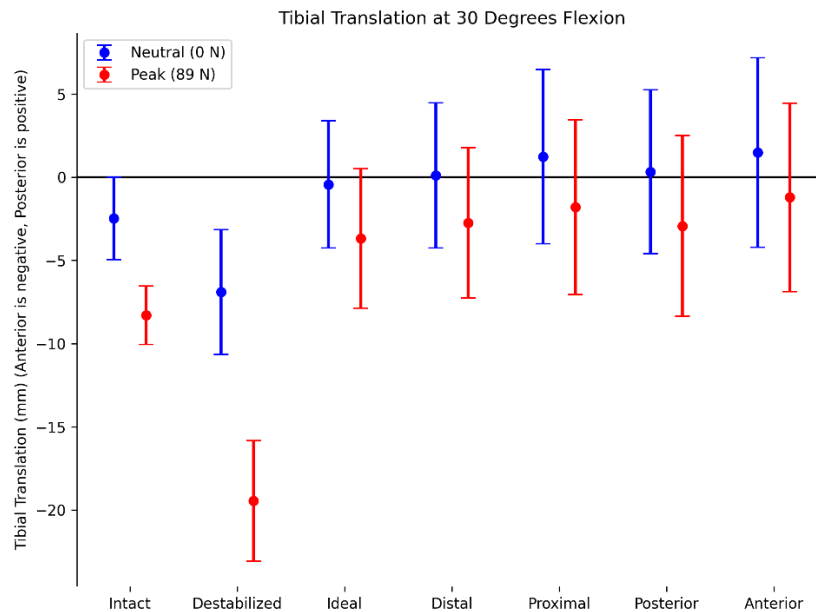


Figure 48. Tibial translation maximums and minimums by condition at 30 degrees flexion.

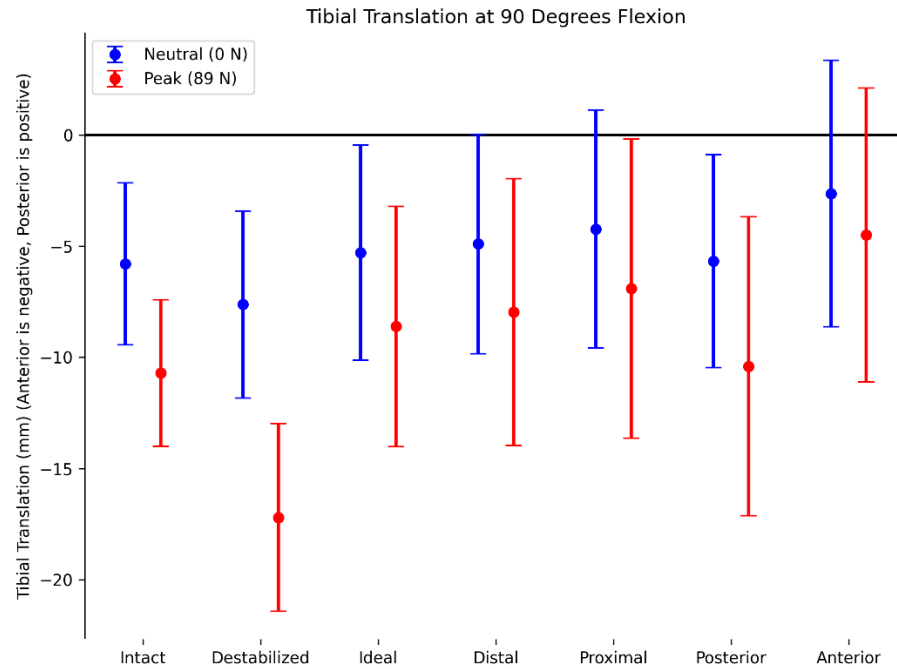


Figure 49. Tibial translation maximums and minimums by condition at 90 degrees flexion.

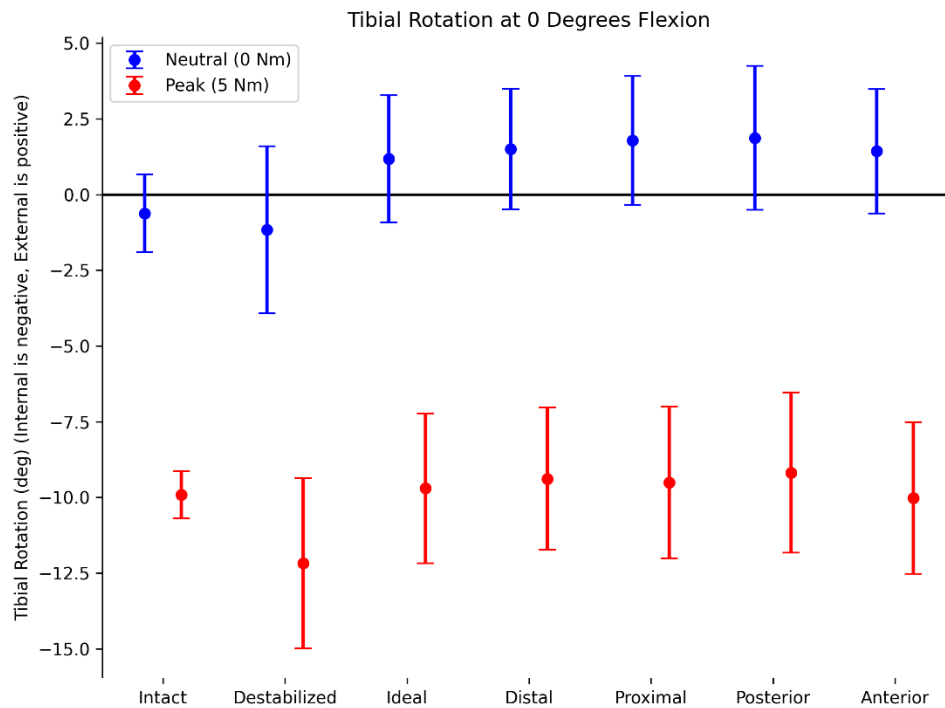


Figure 50. Tibial rotation maximums and minimums by condition at 0 degrees flexion.

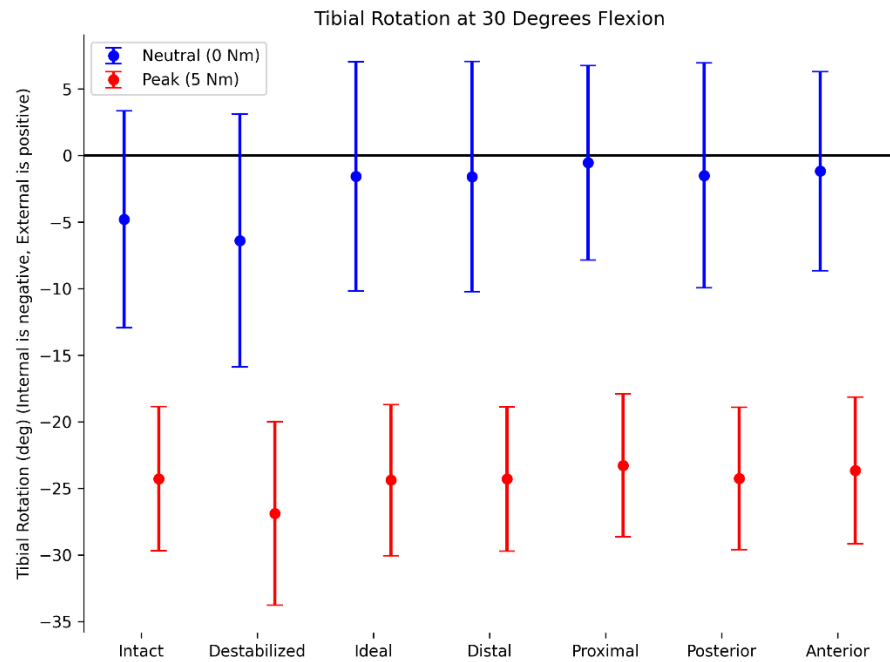


Figure 51. Tibial rotation maximums and minimums by condition at 30 degrees flexion.

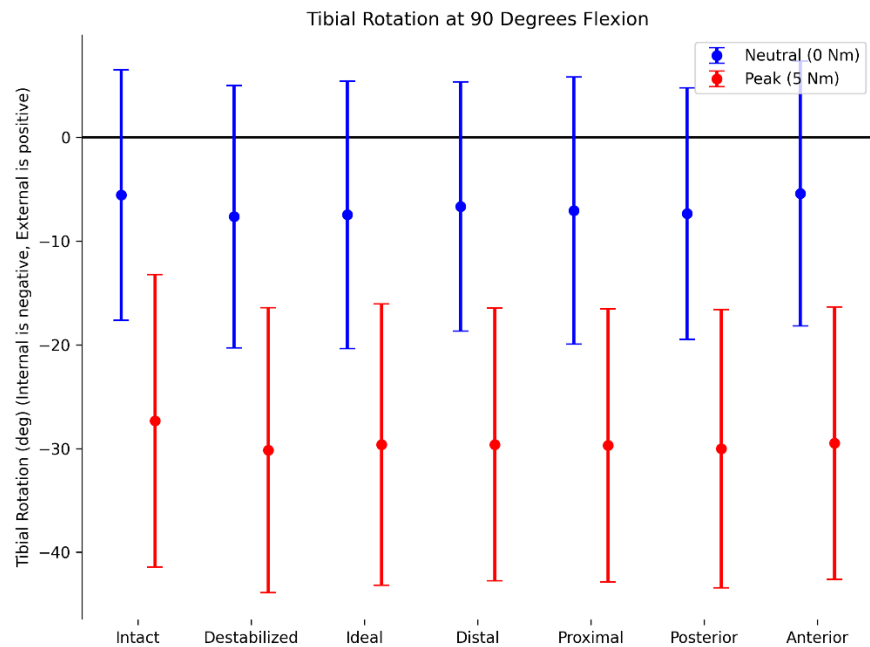


Figure 52. Tibial rotation maximums and minimums by condition at 90 degrees flexion.

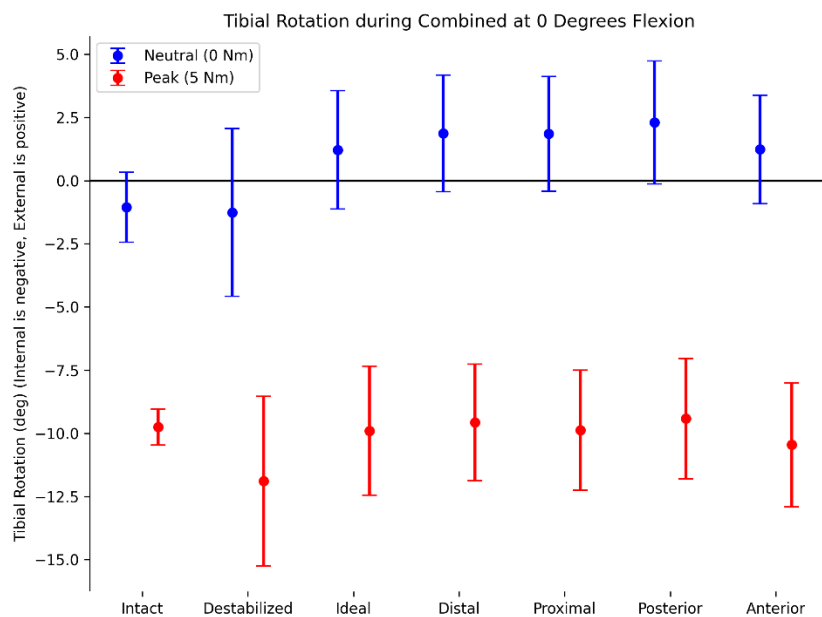


Figure 53. Tibial rotation during combined maximums and minimums by condition at 0 degrees flexion.

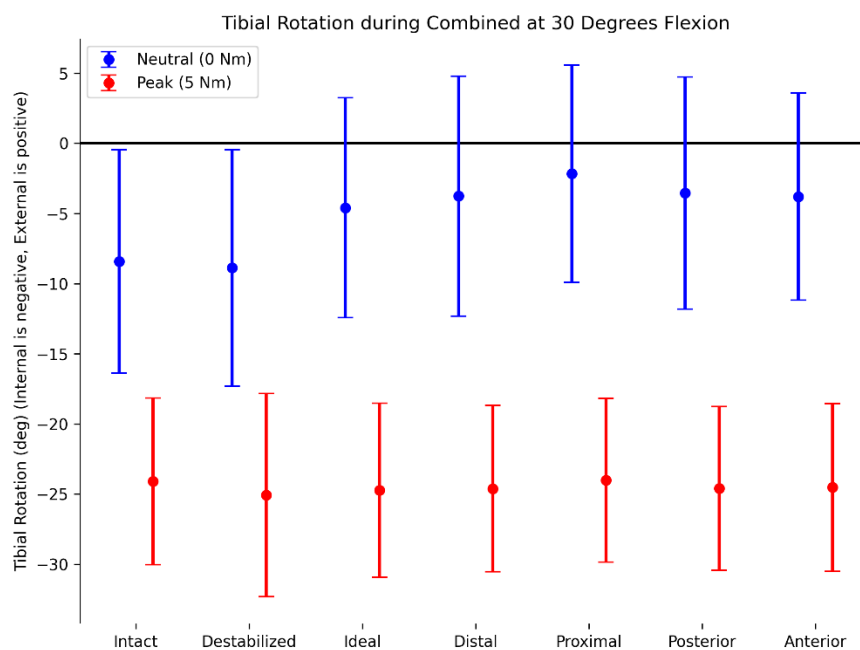


Figure 54. Tibial rotation during combined maximums and minimums by condition at 30 degrees flexion.

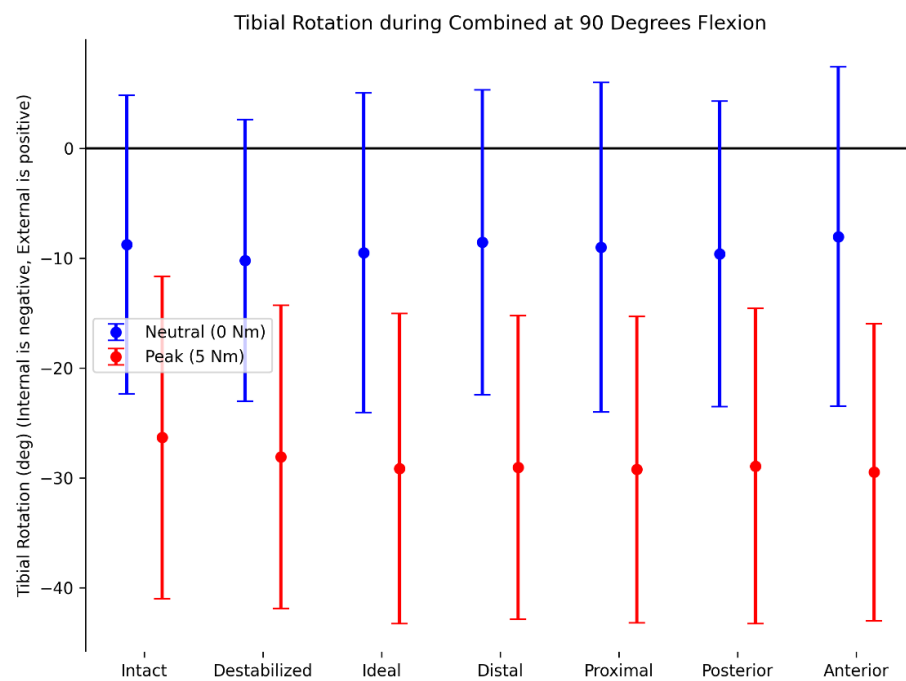


Figure 55. Tibial rotation during combined maximums and minimums by condition at 0 degrees flexion.

Appendix B

Graft Length Analysis Additional Figures, By Specimen

Figure 56, Figure 58, Figure 60, Figure 62, Figure 64, and Figure 66 show the graft length (mm) of each specimen compared to the knee flexion angle. On each graph, as well, is the zero load length (slack length). When the graft length is greater than the zero load length, the virtual ligament should carry a force. When the graft length is less than the zero load length, the virtual ligament should have no force. To easily compare the calculated slack lengths to the force in each virtual ligament condition by angle, a graph of the initial force grouped by angle is included beside each graft length plot, shown in Figure 57, Figure 59, Figure 61, Figure 63, Figure 65, and Figure 67.

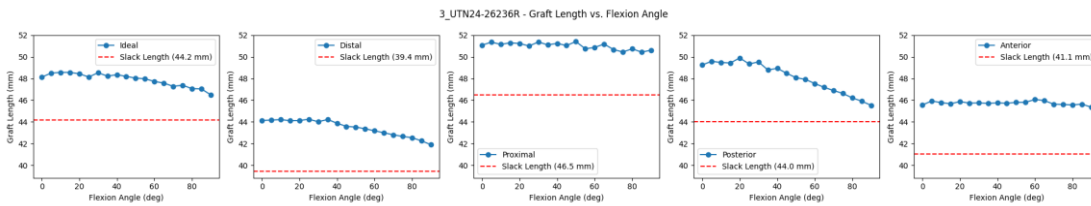


Figure 56. Graft length vs. knee flexion angle, specimen 3.

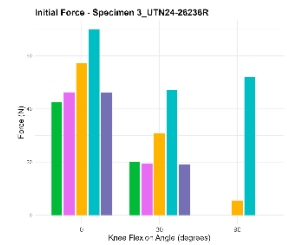


Figure 57. Initial Force by angle, specimen 3.

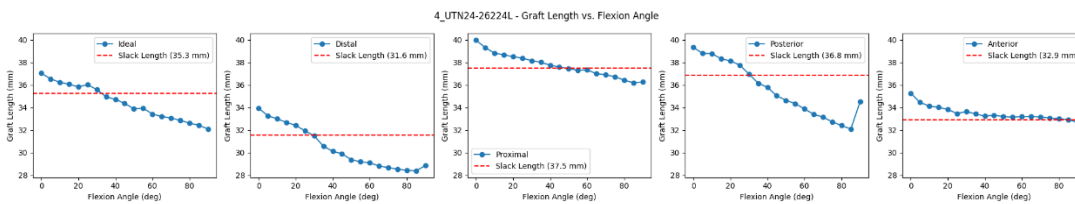


Figure 58. Graft length vs. knee flexion angle, specimen 4.

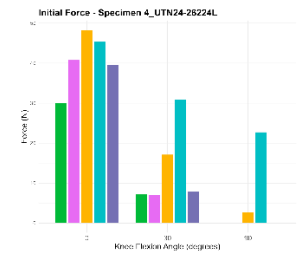


Figure 59. Initial Force by angle, specimen 4.

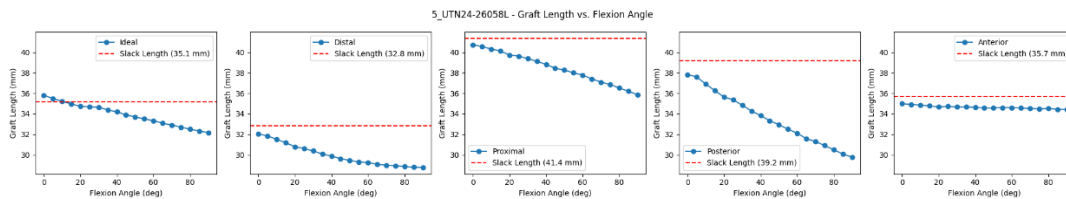


Figure 60. Graft length vs. knee flexion angle, specimen 5.

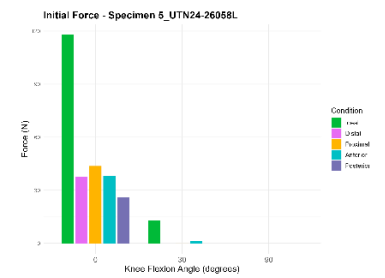


Figure 61. Initial Force by angle, specimen 5.

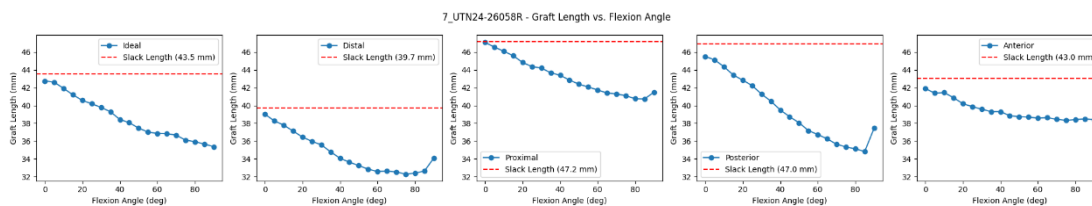


Figure 62. Graft length vs. knee flexion angle, specimen 7.

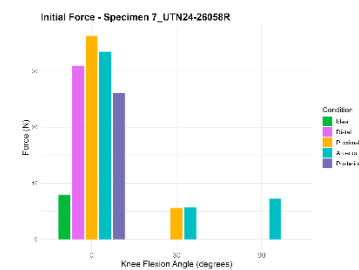


Figure 63. Initial Force by angle, specimen 7.

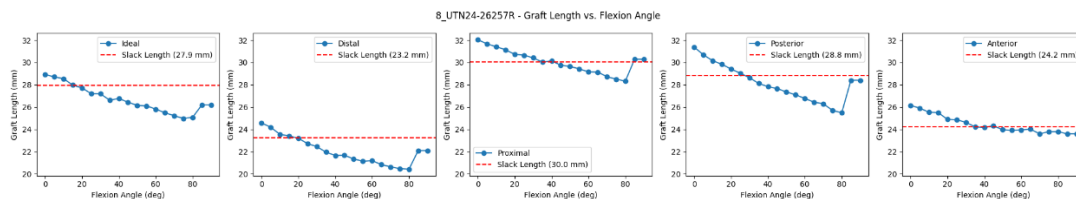


Figure 64. Graft length vs. knee flexion angle, specimen 8.

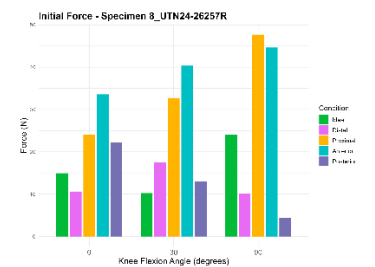


Figure 65. Initial Force by angle, specimen 8.

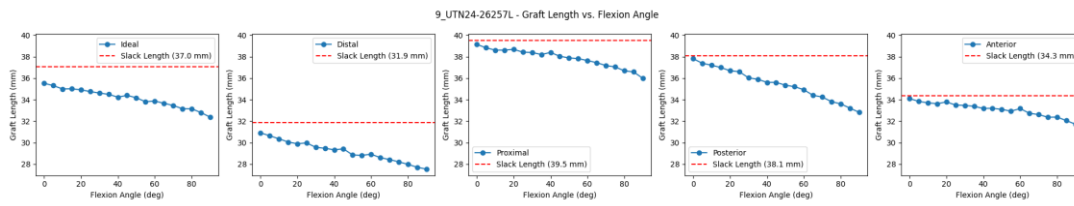


Figure 66. Graft length vs. knee flexion angle, specimen 9.

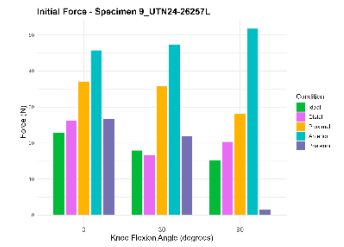


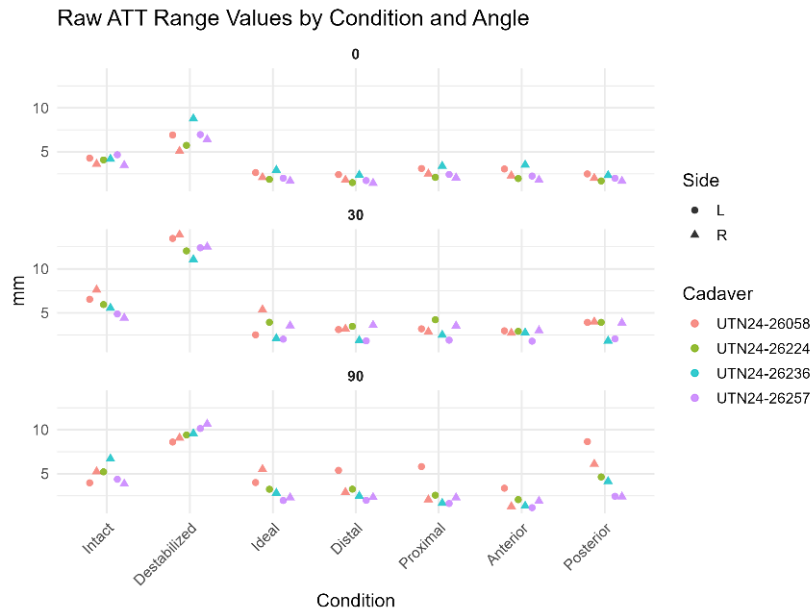
Figure 67. Initial Force by angle, specimen 9.

Appendix C

Statistical Analysis Additional Tables and Figures

This appendix features additional figures including figures of the raw data for each specimen, plots of the normality assumption checks, the ANOVA tables, and the significant differences for each LMM.

ATT:



S

Figure 68. Raw ATT range values, separated by cadaver number.

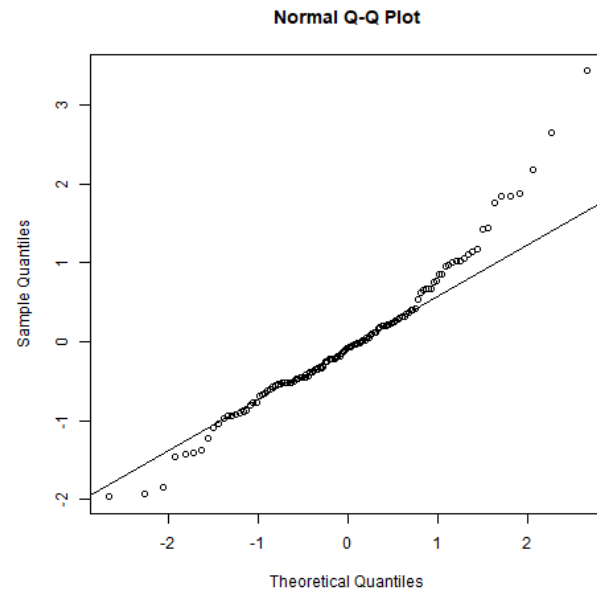


Figure 69. ATT range model Q-Q plot

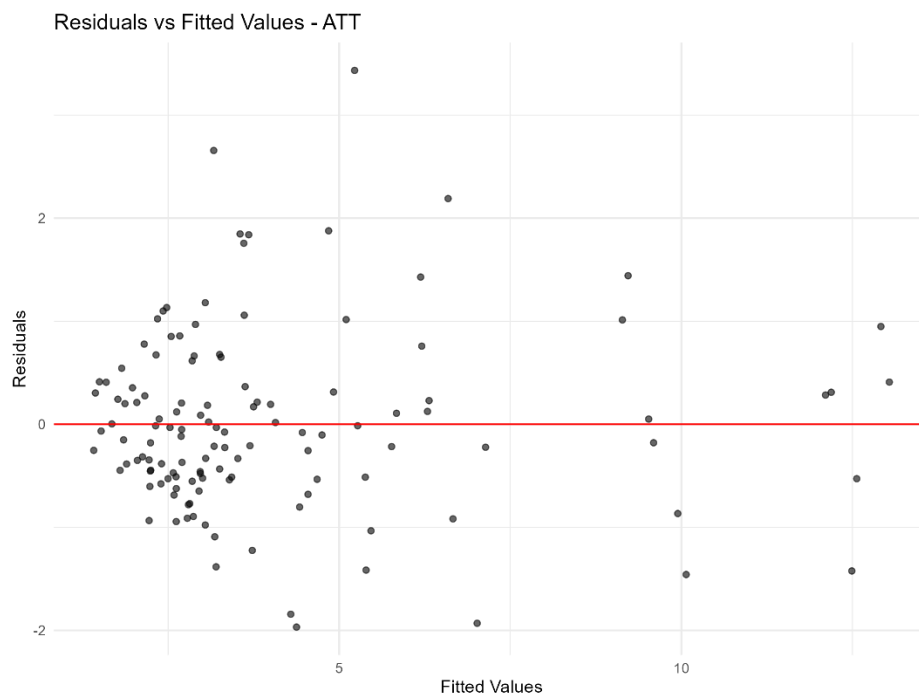


Figure 70. ATT range model residuals vs. fitted values

Table 7. Fixed Effects Summary (ATT Range Kinematics Model ANOVA)

Effect	Sum of Squares	Mean Squares	Numerator Degrees of Freedom	Denominator Degrees of Freedom	F Value	Pr(>F)
Condition	719.0874	119.8479	6	100	126.8626	1.83E-44
Angle	60.1597	30.07985	2	100	31.84043	2.00E-11
Condition: Angle	87.71998	7.309998	12	100	7.737854	5.50E-10

Table 8. Significant Comparisons: ATT Range

Group 1	Group 2	Angle	Estimate	SE	P Value
Intact	Destabilized	0	-2.59402	0.561161	0.000224
Intact	Ideal	0	1.846711	0.561161	0.022619
Intact	Distal	0	2.175267	0.561161	0.003484
Intact	Posterior	0	2.017692	0.561161	0.008847
Destabilized	Ideal	0	4.44073	0.561161	1.95E-10
Destabilized	Distal	0	4.769286	0.561161	1.25E-10
Destabilized	Proximal	0	4.046836	0.561161	2.38E-09
Destabilized	Anterior	0	4.163865	0.561161	9.43E-10
Destabilized	Posterior	0	4.611711	0.561161	1.38E-10
Intact	Destabilized	30	-6.72547	0.561161	1.21E-10
Intact	Ideal	30	2.583874	0.561161	0.000241
Intact	Distal	30	2.985446	0.561161	1.30E-05
Intact	Proximal	30	2.794205	0.561161	5.38E-05
Intact	Anterior	30	3.141077	0.561161	3.93E-06

Intact	Posterior	30	2.564759	0.561161	0.000275
Destabilized	Ideal	30	9.309344	0.561161	1.21E-10
Destabilized	Distal	30	9.710916	0.561161	1.21E-10
Destabilized	Proximal	30	9.519675	0.561161	1.21E-10
Destabilized	Anterior	30	9.866548	0.561161	1.21E-10
Destabilized	Posterior	30	9.29023	0.561161	1.21E-10
Intact	Destabilized	90	-4.67715	0.561161	1.31E-10
Intact	Distal	90	1.841482	0.561161	0.023248
Intact	Proximal	90	2.227024	0.561161	0.002532
Intact	Anterior	90	3.047705	0.561161	8.07E-06
Destabilized	Ideal	90	6.269442	0.561161	1.21E-10
Destabilized	Distal	90	6.518633	0.561161	1.21E-10
Destabilized	Proximal	90	6.904175	0.561161	1.21E-10
Destabilized	Anterior	90	7.724856	0.561161	1.21E-10
Destabilized	Posterior	90	4.845336	0.561161	1.24E-10
Proximal	Posterior	90	-2.05884	0.561161	0.006977
Anterior	Posterior	90	-2.87952	0.561161	2.87E-05

IR:

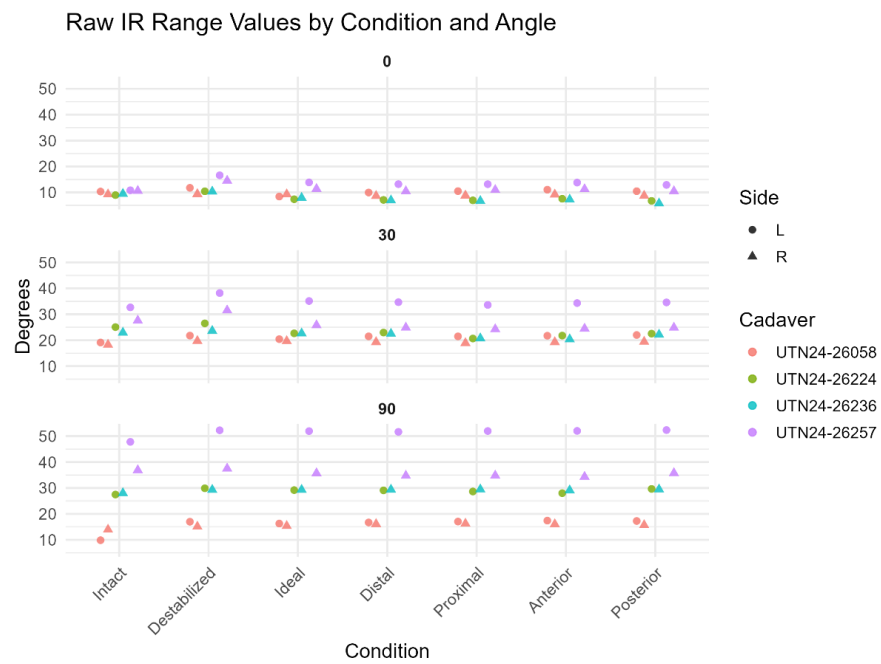


Figure 71. Raw IR max values, separated by cadaver number.

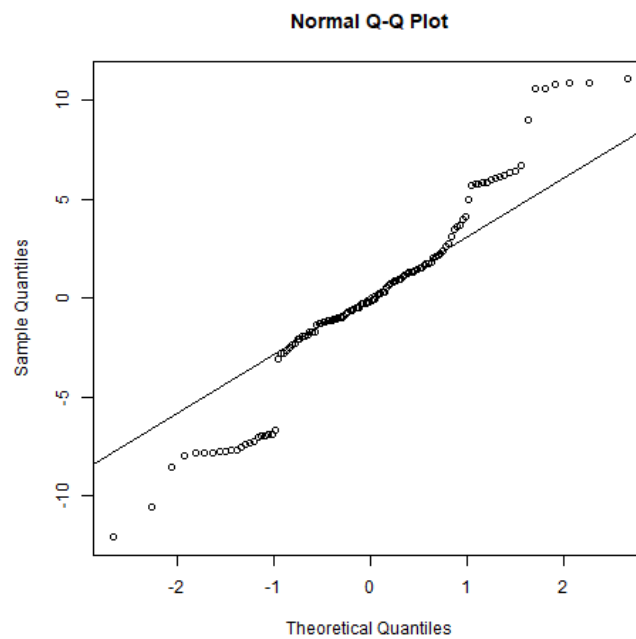


Figure 72. IR max model Q-Q plot

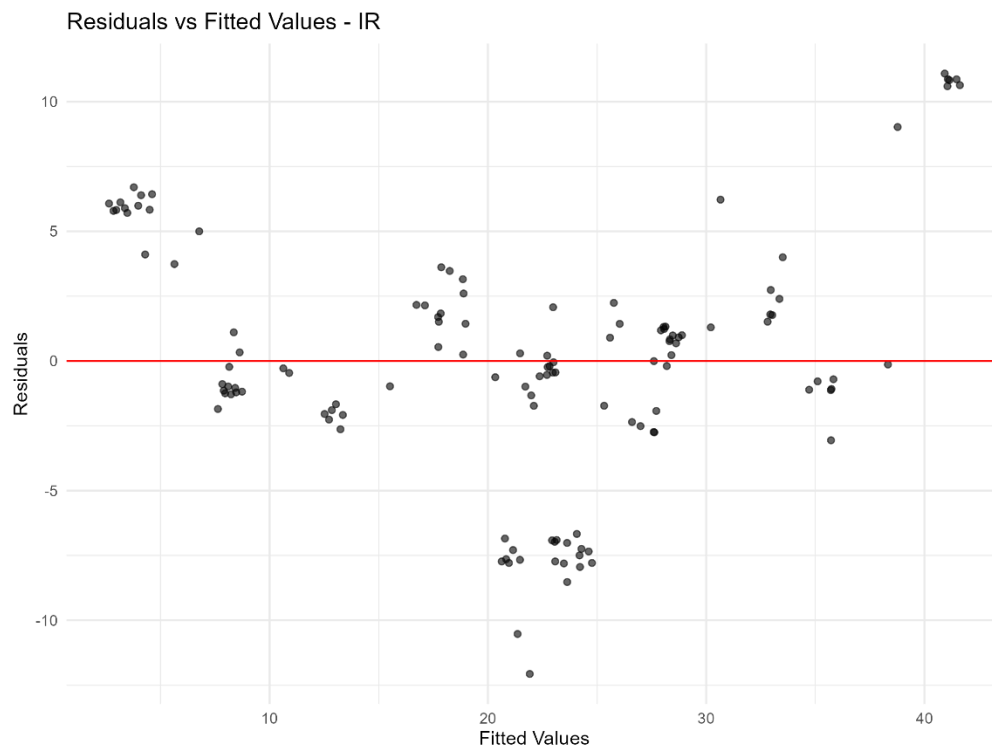


Figure 73. IR max model residuals vs. fitted values

Table 9. Fixed Effects Summary (IR Max Kinematics Model ANOVA)

Effect	Sum of Squares	Mean Squares	Numerator Degrees of Freedom	Denominator Degrees of Freedom	F Value	Pr(>F)
Condition	74.68258	12.4471	6	99.99999	0.442581	0.84865
Angle	8555.745	4277.873	2	99.99999	152.108	4.67E-31
Condition: Angle	43.10091	3.591743	12	99.99999	0.127711	0.999818

IR Max During Combined:

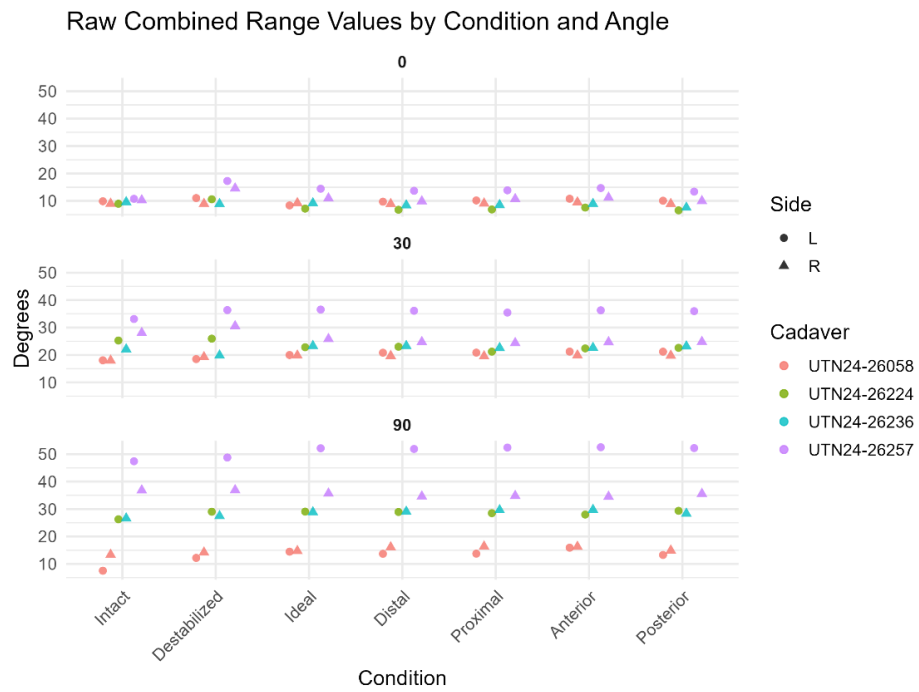


Figure 74. Raw IR Max During Combined values, separated by cadaver number.

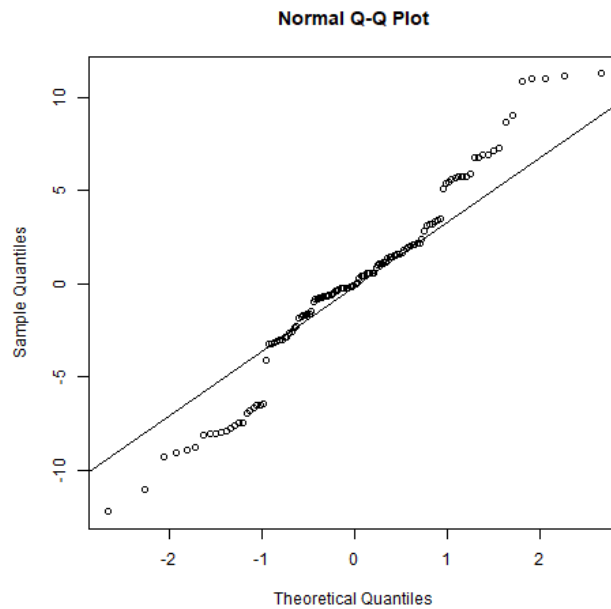


Figure 75. IR max during combined model Q-Q plot

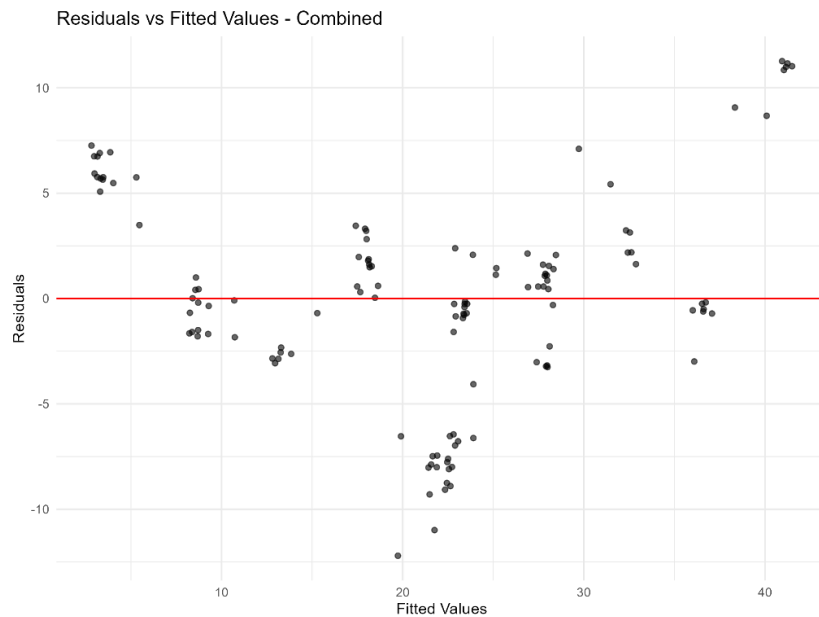


Figure 76. IR max during combined model residuals vs. fitted values

Table 10. Fixed Effects Summary (IR Max During Combined Kinematics Model ANOVA)

Effect	Sum of Squares	Mean Squares	Numerator Degrees of Freedom	Denominator Degrees of Freedom	F Value	Pr(>F)
Condition	29.11723	4.852872	6	100	0.160293	0.9865
Angle	7920.583	3960.292	2	100	130.8106	1.22E-28
Condition: Angle	44.76699	3.730582	12	100	0.123223	0.999849

Initial Force:

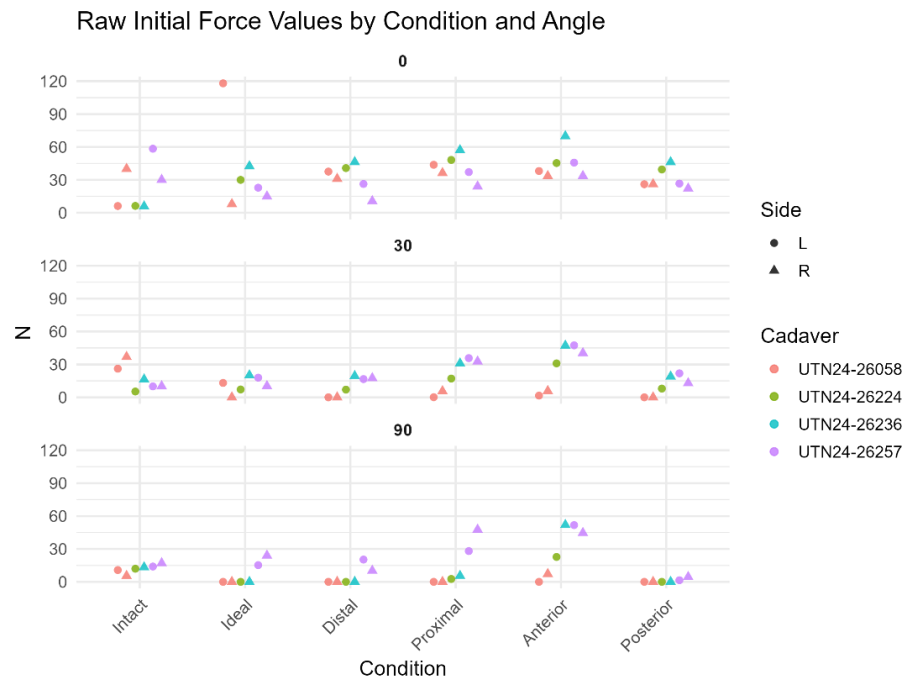


Figure 77. Raw initial force values, separated by cadaver number.

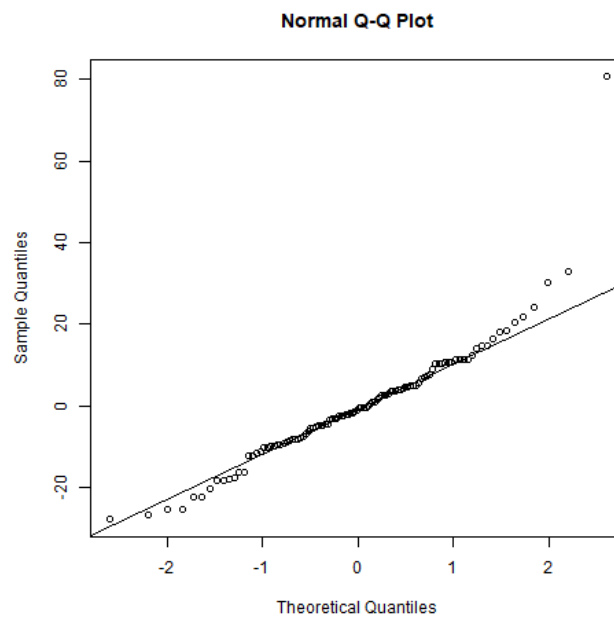


Figure 78. Initial force model Q-Q plot

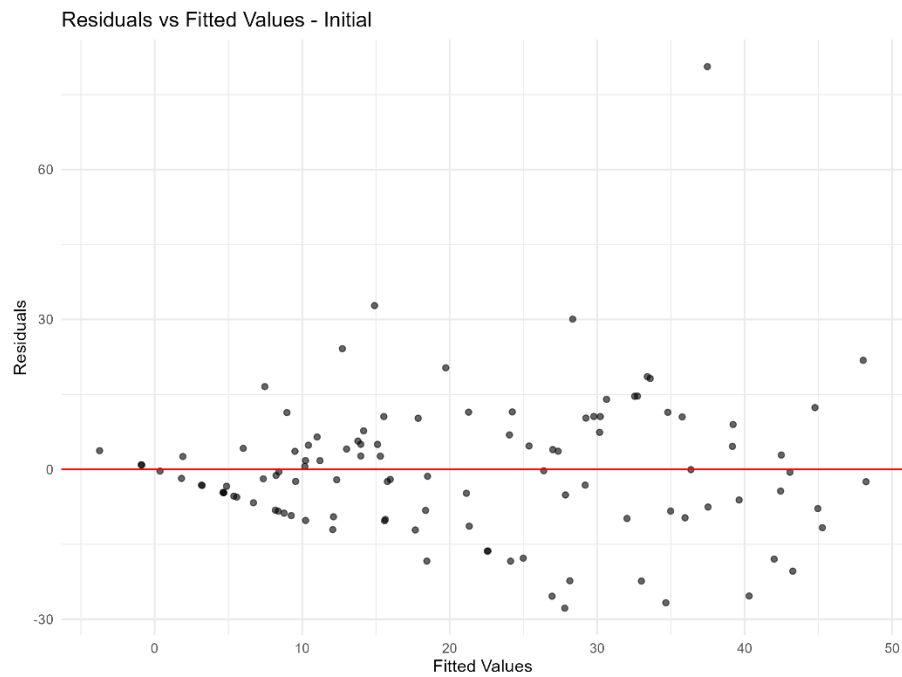


Figure 79. Initial force model residuals vs. fitted values

Table 11. Fixed Effects Summary (Initial Force Model ANOVA)

Effect	Sum of Squares	Mean Squares	Numerator Degrees of Freedom	Denominator Degrees of Freedom	F Value	Pr(>F)
Condition	5066.524	1013.305	5	85	4.099279	0.002206
Angle	11546.68	5773.339	2	85	23.35578	8.25E-09
Condition: Angle	1326.225	132.6225	10	85	0.536518	0.859655

Table 12. Significant Comparisons: Initial Force

Group 1	Group 2	Angle	Estimate	SE	P Value
Anterior	Posterior	90	28.72139	9.077279	0.025442242

Force During ATT:

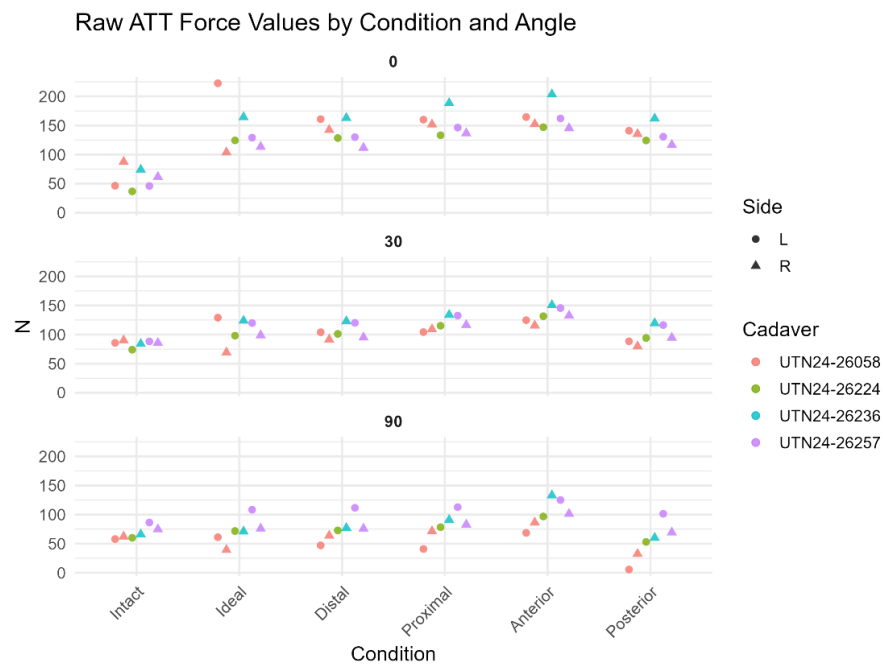


Figure 80. Raw force during ATT values, separated by cadaver number

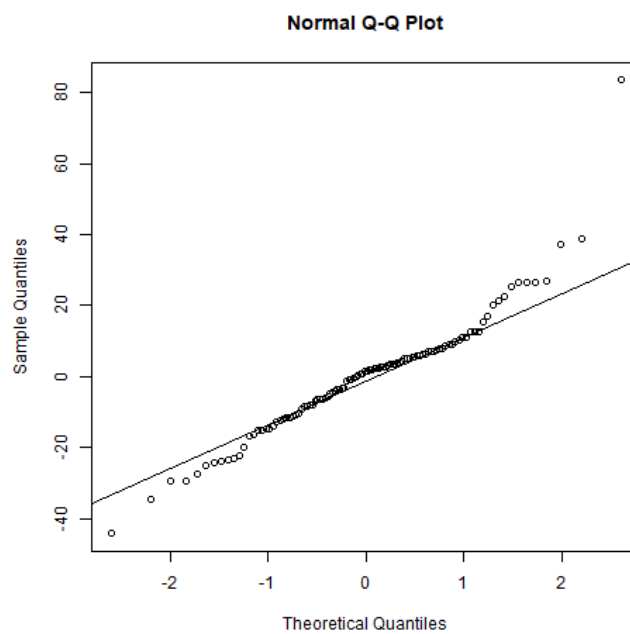


Figure 81. Force during ATT model Q-Q plot

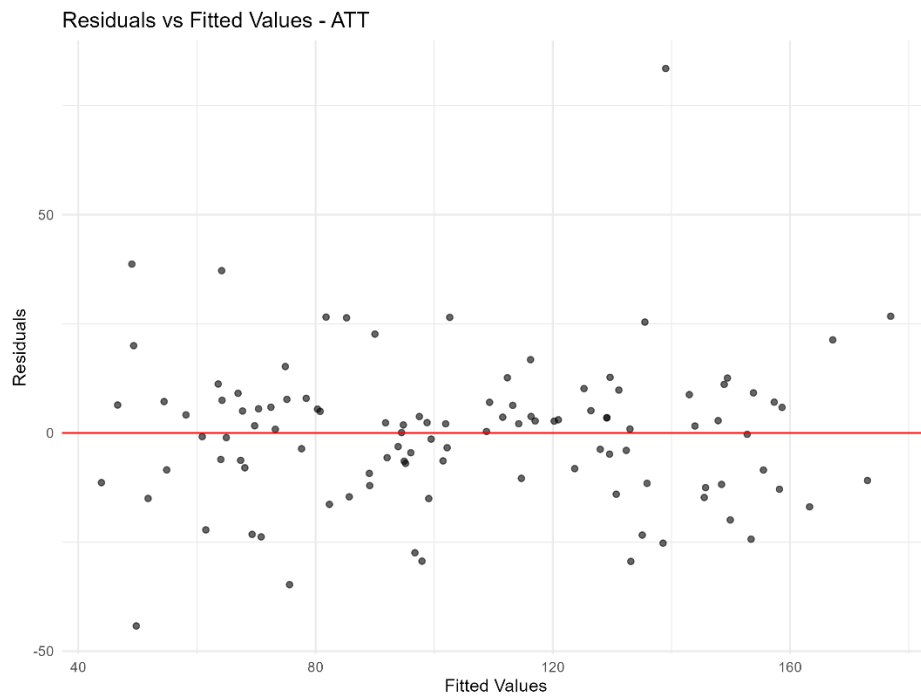


Figure 82. Force during ATT model residuals vs. fitted values

Table 13. Fixed Effects Summary (Force During ATT Model ANOVA)

Effect	Sum of Squares	Mean Squares	Numerator Degrees of Freedom	Denominator Degrees of Freedom	F Value	Pr(>F)
Condition	39397.82	7879.564	5	84.99977	22.5309	2.55E-14
Angle	59202.05	29601.03	2	84.99977	84.64146	5.95E-21
Condition: Angle	17982.78	1798.278	10	84.99977	5.142013	7.47E-06

Table 14. Significant Comparisons: Force During ATT

Group 1	Group 2	Angle	Estimate	SE	P Value
Intact	Ideal	0	-84.1287	10.79695	4.30E-10
Intact	Distal	0	-80.6248	10.79695	1.19E-09

Intact	Proximal	0	-93.9949	10.79695	2.12E-10
Intact	Anterior	0	-103.754	10.79695	2.08E-10
Intact	Posterior	0	-76.2364	10.79695	6.50E-09
Intact	Proximal	30	-33.9231	10.79695	0.027085
Intact	Anterior	30	-48.7916	10.79695	0.000281
Anterior	Posterior	30	34.62837	10.79695	0.0225
Intact	Anterior	90	-33.9149	10.79695	0.027143
Anterior	Posterior	90	48.1742	10.79695	0.000348

Force During IR:

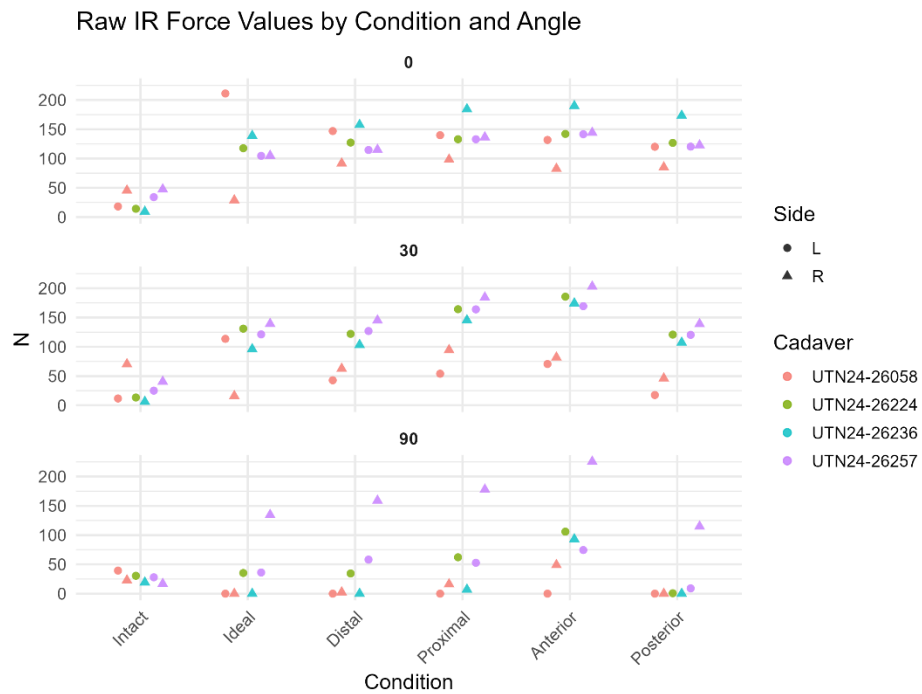


Figure 83. Raw force during IR values, separated by cadaver number

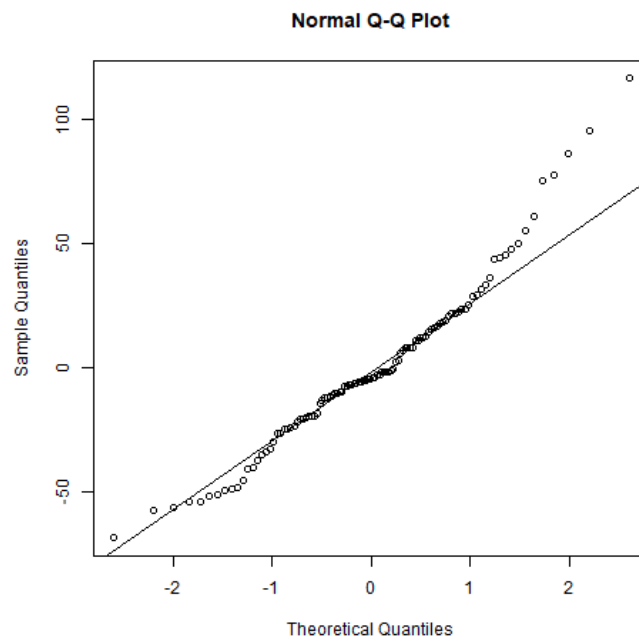


Figure 84. Force during IR model Q-Q plot

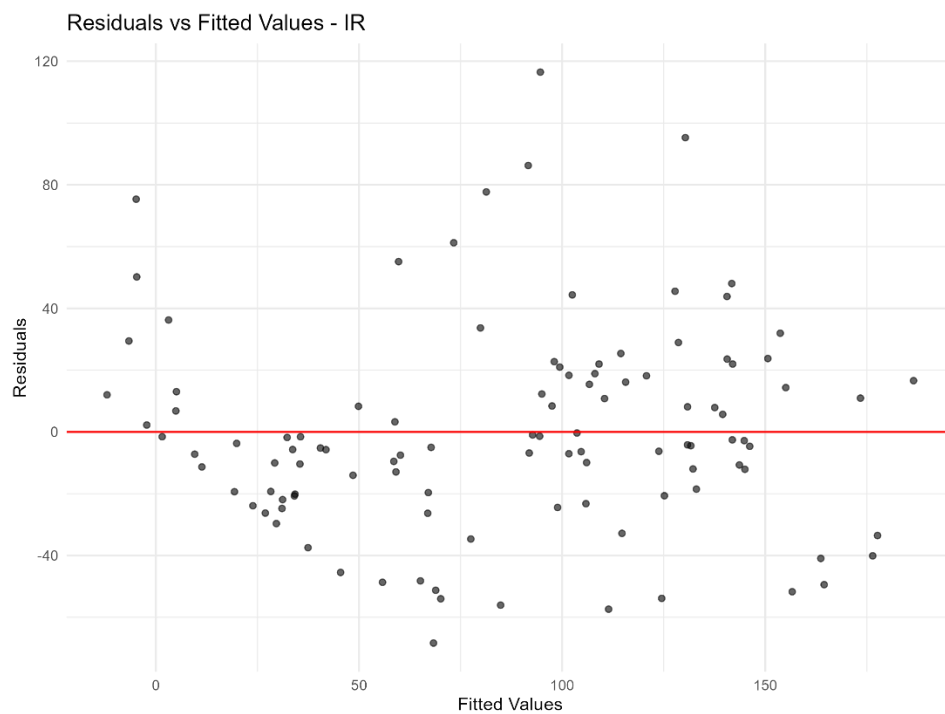


Figure 85. Force during IR model residuals vs. fitted values

Table 15. Fixed Effects Summary (Force During IR Model ANOVA)

Effect	Sum of Squares	Mean Squares	Numerator Degrees of Freedom	Denominator Degrees of Freedom	F Value	Pr(>F)
Condition	100348.6	20069.72	5	85	14.2823	3.88E-10
Angle	93986.67	46993.33	2	85	33.44206	1.93E-11
Condition: Angle	24159.72	2415.972	10	85	1.719288	0.089385

Table 16. Significant Comparisons: Force During IR

Group 1	Group 2	Angle	Estimate	SE	P Value
Intact	Ideal	0	-89.561	21.64268	0.001124
Intact	Distal	0	-97.4252	21.64268	0.0003
Intact	Proximal	0	-109.375	21.64268	3.55E-05
Intact	Anterior	0	-110.56	21.64268	2.85E-05
Intact	Posterior	0	-96.6004	21.64268	0.000346
Intact	Ideal	30	-74.9708	21.64268	0.010489
Intact	Distal	30	-72.5979	21.64268	0.014624
Intact	Proximal	30	-106.508	21.64268	6.00E-05
Intact	Anterior	30	-119.576	21.64268	5.19E-06
Intact	Posterior	30	-63.9496	21.64268	0.045128
Intact	Anterior	90	-65.202	21.64268	0.038667
Anterior	Posterior	90	70.57375	21.64268	0.01927

Force During Combined

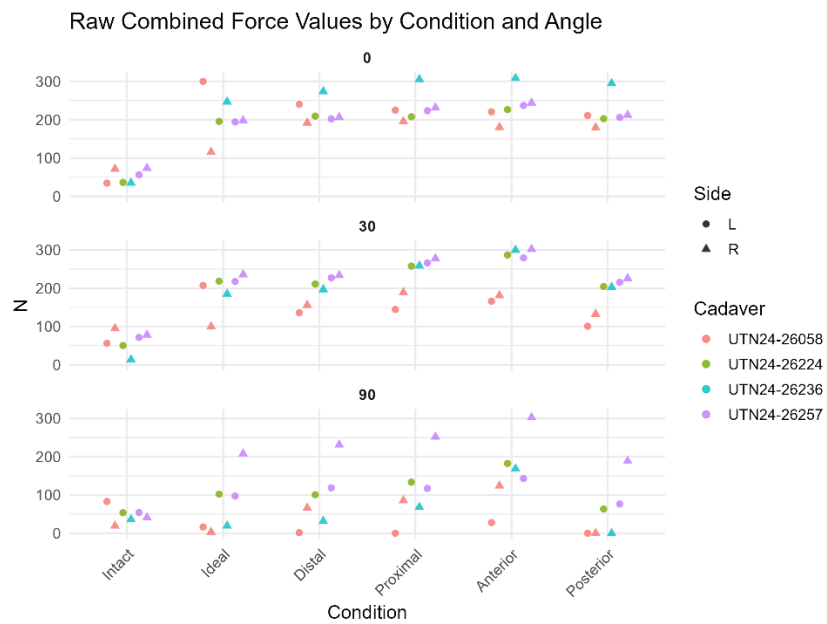


Figure 86. Raw force during Combined values, separated by cadaver number

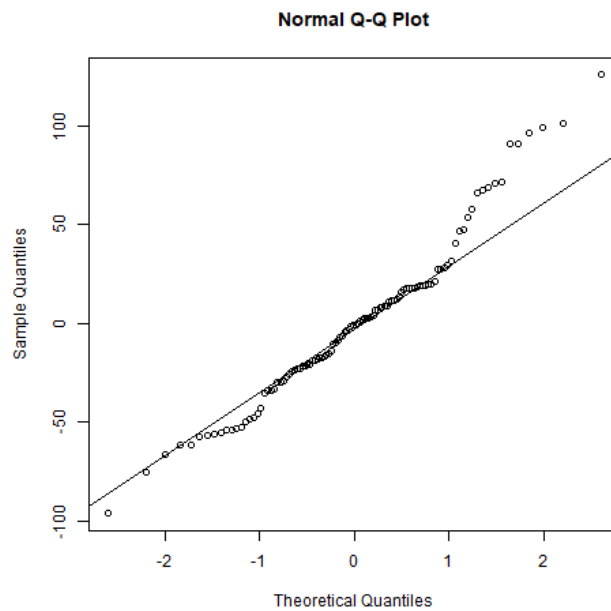


Figure 87. Force during combined model Q-Q plot

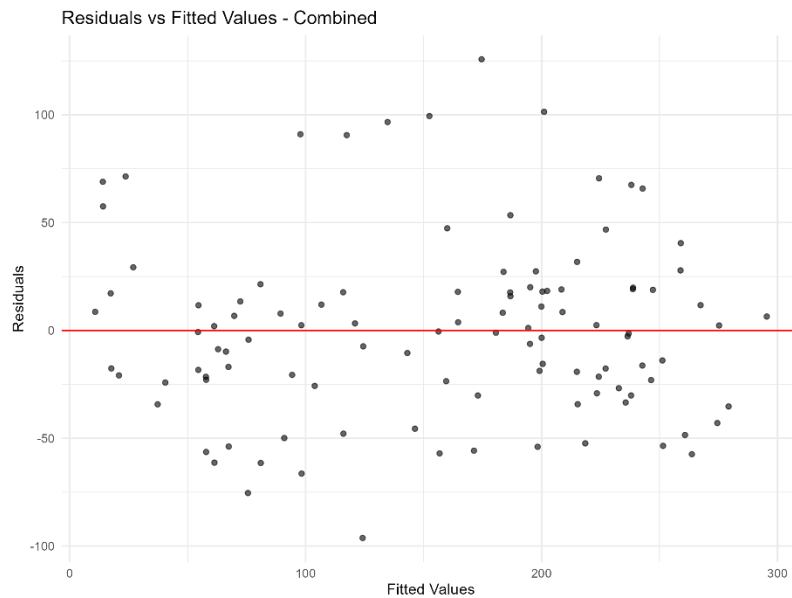


Figure 88. Force during combined model residuals vs. fitted values

Table 17. Fixed Effects Summary (Force During Combined Model ANOVA)

Effect	Sum of Squares	Mean Squares	Numerator Degrees of Freedom	Denominator Degrees of Freedom	F Value	Pr(>F)
Condition	279266	55853.21	5	85	25.87129	8.82E-16
Angle	244051.4	122025.7	2	85	56.52248	2.44E-16
Condition: Angle	54936.77	5493.677	10	85	2.544679	0.009783

Table 18. Significant Comparisons: Force During Combined

Group 1	Group 2	Angle	Estimate	SE	P Value
Intact	Ideal	0	-157.242	26.82591	1.25E-06
Intact	Distal	0	-169.45	26.82591	1.75E-07
Intact	Proximal	0	-180.242	26.82591	2.95E-08
Intact	Anterior	0	-185.021	26.82591	1.34E-08
Intact	Posterior	0	-166.546	26.82591	2.80E-07
Intact	Ideal	30	-133.101	26.82591	5.12E-05
Intact	Distal	30	-132.653	26.82591	5.46E-05
Intact	Proximal	30	-171.418	26.82591	1.27E-07
Intact	Anterior	30	-191.638	26.82591	4.53E-09
Intact	Posterior	30	-119.43	26.82591	0.000361
Intact	Anterior	90	-110.145	26.82591	0.00126
Ideal	Anterior	90	-83.6616	26.82591	0.028906
Anterior	Posterior	90	103.3	26.82591	0.003024

Graft Length Change

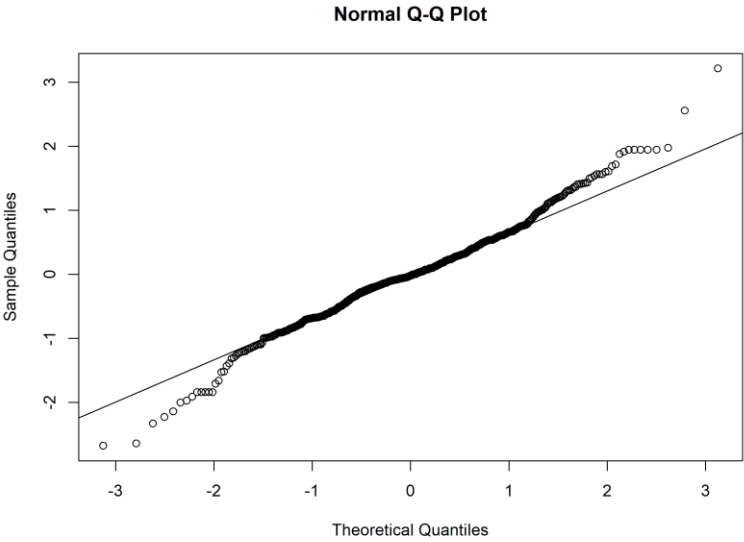


Figure 89. Graft length change model Q-Q plot

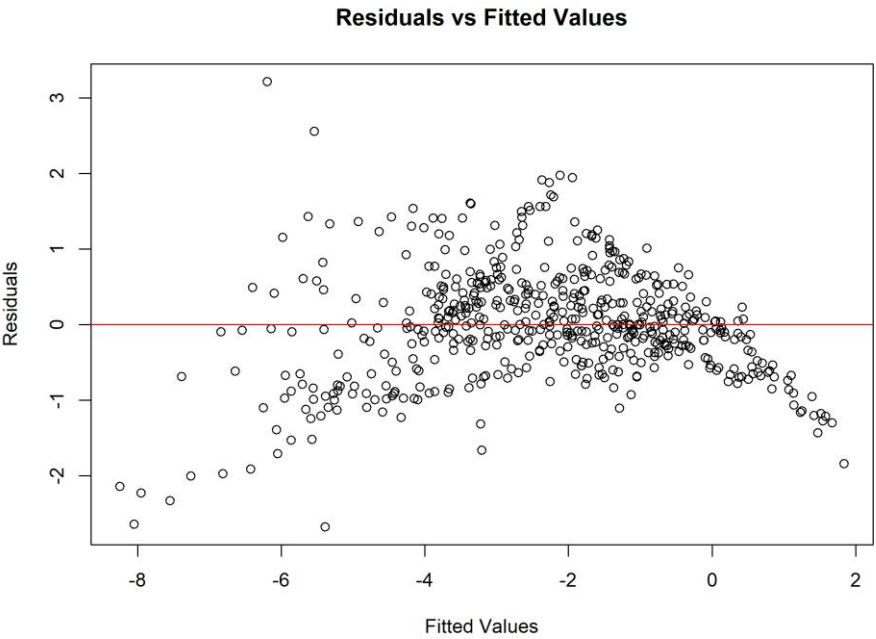


Figure 90. Graft length change model residuals vs. fitted values

Table 19. Fixed Effects Summary (Graft Length Change ANOVA)

Effect	Sum of Squares	Mean Squares	Numerator Degrees of Freedom	Denominator Degrees of Freedom	F Value	Pr(>F)
Condition	315.0074	78.75184	4	470.0003	112.2314	4.32e-67
Angle	856.8764	47.60424	18	470.0003	67.84209	4.20e-118
Condition: Angle	145.6976	2.023577	72	470.0003	2.883855	8.64e-12

Table 20. Significant Comparisons: Graft Length Change

Group 1	Group 2	Angle	Estimate	SE	P Value
Anterior	Posterior	35	1.688445	0.483629	0.004763
Ideal	Posterior	40	1.352215	0.483629	0.042693
Distal	Anterior	40	-1.32884	0.483629	0.048759
Proximal	Posterior	40	1.549859	0.483629	0.012517
Anterior	Posterior	40	1.98658	0.483629	0.000452
Ideal	Posterior	45	1.575188	0.483629	0.010558
Distal	Anterior	45	-1.45947	0.483629	0.022439
Proximal	Posterior	45	1.653704	0.483629	0.006116
Anterior	Posterior	45	2.375886	0.483629	1.23E-05
Ideal	Posterior	50	1.651584	0.483629	0.006209
Distal	Anterior	50	-1.698	0.483629	0.004442
Proximal	Posterior	50	1.951469	0.483629	0.000607
Anterior	Posterior	50	2.681799	0.483629	4.89E-07
Ideal	Posterior	55	1.838712	0.483629	0.001521

Distal	Anterior	55	-1.84731	0.483629	0.001421
Proximal	Posterior	55	2.029019	0.483629	0.000314
Anterior	Posterior	55	3.001446	0.483629	1.19E-08
Ideal	Anterior	60	-1.45686	0.483629	0.022809
Ideal	Posterior	60	2.003333	0.483629	0.000391
Distal	Anterior	60	-1.99489	0.483629	0.000421
Distal	Posterior	60	1.465302	0.483629	0.021635
Proximal	Posterior	60	2.270048	0.483629	3.46E-05
Anterior	Posterior	60	3.460193	0.483629	5.49E-11
Ideal	Anterior	65	-1.5782	0.483629	0.010344
Ideal	Posterior	65	2.25535	0.483629	3.98E-05
Distal	Anterior	65	-2.1303	0.483629	0.000128
Distal	Posterior	65	1.703254	0.483629	0.004274
Proximal	Posterior	65	2.564041	0.483629	1.76E-06
Anterior	Posterior	65	3.833554	0.483629	2.26E-11
Ideal	Anterior	70	-1.59812	0.483629	0.009027
Ideal	Posterior	70	2.336633	0.483629	1.81E-05
Distal	Anterior	70	-2.09049	0.483629	0.000183
Distal	Posterior	70	1.844266	0.483629	0.001455
Proximal	Anterior	70	-1.36492	0.483629	0.039676
Proximal	Posterior	70	2.569838	0.483629	1.65E-06
Anterior	Posterior	70	3.934754	0.483629	2.25E-11
Ideal	Anterior	75	-1.77678	0.483629	0.002463

Ideal	Posterior	75	2.50018	0.483629	3.45E-06
Distal	Anterior	75	-2.19104	0.483629	7.30E-05
Distal	Posterior	75	2.085928	0.483629	0.00019
Proximal	Anterior	75	-1.50837	0.483629	0.016439
Proximal	Posterior	75	2.768591	0.483629	1.84E-07
Anterior	Posterior	75	4.276964	0.483629	2.25E-11
Ideal	Anterior	80	-1.92089	0.483629	0.000783
Ideal	Posterior	80	2.661512	0.483629	6.11E-07
Distal	Anterior	80	-2.27734	0.483629	3.23E-05
Distal	Posterior	80	2.305061	0.483629	2.47E-05
Proximal	Anterior	80	-1.71392	0.483629	0.003951
Proximal	Posterior	80	2.868485	0.483629	5.83E-08
Anterior	Posterior	80	4.582401	0.483629	2.25E-11
Ideal	Anterior	85	-1.81987	0.483629	0.001764
Ideal	Posterior	85	2.475728	0.483629	4.44E-06
Distal	Anterior	85	-1.98661	0.483629	0.000451
Distal	Posterior	85	2.308986	0.483629	2.37E-05
Proximal	Anterior	85	-1.46961	0.483629	0.021057
Proximal	Posterior	85	2.825988	0.483629	9.55E-08
Anterior	Posterior	85	4.295599	0.483629	2.25E-11
Ideal	Anterior	90	-1.95038	0.483629	0.000613
Ideal	Posterior	90	1.526077	0.483629	0.014648
Distal	Anterior	90	-1.59463	0.483629	0.009246

Distal	Posterior	90	1.881827	0.483629	0.001077
Proximal	Posterior	90	2.180784	0.483629	8.03E-05
Anterior	Posterior	90	3.476457	0.483629	4.84E-11

Glossary

Anterior	Front or forward part of the body or structure.
Anterior Tibial Translation	Movement of the tibia forward relative to the femur.
Biomechanics	The study of mechanical principles that define the way joints and muscles move together.
Degree of Freedom	Number of independent movements a joint can make.
Distal	Farther away from the center of the body.
Extension	Movement that increases the angle between two body parts.
External Rotation	Movement where a body part rotates out, away from midline of the body.
Flexion	Movement that decreases the angle between two body parts.
Internal Rotation	Movement where a body part rotates inwards towards the midline of the body.
In Vitro	Outside a living organism, typically in a laboratory setting.
Isometric	The distance between ligament insertions is constant over a joint range of motion.
Ligament	Fibrous connective tissue that connects bone to other bones.
Neutral	Unstrained position of a joint or a body part.
Posterior	Back or towards the back of the body or structure.
Posterior Tibial Translation	Movement of the tibia backwards relative to the femur.
Proximal	Closer to the center of the body.
Subluxation	State where a bone is partially pulled or pushed out of place.