A Computational Assessment of Lisfranc Injuries and their Surgical Repairs

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A COMPUTATIONAL ASSESSMENT OF LISFRANC INJURIES AND THEIR SURGICAL REPAIRS

A thesis submitted in partial fulfillment of the requirements for the degree of Masters of Science in Biomedical Engineering at Virginia Commonwealth University.

by

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List of Abbreviations

General Abbreviations

2D Two-Dimensional
3D Three-Dimensional
ADAMS Automatic Dynamic Analysis of Mechanical Systems
AOFAS American Orthopedic Foot and Ankle Score
AP Anteroposterior
BW Body Weight
CADD Computer-Aided Drafting and Design
CT Computerized Tomography
DICOM Digital Imaging and Communications in Medicine format
DOF Degree of Freedom
DP Dorsoplantar
EMG Electromyography
FEA Finite Element Analysis
GSTIFF Gear Stiff Numerical Integrator
HU Hounsfield Unit
ISB International Society of Biomechanics
MCO Medailizing Calcaneal Osteotomy
mm Millimeter
N/mm Newtons per Millimeter
RBM Rigid Body Motion
STL Stereolithographic

Anatomical Abbreviations

Cal Calcaneus
CN1 Medial Cuneiform
CN2 Intermediary Cuneiform
CN3 Lateral Cuneiform
MT1 First Metatarsal
MT2 Second Metatarsal
Nav Navicular
DEL_* Denotes structure is part of the deltoid tissue
DIST_* Denotes structure is part of the distal metatarsal tissue
DOR_* Denotes structure is part of the dorsal ligament network of the foot
IOL_* Denotes structure is part of the talocalcaneal interosseous tissue
IOM_* Denotes structure is part of the tibiofibular interosseous membrane
LCL_* Denotes structure is part of the lateral collateral ligaments
MCL_* Denotes structure is part of the medial collateral ligaments
PLAN_* Denotes structure is part of the plantar ligament network of the foot
PROX_* Denotes structure is part of the proximal tibiofibular capsule
FDL Flexor Digitorum Longus
FHL Flexor Hallucis Longus
IMCN Inferomedial Calcaneonavicular
INTCn Intercuneiform Ligament
INTMt Intermetatarsal Ligament
IOL Talocalcaneal Interosseous Ligaments
IOM Interosseous Membrane
LCL Lateral Collateral Ligaments
LPL Long Plantar Ligament
MCL Medial Collateral Ligaments
MCN Middle Calcaneonavicular Ligament
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<tr>
<td>PB</td>
<td>Peroneus Brevis</td>
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<tr>
<td>PL</td>
<td>Peroneus Longus</td>
</tr>
<tr>
<td>PT</td>
<td>Posterior Tibialis</td>
</tr>
<tr>
<td>SFR</td>
<td>Superficial Fibular Retinaculum</td>
</tr>
<tr>
<td>SMCN</td>
<td>Superomedial Calcaneonavicicular Ligament</td>
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While Lisfranc injuries in the mid foot are less common than other ankle and mid foot injuries, they pose challenges in both properly identifying them and treating them. When Lisfranc injuries are ligamentous and do not include obvious fractures, they are very challenging for clinicians to identify unless weight bearing radiographs are used. The result is that 20%-40% of Lisfranc injuries are missed in the initial evaluation. Even when injuries are correctly identified the outcomes of surgical procedures remain poor. Existing literature has compared the different surgical procedures but has not had a standard approach or procedures across studies. This study uses a computational biomechanical model validated on a cadaveric study to evaluate factors that impact injury presentation and to compare the different procedures ability to stabilize the Lisfranc joint after an injury. Using SolidWorks® a rigid body kinematic model of a healthy human foot was created whereby the 3D bony anatomy, articular contacts, and soft tissue restraints guided biomechanical function under
the action of external perturbations and muscle forces. The model was validated on a cadaveric study to ensure it matched the behavior of a healthy Lisfranc joint and one with a ligamentous injury. The validated model was then extended to incorporate muscle forces and different foot orientations when simulating a weight bearing radiograph. The last section of work was to compare the stability of four different surgical repairs for Lisfranc injuries. These procedures were three open reduction and internal fixation (ORIF) procedures with different hardware (screws, screws and dorsal plates, and endobuttons) and primary arthrodesis with screws. They required use of finite element analysis which was performed in Ansys Workbench. For the presentation of injuries, both muscle forces and standing with inversion or eversion could reduce the diastasis (separation) observed for weight bearing radiographs and thus confuse the diagnosis. When comparing the different surgical procedures, the ORIF with screws and primary arthrodesis with screws showed the most stable post-operative Lisfranc joint. However, the use of cannulated screws for fixation showed regions of high stress that may be susceptible to breakage. A challenge in the literature has been the use of different experimental designs and metrics when comparing two of the possible procedures for a Lisfranc injury head to head. This study has been able to benchmark four procedures using the same model and set of metrics. Since none of the existing procedures showed consistently good to excellent patient outcomes, more procedures could be proposed in the future. If this were to occur, this study offers a standard procedure for benchmarking the new procedure’s post-operative mechanical stability versus those procedures currently in use.
Chapter 1: Introduction

1.1 Overview of Biomechanics Research

The human body is a complex and dynamically changing system that is constantly experiencing physical forces and pressures during day to day activities. The resulting stress and strain that the body’s structures experience have long been known to impact how the structures respond and remodel which can either be beneficial or detrimental [1]. The normal cyclic loading of tissue such as bone or tendon promotes natural growth and remodeling of these tissue [2]. Higher blood pressure however leads to hypertrophy of the heart [3]. Beyond experiencing external stimuli, the body also generates forces and motion. Biomechanics is a multiscale study of living organisms in response to both external forces and those generated by the organism itself. While today biomechanics is used in studying both healthy and diseased tissues in many systems of the body, one of the earliest areas of biomechanical study was in the musculoskeletal system.

Since the musculoskeletal system provides the structural frame for supporting the body and the network of force generating components, biomechanics can describe these two roles of the system. When the biomechanics of a portion of the musculoskeletal system is disrupted as a result of an injury or pathology, understanding the change becomes important in order to reverse the negative impacts that may result. For example, sprained ligaments have reduced stiffness meaning without proper healing and treatment, the joint spanned by the ligament may be under-constrained or unstable. For lateral ankle sprains in athletes, up to 80% of individuals will have chronic laxity in the ankle joint [4]. Research into the biomechanics of the body focuses on understanding these normal and injured states in order to have improved patient outcomes for treatments. Traditional experiments using animal, human, and cadaver models have all been used to study biomechanics. However, as more information about
material properties of the biological structures have become available and computational software has expanded options, computational models have used this information to further study biomechanics in computational experiments.

1.2 Computational Modeling

Computer aided drafting and design (CADD) uses computer software to design biomechanical computational models. These models combine physical properties, object geometries, and forces to create a mathematical representation of the system being studied. These models are then used in simulations to determine the responses of the system under normal conditions and then to an injury/change. The increased computational power available to researchers as well as more software suites that support drafting have lowered the barrier to entry for many laboratories. In addition, larger databases of tissue geometry and material properties mean that researchers have greater access to foundational information that can be used in the creation of a computational model.

Besides the lower barriers to using computational models, they have practical advantages. These include the ability to re-run simulations with single small changes. They also allow tissue and simulation parameters to be more easily controlled than in a cadaver or patient study. Lastly, there are lower costs with running computational simulations than running a wet experimental lab. In this study, the advantages of computational models were used to further the understanding of the Lisfranc joint in the foot.

1.2.1 Types of Computational Modeling

In all computational models, certain known parameters must be available in order to solve for the unknowns in the simulation. Two types of simulations used in computational biomechanics are inverse dynamics and forward dynamics. The difference between these simulations is what is known versus what is unknown. Inverse dynamics uses known position
information of a specific point on the body tracked throughout a motion to solve for the forces required to create the recorded motion. Forward dynamics by comparison is focused on predicting a motion as a result of known forces.

Another variation of computational mechanics is whether a part in the model is permitted to undergo deformations or not. Some materials such as elastics show large deformation without failing while others such as concrete largely maintain their dimensions up until the point of failure. When capturing the deformation or stress in the structures of a computational model is vital, a technique called finite element analysis is used. This technique will allow forces to deform parts of the model if the forces are sufficiently large. By comparison, if the goals of the analysis are not comprised by not allowing parts to be deformed, then rigid body motion analysis can be used.

This thesis is focused on forward dynamic modeling with both rigid body modeling (RBM) and finite element analysis (FEA) to answer different questions of the investigation.

1.2.2 Software Packages

Since biomechanics requires solving a set of equations for many different points of time or locations in space, computers are the only viable option to use for the repeated calculations. Computers can repeatedly and efficiently solve the constitutive equations of a biomechanics study. As a result, early computers were used in the large finite element studies. Today, several different software packages are used in the biomechanical research space. Packages like OpenSim and AnyBody are used in inverse dynamics problems in the human body using scaled model anatomy (Figure 1.2-1). Other CADD programs such as SolidWorks® and Adams MSC have been used to generate patient specific models which can then be used in forward dynamic and inverse dynamic simulations.
The CADD programs that use patient specific bone or tissue geometries must be able to extract the unique patient bone geometries from imaging data. The source of this data is usually computerized tomography or magnetic resonance imaging scans. Segmentation software packages allow the scan to be represented in 3-D which allows the user to isolate the bone or other tissue from the rest of the scan. These segmentation packages can then save the extracted bone or tissue in a format that CADD programs can use.

This thesis used a software suite called Mimics (Materialise, Leuven BE) to extract bone geometry from computerized tomography scans. These geometries were then used in SolidWorks® 2016 (Dassault Systemes, Waltham MA) to perform forward dynamic studies of rigid bodies, also called rigid body modeling. Finite element analyses were performed using Ansys Workbench (Ansys, Canonsburg PA).
1.3 Foot and Ankle Modeling

In the past 15 years, research groups like Dr. Haut’s Orthopaedic Biomechanics Laboratories, Dr. Zhang’s research group, and Dr. Wayne’s Orthopaedic Research Laboratory have published on the use of computational models to understand the biomechanical behavior of healthy and injured lower limbs [5–9]. Some of the major areas of focus have been on ankle instability [5, 10], adult developed flat-foot deformities [6, 7, 9, 11] (Figure 1.3-1), and stresses in the sole of the foot [8]. Together these research groups have not only proven the predictive capabilities of computational models in the foot and ankle, but have also provided significant guidance on the approaches and pit falls of this area of work.

Figure 1.3-1: Bony anatomy of a previous foot and ankle model used to study adult developed flatfoot deformities [6].
1.3.1 Mid Foot Injuries

Thus far, foot and ankle models have focused on modeling high frequency injuries or deformities. In the mid foot this means most studies have focused on adult developed flatfoot deformities [7,11]. However, this has left other less common injuries without models or information on their kinematic behavior. One such injury in the mid foot is called a Lisfranc injury (Figure 1.3-2). This injury includes dislocations and fractures of the bones as well as soft tissue injuries in specific joints [12]. Because of unique anatomical structures in the foot, the injuries are usually concentrated between the medial cuneiform and second metatarsal. This set of injuries is challenging because it is often overlooked in initial evaluations and results in large degrees of pain and discomfort for the afflicted individual [13,14].

With injuries to the ligamentous structures of the Lisfranc joint, treatment is usually surgery to temporarily or permanently secure the injured joint in its anatomical orientation [15,16]. However, patient satisfaction ratings for these surgeries are low; in some cases as few as 50% of patients are satisfied with the outcome [15]. An additional challenge is that for the more severe injuries (such as dislocations) anywhere between 40% to 94% of patients develop osteoarthritis [17–20]. This has resulted in competing procedures and hardware to be proposed for the internal fixation of the Lisfranc injury. Recent literature has compared a range of surgical repairs with conflicting recommendations on which procedure provides the optimal environment to promote healing and avoid implant failures [15,19,21–24]. The combination of multiple procedures and sub-optimal outcomes in surgical corrections demonstrates the need for a controlled examination of all procedures in a single study to quantify specific biomechanical characteristics of the postoperative joint. Computational modeling of joints has been shown to provide the controlled environment for comparing both the healthy, injured and post-operative mechanics of the foot and ankle [5,10,11].
1.4 Objective

There are two challenges relating to Lisfranc injuries that this thesis is geared to address: the high rates of missed diagnoses and the conflicting recommendations on surgical procedures to repair the injuries. By identifying factors that can reduce the key indicators of a Lisfranc
injury, clinicians can have a better idea how they can take radiographs that will have the most obvious indication of an injury if one exists. Also, given the subpar patient outcomes for surgeries to repair Lisfranc injuries, this study looks to help better understand the biomechanics of the different procedures immediately after the surgery. By capturing the foot’s biomechanics after surgery, we can better identify where the different procedures are lacking in terms of returning the foot to an anatomically normal orientation and behavior.

The surgical procedures to be evaluated fall into two groups. The first group of surgeries are called open reduction and internal fixation (ORIF). At a high level all of these surgeries have open incisions made to the foot so the surgeon can reduce any excessive displacement of the bones away from their normal anatomical position [19,21,25]. The difference then comes from the hardware used to fix the bones in the correct position. The three different internal fixation setups to be tested are the transarticular screws, endobuttons, and dorsal plates with screws. After the injury has healed the hardware may be removed though sometimes it is left in place [15]. The second group of treatments is arthrodesis of the joint. In this case, the articular surfaces of the bones are debrided before the bones are placed in the correct orientation. Then screws are used to permanently fix the bones in place [19,26].

The objectives of this study are three fold. First, a computational model for Lisfranc injuries needs to be validated based on previous cadaver studies. Next, the model must incorporate muscle forces and other environmental factors to consider their impact on the injured foot’s mechanics as well as how the injury would present to clinicians. Lastly, the model needs to compare the ability of the four different surgeries to repair and secure the injured joint. This study is geared to provide surgeons and other healthcare professionals with strategies to help identify these injuries and then recommendations on the different corrective surgeries to treat Lisfranc injuries. These recommendations are based on how well the surgery returns the injured foot to a healthy orientation because this is a key indicator for successful outcomes [18].
Chapter 2: Anatomy Overview

2.1 Introduction

Human walking is a complex and vital mechanism for activities of daily living. The dynamic structure of the foot and ankle incorporates a large number of bones and soft tissues which allow it to accommodate different terrain while walking. The structures also allow the transfer of force into the ground to drive propulsion [27]. Since the foot and ankle bears the body’s weight and support millions of cycles, the structures must be resilient and reliable.

For normal function, the large number of bones, ligaments, and muscles (Figure 2.1-1) all must work together. When there is injury or deficiencies in one structure, then the overall function of the foot can be impacted. As a result, it is important to have a baseline understanding of the anatomy of the foot. This allows for better understanding of the mechanisms of injury and the treatment goals for Lisfranc injuries. Understanding these structures is also important to consider when representing the anatomy in a computational model.
2.2 Bony Anatomy

Bones are the structural base of the foot. They provide the framework for the rest of the tissues to attach and carry the body’s weight. The tibia and fibula are the long bones of the lower leg that form the ankle (talocrural) joint with the rest of the foot. The tibia and fibula run parallel with each other. The larger tibia has a plateau roughly in the transverse plane on its proximal end. This tibial plateau has medial and lateral depressions where the condyles of the distal femur articulate. These two depressions are separated by a ridge referred to as intercondylar eminence. The tibia and fibula do not move significantly relative to each other but they do have articulates at two points along their length: one at the proximal end and one at the distal end. The proximal articulation is formed by a concavity in the lateral tibia that holds the proximal end of the fibula. A thick capsular structure of ligaments prevents most motion this articulation. Along the length of the tibia and fibula,
the bones are connected by an interosseous membrane. The distal articulation is a notch in
the lateral tibia where the fibula is again secured with a capsule of ligaments. The inferior
surface of the distal tibia has a concavity that runs mediolateral. On the medial side of this
concavity, the tibia has a rounded portion of bone that extends inferiorly while on the lateral
side of the concavity the fibula extends inferiorly. These inferior portions of the tibia and
fibula on either side, called the malleoli, along with the inferior concavity of the tibia form
the mortise like structure of the proximal ankle (Figure 2.2-1).

The talus alone forms the distal portion of the ankle joint. The superior portion of the
talus is cylindrical running from medial to lateral and is called the talar dome. The dome
sits between the malleoli and rotates in the sagittal plane against the inferior concavity of
the tibia. The malleoli also articulate on concave portions on either side of the talar dome.
This geometry allows for significant flexion and extension of the foot when the dome rotates
within distal end of the tibia and fibula. The inferior portion of the talus widens outward
and forms the proximal portion of the subtalar joint. Anteriorly, the talar neck extends
before forming a hemispherical articulation with the navicular bone [29].

Figure 2.2-1: Anteroposterior (AP) view of the talus sitting between the malleoli.
Inferior to the talus is the largest tarsal bone, the calcaneus. This bone is dominated by the posterior tubercle where the Achilles tendon inserts. While the long axis of the bone runs anterior to posterior, there is a medial projection called the sustentaculum tali which sits under a portion of the talus. This is one of the articulations between talus and calcaneus which form the subtalar joint [30]. The anterior portion of the calcaneus narrows to a slightly concave surface that articulates with the cuboid. The inferoposterior portion of the bone has an important tubercle that is an anchor for the plantar fascia, an important soft tissue structure that acts as a bow string for the foot.

The anterior articulations of the talus and calcaneus with the navicular and cuboid bones, respectively, form a two joint complex named the transverse tarsal joint (Figure 2.2-2). The spheroid portion of the talus sits in a concavity in the posterior side of the navicular [29,31]. The transverse tarsal joint plays an important role in normal gait. Depending on the amount of inversion or eversion of the foot, the axes of the two unique joints may either be aligned parallel or misaligned. When aligned, the transverse tarsal joint as a whole is moveable which is important when the foot is adjusting to uneven surfaces. However, a locking mechanism occurs when the axes are misaligned causing a rigid connection across the transverse tarsal joint [32].

The bones anterior to the navicular are the three cuneiforms. These three bones along with the cuboid form the proximal portion of the tarsometatarsal (Lisfranc) joint. These four bones are fairly cube shaped in nature with significant portions of their surface being flat and covered with articular cartilage. By comparison, the distal end of the Lisfranc joint is formed by five long bones that are called the metatarsals. The two lateral metatarsals (numbered four and five) articulate with the cuboid while the other three each articulate with one of the cuneiforms. All five bones have a proximal base that articulates in the Lisfranc joint followed by a long hollow diaphysis that ends at the distal head where the phalanges articulate. The Lisfranc joint runs medial-laterally and is almost all in a single coronal plane except for a
disruption in the cuneiforms in the medial portion of the joint [15,33]. When looking at the three cuneiforms from a dorsal view, the anterior surface of the intermediary cuneiform is recessed compared to the other two. Since the first three metatarsals each articulate on one of the cuneiforms, this means the second metatarsal sits in a mortise formed by the three cuneiforms. This indentation disrupts the transverse soft tissue structures that run across the joint.

Figure 2.2-2: Superior view of the foot highlighting the transverse tarsal joint (outlined in red) and the tarsometatarsal joint (outlined in blue).

The five metatarsals and the more anterior phalanges make up the forefoot. The distal end of the metatarsals is a spherical head. The inferior surfaces of the metatarsal head contact the ground and the anterior portion is where the first phalanges articulate. Metatarsals two through five have a series of three phalanges while the first metatarsal has two phalanges in series.
2.2.1 Mid Foot Arches

The bones in the foot form three arches. Two run anterior to posterior (front to back) and are called the medial and lateral longitudinal arches. The third arch is the transverse arch and is formed by the three cuneiforms and the cuboid bone (Figure 2.2-3). The transverse arch can be compared to a Roman arch where the intermediary cuneiform acts as a keystone preventing the arch from collapsing. There is some deformation and flattening of the arch during normal gait cycles [34]; however, Lisfranc injuries can cause additional collapse in this arch. Some lateral radiographs can identify this collapse and are used to diagnose a Lisfranc injury in limited cases [16].

![Figure 2.2-3: Anteroposterior (AP) view of the transverse arch, the arch Lisfranc injuries impact. Metatarsals are not shown for clarity.](image-url)
2.3 Soft Tissue Anatomy

Bones form the structural architecture of the foot and ankle and can constrain joint motion based on their geometry. However, the soft tissues of the foot and ankle provide a large amount of the constraints in joint motion (ligaments), provide smooth surfaces for articulation of joints (cartilage), and generate the forces to move the foot and ankle through different motions (muscles). These soft tissues along with the bone, define the function of the foot.

2.3.1 Ligaments

The sheer number of bones in the foot and the need to maintain stable positioning of the bones relative to each other means that there are a huge number of ligaments which may span multiple joints. Ligaments in the foot and ankle may be distinct bands of tough tissue or may be structures associated with the joint capsules. Regardless of the structure, the function remains the same: preserve anatomic orientation of the bones and constrain certain motions in the foot.

Starting with the proximal tibiofibular articulation, the fibula is constrained by a capsular structure that connects the anterior, superior, and posterior portions of the proximal fibula to the lateral tibia. This stiff structure resists most motion in this joint. The interosseous membrane that connects the diaphyses of the tibia and fibula is a long membrane (Figure 2.3-1) that keeps the two bones in parallel but allows some separation between the bones, like when the malleoli are pushed apart by the talus during ankle extension. At the distal end of the two bones, anterior and posterior tibiofibular ligaments hold the fibula in the tibial notch on the lateral side of the bone. This capsular structure of ligaments also resists most motion like its proximal equivalent; however, there is more range of motion than the in proximal joint.
Figure 2.3-1: Tibia and fibula with ligament structures between the two [28].

The ankle is the joint with the widest range of motion and the joint capsule contains many areas of thickened ligamentous structures. These collateral ligaments are broken into the medial (MCL) and lateral collateral ligament (LCL) (Figure 2.3-2). The MCLs that have origins in the medial malleolus of the tibia are collectively called the deltoid ligaments. These include the anterior and posterior tibiotalar ligaments (TiTa-a, TiTa-p), tibionaviculacal ligament (TiNa), tibiocalcaneal ligament (TiCa), and tibiospring ligament (TiSp). The other MCLs that are not a deltoid ligament include the medial and posterior talocalcaneal ligaments (TaCa-m, TaCa-p). Together, these ligaments guide flexion and extension in the ankle while also resisting eversion of the ankle [35].

The LCLs provides the same constraints on flexion and extension of the ankle but they also resist inversion of the ankle. The LCLs include the anterior and posterior talofibular
Inferior to the talus are additional ligament structures that impact the range of motion for the subtalar joint and the transverse tarsal joint. On the medial side, there is the spring ligament complex. The spring ligament complex has been characterized as three different bands of tissue running along the plantar surface of the foot from the medial and anterior surface of the calcaneus to most of the inferior surface of the navicular. The three bands are the superomedial calcaneonavicular (SMCN), middle calcaneonavicular (MCN), and inferomedial calcaneonavicular (IMCN). The SMCN has the most medial insertions on the calcaneus and the navicular of the three while the IMCN has the most lateral insertions. The SMCN and the MCN are inferior to the talar head and as a result, receive near constant compression of the ligament perpendicular to their long axis. This loading can lead to some unique changes to the ligaments’ body which may incorporate cartilage or mineralized collagen structures [37].

Continuing anteriorly to the mid foot and forefoot, a dense network of short ligaments constrains most motion of the bones (Figure 2.3-3). These ligaments are mostly thickening of the joint capsules and have plantar and dorsal components that run perpendicular to the
different joint spaces. Starting with the transverse tarsal joint, there are the talonavicular (TaNa), calcaneonavicular (CaNa), and calcaneocuboid (CaCu) ligaments. In the rest of the mid foot are the naviculocuneiform (NaCn), naviculocuboid (NaCu), cuboicuneiform (CuCn), and intercuneiform (INTCn) ligaments.

![Artistic view of dorsal ligament structures from a superior view onto the talus and foot](image)

**Figure 2.3-3:** Artistic view of dorsal ligament structures from a superior view onto the talus and foot [28].

### 2.3.2 Connections Around the Tarsometatarsal Joint

The last set of short interosseous ligaments relevant to this work are across the tarsometatarsal joint in the forefoot (Figure 2.3-3). The cuneometatarsal (CnMt) and cuboimetatarsal (CuMt) ligaments connect the mid foot with the forefoot. There are also proximal and distal transverse intermetatarsal (INTMt) ligaments spanning from one metatarsal to another. However, the recession of the second metatarsal base between the three cuneiforms does impact the proximal intermetatarsal connections. Since the base of the second metatarsal base is posterior to the first metatarsal base, there is no ligament connection between the two bones [38–40]. Instead the first metatarsal only has ligament connections to the medial
cuneiform at its proximal end. As a result, the proximal mediolateral connection for the second metatarsal is with the medial cuneiform. This ligament, which has dorsal interosseous, and plantar components [40], technically spans the tarsometatarsal joint but because of its common involvement with Lisfranc injuries, it is more often called the Lisfranc ligament.

The missing ligament between the proximal first and second metatarsal is thought to be a contributing factor to the prevalence of Lisfranc injuries to occur between the first and second rays. Arguments have been made that the amount of stress placed on the Lisfranc ligament leads to dislocations and ligament injuries that are seen in this location [15]. There are also suggestions that the shallower the mortise that the second metatarsal base sits in the higher the future risk of a Lisfranc injury [41]. The mortise is highlighted in Figure 2.3-4.

![Figure 2.3-4: Superior view of the tarsometatarsal (Lisfranc) joint and the unique anatomy with mortise formed by the cuneiforms highlighted in red.](image)

### 2.3.3 Plantar Fascia

The last ligamentous type structure of relevance in this work is the plantar fascia. As the name suggests, this structure covers the full sole of the foot. The proximal origin is at the
inferior calcaneal tuberosity. The band of tissue is fairly superficial and extends anteriorly before branching into five bands and inserting into the proximal phalanges of each toe. When the toes are extended, the plantar fascia is tensioned by wrapping around the metatarsal head in what is called the windlass mechanism [42]. Other muscles also contribute to this mechanism but the tension along the plantar surface of the foot pull the metatarsal heads and calcaneus together and raise the medial longitudinal arch of the foot.

2.3.4 Articular Cartilage

The soft tissue that covers the areas of a bone’s surface that contacts another bone is articular cartilage. There are two main functions of this tissue in the biomechanics of the foot, as well as the rest of the body. The first is to provide a low friction, lubricated surface to enable smooth joint movement and reduce wear that the bones experience. The second functional purpose is to distribute contact forces across larger areas of the subchondral bone. As cartilage deforms, the fluid in the tissue is pressurized allowing forces to be distributed [43].

In the foot, there are bones that are part of three or more joints. Thus, several bones, like the cuneiforms and talus, have a large portion of their surface covered in articular cartilage. Besides there being a large number of articulations for some bones, the individual regions themselves are large to ensure smooth translation throughout the full range of motion exhibited by the individual joints. For instance, the metatarsal heads have a large region of cartilage to allow full flexion and extension of the metatarsophalangeal joint in the sagittal plane [44]. The talar dome also has a large cartilage coverage since the tibia rotates along the superior surface and the malleoli translate on either side of the dome (2.2-1). In healthy individuals, cartilage thickness in the foot ranges from 0.6 mm to 1.2 mm depending on location [45, 46]. Cartilage on the talus and distal tibia show higher thickness while values decrease in the mid foot between the cuneiforms and navicular [46].
2.3.5 Muscles

Flexion, extension, inversion, eversion, and internal and external rotation are the different motions the foot can perform with the forces generated by muscles. There are dozens of muscles that help accomplish these motions and many of them act as antagonists to each other. For this study, the muscles of the most importance are the flexor muscles: triceps surae, posterior tibialis, flexor hallucis longus, flexor digitorum longus, peroneus longus, and peroneus brevis. These muscles provide the propulsion during walking and resist ankle rotation while in a heel raise stance.

The largest of these muscles is the triceps surae which includes three muscles: the lateral and medial gastrocnemius and the soleus. The three muscles originate in the posterior calf and converge into the Achilles tendon, a dense tough tendon that inserts into the posterior portion of the calcaneus. The triceps surae is a major contributor to ankle flexion during gait.

The posterior tibialis (PT) muscle also originates in the posterior calf and its tendon wraps around the medial malleolus before continuing to the anterior portion of the mid foot (Figure 2.3-5). The tendon then branches into a network of bands that insert into the navicular, medial cuneiform, and some of the metatarsal bases [47]. Since the PT tendon wraps under the ankle joint before inserting into the medial arch, this muscle supports flexion as well as inversion of the foot.
Other muscles in the posterior calf have their tendon wrap around the talus/sustentaculum tali or malleoli before inserting into anterior portions of the plantar foot. The flexor hallucis longus (FHL) and the flexor digitorum longus (FDL) both continue this motif. The FHL and FDL wrap around the posterior portion of the medial malleolus with the FDL deeper than the PT and the FHL deeper than the other two. The FHL runs along the plantar surface of the foot before inserting into the distal phalanges of the big toe. In comparison, the FDL travels anteriorly until it branches slightly before the forefoot into 4 different bundles. Each bundle inserts into the distal phalanges of toes two through five. Both the FHL and FDL contribute to the tension generated between the heel and forefoot like the plantar fascia. Together this heightens the medial arch of the foot [49,50].

The peroneus longus (PL) and brevis (PB), also known as the fibularis longus and brevis, contribute to flexion but are located on the lateral side of the foot and leg. The muscles are in the lateral portion of the calf and both tendons travel down, around the lateral malleolus,
and along the lateral edge of the calcaneus (Figure 2.3-6). From there, the paths of the two
tendons diverge. The PB continues anteriorly and inserts into the base of the fifth metatarsal.
By comparison, the PL goes medial-anterior travelling on the plantar surface of the cuboid
before inserting into the lateral aspect of the medial cuneiform and first metatarsal base [47].
Together, these muscles induce plantar flexion at the ankle and eversion of the foot.

![Figure 2.3-6: Lateral view of the PL and PB course and inferior view of PL course.](image)

### 2.4 Lisfranc Injuries

As discussed previously, the unique anatomy of the tarsometatarsal joint between the
first and second ray compared to the second through fifth ray has been hypothesized as a
contributing factor to the susceptibility of this area to injury (Figure 2.4-1). There are a wide
range of mechanisms that give rise to Lisfranc injuries. Many are high energy events such as
falling from a height or contact between the first metatarsal head and the brake pedal during
a car accident. Other times, these could be sports injuries such as abduction/adduction of
the foot while it is plantarflexed. While not a common risk today, people previously would get this type of injury when their foot would get stuck in their stirrup when dismounting a horse. Lastly, diabetics are at risk of developing this injury. When Charcot foot develops in a diabetic, the degradation of the somatic nervous system in the foot can result in micro injuries that the individual does not perceive. Lisfranc injuries are the most common mid foot injury for those with Charcot foot [51,52].

![Dorsoplantar (DP) radiograph of a left foot with a Lisfranc injury](image)

**Figure 2.4-1:** Dorsoplantar (DP) radiograph of a left foot with a Lisfranc injury [53]. Arrow points to the increased diastasis between the medial cuneiform (CN1) and the second metatarsal (MT2).

The different injuries of the Lisfranc joint include fractures, dislocations, and ligamentous injuries. Myerson et al. has classified the injuries into three groups. First is total incongruity where all 5 metatarsals move away from the mid foot as one unit. Next is partial incongruity where a subset of metatarsals are dislocated but other metatarsals remain reduced in their
anatomical location. The last type is divergent, which is when the dislocation of the first metatarsal is medial and the second metatarsal (and potentially additional metatarsals) dislocates laterally [17].

Diagnosis of Lisfranc injuries are often challenging. Major fractures and dislocations are usually obvious with radiographs where the most significant finding is diastasis (separation) between the base of the first and second metatarsals or between the medial and intermediary cuneiform [13]. The challenge comes from injuries without gross deformation or bone fractures. These are more common in low energy injuries like those seen from sport injuries. In these cases, the soft tissues and ligaments of the mid foot are injured and result in joint instability and pain. However, 20-40% Lisfranc injuries are missed in the initial evaluation because diastasis in the joint is not easily discerned [13, 14]. While radiographs are the standard diagnostic tool, up to 50% of those with Lisfranc injuries will not show an opening in the joint unless the radiographs are taken with weight bearing [54]. Even weight bearing radiographs have been shown to miss up to 15% of cases with Lisfranc injuries [55].

The treatment options for Lisfranc injuries are either conservative or surgical. Conservative treatments are only recommended for stable, non-displaced injuries. Even subtle Lisfranc joint injuries, characterized as 1-5 mm of diastasis between the medial cuneiform and second metatarsal, are recommended for surgical intervention [13, 15, 16]. Lisfranc injuries are not overly common, representing only 0.2% of fractures and have a prevalence of 1 in 55,000 per year [15]. The high rate of surgical repair for Lisfranc injuries coupled with the poor patient ratings (Figure 2.4-2) for the different procedures make this a unique and relevant challenge in orthopaedics.
Table 2: Clinical studies for operative treatment of Lisfranc injuries

<table>
<thead>
<tr>
<th>Study</th>
<th>Year</th>
<th>Study Level</th>
<th>Number of patients</th>
<th>Treatment</th>
<th>Outcome</th>
</tr>
</thead>
<tbody>
<tr>
<td>Myerson et al.</td>
<td>1986</td>
<td>IV</td>
<td>72 patients, 76 feet</td>
<td>ORIF</td>
<td>49% good to excellent, 51% fair to poor</td>
</tr>
<tr>
<td>Amiz et al.</td>
<td>1988</td>
<td>IV</td>
<td>41</td>
<td>ORIF</td>
<td>34/41 good to excellent, 6/41 fair to poor</td>
</tr>
<tr>
<td>Kuo et al.</td>
<td>2000</td>
<td>IV</td>
<td>48</td>
<td>ORIF</td>
<td>AOFAS score 77 points</td>
</tr>
<tr>
<td>Richter et al.</td>
<td>2001</td>
<td>IV</td>
<td>49</td>
<td>ORIF, closed,</td>
<td>AOFAS score 71 points</td>
</tr>
<tr>
<td>Calder et al.</td>
<td>2004</td>
<td>IV</td>
<td>46</td>
<td>ORIF</td>
<td>Poor prognosis associated with delay in treatment or compensation claim</td>
</tr>
</tbody>
</table>

*Richter et al. study did not distinguish results between Chopart versus Chopart-Lisfranc injuries in the AOFAS score.*

**Figure 2.4-2:** Table of past clinical studies showing the patient outcomes of the surgery [15]. American Orthopedic Foot and Ankle Scores (AOFAS) of 80+ are considered “good” results [56].
Chapter 3: Development and Validation of a Computational Foot and Ankle Model for Application to Lisfranc Injuries

3.1 Introduction

The methods and procedures presented here were used to develop a new foot and ankle computational model that was then validated against a cadaver study of Lisfranc injuries. This rigid body kinematic model was used in further studies of the Lisfranc joint detailed in later chapters. The development of the model (Figure 3.1-1) was based heavily on previous computational foot models by Iaquinto and Spratley [6, 7]. Model development began with computerized tomography (CT) scans of the lower limb and foot which allowed extraction of solid body parts representing the bones in the foot and ankle. Because of the dynamic nature of the foot while walking, all the bones of the foot along with the tibia and fibula were included in the model. Additional parts were added to the model to help simulate loading the lower leg and foot, wrapping muscles around bony anatomy, and ligament connection points. Tension constraints were added between the solid bone bodies to simulate ligament constraints. This basic model was then loaded with body weight forces and measurements of the model were made. These measurements were compared with those collected for cadaveric feet with healthy and injured Lisfranc ligaments in a study by Bansal et al. [57].
3.2 Cadaveric Validation Study Design

The computational model was oriented and set up to allow comparison and validation with a previous experiment looking at ligamentous Lisfranc injuries in lower leg cadavers [57]. The experiment examined 10 left and right paired cadaveric lower legs and used triad optical tracking probes to measure the diastasis of the Lisfranc joint for both healthy and injured feet. Measurements were made for the dorsal and plantar portions of the Lisfranc joint space and between the first two metatarsal’s midpoints along the dorsal portion of the diaphyses (Figure 3.2-1). The tibia, fibula, talus, and calcaneus were secured in 30° plantarflexion with screws.
In the healthy experiments, the lower leg was loaded with 20 N to simulate non-weight bearing and then 343 N to simulate 50% partial weight bearing. Loads were held for a minute to allow ligament creep to reach steady states. The optical tracking data of the bone triads was recorded. The researchers then took the cadaver foot and transected the dorsal, interosseous, and plantar Lisfranc ligaments as well as the intercuneiform ligaments of the medial and intermediary cuneiforms, similar to other experiments [21]. Again, motion tracking data of the bones were captured during 20 N and 343 N loading.

The study performed further tests on open reduction and internal fixation with screws and dorsal plates. For the purposes of this validation work, only the healthy and injured test results for the cadavers were used (Figure 3.2-2).
3.3 Computational Model Geometries

The bones of the foot and lower leg were the geometries used in the CADD models in this work. These bony geometries had to be created from an anatomical source and process before they could be used in any computational simulations.

3.3.1 Bone Isolation and Calcaneus Repair

The rigid bodies of the kinematic model were the 3D bones based on a patient’s anatomy. In this study the cadaveric leg of a 61 year old female received a CT scan. The individual had previously received a medializing calcaneal osteotomy (MCO) but had not previously listed any Lisfranc injuries [6]. The MCO was reversed in later processing. The scan was loaded into the commercial medical image software MIMICS® (v16.0, Materialise’s Interactive Medical Imaging Control System, Materialise, Ann Arbor MI). DICOM (Digital Imaging and Communications in Medicine) data stacks of the scan were used in MIMICS®. These data stacks were represented as three sets of 2D images. Each of the three sets of the images

Figure 3.2-2: Results of healthy and injured cadaver diastasis measurements in Bansal et al. [57].
was a plane of view representing the CT data as slice-by-slice images throughout the depth of the scan region.

Since the grayscale value of each pixel in the DICOM is related to the radiopacity of the tissue based on Hounsfield units (HU), MIMICS® could create masks of all pixels above a certain threshold pixel intensity that defined bone (Figure 3.3-1). This threshold value was roughly 600 HU. Simple thresholding was able to identify all the bones as a single mask. To separate mask into individual bones, semi-automated tools were used to identify lines of demarcation between bones in several slices of a scan and then interpolating the demarcations throughout the other slices of the scan. Once individual masks were identified, MIMICS® provided a tool to fill all pixels interior to the mask. This was important because of the low intensity portions of the diaphysis of long bones that do not meet the HU thresholding value. For the purpose of the solid bodies in the kinematic model, cortical and cancellous bone were not differentiated. Manual review and edits of the meshes compared to the scan in MIMICS® were used to correct any areas of gross difference or error.

![Figure 3.3-1: Images of the raw CT scan and the mask of the talus highlighted in the CT scan.](image)

A shortcoming of isolating bones based on thresholding is that soft tissue in the joint
space such as cartilage is not captured. As a result, the isolated bones appear to have larger
distances between their articular surfaces than they would in vivo. As a correction, the
masks for the isolated bones were scaled. This preserved the distance between the centroids
of the bones, joint geometry, and allowed for cartilage volume to be represented as bone
volume. MIMICS® allowed this scaling function to expand the mask by one pixel in each
direct. This lined up well with the two pixel gaps (~1 mm) seen in the joints space when
masks were not scaled. This last assumption of treating cartilage and bone the same is not
biomechanically accurate but since the simulation is rigid body modeling (RBM), the more
compliant cartilage would not deform in this simulation even if it was modeled separate from
the bone. However, the low friction function of cartilage was reproduced during articulation
between adjacent bones.

With the mask representation of the individual bones isolated from each other, MIMICS®
provided a tool to make a shell of triangles to represent the surface of the mask. This process
of generating triangulated representations of the masks is called meshing. MIMICS® allows
these meshes to be saved as their own stereolithographic (STL) file. Depending on the
resolution used in the meshing process, some of the anatomy of the mask may be simplified.
One intentional simplification was to treat all of the phalangeal bones for a given toe as
one body. While the meshes of the phalanges were not ultimately used in the model, they
remained as suppressed bodies that could be added into the simulation. All of the meshes
that were generated maintained the coordinate system and origin of the original CT scan.

The meshing process could end up generating millions of triangulated surfaces to represent
a given bone. This number of data points posed a challenge for other software used subsequently.
To handle this, a Materialise software called 3-matic was used to refine the mesh representations
of the bones (Figure 3.3-2). The STL files from MIMICS® were each loaded into 3-matic.
Automated tools were used to remesh the bones to a manageable number of triangular
surfaces (less than 20,000 for tibia and fibula and less than 4,000 for other bones). This
involved balancing reduction of the number of triangles with preserving the anatomic geometry of the bone. Once the file size had been reduced to the number of triangular surfaces listed above, each bone was saved as a unique STL file.

![Triangle reduction and smoothing](image)

**Figure 3.3-2:** 3D mesh of the talus before and after the smoothing process.

SolidWorks® was the software chosen for RBM of the foot’s biomechanics because of its multifaceted capabilities to handle engineering design and kinematic testing in the same interface. Add-ons in SolidWorks allow for models to be used in rigid body motion simulations, which was most important for model validation portion of the work. Each bone was imported into SolidWorks® where the STL files were converted into SolidWorks parts file types.

Since the STL files were saved with the same coordinate system from the CT scan, it was straightforward to orient all the bones automatically once they were imported into a SolidWorks® assembly. All bones were oriented into the same neutral position they were in during the CT scan by lining up all the origins of the part files. This was done with SolidWorks®’s mate command. While the phalangeal bones were imported into the model for future use, they were suppressed in the current model from contributing to function.

At this point in the model build, the bones were all in the neutral stance from the original CT scan. The foot needed to be positioned in 30° plantarflexion in order to match the cadaveric study used for model validation [57]. The axis of rotation for the ankle was determined using another function of the 3-matics tool. By selecting all of the vertices in the
mesh for the talar dome, a cylinder was fit to the selected points. The axis of the cylinder was identified and used as an approximation of the ankle’s axis of rotation like in previous studies [7]. Back in SolidWorks®, the tarsal and metatarsal bones were locked relative to each other while the tibia and fibula were locked with each other. The foot and lower leg were then both mated to a line representing the axis of rotation for the ankle. The foot as a whole was rotated into plantar flexion so that the transverse planes of the tibia and talus were 30° relative to each other (Figure 3.3-3).

![Bony model in 30° plantar flexion](image)

**Figure 3.3-3:** Bony model in 30° plantarflexion.

The last two changes to align with the cadaveric study was to constrain the tibia/fibula complex to only translate vertically along its long axis and lock several bones relative to each other. This was accomplished by aligning the anatomical axes of the coronal and sagittal plane for tibia/fibula complex (which are explained below) in parallel with the y-x and z-y axes of the assembly using SolidWorks mates. Lastly, the cadaveric study placed locking screws to keep the tibia, fibula, talus, and calcaneus in the same position of 30° plantarflexion.
This was done in the model by locking the four bones relative to each other using mates again.

Since the patient from which the bony anatomy was obtained had previously had an MCO from a previous experimental study, we wanted to reverse this procedure to model a healthy bone anatomy. SolidWorks and 3-matic were both used to help this process. First, the cut plane of the initial osteotomy was identified in SolidWorks®. The plane of the initial osteotomy was used to cut the bone in two portions in SolidWorks®. The posterior section of the cut was translated back into alignment with the anterior portion. The lofting tool in SolidWorks® was then used to connect the two portions of the bone together and form one single body (Figure 3.3-4). To clean up the results, the bone was again remeshed in 3-matic in order to smooth out some of the rough cut edges.

![Figure 3.3-4: Superoposterior view before and after models for the MCO repair.](image)

### 3.3.2 Non-Anatomical Model Parts

Once the bones of the foot were oriented correctly, the SolidWorks® parts that provided the plantar plate and allowed loading were added. To begin placing non-anatomic parts in the model, a coordinate system relative to the anatomy of the bones needed to be specified. From previous work by Wu et al. and the International Society of Biomechanics, a standard system has been defined for the lower leg [58] (Figure 3.3-5). This coordinate system was
important to ensure the placement of these non-anatomical parts would be correctly oriented relative to the bones. First, points of interest for determining a coordinate system on the proximal tibia were identified. These included the most medial and lateral points of the tibial condyle, MC and LC, respectively. The halfway point between the MC and LC was labeled as the inter-condylar point (IC). The points of interest for the distal tibia/fibula were the tips of the medial and lateral malleoli (MM and LM) and the halfway point between them (inter-malleolar point; IM).

Figure 3.3-5: Coordinate points and axes for the lower leg based on ISB publications [58].

Once the points had been identified, the axes could be created as planes in the SolidWorks® assembly. The frontal plane was defined to contain the MC, LC and IM. The sagittal plane was perpendicular to the frontal plane and contained the IC and IM points. Lastly, the
coronal plane was perpendicular to the first two planes. The origin was the IM with the positive x anteriorly, positive y superior, and positive z laterally.

In order to model weight bearing in the computational model, a plantar plate needed to be placed inferior of the model. Some of the design decisions for the plantar plate were driven by the desire to make the model extend-able to other model setups. For studying a 30° plantarflexed foot, a simple flat surface would be sufficient since only the metatarsal heads are contacting the plantar plate. However, for a neutral foot position, the different thickness of the fat pads beneath the calcaneus and the forefoot required consideration. In the case of the patient used in the model creation process, the fat pad was 4.5 mm thicker at the calcaneus than in the forefoot as measured from the CT scan. As a result, the posterior portion of the plantar plate was raised 4.5 mm to preserve the vertical positions of the hind and forefoot when a complete foot contacts the ground like in the original CT scan. As a note, the CT scan was performed with a cadaveric leg that had a neutral foot position even though it was not loaded to simulate body weight. This is a strategy that has been used for previous foot and ankle models studying flatfoot deformities [6, 7].

Also, the plantar plate in the cadaveric study had a screw bolt that was placed between the first and second toes. The pillar helped prevent excessive anterior translation of the foot when loaded. While not in the first iteration of the model’s plantar plate, a pillar was identified as the best solution to represent the experimental setup used for validation. More details are in Section 3.5. The plantar plate was then placed parallel to the transverse plane that was previously defined using the ISB coordinate system of the tibia/fibula complex. Special care was taken to align the surface in the transverse plane so that the pillar was slightly anterior to the heads of the first and second metatarsals

In order to apply body weight forces to the model, an assembly was placed in the tibial plateau (Figure 3.3-6). A 12.7 mm diameter rod was inserted ~38 mm into the tibial plateau. The rod was perpendicular to the transverse plane and was superior to the IC and IM points.
which formed the long axis of the tibia/fibula complex. When a force was applied to the superior surface of the rod, it ensured that the loading was perpendicular to the plantar plate. The rod also had a key channel and small rod extending from its posterior surface. The posterior surface key channel was mated to be parallel to the frontal plane to ensure the small posterior rod was directed posteriorly. This rod is discussed more in the discussion on muscle modeling. Lastly, a square collar with a hole matching the loading pin’s cross section was mated to the pin. This setup was similar to those used in both computational and cadaveric studies in the Orthopaedic Research Laboratory in the past [59,60].

![Image](image.png)

**Figure 3.3-6:** Proximal assembly to enable simulation of weight bearing.

### 3.4 SolidWorks Motion Study: Rigid Body Modeling

The SolidWorks® add-on that enables the forward dynamics RBM used in this study is called SolidWorks Motion. This integrated add-on allows various forces and mechanical constraints to be applied to the assembly previously discussed. SolidWorks Motion utilizes the geometries of the SolidWorks® assembly, the equations of motion, and the user defined
forces and constraints as inputs for a built-in rigid body motion solver produced by Automated Dynamic Analysis of Mechanical Systems (ADAMS) (MSC Software Corp., Santa Ana, CA).

The ADAMS solver uses what is called a “Gear Stiff” (GSTIFF) integrator to efficiently solve higher order differential equations. It is aimed to handle systems that have varying rates of change. The step size of the integrator is determined by user defined parameters for the acceptable values for error. In other words, when the solution for an equation is changing slowly, step sizes will be set larger. Conversely, when solutions are changing quickly, the step sizes will be smaller to ensure that solutions are within the error threshold amount. This is important because the end model has both high frequency forces (tension only springs) and lower frequency forces (contact forces). This means that the solver can adjust to efficiently and accurately handle a wide range of inputs. Figure 3.4-1 has the parameters used in these models.
The constraints (mates) that were carried over from the SolidWorks assembly were the mates for the loading pin, the tibia, fibula, talus, and calcaneus being locked together, the tibia constrained to vertical displacements, and the plantar plate being locked in place. In SolidWorks Motion contact forces between different components in the model as well as ligament and muscle forces were added. The following sections discuss how these additions to the model were implemented in SolidWorks Motion.

3.4.1 Contact Parameters

A common practice in computational biomechanical models is to simplify joints to a simple hinge or ball and socket joint. Computational software such as OpenSim use this
approach with great success when performing inverse dynamics. However, this greatly limits certain types of studies, particularly those whose goals are to analyze the effects of osteotomies, soft tissue deficiencies, etc. Further, studies of the ankle have shown that the axis of rotation of the joint changes throughout the range of motion [61]. Even further, the ankle has more degrees of freedom than a simple hinge [62]. In the mid-foot, many of the joints are planar. In order to capture the motion in these joints, all the bones other than the tibia, fibula, talus, and calcaneus were constrained by 3D physiological constraints of the bones and forces applied by soft tissue. Therefore, they had six degrees of freedom (DOF) and would be held in anatomical alignment with soft tissue constraints and contact forces with other bones. The contact forces in the model were generated based on interference detection between two parts (bones).

In SolidWorks Motion, contact settings require the user to select the parts that require contact detection and a set of parameters used in the contact force calculations. The calculation of contact force is based on the amount of overlap of the two parts and the stiffness of the parts that the user defined. In rigid body motion, deformation is assumed to be negligible. So the stiffness values are set very high in order to heavily penalize any overlap of parts. This ensures that the steady state solutions will have minimal overlap while still ensuring that final positions are dependent on contact with other parts.

In order to reduce computational complexities, contact sets were only defined for a part with their closest neighbors. For example, the talus should not be contacting the metatarsals (i.e. no contact set required) but it should contact the navicular. This resulted in 12 contact sets defined in the model. An example of one of these contact sets is shown in Figure 3.4-2.
After defining the geometries whose overlap governs the contact force, the remaining parameters for the force’s magnitude need to be defined by the user. All contact sets used the same contact parameters shown in Figure 3.4-3. The restoring force to separate two parts is dependent on the stiffness ($k$) and the exponential ($e$) of the overlapping distance ($g$) (Equation 3.1) where the stiffness and exponential value is user defined. These values were taken from previous studies which had optimized these values so that at equilibrium, overlap was an infinitesimal fraction of the total bone volume [6, 7].

The last set of user defined parameters for contact was used to define a damper that reduced the contact force as a function of how fast the parts are moving ($dg/dt$). In Equation 3.1 these parameters are the max damping ($d_{\text{max}}$) and max penetration ($c_{\text{max}}$). This will prevent large spikes in contact force that could “ping-pong” parts between each other. The reduction of the contact force when parts are rapidly moving allows for more stable simulations that do not have a large amount of high frequency changes.
\[ F_n = k \cdot g^e + \text{step}(g, 0, 0, d_{\text{max}}, c_{\text{max}}) \cdot \frac{dg}{dt} \] (3.1)

**Figure 3.4-3:** Contact set parameters to prevent part overlap in SolidWorks Motion.

### 3.4.2 Soft Tissue Constraints

Even though the model validation required the lower leg and hind foot to be locked relative to each other, the soft tissues across these joints were included in the model. Even though these tissues did not impact the results in this study, the goal was to create a model that could be validated for other foot or ankle studies and be ready for use after validation. For that reason, information is included on how these structures were modeled and setup.

#### 3.4.2.1 Ligaments

SolidWorks Motion does allow simple mechanical components like springs and dashpots; however, ligaments are resistant to tension but act like a rope when compressed. Spring elements could model the resistance of two bones being pulled apart, but they would also introduce resistance when the bones are pushed closer together. To solve this, SolidWorks Motion allows forces between points to be coded as a function of the distance between the two points. This allows for the equation below (Equation 3.2) to be used to only generate a tensile force when two points are stretched beyond the specified length and otherwise provide no force.
\[ F = IF(\text{Length} - L_0 : 0, 0, -S \times (\text{Length} - L_0)) \]

With this functionality, ligament behavior can be modeled if the origin and insertion of the ligaments are known. One additional challenge to consider is that the length where ligaments generate no force \(L_0\) is not the distance between the origin and insertion in a CT or magnetic resonance imaging scan. In those scans, the ligaments are already taut under an in-situ strain. To account for this, all measured ligament lengths were considered to be under 3% strain compared to the \(L_0\) length of the ligament. While this argument stands up to observed behavior of ligaments being cut \textit{in vivo} \[63\], it also helps prevent laxity and instability in the model.

As discussed previously, the foot and ankle have a complex network of unique ligament bands as well as areas of thickened synovial capsules. The methods described above only allow forces to be point to point. Therefore, most ligaments which have origins and insertions over an area were modeled as multiple elements (usually 2-3) with insertions on the outer edges of the ligament-bone interface area. In total, the model contained 176 tension only elements representing 54 ligament structures. A full list of the individual elements with their stiffness and abbreviations are listed in Appendix A. Selection of the ligament origin and insertion locations as well as stiffness values were informed by previous foot and ankle models as well as additional sources from the literature \[6, 7, 36, 49, 64–67\].

The ligament structures between the tibia and fibula are broken into three groups: proximal articular capsule (PROX\_TiFi), interosseous membrane (IOM\_TiFi), and distal articular capsule (LCL\_TiFi). While these structures did not impact model performance because of the constraints between the tibia and fibula, they were included for model completeness. The proximal articular capsule permits minimal movement, so the structure was represented by six elements that connected the edge of the proximal fibula to the lateral side of the tibia. Previous models, literature, and the fact that the articulation has very little
movement lead to a stiffness of 200N/mm being applied to each element [6,9]. The IOM_TiFi was represented as seven elements running parallel to each other between the fibula and tibia with insertions and origins along the diaphyses of the long bones. Previous reviews of the interosseous membrane in the forearm resulted in other models using a stiffness of 880 N/mm divided evenly among the 7 elements [64]. The last tibiofibular ligament structures are the three bands at the distal end of the two bones. The two anterior and one posterior bands had stiffness values from 90-120 N/mm based on work by Siegler et al. and Attarian et al. [36,67].

The LCL_TiFi is often considered part of the lateral collateral ligaments (LCL). The other structures in the LCL are the calcaneofibular ligament (LCL_CaFi), the anterior and posterior talofibular ligaments (LCL_TaFi), the lateral talocalcaneal ligament (LCL_TaCaL) and the superficial fibular retinaculum (LCL_SFR). These are shown in Figure 3.4-4. Stiffness values for these elements were also identified and subsequently validated by the same sources as those for the LCL_TiFi [36,67].

![Image of the lateral aspect of the foot and ankle model highlighting the LCL structures.](image-url)
The deltoid ligament group on the medial side of the ankle joint used two elements to represent each of the tibiocalcaneal (DEL_TiCa), tibionaviculbar (DEL_TiNa), and tibiospring (DEL_TiSp) ligaments. Other deltoid group structures included the anterior and posterior tibiotalar bands (DEL_TiTa) and the medial and posterior talocalcaneal (DEL_TaCa) ligaments (Figure 3.4-5). The stiffness of individual ligament structures ranged from 80-400 N/mm based on past studies [6, 36]. The last ligament structure in the ankle is the talocalcaneal interosseous ligament which is represented by three elements with stiffness values of 90 N/mm.

Figure 3.4-5: Image of the medial aspect of the foot and ankle model highlighting the deltoid group structures.

The next set of ligament structures connected the hind foot to the mid foot. Along the dorsal surface, there is an expansive network of bands of thickened capsular tissue that form the ligaments. A visual representation of the overall dorsal network is shown in Figure 3.4-6. The plantar surfaces also have a similar network of thickened capsular tissue that form the ligaments. However, the plantar surface also has distinct structures such as the
spring ligaments and the long plantar ligaments and plantar fascia. The spring ligament was represented by three different structures (inferomedial, middle, and superomedial calcaneonavicular ligament), each with two to four elements. The approach matched the setup used by Spratley et al. [7]. Between all the structures of the spring ligament, a total stiffness of 200 N/mm was used.

**Figure 3.4-6:** Image of the network of dorsal ligaments from a superior view of the foot and ankle model.

The long plantar ligaments and plantar fascia are inferior to the other interosseous plantar ligaments in the foot. The long plantar ligament (LPL) and the plantar fascia were represented using sets of elements from the calcaneus to the different insertion points along the metatarsals (Figure 3.4-7). In the case of the long plantar ligament, there is a central band that was modeled with 6 elements originating along the inferior surface of the calcaneus and inserted into the bases of metatarsals two through four. The lateral band of the LPL (i.e. more superficial to the skin of the plantar foot) has a similar origin but inserted into the base of the fifth metatarsal. The plantar fascia which is inferior to the LPL has a similar arrangement for its central and lateral bands with the exception that the
central bands inserted near the metatarsal heads. The challenge with these long ligaments is that they wrap to follow the longitudinal arch of the foot. To model this wrapping shape, intermediary bodies were used to direct the bands superiorly before descending inferiorly as they extend towards the anterior foot. The intermediary bodies were constrained close to the inferior surface of the longitudinal arch and were bars that allowed the multiple elements of each band to insert over a wider mediolateral range (roughly 25 mm). Both the LPL bands and the plantar fascia have been well characterized by Huang et al. and Kitaoka et al., respectively [49, 65]. The stiffness values from these works have been used in other models and were again used in this study.

The last set of structures are the deepest plantar network of ligaments and the dorsal ligament networks. These ligaments are not well characterized in terms of their stiffness since they are mostly incorporated within the joint capsule. Other possible reasons for their absence from studies characterizing mechanical properties could include the challenge of
working with these small structures and large variances between individuals. As a result, past models where stiffness values were assigned by a structure’s size were used since these models validated with experimental data [6, 7]. The stiffness values ranged from 90-270 N/mm.

Since the focus of this model is to study the ligaments of the Lisfranc joint, a specific review of the anatomy and its previous representation of this soft tissue in models was performed. Dissections of the Lisfranc joint identified three different structures throughout the depth of the joint [39, 40]. These are the dorsal, interosseous, and plantar ligaments connecting the medial cuneiform to the base of the second metatarsal (Figure 3.4-8). The interosseous ligament has been noted as the thickest of the bands [39]. The reviews of past models showed that dorsal and plantar ligament elements were used to connect the second and first metatarsal bases (i.e. proximal intermetatarsal connection). The ligamentous structures are not present in vivo. And while the previous models did have dorsal and plantar ligaments spanning the medial cuneiform and the second metatarsal base, they did not include an interosseous element to represent the third ligamentous band of the Lisfranc joint. Therefore, no proximal intermetatarsal ligaments between the first and second metatarsals were included. Also, a third Lisfranc ligament element with a higher stiffness (120 N/mm) was included in the interosseous space with origins and insertions based on Panchbhavi et al [39].
3.4.2.2 Muscles

The muscles and tendons that contribute to plantar flexion in the foot were incorporated into the model. These muscles are the flexor hallucis longus, flexor digitorum longus, tibialis posterior, peroneus longus, and peroneus brevis. The muscles that terminate in the Achilles’ tendon (soleus and gastrocnemius) were also included in the model; however, they did not play a role in this study because the lower leg and hind foot were locked relative to each other.

These additions posed a challenge because of their geometries. The muscles are located on the posterior of the lower leg, but their tendons descend before wrapping around the posterior of the ankle (talocrural) joint and continuing anteriorly along the lateral or inferior surface of the foot where they insert into the mid and forefoot. However, the model was validated based on a cadaveric study where no muscle activity was simulated. Therefore, the
details for calculating the muscle forces for a 30° plantarflexed stance and the implementation of the muscle wrapping are discussed in Section 4.2.

3.4.2.3 Cartilage

Cartilage was not a separate component in the model. Instead, the volume of the cartilage in the joint spaces were captured by the scaling of the bones discussed previously. However, it was important to capture the low friction that cartilage provides between bones. This was achieved through the contact parameters in SolidWorks Motion. When defining contact groups between bones, the friction option was turned off. This way, bone on bone contact in the model would behave the same as if there was the nearly frictionless cartilage and synovial fluid in the joint space. The shortcomings in this approach to representing cartilage was that the cartilage volume was treated as part of a non-deformable rigid body. Given the small volume of articular cartilage and the macroscopic motion that is being observed in the model, the lack of cartilage deformation was deemed an acceptable assumption.

3.5 Model Validation

The model of the foot and ankle was setup to imitate the experimental design by Bansal et al. [57]. The bones of the lower leg and hind foot (talus and calcaneus) were locked relative to each other in 30° plantarflexion. Because the cadaver experiment did not apply forces to simulate muscle contraction, no muscle forces were incorporated into the validation studies. Model runs were performed for the 20 N and 343 N loading setups to compare the Euclidean diastasis values for the three locations reported in the experimental study: Dorsal Lisfranc, plantar Lisfranc, and intermetatarsal (Figure 3.2-1). The model simulated a 12 second period of time where the body force was either kept at 20 N or ramped up from 20 N to 343 N in the first 2.5 seconds. After the model completed running, the X, Y, and Z coordinates of specific points were extracted from the model.
With the extracted data points the steady state time period of the simulation for the model was determined by when most points were no longer moving. At this point, the coordinates of the raw data were averaged and used in calculations in MATLAB to output the measurement values needed.

3.5.1 Model Variations

A common error with this model and others developed in the past has been the model “exploding.” In these cases, the simulation begins but have bones fly apart from each other and the simulation stops before completion [6, 7]. In order to ensure the model was able to complete its calculation, some of the model settings were altered until the models ran to completion with the values shown in Figure 3.4-1.

With models that were running to completion, the largest validation challenge was to ensure the cadaveric setup was adequately reproduced in the computational model. Different variants of the plantar plate and distal intermetatarsal ligament setups were tested before a final model validated with the experimental data.

The first plantar plate did not have a pillar to prevent the foot from translating forward. The problem from a qualitative sense was that the foot was translating forward when it was loaded since there was no frictional force between the bones and the plantar plate. By comparison, the videos of experimental tests showed the foot relatively stationary in the anterior posterior direction because of the screw bolt between the first two toes.

The second attempt to constrain the anterior-posterior motion of the model was to place a wall along the front of the plantar plate to prevent forward translation. While this did stop the forward translation in the model, the metatarsal heads would slide medially after contacting the wall.

The next attempt was to use a pillar to represent the screw bolt from the experimental study. Since the soft tissue between the toes was not physically represented in the model, the distal ligaments of the first two metatarsals were altered. Instead of the distal intermetatarsal
ligaments running from one metatarsal to the other, they each connected to the pillar of the plantar plate (Figure 3.5-1). The stiffness values were adjusted so that the same amount of force would be generated by the ligament connections as they would have been as one element.

![Figure 3.5-1: Connections between the metatarsals and the pillar of the plantar plate.](image)

Ultimately, this model setup resulted in limited movement of the foot along the plantar plate that was observed in the experimental tests. The intermetatarsal diastasis values in the injured model was the only measurement that was outside the standard deviation of the experimental data. This was because the diaphyses of the two metatarsals were not separating enough.

The next variation in the connection between the metatarsal heads and the plantar plate came after a review of the distal intermetatarsal structures. The distal transverse intermetatarsal ligament is described as having a dorsal and thicker plantar band that runs from the medial first metatarsal to the lateral fifth metatarsal [68]. Previous models had only used a dorsal element to model this structure. In order to better represent the distal intermetatarsal ligament, several changes were made. First, the model was reverted
to the setup where ligaments originated and inserted into the metatarsal heads (i.e. no direct connection to the pillar). Second, the dorsal stiffness value was changed from 90 N/mm to 30 N/mm. Lastly, a plantar ligament element with a stiffness value of 90 N/mm was incorporated into the model. This increased the stiffness of the distal intermetatarsal ligament by 30 N/mm overall.

After these changes to the model, simulations still showed the minimal anterior translation qualitatively observed in the experimental work. More importantly, the model showed validation in all three measurements for both the healthy and the injured models. The heads of the first and second metatarsals contacted the sides of the plantar plate’s pillar from the posterior direction but the intermetatarsal ligament prevented the heads from separating enough to allow additional anterior translation. These results are discussed further in 3.6.

After a validated model was generated, a variation in the plantar ligaments was evaluated to see if the model would still be in the experimental result’s range of values. This change was to leave the plantar ligaments in the Lisfranc joint and intercuneiform joint intact but with a reduced stiffness. The motivation was to model the fact that the experimental transections of the joint were done from a dorsal aspect which left the cutaneous tissue and other non-ligamentous structures intact in the plantar portion of the foot.

The model was run with 50%, 25%, and 10% of the plantar stiffness still intact. For the plantar Lisfranc ligament and intercuneiform ligament, the element stiffness values were 45, 22.5, and 9 N/mm. In the case of the 25% residual stiffness in the plantar ligaments, it is interesting to note that the dorsal diastasis of the Lisfranc joint is larger than the plantar diastasis; however, the intermetatarsal diastasis is below the standard deviation of values observed experimentally (Figure 3.5-2). So while models with residual plantar stiffness allowed certain motifs to be replicated, they were not able to be validated with the cadaver model.
3.6 Final Model Results

The final model emulated the experimental setup by Bansal et al. in a realistic manner [57]. The results in Figure 3.6-1 show that both the healthy and injured model results were within the standard deviation of the experimental tests. The healthy model showed very close agreement between the average cadaver study and the model results. They also showed a similar order in the size of diastasis going from smallest in the dorsal portion of the Lisfranc joint to the largest in the diastasis of the metatarsal diaphyses.
In the injured model, the diastasis values were in the standard deviation range of the experiment, but the metatarsal diastasis was just on the lower edge. Also, the model showed larger plantar diastasis than dorsal diastasis in the Lisfranc joint. The average experimental result had the opposite relationship between the two points on the Lisfranc joint.

### 3.7 Discussion

The validated model of the lower leg and foot was able to accurately represent (within one standard deviation) the biomechanical behavior of the Lisfranc joint and mid-foot for both healthy and injured cadaver foot. The bony anatomy was captured and accurately represented in the model using a work stream that takes CT data and extracts out the geometry of specific bones. Ligament connections of the foot and ankle were included as tension only force elements that used intermediary bodies to help represent the wrapping of some structures in the inferior portion of the foot.
While the injured model does have diastasis values within the experimental range, there was one pattern that was not seen in the simulation results. This was the fact that the average dorsal diastasis was larger than the plantar diastasis in the cadaver study. This was contrary to the original hypotheses as the plantar structures are notably stiffer than the dorsal structures. Exploration with the computational model set up to represent a dorsal dissection showed that the model would have larger dorsal diastasis than plantar diastasis if there was residual stiffness in the plantar Lisfranc (i.e. between the joint marked in red in Figure 1.3-2) because of the undisturbed soft tissue and skin. This is shown by the results in Figure 3.5-2. However, this is only a directional hypothesis that the plantar soft tissue that was not disturbed by the ligament transection could be the reason for the experimental results. The simulation for these cases of residual plantar connections did not validate on all three experimental diastasis values. All further simulations were run with complete ligament transection for the plantar Lisfranc and intercuneiform ligaments to represent a complete disruption of the structures.

Since validation was performed on cadaveric experimental data, muscle activity and contraction were ignored. This shortcoming in representing in vivo conditions is addressed by additional simulations described in the next chapter as incorporating muscle contractions.

Ultimately, this work has shown that a computational model using the approach laid out here can validate well with the mechanical behaviors of the healthy and injured Lisfranc joint of the foot. The validation provides confidence that the extension of this model to different surgical procedures to repair the Lisfranc injury will provide meaning and relevant results. This will provide the ground work for evaluating the different surgical treatments for Lisfranc injures and provide a frame of reference for the proper biomechanics of a healthy foot.
Chapter 4: Impact of Muscle Contraction and Foot Orientation on Injury Presentation

4.1 Introduction

Lisfranc injuries are challenging to identify in the clinical setting. To explore possible reasons for this fact, several changes to the computational model were implemented to consider how two important contributions - active muscle contractions and loading orientation - could impact the way that the injury presents in weight bearing radiographs. By including muscle forces, the computational model was able to overcome a shortcoming of many cadaver studies. Additionally, looking at different loading and foot orientations a patient might use can help identify if the clinicians should pay special attention to how patients position their foot in weight bearing radiographs.

Cadaver studies are common when trying to capture data that would otherwise be overly invasive in a living subject. If the cadaver is properly prepared and handled, then it is reasonable to relate a cadaveric study’s results to \textit{in vivo} behavior [69,70]. For understanding the biomechanics of the body, cadavers are extremely useful in identifying the impact of a myriad of conditions in the body. The challenge arises from incorporating the active forces of muscle contraction. Unless the experiment has a clamping mechanism (usually connected to an actuator) to attach to the tendon for the purpose of applying a tensile load, it is not possible to study the impact of muscles contraction on a cadaveric study. In the study used for validation data of the computational model, active muscle contributions were not simulated.

With the validated foot and ankle model for the Lisfranc injuries, forces were added in SolidWorks Motion to represents active muscle contraction while standing on two feet
with $30^\circ$ plantarflexion. The muscle and tendon paths were defined by the muscle origin and insertion along with surfaces and beads to help enable the wrapping of forces around different bony anatomy. Muscle force values were based on a combination of studies in the literature looking at maximal force generation in different ankle orientations and electromyography data on specific muscle activity. With the addition of muscle forces, the computational model could identify where cadaveric data may be overstating or understating in vivo behavior.

Next the computational model was used to understand how the injury would present in the clinical setting. Dorsoplantar (DP) radiographs of the foot are used to identify Lisfranc injuries. The radiographs can be non-weight bearing or weight bearing. In non-weight bearing the patient is on their back and the sole of the foot is flat on the ground resulting in the ankle being flexed. However, weight bearing is taken while the patient is standing with a neutral ankle position. Weight bearing radiographs have shown indications of Lisfranc injuries that are not visible in non-weight bearing cases [54].

Therefore, the model was set up to simulate a patient taking a weight bearing radiograph where the foot is flat on the ground. To accomplish this the computational model was placed in a neutral foot position (i.e. no flexion/extension) and loaded with a compressive load to the tibia to simulate weight bearing, both with and without muscle forces. Different foot orientations of inversion and eversion were included to simulate how a patient may try to offload portions of the foot due to pain. These studies helped elucidate the relationship between an injury’s presentation (i.e. magnitude of the Lisfranc joint separation) and additional factors during a weight bearing radiograph.

4.2 Incorporating Flexor Muscles

Only the flexor muscles of the foot and ankle were incorporated into the computational model. This approach has been used in previous foot and ankle models where understanding the steady state positions of the foot was the focus [6,7]. The muscles in the model were the
triceps surae, flexor hallucis longus (FHL), flexor digitorum longus (FDL), posterior tibialis (PT), peroneus longus (PL), and peroneus brevis (PB). Each force was identified as a percent of the subject’s total body weight before being converted to a force in Newtons (N).

4.2.1 Muscle Force Magnitudes

Previous computational models selected muscle forces for static flatfoot, neutral stances based on muscle parameters in previous publications [6,7]. This involved the use of previous literature on pennation angle, fiber lengths and electromyography (EMG) to identify force values [7, 71–74]. Using similar sources, muscle forces were determined for the validated model which is in 30° plantarflexion.

First, the 100% body weight muscle forces approximated by Blackman et al.’s cadaver study for the triceps surae (Achilles tendon in the study), FHL, FDL, PT, PL, and PB were used as the starting point for neutral stance muscle forces [71]. The forces were then converted to be expressed as a percent of the individual’s body weight (BW). Then results from Noh et al. were used to scale these neutral stance muscle forces based on the muscle activation in different points in the gait cycle. In this study, the team reported the EMG’s value as a 0 to 1 value of the maximal signal during a maximal isometric contraction throughout a full gait cycle while barefoot [75]. This allowed for an EMG value to be calculated for mid stance (when the foot is largely neutral) and for heel off of the gait cycle (when the foot is plantarflexed). The heel off phase was chosen to represent the static plantarflexed position of the model while the mid stance value represented the neutral foot position value of the Blackman et al. results. This allowed the force as a percent of body weight from Blackman et al. to be divided by the mid stance EMG value from Noh et al. times the heel off EMG value from the same study (Equation 4.1).
\[
\frac{\text{Blackman et al. Max Neutral Force(\% of BW)}}{\text{Mid Stance EMG normalized value}} \times \text{Heel Off EMG normalized value} = \text{Plantar flexion Force(\% of BW)}
\] (4.1)

The results were muscle forces representing the heel off phase in terms of percent of body weight (Figure 4.2-1). Since the weight of the individual from the CT scan used in the model creation is known, the muscle forces were calculated with the values in Figure 4.2-1. A shortcoming of this approach is that Noh et al. is an EMG study of the gait cycle and not for a static mid stance vs static heel rise. This shortcoming was deemed an acceptable assumption given the lack of data and because the ankle would be fixed to prevent the changes in flexion that occur during heel off in gait. Another shortcoming in this approach was that force-length and force-velocity relationships were not considered. This data was also not available to implement. Thus, an approximation of the muscle force magnitudes was calculated.

<table>
<thead>
<tr>
<th>Tendon</th>
<th>Force (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Achilles</td>
<td>816.78</td>
</tr>
<tr>
<td>Tibialis posterior</td>
<td>31.55</td>
</tr>
<tr>
<td>Flexor hallucis longus</td>
<td>36.16</td>
</tr>
<tr>
<td>Flexor digitorum longus</td>
<td>14.05</td>
</tr>
<tr>
<td>Peroneus longus</td>
<td>47.91</td>
</tr>
<tr>
<td>Peroneus brevis</td>
<td>33.66</td>
</tr>
</tbody>
</table>

**Figure 4.2-1:** Muscle force values applied in the foot and ankle model for the heel off stage of the gait cycle.

With the forces of the muscles calculated, the next challenge was how to assign values to muscles that are represented by multiple force elements in SolidWorks Motion. In the case that multiple elements are in series to accommodate tendon wrapping, each of the force
elements was assigned the same force value of the total force generated by the muscle. When a muscle had multiple insertion points represented by elements in parallel, the total force of the muscle was evenly split among each force element.

Muscle forces were not immediately activated in the simulation. Instead, the muscle forces ramped up from 7.5 seconds in the simulation time frame and reached the final force value at 9 seconds. The ramping of the force overtime helped prevent high frequency changes that a step function could create if the muscles were instantly activated. The final 3 seconds of the simulation had all the muscle forces constant and allowed the model to fully settle into a steady state.

4.2.2 Muscle/Tendon Paths and Wrapping

4.2.2.1 Triceps Surae and Achilles Tendon

While the triceps surae muscles did not directly impact the model (at least when the ankle was locked in place), they were still incorporated to make the model more extensible to other studies (see section 4.5). Both the two headed gastrocnemius and the soleus are part of the triceps surae and insert into the posterior calcaneus through the Achilles tendon. Even though the two muscle groups of the triceps surae have different origins, their shared insertion and similarity of the line of force when standing allowed for the two muscles to be modeled as a single structure. This is because when standing, the origin of the gastrocnemius is superior to the soleus’ origin. This assumption may not hold for extreme knee flexion because the gastrocnemius originates on the distal posterior femur while the soleus originates on the proximal posterior portion of the tibia/ fibula complex.

The loading pin assembly on the proximal tibia was used as the triceps surae origin. The rod that extended 55 mm in the posterior direction from the loading pin was used to approximate the origin of the muscle. Four force vectors were then created in SolidWorks Motion from the origin point to insertion points on the superoposterior of the calcaneus
(Figure 4.2-2). The calculated force of 816 N was divided among the four force elements evenly.

![Figure 4.2-2: Superior view of the posterior calcaneus (red) with the insertions of the Achilles tendon highlighted.](image)

### 4.2.2.2 Flexor Hallucis Longus and Flexor Digitorum Longus

Both the FHL and FDL descend before wrapping under the medial posterior portion of the ankle. To create the wrapping effect in the computational model, intermediary connections were used as accomplished with the plantar fascia and LPL. This resulted in proximal and distal portions of the muscle path. For the FHL, two beads were constrained to follow the circular edge of a cylinder that approximated the talocalcaneal tunnel that the tendon follows (Figure 4.2-3). The two beads were constrained to remain 10 mm from each other. The proximal muscle force for the FHL originated on a point marker posterior of the distal tibia and inserted in the proximal bead. The origin was created to match published information on the path of the FHL [47]. The distal portion of the muscle path ran from the distal bead to a point slightly inferior to the first metatarsal head. The forces still acted on
the first metatarsal, but the inferior offset of the insertion was to preserve the line of force since the tendon wraps around the metatarsophangeal joint.

![Diagram showing tendon path and force vectors](image)

**Figure 4.2-3**: Medial view of the tendon path for the FHL. The intermediary beads (red) locked as 10 mm apart from each other and mated to the cylinder (blue) which was locked in position relative to the calcaneus. When the FHL muscle force was applied from the muscle origin to the proximal bead and from the distal bead to the muscle insertion, the line of action of the force wrapped around the posterior of the ankle and also supported the medial longitudinal arch.

A similar approach was followed for the FDL but with only a single intermediary bead. This single bead was constrained to a curve that roughly followed the inferior surface of the navicular (Figure 4.2-4). This allowed the FDL to support the navicular and by extension, the longitudinal arch. The proximal portion of the muscle was modeled as a straight line from a posteriorly offset point on the distal tibia to the intermediary bead. From there, four different elements ran from the intermediary bead to points inferior of metatarsals two through five.
Figure 4.2-4: Medio-inferior view of the tendon path of the FDL. The single intermediary bead (red) is mated to the surface of the cylinder (blue) which is locked relative to the navicular bone. The bead has a force element towards the proximal origin and four forces to the distal muscle insertions.

Ultimately, this arrangement did increase computational cost of the model because of the beads moving on a curved path. However, the advantages were twofold. Firstly, it accurately modeled the path of the muscles which otherwise would have been modeled to pass through bone. Secondly, the setup helped model the windlass mechanism by which the metatarsal heads are pulled posteriorly, leading to changes in the longitudinal arch [42, 49, 50].

4.2.2.3 Posterior Tibialis

The last muscle that wraps around the posterior medial ankle is the PT. Unlike the FHL and FDL, no intermediary beads were used. This was because the PT mostly inserts into the navicular and medial cuneiform (Figure 4.2-5). The other insertions points are on the second metatarsal base. Again, the muscle origin was represented as a point offset from the posterior of the tibia based on published information on the PT’s path [76]. From the origin, four force elements have a straight line to their insertion points on the navicular and
medial cuneiform. Three elements also connected to the metatarsal bases. The 32 N force was shared evenly across the seven elements. Since the PT does not insert as distally as the FHL and FDL, wrapping was not a concern. Even with straight force elements, the PT still maintained a line of action along the medial longitudinal arch, which the PT helps raise when active [50, 76].

![Figure 4.2-5](image)

**Figure 4.2-5:** PT tendon was modeled to transmit force from a point on the posterior of the tibia along the plantar surface of the medial longitudinal arch where it inserts onto the navicular (Nav) and medial cuneiform (CN1) bones.

### 4.2.2.4 Peroneus Longus and Brevis

The PL and PB tendons both descend along the posterior fibula before wrapping along the lateral surface of the calcaneus. From the calcaneus, the two muscles have diverging paths. The PB inserts into the base of the fifth metatarsal while the PL turns sharply to enter the plantar peroneal tunnel under the cuboid. After exiting the tunnel, the PL tendon crosses the rest of the foot to insert on the first metatarsal base and the medial cuneiform.
Again the lines of action would not permit a single straight force element from the posterior “origin” point relative to the fibula to the tendon insertion.

Both force lines of action used two beads for wrapping along the lateral surface of the calcaneus. Both muscle forces “originated” from a point offset from the posterior of the fibula with the PL point being more lateral. From the origin points, muscle forces were connected to the proximal beads. Again, the two intermediary beads for the PL and PB path were constrained to stay 10 mm apart from each other while also remaining on a curve representing the tendon path. For the PB, the distal bead then connected to the base of the fifth metatarsal with three elements of 11.2 N each. The distal bead of the PL tendon path connected to a second set of intermediary beads in the plantar peroneal tunnel. These beads had the same constraints as others discussed. The distal bead in the plantar peroneal tunnel then had four force vectors from it to points on the first metatarsal base and the medial cuneiform.

**Figure 4.2-6:** (Left) A lateral view where the intermediary beads of the PL and PB tendons were mated to the cylinders that were locked relative to the calcaneus. The PB tendon inserts into the fifth metatarsal base while the PL tendon enters the peroneal tunnel on the plantar surface of the cuboid. (Right) The second set of intermediary beads for the PL were mated to the surface of the cylinder which was locked relative to the cuboid. The PL tendon forces then terminate on the medial side of the foot.
4.2.3 Additional Measurements and Model Runs

With the validated models, measurements in addition to diastasis in the Lisfranc joint and metatarsals were captured. These included a third diastasis measurement in the interosseous space of the Lisfranc joint, the transverse arch height, the radius of curvature for the transverse arch, and the intermetatarsal angle for the first two metatarsals (Figure 4.2-7). In the case of interosseous diastasis measurement, the motivation was to have more points of comparison throughout the joint’s depth. Values were determined based on the coordinates for a pair of points during the steady state period of the simulation. A MATLAB script calculated the diastasis measurements based on the time period selected for steady state.

The additional measurements for the transverse arch height and radius of curvature helped identify if transverse arch was collapsing. They have been listed as an alternative indicator of Lisfranc injuries [13, 77]. The transverse arch height was calculated based on the normal distance between the most inferior portion of the intermediary cuneiform and a plane defined in the model. This plane was defined by points offset from the inferior surface of the first metatarsal, fifth metatarsal, and calcaneus. The offsets from each bone were determined based on the thickness of the fat pads between the bone and the plantar plate in the original CT scan. With these three points, a plantar plane was defined.

The radius of curvature was calculated based on the most inferior points of the medial cuneiform, intermediary cuneiform, and cuboid. The three points were projected into the plane to represent the view of an anteroposterior radiograph. With the three points in this plane, a circle was fit to them. The radius of the fit circle was used to measure if the arch was flattening (i.e. larger radius).

The last measurement was the intermetatarsal angle between the first and second metatarsals. This measurement takes the long axes of each metatarsal and traces them posteriorly until the axes cross. The point where the axes cross is the vertex of the intermetatarsal angle when in a dorsal view [47]. The logic was that loss of tissue constraints between the first
and second ray (i.e. Lisfranc ligaments) would lead to changes in the intermetatarsal angle.

Figure 4.2-7: Additional measures: (Left) Transverse arch height as the normal distance (orange arrows) between the plantar plane of the foot and the most inferior point of the intermediary cuneiform. (Center) Intermetatarsal angle as the angle between the diaphysis of the first and second metatarsals (7). (Right) Transverse arch radius as the radius of the circle (red) fit to the most inferior points of the medial cuneiform, intermediary cuneiform, and cuboid projected on the frontal plane. Metatarsals not shown for clarity.

With the addition of muscle forces and the additional measurements, model runs were performed for the intact healthy model as well as for the injured model. Both cases were in 30° plantarflexion as in the model validation simulations. The diastasis measurements were calculated as the difference in separation for the model with muscles minus the separation of the model with 20 N load without muscle forces. The additional measurements (third diastasis value and other mid foot measures) were also extracted from the previous validated healthy and injured models that did not have muscle forces. These additional measurements could be captured without re-running the model and allowed for additional points of comparison with the models incorporating muscle forces.

4.3 Results with Muscle Forces

The incorporation of muscle forces did increase the solve time for models, but model settings did not need to be changed. The muscle forces slightly reduced the amount of diastasis in the simulation of the plantarflexed healthy foot (Figure 4.3-1). In the case of the
dorsal diastasis, the muscle forces actually had a smaller diastasis value than the model with 20 N loading, thus giving it a negative value. As an interesting note, the diastasis values for the intact model with muscles still fell within the standard deviation of the cadaver experiments.

When examining the additional measurements of the mid foot for the healthy model with muscle forces, very little change was noted. The largest change was the increase in the transverse arch radius with muscle forces (Figure 4.3-2), which suggests a flatter arch. Since the radius of the transverse arch is calculated based on a circle fit to plantar points on the foot, when the points form a more linear shape as the arch flattens, then the radius of the calculated circle fit will have a larger radius.

<table>
<thead>
<tr>
<th></th>
<th>Transverse Arch Height (mm)</th>
<th>Intermetatarsal Angle</th>
<th>Transverse Arch Radius (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Healthy</td>
<td>33.8</td>
<td>8.9°</td>
<td>18.4</td>
</tr>
<tr>
<td>Healthy with Muscle</td>
<td>33.9</td>
<td>8.8°</td>
<td>19.1</td>
</tr>
<tr>
<td>Lisfranc Injury</td>
<td>33.1</td>
<td>7.4°</td>
<td>19.4</td>
</tr>
<tr>
<td>Lisfranc Injury with Muscles</td>
<td>34.4</td>
<td>7.7°</td>
<td>18.3</td>
</tr>
</tbody>
</table>

Figure 4.3-2: Additional results captured for the intact and injured models in 30° plantarflexion that characterize the transverse arch and mid foot.
When comparing the results for an injured model with and without muscle forces, again there was a decrease in the diastasis values with muscles (Figure 4.3-1). The transverse arch height and intermetatarsal angles slightly increased for the injured foot with muscle forces while the transverse arch radius decreased slightly. The change in transverse arch radius in the injured model was opposite of what was seen in the healthy model when comparing with and without muscle forces. The strange pattern in the healthy model with and without muscle forces was that there was a slight increase in arch height while the radius also increased. This combination of a higher arch (granted only 0.1 mm higher) and a flatter arch runs counter to what would be expected. Had the increase in arch height been larger, then seeing the concurrent increase in arch radius would be more surprising.

Comparison of healthy vs injured diastasis when muscle forces were active showed a similar pattern as the two models without muscle forces. Injured models had larger diastasis throughout the Lisfranc joint and between the metatarsals. Additionally, injured models decreased intermetatarsal angles whether muscles were active or not. However, the direction of change for the transverse arch height and radius from a healthy foot to an injured foot varied whether muscle forces were active or not. When no muscles were modeled, the height decreased and the radius increased after injury. This behavior would be expected because it signifies that the transverse arch is collapsing and getting flatter. When comparing healthy to injured models with muscle forces, the relationship was reverse with height increasing and radius decreasing. So despite the injury, the transverse arch is being pulled upwards and the arch is narrowing.

4.4 Muscle Force Discussion

The addition of muscle forces into the model demonstrated how stable the intact foot and ankle are. While the diastasis values decreased a small amount, none of the measurements captured changed by more than 0.5 mm of their values in models without muscles. With such
little change in the arch height and intermetatarsal angle, this demonstrates the stability of an intact joint regardless of whether there are muscle forces. The only possible change to consider with muscle forces in the healthy model is how the muscles impact the curvature of the transverse arch. The results indicate that muscle forces may slightly flatten the arch by increasing the radius of curvature. Since transverse arch collapse has been used to identify Lisfranc injuries [13,77], these results suggest a high level of suspicion should be used when relying on this phenotype as a diagnostic tool in order to avoid false positives.

In a more general sense, the presence of muscle forces in a model showed less diastasis than the model without muscle forces regardless of whether it is a healthy or injured model. The smaller amount of diastasis in an injured model with muscle activation may be a contributing factor to the 20-40% of injuries that are missed in initial evaluations [14]. Since most radiographic evaluations are performed with a dorsoplantar (DP) image or a weight bearing DP radiograph, muscle activation in the weight bearing evaluation may obfuscate the indicators from the clinician.

A major reason for including the tendon wrapping around the posterior of the ankle was to ensure the muscle forces would support the longitudinal arch of the foot. The role flexor muscles play in supporting and raising the longitudinal arch have been previously noted in the literature [49,72,78]. The comparison of the models with and without muscle forces suggest that muscle activation has a similar impact on the transverse arch of the foot. Whether the model is for the healthy or injured foot, muscle activation does increase the transverse arch height for the steady state. When considering that the muscles are putting tension between the forefoot and ankle that the muscles wrap around, it makes sense that the whole mid foot and transverse arch would be lifted relative to the plantar plane.

The intermetatarsal angle measurements highlighted how the first metatarsal moves after injury. Regardless of the presence of muscle activation, the intermetatarsal angle decreased when injured. In order for the angle to decrease, the first two metatarsals must move
closer to a parallel orientation. Since the medial cuneiform and first metatarsal are under compression, they are prone to shifting medially or laterally. With the Lisfranc ligaments cut, the two bones are able to shift medially which causes the opening seen in the injured case. The medial shift of the metatarsal’s base while the distal portion stays in place due to the distal intermetatarsal ligament causes the decrease in intermetatarsal angle. This medial shift of the first metatarsal base was qualitatively observed from videos of Bansal et al.’s experimental studies [57].

The incorporation of muscle forces suggests that cadaveric models that do not simulate muscle forces can overstate the observable instability of the foot after a Lisfranc injury. However, cadaveric studies still show the same trends in the Lisfranc diastasis between a healthy and injured foot as the model with muscle forces. This means that the cadaver study will give directional information but not an absolute value for the behavior of a physiologically intact foot in regard to kinematic measurements.

However, the measurements of transverse arch height and radius are dependent on the presence of muscle forces to accurately capture the intact physiological behavior. Without muscle forces, the height and radius of the transverse arch decrease and increase, respectively, when the foot is injured. However, the opposite happens when muscle forces are accounted for. A possible reason for this is due to the key role muscles play in adding tension to the longitudinal arch which raises the arched of the mid foot. Also, muscles such as the PT and PL directly insert into bones in or near the transverse arch. The additional forces in such proximity to the arch would naturally cause changes to the two measurements in question. But overall many of the changes in measurements of the transverse arch were around 1 mm or less. Unless these changes result in a step off seen in a lateral radiograph on the dorsal portion of the foot, it is unlikely that clinicians would consider this an indication of a Lisfranc injury [79].
4.5 Alternative Foot Positions

As previously discussed, Lisfranc injuries are notoriously hard to identify and are often missed in the initial clinical review of the patient [14]. The model showed that active muscle contraction while the foot is in a plantarflexed position can obfuscate the key signs of injury in weight bearing simulations. Weight bearing radiographs without flexion/extension have been recommended for identifying Lisfranc injuries since non-weight bearing radiographs will not show indications of injury for up to 50% of patients [54, 80]. A question was posed that if the patient with a Lisfranc injury does not maintain a neutral foot position (i.e. introduces inversion/eversion while maintaining no flexion/extension), for a weight bearing radiograph how may the presentation of the injury change? The speculation was that inverted or everted foot positions may be used by patients to reduce pain if regional anesthesia is not used during weight bearing radiographs. To evaluate the hypothesis that alternative foot positions (inversion/eversion) could change injury presentation, healthy and injured models were run with the foot loaded in the neutral position, with 10° inversion, or with 10° eversion provided. None of the models had flexion or extension of the foot. The diastasis values of the healthy neutral model and injured model in the different foot orientations were used to see which positions displayed the largest difference and therefore should be easier to identify in DP weight bearing radiographs.

4.5.1 Neutral Foot Position

To simulate a weight bearing radiograph, the model needed to be changed from 30° plantarflexion to a neutral position. Since the original CT scan for all the bones was in a neutral foot position, the origin and axes of the bones in the foot simply needed to be aligned (Figure 4.5-1). All other components (i.e. plantar plate and muscle wrapping beads and surfaces) either moved with the bones or were already correctly oriented.

In the neutral position, all of the bones of the foot were free to move with their only
constraints being the ligament/muscle forces and the contact geometries. This was different from the previous simulations reported in Chapter 3 and earlier in this chapter where the talus and calcaneus were locked relative to the tibia/fibula. As a result, ligaments such as the anterior talofibular and muscles such as the triceps surae actually did impact the motion in the model.

Since the model was now in a neutral stance, muscle force values required updating. Past computational models in the Orthopaedic Research Laboratory have focused on modeling foot biomechanics in neutral positions and have identified forces as a percent of body weight for the flexor muscles in the foot [6,7]. The values in Figure 4.5-2 were used in the following models of the different foot positions without any flexion or extension.
4.5.2 Inversion and Eversion of the Model

Inversion and eversion were simulated by rotating the plantar plate along its long axis by 10°. This approach was sufficient since the study was focused on the steady state positions and not on how the model reached that point. As the model was loaded with the body weight force, the tilted surface and free movement of the bones caused the foot to be invert or evert relative to the tibia/fibula which could still only translate superiorly or inferiorly.

4.5.3 Model Runs and Data Collection

In the three different foot orientations, models were run for the healthy and injured models with and without muscle forces. The three Lisfranc diastasis values were collected from the steady state period of the model run. The three measurements to characterize the transverse arch and mid foot were also captured. The difference between the injured models in different foot orientations were compared to the neutral healthy foot values. The goal was to identify the foot orientation that would have the maximal difference between the healthy and injured models and by extension should be the easiest to identify in the initial evaluation. This comparison was performed both with active muscle force and without. Figure 4.5-3 list the different model runs and comparisons for this study.
4.6 Results with Alternative Foot Orientations

For the models without muscle forces, the diastasis of the injured models varied based on the foot orientation with the inverted stance having the smallest difference in dorsal diastasis when compared to a healthy model (Figure 4.6-1 left). In the case that the muscle forces were included, the neutral stance showed the largest increase in the interosseous and plantar Lisfranc diastasis for the injured model compared to the healthy model (Figure 4.6-1 right).

The no muscle results show a smaller difference in injured vs healthy transverse arch height and radius of curvature when the foot is inverted (Figure 4.6-2). The intermetatarsal angle did not change as much during eversion for the injured model continuing the pattern of alternative foot positions showing less instability in joint motion. However, when muscle forces were incorporated, the absolute difference in transverse arch height and radius when injured was actually less than what was seen with inversion and eversion (Figure 4.6-2).
Figure 4.6-1: (Left) The difference in the diastasis for injured models compared to a healthy neutral model with muscle forces excluded. (Right) Same comparison of diastasis values but when muscle forces were active.

<table>
<thead>
<tr>
<th></th>
<th>No Muscle Forces</th>
<th>Muscle Forces</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Neutral</td>
<td>Inversion</td>
</tr>
<tr>
<td>Transverse Arch Height (mm)</td>
<td>-0.65</td>
<td>-0.05</td>
</tr>
<tr>
<td>Intermetatarsal Angle</td>
<td>-0.95°</td>
<td>-1.15°</td>
</tr>
<tr>
<td>Transverse Arch Radius (mm)</td>
<td>-0.73</td>
<td>-0.30</td>
</tr>
</tbody>
</table>

Figure 4.6-2: Table comparing the difference in several mid foot measurements between healthy and injured models for various foot orientations.

When comparing raw diastasis values, muscle forces again showed a reduction for injured models (Figure 4.6-3 shows the dorsal diastasis values). When the injured foot had muscle forces included, the diastasis in the Lisfranc joint, which is the common indicator clinicians look for, was smaller than when the foot had no muscle activation. Not only was this pattern seen in the neutral foot position, but also in the inverted and everted cases.
A last data point that was captured for comparison was an approximation of the center of contact pressure between the foot and the plantar surface. This was calculated for the different neutral foot model states (healthy vs injured) with and without muscle forces. The approximation was performed with a weighted average of coordinates of the contact points for the five metatarsals and calcaneus with the plantar plate. It was noted that across the four different models, the center changed very little. All four models had a center of pressure within 2.5 mm of each other.

4.7 Foot Orientation Discussion

Lisfranc injuries occur on the medial portion of the transverse arch. If a patient is trying to offload the medial portion of the foot because of pain, it would make sense that they would invert their foot. Regardless of whether muscle activation is present, the diastasis of the Lisfranc joint is reduced for an inverted foot position. So while the inverted foot position may off load the Lisfranc joint and reduce pain, it poses additional challenges for clinicians to identify the injury.
The diastasis values showed differences in the injured models with neutral, inversion, or eversion. The differences in the neutral versus eversion models were less than those seen in the neutral versus inversion stances. The measurement of the transverse arch height, radius of curvature, and intermetatarsal angle did not have as clear of a trend. When considering the models without muscle forces, the foot with inversion had transverse arch height and radius similar to the healthy model. But when muscle forces were incorporated, the neutral foot position resulted in the values most similar to a healthy foot. This suggests that the transverse arch collapse is less likely to be identified in neutral foot position. Again, this points to the fact that the muscle forces have a stabilizing impact on the mid foot and transverse arch.

While the alternative measurements in the models with muscle forces suggest that inversion or eversion would show a greater change in the transverse arch, these alternative measurements are not the standard indicators used by clinicians to identify injuries. The transverse arch height and radius are not calculated in patient radiographs [13] and were measures created in this study to try and capture changes to the transverse arch after an injury. Thus these measurements of the transverse arch have not been characterized clinically for Lisfranc injuries in the literature. This appears to be the result of most lateral radiographs for Lisfranc injuries focus on whether the second metatarsal has shifted superiorly from its anatomic position [12,54,79] and is not specifically focused on the geometry of the transverse arch. Additionally, the range of intermetatarsal values (which is a measured value reported in the literature) from the simulations all fell within what is a normal healthy range of less than 9° [47]. So while the alternative measurements of the mid foot are interesting for understanding the behavior of the bones, they would not be the indicator of injury in a clinical setting. Instead, the diastasis values are better options for understanding how easily an injury may be overlooked because of the changes in the foot caused by muscle force and foot orientation.
When approximating the center of pressure across the different models, little difference was noted. In this experimental setup, the presence of muscle forces and Lisfranc injuries did not change the location of the center of pressure on the plantar plate. Ultimately, all neutral models had a center of pressure in the similar area as those reported in the literature for healthy individuals (inferior to the medial-anterior portion of the calcaneus) [81]. The center of pressure was purposefully only captured in the neutral stance models in the hope of identifying if Lisfranc injuries alter the loading of a neutral foot. In reality, the presence of a Lisfranc injury may cause the patient to purposefully alter their foot orientation, as discussed previously, to reduce the amount of pain experienced. Naturally, this change in foot position would also shift the center of pressure for the foot.

Diastasis values were smaller when comparing the injured model with versus without muscle forces. This followed the same pattern as the studies previously discussed in this chapter when the model was in plantarflexion. Since muscle forces reduced the key injury indicator (Lisfranc joint diastasis) applied to all three foot positions, an injured individual that is not in a neutral foot position and generates muscle forces can greatly reduce the diastasis seen in weight bearing radiographs.

A possible solution to reduce the chance of muscles and foot position obfuscating the key indicators is for weight bearing radiographs to always use a regional anesthesia. This would impede muscle contraction and allow foot positioning without pain.

The shortcomings in this study of different foot orientations were that the muscle forces remained the same for the different stances and that a wider range of inversion/eversion angles were not tested. The flexor muscles do provide inversion and eversion moments. For example, the peroneus longus and brevis both cause eversion of the foot. When a patient places their foot in eversion, the peroneus longus and brevis would need to provide a different force while the foot is loaded. One reason this shortcoming was accepted was because the magnitude of the flexor muscles that provide inversion/eversion, is on a much smaller scale.
than the forces seen in the body weight or triceps surae. The other shortcoming was the limited number of angles tested. 10° was selected since it was the upper range of the subtalar’s inversion angle during a gait study [82]. However, additional angles above and below 10° could help better characterize the relationship between foot position versus indications of injury.
Chapter 5: Rigid Body Modeling and Finite Element Analysis of Surgical Repairs

5.1 Introduction

The next extension of the computational model was to consider how repair surgeries impact the post-surgery biomechanics of the foot. The main concern was how well the surgery was at providing a biomechanically stable Lisfranc joint after the surgery. Joint stability after surgery was important to consider since it has been an indicator in patient healing and outcomes. If the injured joint does not maintain a healthy position while performing normal activity after surgery, the patient healing and AOFAS ratings are worse [18]. A secondary concern was whether the stresses in the implanted hardware after surgery would be susceptible to yielding and failures. These cases always require re-operation and result in poor patient outcomes.

Thus far, all the computational models have used rigid body modelling (RBM). This has been sufficient because the main information of interest for those analyses were bone kinematics, ligament strains, contact forces, etc. However, the stresses that are generated in the structures can also be relevant to understanding the effects of injuries or repairs. Further, rigid body modelling assumes a surgical implant is rigidly adhered to the bone, with both acting as one unit. The constraint that bodies cannot deform would mean that the bones that screws pass through could not move relative to each other. The alternative computational technique of finite element analysis (FEA) was selected to allow deformation of the fixation hardware for analysis of the stresses generated. As discussed in Chapter 1, FEA uses the forces, material properties, and geometries of parts to identify the deformation that would occur. This allows for diastasis measurements to be captured based on the deformation of
the bones and hardware used in the surgery. Additionally, FEA calculates stress values for the parts in the simulation. The stress results can be used to identify points of failure in the bone or hardware which could require future surgeries to correct.

The resulting workflow was to first run kinematic rigid body motion simulations with each surgical repair construct in SolidWorks Motion. Then steady state forces and a subset of bones were then transferred into a finite element (FE) model in ANSYS Workbench (Ansys, Canonsburg, Pennsylvania). SolidWorks Motion allowed for the X, Y, and Z components of any contact force, or reaction force due to muscles or ligaments to be extracted. These contact, ligament, and muscle forces were applied to the FE model. The displacement information for the points of interest in the FE model were extracted to identify the diastasis values for the surgical repair.

Below are discussions on the implementation of the four surgeries of interest: open reduction and internal fixation (ORIF) with endobuttons, cortical screws, and dorsal plates and primary arthrodesis with screws. ORIF procedures can have their hardware removed at a later point and because of this, articular surfaces are preserved as much as possible. Arthrodesis by comparison includes debriding the articular surfaces of the bones so that when fixed in contact with each other, the bones will fuse together. ORIF with screws is the current gold standard when fusion of the joint is not required [18, 23, 83–85]. FEA was performed for all of the surgeries except for ORIF with endobuttons because it could be modeled with RBM while the other three procedures needed FEA. For this reason, ORIF with endobuttons is discussed first.

5.2 Open Reduction and Internal Fixation with Endobuttons Modeling

Endobuttons, also called suture buttons, are beginning to find more use and recommendation in orthopaedic surgeries [86, 87]. The material used may vary but they are two buttons with a wire/string running between them. The endobutton is designed so that the buttons are on
either side of two or more bones that have a tunnel drilled through them. The wire passes through the tunnel which is too small for the buttons to fit into. The wire is tightened to apply tension that will bring the different bones closer together. This is designed to fix in place unstable joints or bones.

The advocates of this procedure point out that it allows earlier weight bearing which provide mechanical signals that can help promote tissue growth and remodeling. Also, earlier mobilization can mean patients can begin normal daily activities sooner [21]. Detractors have pointed out that it provides less mechanical stability which could disturb initial healing responses [25].

5.2.1 Kinematic Model

The endobutton model like all of the other surgical repair models was based off the injured model with muscle forces in 30° plantarflexed position. A procedure using two endobuttons was selected based on past studies by Marsland et al. [21]. One endobutton spans from the medial surface of the medial cuneiform laterally to the lateral surface of the intermediary cuneiform. The second endobutton spans from medial surface of the medial cuneiform anterosuperiorly to the lateral surface of the second metatarsal base [21] (Figure 5.2-1).
Figure 5.2-1: Graphical representation of endobutton repairs which ran from the medial cuneiform (CN1) to either the intermediary cuneiform (CN2) or the second metatarsal (MT2). The red brackets identifying the portion of the fiberwire modeled in SolidWorks Motion.

Since the wire passes through holes in the bones, the motion of the bones is constrained relative to the points where the wire exits the bone holes in the interosseous space. Therefore the physical representation of the fiberwires in SolidWorks Motion was from the lateral surface of the medial cuneiform to the medial surface of the intermediary cuneiform or second metatarsal base. The origin and insertion points were selected to be points that fell on the lines previously identified for the endobutton path. The physical body of the endobutton was thus not required to be included in the model.

5.2.2 Endobutton Stiffness

The material properties to define the wire stiffness were based on the Mini TightRope FT-AR8917DS from Arthrex Inc. (Naples, Florida). This product uses a Number 2 FiberWire. Experiments by Najibi et al. determined the stiffness to be 35 N/mm for a 50 mm sample with a cross-sectional area of .37 mm$^2$ [88]. This equates to a modulus of 4.73 GPa. Since
the length of the wire spanning the joint and the cross sectional area were known, stiffness values for one wire across the Lisfranc joint and intercuneiform joint were calculated to have a stiffness of 54.62 N/mm and 76.65 N/mm, respectively. Each value was multiplied by four to get the total stiffness of the four wires spanning the endobuttons. Each fiberwire was modeled as a tension only element, the same way that ligaments were defined. Additionally, a 3% in situ strain was applied to the wire to simulate the tension the surgeon applies before the wire between the buttons is tied off. With these additions to the injured joint, the 30° plantarflexed injured model was ready to run.

5.3 Open Reduction and Internal Fixation with Screws Modeling

ORIF with screws was one of the first procedures used and continues to be the most commonly recommended repair for Lisfranc injuries [18, 83–85]. A standard setup is to use two transarticular, cannulated screws and drill them from the medial cuneiform to the intermediary cuneiform and second metatarsal base. This setup is very similar to the orientations used with the endobutton procedure.

5.3.1 Hardware Models

The screws used in the model were based on 3.75 mm fully-threaded AR-7000 screws from Arthrex Inc. A model was created with the threaded screws and simplified cylinder screws. These variants were used in the FEA work because the high curvature of the threaded screws can lead to challenges in FEA pre-processing. A sketch of the thread profile is shown in Figure 5.3-1. The intercuneiform screw (medial to intermediary cuneiform) was modeled as 23.4 mm and the Lisfranc screw (medial cuneiform to second metatarsal) was 35.5 mm.
5.3.2 Kinematic Model

In SolidWorks® a new assembly was created using the medial and intermediary cuneiform, the second metatarsal, and two screws with threads. The bones were oriented in their neutral position by aligning the origins of the three bones. The long axis of each screw was then mated to the line representing its insertion direction. Special care was taken to ensure that the screws did not intersect with each other. The screws were inserted until the heads began to intersect with the medial cuneiform. A Boolean subtraction from the bones was performed. In any case where the volume of screws and bone overlapped, the bone volume was subtracted. After the subtraction, the screws were replaced with exact copies of themselves that were cannulated so no bone material was in the center of the screw.

The final subassembly of the bones and screws were then added to the plantarflexed
injured model with muscle forces (Figure 5.3-2). The existing medial and intermediary cuneiforms and second metatarsals were replaced by the subassembly version of said bones with screws. In SolidWorks Motion the ligament and muscle forces needed to be redefined to connect with the points on the bones in the subassembly. Contact groups also needed to be redefined. Lastly, an additional set of contact groups were set between the screws and the bones they contacted.

**Figure 5.3-2:** Example of the kinematic model used for ORIF with screws. The fixation screws pass from the medial cuneiform (CN1) to either the second metatarsal (MT2) or the intermediary cuneiform (CN2).

Once the elements in SolidWorks Motion were updated, the simulation was run. The steady state time frame of the model was determined based on when motion in the bones stopped. Then using the SolidWorks Motion results tools, the force’s X, Y, and Z components were exported for any ligament, muscle, or contact force on the two most medial cuneiforms and the second metatarsal. Using a custom MATLAB script, the average force value during steady state by component for each force was calculated. These forces and components were then used in the model setup for the FEA portion of the work.
5.3.3 Finite Element Modeling

Since the screws passed through the bones, the rigid body motion simulations would not show any movement of the bones relative to each other. Extracting the forces from the kinematic model and using them in finite element modeling allowed simulation for deformation of the bones of the Lisfranc joint relative to each other. Since the points of interest are in the Lisfranc joint, only the geometries of the subassembly with screws from the kinematic model were required in the FE model. The only change to the bone geometries was to incorporate cancellous bones into the model. Incorporating cancellous bone was important because of the different material properties of the cancellous versus cortical bone. Cortical has a modulus that is an order of magnitude higher than cancellous. If the bones were treated entirely as cortical bone, then bone deformation would be underestimated compared to true behavior.

The three procedures in this study that utilized FEA followed the common workflow outlined below. To prevent repetition, the other two FEA studies will only focus on portions of the FE model setup that were different.

5.3.3.1 Cancellous Bone Model

The cancellous bone of the medial and intermediary cuneiform was created using the base masks in MIMICS. Since the cortical thickness of these bones are fairly consistent throughout, the masks of these bones were eroded with a morphological command. The resulting mask fit entirely within the original mask with the difference between the walls of the masks as $\sim 0.75$ mm. A similar procedure to those described in Section 3.3.1 were used to smooth the mask and create SolidWorks bodies.

The erode approach would not work for the metatarsal since the cortical thickness varies by region. Instead the cancellous bone model for the second metatarsal was generated using MATLAB. The existing STL file for the external surface of the second metatarsal was loaded
in MATLAB. Then the vertices of the STL file were moved .75 mm to 2.4 mm in the opposite direction of the external surface’s normal direction at that point. A grey scale image was used to define the magnitude of the move where the gradient changed along the anterior to posterior direction. The area of largest displacement on the original surface was in the diaphysis of the metatarsal and the smallest change was in the head and base of the bone. The result was a body representing the cancellous and intramedullary canal of the second metatarsal (Figure 5.3-3). The MATLAB result was exported as an STL file where it was processed in 3-Matic before it was used to create a SolidWorks part.

![Diagram showing the process](image)

Figure 5.3-3: Example of the process to create the volume for the cancellous and intramedullary canal that could then be subtracted from the original bone to generate the volume of cortical bone.

Once the cancellous bones were created as their own unique parts, they were added to a copy of the fixation subassembly used in the kinematic model. The cancellous bones were set so that their origin and coordinate axes lined up with the other bones. Additional Boolean subtractions were then performed to make sure that the cancellous bones and existing bones did not overlap. The result was the cancellous bones and a cortical shell that surrounded the cancellous bone. The metatarsal required an additional change to model the intramedullary canal. The proximal 15.4 mm and distal 17.1 mm of bone along the long axis were preserved.
while the rest was removed. The result was a volume of cancellous bone in the head and base of the metatarsal, but the intramedullary canal was void of material.

5.3.3.2 Forces and Boundary Constraints

Ansys Workbench was selected to organize the FEA study. SolidWorks® offers a built in functionality to perform FEA; however, Ansys offered a more robust suite of tools that provide more options when generating meshes to represent geometries or setting parameters for contact and solver variables. The subassembly of the fixated bones and screw were loaded into Ansys Workbench using the DesignModeler module.

The module for pre-processing and running the FEA was Ansys Mechanical. Once the geometry was in the Ansys workflow, Ansys Mechanical automatically identifies contact groups based on part proximity. For a static study like this, it meant that no additional contact groups were needed. The cancellous core of each bone was treated as bonded to the cortical shell. The threaded screw was treated as bonded to the bone that it contacted. The model variants where simplified screws were used had rough contact sets that prevented the screws from sliding relative to the bone. The contact between the cortical shells of two bones were treated as frictionless due to the articular cartilage that cover these surfaces.

The contact sets have a variety of control parameters that can be designated in Ansys Mechanical (Figure 5.3-4). If no selection is made by the user, Ansys will select from the options based on the overall study when the analysis runs. The one setting that was always controlled was the “Type” of contact. The options were bonded, frictionless, frictional, rough, and no separation. Frictional and frictionless allowed the parts to separate from each other as well as slide relative to each other. Rough allowed parts to separate but treated the surfaces as having infinite coefficients of friction. No separation allowed sliding, but parts had to remain in contact and lastly, bonded parts could not separate or slide relative to each other. When contacts were modeled as bonded, the rest of the settings were left as programmed controlled. In the cases that frictionless or rough contact types were used, the
The formulation setting was changed to Augmented Lagrange. This formulation improves solve
times when contact settings allow for separation of the two bodies from one another [89].

![Table of parameters](image.jpg)

**Figure 5.3-4:** Example of the set of control parameters in Ansys Mechanical that define contact settings. This example has defined the contact type as rough (i.e. the parts can separate but are not able to slide against each other.)

The next step of the pre-processing of the FEA was to generate meshes for the geometries
in the model. Meshing converted the geometry into a collection of points (i.e. nodes) that
make up 3D normal shapes (i.e. tetrahedron) that are called elements. In this case,
the elements were 4 noded tetrahedral. Ansys Mechanical allows for a Mesh Control called
Sizing to factor into the size of the elements. This functionality was used to ensure that
the screws with threads were able to be meshed. Depending on whether the model was run
with threaded screws or simplified screws, the number of elements in the model ranged from 12,831 (without threads) to 77,372 (with threads).

The next portion of the pre-processing work in Ansys Mechanical was assigning forces and fixation constraints. Since the model was in the same orientation as the kinematic model, the different forces could be defined based on the X, Y, and Z components of each force vector. The regions of the bone surfaces where the force acted were selected manually. 37 different forces were used in the ORIF screw FEA. The only fixed surface that could not move in the simulation was the plantar surface of the second metatarsal. Figure 5.3-5 shows the different forces and fixation points on the FE model. Forces used in the model are listed in Appendix B.

![Figure 5.3-5: Visual representation of the contact muscle and ligament forces on the FE model consisting of the second metatarsal and medial and intermediate cuneiform. The magnitude and direction of the forces were extracted from the RBM done in SolidWorks Motion.](image)

### Material Properties and Solver Parameters

The last portion of the FEA setup was to provide the properties for the different materials in the model and set some of the solver parameters. The materials in the model were cortical bone, cancellous bone, and titanium. All of the materials were treated as isotropic and
elastic. While this is not a perfect description of bone, for the purposes of this study, they were sufficient. The values that were needed to define the material were the density, Young’s modulus, Poisson’s ratio, and yield strength. These values are listed in Figure 5.3-6.

<table>
<thead>
<tr>
<th>Material</th>
<th>Density (kg/m³)</th>
<th>Young’s Modulus (Gpa)</th>
<th>Poison’s Ratio</th>
<th>Yield Strength(Mpa)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cortical Bone</td>
<td>1850</td>
<td>15</td>
<td>0.3</td>
<td>100</td>
</tr>
<tr>
<td>Cancellous Bone</td>
<td>1100</td>
<td>1.5</td>
<td>0.2</td>
<td>2</td>
</tr>
<tr>
<td>Titanium</td>
<td>4425</td>
<td>104.8</td>
<td>0.31</td>
<td>827.37</td>
</tr>
</tbody>
</table>

Figure 5.3-6: Material properties used in the FEA.

Ansys Mechanical has several solver variants that perform better for different studies. Tests with both direct and iterative solvers were tested. Ultimately, the direct solver was robust enough to handle the ORIF with screws FEA, but some of the other surgical simulations required the iterative solver to reach model convergence. The rest of the settings for the solver could be set to program controlled. The solve time was significantly longer for the model with threads. As an example the ORIF with threaded screws solved in roughly two and a half hours while the simplified screw model solved in ten minutes when running on an Intel i7 CPU with 32 GB of RAM.

5.3.3.4 Data Collection

With the pre-processing work completed for the FEA, the simulation is ready to run. Once the simulation completed running, the original and final coordinates of all the nodes were exported. Nodes correlating to points on the dorsal, interosseous, and plantar portion of the Lisfranc joint were identified. The diastasis for each depth of the Lisfranc joint was calculated based on the nodes’ original and final coordinates.

Additionally, von Mises stress contours were generated for the different materials in the model. Von Mises stress uses the components of the 3-by-3 stress tensor for the body in question to calculate a single stress value that can be compared to the yield criteria [90]. While these stress contours were used to evaluate whether bone would yield and potentially
lead to screw loosening, the main focus was to capture the stress distributions in the screws. Screw breakage in the interosseous space of the Lisfranc joint is a major failure reason for ORIF with screw [18, 19].

Three variants of the FEA study were run: threaded screws bonded to the bone, simplified screws without threads with rough contact to the bones, and simplified screws but with a mesh with more elements. The last model where the mesh of the model was changed was a convergence test. Its purpose was to identify whether the results of the model were dependent on the mesh used. Since the meshing process simplifies the geometries, convergence testing is to make sure that the results for a mesh with fewer elements does not give drastically different results because of the size of the elements in the mesh. The two model runs had 12,831 nodes and 8,239 nodes. Specific results that were evaluated included diastasis measurements in the Lisfranc joints, the locations and maximal stresses in the different parts and a more subjective visual examination of whether the color maps of the stress contours showed the same patterns.

5.4 Open Reduction and Internal Fixation with Dorsal Plates Modeling

The final ORIF procedure to be modeled was using a dorsal plate and four screws. This procedure is less common than transarticular placement of cannulated screws, but has been noted to not disturb or further injure the articular surfaces of the Lisfranc and intercuneiform joint spaces [23, 83, 85]. The dorsal plate of interest in this study used four screws that are drilled into the medial and intermediary cuneiforms and the first and second metatarsals. The main concern with these types of hardware is whether they provide sufficient stability in the transverse plane through the full depth of the injured joint [83, 85].
5.4.1 Hardware Models

The dorsal plate and screws were modeled after the Arthrex medium Lisfranc plate (AR-8951). This is a four holed plate that allows for locking screws in three of the holes and a bone fixation screw in the fourth. Dimensions of the dorsal plate were collected from physical measurements. The largest challenge was to capture the curvature of the plate. In a surgery, the plates are physically bent to allow the plate to sit roughly flush on the dorsal surface of the four bones. This involved updating the curvature of the plate when placing it in the model. The model of the dorsal plate is in Figure 5.4-1.

![Model created in SolidWorks based on the dorsal plate AR-8951](image)

**Figure 5.4-1:** Model created in SolidWorks based on the dorsal plate AR-8951 [91].

The screws used in the model were 3.5 mm locking and fixation screws. These screws unlike those used in the ORIF with screws were not cannulated. 22 mm long screws were used to connect the dorsal plate to the cuneiforms and the first metatarsal while a 15 mm long screw was used in the second metatarsal. The shorter screw for the second metatarsal was to make sure that the screw did not break through the plantar surface of the bone.
5.4.2 Kinematic Model

A new assembly of the four bones with the dorsal plate placed in a neutral position was created in SolidWorks®. The dorsal plate was updated to fit roughly flush on top of the four bones and the four screws were inserted along the axis of the dorsal plate holes into the bones. Boolean subtraction from the bones were performed so that the screws did not overlap with the bone volumes. The final fixation subassembly was added to the injured foot and ankle model with muscle forces (Figure 5.4-2).

![Kinematic model in SolidWorks Motion for the dorsal plate ORIF. Screws are drilled towards the plantar surface of the bones](image)

**Figure 5.4-2**: Kinematic model in SolidWorks Motion for the dorsal plate ORIF. Screws are drilled towards the plantar surface of the bones

In the validated kinematic model, the first two metatarsals and the medial and intermediary cuneiforms were removed and replaced with the assembly discussed above. The different force elements for muscles and ligaments were updated to point to the bones of the assembly with the dorsal plate instead of the previous bones that were removed. Contact sets were also redefined to include the parts of the new subassembly. The last update was to create contact
sets between screw, dorsal plate, and bones.

The kinematic model was then run in SolidWorks Motion using previously discussed model parameters. X, Y, and Z components for forces acting on any part in the dorsal plate fixation assembly (i.e. dorsal plate, screws, or any of the four bones the screws entered) were extracted from the completed kinematic model using the same process as section 5.3.2. The contact with other bones and reaction forces from soft tissue or muscle that were extracted from the kinematic model in SolidWorks® were used in the FEA study to help determine the stability of the Lisfranc joint with the dorsal plate and the stresses in the structure.

5.4.3 Finite Element Modeling

The same workflow was performed for the dorsal plate ORIF FEA as the one used in the ORIF with screws. As a reminder, this involved incorporating cancellous bone volumes into the SolidWorks® fixation subassembly, importing the subassembly geometry into Ansys Workbench, using Ansys Mechanical to define contact parameters, meshing the geometries, adding the forces and fixation points to the different parts, and finally running the FEA.

The only additional cancellous bone that needed to be generated was for the first metatarsal. In order for the cortical shell to be thicker in the diaphysis of the metatarsal, the same MATLAB program for the second metatarsal cancellous bone volume was used. This moved the vertices of the STL file in the opposite direction of their normal vector by a value scaled based on a grey scale image varying from anterior to posterior anatomical locations.

Once the parts of the SolidWorks® model were placed and Boolean subtraction was performed to remove screw and bone overlap, the model was imported into Ansys Workbench. In the Ansys Mechanical module, the contact sets were updated so that any cortical bone on cortical bone contacts were set to frictionless. In the model case where the screws were threaded, any contact sets with the screws were left as bonded. When the screws were simplified to simple cylinders, the contact sets were set to rough (i.e. allowed separation
but no sliding of the parts relative to each other). 31 forces from the kinematic model were included in the FEA (Figure 5.4-3)

![Figure 5.4-3: FEA model in Ansys Mechanical for the dorsal plate ORIF consisting of the first and second metatarsals and the medial and intermediary cuneiforms. The contact, muscle and ligament forces which were taken from the RBM in SolidWorks Motion are shown in red arrows.](image)
threaded screws. Von Mises stress contours were generated for the different materials in the model to understand any areas susceptible to yielding.

5.5 Primary Arthrodesis Modeling

Primary arthrodesis is the recommended procedure for when the bones in the foot develop end stage arthritis [26]. The advantage of arthrodesis procedures is that they are designed to promote multiple bones to fuse together into a single body as part of the healing process. The final state of bones rigidly fused together has become an option for treating Lisfranc injuries as well [19, 22, 26]. Exact screw placement and which bones are fixed in place vary by the patient’s injury and clinician’s preference [22]. However, the core of the procedures remains the same. First, the articular surfaces are debrided to expose the subchondral bone for any joints that are being fused. An autograft scaffold may be placed in the joint space to promote fusion [22], but because of the small joint spaces of the mid foot, autografts are not used by all clinicians [92]. In this model, no scaffold was added. The bones that are meant to be fused are then brought in contact with each other. The bones are then fixed in place with screws.

In this study, an arthrodesis procedure presented by Cochran et al. was utilized [22]. The number of screws and bones involved agreed with other studies on Lisfranc joint arthrodesis [19, 26, 92, 93]. The procedure was further confirmed with notes from Dr. Robert Adelaar of the McGuire VA Medical Center on how he has performed these procedures. In this case, all three cuneiforms and the first two metatarsals were fixated for fusion (Figure 5.5-1). One screw ran from the medial cuneiform through the other two cuneiforms. Two screws passed in opposite directions from the dorsal surfaces of the medial cuneiform and the first metatarsal. A fourth screw ran from the medial cuneiform across the Lisfranc joint to the second metatarsal. The last screw passed from the second metatarsal to the intermediary cuneiform.
5.5.1 Hardware Models

The same cannulated screw design (Arthrex AR-7000) used in the ORIF with screws was also used for the arthrodesis study. There were two screw lengths used in the model. Four of the screws were 34 mm long while one screw which crossed the Lisfranc joint was 38.1 mm. These lengths were able to span the required bones and had insertion depths similar to those in Cochran et al. based on visual inspection [22].

5.5.2 Kinematic Model

A new assembly in SolidWorks was created using the three cuneiforms, two metatarsals and five screws. The bones were placed in their neutral position by mating their origins and axes. A guideline was created for each screw to represent how they would be drilled into
the bones. To imitate the debriding process \textit{in vivo}, the bones were translated along the different guidelines to bring the articular surfaces of the bone in contact with each other. When the majority of the articular surface was in contact with each other for a joint, Boolean subtraction operations were performed to remove the overlap and represent the removal of the articular surface of the bones in the arthrodesis procedure.

Once the bones were in place and the debriding had been simulated, the screws were added to the model and mated to fit on the guide lines. The screws were “drilled” into the bone until the head of the screw began to overlap with the bone. Again Boolean subtraction was used to remove the bone overlap with screws.

The new subassembly was added to the injured model with muscle forces. The original cuneiforms and first two metatarsals in the model were suppressed and replace by the new fixation subassembly. Once the force elements and contact groups in the model were updated to incorporate the replaced parts, the kinematic model was ready to run. The steady state time period of the model run was determined based on when bones in the model stopped moving. With the steady state time period know, any forces that impacted the three cuneiforms or first two metatarsals were extracted from SolidWorks Motion.

![Image of bone model](image)

**Figure 5.5-2:** The arthrodesis RBM in SolidWorks Motion based on the procedure used in Cochran et al. [22].
5.5.3 Finite Element Modeling

The same procedures as discussed in Section 5.3.3 were used to generate the FE model. Cancellous bone was added to the arthrodesis subassembly. Cancellous bone needed to be generated for the lateral cuneiform but a simple erode operation in MIMICS sufficed since the cortical thickness of the bone in the CT scan appeared largely uniform in thickness. Cancellous bone for the other four bones was reused from the previous models.

Contact parameters between cortical bones were different than those used in the other FEAs. For arthrodesis the contact between bones was modeled as rough which did not permit sliding of the bones relative to each other. This was to represent the removal of the articular surface of the bones when they were debrided.

Like the FEA for the dorsal plate, special attention was needed for mesh sizing control to ensure the number of elements remained under the 200,000 limit. The FE model with threaded screws contained 185,000 elements while the simplified screw models had less than 60,000 elements. The 38 forces from the kinematic model were added into Ansys Mechanical and the plantar surfaces of the two metatarsal heads were fixed in the model (Figure 5.5-3). All of the screws were modeled as titanium and the rest of the bodies in the model were cortical or cancellous bone.
Three variants of the model were again performed. The first one used screws with threads where the screw was bonded to bone contacts. The other two both used simplified screws with a rough contact with bones. The difference in the model was in the number of elements and was done to convergence test the model (26,680 nodes versus 16,776 nodes). Lisfranc diastasis throughout the depth of the joint was captured as well as von Mises stress maps for the materials in the model.

5.6 Results

Diastasis measurements at three depths (dorsal to plantar) in the Lisfranc joint were the only measurements that were captured in both the RBM endobutton simulations and the FEA work for the other three surgical procedures. This was acceptable because joint diastasis is the key indicator of injury and anatomical reduction. Depending on whether the procedure could be modeled as with only RBM (ORIF with endobuttons) or required an FEA (other three surgical procedures) determined what set of measurements were captured.
The diastasis of the Lisfranc joint could be compared across all four surgeries and with a healthy and injured foot (Figure 5.6-1). The negative values in the graph means that the separation between the bones decreased in the partial weight bearing simulation. The additional results such as the stress contours were captured in the FE models but were more indicative of hardware failure or locations susceptible to fatigue than joint mechanics.

![Diastasis Values by Surgical Procedure](image)

Figure 5.6-1: The three Lisfranc joint diastasis measurements for the different surgical repairs of the Lisfranc joint compared with the healthy and injured foot. The Healthy, Injured, and Endobutton models had their data captured from RBM in SolidWorks. ORIF with screws, ORIF with dorsal plates, and arthrodesis results are from FEA in Ansys Workbench for models with threaded screws that are bonded to bone.

Lastly, contact forces with the plantar surface and in several of the adjacent joints were captured for the four different procedures as well as the healthy and injured models. For the plantar plate, the contact forces of the five metatarsals were captured from the rigid body modeling and compared the percent of the total contact force each metatarsal experienced. Figure 5.6-2 shows how the distribution of the plantar contact across the metatarsals changed for the different models. The joint contacts that were collected were in the joints proximal to the injury/fixation (navicular-medial cuneiform and navicular-intermediary cuneiform) and on the lateral portion of the transverse arch (cuboid-lateral cuneiform). Figure 5.6-3 shows how the injury and different surgical procedures change the adjacent joints’ contact forces.
Figure 5.6-2: Distribution of the plantar contact force across the different metatarsals for the healthy, injured, and surgical procedures. Contact forces were captured from the RBM in SolidWorks.

Figure 5.6-3: Joint contact force for navicular-medial cuneiform (Nav-CN1), navicular-intermediary cuneiform (Nav-CN2), and cuboid-lateral cuneiform (Cub-CN3) given a foot is healthy, injured, or received a surgical repair. All contact forces were collected from RBM in SolidWorks.

5.6.1 ORIF Endobutton RBM Results

ORIF with endobuttons has been lauded for allowing patients to return to weight bearing earlier [21]. While this might be true, the procedure only demonstrated limited success in providing mechanical stability in a simulation of immediately after the surgery. The
endobutton was able to reduce the interosseous diastasis seen in the injured model. However, the large decrease in the dorsal diastasis and the large increase in the plantar diastasis show that the bones may rotate relative to each other with the wire acting as the axis of rotation (Figure 5.6-4).

The endobutton repair from the rigid body modeling simulations was the only procedure where all the secondary measurements of the mid foot could be captured, and the results were mixed. The steady state intermetatarsal angle returned closer to the behavior in the healthy model while the transverse arch height and radius continued to deviate from the healthy behavior of the foot.

The stresses in the endobutton wires could be calculated using the forces of the tension only elements in SolidWorks Motion and the cross-sectional areas of the wires. This yielded steady state stresses of 16.2 MPa and 54 MPa for the intercuneiform and Lisfranc endobutton wires, respectively. These values were well within the yield strength of 762 MPa which was calculated from the results of Najibi et al. [88].

### 5.6.2 ORIF Screw FEA Results

The diastasis values in the ORIF with screw procedure showed some magnitude dependence based on whether the screws were modeled with bonded threads or a simplified (cylindrical) screw with rough contacts. The threaded and simplified screws showed the same overall
behavior. The absolute magnitude of the diastasis was larger for each of the three measures with the simplified screws. The dorsal diastasis increased more than the interosseous space compared to the injured model. Meanwhile there was a slight reduction in the plantar diastasis of the joint compared to the healthy model (Figure 5.6-5). Ultimately, the threaded model showed that ORIF with screws had greater restraint in the model’s motion than the endobutton.

The ORIF with screw procedure was a complete success in stabilizing the joint for the threaded screw or a partial success for the simplified model. The threaded screw setup showed all three measurements within 0.7 mm of the healthy model. The simplified model was also in the same range of the healthy model, but the dorsal diastasis actually increased beyond the increase seen in the injured model. This additional diastasis is the main reason the simplified screw simulation is only a partial success in returning healthy stability levels to the joint.

![Diastasis Values for ORIF with Screws Procedure](image)

**Figure 5.6-5:** Diastasis values of the ORIF models with screws. The threaded screws model used screw geometries with threads that were bonded to the bone interfaces. The other two ORIF with screw model runs used a simple cylinder with a rough bone interface contact setting that prevented sliding. The simple screw convergence test had a finer mesh used than the simple screw model.
Unsurprisingly, the stress contours of the fixation screws varied depending on whether threads were present. The maximal von Mises stress for the threaded Lisfranc screw greatly exceeded the yield strength of titanium. However this was a small point on one of the threads. The rest of the model did not show signs of exceeding yield stress. By comparison, the simplified model did not show stress spikes above the yield strength. The peak stresses arose around the screw’s insertion to the second metatarsal. Looking back at the threaded screw, the insertion point of the screw into the second metatarsal also showed a higher stress area in the same location (Figure 5.6-6). Whether the screws had threads or were simplified, the screw spanning the Lisfranc joint still experienced higher von Mises stresses than the intercuneiform screw. The cortical bone and cancellous bone did not show areas of yielding though stresses around the screw holes were high and may be susceptible to fatigue.

The convergence test performed for the simplified screw models was compared with the other simplified screw for Lisfranc joint diastasis measures, location and magnitude of peak stresses and patterns of stress concentration. No major differences in the diastasis values (i.e. less than 0.01 mm in differences) or with the stress contours were noted. Additionally, the peak stresses in the fixation screws were in the same location and were 358 MPa compared to 329 MPa. This gives confidence in the model being a unique solution that is not dependent on the mesh’s approximation of the geometry.
5.6.3 ORIF Dorsal Plate FEA Results

Dorsal plates were not able to provide the same level of stability throughout the full depth of the Lisfranc joint. Unsurprising, the smallest diastasis was in the dorsal aspect of the joint. The diastasis values increased for the deeper portions of the joint. Depending on how the screws were modeled, the interosseous and plantar diastasis values were either at or above the diastasis seen in injured models except for one (plantar diastasis for threaded screw) (Figure 5.6-7).

![Figure 5.6-7](image_url)

Figure 5.6-7: Diastasis values of the ORIF models with screws. The threaded screws model used screw geometries with threads that were bonded to the bone interfaces. The other two ORIF model runs with dorsal plates used a simple cylinder with a rough bone interface contact setting that prevented sliding. The Simple screw convergence test had a finer mesh used than the simple screw model.

The von Mises stress contours on the fixation hardware showed a similar pattern even though there were some high stress point increases in the metal on metal contact. Regardless of the screw geometry used, the screws showed areas of higher stress where the screws entered the dorsal portion of the bone. Specifically, the first metatarsal screw and the medial cuneiform screw showed the higher stress values. The dorsal plate around the hole
for the first metatarsal also showed a region of higher stress (Figure 5.6-8). Maximal stress values were located at the screw-plate interfaces. These interfaces were modeled as bonded which may be the reason for these stress points. Some of the peak stresses on the bonded surfaces are above the yield strength of titanium (827 MPa) but were isolated to small points.

The majority of bone experienced stress below the yielding value. However, the screw insertion point for the first metatarsal showed regions of stresses above yielding strength for both variants of screw geometry.

The convergence test for the simplified screws had similar results so there is confidence that changing the meshes does not change the takeaways of the model.

### 5.6.4 Primary Arthrodesis FEA Results

Regardless of the screw’s geometry, the five screw arthrodesis approach showed very small diastasis. The diastasis values were all less than 0.07 mm for both the threaded and simplified screws (Figure 5.6-9). The arthrodesis clearly reverses the increased interosseous and plantar diastasis from an injury. Unlike the healthy foot, the dorsal diastasis did not reduce, which makes sense since the screw across the Lisfranc joint also resists compression.
Figure 5.6-9: Diastasis values of the Arthrodesis ORIF models. The threaded screws model used screw geometries with threads that were bonded to the bone interfaces. The other two arthrodesis model runs used a simple cylinder with a rough bone interface contact setting that prevented sliding. The simplified screw convergence test had a finer mesh used than the simple screw model.

The five screw arthrodesis procedure showed some regions of high stress concentration where implant failure might be a concern. The screw running through the three cuneiforms experienced the highest stress in all variants of the model. The stress values were near the yield strength of titanium (Figure 5.6-10). The other screws all had respectable safety factors of ~3x below the yield strength. When considering the stress contours of the cortical and trabecular bone, there were only small isolated areas near the yield strength of the material. For the cortical bones, this was in the debrided joints and usually occurred where the cortical thickness had been greatly reduced by the debriding process. The highest regions of stress in the cancellous bone was at the screw/bone interfaces.

The convergence tests for the simplified screw model showed similar results to the model with a less refined mesh. This gives confidence that the results of the model were not dependent on the mesh.
Figure 5.6-10: (Left) Von Mises stress contours of the intercuneiform threaded screws for arthrodesis from a superoposterior angle. (Right) The same stress contours but for a simplified screw model.

5.7 Discussion

This study sought to characterize the stability of the Lisfranc joint and mid foot after a repair surgery and the stresses in the structures that result from finite element analysis. The results showed that all four procedures reversed some portion of the post injury instability, but the biomechanics still had differences from the healthy foot. The study was also able to highlight the concerns that have been raised about the different procedures. For the endobuttons, this is the larger amount of movement that is permitted in the joint. For the procedures with screws and plates, the concerns are that the hardware could fail. ORIF with dorsal plates also has concerns about them being unstable in the plantar portion of the joint.

The less rigid fixation of the Lisfranc joint with the endobuttons is hard to state as an outright negative outcome. From the biomechanics sense, the endobutton having a single tension element holding the medial cuneiform and second metatarsal will not be as stable as with the three native ligaments that support this joint in a healthy foot. The result is that there are some abnormal movements of the bones, like rotation about the bone tunnels for the wire seen in the simulation. This is not to say that the endobutton fails to reduce the injury (i.e. return bones to their anatomical orientation). But once in the original position, the bones of the Lisfranc joint have more freedom to move relative to each other once under
the effects of muscle and body loading.

However, the additional movement in the joint space has been seen as a positive for healing response [21]. The idea is that the additional movement helps promote remodeling of the injured and scarred tissue. Unfortunately, these claims have only been made because of improved patient scores for endobutton procedures where clinicians have noted the larger amount of motion in the joint. The amount of motion before adverse results occur have not been quantified. In terms of return to normal function for endobutton procedures, clinicians have allowed patients to begin weight bearing at earlier time points because there is less concern of the implant failing when compared to screws [21]. These results seem to support that the devices are not near failure when loaded in a plantarflexed foot.

ORIF with screws was the temporary fixation technique that provided the most similar diastasis values to a healthy foot throughout the depth of the joint. Figure 5.6-1 shows that whether the screw is modeled with threads or as a cylinder, the ORIF with the screw brought the injured foot’s diastasis value closer to the healthy model than other ORIF procedures. In the dorsal measurement, all three ORIF increased the diastasis from the injured model but they were all by similarly small amounts. However, the diastasis value for the injured model (and even the healthy model) is negative (i.e. the bones in the injured model come closer together). This behavior in the injured model appears to be because there was no material present to resist additional reduction of the dorsal Lisfranc joint when muscles were active. Meanwhile, the rigid structures in screw and plate fixation also resist this reduction in the dorsal aspect of the joint just as they would resist separation.

The stress contours for the ORIF with screw procedure agreed with hardware failures reported in the literature [21]. Most ORIF screw fixation failures occur in the Lisfranc screw where the screw fracture occurs in the portion spanning the joint space. Regardless of how the screws were modeled, the portion of the screw as it entered the second metatarsal showed areas of increased stress. These regions did not show yielding on this steady state
simulation. But with repetitive loading, fatigue failure could be a concern. For this reason, weight bearing activity should be strictly controlled until ligament or capsular tissue begins to regain organizational alignment and increased thickness. While these are qualitative assessments for the soft tissue, they should help support some of the stress across the joint that can cause screw failures.

ORIF with dorsal plates showed signs of the plantar portion of the Lisfranc joint splaying apart. Similar behaviors and magnitudes of separation have been observed by Bansal et al. [57]. Specifically, Bansal et al. found an increase of ~1.5 mm in the plantar portion of the Lisfranc joint compared to a healthy foot while this study found ~1.8 mm in increase separation. Ultimately, the fact that the plate fixation only spans the joint space in the dorsal aspect of the joint means that even small bending of the dorsal plate can result in much larger separations in the plantar part of the joint. This somewhat mimics the behaviors of the endobutton where the wire running through the midsection allows motion in the dorsal and plantar portions of the joint. Ultimately, these behaviors may contribute to why the ORIF procedure with screws is still the gold standard for temporary fixation [18,23,83–85].

There are several factors that likely contribute to the arthrodesis procedure producing almost no diastasis. The first is the sheer volume of metal that is implanted as part of the procedure. The larger modulus of the screws compared to the bone as well as the physical connection between the different bones resist movement. Additionally, the debriding process before bone fixation removes the low friction cartilage so the bones cannot glide easily across each other (hence using the rough contact setting in the FEA). Besides more force needed to move the bones relative to each other, the debriding process brings larger surface areas in contact with each other. The larger amount of contact area in turn limits the possible motion of the bones.

With five screws spanning five different joints, the implanted screws were being loaded in many different orientations. The stress contours showed the intercuneiform screw experiencing
the highest stress values. In both the threaded and simplified screw models, the highest stress values which were around yield strength occurred where the screws crossed the intercuneiform joint space. The high stress regions suggest that there might need to be a second intercuneiform screw, which is seen in procedures by some researchers and clinicians [94]. Additionally, non-cannulated screws may be able to handle the forces in the intercuneiform joints without the high stress values.

Depending on the surgical procedure, there were changes in the distribution of the plantar contact force across the five metatarsals. When the injury was introduced to the model, contact force distributions changed very little. Even the ORIF with endobutton procedure had very little change in the contact force distribution. However, the rigid fixation procedures with screws and plates greatly increased the percent of the plantar contact that the second metatarsal carried. The larger contact forces that result in these fixation procedures lead to the hardware inserting into the second metatarsal having to carry larger forces. This could play an important factor in the ability of the hardware to survive without breakage or failures.

When joints are fixed in place in the foot, there are higher rates of osteoarthritis (OA) in the adjacent joints [95]. A possible reason for this is because of changes in the joint contact forces after a fixation procedure. When looking at the joints proximal to the surgery location (cuneonavicular joint), there are indications that the medial portion of the joint has increased contact force after a rigid fixation procedure (ORIF with screws/plates or arthrodesis). The lateral portion of the transverse arch (cuboid and lateral cuneiform) does not have large changes in contact force and may after an ORIF procedure (Figure 5.6-3). All of these joint contact results suggest that the medial cuneonavicular joint may have higher risk of developing OA. However, this does not consider the changes that a patient may make to their gait kinematics as a result of fixation or pain from the injury. In reality, the patient may adopt a gait pattern that generates more or less force on these adjacent
joints. Additionally, this only considers the post-operative state. Overtime, soft tissue will heal and in many ORIF procedures, the hardware will be removed which would change the mechanics of the adjacent joints to the Lisfranc joint.

In any computational model that simplifies geometries with a mesh, there are questions of validity of computational results. Part of the confidence in the FEA results comes from the convergence testing done in each case. The convergence tests showed that by changing the mesh that represents the geometries, the trends and takeaways for the simplified screws remained the same. By extension, the models with the threaded screws also showed the same trends as the simplified screws; however, some of the magnitudes differ. So while exact values may have some error or variance like any simulation, the directional trends remain consistent across models. However, if the underlying modeling assumptions (i.e. material properties, contact types, etc.) have errors, there would be concerns the simulations do not reflect reality.

In conclusion, these studies have created a benchmark for the mechanical stability of the different Lisfranc repair surgeries; however, there are some additional considerations that clinicians use to decide on the procedure they perform. Arthrodesis provided the most mechanically stable (or rigid) mid foot and Lisfranc joint. However, this procedure is permanent and will result in fused bones and is not geared at producing an environment for soft tissue healing. Additionally, the rigid fusion of other joints in the foot have been noted to lead to arthritis developing in adjacent joints after fusion [95]. Of the three ORIF procedures, ORIF with screws provided the most mechanically stable post-operative joint.

This model and procedures can hopefully also function as a benchmark for any new or proposed Lisfranc repairs. Since none of the surgeries in this study have received universal approval in the literature or patient ratings [15], healthcare providers are likely to continue trying new surgeries. If this model is used in the future to compare new surgeries, there are already a standard set of measurements and values for the existing procedures that the
new surgery can be compared to. Ultimately, the results of these studies suggest the use of cannulated screws for ORIF procedures to treat Lisfranc injuries, since the screws provide the most mechanically stable environment after the surgery. This should translate to improved healing assuming reduction of the injury is properly performed. Unfortunately this study only looks at the mechanical stability and its role in healing. *In vivo* there are additional factors that determine outcomes of healing. For example this recommendation does not take into account the positive effects of earlier loading of the foot that can be achieved with the endobutton procedures.
Chapter 6: Conclusion and Future Direction

Lisfranc injuries, especially the ligamentous variety, are a challenging orthopaedic injury. They can be hard to identify and when they are identified, surgery is usually required to allow healing and/or to restore stability. Given the patient outcome ratings of the surgeries are subpar compared to other procedures in the foot and ankle (i.e. below 75 for AOFAS), more surgical approaches have continued to be proposed without any consensus in the literature on which is best. The work discussed in this thesis explains how computational models using rigid body modeling (RBM) and finite element analysis (FEA) were used to evaluate the challenges in identifying Lisfranc injuries and in comparing the mechanical stability of four different surgical repairs.

6.1 Work Completed

In this study a computational model was created for a healthy foot and ankle based on the CT scan of a healthy individual. Bones, ligaments, and muscles were all modeled based on anatomical data. RBM was used to validate that the forward dynamics of the model matched observed results in a cadaveric study of the foot with and without a Lisfranc injury. After the validation was completed, variations in muscle activity and foot position were evaluated to determine their impacts on the presentation of Lisfranc injuries. The hope was to identify behaviors or factors that could make the identification of Lisfranc injuries more challenging. All these studies were completed with RBM where the bones other than the tibia, fibula, talus, and calcaneus had 6 degrees of freedom (DOF). Therefore, bones were constrained based on contact with other bones and forces generated by the ligaments and muscles in the foot.

The model was then extended to compare the results of different surgeries to repair the
Lisfranc joint. The aim of this work was to help identify the mechanical stability of the Lisfranc joint after the surgery. Since screws and other material like plates formed a rigid connection between the bones, RBM could not be used. In order to capture the deformation of bones and screws which allowed for diastasis in the Lisfranc joint after surgery, FEA had to be used in some procedures where RBM would not allow any movement of bones relative to each other (and therefore would have shown no diastasis). Additionally, FEA can access the stresses in the different implanted hardware. This can identify where the device is most likely to fatigue or fail.

The comparisons between the four surgical procedures was the first study that we are aware of to compare multiple procedures for Lisfranc surgical repair. Other comparisons in the literature horse-race two procedures but experimental setups and metrics vary from one study to the next [18, 19, 21, 23, 25, 85]. With a single experimental design and a consistent set of metrics, comparison on the mechanical stability of the Lisfranc joint after surgery were made across four procedures in this work.

6.2 Key Takeaways

The work associated with this thesis accomplished several things. First it validated the ability of computational models to represent the biomechanics of the foot and ankle for both healthy and injured cases. Secondly, it identified potential factors that contribute to the challenges with identifying Lisfranc injuries in the patient population. Finally, it compared four different repair surgeries and reported on the ability of the surgeries to return the Lisfranc joint to a healthy position when loaded. The results also gave insight into points of risk for different hardware implanted in the Lisfranc repair surgeries.

Computational models have been used to simulate cadaveric biomechanics of the foot and ankle in previous studies [6,7,11]. Using similar techniques, a computational model was able to be validated on Lisfranc injuries observed in a cadaveric study. Additionally, the
model was able to point to possible shortcomings of cadaveric studies that do not simulate muscle forces. The anatomy of bones and the passive forces of the ligaments/soft tissue were modeled using SolidWorks Motion. When the healthy and injured versions of the model were loaded in 30° plantarflexion, the diastasis measurements of the Lisfranc joint and the intermetatarsal diastasis measurements were within the standard deviation of Bansal et al. [57]. When muscle forces were added to the 30° plantarflexed model, injury measurements reduced. This suggested that cadaveric models without muscle simulation may overstate the observable instability of ligamentous Lisfranc injuries.

The validated model allowed for additional studies into the factors that could obfuscate key risk indicators of an injury when evaluating a patient with a weight bearing radiograph. The two factors considered were the presence of muscle contraction and the inversion/eversion of the foot. The results suggested that both muscle contraction and positioning the foot with inversion or eversion from a neutral stance would reduce the diastasis observed in the Lisfranc joint. Thus a clinician would see the largest diastasis amount throughout the full depth of the injured joint (dorsal to plantar) if the patient had no muscle activation and their foot was placed in a neutral position. Regional anesthesia might address both of these challenges. Without the pain of the injury, patients could be more willing to place their foot in a neutral position instead of trying to offload the injury by putting the foot in inversion. Additionally, the anesthesia can prevent muscle activation and their impacts on the presentation of the injury though it is an additional intervention beyond that required for radiology.

When modeling the four different repair surgeries, strengths and shortcomings of each procedures were identified. ORIF with screws and primary arthrodesis showed the most similar performance to a healthy joint when loaded. However, high stress concentrations were found in the implanted screws of each set up. In the ORIF with screws, this was in the joint space for the screw spanning from the medial cuneiform to the second metatarsal. With arthrodesis, the highest stress was on the screw passing through the cuneiforms and
occurred in the intercuneiform joint space. While \( \sim 5\% \) below the yield strength of the titanium screws, the high stress values may make the screw susceptible to fatigue failure. By comparison, ORIF with endobuttons offered less stability in the Lisfranc joint, but the lack of rigid hardware has been reported to help allow earlier weight bearing activities since instability and implant failure have not been seen with repeated loading in cadaver studies [21]. However, the surgeon must decide whether up to 1 mm in motion is an acceptable outcome for endobutton fixation procedures. ORIF with dorsal plates was less stable in the plantar portion of the Lisfranc joint (i.e. showed larger diastasis values); however, the setup does not require disruption of the articular surfaces like the other three procedure. Each of the other three procedures have hardware or holes drilled through the subchondral bone and articular cartilage.

### 6.3 Limitations and Shortcomings

As most computational models usually do, assumptions were made to either simplify the simulation or to overcome limitations in data. While these assumptions were reviewed and deemed reasonable, they still introduce uncertainty and cause limitations in what the model can be used to predict.

One assumption that was taken from past validated foot and ankle models was that similar sized ligament structures had similar stiffness values. While many of the ligament structures had stiffness values from literature and previous material testing, the number of small ligament structures in the mid and forefoot meant that every single ligament did not have a reported value. Instead similar stiffness values were used for ligaments that have similar dimensions to known ligament stiffness values. Since this technique of interpolating stiffness values from other structures had been used in other validated foot and ankle models, these assumptions were reasonable.

Since the model creation to represent muscle and ligaments is currently very manual,
only one set of patient anatomy was tested. While this was a healthy anatomy, it still may have had some unique geometries of the bone that impact the biomechanics of the studies. No gross deformations or abnormalities were noted in the Lisfranc joint but the addition of models for other healthy anatomies could improve the confidence in some of the behaviors observed. The MCO that was previously performed on the calcaneus was reversed and could have some impacts on the kinematics of the weight bearing radiograph simulations. However, all other studies had the calcaneus remain locked relative to the talus and tibia/fibula.

There were additional shortcomings for the study on the impact muscle contraction and foot position have on the presentation of Lisfranc injuries for weight bearing radiographs. Namely, foot positions in neutral, $10^\circ$ inversion, and $10^\circ$ eversion were the only positions tested. Patients may use other alterations to their stance in order to reduce the pain of a weight bearing radiograph when they have a Lisfranc injury. Another challenge was that the muscle forces did not change depending on foot position. This is not likely to occur in vivo. However, the lack of data to ground how the muscle force changes with these small variation in positioning led to the same forces being used in all three positions.

Limitations in the study comparing surgical procedures had similar limitations to the RBM work as discussed above (i.e. uncertainty on ligament material properties and use of a single anatomy). However, additional limitations came from FEA simplifications of the model and software limits. One simplification for the FEA was the geometry of the screws. The tip of the screw was not modeled exactly the same as the screws used in surgery. This was because the complex geometry could not be manually measured and would have required a scan or specification sheet with dimensions which were not provided by the manufacturer. While this limitation was a design decision, there were some limitations with the software used. The largest limitation was that Ansys Mechanical limited FE meshes to 200,000 elements for the student edition of the software. While this did limit the resolution of the mesh in some model runs that used threaded screws, the fact that the simplified screw models
and convergence tests provided the same takeaway makes this limitation less concerning. The largest challenge with the FEA work was that the models with threaded screws could only be solved when the contacts between the screws and bone were treated as bonded. Friction, frictionless, and rough contact settings all failed to converge unless a simple cylinder screw was used. In reality, modeling screw contact as frictional may be a more accurate option if the solver could converge.

6.4 Future Direction

There are three main areas where this work could continue. The first is to continue similar studies as those presented here to better characterize relationships between Lisfranc injuries and their clinical presentation. Next the model building and generation could be streamlined to allow for faster generation of patient specific models that could offer clinical support for surgeons. Lastly, the model could be extended to new studies of the foot and ankle for Lisfranc injuries. This could be extension of the model to evaluate new procedures and hardware for implants or modeling a dynamic action such as gait.

As noted in the limitations of this work, when evaluating the impact of foot position on the injury presentation, single positions of inversion or eversion were used. Future work could better characterize how the range of foot positions impact the injury that is visible to the clinician. Additional work in the same vein as what has been presented would be to model other potential surgical repairs for the Lisfranc joint. There are other dorsal plate designs and alternative screw positions that could be evaluated and compared to the results in this thesis. And since the existing surgeries have continued to have subpar results, there is a possibility that surgeons will continue to explore new fixation options for Lisfranc injuries. If this occurs, the model could be used to benchmark the latest proposed surgeries against those already used.

Two areas of additional work fit together quite well. First would be reducing the
time required to create these models. This could involve automating some of the steps of identifying muscle and ligament insertion and origin points. If models can be quickly generated using some of the automated processes of non-rigid registry and other algorithms, then the second area of future work naturally flows. This would be creating patient specific models in a clinical environment to be used for pre-operative planning. Even further, the models could be used to evaluate the expected outcomes for different surgical repairs. The speed of delivery and the streamlining of the model could help augment surgeons’ decision making process and would rely on procedures and methods outlined in this thesis.

Other lines of work that are separate from what has been presented here include the use of computational models in studying dynamic motions. For example, modeling the Lisfranc joint after surgery during the gait cycle could help inform on how implants fail and whether the joint is stable during the entire cycle. The largest challenge with using forward dynamics for these dynamic simulations would be determining the forces through different time points. Past work in the Orthopaedic Research Laboratory has been able to accomplish this with iterative testing since extracting specific muscle forces is challenging to capture in vivo [96]. Instead, an inverse dynamics approach like those used in OpenSim may be required to take motion tracking data for the foot and then calculate the forces in joints and the implanted hardware.

Regardless of future directions of this work, the existing work provides useful information regarding the challenges of identifying Lisfranc injuries and the outcomes of four different surgical procedures.
References


[75] Byungjoo Noh, Takeo Ishii, Akihiko Masunari, Yuhei Harada, and Shumpei Miyakawa. Muscle activation of plantar flexors in response to different strike patterns during


Appendix A: Ligament Properties
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**Appendix A:** Table of the properties of the soft tissue elements used in the creating the model representations of the ligaments and other soft tissue structures. Lo is the undeformed length given all model measurements based on CT positions assume a 3% in situ strain.
Appendix B: FEA Muscle, Ligament, and Contact Forces
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Forces extracted from the kinematic model of the ORIF with screw procedure that was then added to the FEA of the procedure.
### Dorsal Plate FEA Forces

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Forces extracted from the kinematic model of the ORIF with a dorsal plate procedure that was then added to the FEA of the procedure.
### Arthrodesis FEA Forces

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Forces extracted from the kinematic model of the arthrodesis procedure that was then added to the FEA of the procedure
Vita

M. Tyler Perez was born May 15th, 1991 in Fairfax, Virginia and grew up in Manassas, Virginia. He graduated Osbourn High School in 2010 as salutatorian. Tyler attended Washington University in St. Louis to study biomedical engineering. Tyler first began working in research with Dr. Spencer Lake’s Musculoskeletal Soft Tissue Laboratory. As an undergraduate student, Tyler was a member of the Men’s Swim Team and was a captain his senior year. He was also a board member of the First Year Center where he organized orientation and other programs geared towards introducing students to the wider St. Louis community. He graduated cum laude in 2014 with a Bachelor of Science. From July 2014 to August 2017, Tyler worked as a business analyst for Capital One Financial and volunteered as an emergency medical technician in Arlington, Virginia. In Fall 2017, Tyler returned to studying biomedical engineering as a master student at Virginia Commonwealth University. He joined the Orthopaedic Research Laboratory under Dr. Jennifer S. Wayne to focus on foot orthopaedic surgeries. Following graduation, Tyler will be moving to Boston to work for Simon-Kucher & Partners as a consultant for their life science and medical device clients.