

**Development and Implementation of a Forward Solution Hip
Mathematical Model to Determine In Vivo Mechanics and
Predict Hip Separation**

A Dissertation Presented for the
Doctor of Philosophy
Degree

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DEDICATION

This dissertation is dedicated to my parents, Ann and Bruce, for their continual support throughout my life. From the very beginning, I knew that I was loved and supported, no matter what direction I chose, and I promise to pay that forward to my children. The confidence of knowing that they always had my back, both physically and mentally, allowed me to take risks in life that I would otherwise not have been able to do, and because of that I am graduating with my PhD. I am extremely excited for the path that lies ahead of me. Thank you for everything, Mom and Dad.

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ABSTRACT

Osteoarthritis, signified by excessive wear of joint cartilage, is most common in hip, knee, and spinal joints [1]. Current treatment for extreme osteoarthritis is total joint arthroplasty, where the damaged bone and cartilage are removed and replaced by artificial bone representations separated by artificial cartilage representations. While this treatment does generally improve patient quality of life, current joint replacement systems unfortunately still yield atypically high forces, premature component wear, and abnormal kinematics compared to native joints [2] [3] [4] [5].

For total hip arthroplasty, one common complication is in vivo separation and dislocation of the femoral head within the acetabular cup [6] [7]. Determining a successful solution to this issue revolves around developing progressive new implant designs, establishing the least destructive surgical methods, and determining and executing ideal intraoperative component alignments. However, ethical issues and extravagant expenses prevent surgeons and implant companies from experimenting with unknown or risky concepts. Fortunately, an alternative approach of developing and utilizing mathematical models is available, providing a solution to these issues. A forward dynamics mathematical model of the hip allows users to virtually insert a hip implant into a hypothetical patient and observe the results. This will allow design companies to implant new, innovative designs or incorporate untested surgical procedures on the “patient” without the risk of harm or failure, ultimately progressing towards a solution to eliminate postoperative hip instability.

The objectives of this dissertation are to develop a fully functional forward solution mathematical model of the hip that allows for a comparison between various implant designs and a determination of factors leading to in vivo hip separation, instability, and edge loading. Specifically, this includes: development of a forward solution mathematical model of the hip that incorporates detailed articulating surface geometry and patient-specific kinematics; customization of the model to accept multiple implant designs and predict occurrences of hip separation, instability, and edge-loading; implementation of the model to conduct detailed simulations to compare various implant designs and determine ideal component alignment zones; and design of a graphical user interface to merge the model with intraoperative alignment tools, yielding intraoperative predictions of postoperative mechanics and stability.

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LIST OF ATTACHMENTS

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CHAPTER 1: INTRODUCTION

1.1 Anatomy of the Hip

This dissertation investigates the in vivo mechanics and stability of the hip joint and the surrounding soft tissues. Hence, it is important to have a fundamental knowledge of hip joint anatomy to fully understand the development of the mathematical model and the relevance of the results that are discussed.

1.1.1 Joint Motion

Before discussing the hip joint anatomy, it is important to understand the basic motions of the hip joint. The body is divided into three planes – the coronal (frontal), the sagittal, and the transverse (horizontal) plane – and contains three corresponding axes – the superior/inferior (S/I) axis, the medial/lateral (M/L) axis, and the anterior/posterior (A/P) axis – all shown in Figure 1. Hip joint rotation about the sagittal plane, shown in Figure 2a, is known as flexion/extension of the hip. Hip joint rotation about the coronal plane, shown in Figure 2b, is known as abduction/adduction of the hip. Hip joint rotation about the transverse plane, shown in Figure 2c, is known as internal/external rotation [8].

1.1.2 Bones

The hip joint serves as the articulating surface between the femur and the pelvis bones. The femur bone is the longest and strongest bone of the human body and is shown in Figure 3. The proximal portion of this bone, the femoral head, serves as the articulating portion of the bone and sits within the acetabulum. The femoral neck connects the femoral

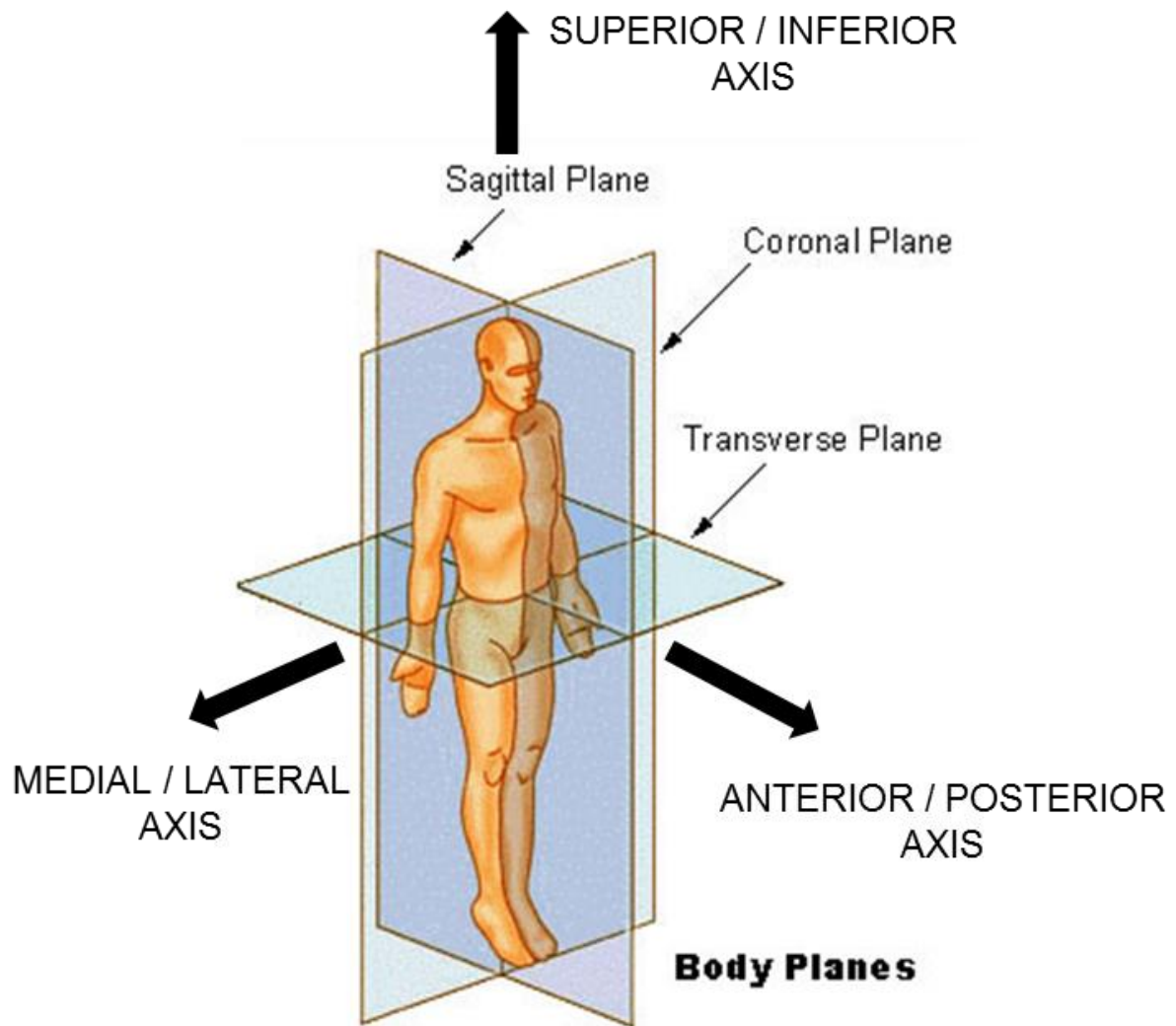


Figure 1: Reference planes and axes for the body. Image modified from (athletics.wikia.com).

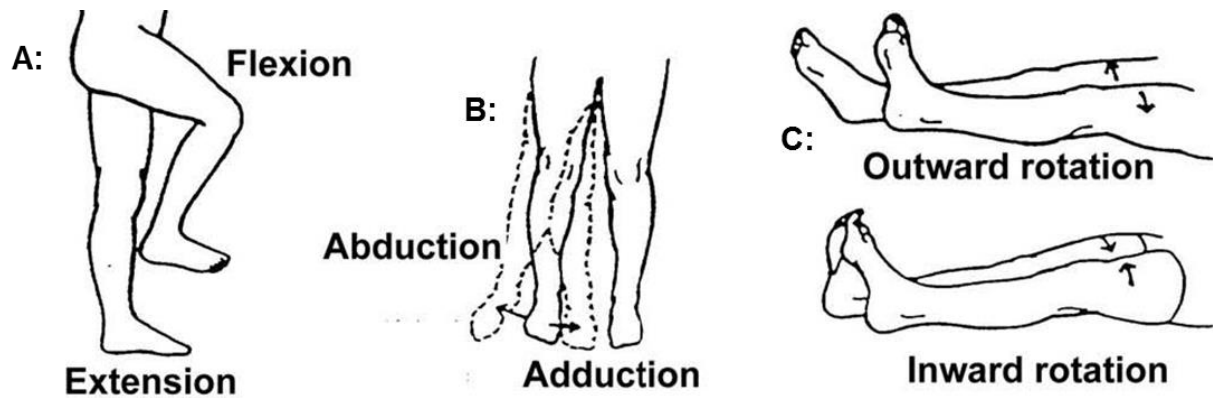


Figure 2: (a) sagittal plane rotation, showing flexion and extension directions about the hip joint; (b) coronal plane rotation, showing abduction and adduction directions about the hip joint; (c) transverse plane rotation, showing internal and external rotation directions about the hip joint. Image from (b-reddy.org).

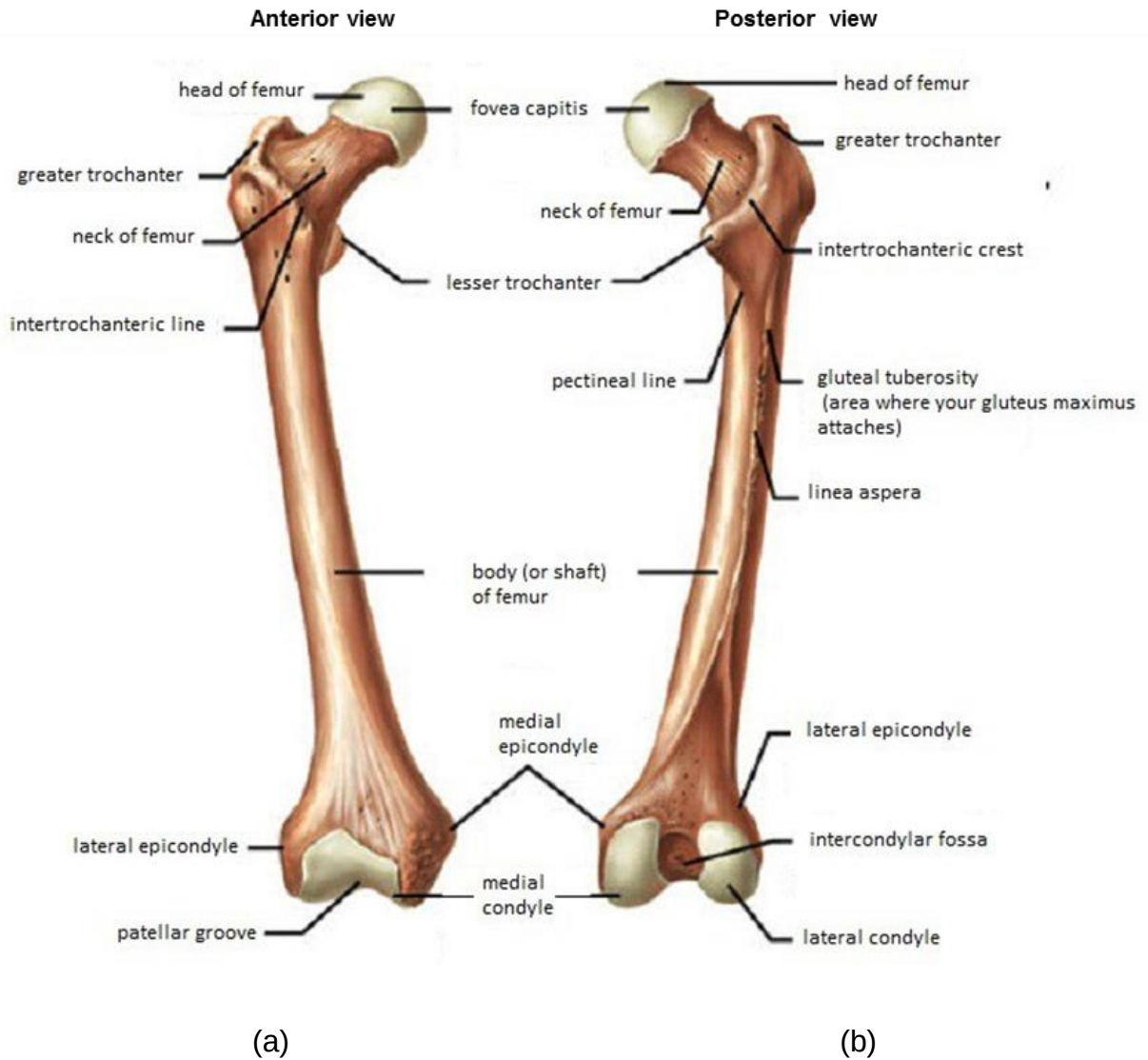


Figure 3: Femur bone schematic showing anatomical landmarks from (a) an anterior view and (b) a posterior view. Image from (boneandspine.com).

head to the long shaft of the femur. The femoral shaft serves as the primary weight-bearing portion of this bone and contains many bony landmarks, including the greater and lesser trochanter, that serve as attachment sites for various muscles and ligaments [8].

The pelvis is a combination of three bones: the ilium, the ischium, and the pubis, shown from a lateral view in Figure 4 and from an anterior view in Figure 5. These bones, along with the sacrum, together form what is conventionally called the pelvis, and the muscles and ligaments of the hip joint attach to one of these four bones. The junction of the ilium, ischium, and pubis forms the acetabulum, or acetabular cup, that serves as the mating articulating surface for the femoral head [8].

1.1.3 Ligaments

The primary ligamentous structures that surround the hip joint are together known as the hip capsule. On the anterior side, this includes the iliofemoral ligament and the pubofemoral ligament, shown in Figure 6a. On the posterior side, this includes the ischiofemoral ligament, shown in Figure 6b. In addition, there is an internal ligament present within the acetabulum, known as the ligamentum teres, shown in Figure 7. The primary purpose of the hip capsule is to maintain tension at the hip joint. The spiraling nature of the capsule yields complex profiles of tension in various joint positions. Simply put, in full extension, the hip capsule essentially “spirals” around the femoral head, yielding a higher amount of joint tension. When the hip is flexed, many (but not all) of the ligament fibers relax, but the capsule overall remains in tension [8].

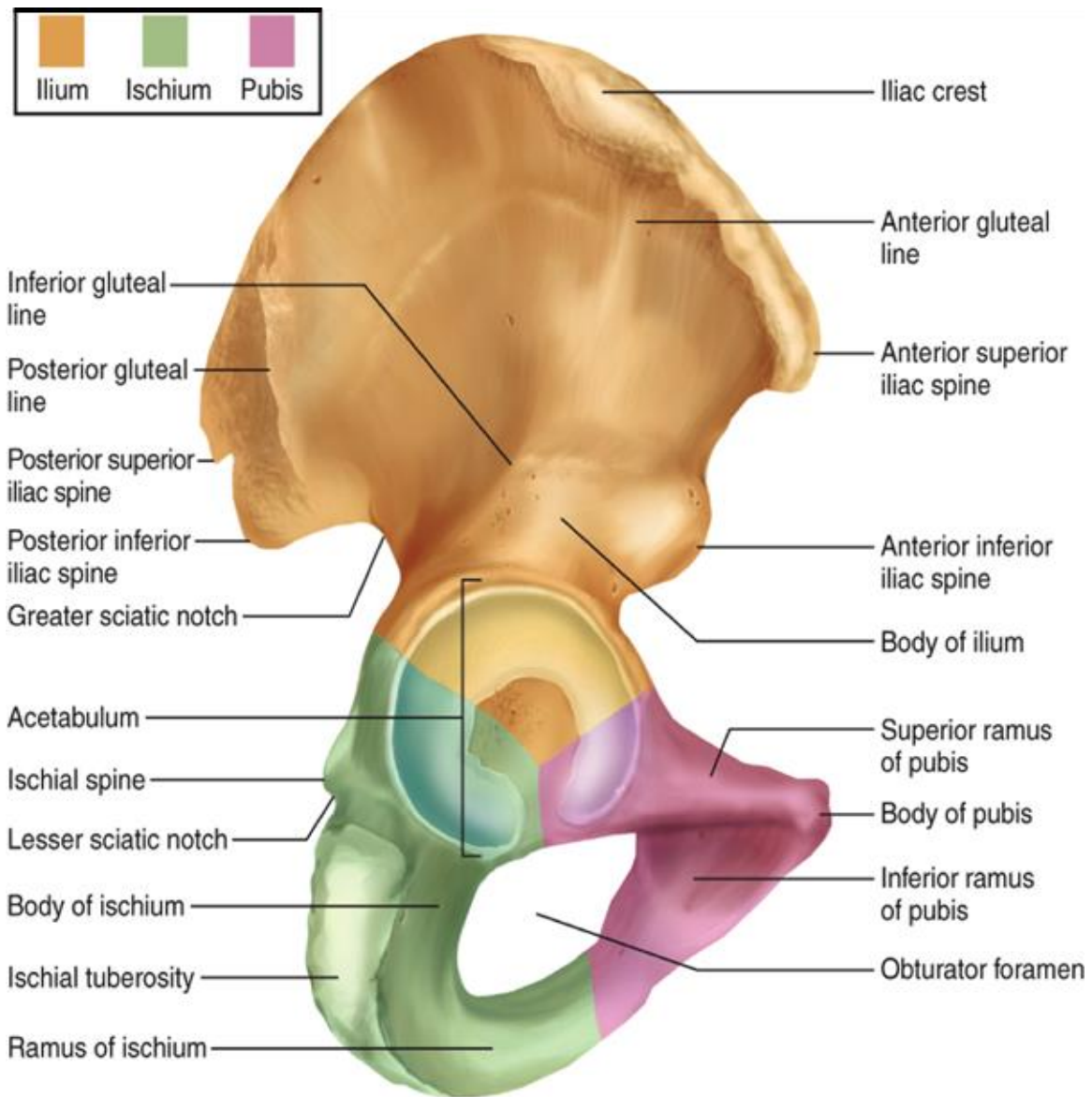


Figure 4: Pelvis bone schematic showing anatomical landmarks from a lateral view.

Image from (studyblue.com).

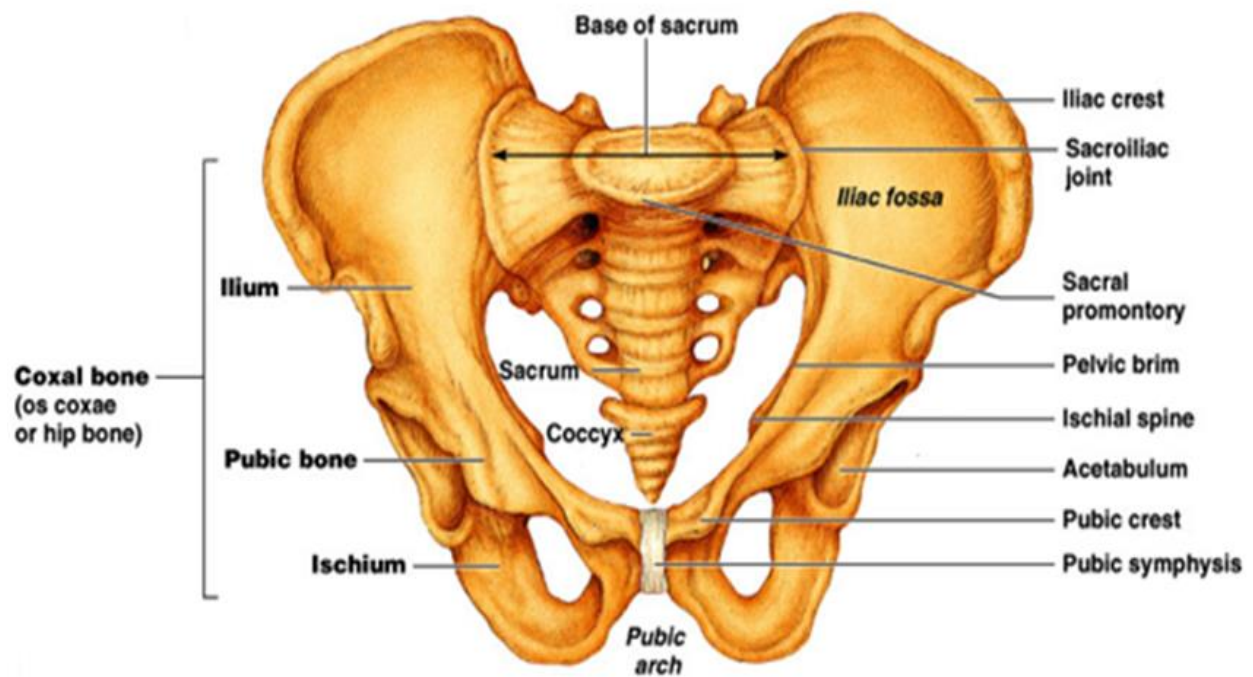


Figure 5: Pelvis bone schematic showing anatomical landmarks from an anterior view.

Image from (wadsworthosteopaths.co.uk).

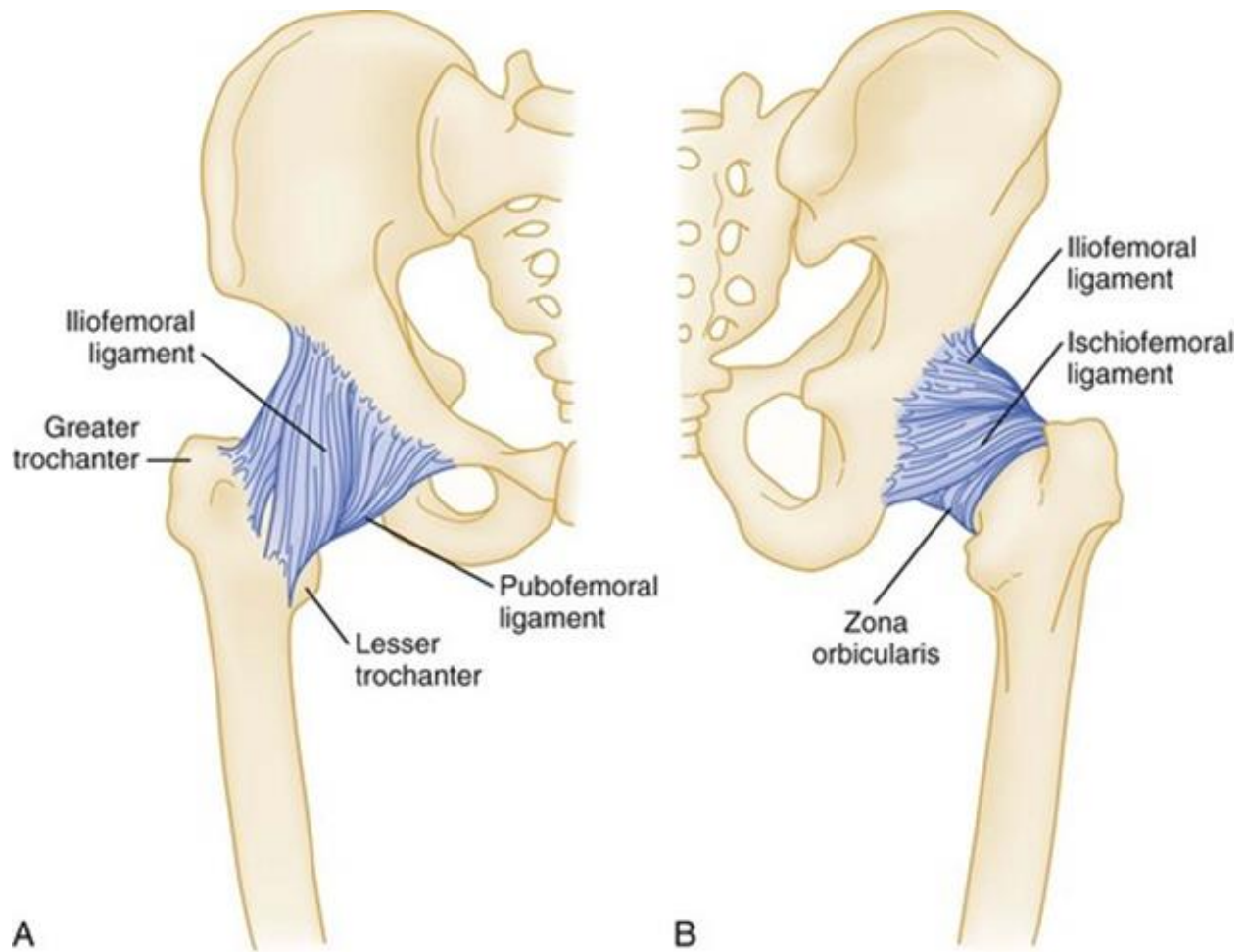


Figure 6: (a) anterior capsule, showing the iliofemoral and the pubofemoral ligaments, and (b) posterior capsule, showing the ischiofemoral ligament. Image from (clinicalgate.com).

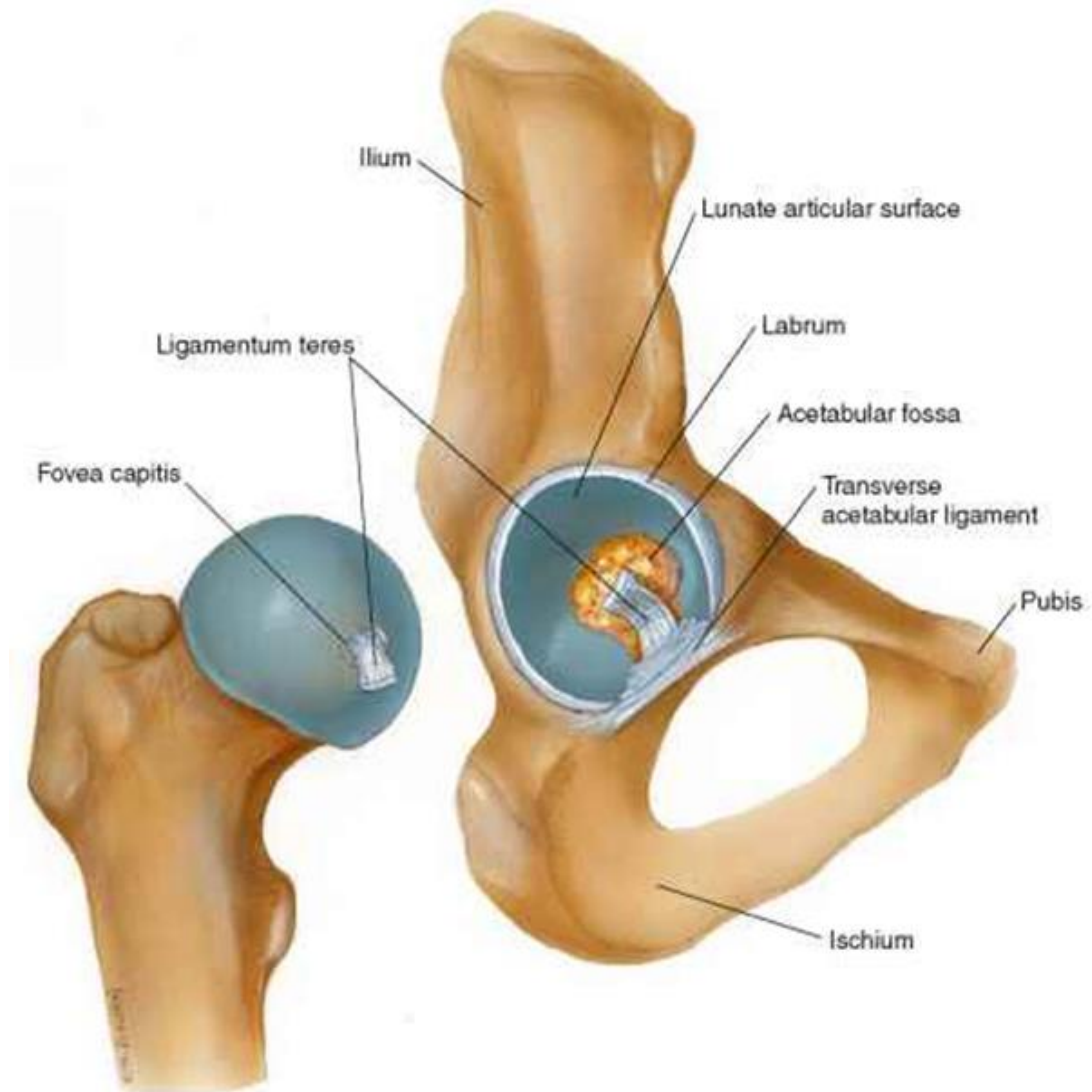


Figure 7: View of the acetabulum, showing the ligamentum teres. Image from (alpmedical.info).

1.1.4 Muscles

The muscles acting across the hip joint work together to collectively control the six degrees of freedom presented in Figure 2. An overview of the anterior muscles of the hip joint can be found in Figure 8a and a schematic of the posterior muscles of the hip joint can be found in Figure 8b. There are over 20 muscles that surround the hip, so the function of each individual muscle will not be discussed here. In general, the primary function of each muscle is based on the line of action of the muscle with respect to the hip joint. In other words, muscles which attach anteriorly to the hip joint may serve the purpose of generating joint flexion. However, because there are so many muscles at the hip, muscular control becomes very complex. Many muscles control multiple directions of rotation at once, some muscles are larger and are the primary actuators of the joint, and other muscles are much smaller and mainly account for joint stability. In general, muscles with larger physiological cross-sectional areas are capable of yielding larger forces [8]. Similarly, muscles with longer moment arms (i.e. more superficial muscles) can yield larger torques. Hence, a large, superficial muscle such as the gluteus maximus is very effective at controlling hip rotation, while a smaller muscle such as the inferior gemellus is more effective as a subtle stabilizer. The interaction between the stabilizing muscles and actuating muscles is very intricate. For mathematical modeling purposes, however, it is not necessary to model every single muscle at the hip; instead, only the larger actuation muscles are modeled. The specific function of each muscle used in this model will be discussed later.

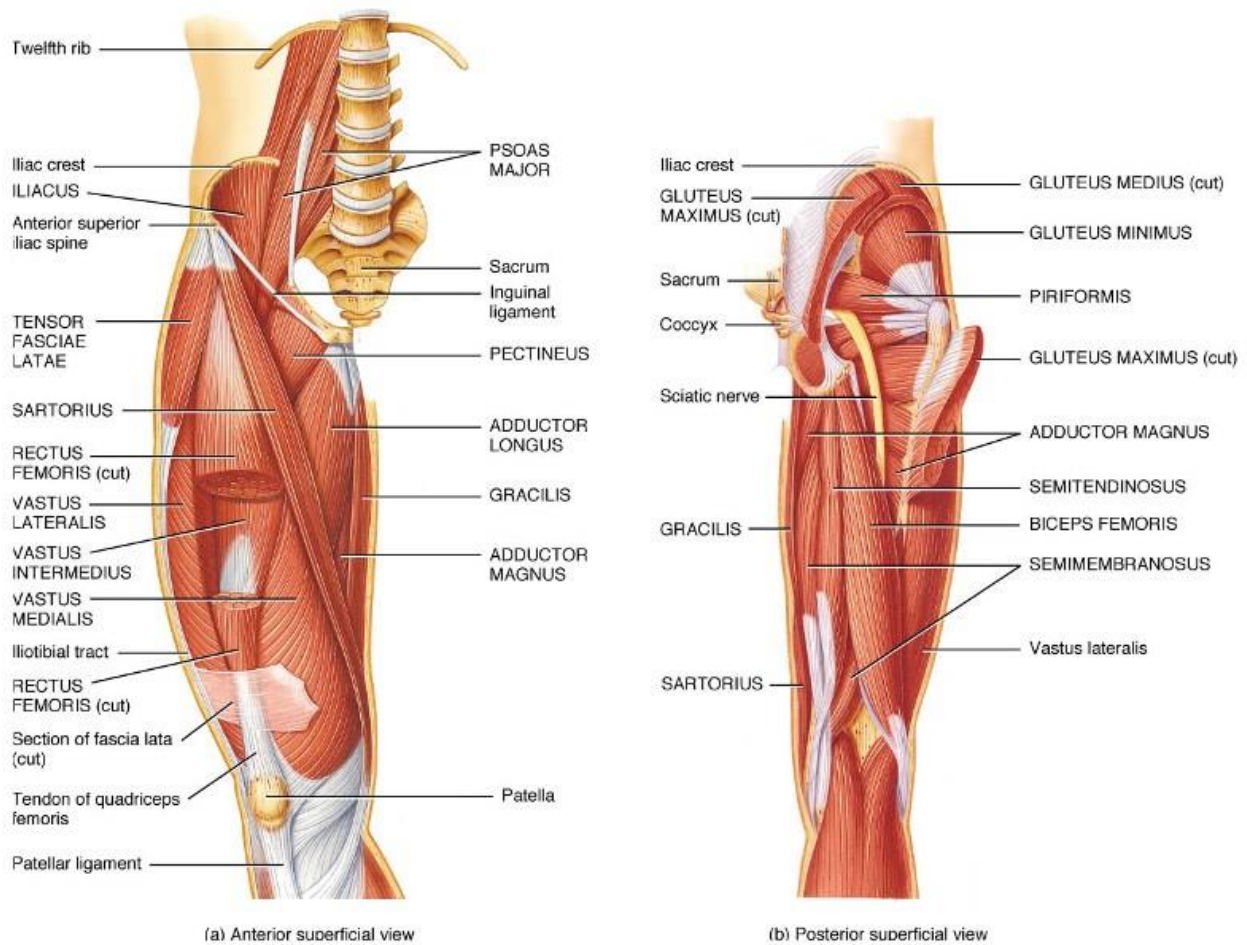


Figure 8: (a) anterior and (b) posterior view of selected hip muscles. Image from (boneandspine.com).

1.2 Total Hip Arthroplasty

Osteoarthritis, the most common type of arthritis, is signified by excessive wear of joint cartilage. It is most common in hip, knee, and spinal joints, and over time the chances of developing this disease increase [1]. When initial treatment attempts are not sufficient, the most successful fallback treatments are total joint arthroplasties, where the damaged bone and cartilage are removed from the joint and replaced by artificial bone representations (generally metal or ceramic) separated by artificial cartilage representations (generally metal, ceramic, or plastic). The portion of the implant that replaces the acetabulum is known as the acetabular cup. Depending on implant type, the articulating surface within the cup (replacing the cartilage) is known as the liner. The portion of the implant that replaces the femoral head is a combination of two components, the stem and the head [9]. Figure 9 shows the individualized components of a standard total hip arthroplasty (THA) and Figure 10 shows a before and after representation of a THA. THA is one of the most common surgical operations performed in the world today [10].

1.2.1 THA Complications: Instability

While the relief provided to patients from these treatments is significantly more desirable than living with osteoarthritis, these artificial joints still fail to fully reproduce the kinematics associated with native joints [2] [3] [11]. Due to these abnormal kinematics, forces present at the joint interface have been shown to be higher than desired [4], leading to premature component wear and/or joint dislocations. Relating specifically to THA, the most common form of complication is separation and dislocation of the femoral head

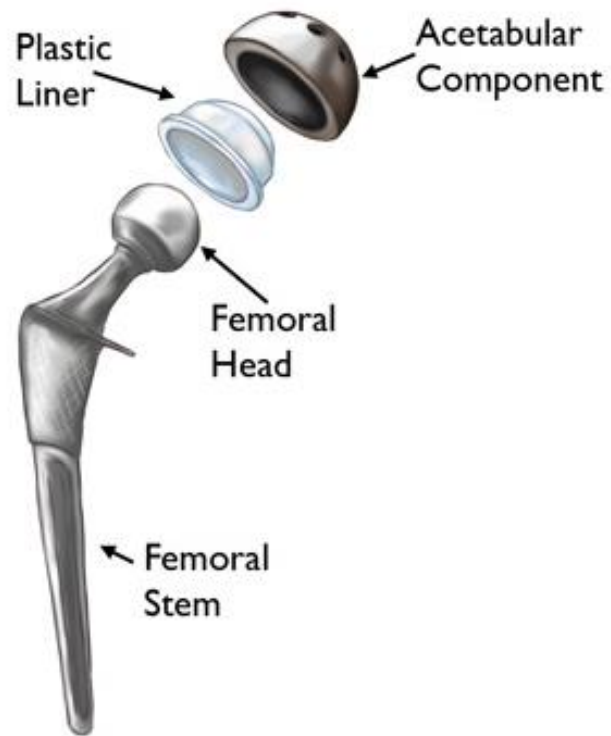


Figure 9: Elements of a total hip replacement. Image from [9].

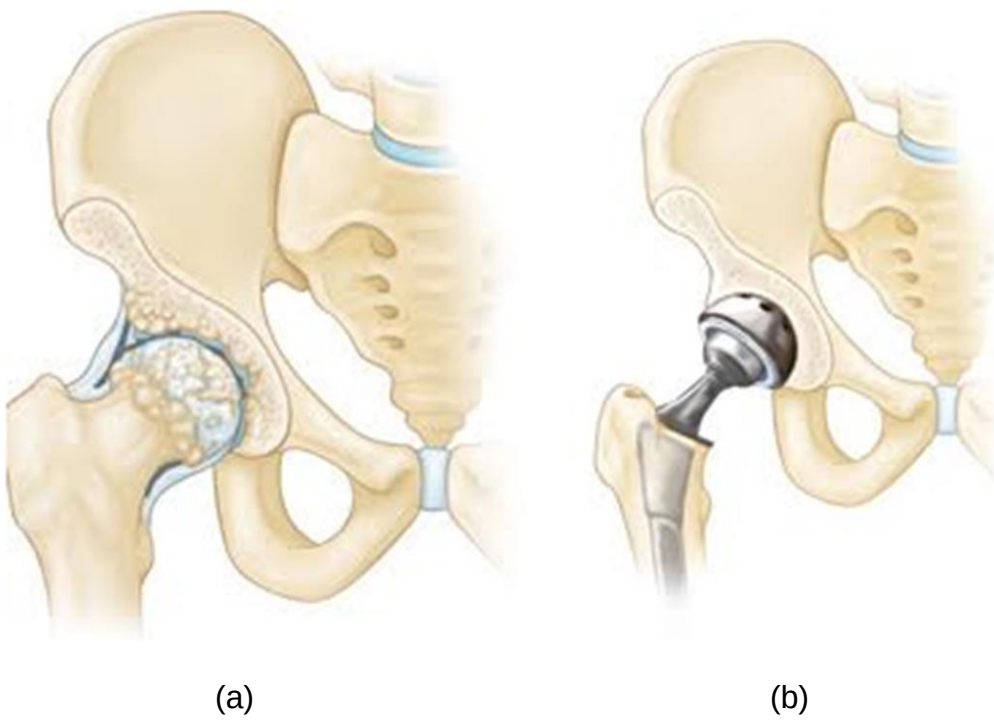


Figure 10: (a) an osteoarthritic hip and (b) a replaced hip. Image from [9].

within the acetabular cup [6] [7] [11] [12] [13] [14]. In a normal hip, the ligamentum teres and the hip capsule function to keep the femoral head properly seated within the cup. During implantation, however, the ligamentum teres must be removed and the hip capsule must be cut. Hence, the risk of hip instability and separation increases. Additionally, variations in component positioning can lead to undesirable postoperative mechanics, including pistoning, microseparation, edge loading, and dislocation. Complete dislocation of the hip joint after THA is a serious complication that often requires revision [7] [12], while smaller microseparation complications are not as severe but commonly lead to edge loading scenarios, decreased articulating contact areas, undesirable wear on the components, and impulse loads [5] [6] [11] [15] [13] [14] [16]. Hip separation can be seen in vivo in Figure 11, and it has been shown that hip pistoning and separation most commonly occur in gait and adduction/abduction activities [6] [11].

1.2.2 THA Analysis

A continuing goal of implant design companies is to develop implants that accurately mimic native joint kinematics and avoid these irregular complications, thereby allowing patients to feel free and unrestricted by the limitations of their artificial joint. Current methods for implant improvement revolve around cadaveric rigs and telemetric implants that can be used to predict the in vivo forces, as well as wear simulators that can be used to predict the stresses and fatigue life of the implants. Additionally, orthopaedic surgeons strive to incorporate the most non-destructive surgical methods possible, thereby allowing patients to retain maximum postoperative soft tissue functionality. Regrettably, due to the expenses required to manufacture and test new implants, the time

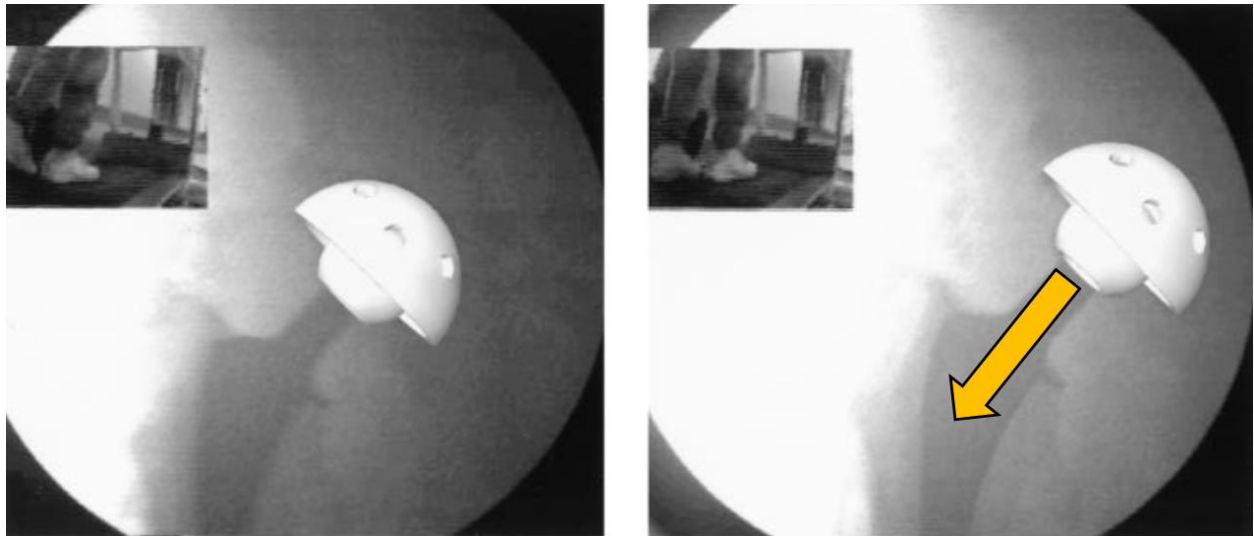


Figure 11: In vivo hip separation, seen in fluoroscopy. Image modified from [6].

associated with proper wear/fatigue analyses, and the ethics associated with unknown or risky designs or surgical techniques, it is difficult to efficiently improve upon existing designs and to discover new ones.

1.3 Mathematical Modeling

Fortunately, mathematical modeling provides an alternative solution to these problems. Mathematical modeling provides users with methods to predict in vivo forces and motions without invasive procedures or expensive equipment. In fact, it is a cheaper, faster way to analyze in vivo joint mechanics because it does not require actual part manufacturing to operate. Additionally, it provides feedback from the simulation in a brief period, generally on the order of minutes, thereby allowing for faster design iterations [17].

Many different methods of mathematical modeling currently exist, including forward, inverse, rigid body, finite element, and more. These will be discussed in more detail in the following section. Most importantly, however, it is necessary to understand that each method of mathematical modeling is accompanied with its own set of benefits, assumptions, and limitations. Thus, it is essential to carefully weigh the assumptions and limitations associated with each modeling technique before selecting the desired method.

The model developed herein is a forward solution mathematical model. Much like the human body, forward solution models accept muscle forces as inputs to the system and provide motions as outputs. Thus, a forward model of the hip will provide accurate representations of in vivo component kinematics and joint stability during desired activities, which are commonly of interest to implant design companies. Additionally, the model developed herein is a rigid body reduction model, meaning all bodies in the system

are assumed to be perfectly rigid (i.e. no deformation), and the system is simplified to avoid complex optimization algorithms. These techniques have been employed to decrease the time it takes to complete a simulation, which is an aspect that will be strongly desired in later stages of this dissertation.

In general, mathematical modeling is an excellent tool that provides users with methods to predict in vivo forces and motions without invasive procedures or expensive equipment. Numerous mathematical models of both knees and hips have been developed to date [4] [16] [18] [19] [20] [21] [22] [23] [24] [25], and the benefits that have arisen from these models is continually being used to improve total hip replacements.

CHAPTER 2: BACKGROUND

2.1 Component Designs

To address various pre- and postoperative complication that arise in THA, there is a vast selection of component designs available on the market. While a detailed explanation of every available commercial design is beyond the scope of this dissertation, the primary purpose and applications of several common designs are briefly discussed below. It is important to note that, to give surgeons more intraoperative control over the surgery, many of the following options can be combined to yield unique scenarios individualized for each patient.

2.1.1 Standard THAs

A schematic of the standard THA can be seen previously in Figure 9. In a standard scenario, the acetabular cup and liner are half-spheres, designed to mate with a spherical femoral head. The surgeon places the cup within the acetabulum and reams the femoral canal to accept the stem. To give the surgeon more versatility, the liners and stems come in several options, including lateralized liners, lipped liners, high offset stems, etc. High offset stems, shown in Figure 12, function to shift the joint rotation center medially but must be accompanied by appropriate acetabular component translations to maintain adequate joint tension [26] [27]. Lateralized liners, shown in Figure 13, can also be used to medialize the cup and increase ligament tension. Lipped liners, shown in Figure 14, provide additional joint stability by incorporating an elevated portion of the rim to prevent dislocation in a specific direction [28] [29]. However, this stability can come at the added

Standard / High Offset Stems

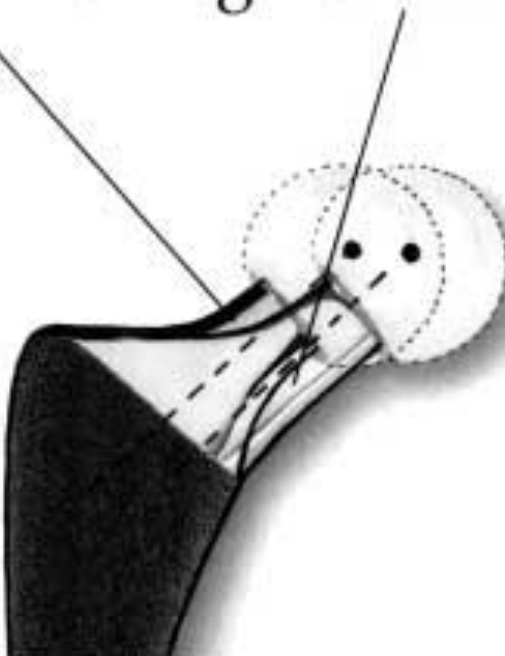


Figure 12: High offset femoral stem. Image from [26].

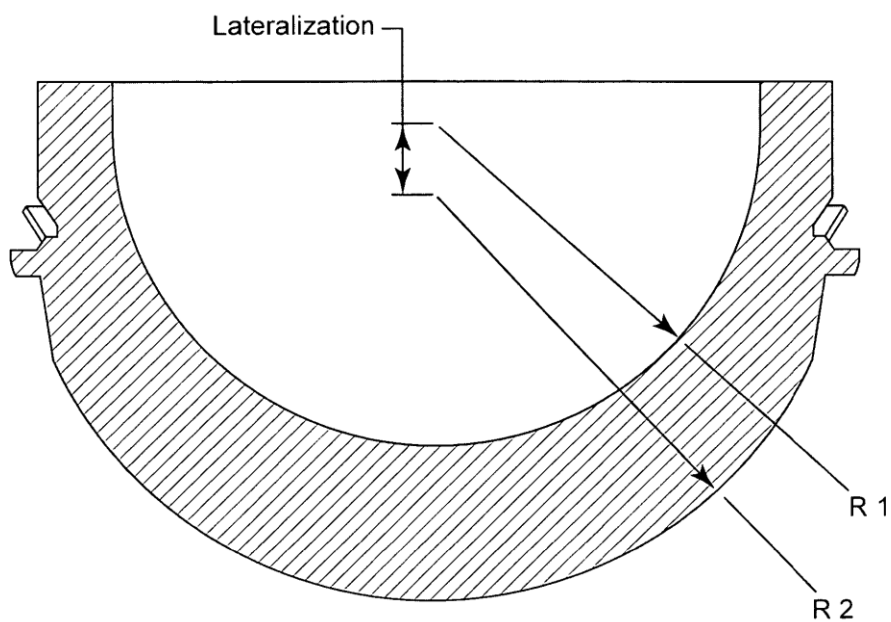


Figure 13: Lateralized liner. Image from [30].

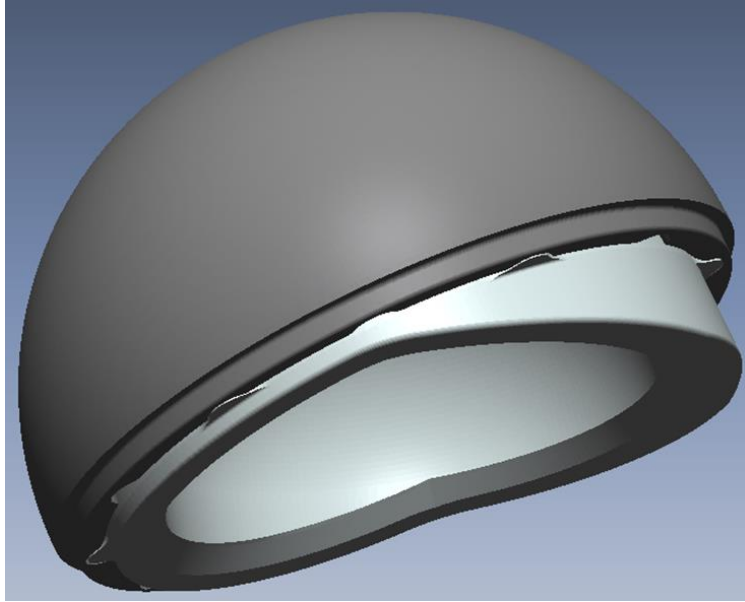


Figure 14: Lipped liner.

expense of a theoretically lower range-of-motion and an increased risk of neck impingement against the elevated portion of the rim [29] [31].

2.1.2 Constrained Liners

In the event of repeated dislocation, surgeons will often use a constrained liner during revision to ensure the femoral head remains constrained within the cup. Constrained liners, such as those shown in Figure 15, require the use of a locking mechanism, usually a metal ring locked to the plastic liner, to prevent warping and dislocation. While these liners are advantageous for patients with recurrent instability problems, these liners can often experience complications of their own, including reduced range-of-motion, re-dislocation due to failure of the locking ring, aseptic loosening, and more [32] [33] [34] [35].

2.1.3 Large-Diameter Heads

A common cause of dislocation after THA is impingement of the neck against the rim of the liner. Hence, surgeons will often turn to a system with a larger head-to-neck ratio, which allows for a larger range-of-motion for the patient and increases the amount of displacement required for dislocation [35]. Large-diameter head systems, such as the one shown in Figure 16, have been shown to reduce dislocation rates [35] [36]. Unfortunately, many older large-diameter femoral head systems were metal-on-metal models, which experienced considerable complications due to metal debris being generated from the articulation, causing adverse reactions to the surrounding soft tissues [37].



Figure 15: Constrained liners. Image from [32].



Figure 16: Large diameter metal-on-metal THA. Image from [36].

2.1.4 Dual Mobility Cups

Like the large-diameter heads, dual mobility bearings also address the issue of dislocation by increasing the head-to-neck ratio of the system [35] [38] [39]. A schematic of a dual mobility bearing can be found in Figure 17a, and an actual implant can be found in Figure 17b. In fact, utilization of dual mobility bearings has led to some of the lowest dislocation rates of all commercial products, and the implants have shown promising survivorship rates thus far [35] [38] [39].

2.1.5 Modular Femoral Stems

Modular stems, shown in Figure 18, offer the theoretical added benefit of more precision regarding the location of the femoral head. In other words, the modularity of these stems allows the surgeon to easily adjust the version, size, and offset of the femoral component regardless of patient anatomy, giving him/her the ability to control exactly where the femoral head should go. However, there are increased risks with modular stems, including extra wear and debris at the modular junction as well as an increased risk of component failure [40] [41]. It has also been shown that there are negligible improvements in femoral head accuracy when using a modular stem versus a standard, non-modular option, indicating that the same positions achieved with a modular stem could have been achieved with a non-modular option [41].

2.1.6 Short Femoral Stems

A recent development in the orthopaedic community is a transition to shorter femoral stem implants, shown in Figure 19. There are many different classifications of



(a)



(b)

Figure 17: (a) dual mobility schematic and (b) dual mobility implant. Images from [35].



Figure 18: Example modular stem options. Image from [40].



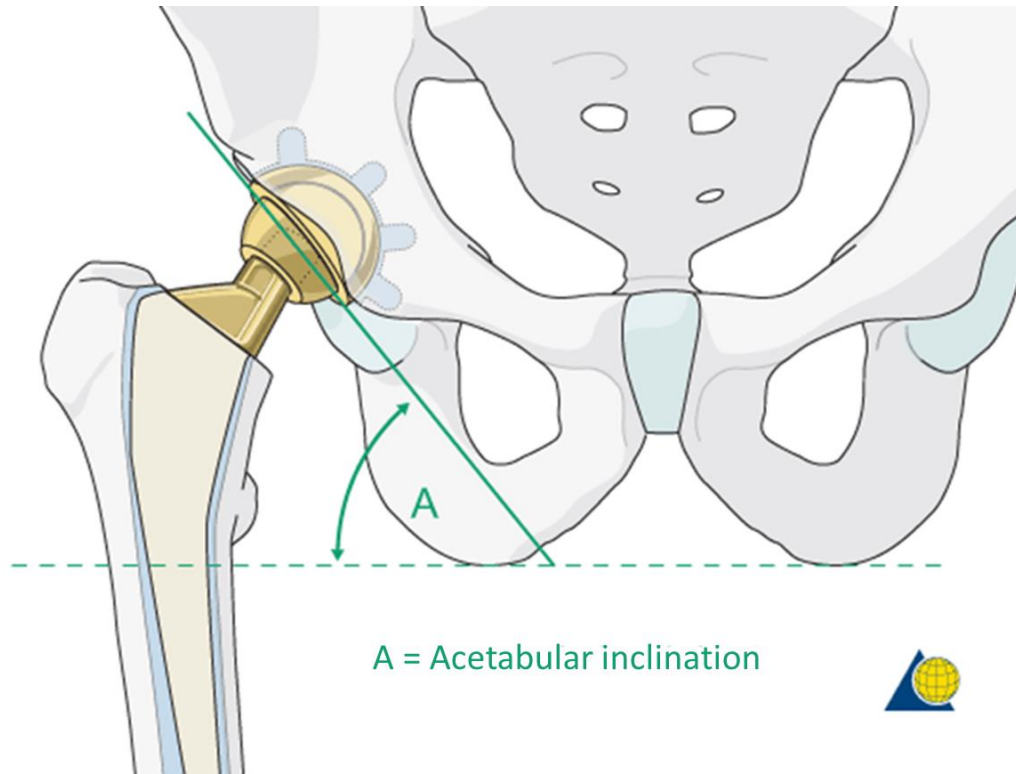
Figure 19: Image of implanted short stem. Image from [42].

“short stems,” but the shared concept among all of them is to shorten the length of the femoral stem and preserve femoral bone in some way. In general, short stem implants hold many potential advantages over conventional stems, including preserved bone stock for future revisions, lowered stress shielding on the proximal femur, improved biomechanics, improved implant fixation, and improved ease of highly technical, minimally-invasive surgical procedures [42] [43] [44]. Additionally, the survivorship of these implants appears to be satisfactory, although many studies are still in early stages [43] [44].

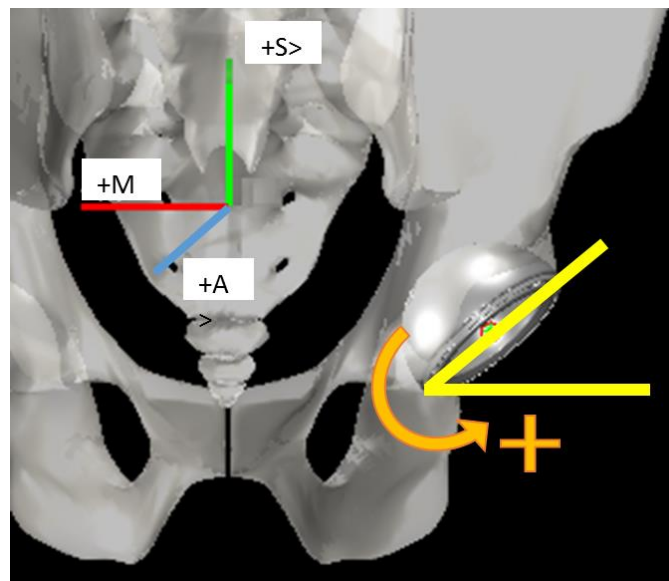
2.2 Component Positioning

2.2.1 Component Rotational Alignments

The intraoperative positioning of hip components has long been an area of study in total hip arthroplasty. Multiple papers [45] [46] [47] [48] [49] [50] [51] [52] [53] have investigated postoperative outcomes and stability of the hip due to a variety of component positions, including cup, stem, and combined angles. Intraoperatively, surgeons generally strive to achieve pre-determined angles of rotation for the cup and stem. Cup inclination, or rotation about the cup or pelvis's A/P axis, is essentially the degree to which the cup opens laterally with respect to the pelvis. Figure 20a shows a visual representation of inclination and Figure 20b shows inclination orientation in the current model. Similarly, cup anteversion, rotation about the cup or pelvis's M/L axis, is the degree to which the cup opens anteriorly with respect to the pelvis. Figure 21a shows a visual representation of anteversion and Figure 21b shows anteversion orientation in the current

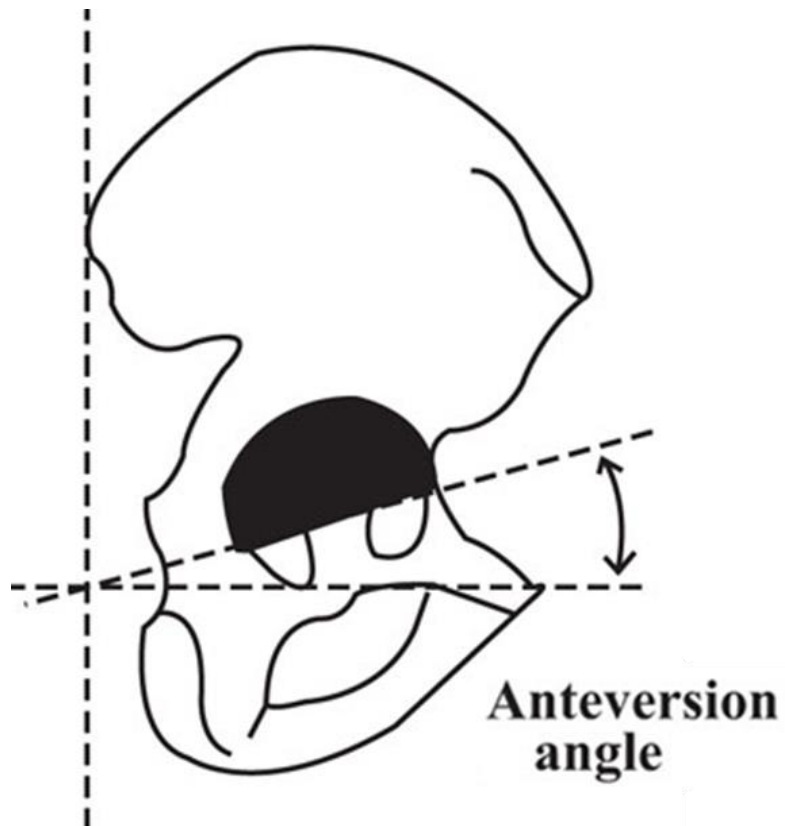


(a)

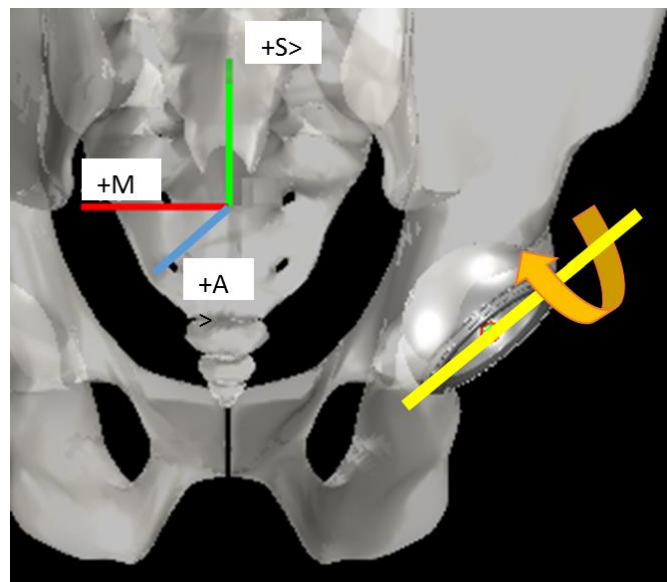


(b)

Figure 20: (a) image depicting acetabular inclination, image from [54], and (b) acetabular inclination in the model.



(a)



(b)

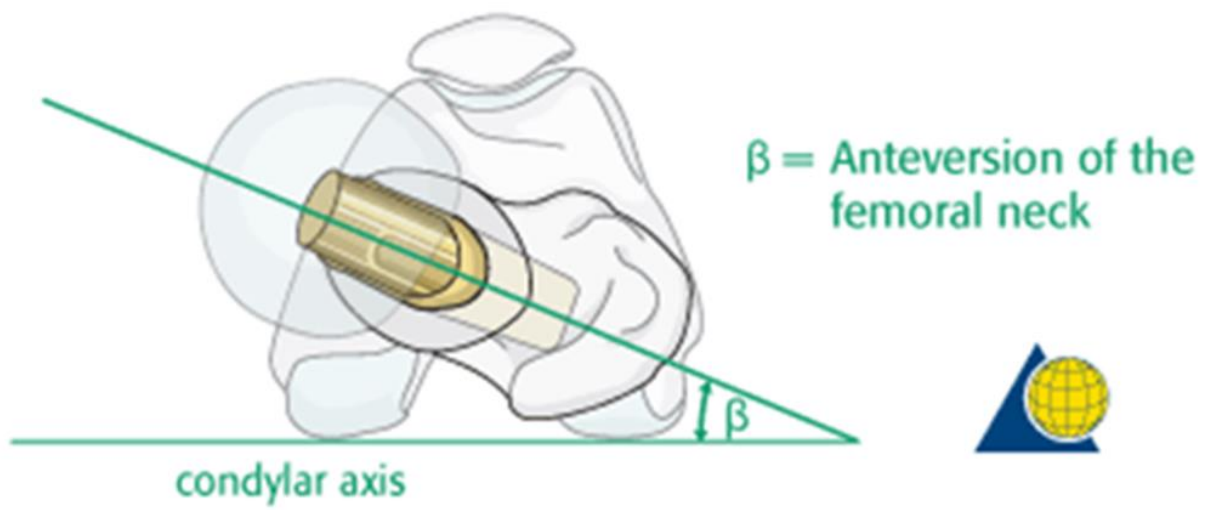
Figure 21: (a) image depicting acetabular anteversion, image from [10], and (b) acetabular anteversion in the model.

model. The goal for the surgeon on the cup side is to appropriately match the natural opening of the native acetabulum. On the other side, femoral anteversion (rotation essentially about the stem or femur's S/I axis, shown in Figure 22) is the internal/external rotation of the stem and is used to better achieve the natural curvature of the femur.

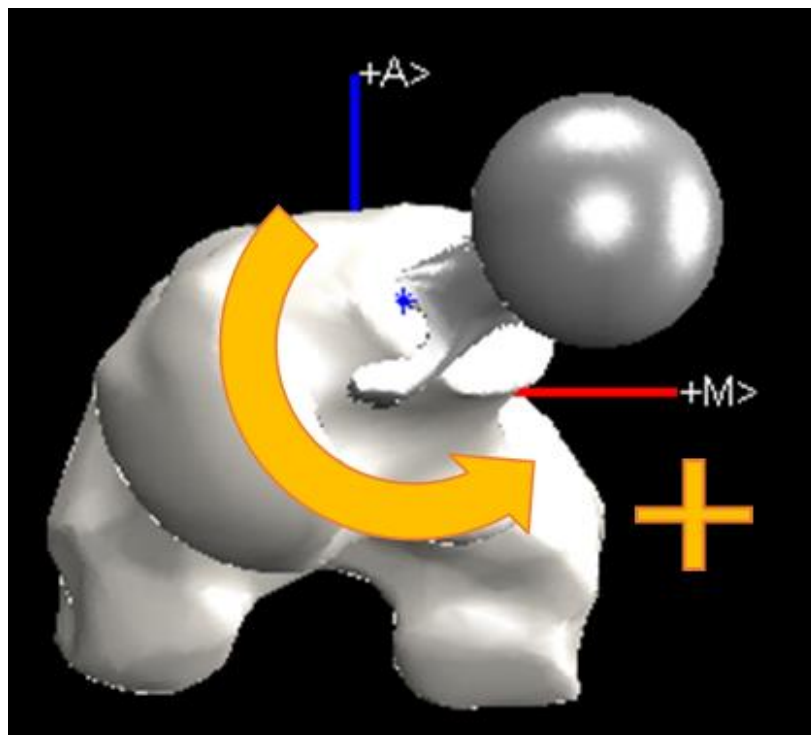
Throughout the literature, various angular “safe-zones” for the acetabular cup have been reported [46] [47] [48] [53], focusing on the stated increase in dislocation rates and impingement for hip patients with cups positioned outside of these zones. These safe zones generally target approximately 35°-45° of cup inclination and 5°-30° of cup anteversion, though many of these safe zones are inconsistent [48]. While surgeons routinely strive to achieve some form of a safe zone, recent studies have arisen questioning the validity of these safe zones, suggesting that the ideal cup placement is specific for each individual patient [50] [52] and that previously published generic safe-zones are arbitrary.

2.2.2 Component Offsets

While most current research focuses primarily on the angular orientation of the acetabular cup, fewer analyses investigate the effects of translational shifts of the joint center. There are many reported benefits to medializing the joint [49] [55] [56] [57], focusing primarily on the theoretical lengthening of the moment arm between the hip joint and the hip abductor muscles and the shortening of the moment arm between the body center of mass and the femoral head. This is done to give patients with weaker musculature more direct control over their abductor function.



(a)



(b)

Figure 22: (a) image depicting femoral anteversion, image from [54], and femoral anteversion in the model.

However, choosing to offset the hip joint can be risky, as both components must be correctly aligned to ensure proper compensation and joint tension. It has been theorized that improper offset compensation, either resulting in a joint that is too tight, too loose, or proprioceptively non-anatomic, may result in hip separation, increased component wear, reduced bone stock, or joint instability [5] [49] [55].

2.2.3 Component Positioning Methods

Given the multitude of surgical technique variations, including exposure angles, incision size, joint clarity, etc. [58], there is often large variability in intended component positioning versus actual component positioning, both rotationally and translationally [57] [59] [60] [61] [62]. Many sources show low repeatability rates for freehand accuracy, with cup inclination ranges as high as 48° [59], cup anteversion ranges as high as 82° [59], joint offset ranges as high as 47mm [60], and successful placement rates within desired safe zones as low as only 26% [59]. Proper alignment is essential for successful postoperative biomechanics, and errors this large are of significant concern to the orthopaedic community.

Hence, there is a large plethora of navigation devices and techniques that have recently been developed to assist the surgeon in a variety of ways, ensuring maximum intraoperative accuracy. A substantial amount of research has been conducted thus far regarding the accuracy of these techniques (i.e. navigation or a specialized jig), and the results overwhelmingly show that some form of navigation yields highly accurate and precise component placements [61] [62] [63] [64] [65] [66]. Such technology available on the market today includes, but is not limited to:

- Patient-specific alignment tools, such as HipXpert (hipxpert.com) and Corin's PS™ Optimized Positioning System (optimizedortho.com), which utilize specialized jigs to assist the surgeon intraoperatively. Preoperatively, a patient-specific plan is determined, resulting in desired component positions. If necessary, patient CT scans and in vivo motion analyses are conducted to determine these positions. Intraoperatively, the jigs are attached directly to the bones to help the surgeon achieve the desired position.
- Computer-assisted surgical devices, such as OrthAlign's alignment technology (orthalign.com), which use specialized navigation technology to intraoperatively track the placement of the surgical jigs and components. These devices are advantageous because they give real-time, operating-room feedback to the surgeon regarding the current placement of the device, allowing the surgeon to change the alignment of the components as needed, while simultaneously ensuring that the new position is exactly what the surgeon desires.
- Robotic-assisted surgical devices, such as Stryker's MAKO robot (stryker.com), which allow the surgeon to create a preoperative surgical plan, and a variety of robotic arms will assist the surgeon to prevent him/her from exceeding boundary conditions and causing undesired problems. Alternatively, the THINK Surgical robot (thinksurgical.com) is a similar product, but in this case, the robot will physically perform the surgery and the surgeon will supervise and take over if necessary.

2.3 Mathematical Modeling Techniques

Multiple types of mathematical models currently exist. Certain models are commercially available (such as OpenSim, AnyBody, and LifeModeler), while other models are simply used for institutional research projects. Before selecting the appropriate modeling technique for the analysis at hand, it is essential to understand the fundamentals and previous research associated with each technique. Despite the considerable number of models that have been created, both commercially and privately, very few of them address the problem of hip joint instability and microseparation.

2.3.1 Commercially Available Models

There are multiple commercially available modeling software packages that exist, such as OpenSim, AnyBody, and LifeModeler. Each software package is powerful in its own way. OpenSim [24], for example, is an open-source software package that uses optimization techniques and can compute both inverse and forward dynamics simulations. This software allows users to create a specific model (via a step-by-step process of input parameters) and then share that model with other users around the world, which is an excellent feature for collaborative biomechanical research. AnyBody [22] is a commercially available software package that uses inverse dynamics, coupled with optimization techniques, to generate in vivo predictions. AnyBody models are very diverse and can be applied to many different industries, including orthopaedics, automotive, aerospace, and more. Finally, LifeModeler is a forward-solution modeling package that can link directly with popular computer-aided design systems to directly import geometries into the model. LifeModeler has multiple plug-in features, including KneeSIM,

LumbarSIM, NeckSIM, and more. In 2012, LifeModeler was acquired by Smith & Nephew, an orthopaedic company [67].

2.3.2 Inverse Solution Modeling

Under the general principle of Newton's Second Law of Motion ($F = ma$), if the mass of a system is known, it is possible to either solve for the forces/torques acting on a body or the associated kinematics (rotations/translations) of the body. With inverse dynamic models [22] [68] [69], the motion of each body segment is known, and thus the model is used to predict in vivo joint forces. Inverse models are commonly used in combination with fluoroscopy studies [70] [71], 3D motion capture techniques [72] [73], or goniometer readings [68]. Inverse solution models are generally more stable than forward solution models, and they are useful for determining the forces within a system, particularly bearing surface forces [17]. However, errors in segment angle measurements (either from skin mounted markers or from fluoroscopy), errors in sensor noise, and errors in body segment parameter choices can all compound and lead to inaccurate predictions [73] [68].

Clinical and orthopaedic applications of inverse models are diverse. Inverse models are often simplified and torque-based. For example, Kim et. al. [72] utilized 3D motion capturing techniques and the commercially available modeling software LifeMOD to analyze knee and hip torques during step up and chair rise activities. This model accurately predicts ground contact forces for the two activities. Additionally, during step up, the model predicts the torques at the hip to be larger than at the knee, but during chair rise, the model predicts the torques at the knee to be larger, indicating the activity-specific

nature of joint reaction force modeling. Their model also incorporates an analysis of these activities using a walking assistant, allowing for an analysis of potentially non-normal activities. However, this model does not transform the joint torques into muscle forces, nor does it observe the resulting joint reaction forces. Lastly, it does not allow for any motion at the joint.

More complex models, such as the ones developed by CS Simpson et. al. [69] using the aid OpenSim, solve for muscle activation profiles instead of joint torques. In this model, the authors developed ranges of muscle activation levels that are required to produce certain motion patterns and compared their model predictions to published EMG data. The authors argue that, due to the over-determined nature of the muscular system, joint torques alone are not adequate for correctly predicting in vivo mechanics. In general, while the authors generally experienced good correlation with existing data, they did find a relatively broad range of “feasible” muscle activation ranges, which further solidifies their argument. Interestingly, the authors determined that almost no single muscle was absolutely “necessary” (from a mathematical perspective, except for the anterior gluteus medius and the tibialis anterior), meaning that the body can compensate for non-activation of most muscles. While this model is undeniably powerful, it primarily focuses on muscle activation data and does not focus on in vivo joint stability.

Unfortunately, inverse models are also subject to many types of errors, most notably from the kinematic inputs, the ground reaction force measurements, surface marker movement, and more. Riemer et. al. [73] present another torque-based model that analyzes the sensitivity of various parameters. In this paper, the authors develop a torque-based model from 3D marker motion capture kinematics. Based on pre-determined

uncertainty set ranges of 5-36% for segment mass, 5-47% for segment moment of inertia, 5-16% for center of mass location, 10-20mm for segment length, 2-8° for segment error, 0.1-0.5% for ground reaction force, and more, the authors concluded that the hip joint can experience resultant torque uncertainties as high as 114-232% and that the knee joint can experience resultant torque uncertainties as high as 50-105%. The authors also determined that the main reason for hip torque uncertainties was due to inaccuracies in segment angles, the distance from the center of pressure to the ankle center of rotation, and the mass of the foot. One again, this model does not transform joint torques into muscle forces, and it also does not allow for any motion at the joint. Most importantly, however, this model demonstrates the high levels of model inaccuracy that can be associated with inverse modeling techniques.

2.3.3 Forward Solution Modeling

Conversely, with forward solution models, the muscle forces in the system are supplied as inputs that are used to drive the system. Because the kinematics of the system are not directly specified, forward solution models are often advantageous when comparing and predicting the behavior of multiple implant types [17]. Forward solution models often incorporate varying contact formulation techniques, such as rigid bodies, elastic foundations, or finite-element models [74]. Depending on the chosen technique, these contact surface definitions can be advantageous when predicting wear patterns for joint replacements. Also, forward solution models often incorporate optimization techniques and can be used to examine the various roles and effects that specific soft tissue characteristics have on a certain activity [17] [74] [75]. In general, the stability and

accuracy of a forward solution model is often dependent on the initial and/or final conditions supplied to the system.

Forward solution models are often used to more accurately analyze the effects that muscle variations have on certain activities. For example, Neptune et. al. [75] used an older version of OpenSim called SIMM to create a forward solution mathematical model of the lower extremity to fully understand how the muscles contribute to the stability and forward motion of the body during gait. The model included many muscles that are also included herein, including the iliopsoas, the gluteus maximum, the vastus muscles, the hamstrings, and more. Their model GRF predictions correlated well with experimental data. This model predicted that, in early stance phase, the knee and hip extensors (vastus, gluteus maximus, hamstrings, and rectus femoris) were the predominant forces, stabilizing the body and absorbing the load, and in late stance, the soleus and gastrocnemius (calf muscles) were the predominant driving force. While this model is certainly powerful, it does not include a methodology to observe joint instability and contact patterns.

Thelen and Anderson [76] demonstrated the mathematical efficiency of computed muscle control (CMC) algorithms in their forward dynamic simulations of gait. In this model, the authors successfully incorporated 92 Hill-type muscle/tendon units into an optimized feedback loop to successfully achieve a forward dynamics simulation based on desired kinematics. Taking approximately 30 minutes per simulation, this model generated kinematics that generally matched within 1° of rotational experimental data and within 10mm or less of translational experimental data. Additionally, their muscle activation profiles closely matched predicted muscle force patterns with known EMG data.

Despite this high degree of accuracy, however, this model does not address hip joint stability and microseparation.

2.3.4 Optimization

Optimization techniques are regularly employed in biomechanical modeling because the human body contains significantly more muscles than it has degrees of freedom [77] [78]. Consequently, the human body is an indeterminate system. Accurate optimization techniques involve determining a reasonable objective function and deriving the solution that best accomplishes the performance criterion. These models may attempt to minimize muscle expenditure/fatigue, or they may attempt to minimize the error between measured and simulation motions [17] [74] [77] [78]. If the objective function of the model is to minimize the muscles forces, then accurate initial and terminal conditions are necessary. If the objective function of the model is to minimize the rotational error, then accurate initial conditions and in vivo motions are necessary. While optimization techniques are indeed powerful, advanced simulations are often computationally expensive [77] and are slightly more time-consuming than desired for this dissertation, as many numerical optimization techniques involve iterative solution convergence. Furthermore, optimization techniques often yield multiple solutions, depending on the nature of the objective function.

In their 2006 publication in *Simulation Modelling Practice and Theory*, Damsgaard et. al. [22] present their powerful commercial mathematical modeling software AnyBody. The AnyBody modeling software uses both optimization and inverse modeling techniques. In this particular paper, the authors analyze a sample gait simulation and

compare model muscle predictions for the gluteus maximus, tibialis anterior, and rectus femoris to known EMG signals. While the normalized magnitudes are occasionally inaccurate, the predictions do accurately predict period patterns. As one of the more publicized commercially available models, AnyBody has clearly outlined four goals that their model must meet: 1) it should be a modeling system; 2) it should facilitate model exchange, cooperation, and scrutiny; 3) it should have sufficient numerical efficiency; and 4) it should maintain realistic levels of complexity [22]. In order to meet these goals, AnyBody has been rigorously validated and shared and has therefore been applied to a multitude of different application, including orthopaedics, automotive, aerospace, and more. While AnyBody is certainly impressive and robust, it does not, to my knowledge, focus on joint stability nor does it allow for hip separation.

Anderson and Pandy [77] developed an optimized forward solution model to predict gait motion based on initial and terminal conditions while minimizing the total metabolic energy expended per unit distance traveled. This methodology is unique because it predicts the motion between two boundary conditions, as opposed to prescribing it using experimental data. In general, this model achieves good agreement between kinematics, ground reaction forces, and muscle profiles compared to known data, indicating that the metabolic energy performance criterion is an accurate choice. However, while extremely powerful, this model focuses primarily on muscle activation, ground reaction forces, and rotational kinematics and does not address hip joint stability and separation.

2.3.5 Reduction

Reduction modeling techniques, such as those employed by Komistek, et. al. [4], D. Glaser [18], as well as in this dissertation, involve simplifying the model such that only primary muscles drive the motion of a body. Ultimately, this keeps the system determinant and therefore only one solution exists for each simulation. While this technique may not be as anatomically accurate as advanced optimization techniques, the differences are often negligible, and reduction techniques often produce solutions faster, which is advantageous for the goals of the dissertation model.

Glaser [18] utilized an inverse, reduced mathematical model to analyze the hip joint during a gait activity based on implant type. Despite using a reduction technique, this model achieved powerful results, predicting maximum quadricep forces of ~300N and maximum hip joint reaction forces between 2.0xBW to 3.0xBW, depending on implant type. Glaser's research is one of the few known sources that directly addresses hip stability, and the results obtained from this model correlate well with in vivo acoustic analyses.

2.3.6 Rigid Body

Rigid body modeling techniques are commonly observed in models that include multiple bodies. With rigid body techniques, it is assumed that the bodies do not deform. When modeling bodies with a high stiffness, such as metallic joint replacement components, this assumption is generally accurate. Rigid body models can involve joint links that are truly rigid [23] and the forces are calculated accordingly, or they can involve joint links that are referred to as "overlap of rigid surfaces," where the forces are computed

from the overlap depth between rigid body surfaces [79]. While rigid body models are extremely useful in determining interactive joint forces and the resulting kinematics, they are unable to accurately predict component deformations or stress distributions, magnitudes, or concentrations, and can therefore only make rudimentary-at-best predictions for component wear. In general, rigid body, multibody modeling simulations are generally much faster than complex, finite-element analyses [79].

Sariali et. al. [16] have developed an experimental simulator as well as a multibody computational model to analyze hip instability. To my knowledge, this simulator research is the most similar research to the hip separation analyses conducted in this dissertation. Experimentally, the authors predict a gradual increase in hip separation during swing phase due to acetabular cup medialization followed by a quick relocation of the femoral head at the beginning of stance phase, at heel strike. Theoretically, the authors have developed a computational model using ADAMS® that incorporates spring/damper representations of the articulating contact surfaces. This model yields a high degree of accuracy compared to the experimental analyses, predicting hip separation values on the order of 0.5 - 2mm [16] [80] [81]. While this model appears relatively similar to the current research, it models an experimental apparatus instead of in vivo conditions and it does not allow for predictions of muscle forces.

2.3.7 Finite Element Analysis (FEA)

While rigid body assumptions will often suffice in these models, in reality all materials deform, depending on the stiffness. Thus, to better understand the detailed contact mechanics that occur at a bearing interaction, FEA is the best modeling technique

of choice. FEA models [25] [82] [83] [81] incorporate material properties for each body, and they are highly effective in predicting deformable body mechanics and stress distributions. The mathematical theory behind FEA is extremely complex, which explains the fact that FEA simulations are generally much more time consuming with regards to initial set up as well as simulation run-time. Due to the computation power required for FEA models, these models are generally more focused on single joints and the detailed interaction that occurs between the surfaces.

Nadzadi et al. [82] used both an inverse model and an FEA model of the hip joint to predict hip dislocation risks during various high-dislocation-risk maneuvers. The inverse modeling techniques were used to determine the angular motion of the components as well as the joint contact forces. This information was then input into the FEA model to determine a “resisting moment” about the cup center as well as the predicted dislocation rates. The model predicts that the maneuver that yields the greatest risk for dislocation is a low sit-to-stand activity, dislocating approximately 87% of the time. In this model, dislocation is frequently accompanied by component impingement. Although this model does address the stability of the hip joint, it adopts a binary “yes” or “no” analysis of hip dislocation and does not analyze the intermediate subluxation phase.

Multiple authors have used FEA techniques to investigate the effects that edge loading scenarios have on the rim of the acetabular cup. For example, Elkins et. al. [83] analyzed the relationship between edge loading contact stress versus edge radius and cup orientation. The authors focused on analyzing metal-on-metal total hip replacements and analyzed a high flexion activity yielding posterior edge loading, specifically a “trunk leaning maneuver, such as when tying shoes.” The authors found a “complex

relationship“ between edge loading stresses/wear and lip geometry/cup orientation. The authors noted that increasing inclination and anteversion both increased the work required for posterior subluxation. Additionally, the authors determined that increasing the cup lip radius decreased the work required for posterior subluxation. However, for larger lip radii, while edge loading was predicted earlier, the intensity of the loading was reduced. In general, this model focuses on edge loading scenarios due to a lack of femoral coverage and does not analyze hip pistoning separation.

Conversely, Liu, Williams, and Fisher [81] have developed a finite element model of the simulator by Sariali et. al. [16], discussed above. This FEA model therefore takes into consideration a similar mechanism of hip separation to what is presented in this dissertation. This FEA model only focuses on the instant of contact between the femoral head and the acetabular cup, corresponding to heel strike. Even with a hip separation magnitude low as 0.1mm, the authors predicted the peak contact stress to be greater than the yield strength of the cobalt-chrome material, and increases in separation magnitudes yielded steep increases in contact stress, with a maximum amount predicted in the study of over 2GPa. This goes to show the significance of femoral head instability within the acetabular cup, predicting plastic deformation patterns on the metal bearing that are similar to what has been observed in retrieval studies. While this model does directly address hip separation, like the discussion above, it does not incorporate a full in vivo model and it does not allow for predictions of muscle forces.

CHAPTER 3: OBJECTIVES AND CONTRIBUTIONS

3.1 Dissertation Objectives

While previous research, both theoretical and experimental, has undeniably yielded large advancements in total hip arthroplasty, most previous investigations focus mainly on analyzing the postoperative accuracy of component placement and retrospectively correlating complication rates to a variety of factors. Additionally, most of the mathematical models have focused on developing the most accurate torque or muscle representations of the lower limb. None of the investigations or models have focused solely on predicting the causes and effects of in vivo hip joint instability and micro-separation, and little advancements have taken place towards generating a viable solution to this problem. Regarding mathematical models specifically, to my knowledge, none of the models are forward solution models specifically focused on determining the causes and effects of hip separation, nor are any of these models currently geared towards implementation with intraoperative alignment software packages.

Hence, the primary objective of this dissertation is to address and analyze hip separation. Specifically, the objectives of this dissertation are to develop a fully functional forward solution mathematical model of the hip that allows for comparisons between various implant designs and determinations of factors leading to in vivo hip separation, instability, and edge loading. This includes:

- Development of a forward solution mathematical model of the hip joint that incorporates detailed articulating surface geometry and patient-specific kinematics.

- Advancement of the forward solution model to predict the causes and effects of hip joint separation, instability, and edge-loading.
- Customization of the model to analyze multiple patients, multiple surgical techniques, multiple surgical alignments, and multiple implant designs.
- Implementation of the model to conduct detailed simulations to compare various surgical techniques and implant designs and to determine ideal component alignment zones.
- Design of a graphical user interface to merge the model with intraoperative alignment tools, yielding intraoperative predictions of postoperative joint mechanics and stability.

3.2 Contributions

As mentioned previously, there are several types of mathematical models of the hip joint that currently exist, and the benefits that are currently arising from these models are profound. Every model that has been developed, either commercially or privately, comes with its own unique set of capabilities, assumptions, and limitations; while a model may excel in one area, it certainly falls short in another area.

Specifically, many of the models that exist assume a ball-and-socket joint at the hip, and such models experience limitations in component shape accuracy and in boundary conditions. Additionally, many of the models do not allow for the determination of hip separation and instead focus on predicting muscle and joint reaction forces of known scenarios. Many of the models are developed to analyze cadaveric rigs, which fail to produce in vivo results. Lastly, to my knowledge, none of the models are specifically

focused on determining the causes and effects of hip separation, nor are any of these models currently geared towards implementation with intraoperative alignment software packages.

Thus, the model developed herein focuses overcoming these shortcomings to intraoperatively predict hip separation, micromotion, pistoning, and edge loading for in vivo scenarios. Specifically, this dissertation brings about five novel contributions to the field. This dissertation contributes:

1. A fully forward mathematical model of the hip joint that incorporates detailed articulating surface contact detection geometry for translational control and accurate in vivo muscle forces for rotational control.
2. A fully forward mathematical model of the hip joint that can predict occurrences of hip separation, instability, and edge-loading.
3. Model customization techniques that can accept multiple implant designs, surgical methods, and patient-specific kinematics.
4. Model analyses that offer detailed comparisons between various implant designs, surgical techniques, and component alignment scenarios, each with the goal of determining the causes and effects of hip separation, instability, and edge-loading.
5. The initial development of an intraoperative tool that could give the surgeon an intraoperative prediction of postoperative mechanics.

CHAPTER 4: METHODS

4.1 General Modeling Methods

The overall goal of this dissertation was to develop a forward solution model of the hip joint that accurately predicts the occurrence of hip separation, instability, and edge loading and offers solutions on how to prevent these problems in the future. This includes developing accurate representations of bones, ligaments, and muscles, as well as developing methods to control the forward model variables, specifically hip joint rotations and hip joint translations. An overall representation of the model can be found in Figure 23.

4.1.1 Autolev and Kane's Method of Dynamics

The two coding programs of choice for this project are the symbolic manipulation program known as Autolev, developed by Thomas R. Kane and David A. Levinson, and the standard coding platform C++. Autolev, a force, motion, and code-generation tool for dynamic systems, functions to build a virtual representation of a human leg and constructs the various mathematical equations that represent the system. This is specifically accomplished by creating bodies within the code, such as the tibia, femur, patella, etc., and populating these bodies with various points of interest, such as muscle/ligament attachment sites, body contact points, etc. Next, the muscle and ligament forces, as well as interactive forces at the joints, are entered into the system to generate the dynamic equations of motion.



Figure 23: Muscles (red), ligaments (green), bones, and implants of the current model.

As a general physics rule, any unconstrained body present in 3-dimensional space contains 6 degrees-of-freedom (3 rotational and 3 translational) that describe its orientation and/or motion. The bones created in this model are no different. For example, if a model is comprised of 4 bodies, a total of 24 equations must be derived to completely describe the system. Therefore, using Autolev, the user can generate a total of 24 equations of motion. To keep the system determinate, the Autolev user could utilize these equations to solve for only 24 different unknowns (for example). Depending on the type of model, these unknowns can be forces and torques (for inverse modeling) or rotations and translations (for forward modeling). It is also possible to solve for a combination of both forces/torques and rotations/translations, depending on what the modeler is interested in determining.

Once the model system has been properly constructed, Autolev generates the equations of motion using Kane's method of dynamics. Kane's method of dynamics is simply an alternative method of dynamics, employed by Autolev and formulated from Newton's Second Law of Motion using the notion that $ma = m(\partial v / \partial t)$. Kane's dynamics revolve around the equation

$$F_r + F_r^* = 0,$$

where F_r is a representation of the generalized active forces on the system (such as gravitational forces and ground reaction forces) and F_r^* is a representation of the generalized inertial forces (forces that do no physical work, such as joint reaction forces) [4] [18] [84] [85]. Kane's dynamics incorporate the concepts of partial velocities and generalized speeds to solve the system. Generalized speeds, represented in the Autolev program as "U's", are essentially a method to pass the forces throughout the entire

system, similar to Newton's notion of "equal and opposite." In the model script, it is necessary to pass appropriate U's from one body to the next to ensure that the forces are being accurately transferred throughout the system.

Once these equations of motion have been developed, Autolev outputs the equations of motion into C++. From there, the PID controllers and the Contact Detection algorithms need to be added to the code, and C++ then solves the equations. The specifics of the PID controllers and the Contact Detection algorithms are discussed in Sections 4.1.5 and 4.1.6, respectively.

4.1.2 Bones

A total of six bodies are represented in the model, shown in Figure 24: the foot, the tibia, the femur, the patella, the pelvis, and a bulk head-arm-torso (HAT) representation (not shown). When applicable, a seventh body is introduced to model a dual mobility THA. This will be discussed later. To model the bones in the system, points are added to the bodies to construct accurate bone representations based on generic bone models present within the Center for Musculoskeletal Research (CMR) bone database, located at the University of Tennessee, Knoxville. More points can be added to specific regions of interest, such as attachment sites or joint interfaces, and fewer points can be utilized where no interactive forces will occur.

The masses of the foot, tibia, femur, and pelvis are modeled as 1.4%, 4.6%, 10%, and 24% of the subject's body weight, respectively. The mass of the patella is considered negligible. Therefore, the patella is modeled as a three-force system that includes the quadriceps force, the patellar ligament force, and the patellofemoral reaction force, in

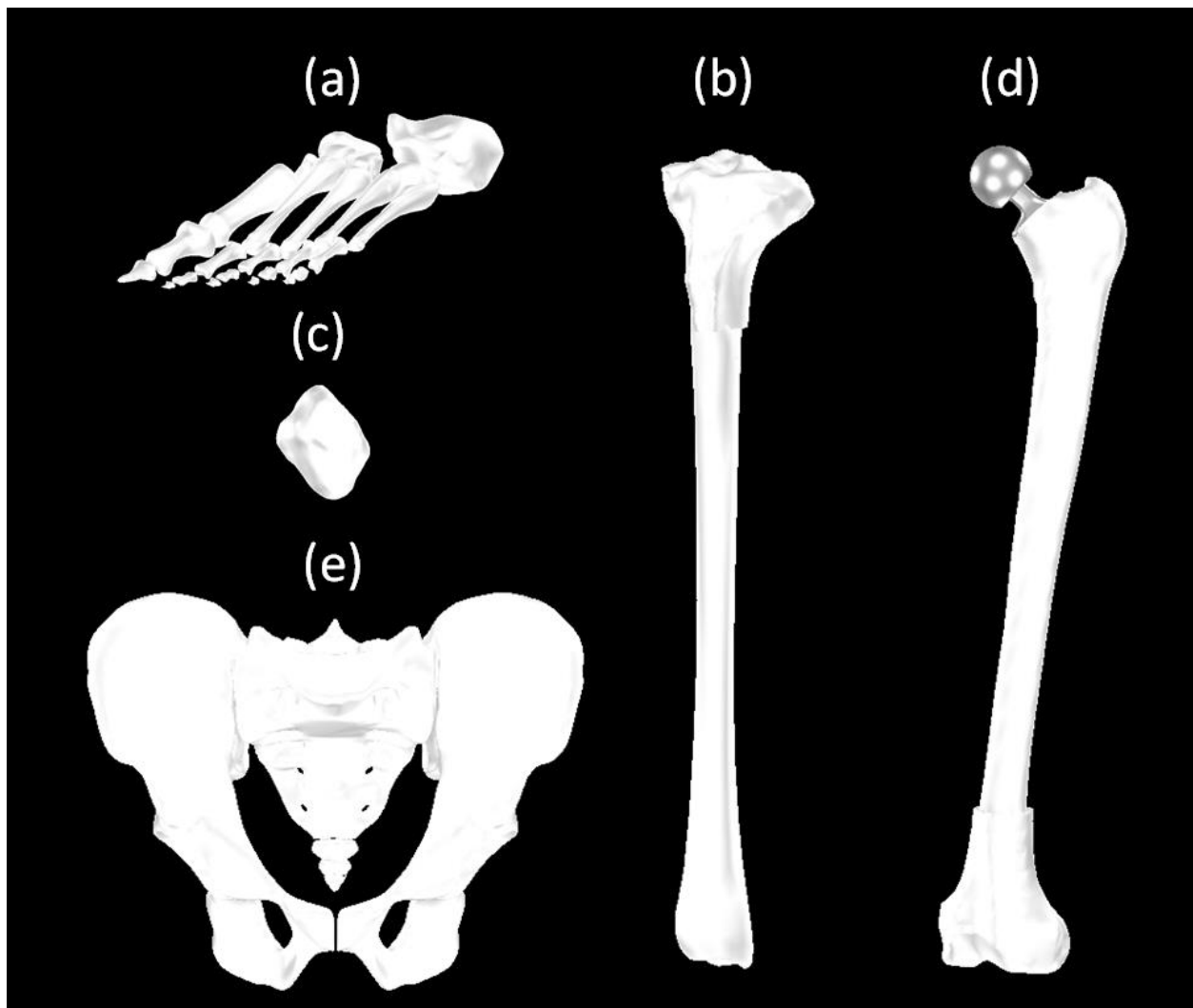


Figure 24: (a) foot, (b) tibia, (c) patella, (d) femur, and (e) pelvis bone representations.

accordance with [18]. The mass of the HAT was determined by taking the entire body weight and subtracting the weight of the pelvis plus twice the mass of the foot, tibia, and femur (to account for two legs), yielding 44% BW.

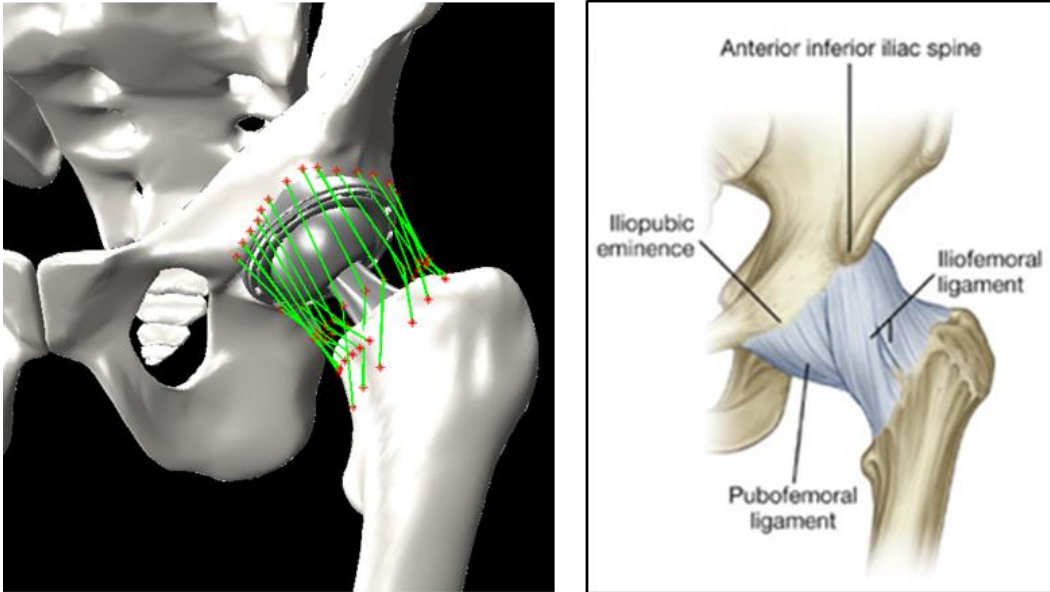
4.1.3 Ligaments

The ligaments included in the model are the primary hip capsular ligaments (the ischiofemoral ligament, iliofemoral ligament, and pubofemoral ligament, as well as the ligamentum teres when applicable). The ligaments are modeled as non-linear springs [86]. Ligament force is given by the equation

$$F = \begin{cases} 0, \varepsilon \leq 0 \\ k/2 (L - L_0), 0 \leq \varepsilon \leq 2\varepsilon_1 \\ k[L - (1 + \varepsilon_1)L_0], 2\varepsilon_1 \leq \varepsilon, \end{cases}$$

where k is stiffness, gathered from literature and existing modeling projects [18] [87], L is the current length of the ligament at each specific time step, L_0 is ligament slack length, assumed for each patient, and ε_0 is reference strain, again gathered from literature and existing modeling projects [18] [86]. The ligaments provide reaction forces to the system based on the relative motion of the bones, and hence they are passive forces and that are primarily dependent on bone motion. The iliofemoral, pubofemoral, and ischiofemoral ligament representations can be found in Figure 25. The ligaments are modeled as multiple fibers that wrap over bones and implant components, and the ligaments can be cut, weakened, or moved to compare surgical techniques or errors.

Anterior



Posterior

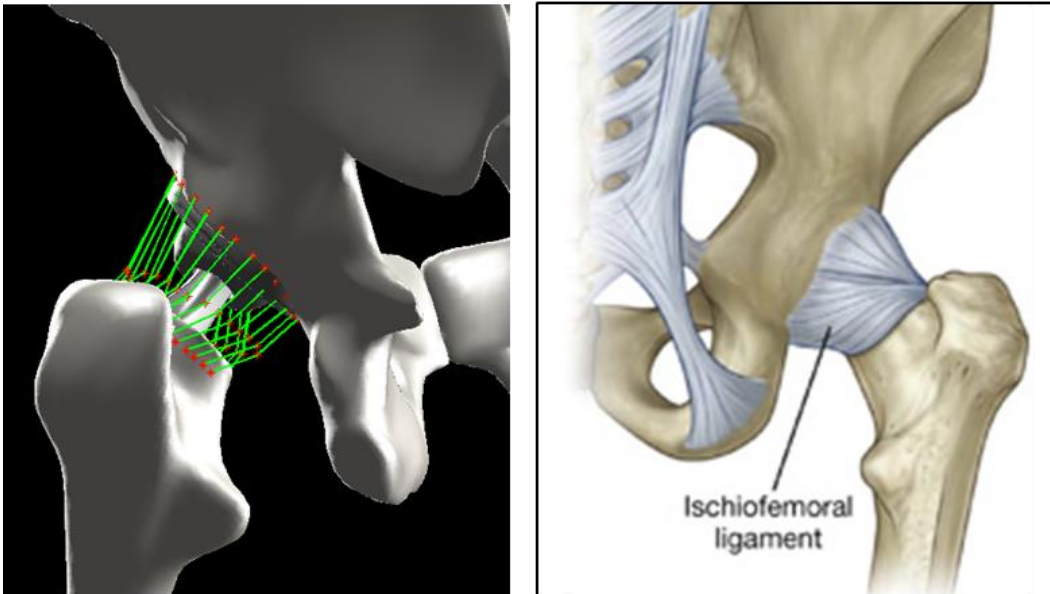


Figure 25: Anterior (iliofemoral and pubofemoral) and posterior (ischiofemoral) hip capsular ligaments included in the model. Right images from (basicmedicalkey.com).

4.1.4 Muscles

The model contains a total of fifteen muscles groups: the vastus medialis, the vastus intermedius, the vastus lateralis, the rectus femoris, the long biceps femoris, the short biceps femoris, the semimembranosus, the iliacus, the gluteus maximus, the gluteus minimus, the gluteus medius, the tensor fasciae latae, the adductor brevis, the adductor longus, and the adductor magnus. The model also contains a torque to represent the oblique abdominal muscles.

The muscles in the model are defined as active forces that are used to drive the motion of the bones. Each muscle in the model is represented as multiple lines, representing fibers, acting over a specific attachment area that can wrap around the bones. The number of fibers for each muscle was determined based on literature-reported physiological cross-sectional areas (PCSAs) of the muscle [88] [89] [90]. The specific number of fibers chosen for each muscle can be found in Table 1.

Before the muscle representations can be explained, it is necessary to understand how the model incorporates torques at each joint. In general, the motion of any body in a dynamic system can be described by six degrees of freedom (three rotational and three translational). Thus, a bare minimum model of the lower extremity (containing no ligaments and no muscles) would have exactly three forces and three torques present at each joint. The forces at the joint are used to control body translations, and the torques at the joint are used to control body rotations. Having these six unknowns at each joint allows the model to remain fully determinate.

Therefore, to incorporate a muscle into the model, a torque must be removed. If the appropriate torque is removed, the respective muscle that is added can recreate the

Table 1: Number of muscle fibers for each muscle.

Muscle	Fibers [#]
Iliacus	11
Gluteus Maximus	18
Gluteus Medius	18
Gluteus Minimus	6
TFL	3
Adductor Brevis	2
Adductor Longus	3
Adductor Magnus	13
Biceps Femoris Short	3
Biceps Femoris Long	6
Semimembranosus	7

no-longer-existing torque (through $T = r \times F$) and can increase the forces across the joint of interest (thereby creating a more accurate representation of the joint forces). For example, at the knee, the firing interaction between the quadriceps muscles and the hamstring muscles cause flexion or extension of the knee about the M/L axis. Therefore, to model these muscles in the current model, the knee torque about the M/L axis must be removed and replaced by the quadriceps and hamstring muscle forces. Thus, as is true in the human body, the quadriceps and hamstring muscles control knee flexion and extension (Figure 26).

4.1.4.1 Muscle Switches

The hip contains significantly more muscles than the knee joint. Muscles crossing a particular joint can induce an increase in the force at that joint. Therefore, it is important to include as many muscles as possible in the hip model to gain a more complete representation of the true compressive hip joint force. Additionally, because human muscles cannot “push” and can only “pull,” a minimum of two opposing muscles are required to completely represent the torque at a joint. To successfully solve for two antagonistic muscle forces, binary switches have been incorporated that turn one muscles “on” and keep the other muscle “off,” therefore only solving for one appropriate muscle group at a time. This allows the system to remain determinate and simultaneously increase the number of muscles that cross the hip joint, but consequently the opposing muscles generally do not fire simultaneously. The schematic shown in Figure 27 demonstrates how one variable (“Force”) can be used to compute two different muscle forces, and it shows a snippet of script that explains how these switches are turned on and off. Specifically, if the model is actively firing the quadriceps muscles and, given the



Figure 26: Quadriceps and hamstring muscles controlling knee flexion/extension.

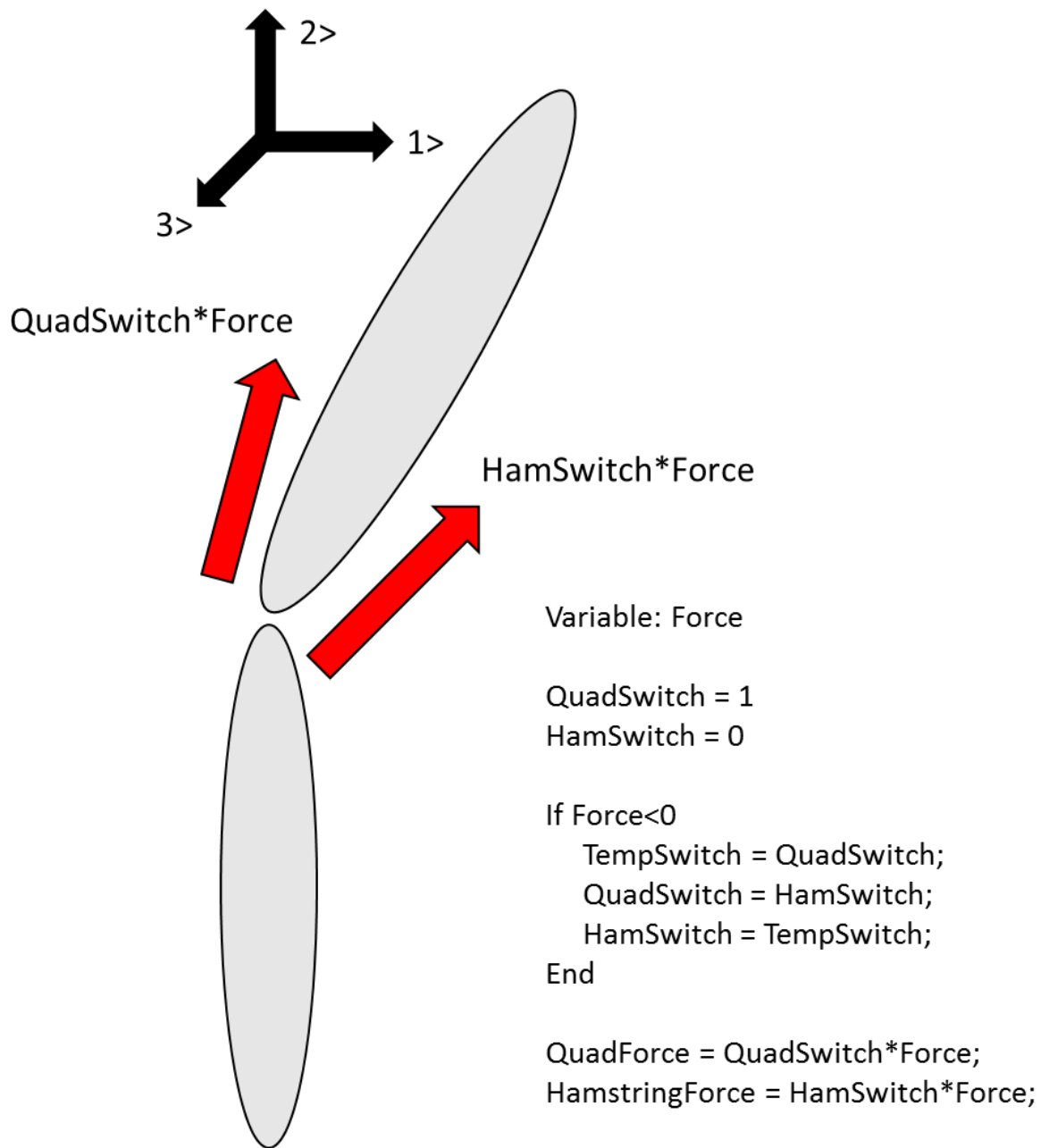


Figure 27: Schematic demonstrating antagonist muscle switches. These calculations allow for complex activities and prevent muscles from firing negatively.

specific activity, the hamstring muscles need to activate to change knee direction, then the model will register that the quadriceps are beginning to fire a negative force. From there, it will switch the QuadSwitch to “off” and the HamSwitch to “on,” and the activity will continue. This allows both flexion and extension to be controlled by one variable and it also prevents muscles from firing negatively.

4.1.4.2 Muscle Contributions

Additionally, due to the complex muscle functions at the hip, muscles generally contribute to more than one degree of rotation. Thus, it becomes more difficult to replace joint torques with muscle representations without using complex optimization algorithms. To accomplish this, the primary function of a specific hip muscle is determined by the line-of-force of the muscle with respect to the location of the hip joint [8], and from there the moment arm of each muscle fiber is used to determine the contribution percent that each fiber will have to the total muscle force. This functions to create more accurate muscle force calculations without complex, time-consuming optimization algorithms.

To better understand this concept (and how it is done in the model), refer to the simplified schematic shown in Figure 28. Crossing the force vector of a muscle fiber with its radius will generate the torque representation of that fiber. From there, dotting that torque with a specific body direction will determine the how much that fiber contributes to the rotation in that specific direction. Muscles with larger radii will have larger contributions. Similarly, muscles with a line-of-action that more directly generates a torque will have larger contributions. The schematic shown in Figure 28 is very simple, only showing 3 muscle fiber contributions about a single axis. According to this diagram, the top muscle fiber would have a contribution of 0.4167, the middle muscle fiber would have

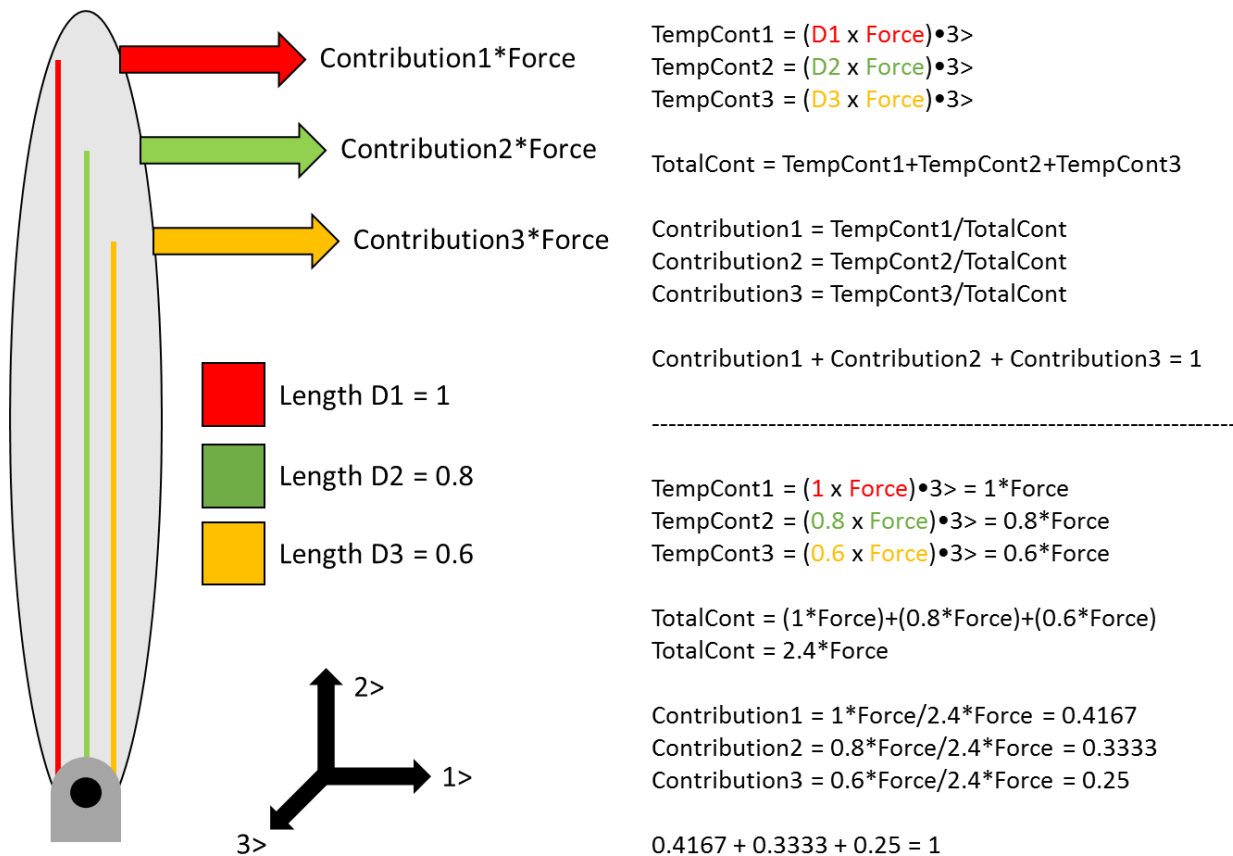


Figure 28: Schematic demonstrating muscle fiber contribution calculations. These calculations function as a pseudo-optimization algorithm to avoid complex, time-consuming optimization algorithms.

a contribution of 0.3333, and the bottom muscle fiber would have a contribution of 0.25. These contributions sum to 1, yielding the force of the entire muscle. In the model, this system is applied to every single muscle fiber, regardless of the specific muscle it belongs to, to determine fiber contribution factors. This quickly becomes very complex, as there are many muscle fibers and they rarely (if ever) lie perfectly in a single plane. Thus, the fibers often contribute to multiple directions of rotation at once. This method is more accurate than assuming equal contributions of each muscle fiber because muscles with a smaller moment arm generally yield lower forces and function more as stabilizer muscles, and muscles with larger moment arms generally yield higher forces and function more as the primary actuators.

For example, the iliacus muscle attaches superiorly and anteriorly to the hip joint with an inferior line of action, and thus this muscle predominantly controls hip flexion. Conversely, the gluteus maximus muscle attaches superiorly, laterally, and posteriorly to the hip joint with a line of action in both the inferior and anterior directions. Thus, the gluteus maximus will function to control both hip joint extension and abduction due to the more complex lines of action.

4.1.4.3 Specific Model Muscles

In the model, the quadriceps muscle group consists of 4 muscle fibers, shown in Figure 29. The quadriceps primarily function to control knee extension. The secondary function of the quadriceps, in particular the rectus femoris, is to control hip flexion. The hamstring muscle group consists of 16 muscle fibers, shown in Figure 30. The hamstrings primarily function is to control knee flexion. The secondary function of the hamstrings, in particular the long biceps femoris and the semimembranosus, is to control hip extension.

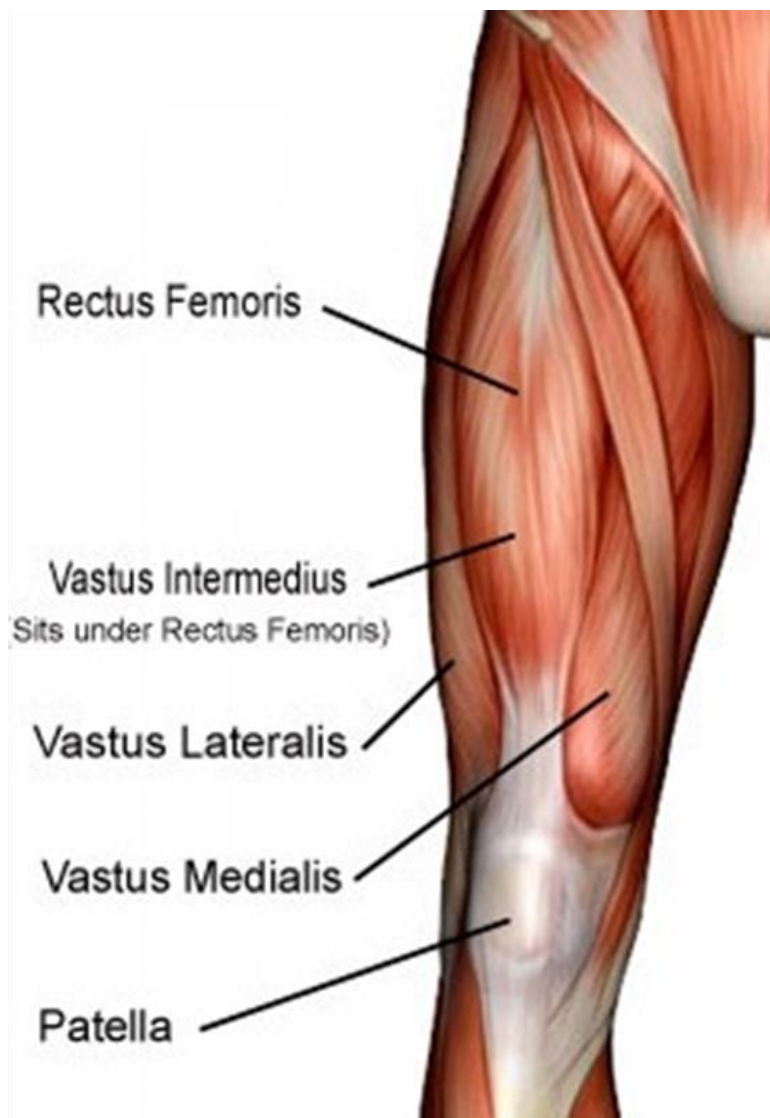
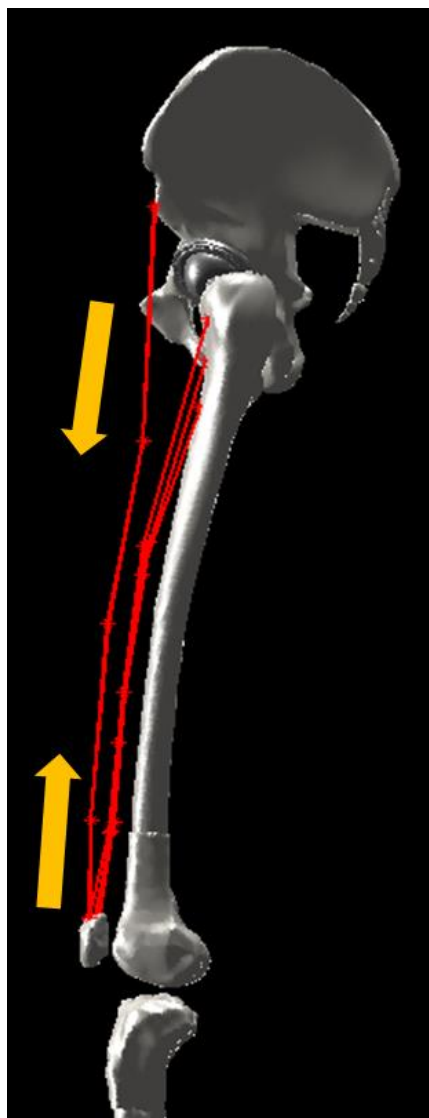


Figure 29: Quadriceps muscles controlling knee extension and pelvic flexion. Right image from (floodta.com).

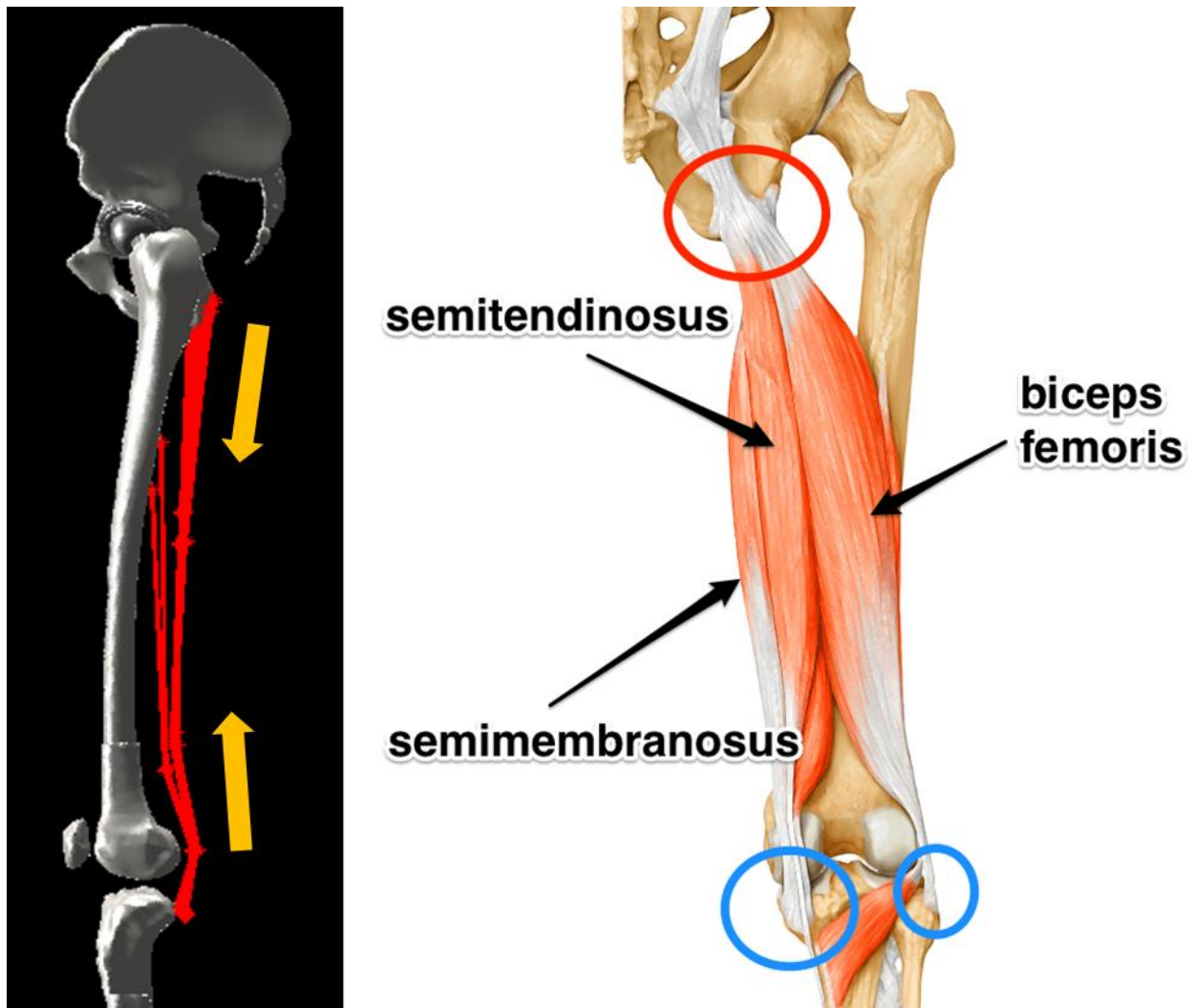


Figure 30: Hamstring muscles controlling knee flexion and pelvic extension. Right image from (yoganatomy.com).

The iliacus muscle consists of 11 fibers, shown in Figure 31, and functions to control hip flexion. The gluteus maximus muscle consists of 18 muscle fibers, shown in Figure 32. The gluteus maximus primarily functions to control hip extension. The secondary function of the gluteus maximus, particularly the middle and upper fibers, is to control hip abduction. The gluteus medius muscle consists of 18 muscles fibers, shown in Figure 33. The gluteus medius functions to control hip abduction. The gluteus minimus muscle consists of 6 muscle fibers, shown in Figure 34. The gluteus minimus also functions to control hip abduction. The tensor fasciae latae (TFL) muscle consists of 3 muscle fibers, shown in Figure 35. The TFL primarily functions to control hip abduction. The secondary function of the TFL is to control hip flexion. The adductor brevis consists of 2 muscle fibers, shown as the 2 superior fibers in Figure 36. The adductor longus consists of 3 muscle fibers, shown as the 3 inferior fibers in Figure 36. Both muscles function to control hip adduction. The adductor magnus muscle consists of 13 muscle fibers, shown in Figure 37. The adductor magnus primarily functions to control hip adduction. The secondary function of the adductor magnus is to control hip extension. The femoro-pelvic internal and external rotations are controlled by an external torque on the pelvis (Figure 38) as well as by a specified contralateral leg force that is applied at the center of the opposing acetabular cup.

4.1.5 Hip Joint Rotational Control: PID Controllers

The three rotational degrees of freedom at the hip joint (A/P, S/I, and M/L rotations) are controlled by the hip muscles. Correct muscular control is ensured by placing

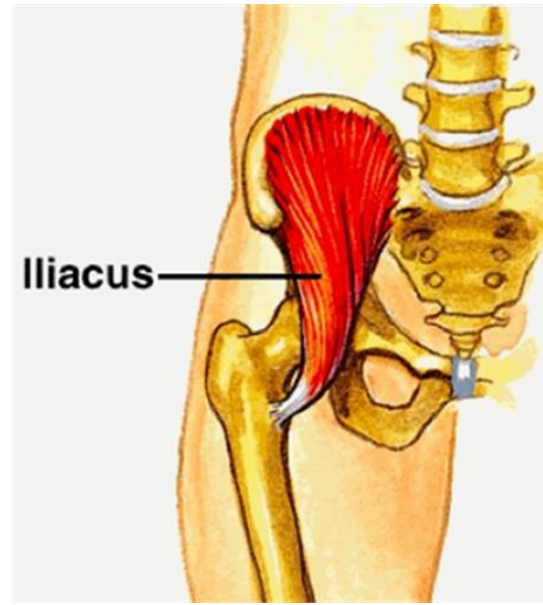
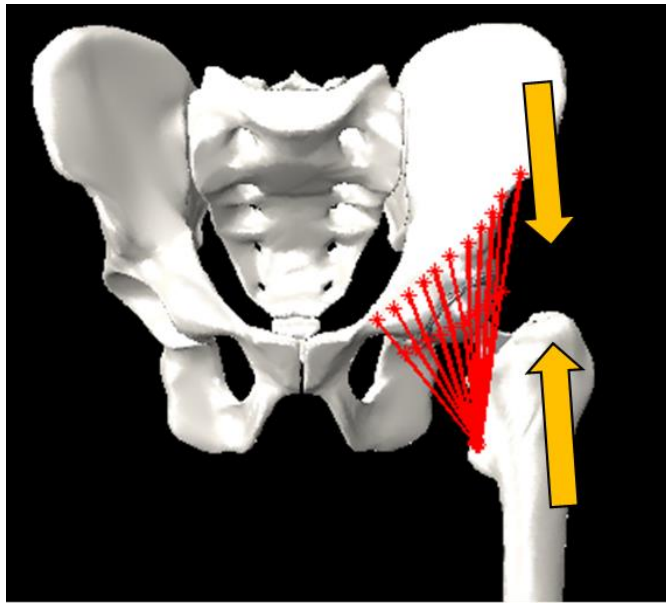


Figure 31: Iliacus muscle controlling pelvic flexion. Right image from (painreliefvermont.com).

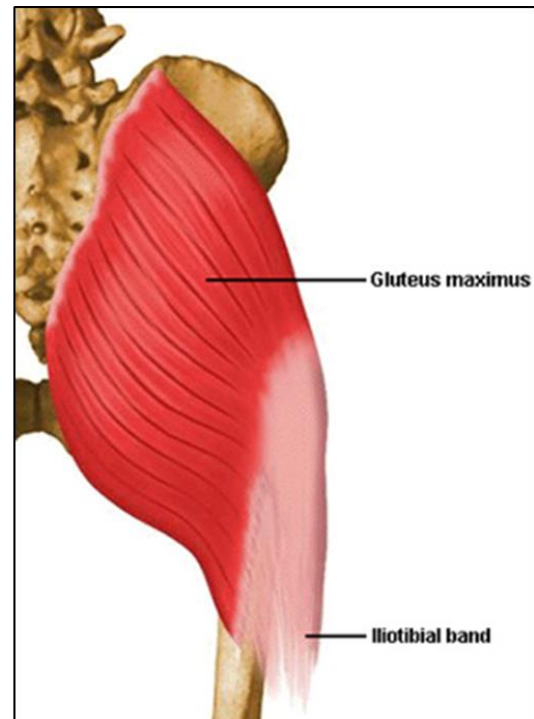
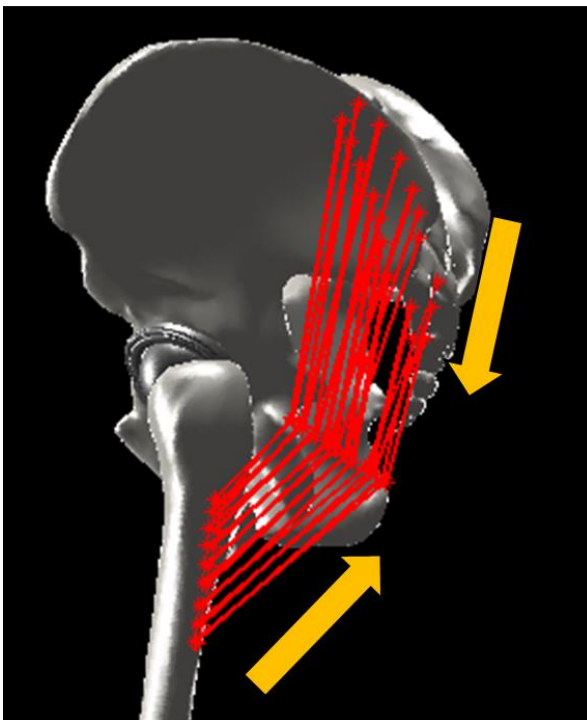


Figure 32: Gluteus maximus muscle controlling pelvic extension and hip abduction. Right image from (mikereinold.com).

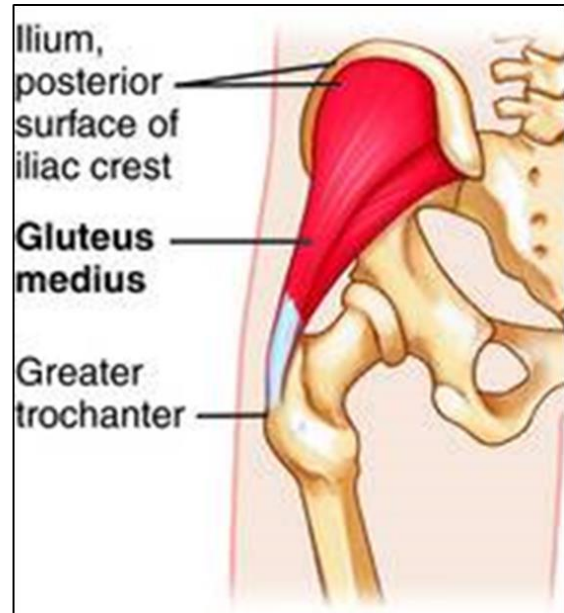
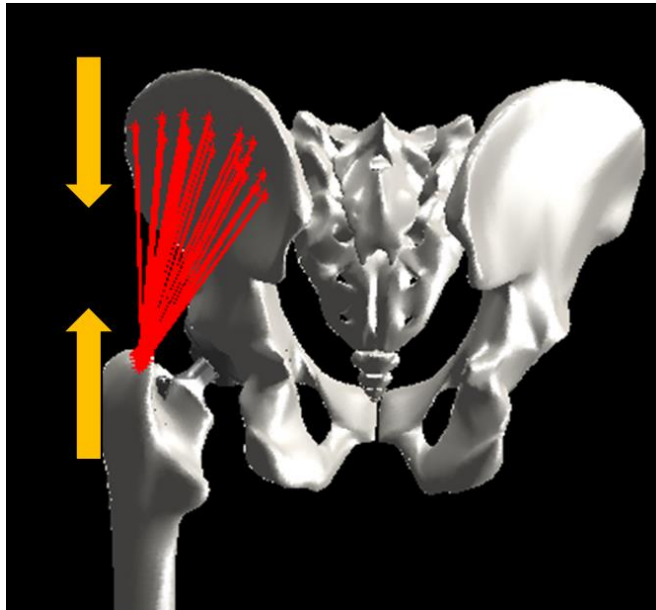


Figure 33: Gluteus medius muscle controlling hip abduction. Right image from (medical-dictionary.thefreedictionary.com).

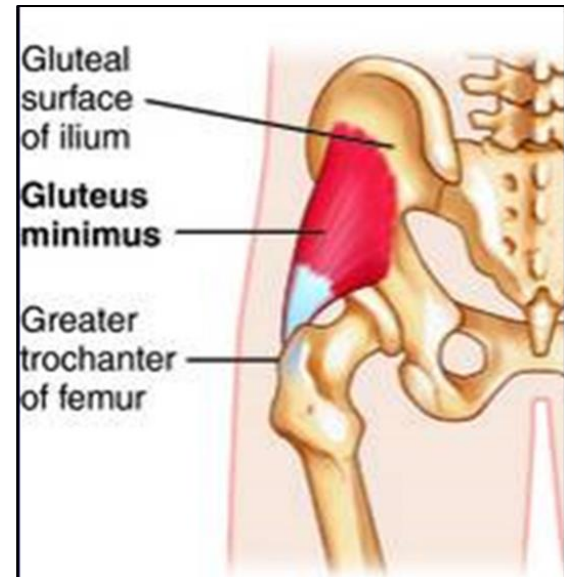
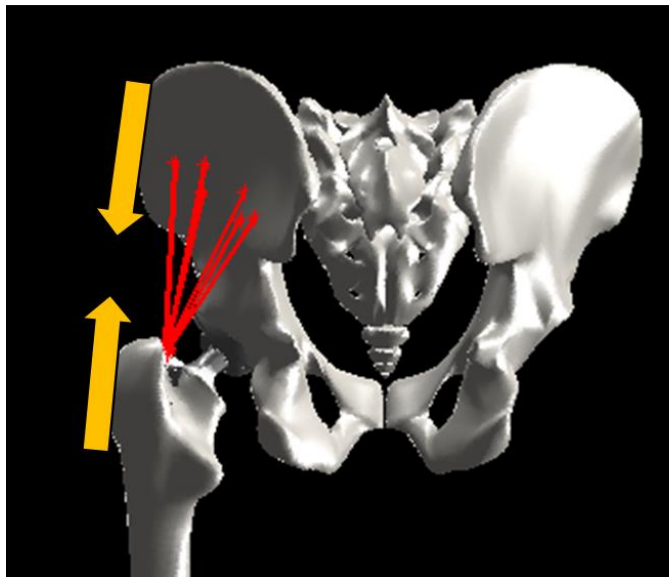


Figure 34: Gluteus minimus muscle controlling hip abduction. Right image from (medical-dictionary.thefreedictionary.com).

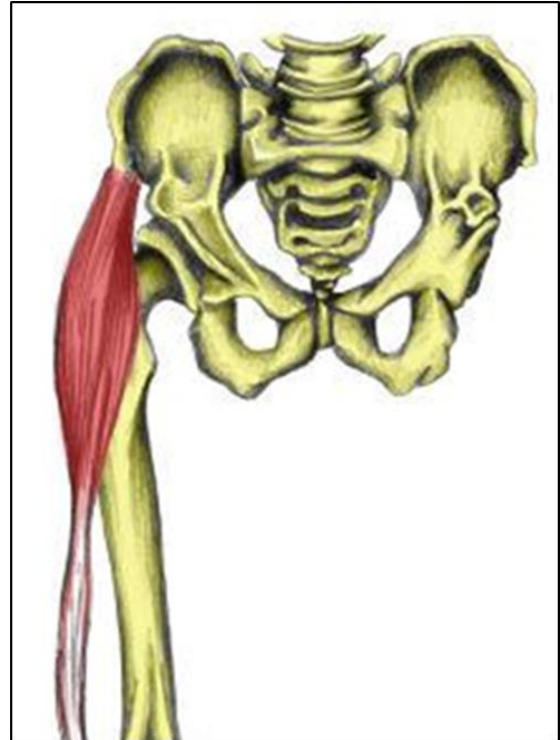
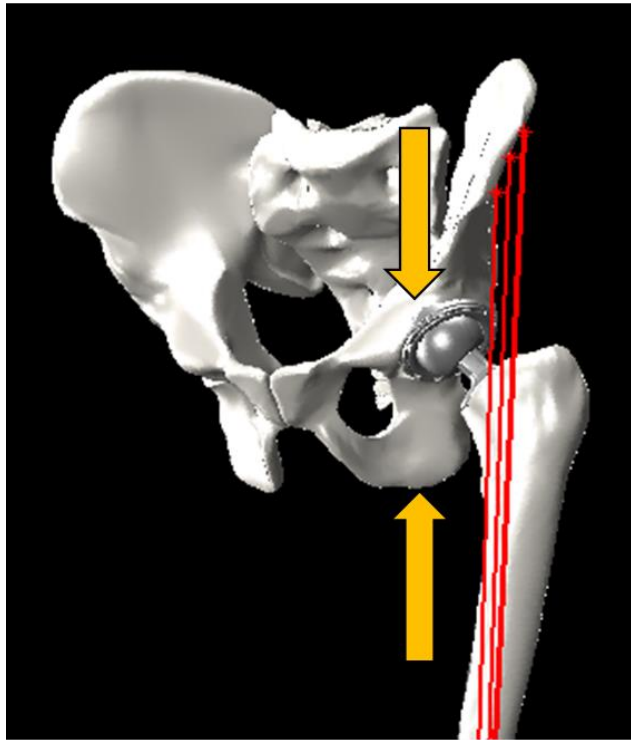


Figure 35: Tensor fasciae latae muscle controlling hip abduction and flexion. Right image from (sportsinjuryclinic.net).

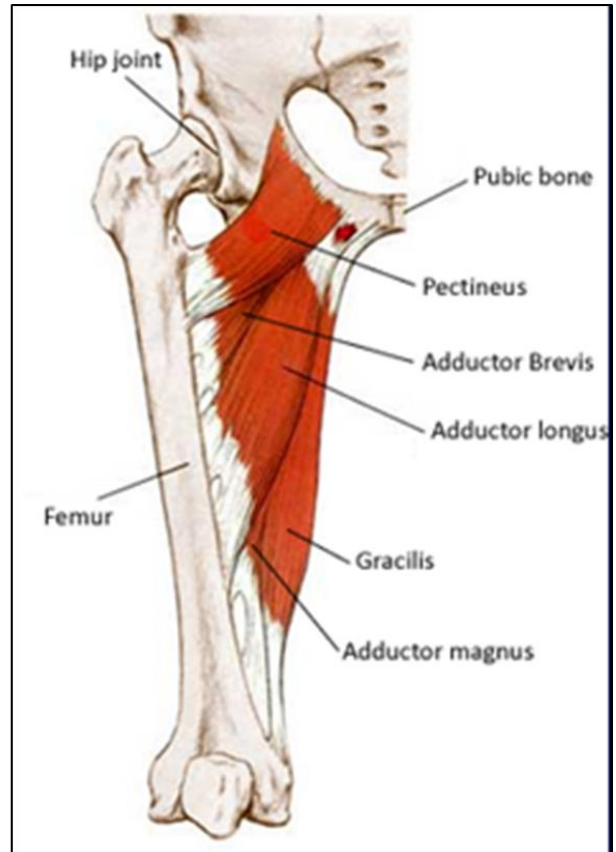
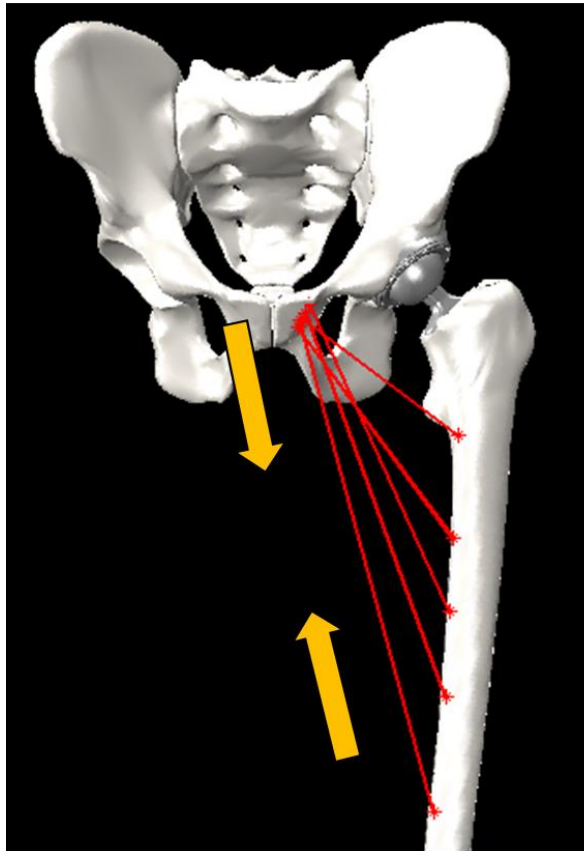


Figure 36: Adductor brevis and adductor longus muscles controlling hip adduction. The adductor brevis is shown as the 2 superior fibers, and the adductor longus is shown as the 3 inferior fibers. Right image from (movementenhanced.com.au).

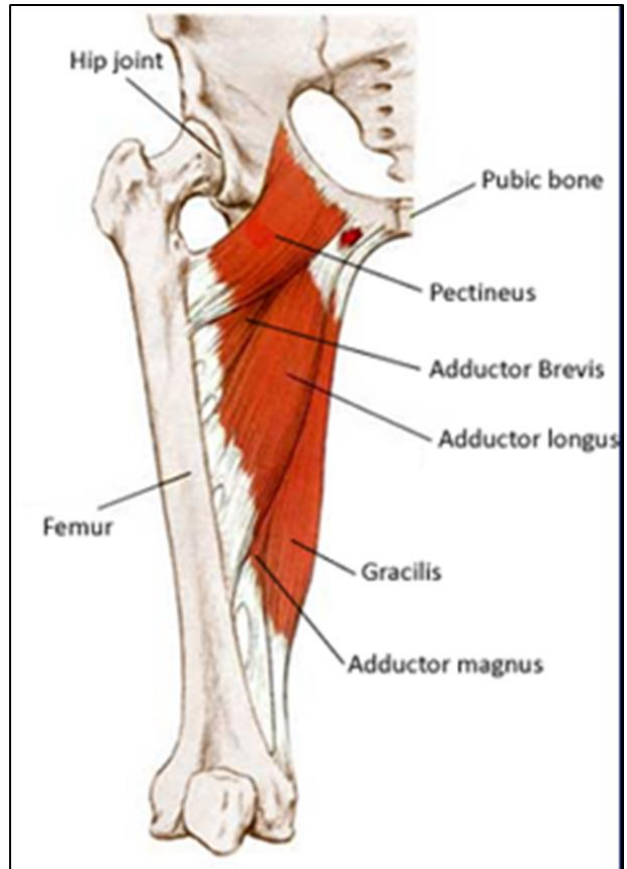
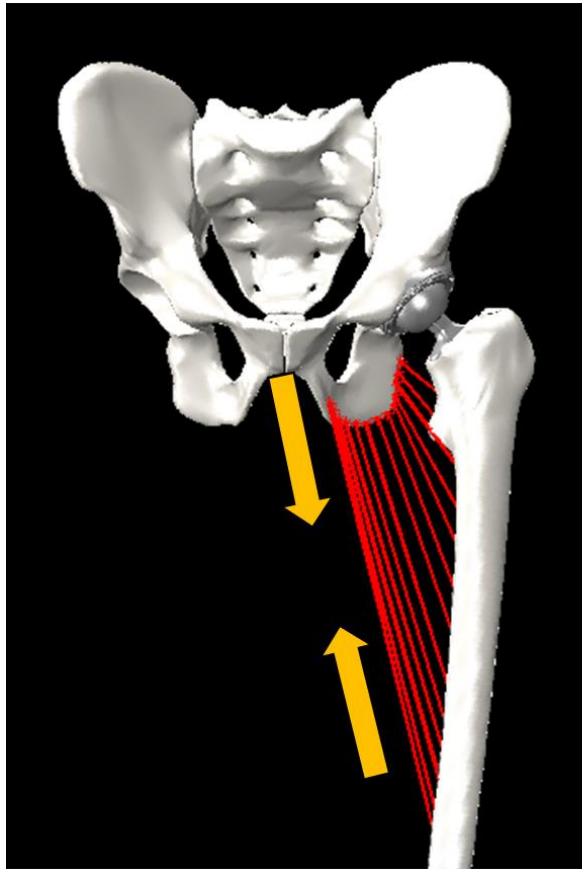


Figure 37: Adductor magnus muscle controlling hip adduction and extension. Right image from (movementenhanced.com.au).



Figure 38: Top view showing S/I axis torque rotation.

proportional-integral-derivative (PID) controllers on the joint muscles to achieve the desired motion. A schematic of the current PID controller can be seen in Figure 39.

These controllers adjust the muscle forces at each time step to ensure that correct joint rotations are achieved. In Autolev, the muscle variables are coded using the following notation:

$$Force = MuscAddFact * (C0 + C1 * T)$$

where $C0$ can, if desired, be set to an initial muscle force value (predicted based on the activity of interest), and $C1$ can be set to zero. The variable $Force$ is passed throughout the Autolev code as explained in Figure 27 and Figure 28, discussed above, and these schematics are combined accordingly to generate a more complex representation of muscle fiber forces.

At each time step, the system determines the error between the actual motion velocity and the desired motion velocity. These computed errors are multiplied by their respective gains and summed together to adjust the “Muscle Addition Factor” ($MuscAddFact$) that adjusts the corresponding muscle forces accordingly. The respective gains (Proportional Gain, $PGain$; Integral Gain, $IGain$; Derivative Gain, $DGain$) can be adjusted appropriately to minimize rotational error.

The Proportional Error is calculated using the equation:

$$ProErr = CurrentRotVel - DesiredRotVel$$

The Integral Error is calculated using the equation:

$$IntErr = OldIntErr + ProErr * TimeStep$$

The Derivative Error is calculated using the equation:

$$DerErr = (ProErr - OldPosErr)/TimeStep$$

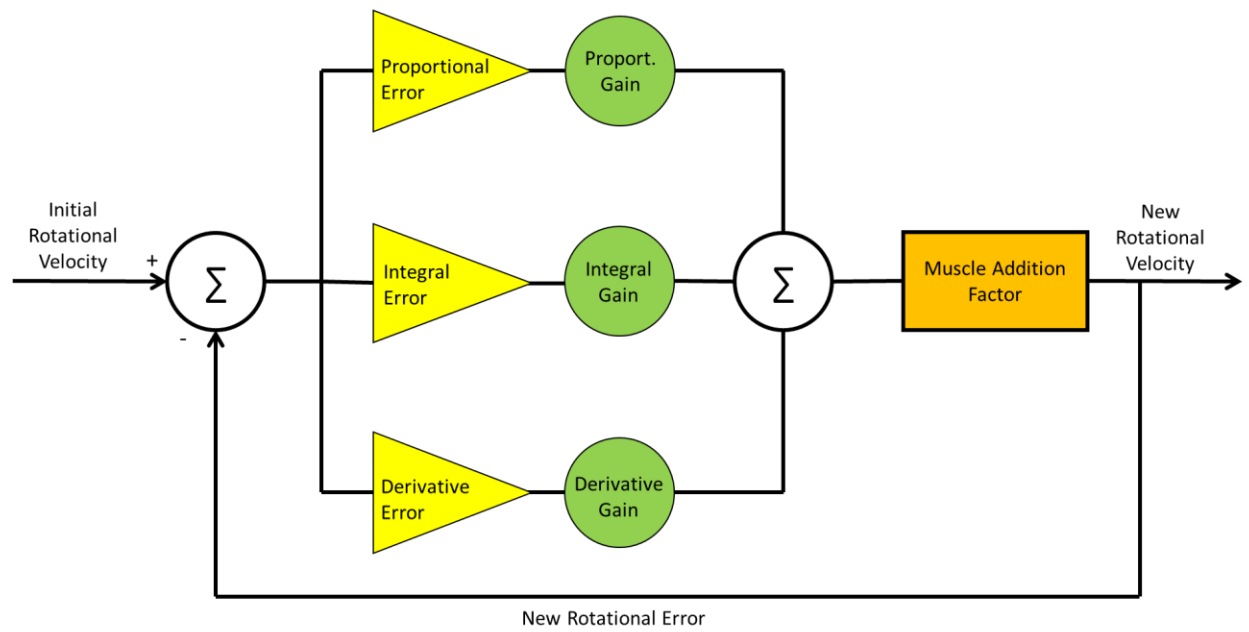


Figure 39: Schematic of the current PID controller acting on rotational velocity.

The errors are summed according to basic PID notation:

$$AddToMuscAddFact = PGain * ProErr + IGain * IntError + DGain * DerErr$$

Lastly, the Muscle Addition Factors are modified according to:

$$MuscAddFact = MuscAddFact + AddToMuscAddFact$$

In the model, a total of 3 PID controllers are present in the model, one to control hip joint flexion/extension, one to control hip joint abduction/adduction, and one to control hip joint internal/external rotation. Thus, specific muscles that contribute to multiple rotations are affected by multiple PID controllers.

4.1.6 Hip Joint Translational Control: Contact Detection

Three translational degrees of freedom at the hip joint, in the acetabular cup reference frame (Figure 40), are controlled by a “contact detection” algorithm. The model computes the interactive hip force using a spring-damper representation (Figure 41) on the acetabular surface. Large magnitudes of stiffness and damping are chosen to simulate a rigid contact surface, and the “penetration” computed by the model simulates sub-millimeter cartilage or polyethylene deformity. Specifically, the model includes a detailed point cloud to represent the femoral component and piecewise polynomial surfaces to represent the acetabular cup, and it tracks the motion of the femur points with respect to the acetabular polynomials.

The femoral head point cloud representation is based on the mesh from a CAD model of the component (Figure 42). The acetabular liner surface is represented by a 3-dimensional polynomial surface that best approximates the shape of the liner. To fully

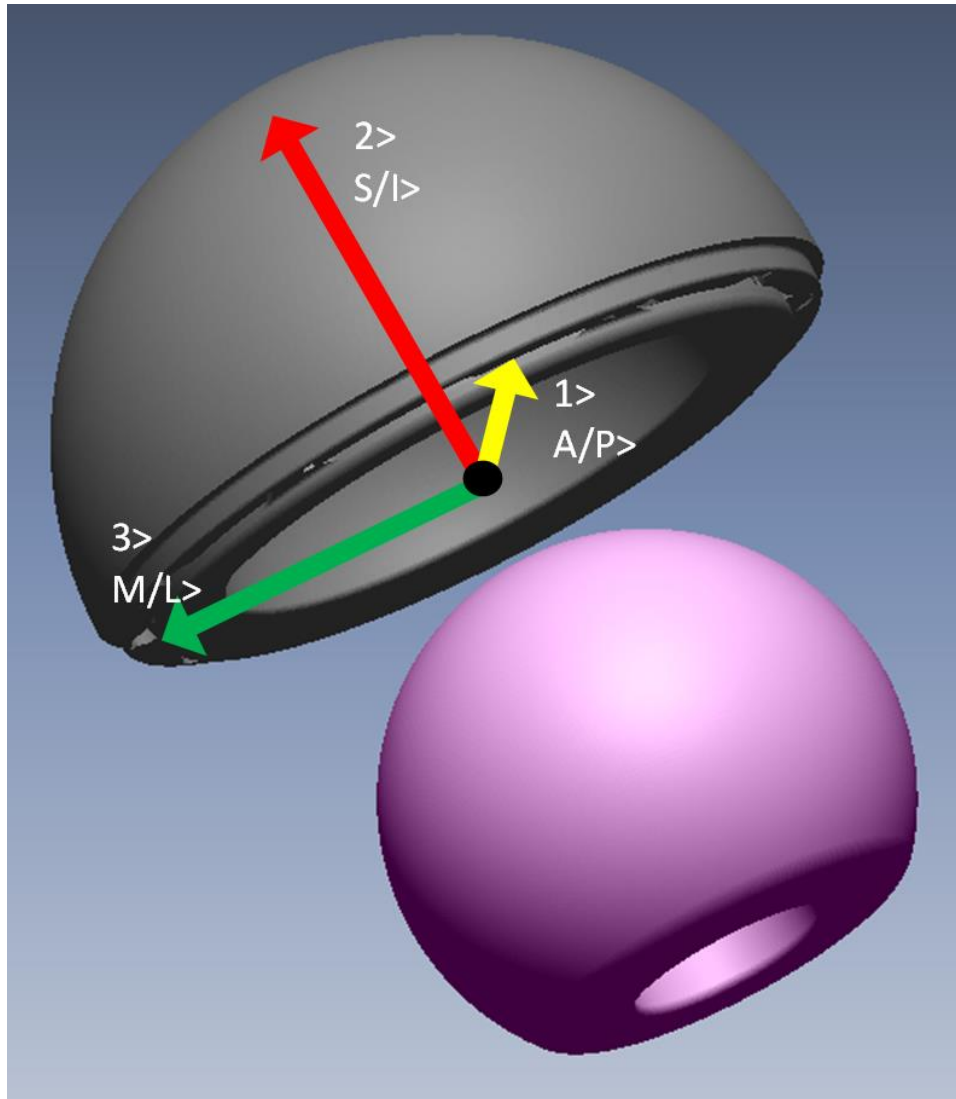


Figure 40: Acetabular cup axis system.

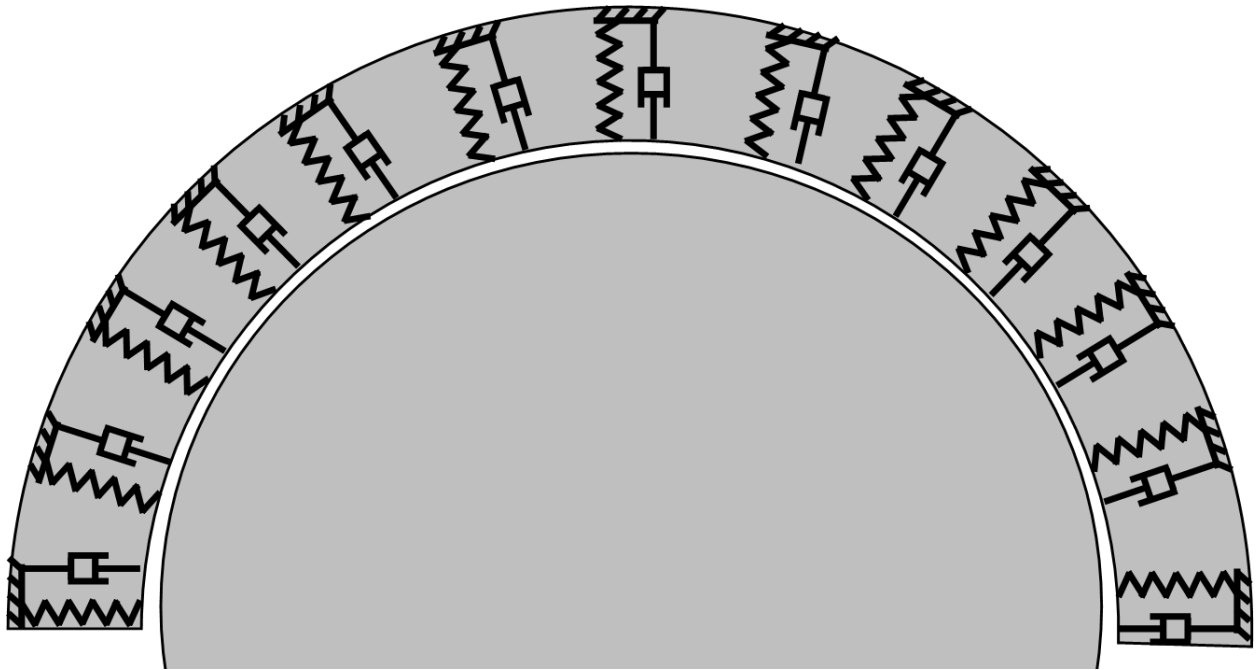
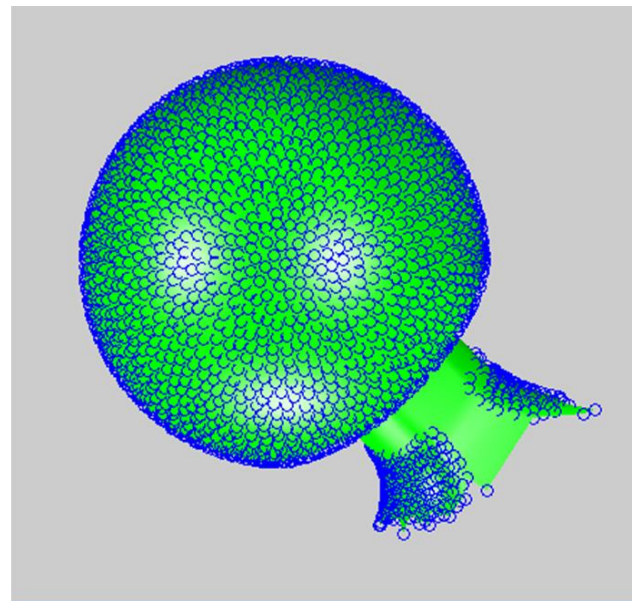


Figure 41: Spring-damper representation.



(a)



(b)

Figure 42: (a) sample femoral component CAD model, and (b) associated vertices to be exported into text file for contact detection.

define the liner surface, a 3-part surface has been constructed. Figure 43 shows the three parts. For lipped liners, the surfaces have each been subdivided, yielding six parts (Figure 44). The spherical liner region fit can be seen in Figure 45a. The cylindrical bore region is defined as a perfect cylinder with matching spherical radius and a specified depth. The rim region (Figure 45b) is defined using a polynomial surface fit and is used for joint stabilization aid in extreme cases of hip separation as well as for predicting femoral neck impingement. The appropriate polynomials are stitched together in a piecewise fashion to form the complete liner representation. In the model, contact detection of the spherical sections are done in spherical coordinates, contact detection for the bore sections are done in cylindrical coordinates, and contact detection for the polynomial sections are done in Cartesian coordinates.

Throughout the activity, the model tracks the position of all of the femoral points in the acetabular cup's reference frame. At each time step, the model inputs the femoral point vertices into the cup polynomials and compares the actual values to the predicted values, yielding a computation of whether the femoral head is in contact. Specifically, Figure 46 shows 2 points present in space (shown in yellow). The top point, P1, has an X-value of X_1 and a Y-value of Y_1 . When X_1 is entered into the shown polynomial of $Y = F(X)$, it can be seen that the actual point's Y-value (Y_1) is greater than the polynomial's corresponding Y-value of $F(X_1)$. Therefore, this point is above the polynomial line and is not in contact. Conversely, P2 also has an X-value of X_1 but a Y-value of Y_2 . In this case, when X_1 is plugged into the shown polynomial, it can be seen that the actual point's Y-value (Y_2) is less than the polynomial's corresponding value of $F(X_1)$. Therefore, this point is below the polynomial line and is in contact. In the model, this polynomial curve is

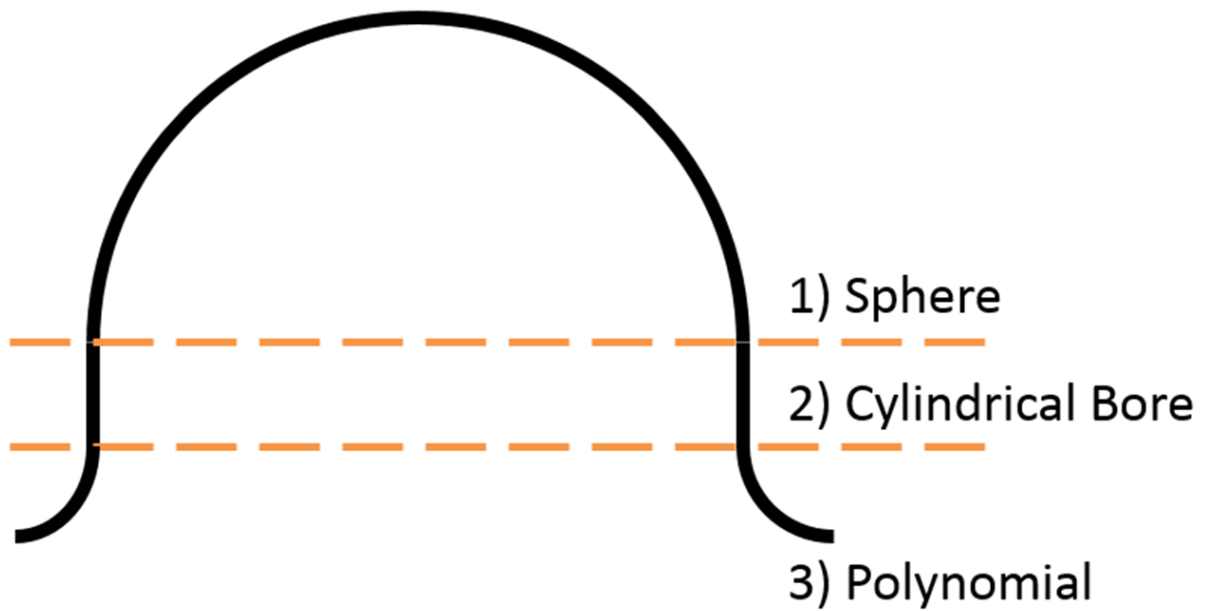


Figure 43: Neutral acetabular liner region definition.

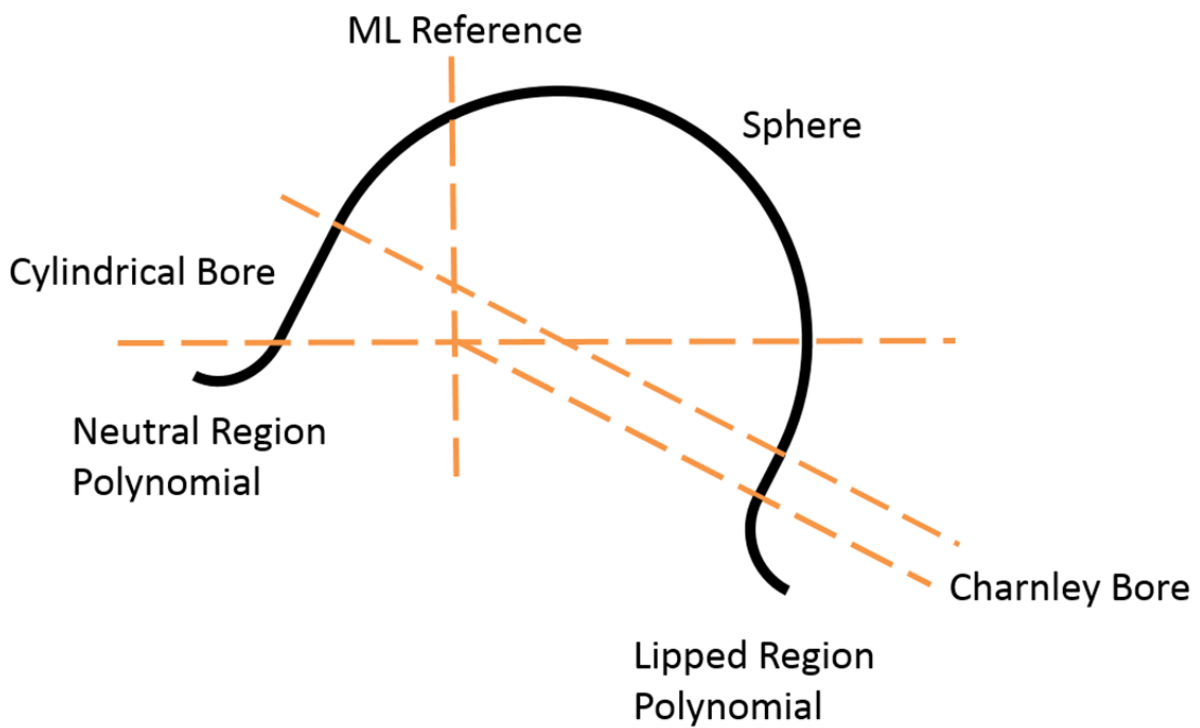
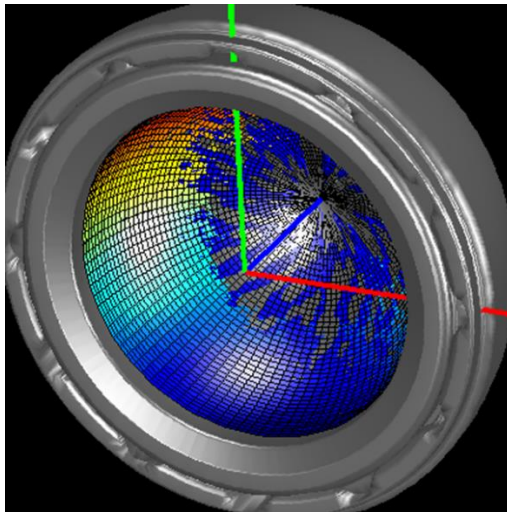
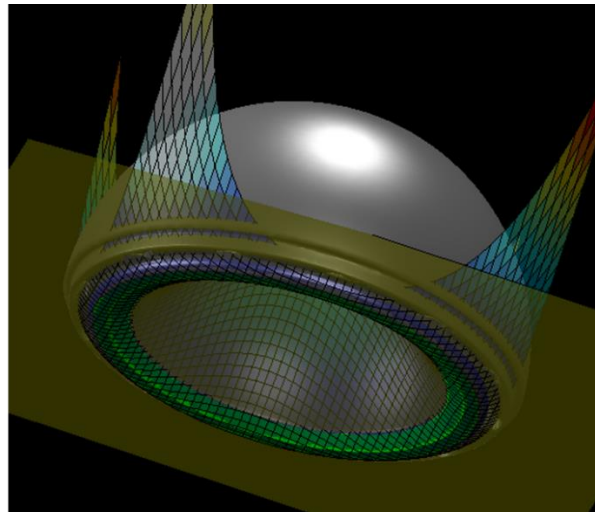


Figure 44: Lipped acetabular liner region definition.



(a)



(b)

Figure 45: (a) spherical liner region fitted with a sphere, and (b) liner rim fitted with a polynomial.

$$Y1 > F(X1) > Y2$$

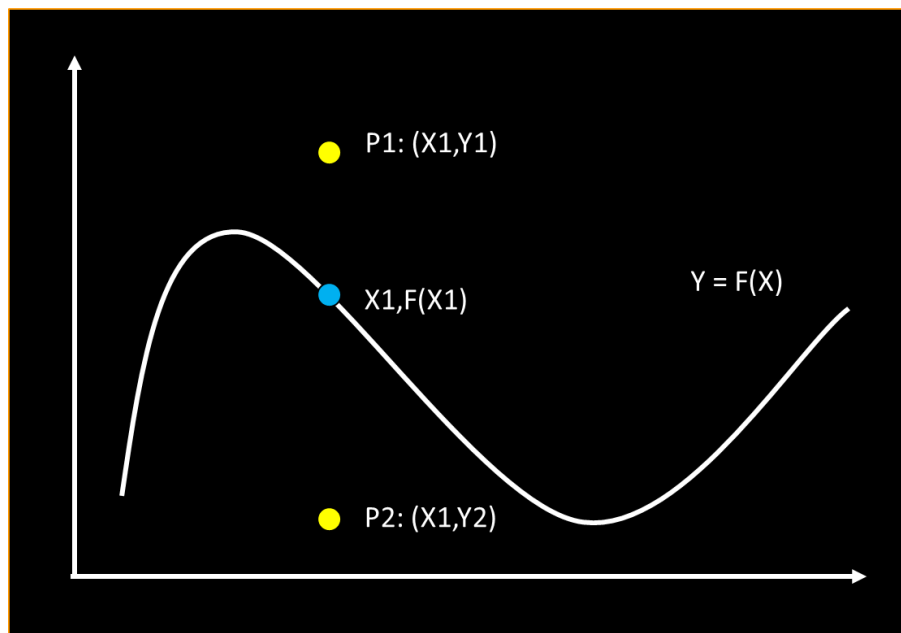


Figure 46: Contact detection explanation using 2 points and a 2-dimensional polynomial of $Y=F(X)$.

expanded into 3-dimensional surfaces in either spherical, cylindrical, or Cartesian coordinates, as mentioned above.

Using this method, the “Rigid Body Overlap” of the femoral component into the acetabular cup can be computed, and a hip force is computed based on the predicted contact penetration between the two components (Figure 47), and this hip force is only applied if the model detects contact between the head and the cup. This hip force is normalized over the computed contact area and applied as a factor of calculated penetration.

This contact detection algorithm can be used to predict and track where the force is being applied on the femoral head and acetabular cup throughout the duration of the activity and how the force is being distributed. This methodology can be further used to aid in predicting component wear patterns and forecast the theoretical longevity of the device.

4.2 Project-Specific Modeling Methods

The above methods sections outline general modeling concepts incorporated into the models used in this dissertation. These concepts hold consistent for all versions of the model. However, as is the case with any form of mathematical modeling, it is often necessary to build models that are specific to the task at hand. In other words, one single version of the forward solution hip model will not suffice to conduct all desired analyses. Thus, the model consists of a closed-chain version, allowing for an analysis of weight bearing activities such as stance phase and deep knee bend, and an open-chain version, allowing for an analysis of non-weight bearing activities such as swing phase. In addition,

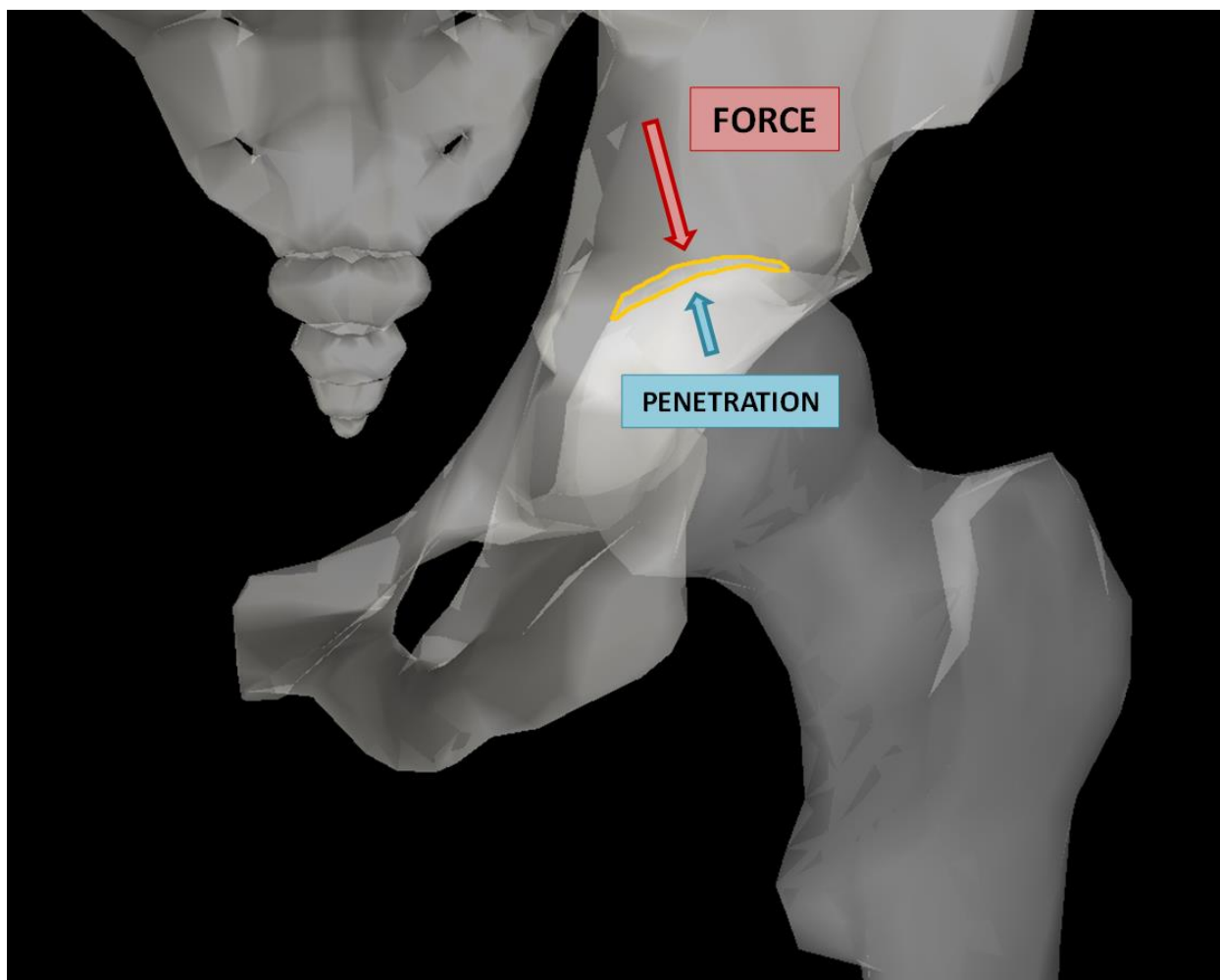


Figure 47: Penetration and applied force representation, exaggerated for clarity.

the model consists of a fixed bearing version, only including six total bodies, and a dual mobility version, which includes a seventh body to represent the dual mobility bearing. Thus, combining swing and stance phase models with fixed and dual mobility models yields a total of four mathematical models that have been developed herein.

4.2.1 Multiple Activities – Swing and Stance Phase

To mathematically model a closed chain system, where the distal segment is fixed to the ground [8] (such as a planted foot during weight bearing stance phase), the model was built in a “ground-up” fashion. More specifically, passing of the U's begins at the interface between the foot and the ground and travel upwards, transferring the GRF and other applicable forces through the system at each joint interface.

Conversely, to model an open chain system, where the distal segment is not fixed to the ground [8] (such as a free-swinging foot during non-weight bearing swing phase), the model was built in a “top-down” fashion. In this case, the U's are passed into the system at the contralateral leg force point, because the contralateral leg is instead fixed to the ground. The U's are then passed downwards through the lower extremity into the foot as well as upwards into the HAT. This approach achieves a similar principle to the stance phase model, passing the forces through the system starting with the fixed segment, i.e. the contralateral leg. Figure 48 demonstrates the direction of flow of the U's to accomplish the desired model.

In general, during stance phase, the model predicts maximum compressive hip forces ranging between 1.4-3.0xBW throughout gait. Figure 49 shows a six-frame representation of the stance phase of gait and Figure 50 shows the corresponding muscle

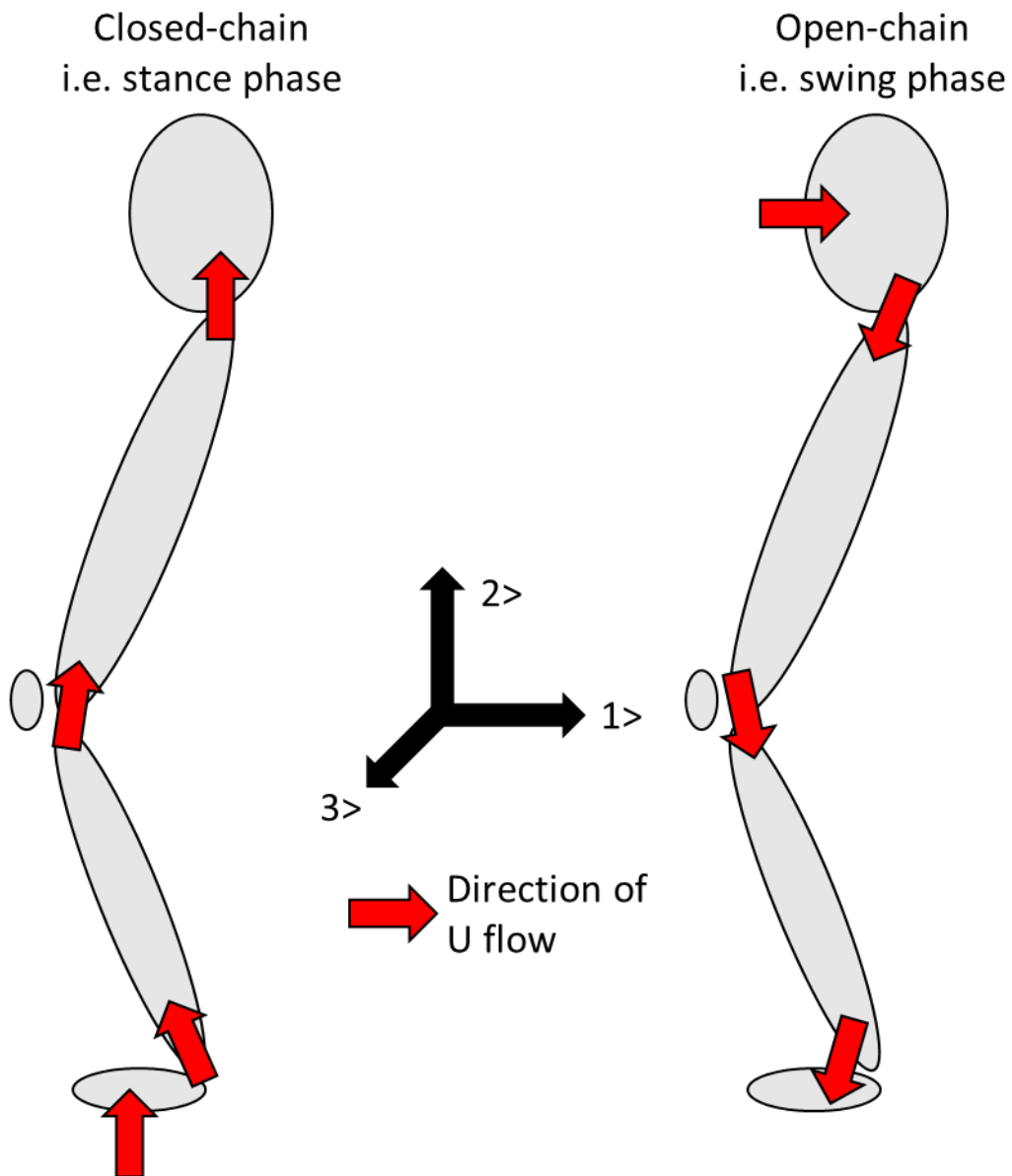


Figure 48: Direction of flow of the U's depending on stance or swing phase.

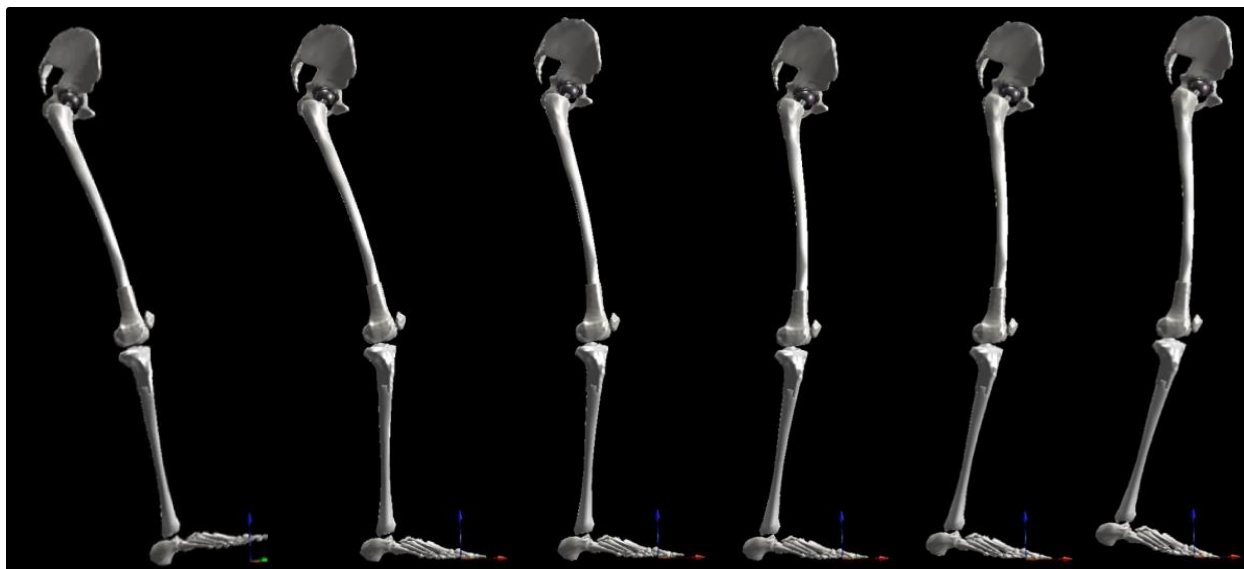


Figure 49: Six-frame stance phase representation.

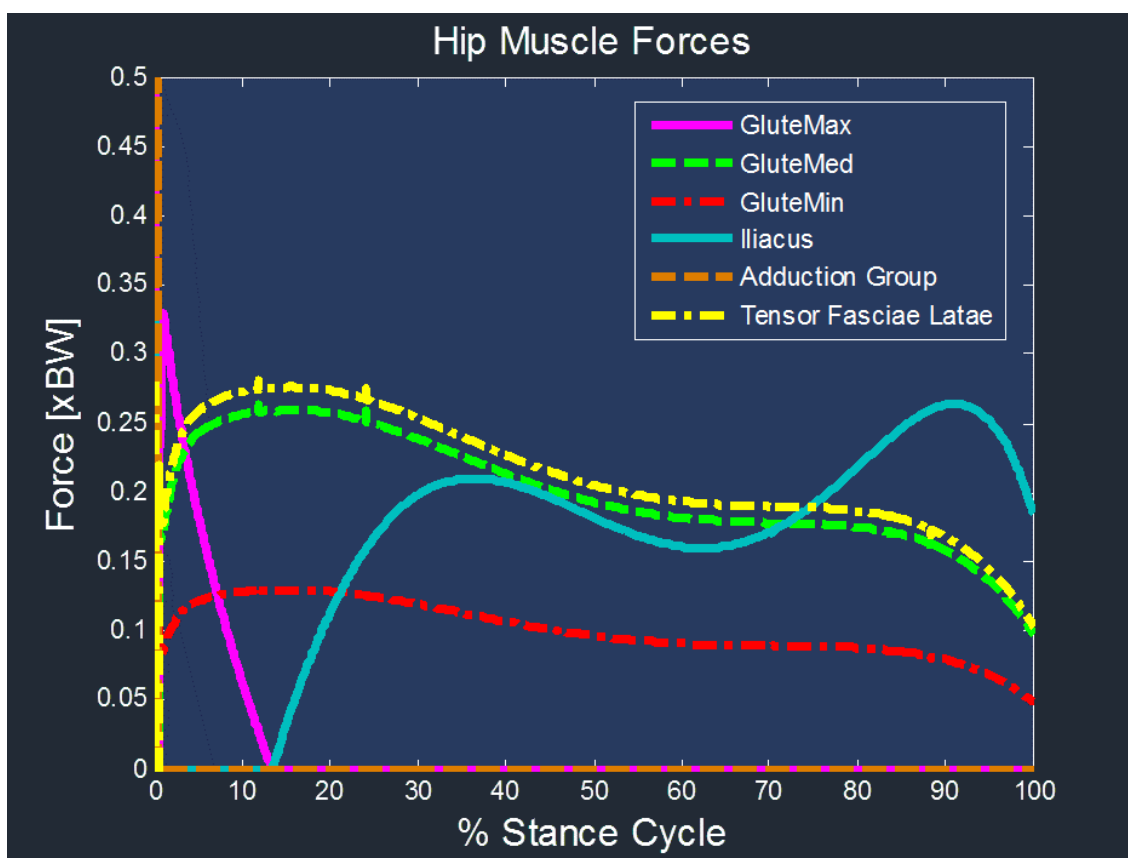


Figure 50: Sample stance phase muscle forces.

forces required to produce the motion. It is clear from Figure 51 that the PID controllers are successfully controlling the motion of the pelvis throughout stance phase. Figure 52 shows a sample simulation with a maximum hip force of $1.4 \times BW$ and a maximum knee force of $1.6 \times BW$.

In general, during swing phase, the model predicts less than $1.0 \times BW$ of maximum compressive hip forces throughout swing phase. Figure 53 shows a six-frame representation of the swing phase of gait and Figure 54 shows the corresponding muscle forces required to produce the motion. It is clear from Figure 55 that the PID controllers are successfully controlling the motion of the hip joint. Figure 56 shows a sample simulation with a maximum hip force of $0.4 \times BW$ and a maximum knee force of $0.2 \times BW$.

4.2.2 Multiple Components – Fixed Bearings and Dual Mobility Bearings

To mathematically model dual mobility cups, another version of the model was built that introduced a dual mobility bearing as a new body. The passing of the U's remains the same as before, depending on swing versus stance phase, except that there are six extra U's throughout the system.

The rotations of the dual mobility bearing were completely unconstrained and remained as such. No form of PID controller was present. Instead, the bearing was allowed to freely rotate within the confines of the cup based on kinematic patterns, friction forces, joint reaction forces, and rim/neck impingement.

To control the translational motion of the bearing, two separate contact detection algorithms are present. First, there is an algorithm between the femoral component and

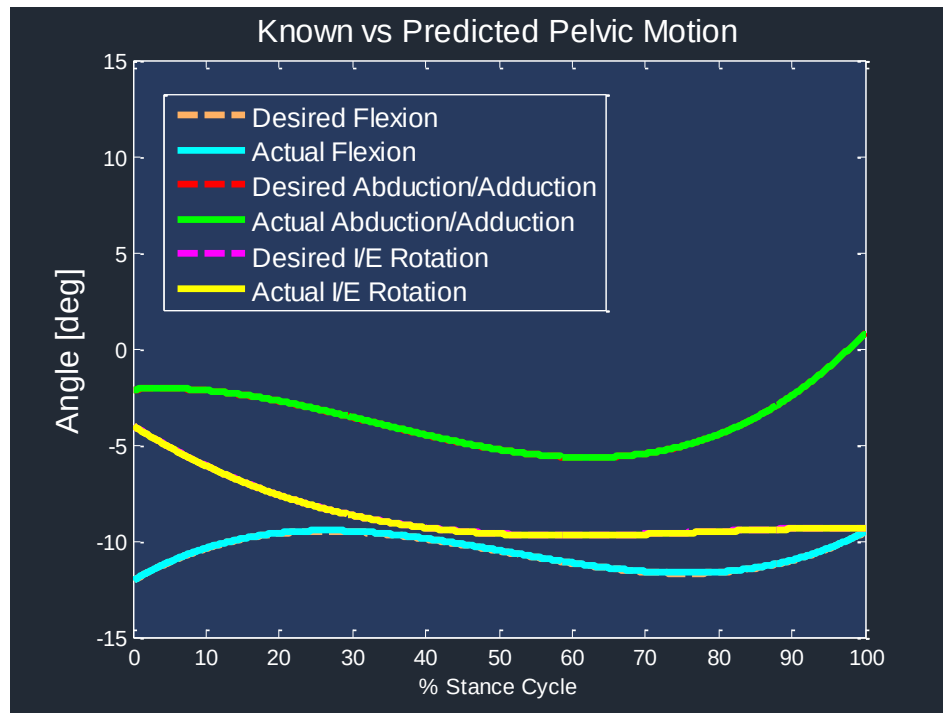


Figure 51: Stance phase rotational PID control for a sample kinematic profile.

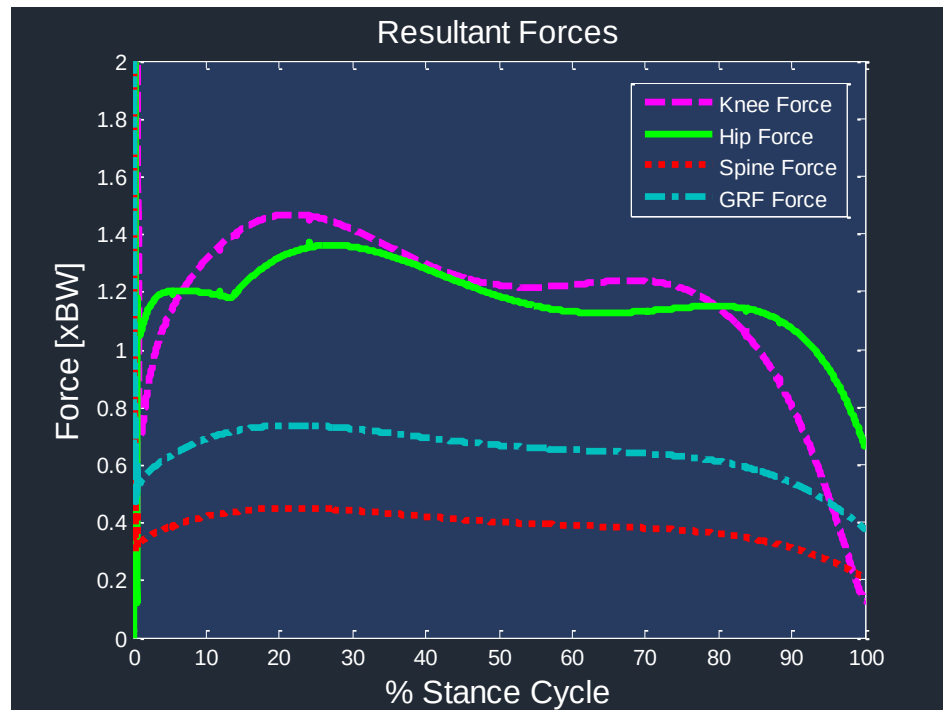


Figure 52: Sample resultant stance phase forces.



Figure 53: Six-frame swing phase representation.

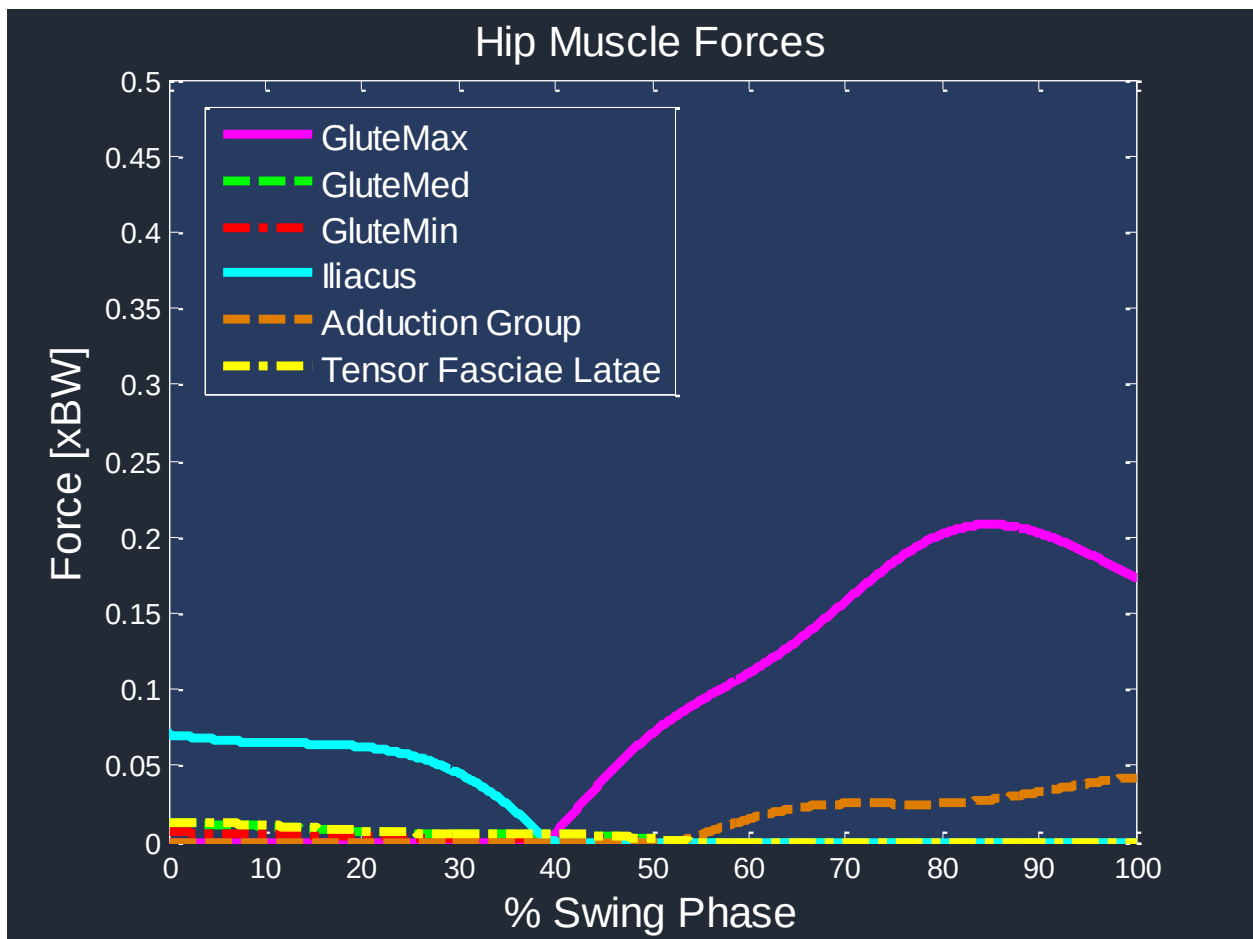


Figure 54: Sample swing phase muscle forces.

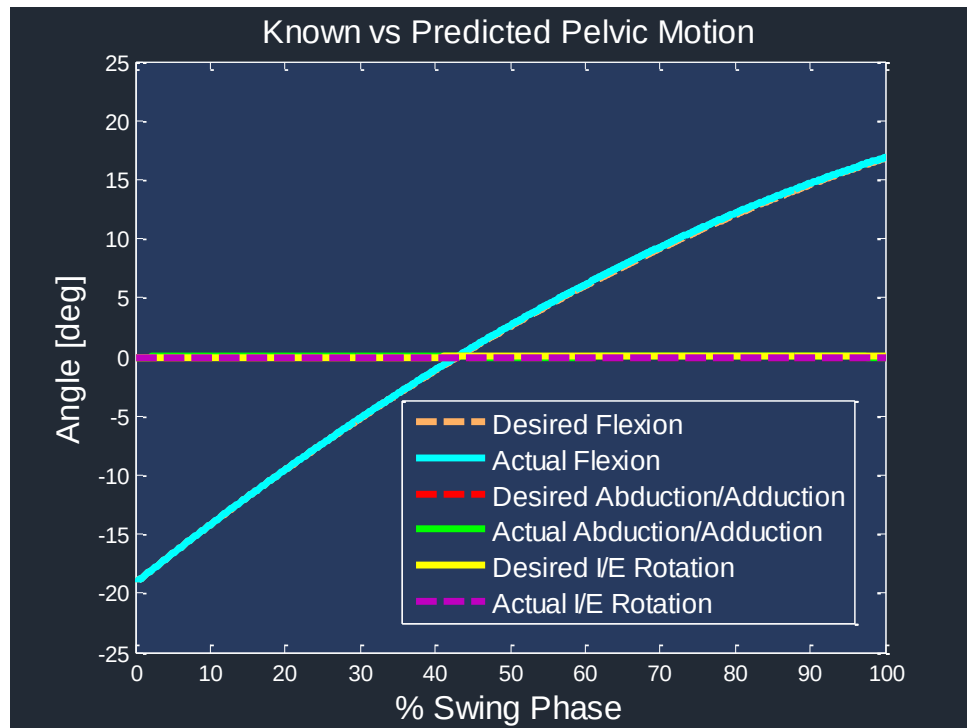


Figure 55: Swing phase rotational PID control for a sample kinematic profile.



Figure 56: Sample resultant swing phase forces.

the bearing. For this algorithm, the translations between the head and the bearing are constrained to zero, because the bearing is a constrained bearing. This assumption saves computational speed and improves model stability. However, there is still a contact detection algorithm present between the head/neck (modeled as a point cloud, as shown previously in Figure 42) and the bearing (modeled using a liner polynomial, shown in Figure 57). This contact detection algorithm allows for impingement between the femoral neck and the bearing rim, to ensure accurate bearing rotations. Second, the traditional contact detection algorithm is present between the outside of the bearing and the internal surface of the cup. In this case, the outside of the liner is modeled identically to the femoral head (i.e. a point cloud, shown in Figure 58) and the internal surface of the cup is modeled as normal, with a sphere and a polynomial, in accordance with Figure 45. The model inherently assumes that the starting position of the bearing is the same as the orientation of the cup, but this assumption can be modified.

4.2.3 Hip Separation

Depending on the activity, hip separation has been successfully induced in a variety of ways, including soft tissue modifications and component malalignments. During stance phase, separation is generated in two ways: via impingement of the femoral neck on the acetabular rim and via surgical cup translational misalignment (i.e. medial/superior shifting). During swing phase, separation is primarily generated via translational misalignment that yields slack in the capsular ligaments. The figures referred to herein

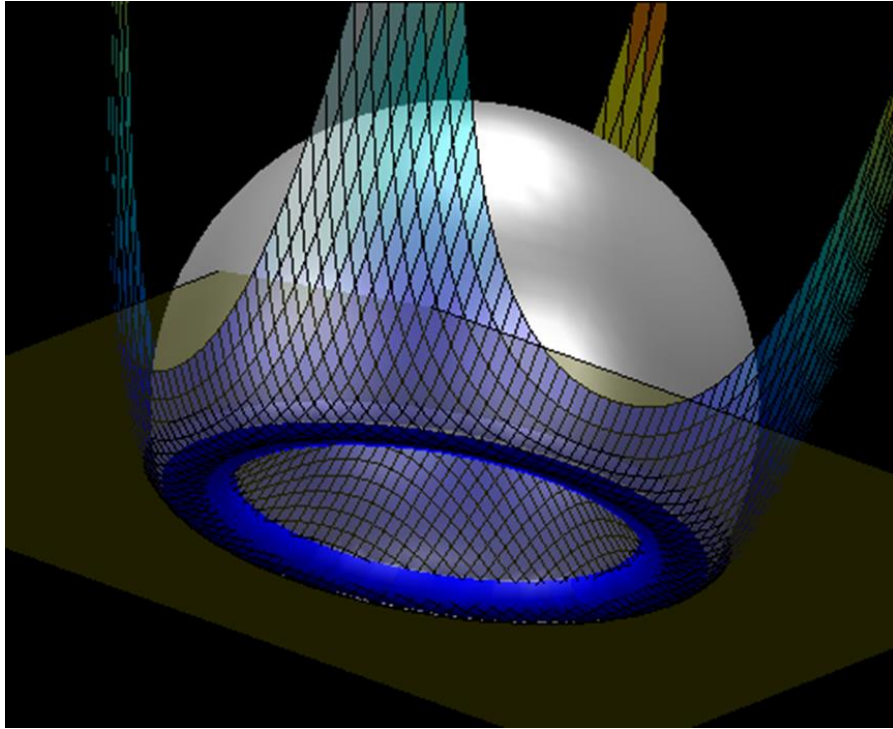


Figure 57: Polynomial fit for the lipped portion of the dual mobility bearing.

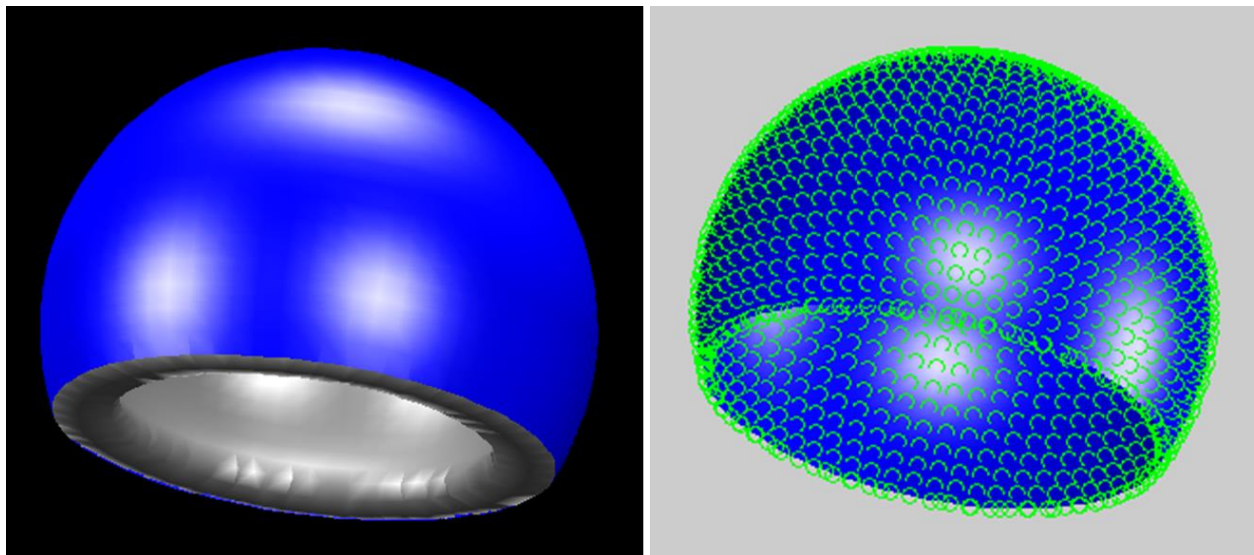
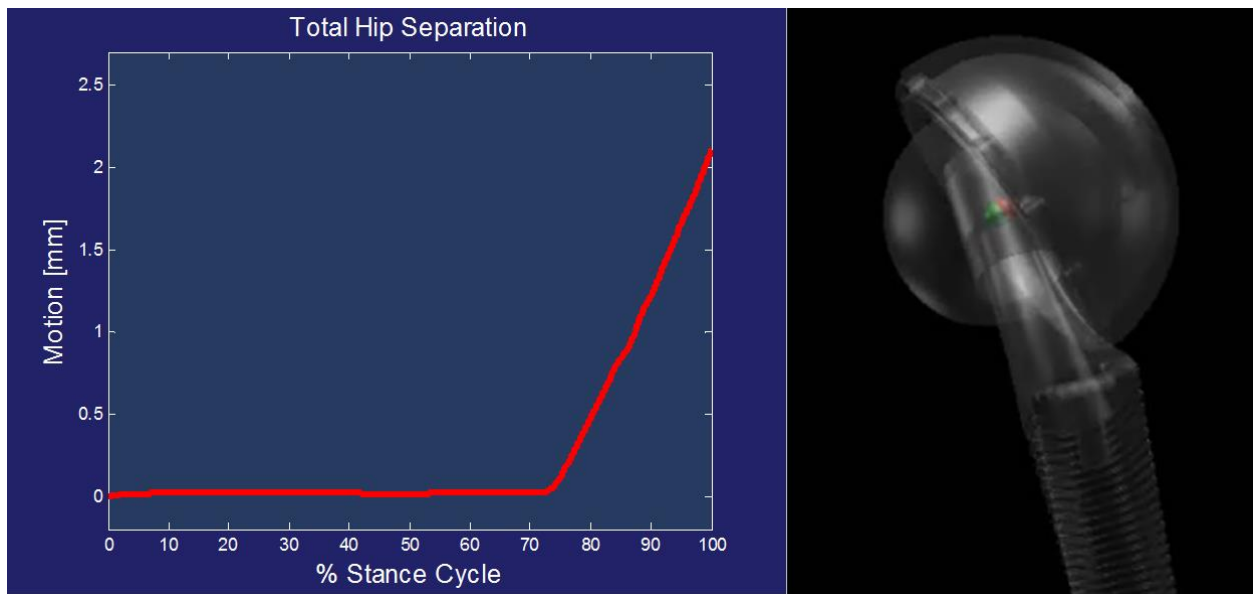


Figure 58: Point cloud for the outer surface of the dual mobility bearing.

show sample scenarios of hip separation. More detailed analyses of hip separation can be found in the results section.

Impingement during stance phase is generally achieved by a lever-torque that occurs when the femur neck points contact the liner rim polynomial (generally due to excessive misalignment of the acetabular cup), thus causing the femoral head to lever out of the liner socket. The occurrence and severity of impingement is patient-specific, depending on individual patient kinematics such as stride length, pelvic rotation, etc. Figure 59 shows approximately 2.1mm of hip separation generated via impingement.

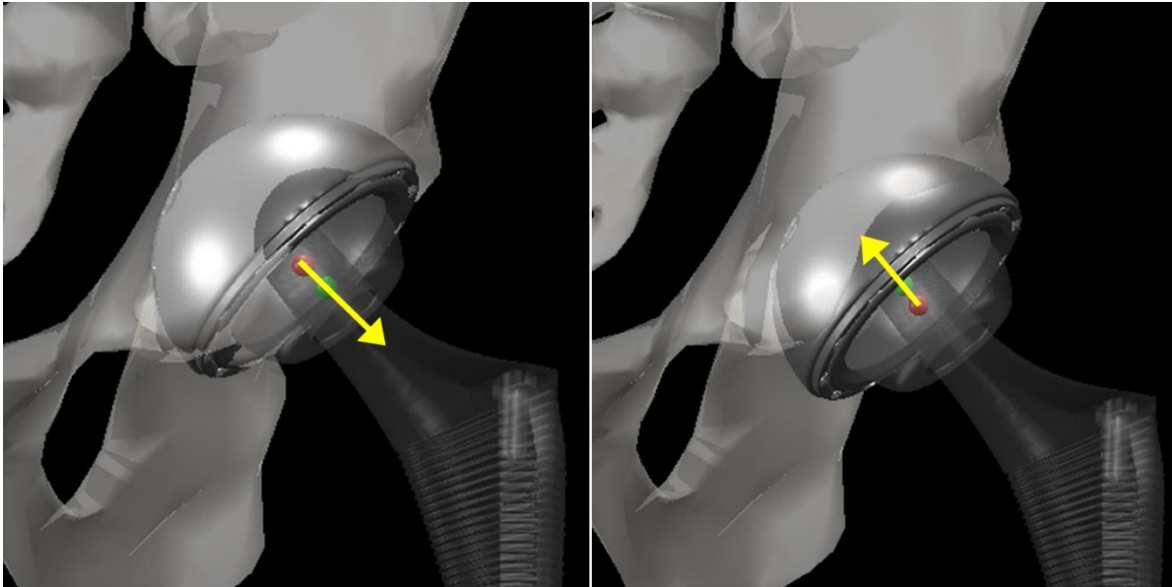
Surgical misalignment separation, occurring during stance phase, is achieved under the proprioceptive hypothesis that the human hip “wants” to rotate about its anatomical center. Thus, if the acetabular cup is medialized (for example), the model attempts to “pull” the current model center back to the anatomical center. This pull force is achieved by introducing a spring-damper force between the component center and the anatomical center, thereby pulling the femoral head center towards the cup anatomical center, potentially yielding hip separation. Figure 60 shows an explanation of the direction of the proprioceptive separation force. The red circle indicates the acetabular component center, and the green circle indicates the anatomical center. In Figure 60a, the acetabular cup has been shifted medially and superiorly. Thus, the proprioceptive force attempts to pull the femoral head laterally and inferiorly, out of the hip socket, generating hip separation. Conversely, in Figure 60b, the acetabular cup has been shifted laterally and inferiorly. Thus, the proprioceptive force attempts to pull the femoral head medially and superiorly, which would not generate separation. Figure 61 shows a sample separation



(a)

(b)

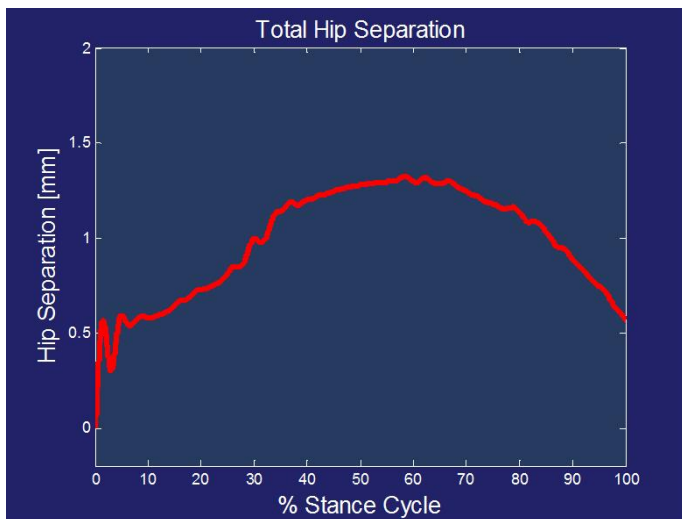
Figure 59: (a) graphical and (b) visual representation of stance phase separation caused by femoral neck impingement.



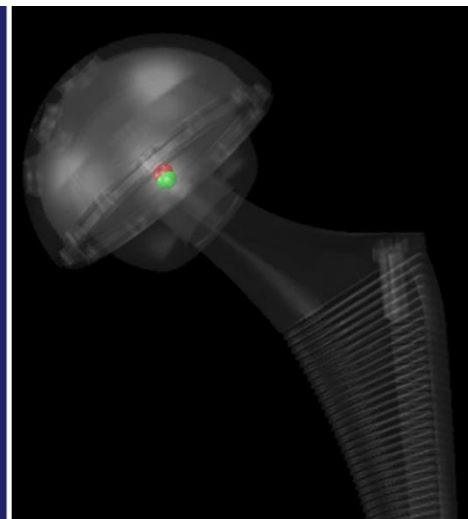
(a)

(b)

Figure 60: Proprioceptive separation force direction given (a) medial and superior shifting of the cup and (b) lateral and inferior shifting of the cup. The red dot represents the cup component rotation center and the green dot represents the anatomical center.



(a)



(b)

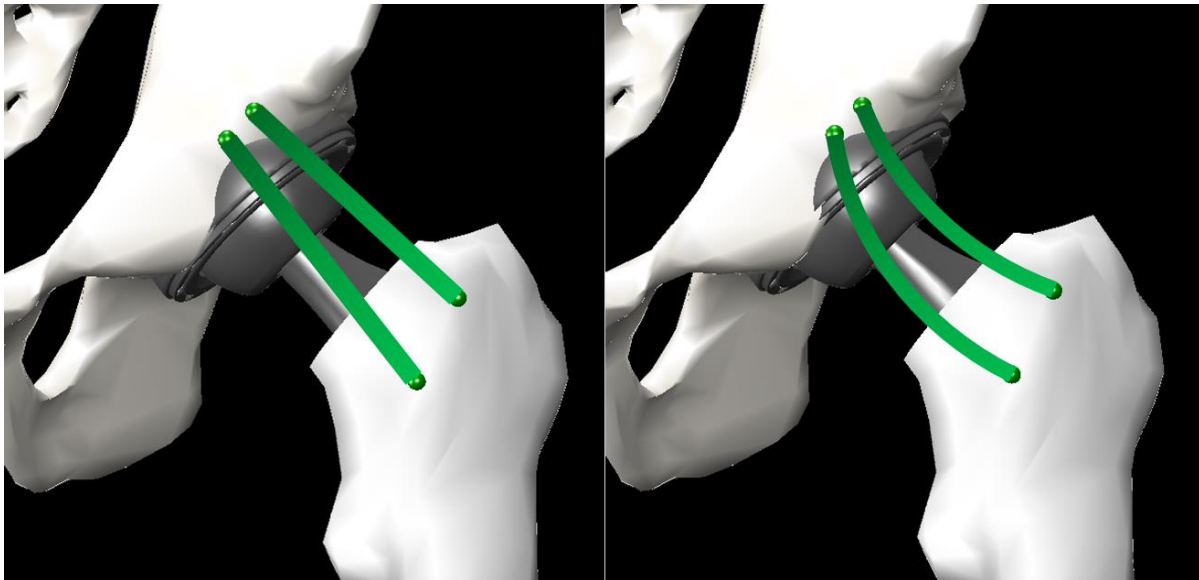
Figure 61: (a) graphical and (b) visual representation of stance phase separation caused by cup medialization.

curve generated by cup medialization. In cases such as this, hip separation generally appears as more of a gradual sliding and edge-loading motion.

Hip separation during swing phase is generated primarily by shifting one component without properly compensating by shifting the other, thus creating loose or improperly balanced capsular ligaments. Figure 62a shows a properly aligned acetabular component with adequate ligament tension, and Figure 62b shows an exaggerated representation of a medialized acetabular component with lesser capsular ligament tension. Logically, the greater the surgical error, the looser the capsular ligaments and thus the greater the hip separation. However, there are certain cases (smaller shifts/errors) where the capsular ligaments, although looser, remain tight enough to keep the femoral head positioned within the cup. Figure 63 shows a sample separation curve generated by cup medialization. In cases such as this, hip separation generally appears as more of a pistoning motion.

4.2.4 Creation of Multiple Patients

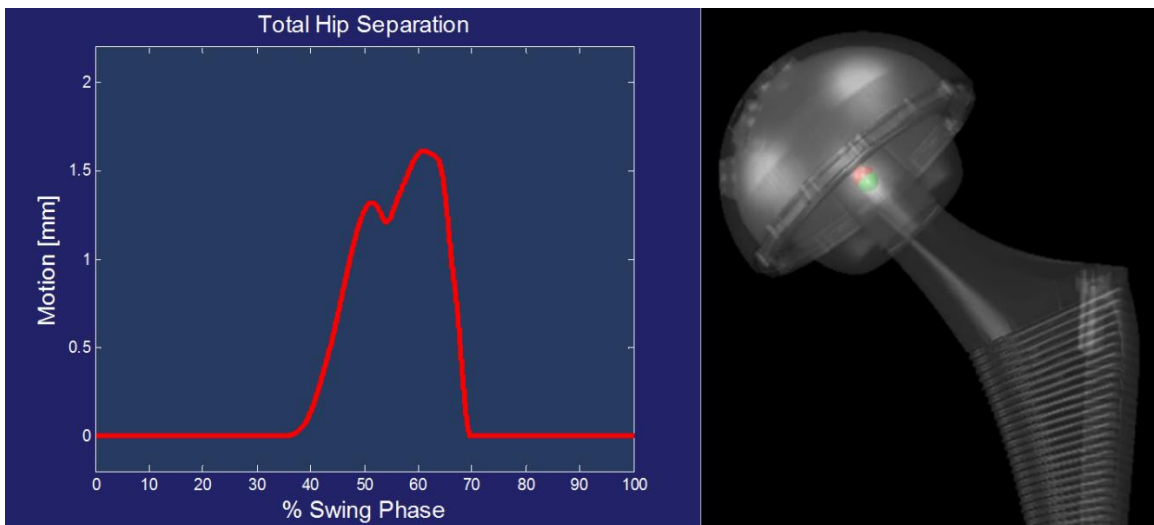
To conduct in-depth analyses, four virtual patients were “created” based on the kinematics extracted from an in-house fluoroscopy study. For stance phase, the patients were taken directly from the fluoroscopy study, chosen to represent diverse kinematic patterns. The in-house fluoroscopy study did not analyze swing phase; thus, for swing phase, the patient profiles were “built” based on simple, accepted kinematic profiles of the hip joint. These profiles were modified to generate patients with longer and shorter stride lengths as well as to generate a patient who walks with an externally rotated foot.



(a)

(b)

Figure 62: An exaggerated representation of hip capsular ligament tension given (a) a properly aligned acetabular cup and (b) a medialized acetabular cup without proper femoral component compensation.



(a)

(b)

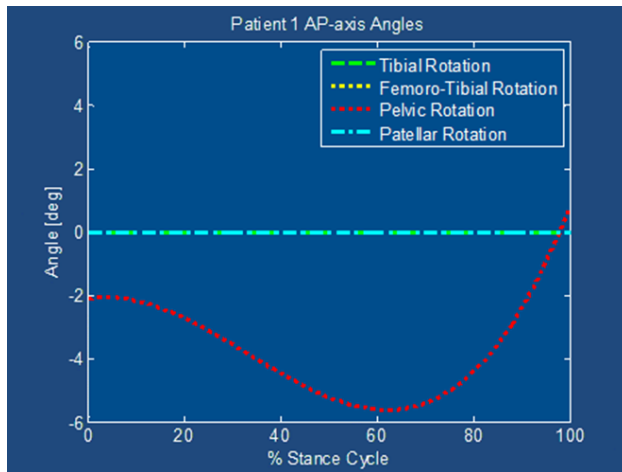
Figure 63: (a) graphical and (b) visual representation of swing phase separation caused by cup medialization.

These virtual patients were frequently used to compare how kinematic diversity affects the resulting mechanics under similar surgical scenarios. It must be noted that these four patients were not used for all simulations presented in the results section. In many cases, only one of the patients was used. For the analyses where all four patients were used, the results section makes this clear. Similarly, for analyses where there is not a four-patient comparison, Patient 1 was generally used. The kinematics for all four patients during stance phase are shown in Figure 64, Figure 65, and Figure 66, and the kinematics for all four patients during swing phase are shown in Figure 67, Figure 68, and Figure 69.

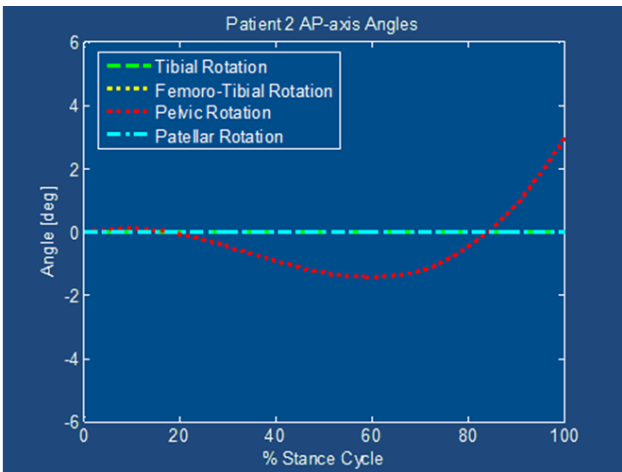
4.3 Validation

The model was validated against telemetric data provided by the OrthoLoad Hip98 data set (orthoload.com [91]). The patient that was chosen for analysis was patient PFL, trial 3, under normal walking conditions [91]. It should be noted that Patient PFL took approximately 1.22 seconds to complete the entire gait cycle. The model's gait cycle is normalized to 1 second, and hence Patient PFL's activity was also normalized to 1 second. This could potentially yield inaccuracies in the model.

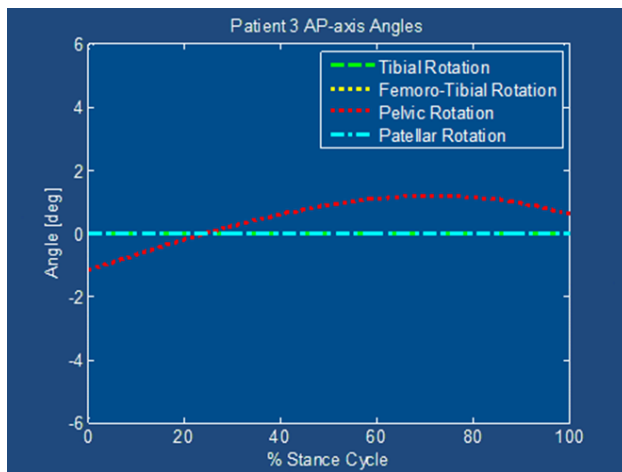
Figure 70 shows the stance phase femoro-pelvic angles that were gathered experimentally versus what was produced in the model, and Figure 71 shows the femoro-pelvic angle comparisons for swing phase. During stance phase, the model closely matched the predicted rotations to the experimental rotations, with the largest sources of error being the AP> and SI> angles. During swing phase, the model struggled to accurately predict the AP> angles. While the maximum abduction angle was similar, the



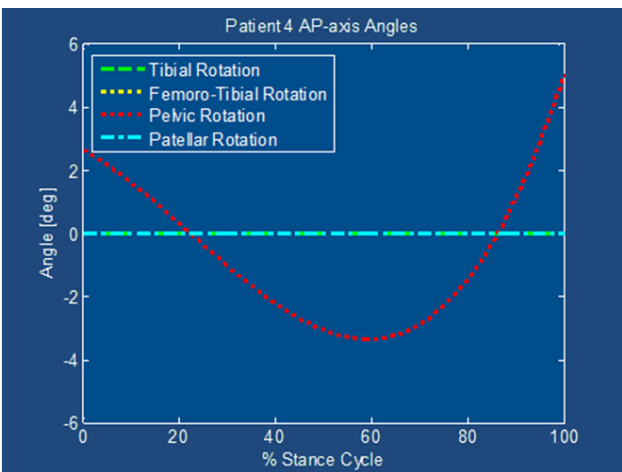
(a)



(b)

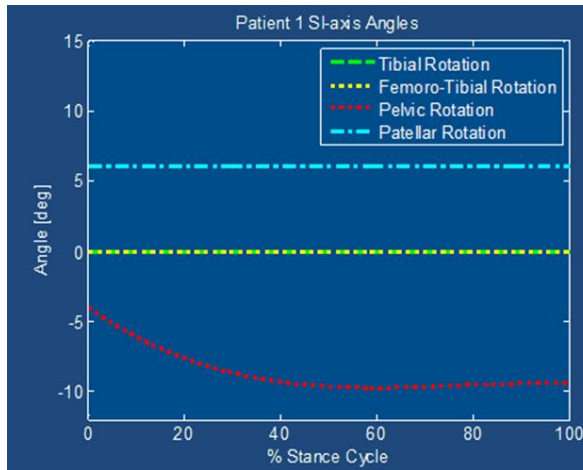


(c)

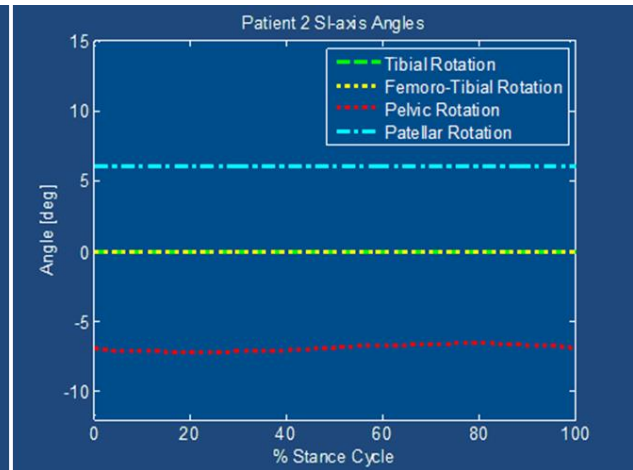


(d)

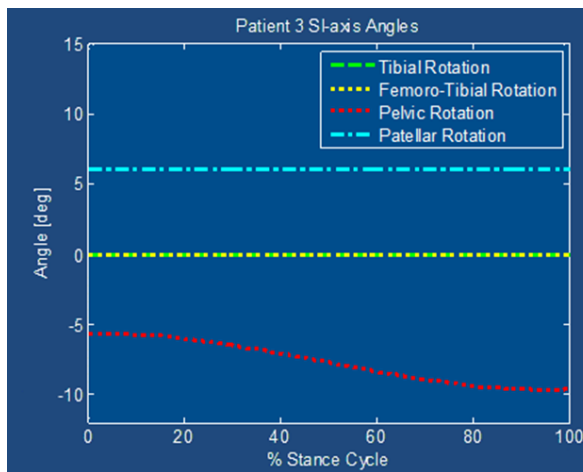
Figure 64: A/P axis stance phase angles for Patients (a) 1, (b) 2, (c) 3, and (d) 4.



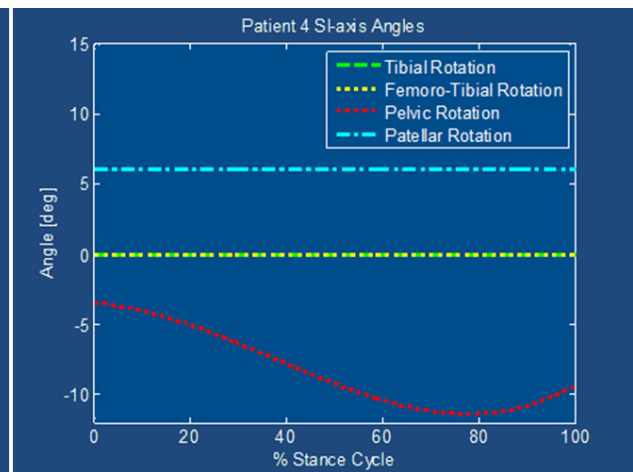
(a)



(b)

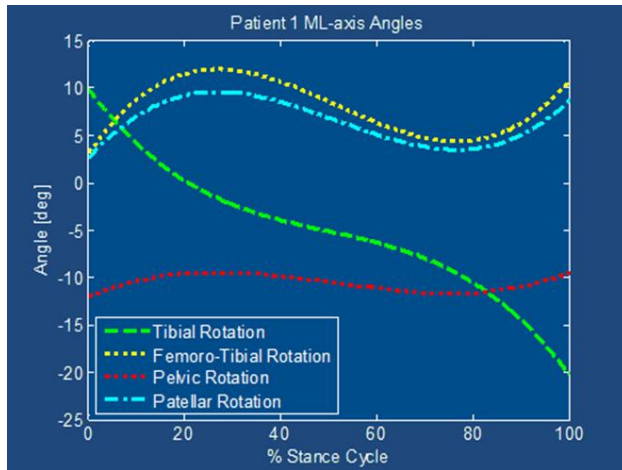


(c)

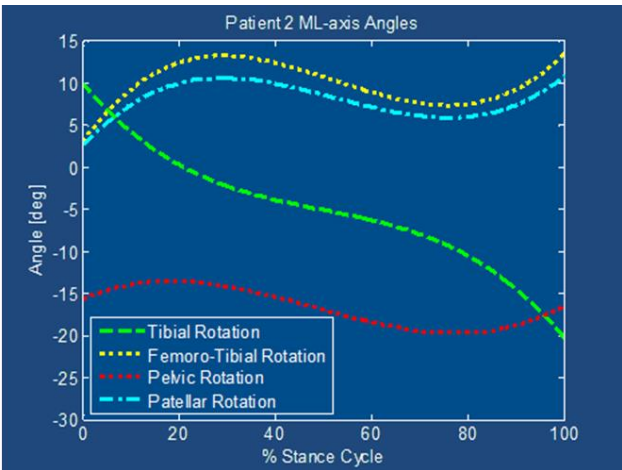


(d)

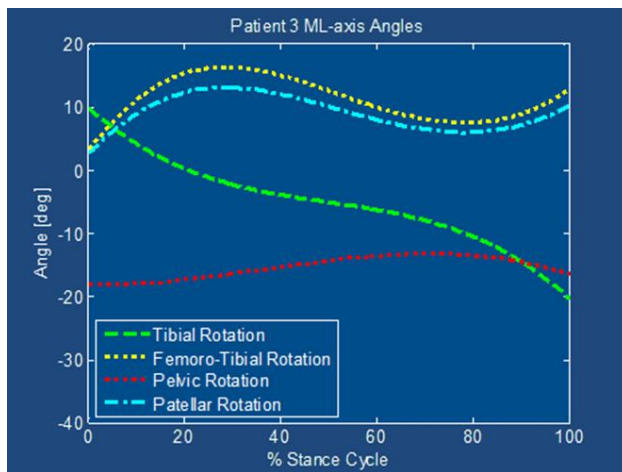
Figure 65: S/I axis stance phase angles for Patients (a) 1, (b) 2, (c) 3, and (d) 4.



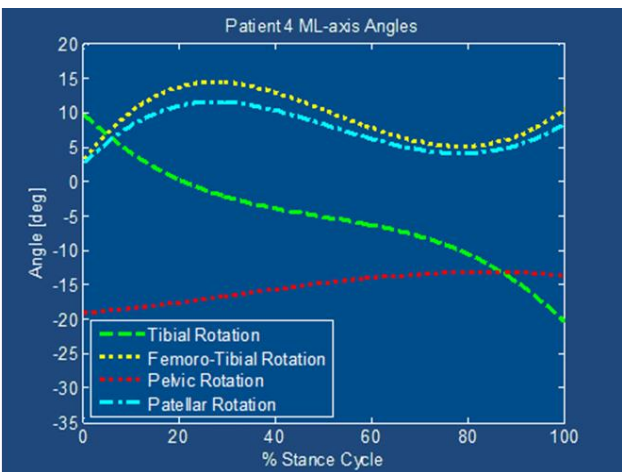
(a)



(b)

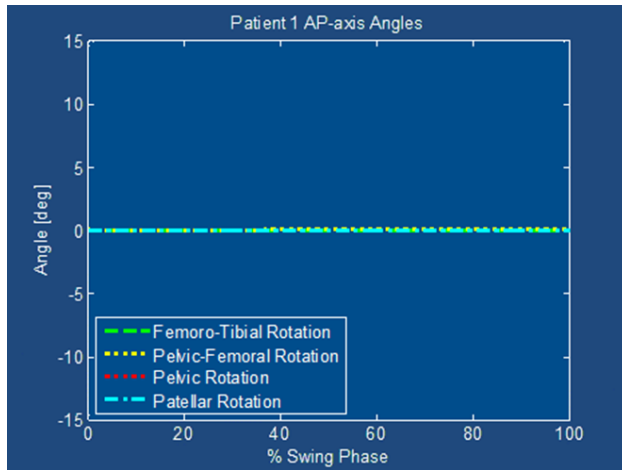


(c)

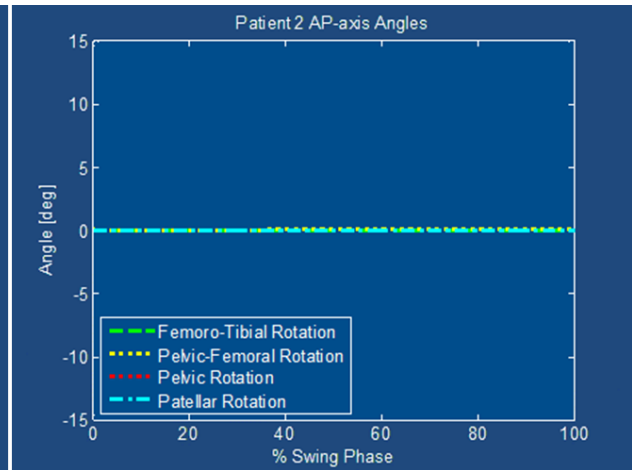


(d)

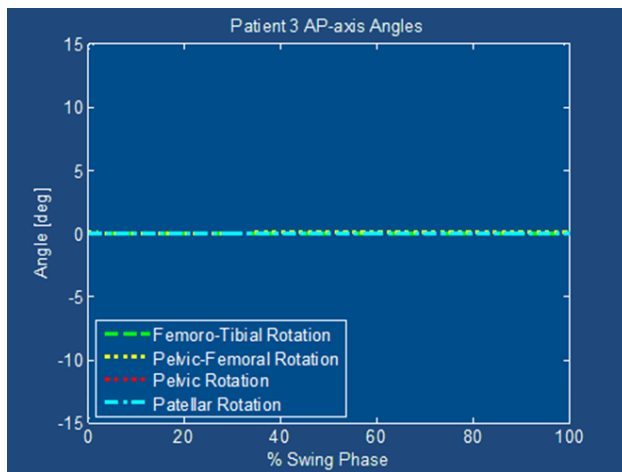
Figure 66: M/L axis stance phase angles for Patients (a) 1, (b) 2, (c) 3, and (d) 4.



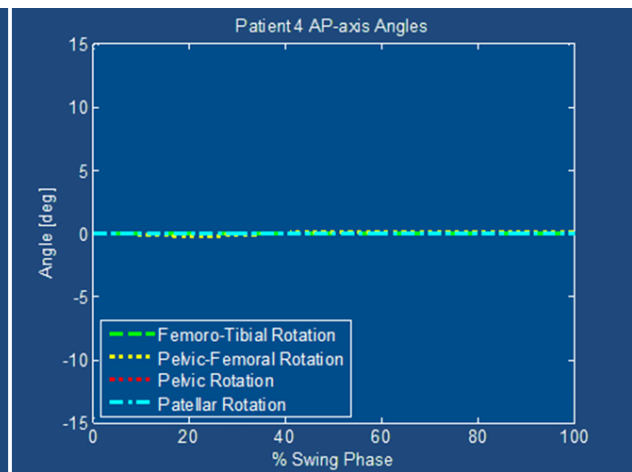
(a)



(b)

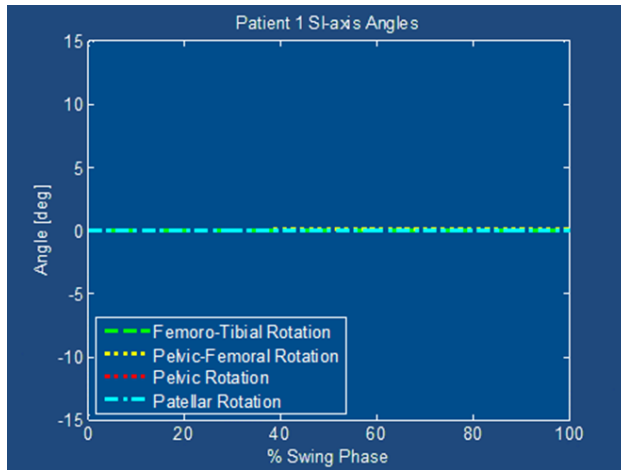


(c)

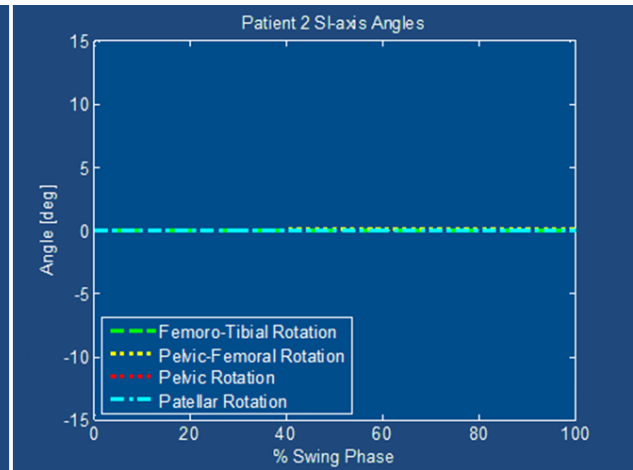


(d)

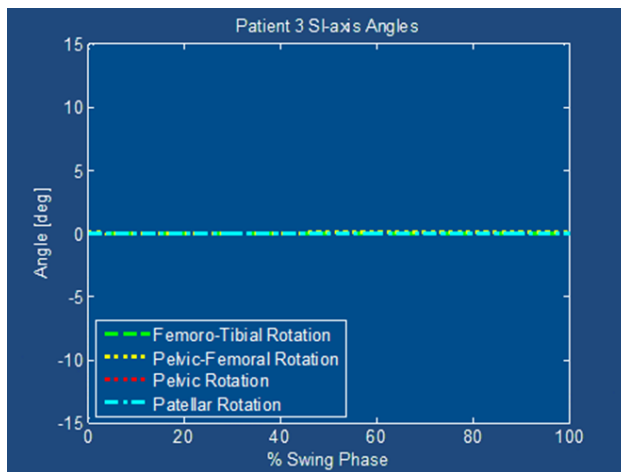
Figure 67: A/P axis swing phase angles for Patients (a) 1, (b) 2, (c) 3, and (d) 4.



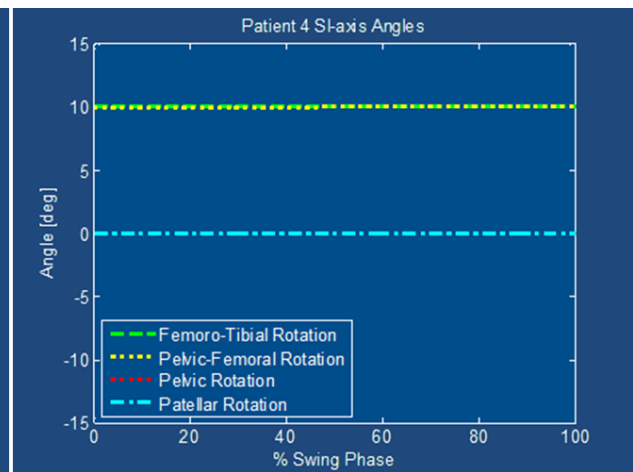
(a)



(b)

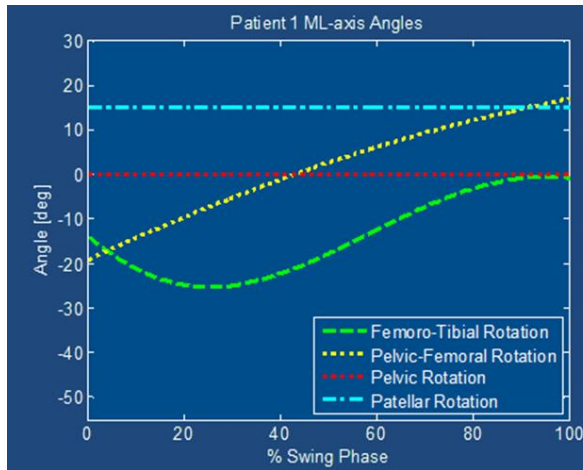


(c)

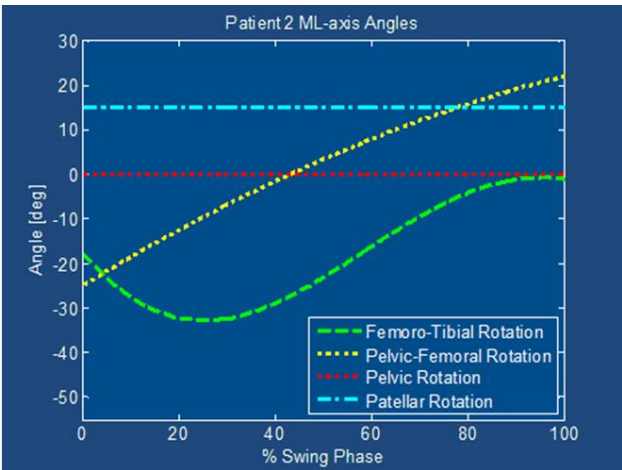


(d)

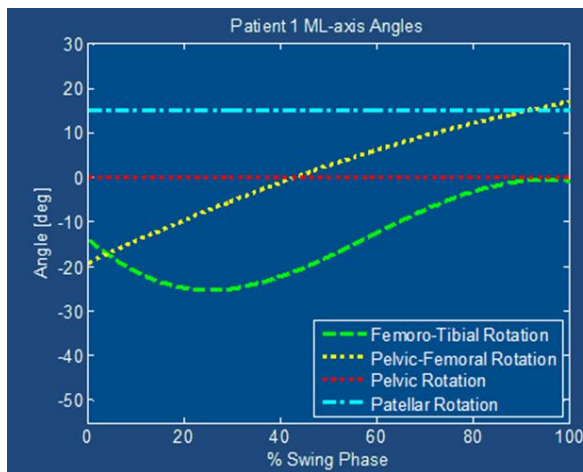
Figure 68: S/I axis swing phase angles for Patients (a) 1, (b) 2, (c) 3, and (d) 4.



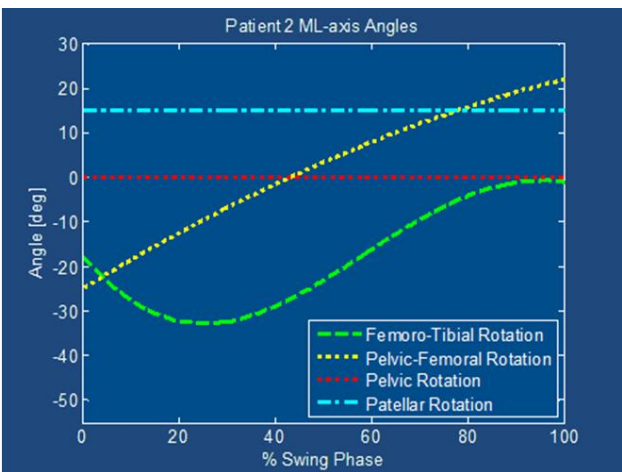
(a)



(b)

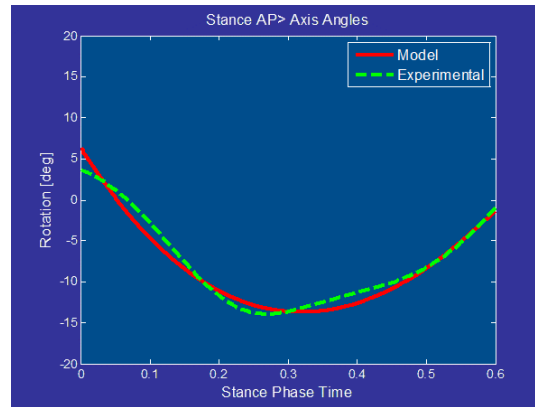


(c)

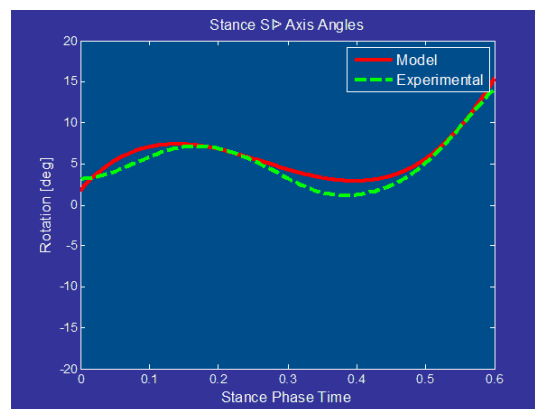


(d)

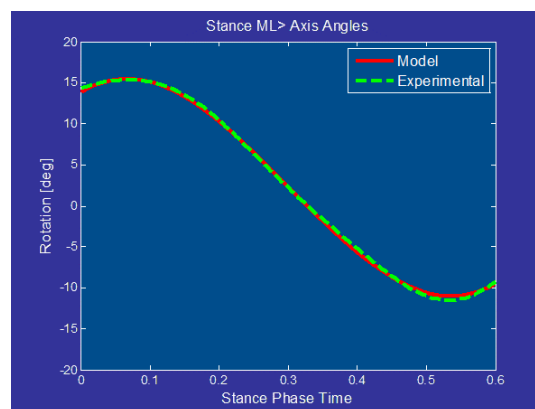
Figure 69: M/L axis swing phase angles for Patients (a) 1, (b) 2, (c) 3, and (d) 4.



(a)

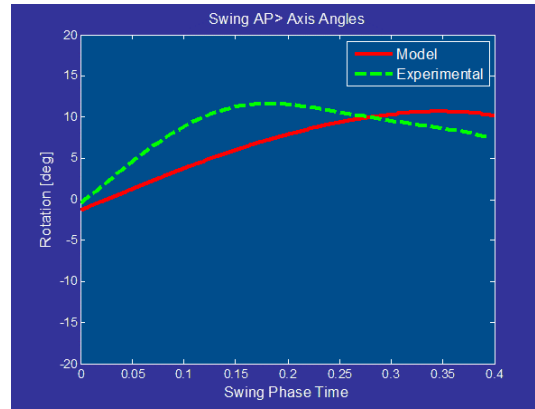


(b)

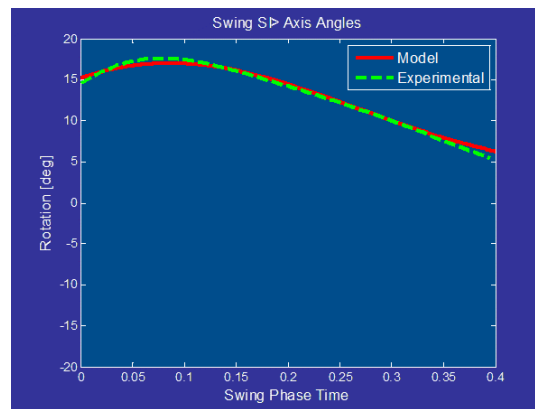


(c)

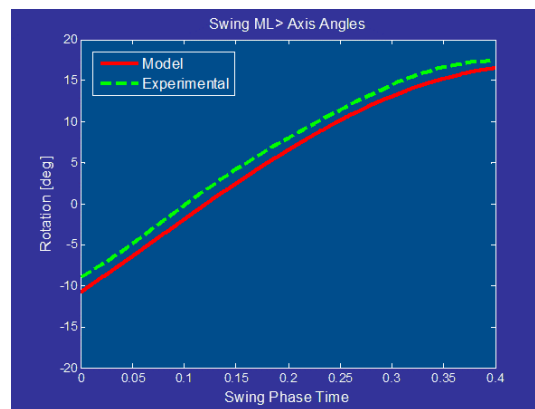
Figure 70: Experimental telemetry versus model-predicted kinematics for stance phase in the (a) A/P, (b) S/I, and (c) M/L directions.



(a)



(b)



(c)

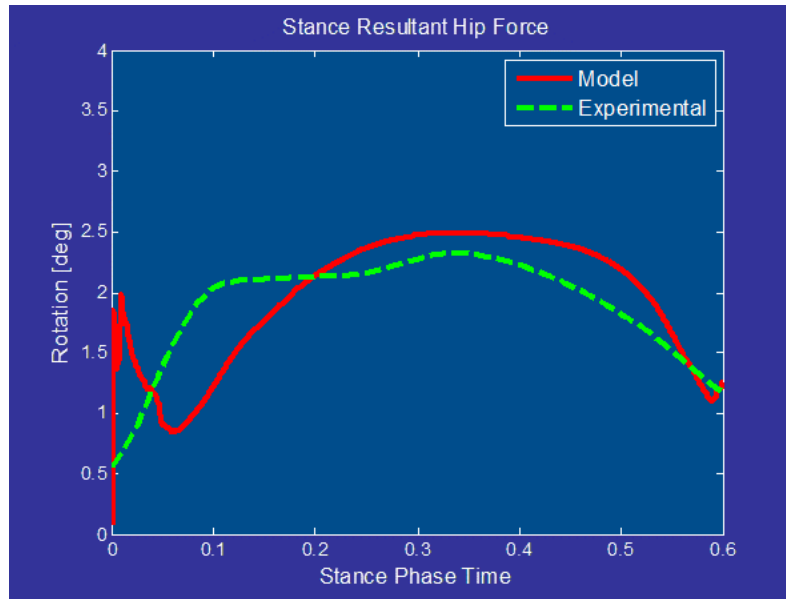
Figure 71: Experimental telemetry versus model-predicted kinematics for swing phase in the (a) A/P, (b) S/I, and (c) M/L directions.

model predicted that this angle was reached later in the activity. The resultant force comparisons can be found in Figure 72. For both stance and swing phase, the model showed moderate agreement with telemetry.

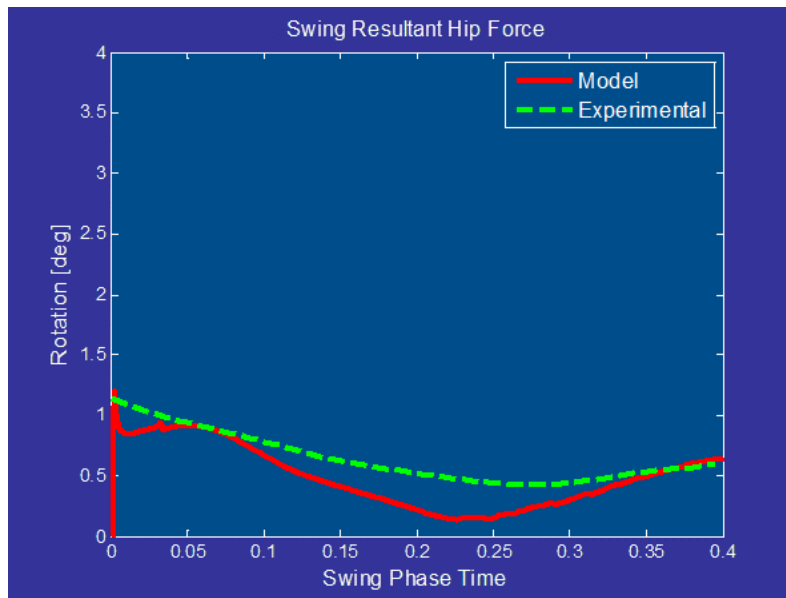
At the beginning of stance phase, the model's initial settling period is clearly visible, yielding inaccuracy compared to telemetry. Additionally, the model predicted that the rise in the resultant hip force would occur slightly later than shown in telemetry. In general, the model predicted higher force values during stance phase than determined through telemetry. The maximum resultant hip force predicted by the model was $2.4895 \times BW$, which occurred at 0.336 seconds into stance phase. The maximum resultant hip force predicted by telemetry was $2.3243 \times BW$, which occurred at 0.33 seconds into stance phase. Comparing resultant force maximums, the model over-predicted telemetry by $0.1653 \times BW$, or 7.11%.

During swing phase, the model accurately predicted the resultant forces at the beginning and at the end of swing phase, but during the middle of swing phase the model predicted lower resultant forces. This is possibly due to the lack of co-contraction by opposing muscles in the model. The minimum resultant hip force predicted by the model was $0.1399 \times BW$, which occurred at 0.2268 seconds into swing phase. The minimum resultant hip force predicted by telemetry was $0.4231 \times BW$, which occurred at 0.28 seconds into swing phase. Comparing resultant force minimums, the model under-predicted telemetry by $0.2832 \times BW$, or 66.93%.

During the validation process, the model was found to be particularly sensitive to three parameters: contralateral leg forces, initial positions of the pelvis, and polynomial over-fitting. During stance phase, the contralateral leg forces are provided as an input to



(a)



(b)

Figure 72: Experimental telemetry versus model-predicted resultant hip forces for (a) stance and (b) swing phases.

the model. Hence, inaccurate contralateral leg forces will yield inaccuracies in the model. To the best of my ability, the contralateral leg forces were input to approximately match the hip forces predicted during swing phase. However, the model proved sensitive to fluctuations in the contralateral leg forces, particularly in the S/I direction.

Second, the telemetric inputs were gathered from 3D motion capture marker systems, and hence there is the possibility of skin-mounted errors. The model proved sensitive to the initial sagittal tilt of the pelvis. If the pelvis was tilted too far in either direction, the relationship between the anterior flexion muscles (iliacus, rectus femoris, and TFL) and the posterior extension muscles (gluteus maximus, adductor magnus, and hamstrings) became skewed (i.e. one set of muscles yielded substantially higher forces than the opposite set). It has been shown by the Optimized Ortho group (Walter, Pierrepont, et. al.) that the patient-specific sagittal tilt of the pelvis can vary substantially [92], as much as 58° while standing [93]. Therefore, it was difficult to confidently input femoro-pelvic flexion/extension profiles directly from OrthoLoad. For the validation patient, it was assumed that the pelvis was tilted anteriorly by approximately 15° , as this initial position yielded an even balance between the flexion muscle forces at the beginning of swing phase and the extension muscle forces at the end of swing phase

Finally, due to the complex nature of gait, using a polynomial to represent this motion can yield highly complicated motion patterns and inaccurate accelerations of the bones. To avoid over-fitting the kinematic profiles, the data was fit with the lowest degree polynomial that sufficiently captured the motion. This polynomial fitting process is a general limitation of the model as a whole and is discussed in the limitations section, Section 6.2.

CHAPTER 5: MODEL GRAPHICAL USER INTERFACE

Overall, the mathematical model presented herein provides an excellent theoretical simulator to aid in analyzing future implant designs, component placement, and surgical techniques. The model contains many valuable resources and features, listed below.

- Capabilities to implant different component designs for direct kinetic and kinematic comparisons
- Capabilities to misalign THA components for surgical accuracy comparisons
- Capabilities to weaken, cut, and/or move attachment locations of muscles and capsular ligaments for surgical technique comparisons
- Capabilities to monitor force distribution and femoral head sliding patterns within the acetabular cup to predict wear patterns
- Capabilities to predict realistic, unconstrained hip joint kinematics and kinetics as well as GRF and contralateral leg forces
- Capabilities to analyze multiple activities
- Capabilities to analyze multiple “patients” for diverse results
- Capabilities to predict hip separation occurrences

Unfortunately, fully capitalizing on these resources without prior knowledge of how to operate the model would be difficult. Thus, a GUI has been built to assist new model users and to ensure that the simulations are correctly setup and that the results are interpreted accurately.

5.1 Graphical User Interface Objectives

The primary function of the GUI is to provide a direct link between the model and any user who should desire to utilize its functions. Because the model is exclusively script-based, it would be unfeasible for others to easily run model simulations without significant training. However, by incorporating user-friendly tools and visualization techniques, it is possible to package the model as a single file for easier simulation management.

More specifically, once the model has been successfully built in Autolev and compiled in C++, the GUI serves as a tool to easily modify the text-based input file that the model reads as well as to easily interpret the results files produced from the simulations. The GUI is coded exclusively in MATLAB. Each forthcoming GUI window serves a unique purpose, either to save and organize model data, to overwrite a specific input variable in the input text file, or to format and plot the results for easy analysis.

The development of the model GUI is an ongoing process that may never be fully completed. Much like commercial software, there will always be improvements and updates to the software. Therefore, certain features should eventually be included in future versions of the software but have not been included at this time. The data files used for the GUI are named “HSIM” files, standing for “Hip Simulation.” These files contain detailed MATLAB structure data related to all aspects of the simulation.

5.2 Pre-Simulation Preparation Windows

Prior to running a simulation, a few specific steps must be taken to ensure the simulation will properly analyze the needs of the user. Once the user has properly loaded the .HSIM file relating to swing/stance phase preferences and fixed bearing/dual mobility

preferences, the user must select the component CAD models that he/she wishes to analyze (using the “Component Contact Detection Window”), implant the components in the desired orientation (using the “Component Alignment Window”), ensure the soft tissue attachment and wrapping points are in the desired location (if applicable, based on surgical technique, using the “Soft Tissue Window”), and implement any activity adjustments that are specific to the hypothetical patient being analyzed (if applicable, using the “Specified Inputs Window”).

5.2.1 Component Contact Detection Windows

The component contact detection GUIs, shown in Figure 73 and Figure 74, serve as means to build the contact surfaces for each component. The acetabular cup window, shown in Figure 73, fits polynomial surfaces to the CAD model vertices in accordance with the contact detection explanations found in Section 4.1.6. The current GUI is robust enough to handle neutral liners (Figure 75), lateralized liners, and lipped liners (Figure 76). A separate window, shown in Figure 77, analyzes dual mobility systems.

The femoral component, shown in Figure 74, is defined using a point cloud. The current femoral component GUI is therefore simpler than the acetabular component GUI, as polynomial fitting is not required. The user must simply create or load a surface representing the desired section-of-interest for the femoral component, and the GUI will automatically save the vertices to the data structure. The same dual mobility window as shown in Figure 77 can be used to generate the appropriate vertex cloud for the bearing, shown in Figure 78.

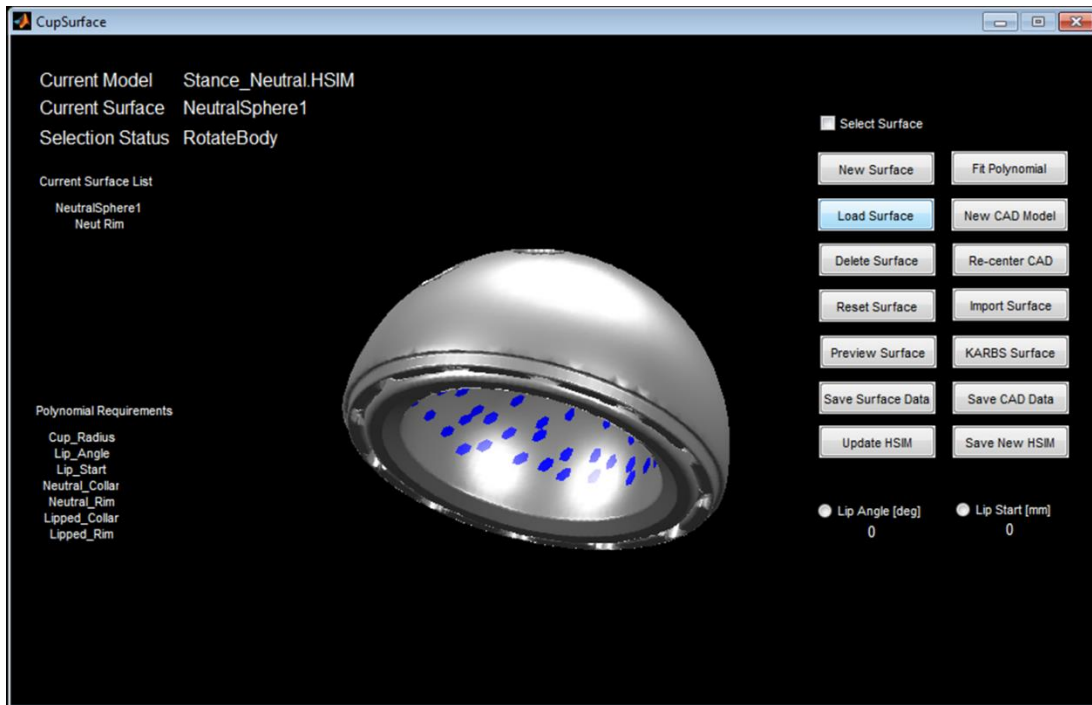


Figure 73: Acetabular component contact detection window, used to export polynomial fits.

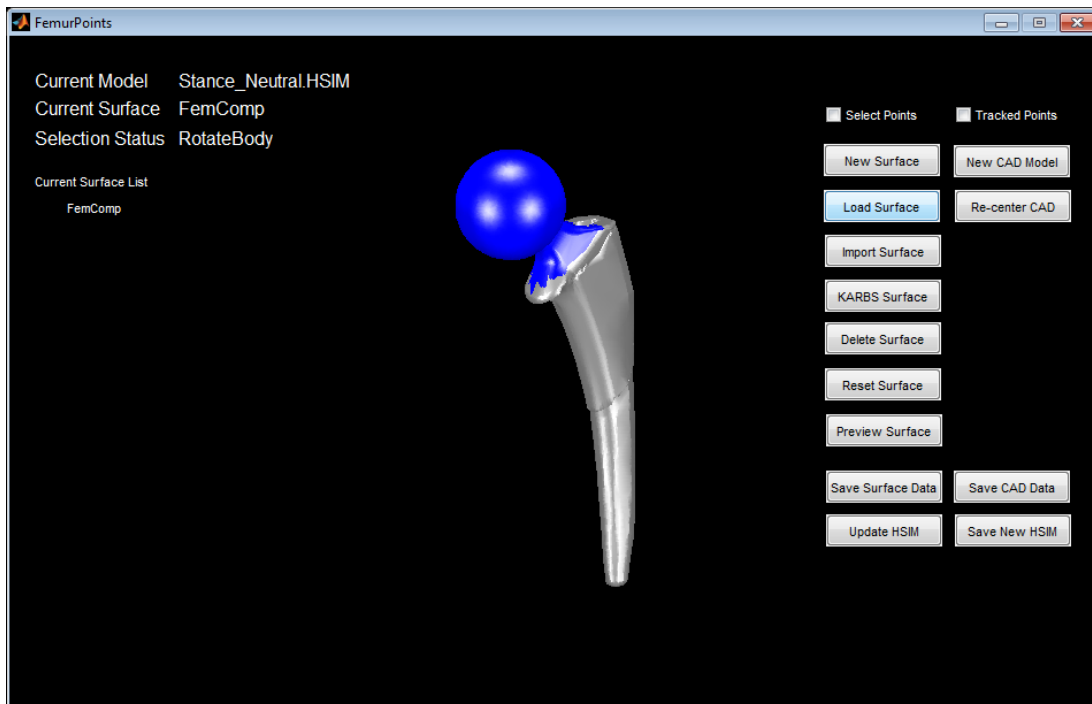


Figure 74: Femoral component contact detection window, used to export a point cloud.

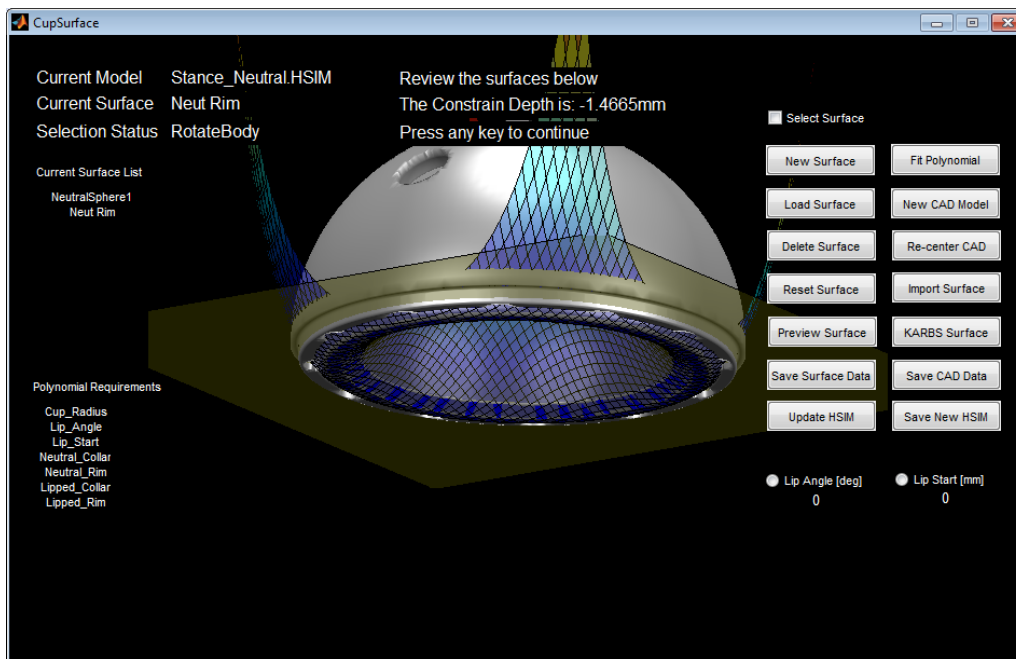


Figure 75: Acetabular component GUI showing a neutral rim polynomial fit.

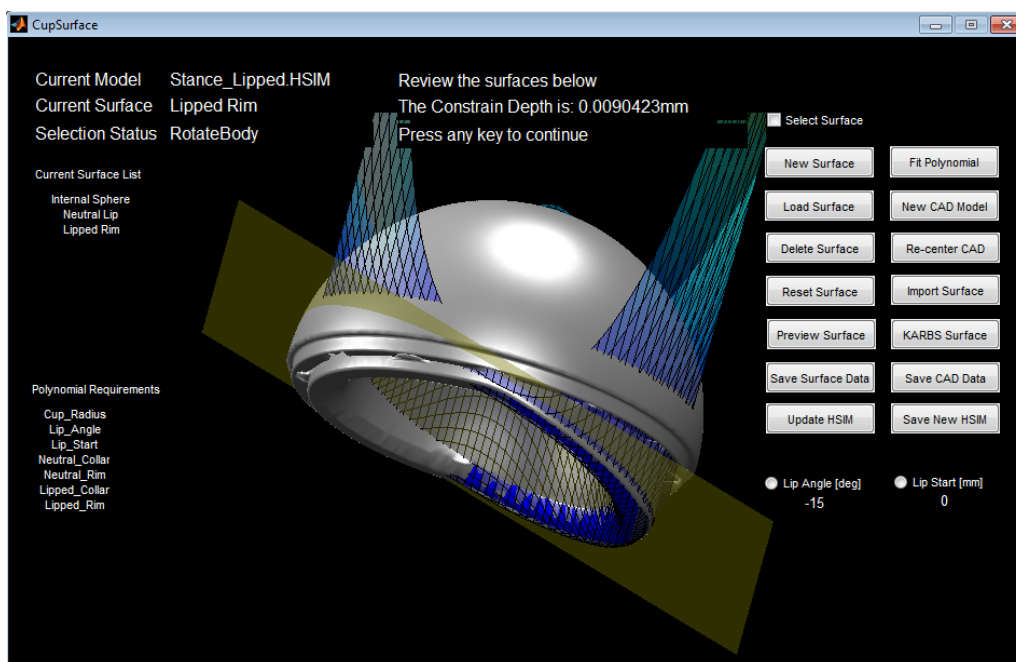


Figure 76: Acetabular component GUI showing a lipped rim polynomial fit. This polynomial becomes stitched with the neutral rim polynomial to form the complete representation.

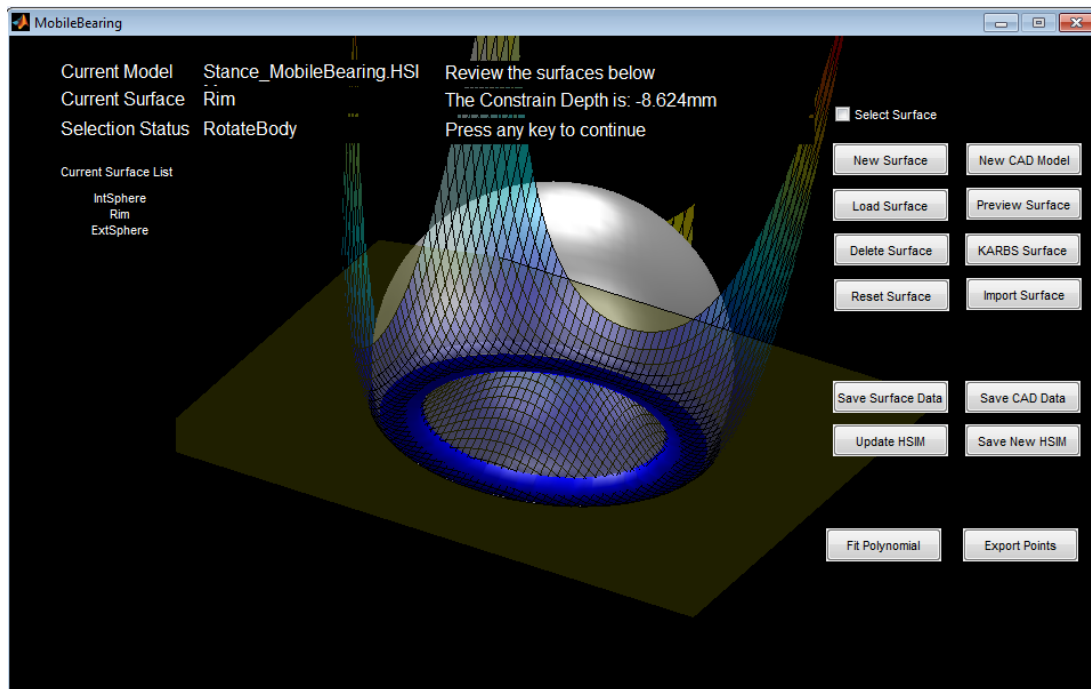


Figure 77: Dual mobility GUI showing the rim polynomial fit.

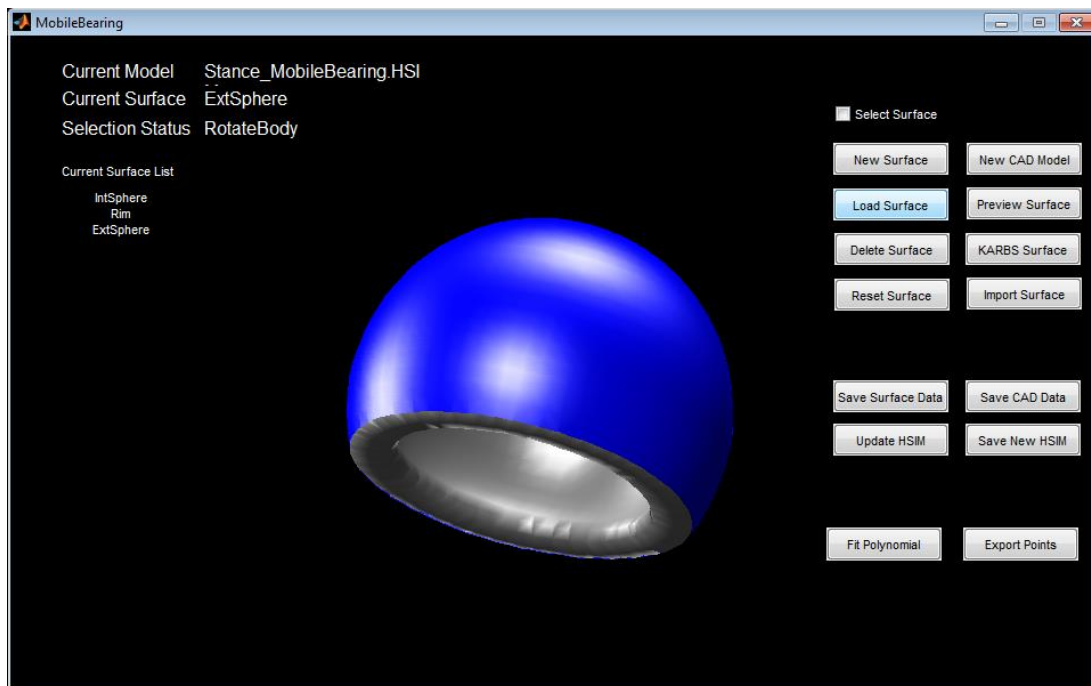


Figure 78: Dual mobility GUI showing the surface selected for the exported point cloud.

These GUI windows make it quick and easy to implant new components. Once a new CAD model is loaded, the existing surfaces are cleared and the checklist (if applicable) shown in the bottom left of the GUI must be completed. To save time, surfaces can be easily saved, managed, and transferred between simulation files. Once complete, selecting “Update HSIM” will automatically save the polynomial fit and/or point cloud vertices to the structure file for future analyses.

5.2.2 Component Alignment Windows

The component alignment windows are perhaps the easiest to use but also the most significant windows included in the software, as these are the tools that virtually implant the components and are therefore used to analyze the effects of component misalignment. Both the femoral head and the acetabular cup have nearly identical windows that can be used to align the components as desired (Figure 79 and Figure 80). These windows allow the user to modify and track component rotations (i.e. version, inclination, etc.) and translations (i.e. medialization, superior shifts, leg length discrepancies, etc.) of both components to meet desired scenarios. These capabilities are similar to the intraoperative alignment tools that are popular in the field of orthopaedics today, but the model’s software package comes with the added capabilities of biomechanical simulations.

Such capabilities allow for easy virtual implantation and theoretical analyses of surgical placement and accuracy. For example, Figure 81 shows an acetabular component misaligned by a translational magnitude of 1cm with radiographic component angles of 27° of inclination and 37° of anteversion. Compared to desired rotation angles

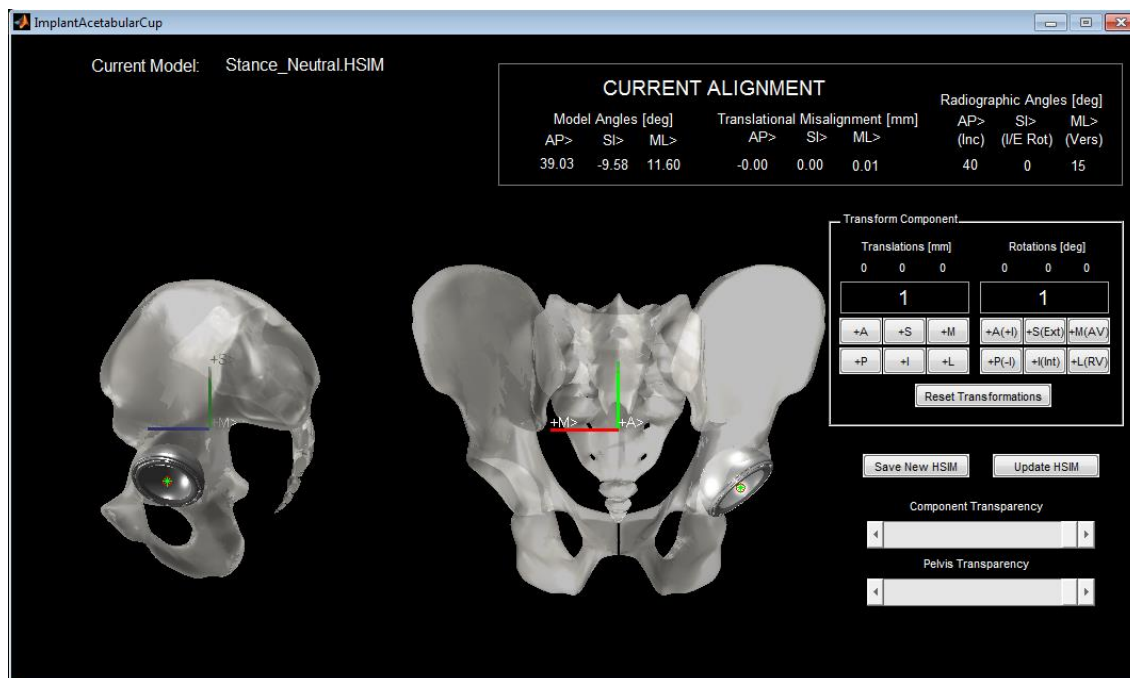


Figure 79: Acetabular component alignment window showing accurate alignment.

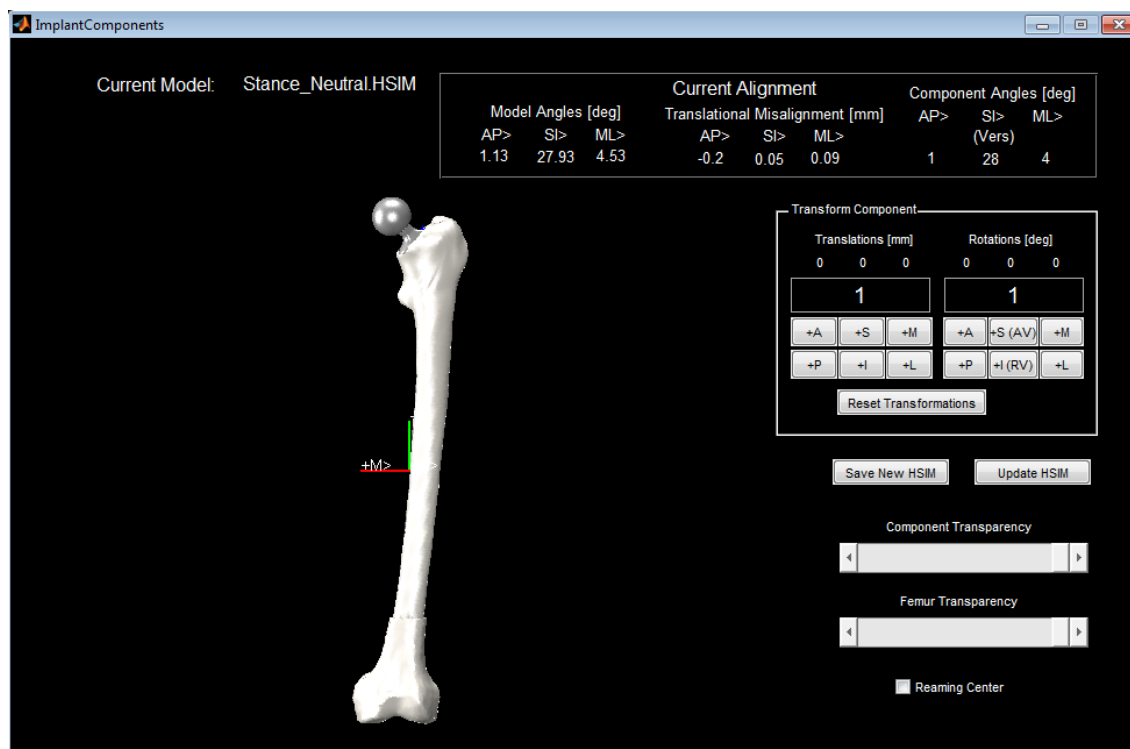


Figure 80: Femoral component alignment window showing accurate alignment.

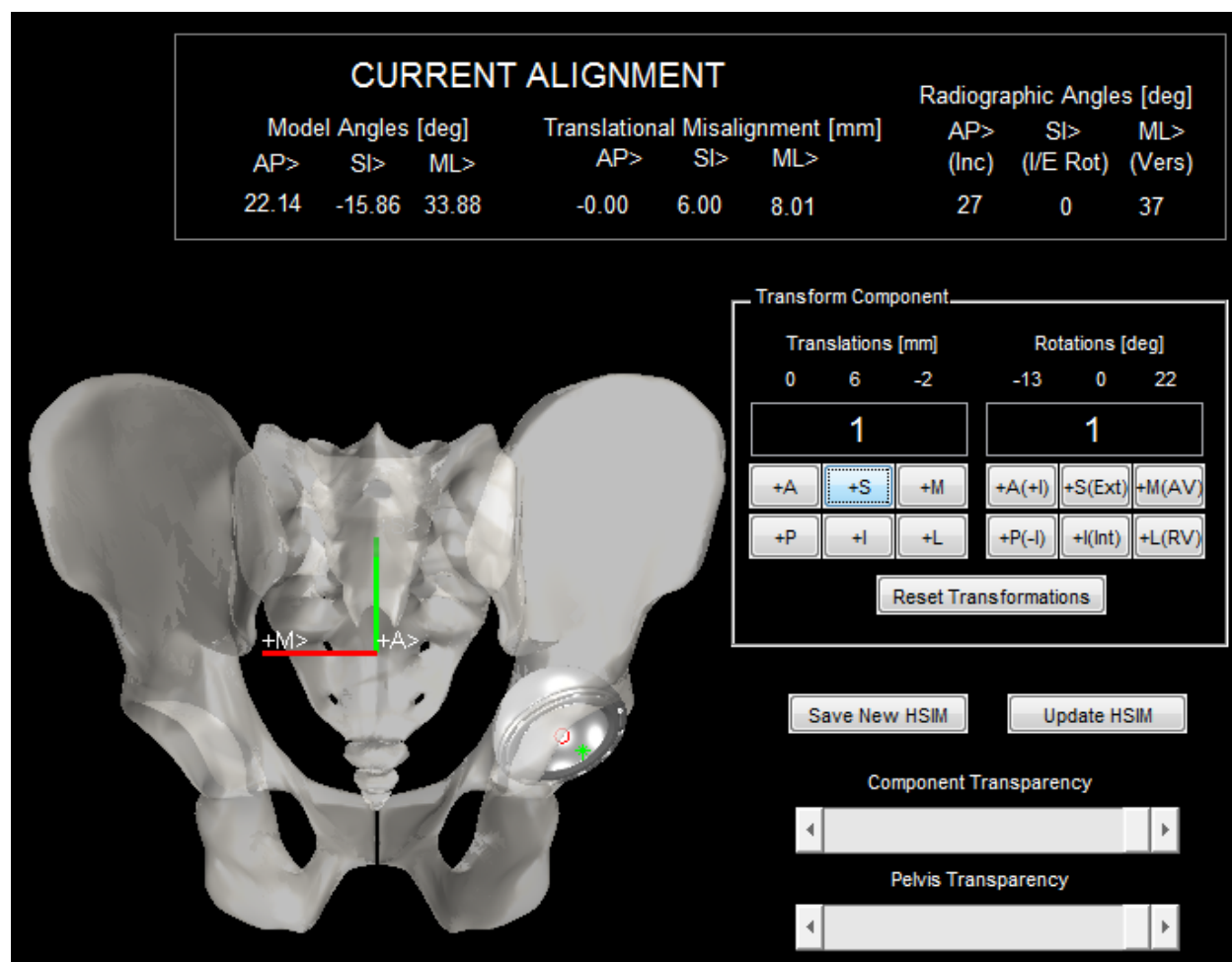


Figure 81: Acetabular component alignment scenario showing 1cm of translational misalignment, 13° of inclination misalignment, and 22° of anteversion misalignment.

of 40° of inclination and 15° of anteversion, this component is clearly misaligned. Similarly, Figure 82 shows a femoral component misaligned by approximately 14° of retroversion, yielding nearly 1cm of translational misalignment. In the femoral component window, the user has the capabilities of changing the theoretical location of the reaming point, shown as the blue point in Figure 82. Thus, when the user adjusts the angles of the femoral component (i.e. version), it will more accurately mimic intraoperative changes.

It is an obvious assumption that no surgeon would ever want to *intentionally* misalign the components to analyze the resulting biomechanics. Unfortunately, components are misaligned unintentionally, leading to concern after implantation, and this mathematical modeling software makes it possible to analyze such scenarios. Once alignment is complete, selecting “Update HSIM” will automatically save the transformation data to the structure file for future analyses.

5.2.3 Soft Tissue Window

Figure 83 shows the soft tissue window, which serves primarily to ensure that the muscle fibers are attached in the correct location. Once this is confirmed, this window is rarely modified. If the surgeon is considering modifying the attachment site of a ligament or muscle intraoperatively, this window could be used to simulate such an event. The first column of check boxes allows the user to visualize any muscle that he/she wants, ensuring that all chosen muscle attachment sites are accurate. If the user wanted to change the muscle attachment or wrapping sites, this can be done using the second column. The user can select the specific muscle fiber of interest and manually modify the location of that point. This can be seen in Figure 84. Once this is complete, selecting

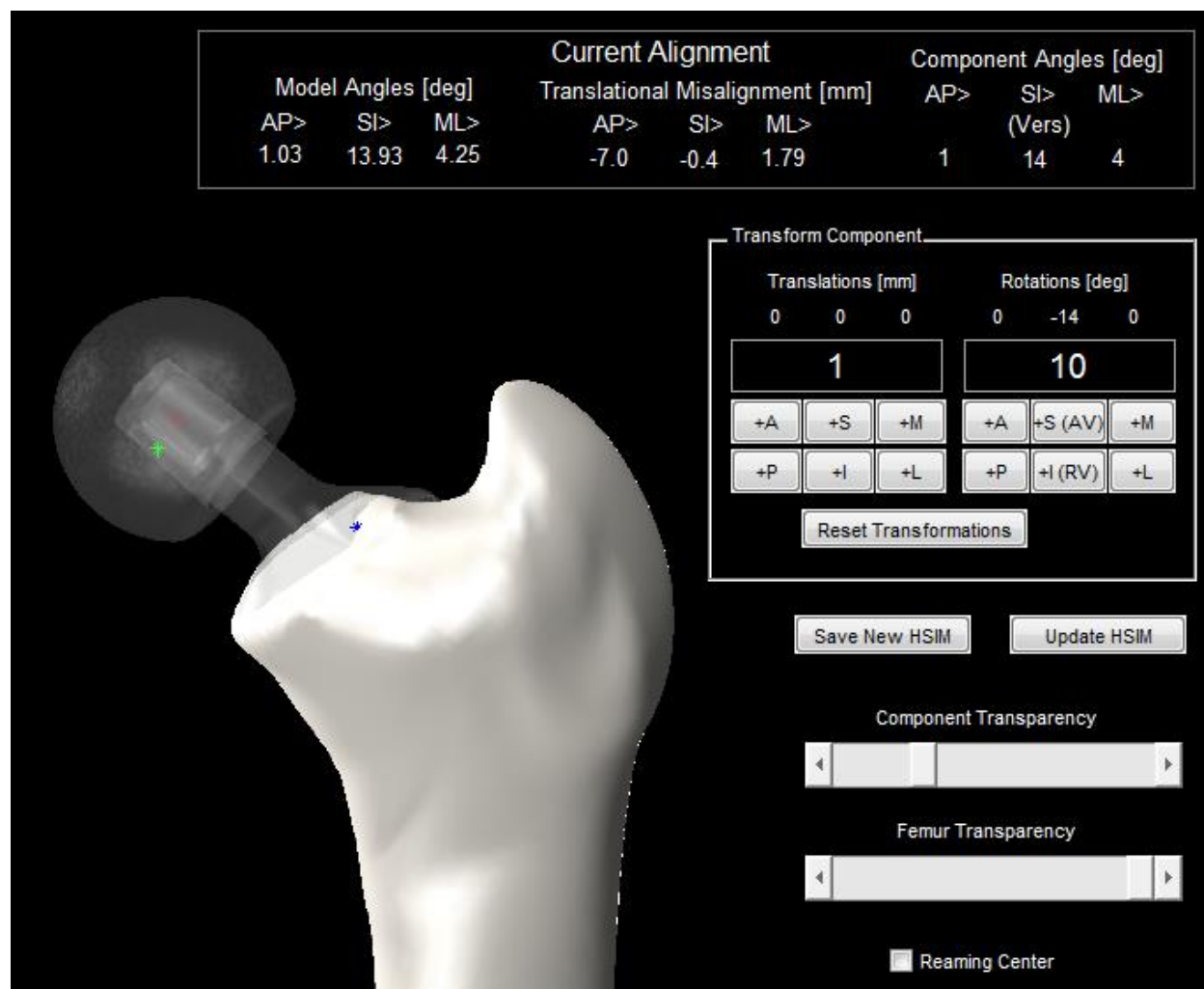


Figure 82: Femoral component alignment scenario showing 14° of retroversion misalignment, yielding nearly 1cm of translational misalignment.

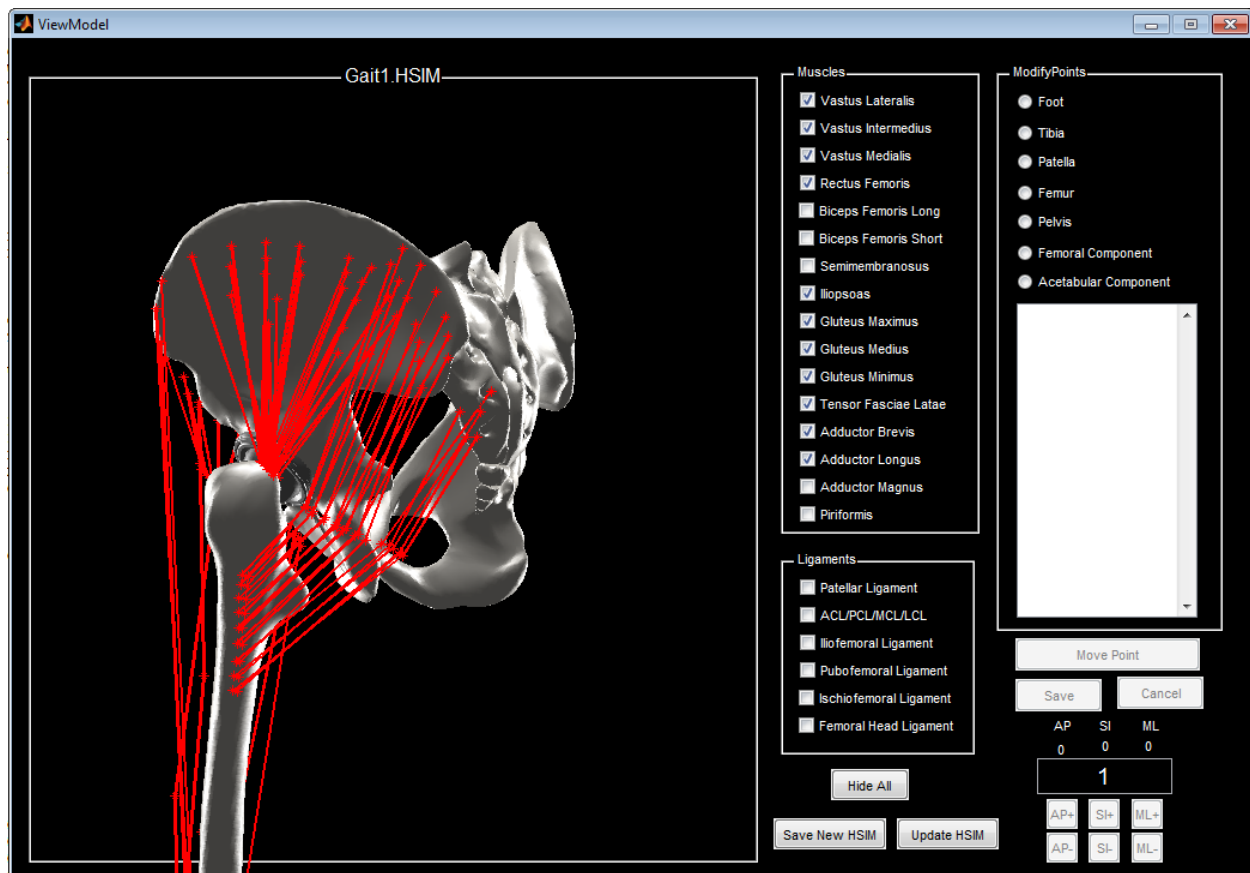


Figure 83: Soft tissue window showing select hip muscles and currently no capsular ligaments.

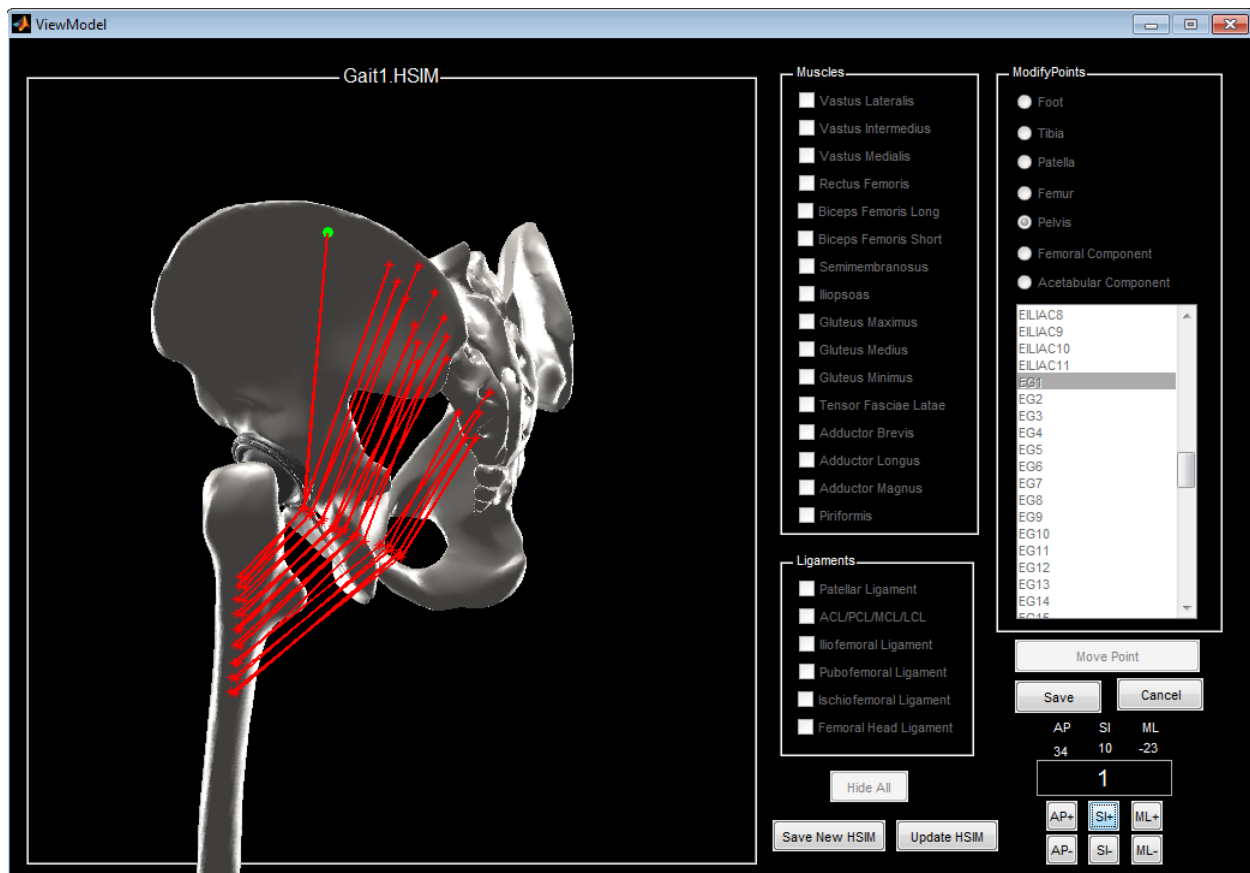


Figure 84: Soft tissue window showing a modified gluteus maximus attachment point.

“Update HSIM” will automatically save the point data to the structure file for future analyses. In future versions, this window will contain methods to resect specific muscle fibers. However, thus far, this must be done manually, using script.

5.2.4 Specified Input Window

The specified input window has been developed to allow the user to specify, using a variety of different methods, desired rotational and translational inputs that are required for the model to run. In other words, this is the window where different patient-specific kinematics can be input or where different activities can be created. The use of this window is relatively technical and requires previous knowledge of the desired activity and/or the patient-specific kinematics.

The user can currently specify polynomial coefficients (up to the 6th degree), manually select/modify points, or scale the inputs as desired. In the future, the user will be able to load tabulated data (gathered from gait labs or goniometers). Figure 85 shows manual modifications to the current tibia motion profile. Note, the shown modifications (in red) are not accurate, they are purely for visualization. Because forces and torques are based on accelerations instead of positions, the window will also generate first and second derivative plots for the current input. Once this is complete, selecting “Update HSIM” will automatically save the input curve data to the structure file for future analyses.

5.3 Post-Simulation Analysis Windows

Once the simulation preparation steps have been completed, the user must select “Update and Run” from the “Run Simulation” tab of the main GUI. This will automatically

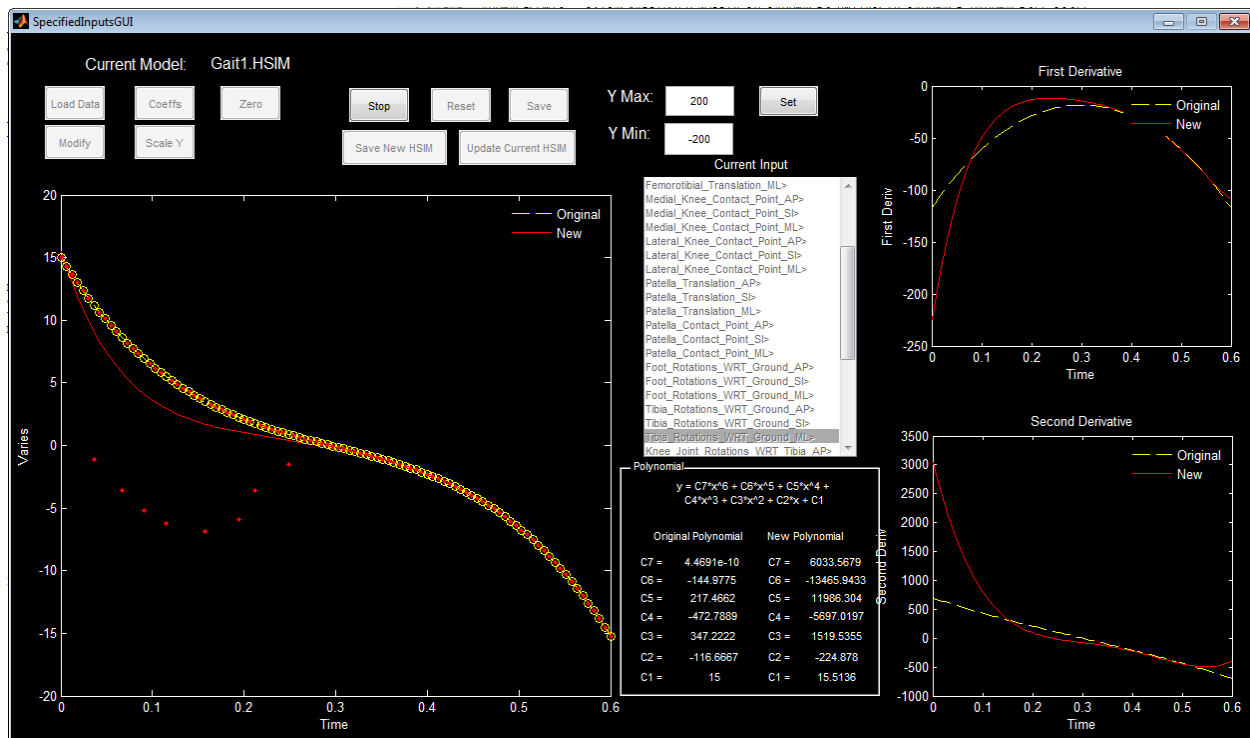


Figure 85: Specified input window showing a modification of tibial rotation.

update the appropriate input file and begin running a new simulation. Depending on the complexity of the simulation, it will generally take between 1-5 minutes to complete. The user could then use the model to store the results in the HSIM structure, allowing for rapid analyses in the future. Once complete, the user has a handful of tools available to aid in interpreting the results, including: (1) a window to visualize four specific results of interest (using the “Results Visualization Window”), (2) a window to compare the current simulation results to a previous simulation (using the “Baseline Comparison Window”), (3) a tool to animate the activity (using the “Animation Tool”), (4) a window to easily visualize the in vivo separation and contact region of the hip joint (using the “Contact Map Window”), (5) a window to easily visualize the motion of fixed femur points of interest with respect to the acetabular cup (using the “Track Femur Points Window”), and (6) a menu to save specific results of interest, including animations and figures (using the “Save Results Menu”).

5.3.1 Results Visualization Window

A visualization window has been developed to aid the user in plotting and interpreting the results. The GUI will automatically interpret and format the text-based results to present clear, easy to understand figures. It currently provides four plots (hip joint reaction forces, hip capsular ligament forces, hip muscle forces, and hip separation). Figure 86 shows the window for a stance phase activity with properly aligned component, yielding $<0.2\text{mm}$ of hip separation and $\sim 2.3\text{xBW}$ of maximum hip joint reaction force.

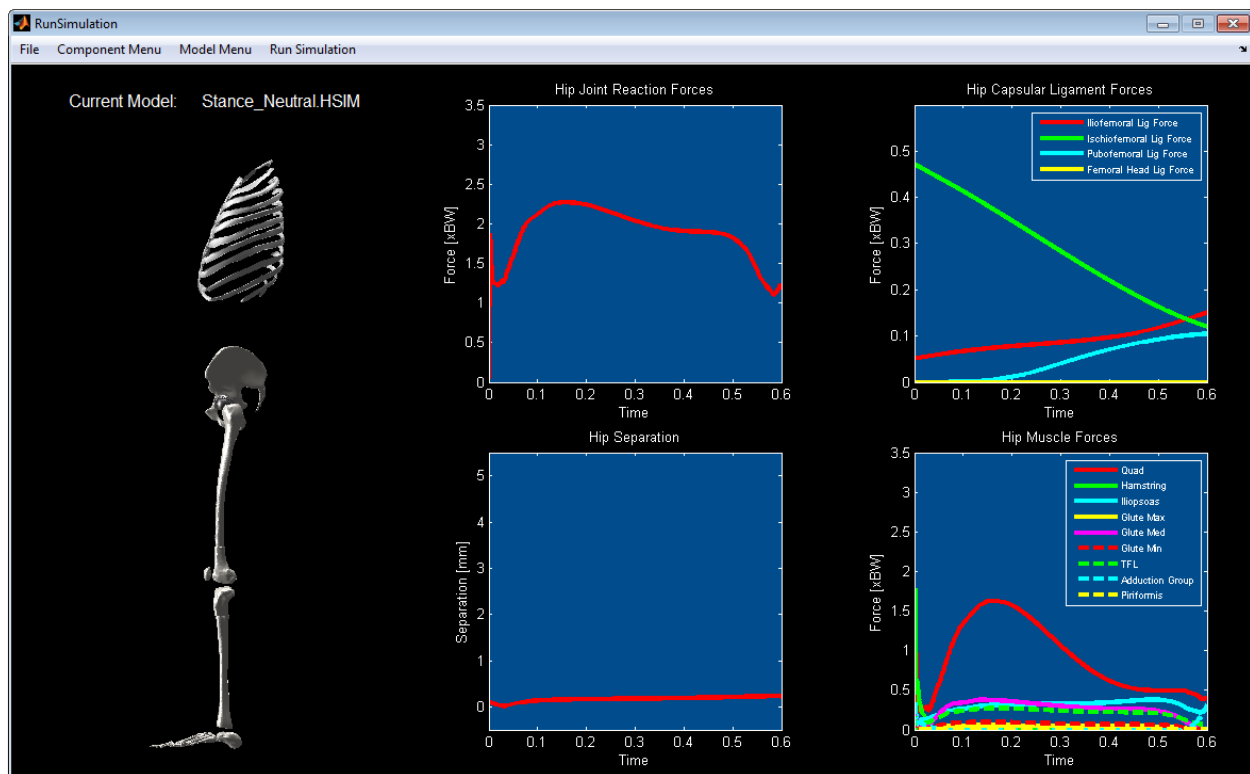


Figure 86: Results visualization window showing in vivo mechanics.

5.3.2 Baseline Comparison Window

Like the results visualization window, a similar window has been developed that allows the user to compare results of interest to an established “baseline” simulation. More specifically, this allows for a direct comparison between, for example, a separation and a non-separation instance. Similarly, this can allow for a direct comparison between two different component models or two different sets of mechanics. The user must first establish the baseline simulation by selecting “Set Baseline” under the “Run Simulation” tab. Once established, this baseline simulation will be saved indefinitely and can be compared to any new simulation that is conducted. The window will compare hip joint reaction forces, hip joint contact areas, hip separations, and hip joint contact stresses of the current simulation to the baseline simulation. Figure 87 shows a sample comparison for a swing phase activity. The baseline data, shown by green dashes, is for a properly aligned component, generating <0.2mm of hip separation, while the current simulation data, shown by the red solid line, is for a misaligned component, generating over 3mm of hip separation. The corresponding decreases in contact area and increases in contact stress are clearly visible.

5.3.3 Animation Tool

The animation tool allows the user to visualize the activity that has been simulated, providing a method to ensure accurate inputs as well as observe the kinematic outputs. For clarity, the animation capabilities include a representation of a floor. Figure 88 shows both a stance and swing phase animation.

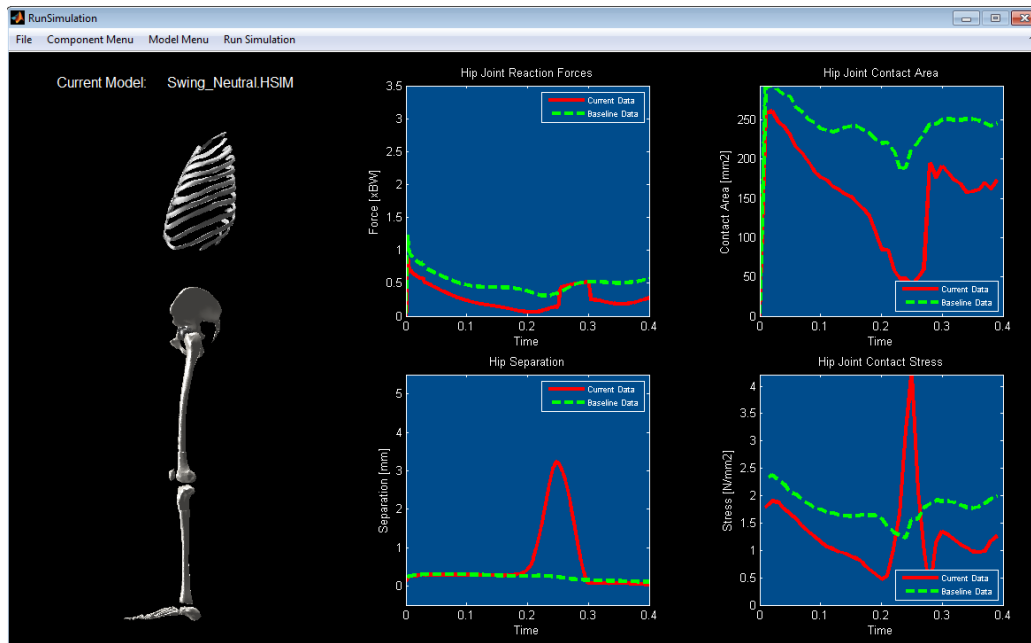


Figure 87: Baseline comparison window comparing swing phase separation and non-separation scenarios.

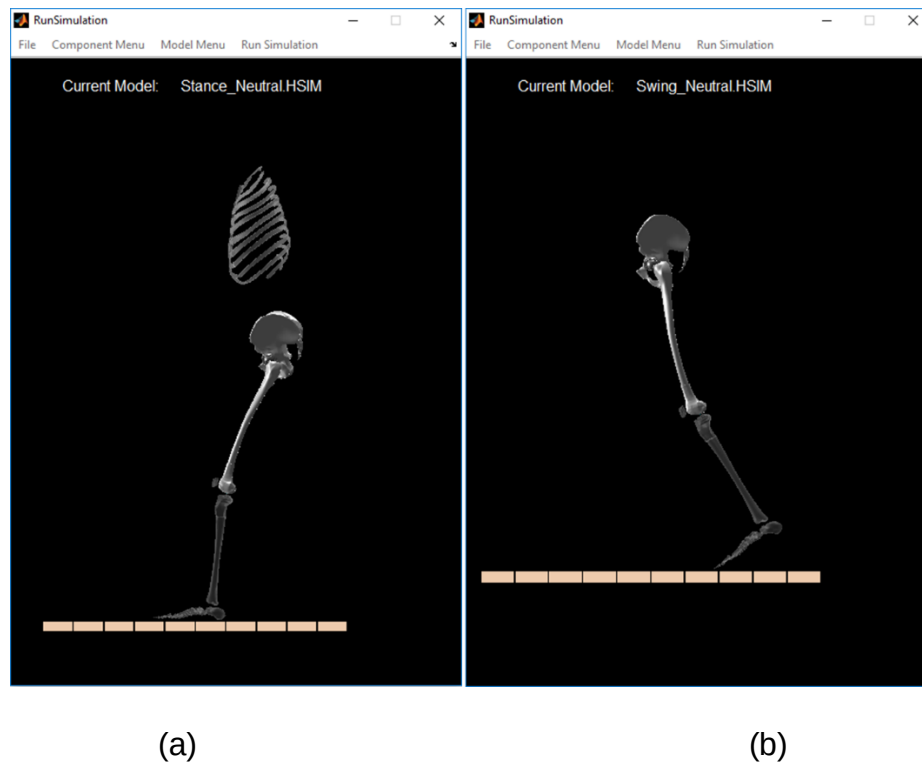


Figure 88: Animation tool showing (a) stance phase and (b) swing phase.

5.3.4 Contact Map Window

The contact map window allows the user to better visualize the contact map, area, percent, and stress distributions that are occurring throughout the simulation. Figure 89 shows the separation visualizations and contact data for a stance phase, non-separation scenario. From this visualization, contact clearly occurs on the superior aspect of the cup. Figure 90 shows the separation visualizations and contact data for a stance phase, separation scenario. From this visualization, contact clearly occurs laterally, near the rim of the cup (edge loading), similar to the wear region seen in polyethylene liner retrievals [5]. Comparing these 2 figures, it is also clear that contact area during separation instances decreases, and therefore contact stress increases. Figure 91 shows a stance phase, non-separation scenario for a dual mobility implant.

The contact map is colored based on the depth of penetration between the femoral head and the acetabular component, with deeper penetration shown in red. The contact area is computed from the femoral component. Specifically, every CAD vertex and triangular face that is determined to be in contact is categorized accordingly. From there, the area of each face, using the vertex coordinates, is computed and summed, yielding the total contact area. The contact percent is determined by dividing this contact area by the total surface area of the liner articulating surface, excluding the rim. This area is determined using standard geometry equations by treating the internal sphere, the Charnley Bore, and the Cylindrical Bore as three separate regions and summing them together. The algorithm is robust enough to handle neutral liners, lipped liners, and augmented lipped liners. Finally, the contact stress is determined by dividing the total hip force at each time step by the computed contact area at each time step.

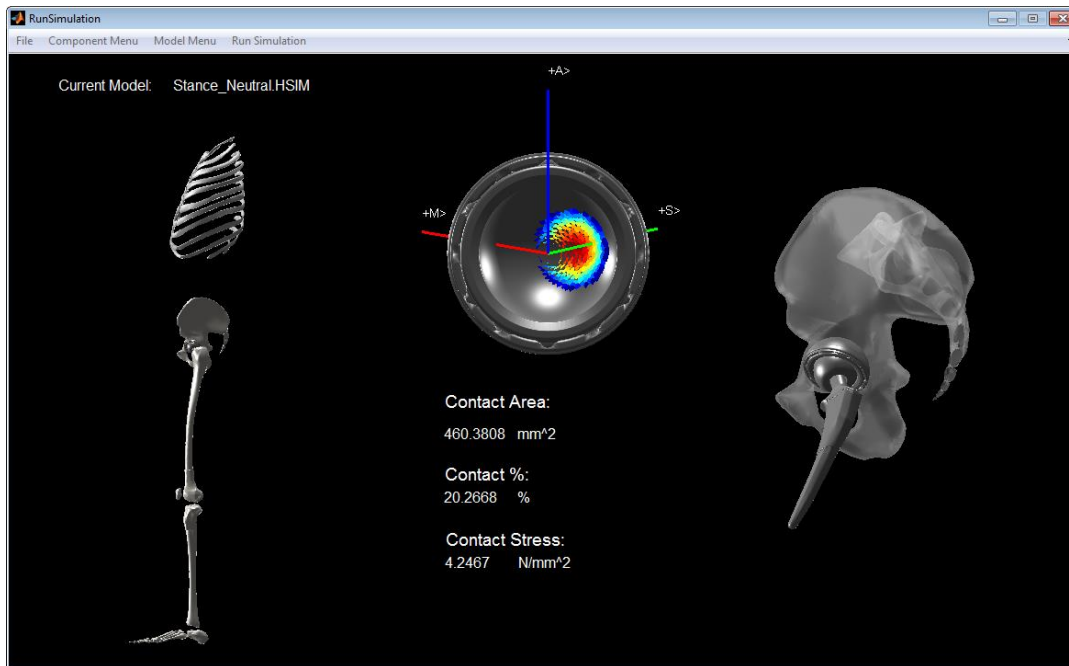


Figure 89: Contact map window showing non-separation stance phase results.

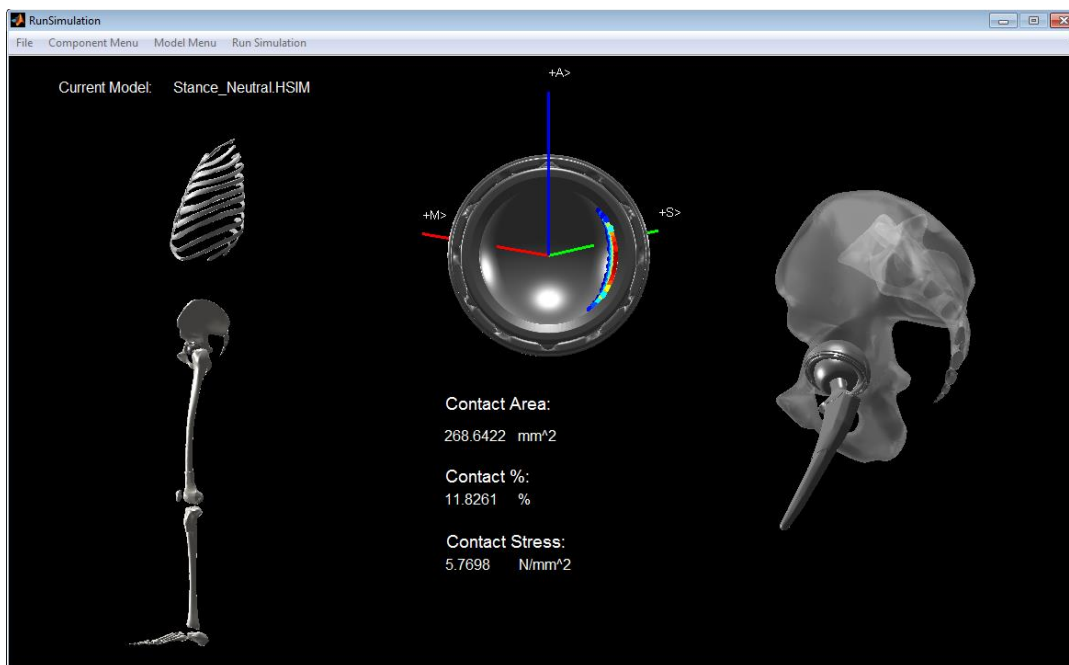


Figure 90: Contact map window showing a separation and edge loading during stance phase.

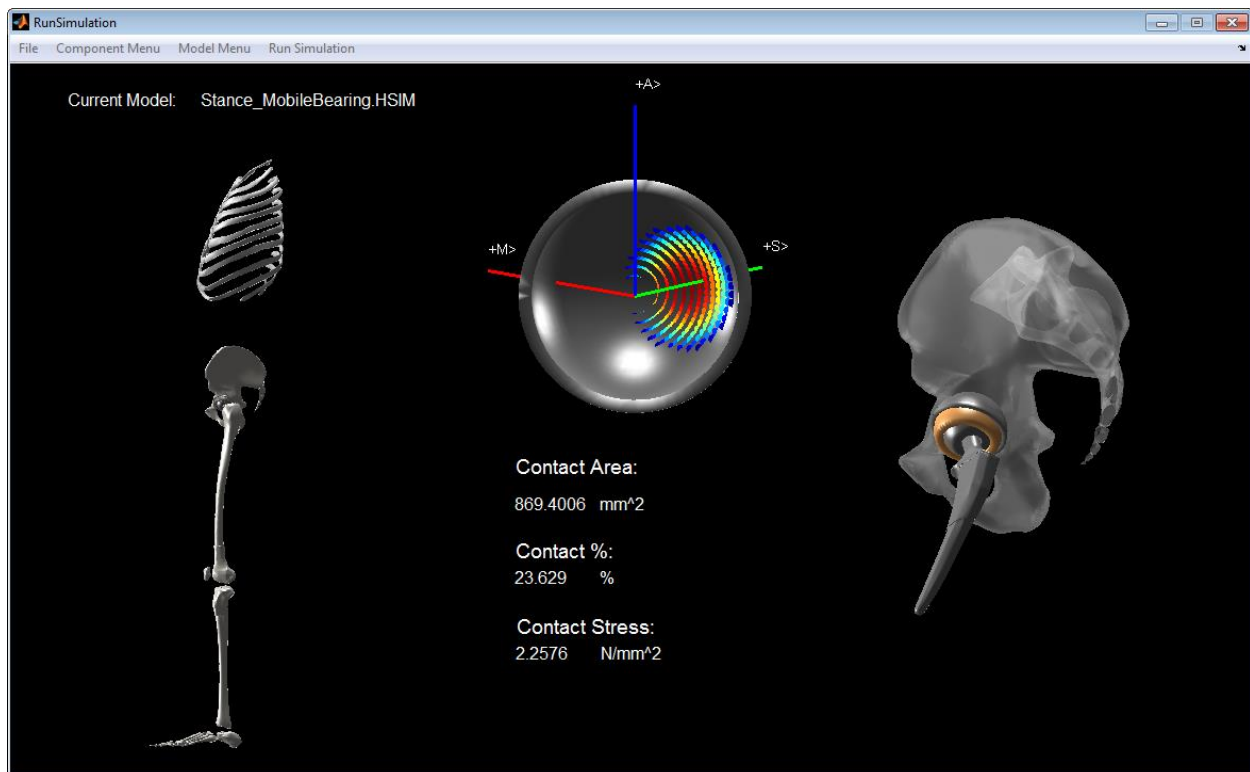


Figure 91: Contact map window showing a dual mobility implant in stance phase with no separation.

5.3.5 Track Femur Points Window

In addition to predicting kinetics, rotational motion, and hip separation, the forward model also has the capabilities of tracking certain points of interest on the femoral head with respect to the acetabular cup. This, along with the contact detection algorithm, can provide insight into the wear patterns on the THA bearing.

Specifically, 5 fixed points of interest can be selected on the femoral component, shown on the left in Figure 92. The location of these points can be easily modified depending on the scenario of the simulation. The movement of these points can be tracked throughout the entire activity, as shown on the right.

5.3.6 Save Results Menus

Another feature included in the GUI is the ability to automatically generate .JPG versions of the results figures as well as save four different animations (Figure 93). Selecting “Generate Figures” will save a total of 10 images, including all the figures shown in the “Plot Mechanics” and in the “Compare Baseline” menu options. Selecting “Save Animation” will give the user the option to save one of the four animations that occur throughout the GUI, yielding videos that are smoother, faster, and therefore more accurate than the in-GUI animations.

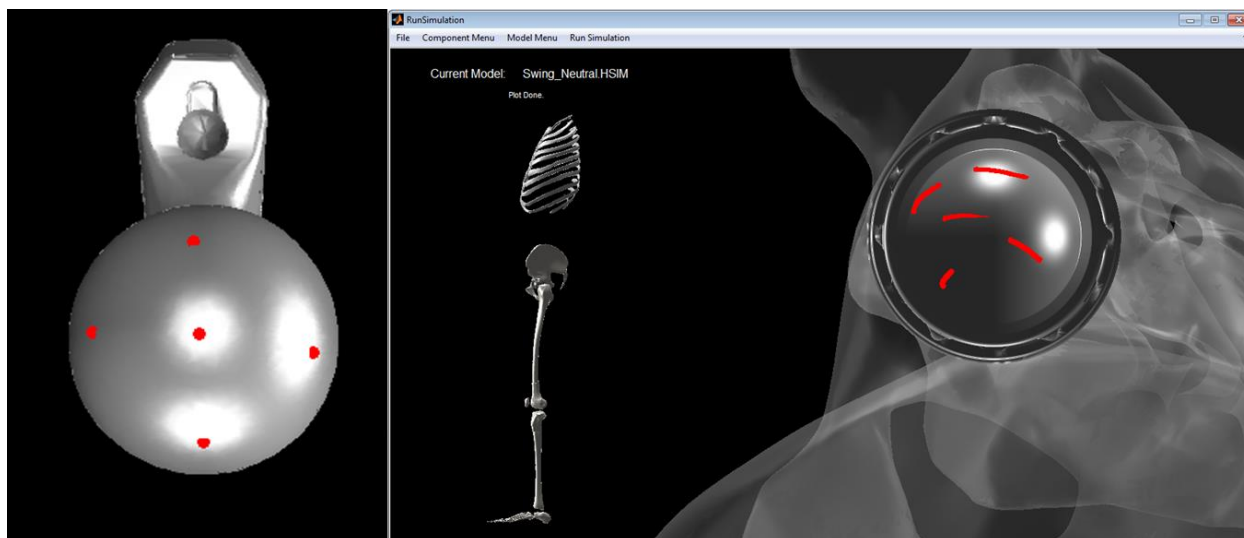


Figure 92: Track femur points window showing the points on the left and their in vivo motion on the right.

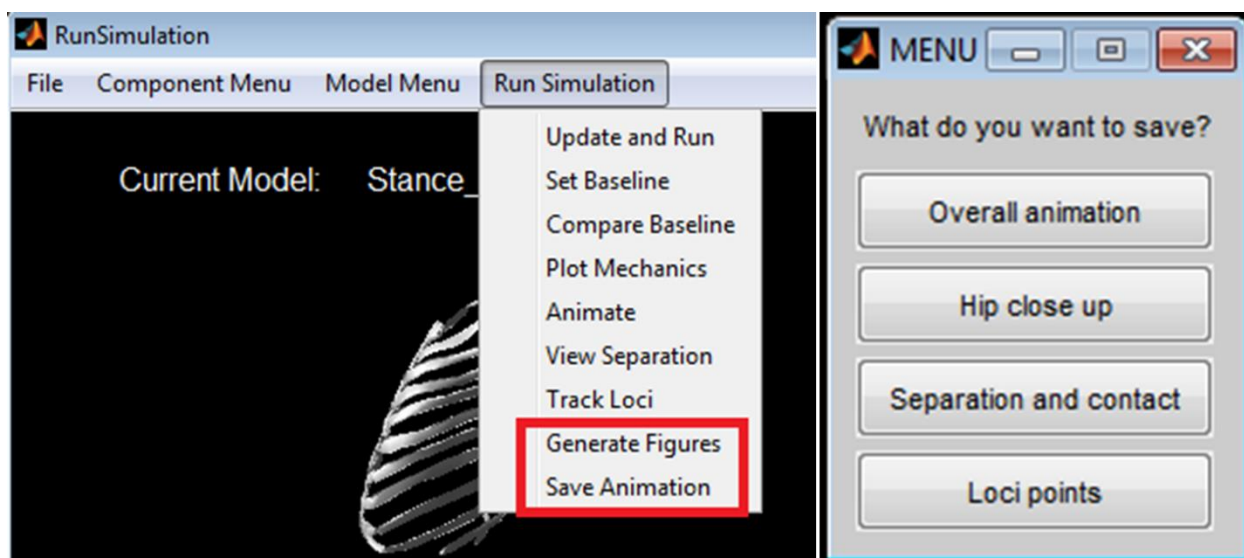


Figure 93: Save results menu options in the GUI.

CHAPTER 6: ASSUMPTIONS AND LIMITATIONS

6.1 Assumptions

For this model, there were a variety of assumptions that were made, outlined as follows:

1. Rather than using moment of inertia (MOI) integration methods to compute bone MOIs, the MOIs were calculated from cylinder or rectangular plate equations. Thus, bone MOIs may be slightly inaccurate. Alternatively, the MOIs could be hard-coded, based on literature, but this does not allow for robust scaling or re-sizing of the bones, as analyzing larger patients would require additional literature research.
2. The masses of the foot, tibia, femur, and pelvis are modeled as 1.4%, 4.6%, 10%, and 24% of the subject's body weight, respectively. The mass of the patella is considered negligible.
3. The patella is modeled as a three-force system that includes the quadriceps force, the patellar ligament force, and the patellofemoral reaction force, in accordance with [18].
4. The muscles control a prescribed rotation *direction*. In other words, the gluteus maximus will always contribute to hip *extension* and never hip *flexion*. There are certain in vivo scenarios where a muscle may control one direction of rotation in one position and then the opposite direction in another position. Specifically, the piriformis (not modeled) controls external rotation at full hip extension and internal rotation at 90° of hip flexion [8]. These scenarios are not accounted for herein.
5. For modeling surgical techniques, the model applies the weakened muscle fiber contributions by an assumed factor.

6. The S/I rotation of the hip joint is directly controlled by a torque and not muscles.
Although out-of-plane muscles contribute to this rotation, they do not control it.
7. The forward solution model developed herein is an *a priori* forward model, meaning it incorporates prior knowledge regarding the activities. Optimization algorithms are required to remove this assumption.
8. The initial conditions of the bodies are chosen based on inverse modeling, and the initial condition of the dual mobility bearing is unknown. Therefore, the initial condition of the dual mobility bearing is assumed to match the initial condition of the cup, and thus the initial condition of one gait phase (i.e. stance) may not match the end condition of the other phase (i.e. swing), and vice versa.
9. The swing phase motion is simplified. Minimal in-house experimental data was available regarding swing phase transformations. Thus, the swing phase transformations were determined and simplified from literature.
10. The proprioceptive notion that the hip “wants” to rotate about its anatomical center is an assumption that can be turned “off,” if desired.
11. The contact area percent calculation excludes the surface area of the rim region.
It only includes the spherical and bore regions.
12. It is assumed that the muscles are always wrapped around prescribed wrapping points.
13. The contralateral leg forces during stance phase are input and assumed based on model data from swing phase.

6.2 Limitations

There are a variety of limitations associated with this research, outlined as follows:

1. In general, one of the most notable limitations for all mathematical models of the human body, not just the one presented herein, is the ligament properties. Determining proper stiffnesses and slack lengths of in vivo ligaments is difficult when doing patient-specific modeling. The ligament properties must either be assumed from literature or cadaveric studies (as was done here) or determined iteratively by having the patient conduct multiple activities and using one of the activities as a control to tune.
2. The algorithms used to compute the muscle forces in this model are relatively simple. While this is advantageous in the interest of speed, muscle optimization algorithms could be incorporated in the future to increase the accuracy of the muscle profiles and to allow for more muscles to be incorporated into the model.
3. All four patients that were used for the analyses contain the same soft tissue attachment sites. Modifying the attachment sites of the soft tissues would require CT/MRI data for each patient, which was unavailable for this project. In addition, all four patients are the same height. Accurately modifying this information would again require CT/MRI imaging data. Alternatively, the bone sizes and attachment sites could be scaled to accommodate different heights. This was done for a few informal simulations, but it was determined that this method was inaccurate.
4. The current research did not analyze non-implanted patients. It would be advantageous to analyze non-implanted patients to serve as a control. However, adequate data for non-implanted patients was not available.

5. The model does not incorporate muscles that control ankle torques (i.e. tibialis anterior, gastrocnemius, or soleus muscles). Incorporating these muscles may affect the kinetic results formulated herein.
6. The polynomial representations that are used to describe the acetabular component surfaces may allow for fast modification of components, but they do not capture sharp curvatures very well. Furthermore, polynomials enable the possibility of over-fitting the surface, yielding inaccurate surface representations. Alternatively, point clouds would remove over-fitting and would better capture sharp corners, but implant design modifications would be tedious. Modifications would require a new CAD model to be loaded and analyzed within the GUI.
7. There was no femoral head size analysis that was conducted. Because the femoral head is described using a point cloud, modifying its geometry would require new CAD models, which were unavailable.
8. The model is sensitive to the contralateral leg forces. These forces are supplied as an input to the stance phase model, so inaccuracies in these forces will yield inaccuracies in this model.
9. It is currently not possible to resect muscles or ligaments in the GUI. This must instead be done manually, within the model's code. This is tedious and should be modified in the future. Similarly, there is currently no way to load new bones in the GUI. This should be modified in the future, as well.
10. Using polynomial fits to input kinematics can yield inaccurate forces, as polynomials cannot capture highly complicated motion patterns and consequently may capture inaccurate accelerations of the bones.

CHAPTER 7: RESULTS AND DISCUSSION

Overall, as previously stated, the mathematical model presented herein provides an excellent theoretical simulator to aid in analyzing future implant designs and surgical techniques. The focus of the research has been to determine the effects of modifying all six degrees of freedom (three translational and three rotational) of both components, with an emphasis on analyzing different liner designs to determine the best means to minimize hip separation and instability. Specifically, the model has been used to conduct multiple in-depth analyses, including:

- A 30-patient fluoroscopy analysis to determine the effects of surgical technique variations on hip forces and hip separation.
- A detailed acetabular cup and femoral component position analysis during swing phase to determine ideal component positioning to minimize hip separation.
- A detailed cup and femoral component position analysis during stance phase to determine ideal component positioning to minimize hip separation.
- A comparison of multiple different liner designs to determine the pros and cons associated with each design.

7.1 Inverse Model Analysis: Surgical Technique and Hip

Separation Analyses

The first version of the model that was developed was an inverse model, used to complement an in-house fluoroscopy study unrelated to this dissertation. With this older

model, three sub-analyses were conducted to determine the effects that various surgical techniques and their associated soft tissue disruptions have on hip and muscle forces. The following three analyses were conducted using an inverse model where the inputs and translations were determined via this fluoroscopic analysis.

7.1.1 Theoretical Surgical Technique Comparison Analysis

This analysis was conducted on a single patient with a low-end maximum hip force. The input kinematics for each approach are identical, and surgical technique variations were induced simply by weakening various muscle fibers of the appropriate muscles based on what has been reported in the literature. Specifically, if a muscle is cut during surgery, the current analysis assumes that the maximum strength of the entire muscle reduces by approximately 20%. Additionally, the specific fibers that are directly cut are weakened by approximately 60%, and the neighboring muscle fibers are weakened by approximately 40%. These are the chosen values for this study, but the user could modify the reduction strength values to any percentage. Specific figures showing the affected muscles for each surgical technique are referred to in their respective sections below.

7.1.1.1 Direct Anterior Approach

During the direct anterior surgical approach, no muscles are cut [58]. Hence, the model's muscles are unaffected, and this technique can therefore serve as a control upon which to analyze the other two techniques. Figure 94 and Figure 95 respectively show the hip muscle forces and resultant joint forces for this technique.

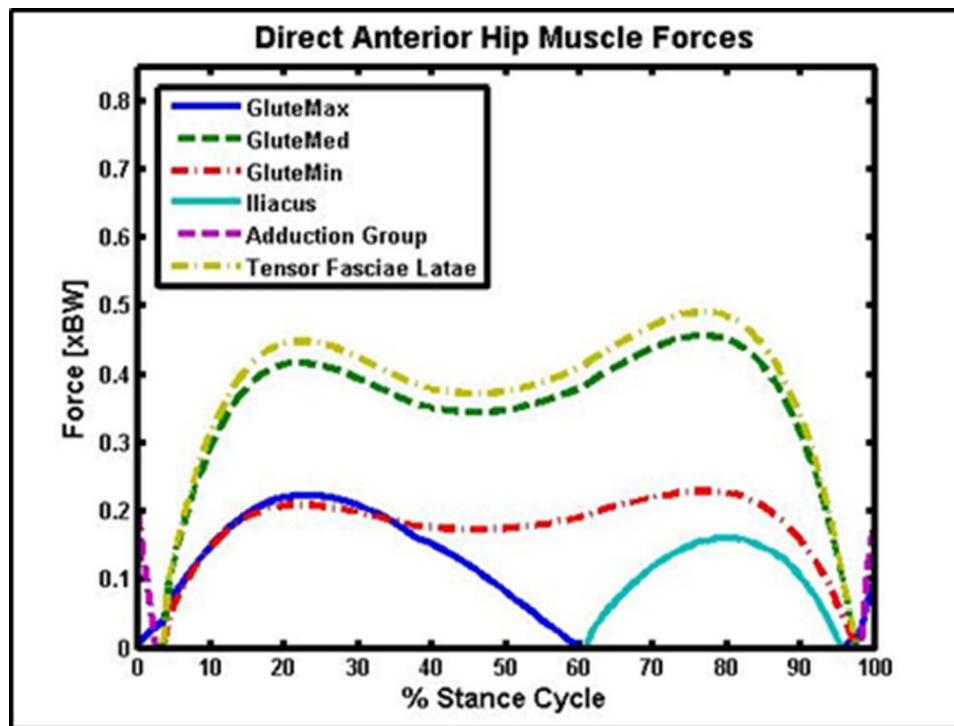


Figure 94: Muscle forces for the direct anterior surgical approach.

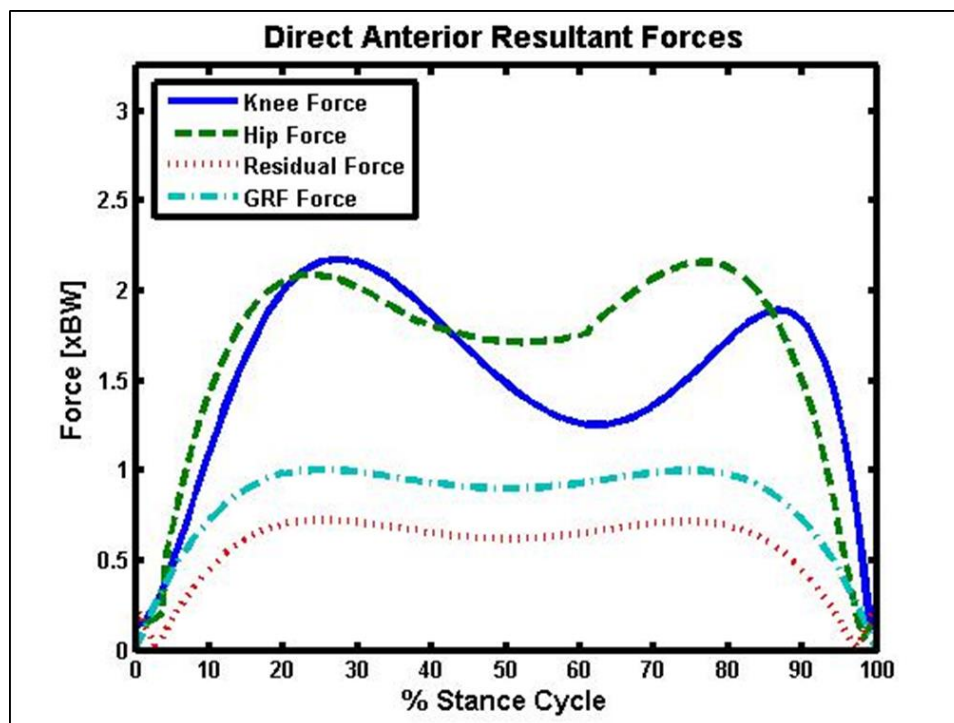


Figure 95: Resultant joint forces for the direct anterior surgical approach.

7.1.1.2 Anterior Lateral Approach

During the anterior lateral surgical approach, the posterior fibers of the tensor fasciae latae and the anterior fibers of the gluteus medius are incised [58]. Figure 96 shows the directly-affected fibers in blue (weakened by 60%), the indirectly-affected fibers in purple (weakened by 40%), and the remaining muscle fibers in red (weakened by 20%). Figure 97 and Figure 98 respectively show the hip muscle forces and resultant joint forces for the anterior lateral surgical technique. Due to the weakened TLF and gluteus medius muscles, the gluteus minimus muscles experience a force increase to compensate (Figure 97).

7.1.1.3 Posterior Lateral Approach

During the posterior lateral surgical approach, the center fibers of the gluteus maximus are incised [58]. Figure 99 shows the directly-affected fibers in blue (weakened by 60%), the indirectly-affected fibers in purple (weakened by 40%), and the remaining muscle fibers are shown in red (weakened by 20%). Figure 100 and Figure 101 respectively show the hip muscle forces and resultant joint forces for the posterior lateral surgical technique. To compensate for the weakened gluteus maximus muscles, the gluteus medius and opposing iliacus muscles both experience force increases (Figure 100).

7.1.1.4 Resultant Hip Force Comparison

A direct comparison of resultant hip forces for all three surgical techniques can be found in Figure 102. According to the results in this analysis, the direct anterior approach had the lowest maximum hip force ($2.16 \times BW$), the anterior lateral approach had the middle maximum hip force ($2.23 \times BW$), and the posterior lateral approach had the largest

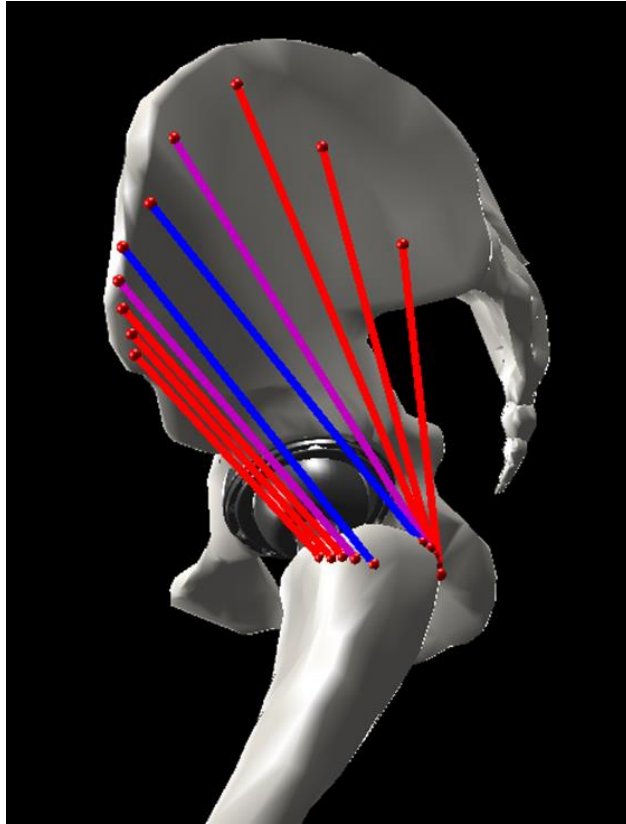


Figure 96: Muscles affected by the anterior lateral surgical approach.

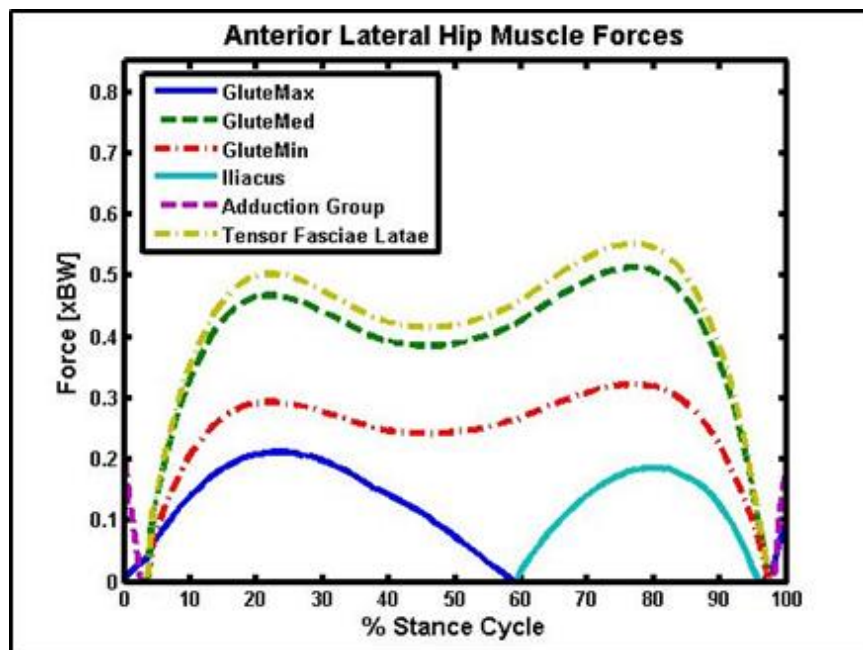


Figure 97: Muscle forces for the anterior lateral surgical approach.

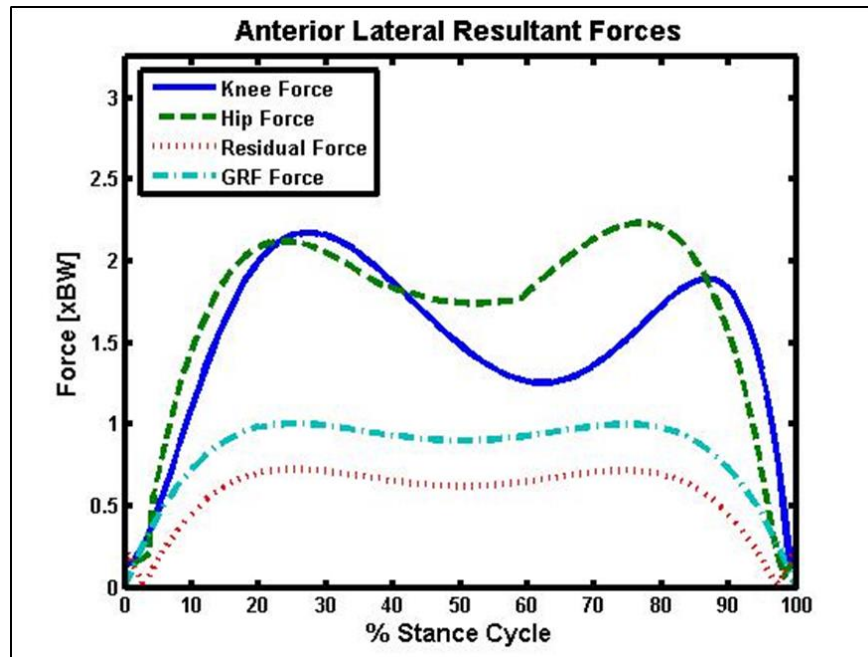


Figure 98: Resultant joint forces for the anterior lateral surgical approach.

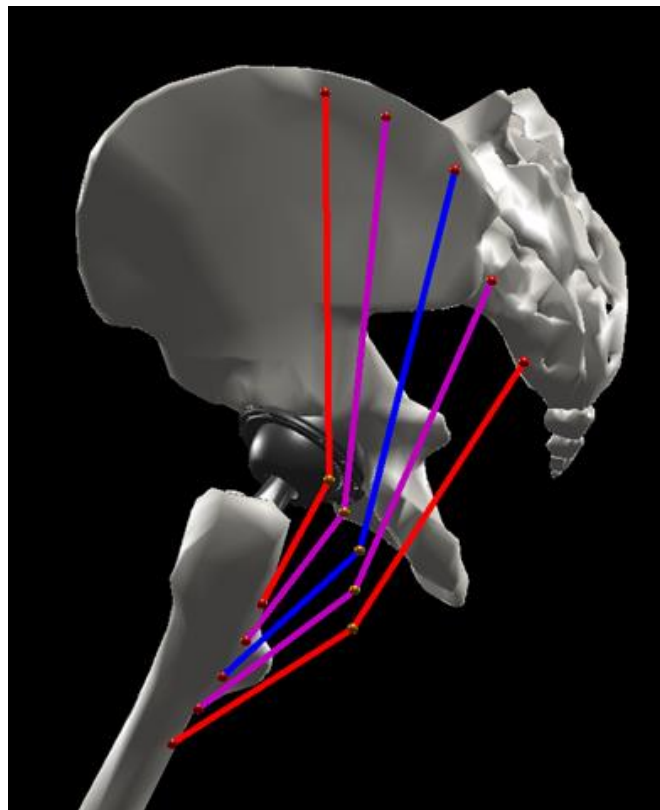


Figure 99: Muscles affected by the posterior lateral surgical approach.

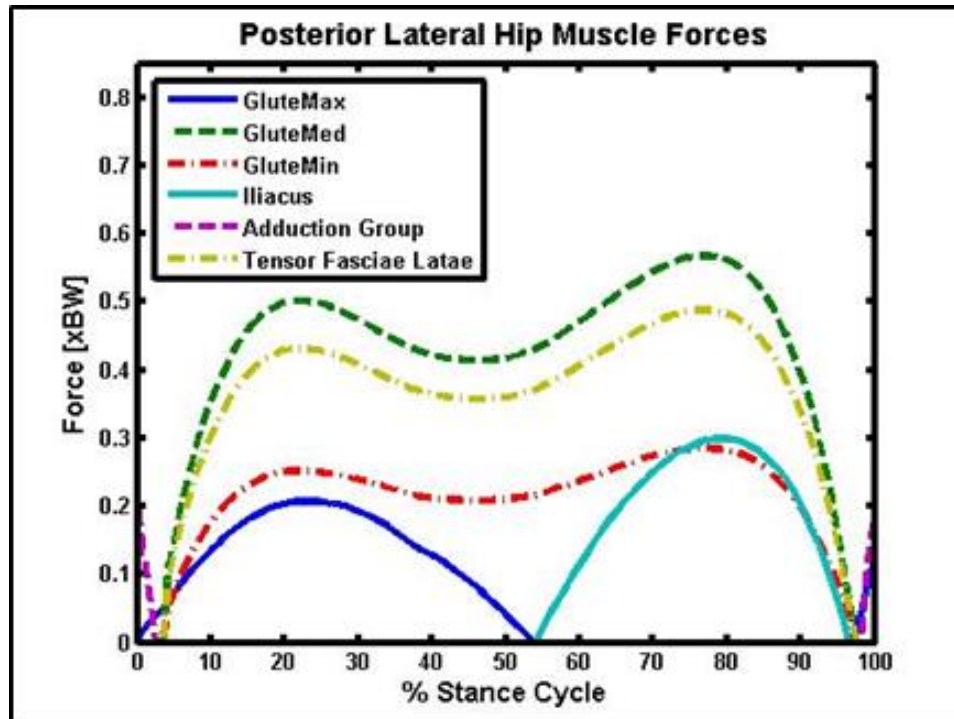


Figure 100: Muscle forces for the posterior lateral surgical approach.

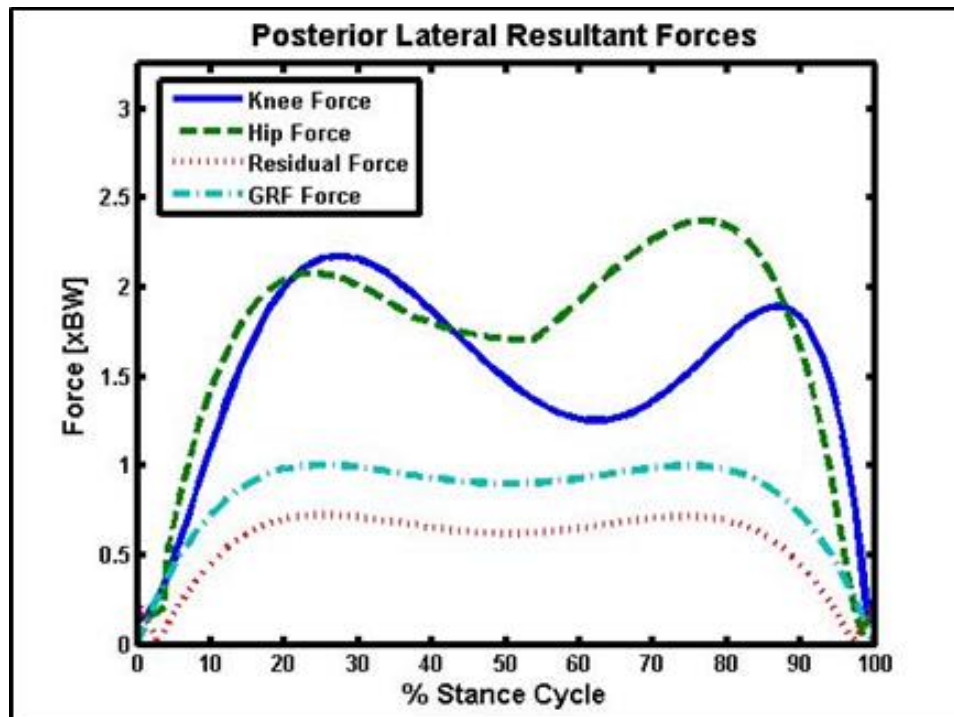


Figure 101: Resultant joint forces for the posterior lateral surgical approach.

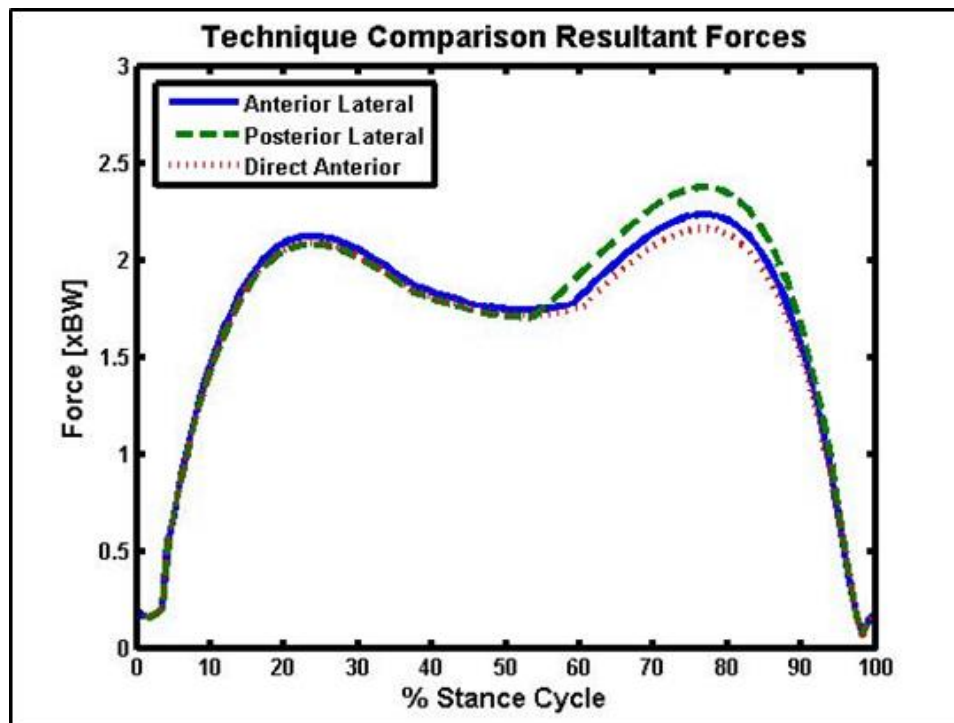


Figure 102: Technique comparison resultant forces during the stance phase of gait.

maximum hip force ($2.37 \times BW$). The increases seen in the anterior lateral and posterior lateral resultant forces can be attributed to the increased forces in the gluteus minimus and iliacus muscles, respectively. Similarly, by not cutting through any muscles (the direct anterior approach), the hip muscles operate closer to their maximum efficiency, and hence there are no hip force increases due to excess muscle compensations. Overall, by simply weakening various muscle groups, the exact same patient can experience potential resultant hip force increases of up to $0.21 \times BW$.

7.1.2 Surgical Technique Comparison Analysis for 30 Patients

A second surgical technique comparison analysis was conducted to further analyze the effects of different surgical techniques. This time, however, the results were based purely on the kinematics of patients analyzed by fluoroscopy, and muscle weakening was not incorporated into the model. In other words, the results found in this analysis were obtained purely from motion differences and not from hard-coded muscular differences. Kinematics during the stance phase of the gait activity, obtained using fluoroscopy and a 3D-to-2D registration technique, were input for 30 patients.

Fluoroscopically, the study found that there was no statistical difference in hip separation between the three surgical techniques. During swing phase, 3/10 Direct Anterior patients, 1/10 Anterior Lateral patients, and 4/10 Posterior Lateral patients experienced separation greater than 1.0mm, and there was no statistical difference in average maximum separation between surgical methods. During stance phase, 3/10 Direct Anterior patients, 1/10 Anterior Lateral patients, and 6/10 Posterior Lateral patients experienced separation greater than 1.0mm. During stance phase, there was no

statistical difference in average maximum separation between surgical methods. Swing and stance phase data for all three groups can be found in Table 2.

Based purely on kinematics, there was no statistical difference in maximum compressive hip forces due to the three surgical techniques. In general, the peak compressive hip force was slightly lower during the first half of stance phase and slightly higher during the second half of stance phase. The Posterior Lateral approach yielded an average maximum hip force of $2.44 \pm 0.38 \times BW$, the Direct Anterior approach yielded an average maximum hip force of $2.70 \pm 0.27 \times BW$, and the Anterior Lateral approach yielded an average maximum hip force of $2.54 \pm 0.57 \times BW$. Figure 103a shows a comparison of the cumulative average for each approach, determined by averaging all patient force magnitudes together at each individual time step, Figure 103b shows de-identified patient hip force data for all analyzed patients in each group, Figure 103c shows a comparison of the average ligament tension for each approach, and Figure 103d shows a comparison of the average gluteus maximus for each approach.

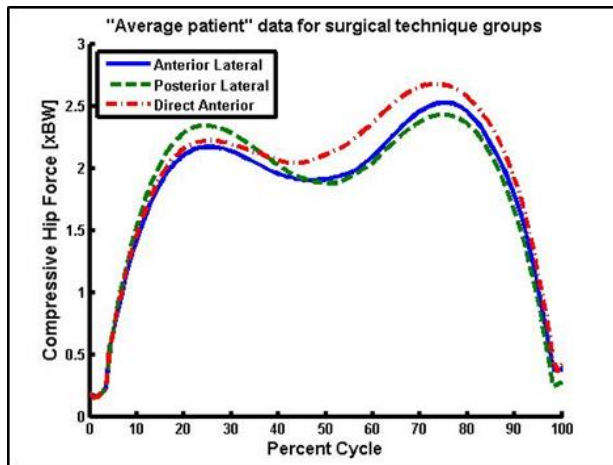
7.1.3 Separation Analysis for 30 Patients

A third analysis was conducted to determine the effects that hip separation during stance phase has on compressive hip forces and tensile capsular ligament forces. To analyze hip separation, patients were divided into two groups: 1) all subjects who experienced $>1.0\text{mm}$ of hip separation, and 2) all subjects who experienced $<1.0\text{mm}$ of hip separation.

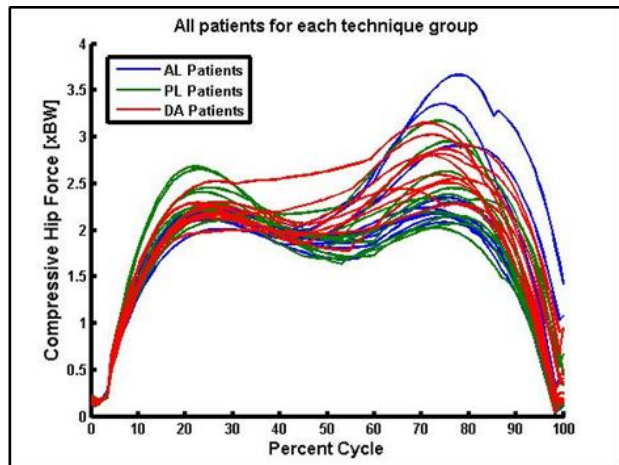
For all patients, the average maximum hip force was $2.62 \pm 0.38 \times BW$. Among all patients, thirty-three percent experienced hip separation greater than 1.0mm . Patients

Table 2: Separation data for the Direct Anterior (DA) group, Anterior Lateral (AL) group, and Posterior Lateral (PL) groups.

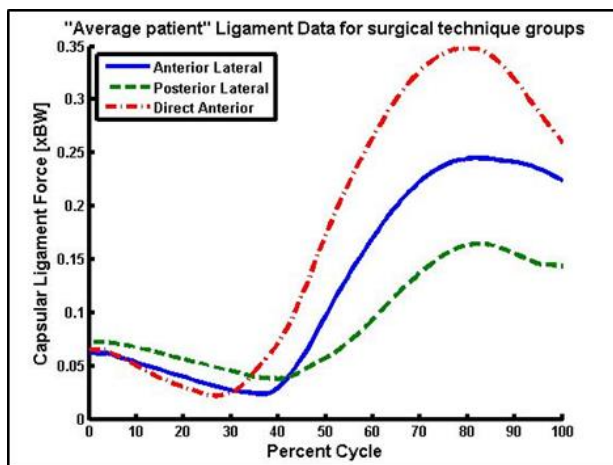
	Swing Phase Sep [mm]			Stance Phase Sep [mm]			Overall Sep [mm]		
	DA	AL	PL	DA	AL	PL	DA	AL	PL
Avg ± St Dev	0.9 ± 0.7	0.6 ± 0.3	1.0 ± 0.4	0.9 ± 0.3	0.8 ± 0.6	1.0 ± 0.3	1.1 ± 0.6	0.8 ± 0.6	1.2 ± 0.4
Maximum	2.5	1.4	1.6	1.4	2.3	1.5	2.5	2.3	1.6
Minimum	0.3	0.3	0.3	0.5	0.4	0.5	0.6	0.5	0.5



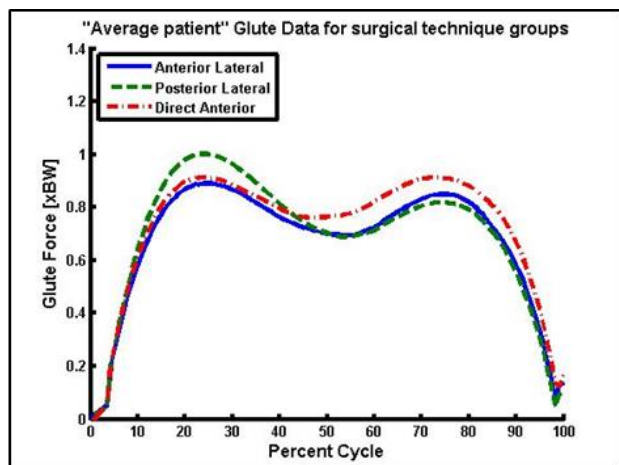
(a)



(b)



(c)



(d)

Figure 103: Comparisons of (a) average compressive hip force, (b) individualized compressive hip forces, (c) average ligament tension, and (d) average gluteus maximus force.

who separated experienced 0.20x BW greater average hip force, 0.13x BW greater average capsular ligament tensile force, and 0.07x BW greater average total glute muscle force than those who did not separate (Table 3).

Additionally, all patients were averaged together to create “average” patient data, shown in Figure 104. For the “average separation patient,” the maximum hip force was 2.64x BW, the maximum ligament force was 0.35x BW, and the maximum total glute force was 0.98x BW. For the “average non-separation patient,” the maximum hip force was 2.49x BW, the maximum ligament force was 0.21x BW, and the maximum total glute force was 0.91x BW, yielding differences of 0.15x BW, 0.14x BW, and 0.07x BW, respectively. Finally, when the patient with the maximum and minimum amount of separation were compared, the dramatic results of hip separation can be seen (Figure 104d).

7.2 Component Positioning Analyses

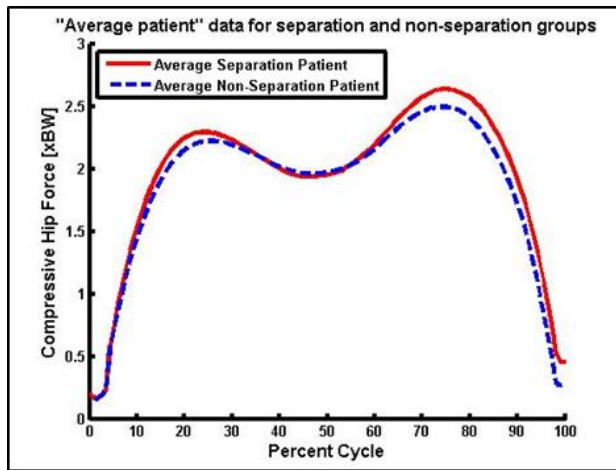
A total of six different detailed analyses were conducted to determine the effects that acetabular cup alignment, leg length discrepancy, hip offset error, and femoral component version have on hip mechanics and separation. Specific data figures can be found in the referred appendices. Presented below are figures to aid visually in understanding the analysis performed, as well as brief summaries of the results.

7.2.1 Cup Alignment during Swing Phase

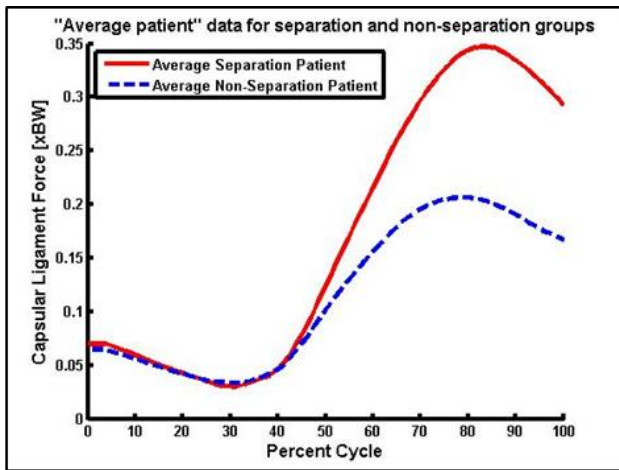
The first in-depth analysis conducted with the forward solution model was to determine the effects that acetabular cup positioning has on hip separation during swing

Table 3: Averages of both maximum compressive hip force, maximum tensile capsular ligament force, and maximum total glute force for patients who experienced >1mm separation and patients who did not experience separation.

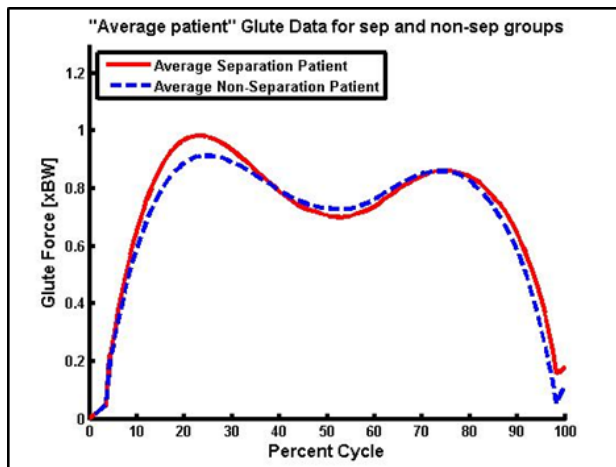
	Separation Group [xBW]	No Separation Group [xBW]
Average Maximum Hip Force	2.75 ± 0.40 (2.24 – 3.67)	2.55 ± 0.37 (2.14 – 3.35)
Average Maximum Ligament Force	0.39 ± 0.33 (0.07 – 1.11)	0.26 ± 0.24 (0.06 – 0.80)
Average Maximum Glute Force	1.02 ± 0.10 (0.85 – 1.21)	0.95 ± 0.10 (0.85 – 1.17)



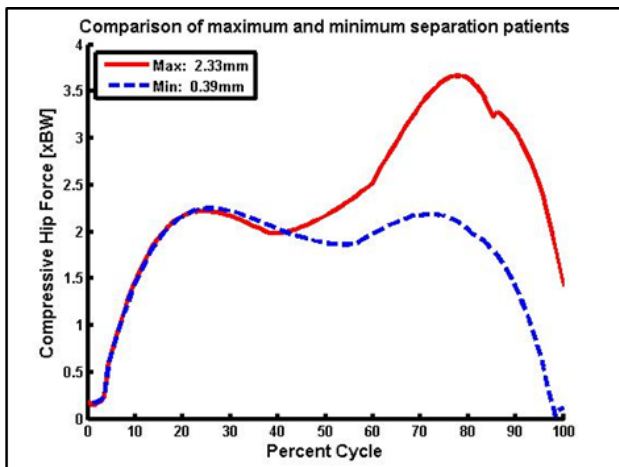
(a)



(b)



(c)



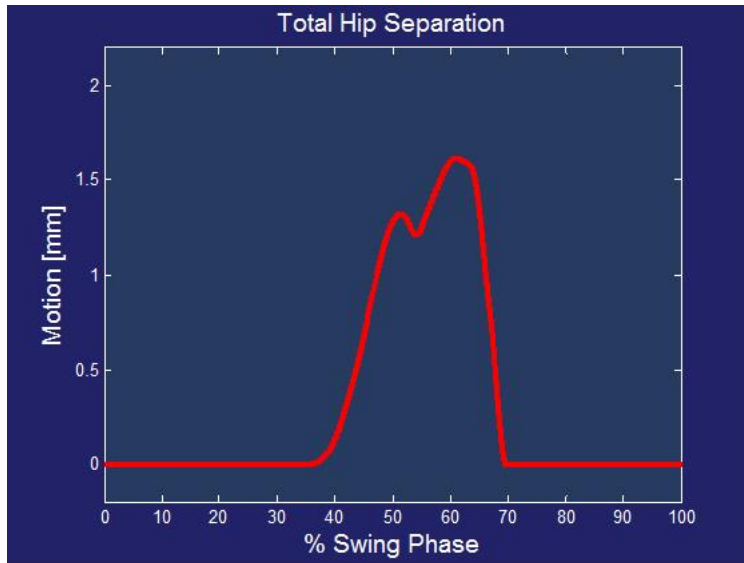
(d)

Figure 104: Comparisons of (a) average compressive hip force, (b) average ligament tension, and (c) average gluteus maximus force, separated by surgical technique, as well as (d) a comparison of patients with the maximum and minimum amount of hip separation.

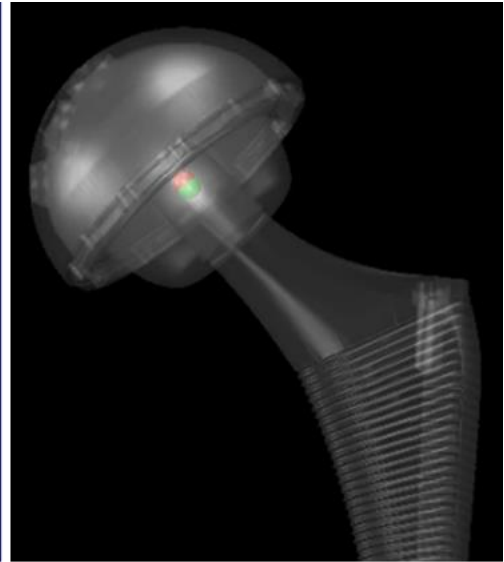
phase. This analysis was conducted on one single “patient” under 11 different alignment scenarios. Separation during swing phase is generated primarily by medial and superior shifts of the acetabular cup. Appendix A, found in the Supplemental Document, shows all figures associated with this analysis; each pair figures in Appendix A is accompanied with an underlying explanation of what scenario is causing the separation. Below, Figure 105 shows the separation curve due to a 10mm medial shift of the cup, and Figure 106 shows the separation curve associated with a 10mm superior shift of the cup. In Appendix A, there are certain cases (such as a 0.5cm medial shift) where the capsular ligaments, although looser, remain tight enough to keep the femoral head positioned within the cup. Table 4 shows a summary of hip separation magnitudes for all analyzed cases. Combinations of both medial and superior shifting appear to yield the largest magnitudes of hip separation.

7.2.1.1 Hip Capsule Management

An additional analysis was simultaneously conducted to determine the importance of repairing the hip capsule after surgery. Even with a properly aligned acetabular cup, poor capsular repair can yield instability and separation during swing phase. Figure 107 shows minor instability (0.15mm of separation) associated with a complete resection of the anterior capsule, Figure 108 shows minimal instability associated with a complete resection of the posterior capsule, and Figure 109 shows complete dislocation associated with a resection of the entire hip capsule.

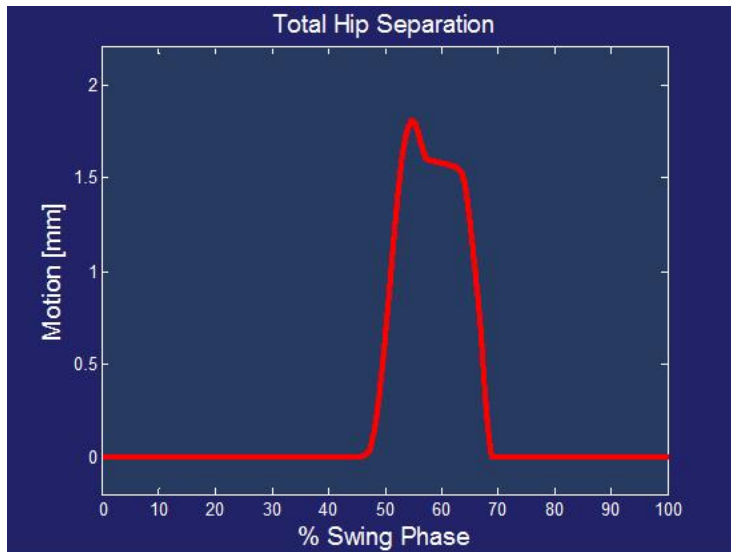


(a)

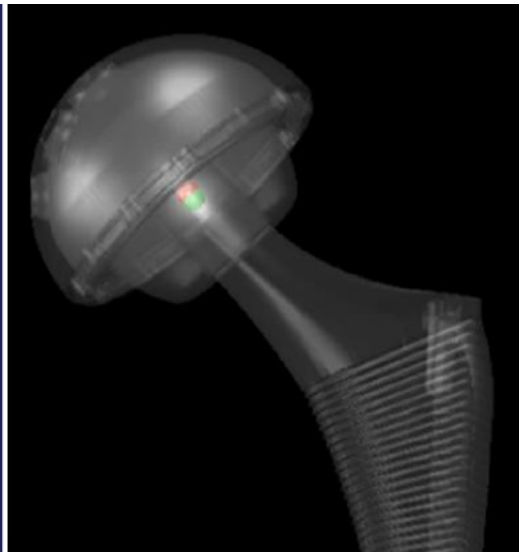


(b)

Figure 105: (a) graphical and (b) visual representation of 1.6mm of swing phase separation. 1cm medial shift of the cup, 40° inclination, 15° anteversion.



(a)

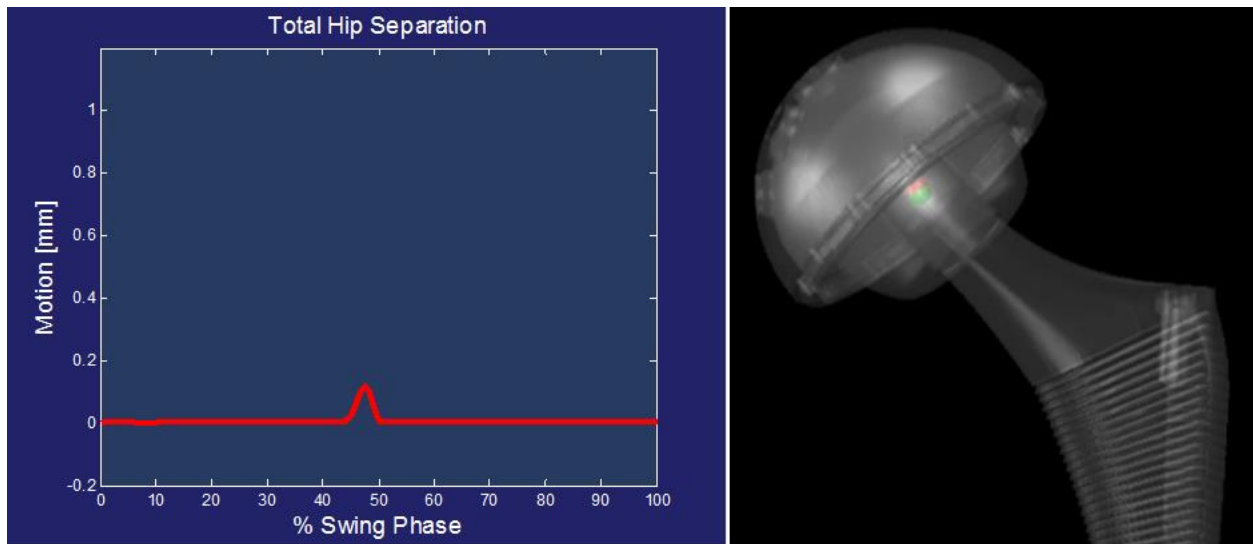


(b)

Figure 106: (a) graphical and (b) visual representation of 1.8mm of swing phase separation. 1cm superior shift of the cup, 40° inclination, 15° anteversion.

Table 4: Separation during swing phase summary.

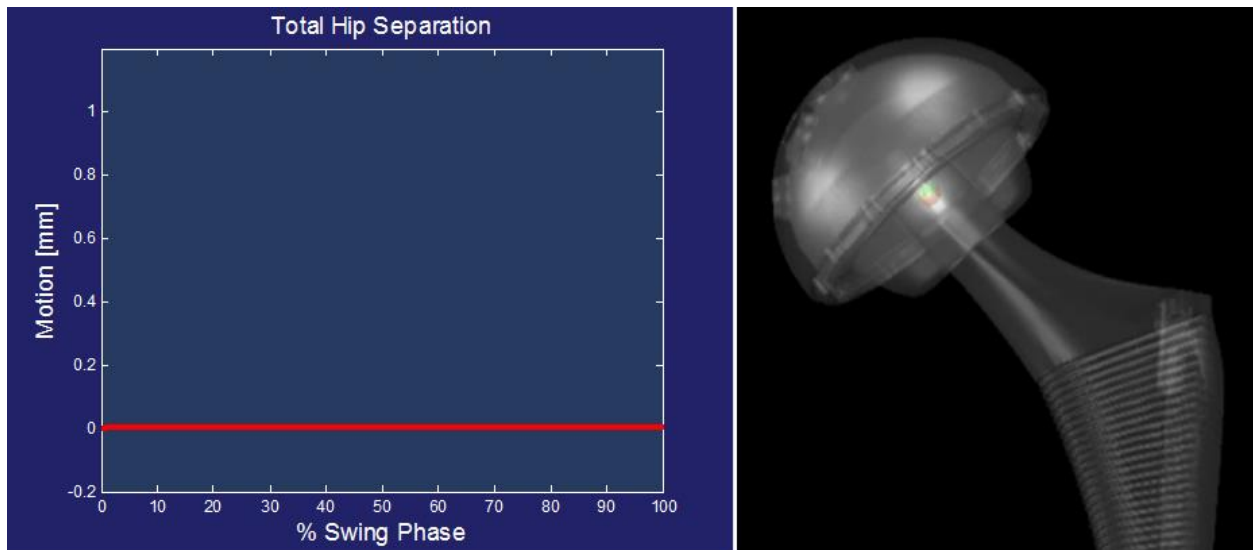
Shift direction	Shift magnitude [cm]	Inclination [degrees]	Version [degrees]	Separation [mm]
None	0	40	15	0.0
Medial	1	40	15	1.6
Medial	1	40	35	3.4
Medial	0.5	40	15	0.0
Medial	0.5	40	35	0.0
Superior	1	40	15	1.8
Superior	1	40	35	3.5
Superior	0.5	40	15	0.8
Superior	0.5	40	35	2.0
37° Supero-medial	0.71	40	15	3.3
37° Supero-medial	0.71	40	35	4.9



(a)

(b)

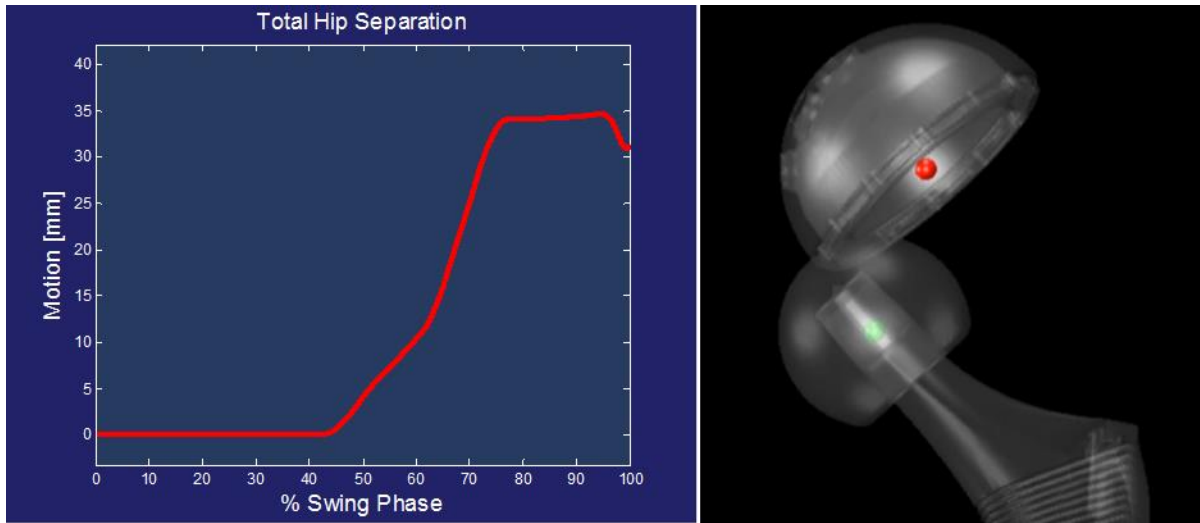
Figure 107: (a) graphical and (b) visual representation of 0.15mm of swing phase separation due to anterior capsule resection.



(a)

(b)

Figure 108: (a) graphical and (b) visual representation of 0mm of swing phase separation due to posterior capsule resection.



(a)

(b)

Figure 109: (a) graphical and (b) visual representation of hip dislocation due to complete capsule resection.

7.2.2 Cup Alignment during Stance Phase

In general, it is hypothesized that stance phase hip separation is of greater consequence than swing phase separation, due to the weight-bearing, higher-force nature of the activity. Hence, the cup alignment during stance phase analysis is more detailed than the cup alignment during swing phase analysis.

For this analysis, the four patients described in the methods section were analyzed under 75 identical conditions, including accurate cup alignment, intended medial and superior shifting of the cup, unintended medial and superior shifting of the cup, and variations in cup version angles, to determine the effects that acetabular component position has on hip separation.

The four patients were each analyzed under three scenarios, depending on what the surgeon “intended” to do versus what he/she “actually” did, outlined visually in Figure 110. Scenario 1 assesses an intended and actual shift purely in the medial direction. Scenario 2 assesses an intended and actual shift in a supero-medial direction, approximately 37° from the medial axis. Scenario 3 assesses an intended medial shift but with the actual reaming direction being in the supero-medial direction. In each scenario, it was assumed that the surgeon intended to shift the acetabular cup either 0mm, 2mm, or 5mm in a specified direction. The overall analysis thus took into consideration 1) whether the surgeon executed the intended shift perfectly or whether he/she accidentally over-reamed the cup, and 2) whether the surgeon reamed in the direction he/she intended or whether he/she accidentally reamed in the incorrect direction.

A sample medial shift is shown in Figure 111, and a sample supero-medial shift is shown in Figure 112. Additionally, each scenario was analyzed with the cup placed at 0°

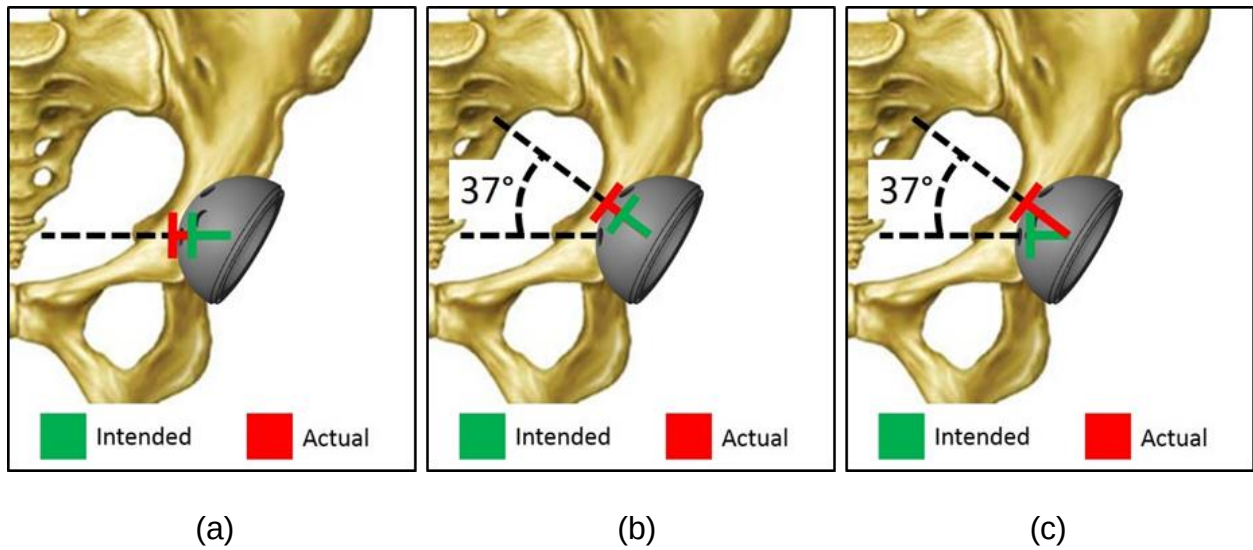


Figure 110: (a) Scenario 1: intended medial shifts, actual medial shifts. (b) Scenario 2: intended supero-medial shifts, actual supero-medial shifts. (c) Scenario 3: intended medial shifts, actual super-medial shifts. Images modified from (humananatomychart.us).

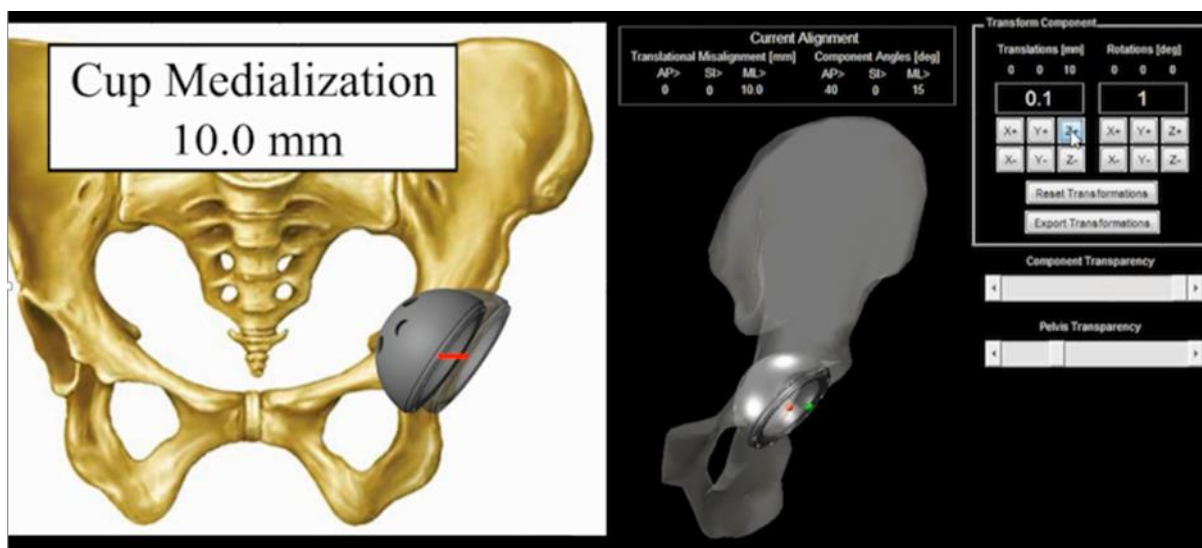


Figure 111: Cup medialization shown visually and in the model GUI. Left image modified from (humananatomychart.us).

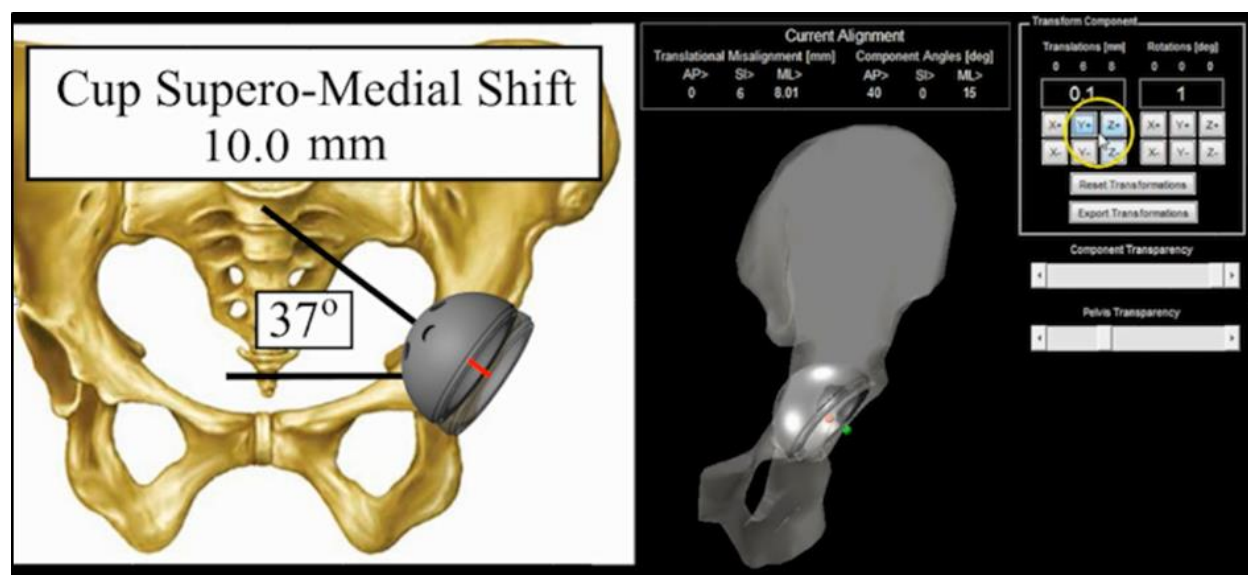


Figure 112: Cup supero-medialization shown visually and in the model GUI. Left image modified from (humananatomychart.us).

Retroversion, 15° Anteversion, and 35° Anteversion (shown in Figure 113) to determine the effects of cup implantation angles.

Figure 114, Figure 115, and Figure 116 show an overall summary of the analysis. In summary, while the direct relationship between component alignment and hip separation varies between patients, it was generally determined that medial and superior shifts of the acetabular cup both have a positive correlation with increased hip separation. Furthermore, the larger the surgeon's intended shift in the acetabular cup, either medially or superiorly, the smaller the margin of error that the surgeon may have before inducing hip separation in the patient post-operatively. In other words, a surgeon who does not intend to medialize the cup at all (a 0mm intended shift) but accidentally over-reams 4mm may still not induce hip separation, but a surgeon who intends to medialize the cup 5mm and accidentally over-reams an additional 2mm may induce hip separation. Graphical representations of the allowable error zones can be seen in Figure 117. The blue shaded areas represent the zones where the surgeon is intending on placing the cup, the green shaded areas represent the regions where the surgeon can actually place the cup and not induce hip separation, and the red shaded areas represent the region where separation will occur. Notice that as the blue region gets bigger, the green region gets thinner. Numeric estimations of intended shifts versus allowable errors can be found in Table 5.

Patient-specific figures for every scenario, including intended and unintended shifts as well as version changes, can be found in Appendix B, which is in the Supplemental Document. In each case, the specific scenarios can be seen in the graphic as well as the figure titles. Additionally, while certain patients appeared to show a

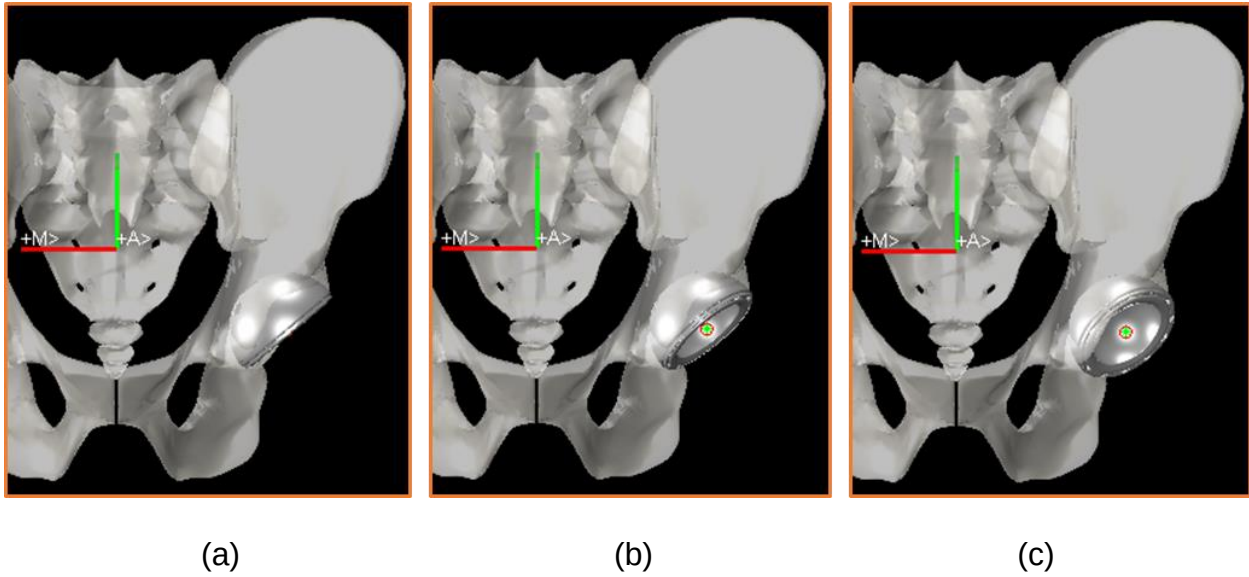


Figure 113: (a) 0° Retroversion. (b) 15° Anteversion. (c) 35° Anteversion.

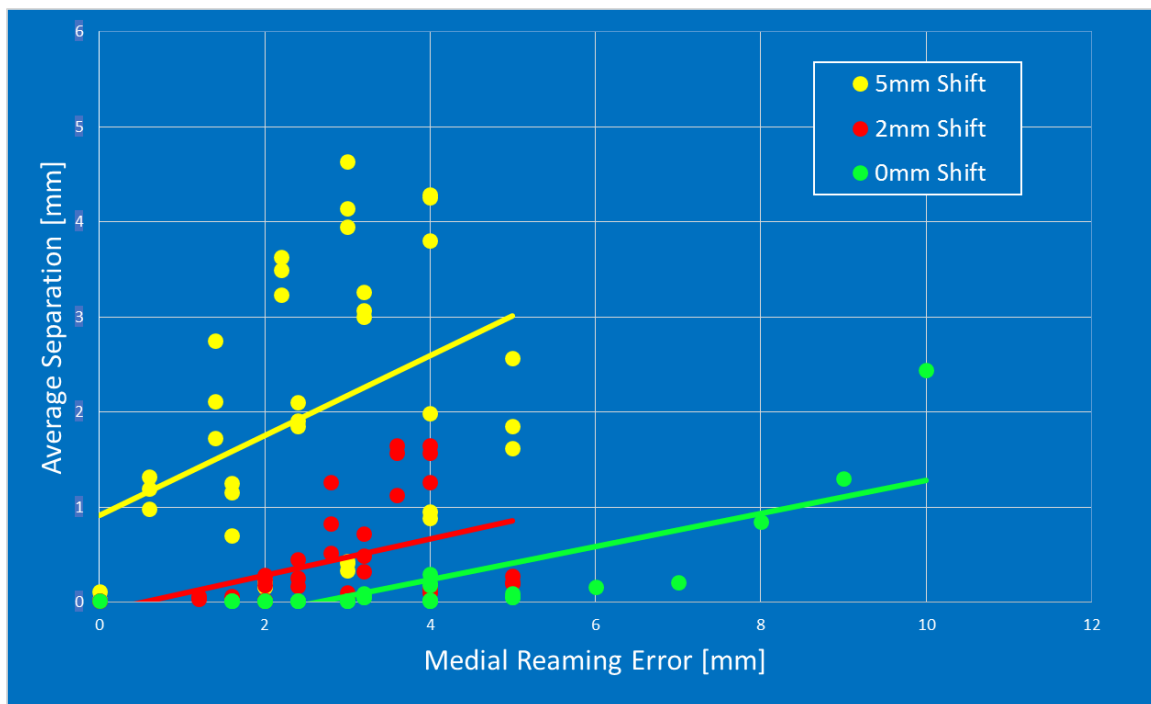


Figure 114: Hip separation versus medial reaming errors.

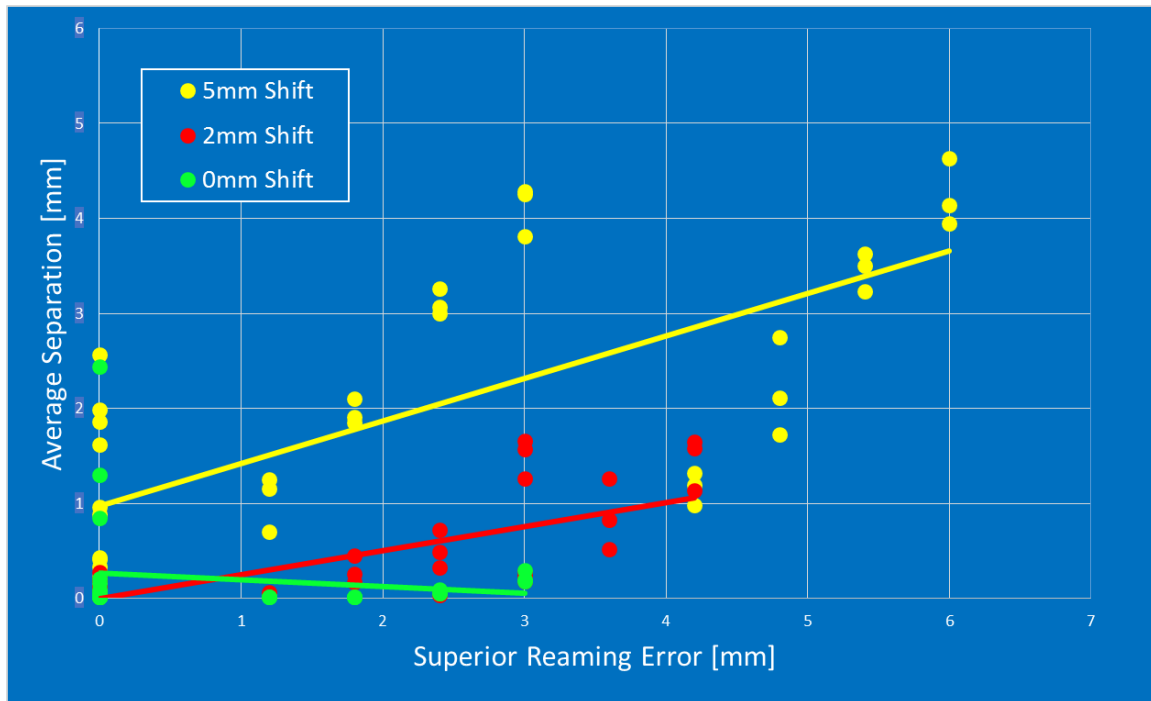


Figure 115: Hip separation versus superior reaming errors.

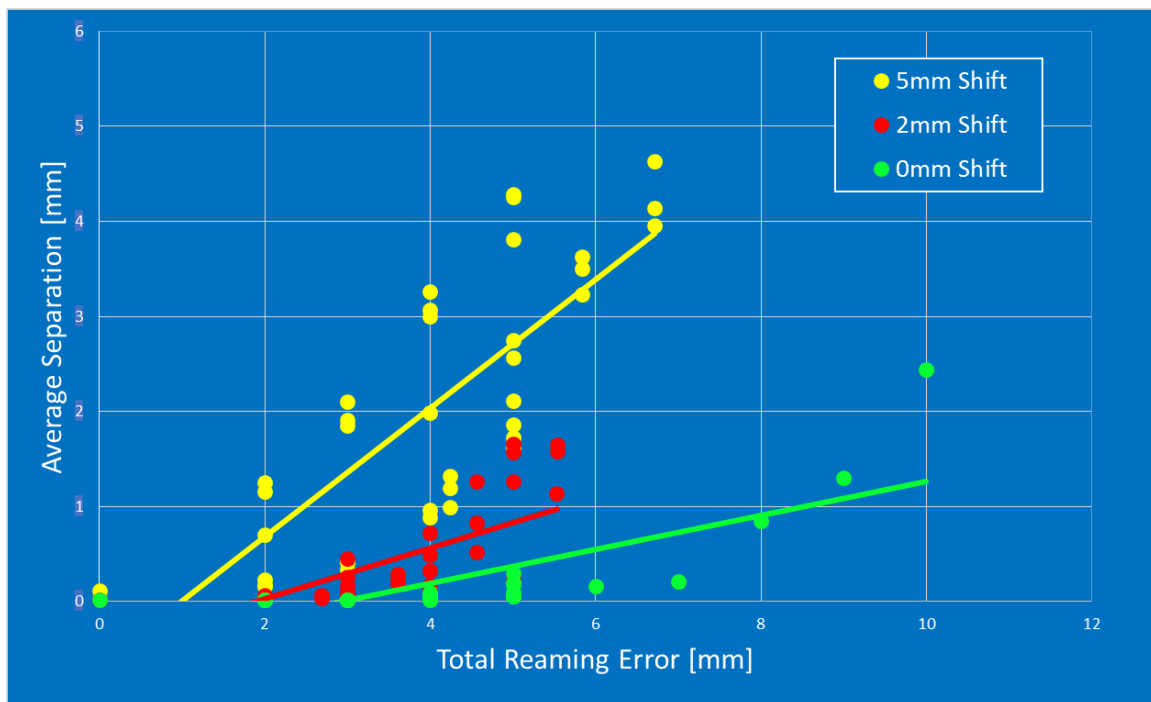


Figure 116: Hip separation versus total reaming errors.

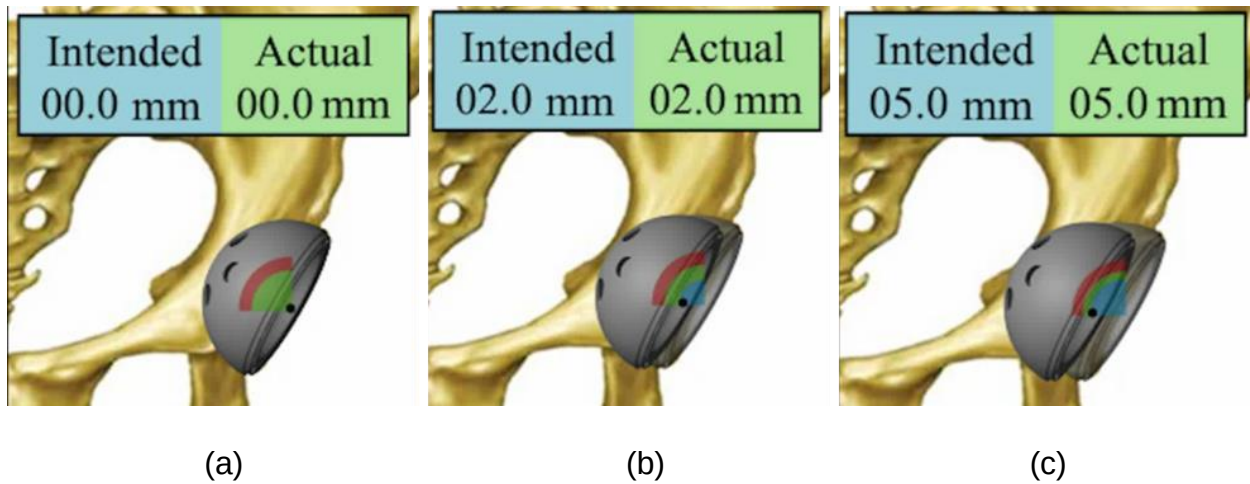


Figure 117: (a) 0mm intended shifts. (b) 2mm intended shifts. (c) 5mm intended shifts.

Images modified from (humananatomychart.us).

Table 5: Intended shifting with the corresponding “Allowable Error” zone.

Intended Shifts [mm]	Allowable Error [mm]
0	6
2	3
5	2

correlation between cup angle and separation, the data was too diverse to claim that cup angle has as clear of an effect on hip separation as cup translations.

7.2.3 Leg Length Discrepancy Analysis

In the model, leg length discrepancies are introduced by errors in femoral component placement, i.e. if the surgeon misplaces the stem in such a way that the center of the femoral head sits superiorly or inferiorly, perhaps by either overreaming or underreaming the femoral canal or by reaming at an incorrect angle. This error would 1) physically change the length of the leg, and 2) change the location of the implanted femoral head center compared to the native femoral head center. A visual example can be seen in Figure 118. The leg length analysis was conducted on both swing and stance phase.

7.2.3.1 Stance Phase: Increasing Leg Length

In general, it was determined that increases in leg length did not have a substantial effect on hip separation during stance phase. However, unintentionally increasing the length of the leg does yield increased stretching and tension in the capsular ligaments, and thus increases in the forces and stresses present on the femoral head and acetabular liner are observed. Figure 119, Figure 120, Figure 121, and Figure 122 show the effects of increasing the length of the implanted leg by 5mm, 10mm, and 20mm for all four patients during stance phase.

7.2.3.2 Stance Phase: Decreasing Leg Length

It was also determined that decreases in leg length did not have a substantial effect on hip separation during stance phase. Unintentionally decreasing the length of the leg

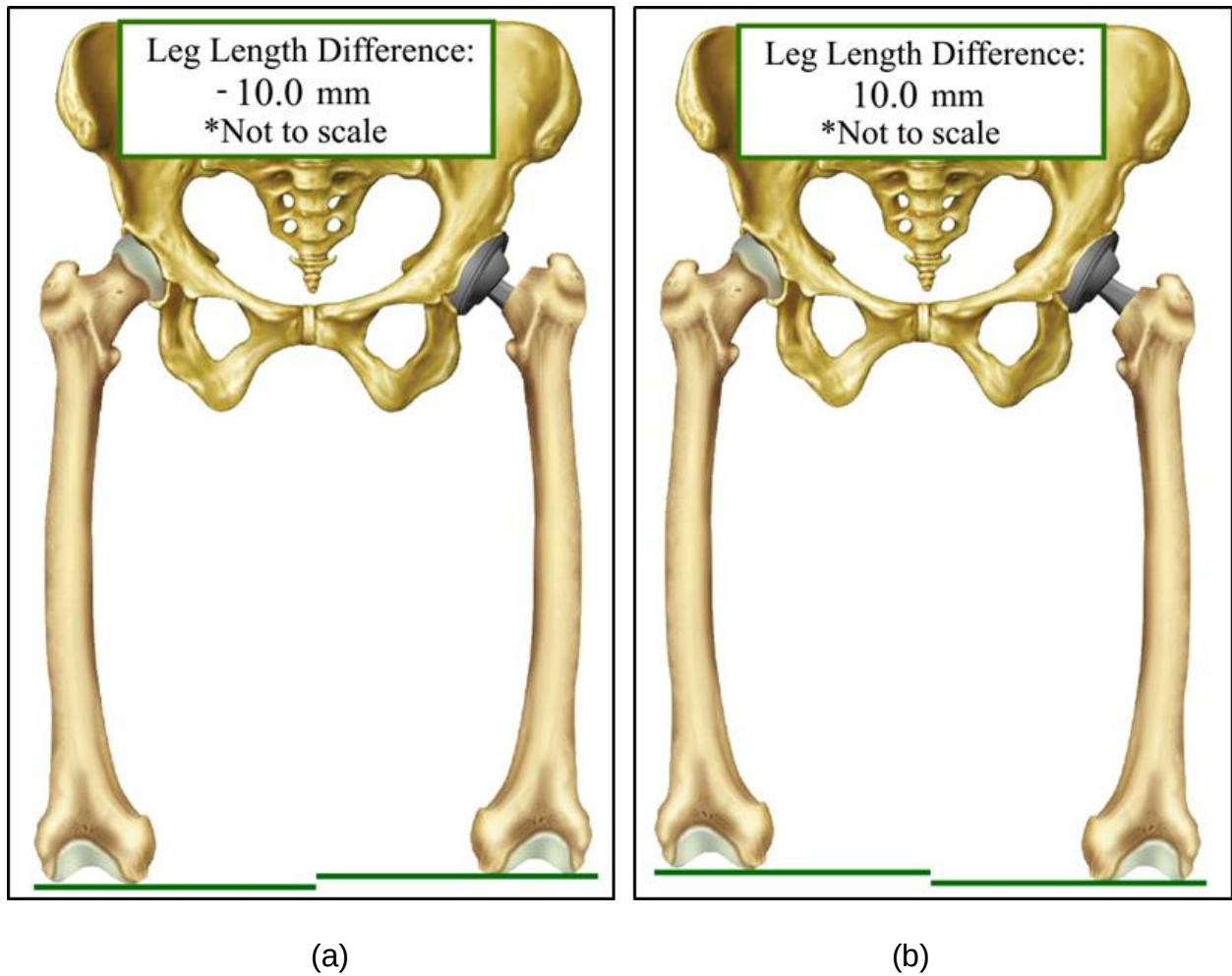
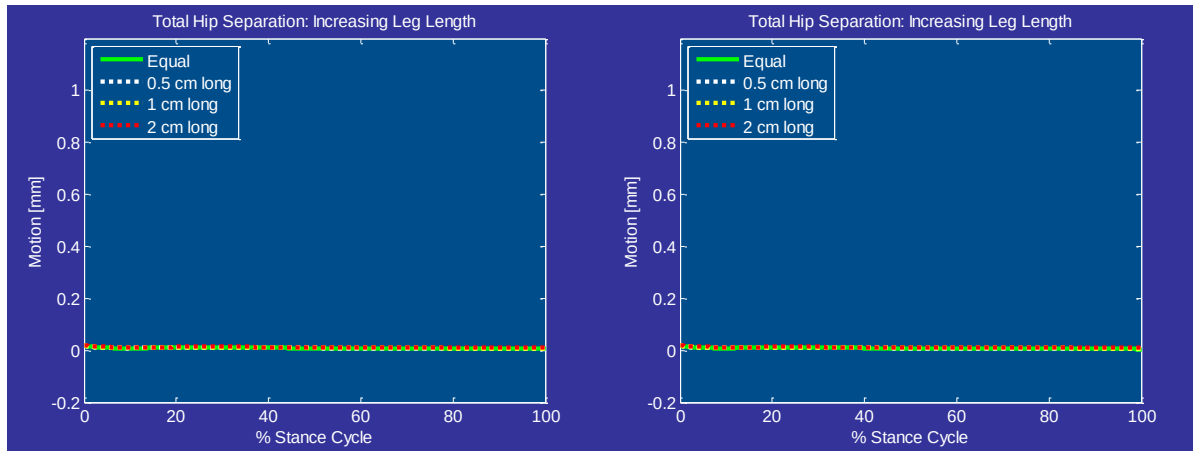
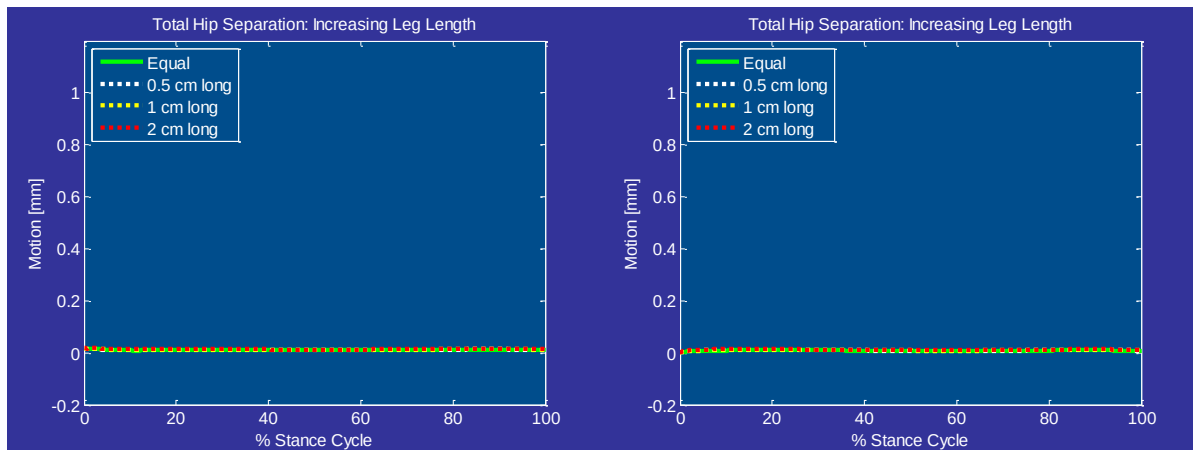


Figure 118: (a) shortening the leg, and (b) lengthening the leg. Images modified from (humananatomychart.us) and (theodysseyonline.com).



(a)

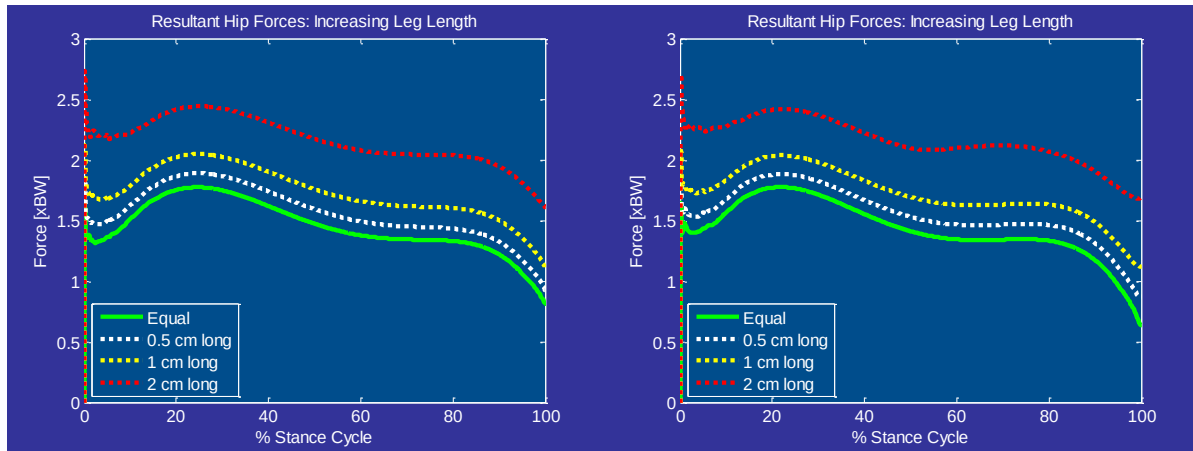
(b)



(c)

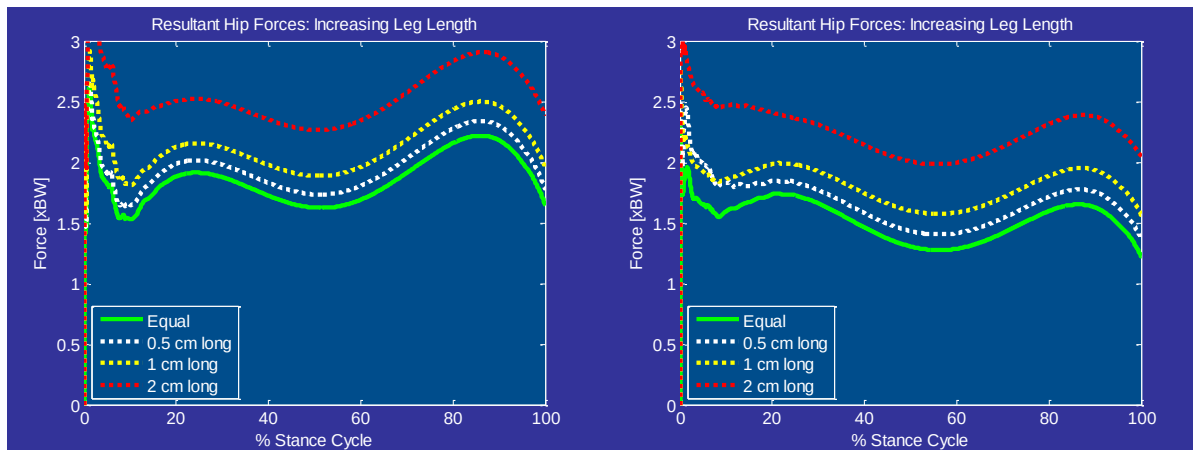
(d)

Figure 119: Leg length analysis, increasing leg length, stance phase hip separation effects for Patients (a) 1, (b) 2, (c) 3, and (d) 4.



(a)

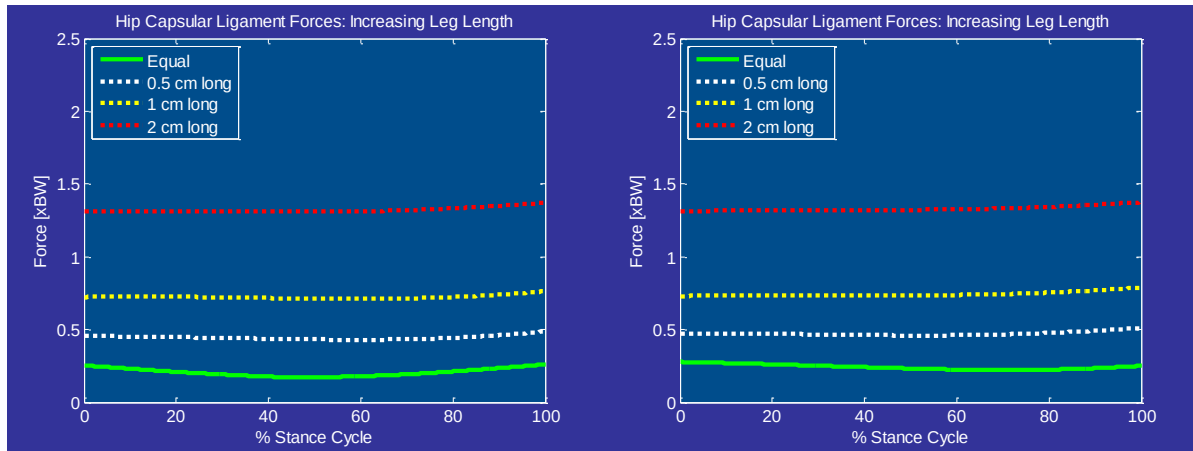
(b)



(c)

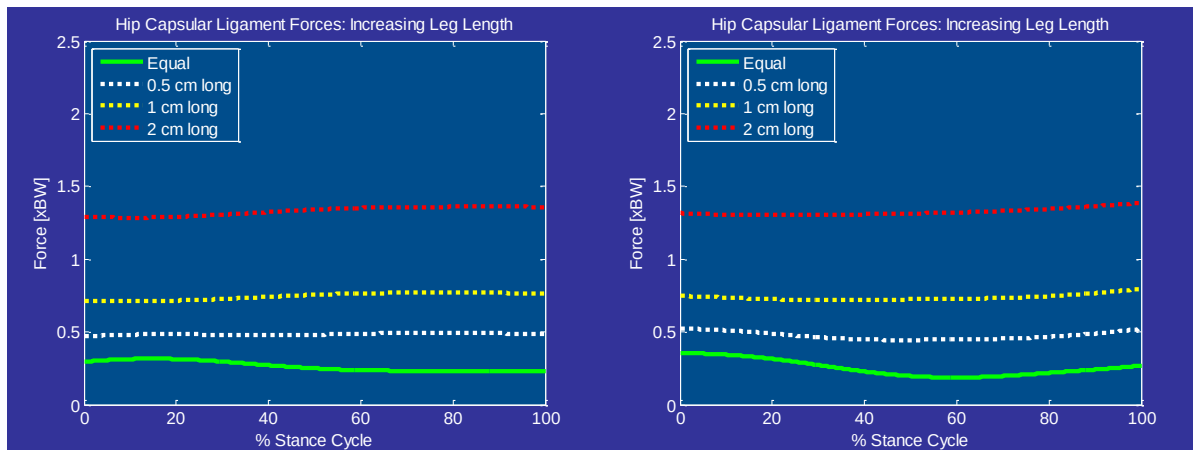
(d)

Figure 120: Leg length analysis, increasing leg length, stance phase hip force effects for Patients (a) 1, (b) 2, (c) 3, and (d) 4.



(a)

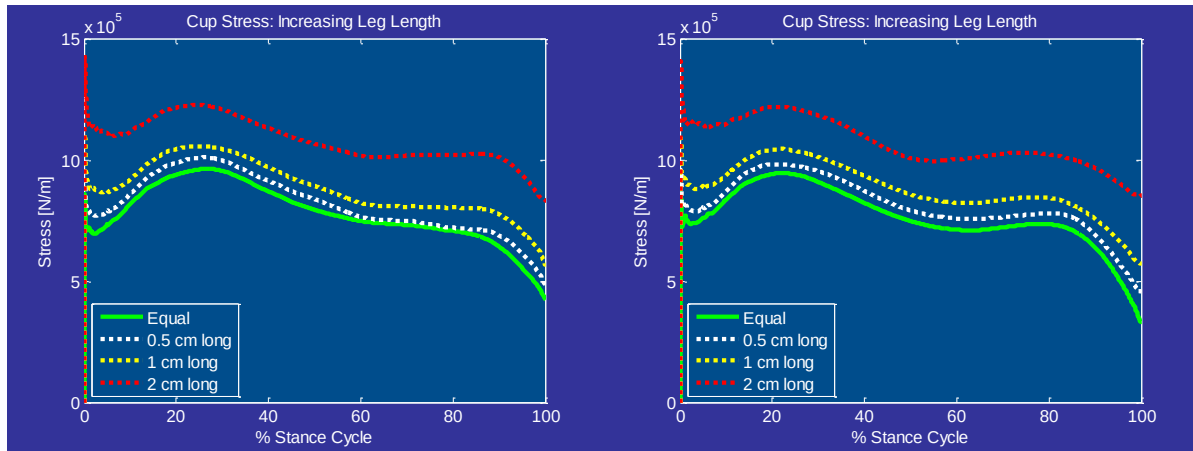
(b)



(c)

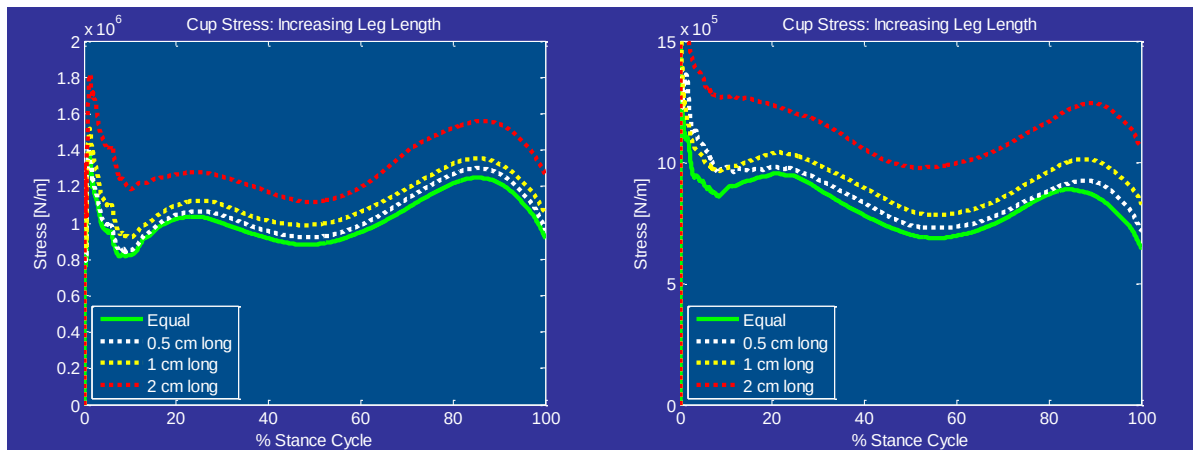
(d)

Figure 121: Leg length analysis, increasing leg length, stance phase capsular ligament force effects for Patients (a) 1, (b) 2, (c) 3, and (d) 4.



(a)

(b)



(c)

(d)

Figure 122: Leg length analysis, increasing leg length, stance phase cup stress effects for Patients (a) 1, (b) 2, (c) 3, and (d) 4.

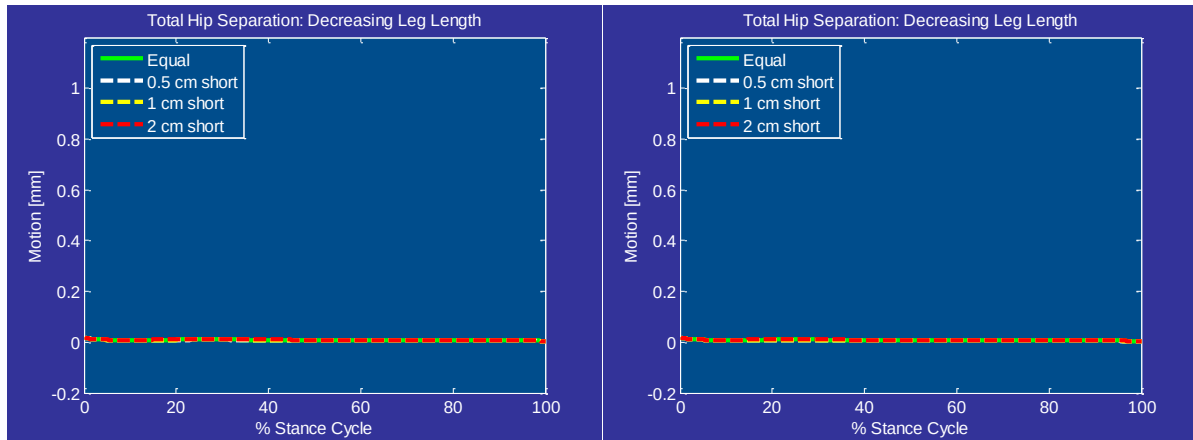
yields increased slack in the capsular ligaments. While this does not appear to have a significant effect on forces or stresses during stance phase, it will be seen later that it does have an effect during swing phase. Figure 123, Figure 124, Figure 125, and Figure 126 show the effects of decreasing the length of the implanted leg by 5mm, 10mm, and 20mm for all four patients during stance phase.

7.2.3.3 Swing Phase: Increasing Leg Length

During swing phase, it was determined that increasing the length of the leg did not yield hip separation. This is because leg length increases yield increases in capsular ligament tension, and hence the system is more resistant to hip separation. However, like with stance phase, these adjustments in leg length and ligament tension do yield to a corresponding increase in cup stresses and resultant hip forces, which can be seen in Figure 127, Figure 128, Figure 129, and Figure 130, which show the effects of increasing the length of the implanted leg by 5mm, 10mm, and 20mm for all four patients during swing phase.

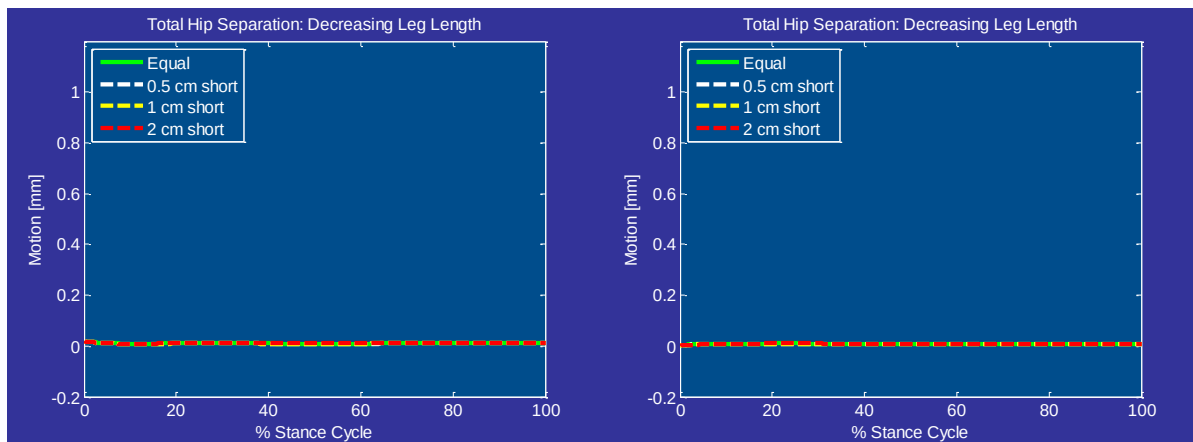
7.2.3.4 Swing Phase: Decreasing Leg Length

Decreasing leg length during swing phase was the only leg length scenario that directly led to the occurrence of hip separation. It is assumed that decreasing the leg length leads to slack in the hip capsule. However, unlike stance phase (where the system is in compression), the system is in tension during swing phase, and thus the tension of the slack capsular ligaments is not enough to keep the femoral head seated inside the cup. Interestingly, a larger decrease in leg length did not always correlate to an increase in hip separation, as one might expect (Patient 3). The reasoning behind this is still being investigated, but it is worth noting that Patient 3 is also the subject who has the longest



(a)

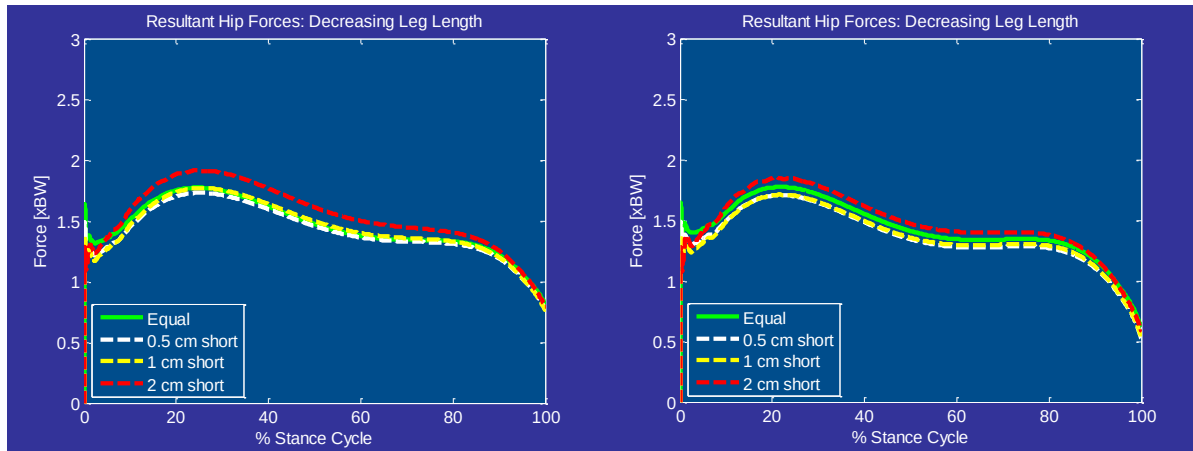
(b)



(c)

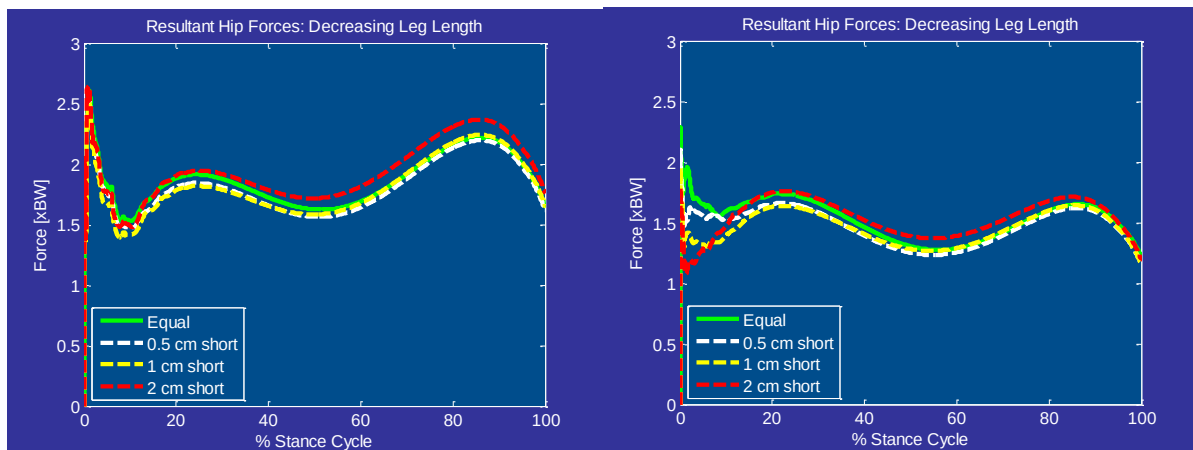
(d)

Figure 123: Leg length analysis, decreasing leg length, stance phase hip separation effects for Patients (a) 1, (b) 2, (c) 3, and (d) 4.



(a)

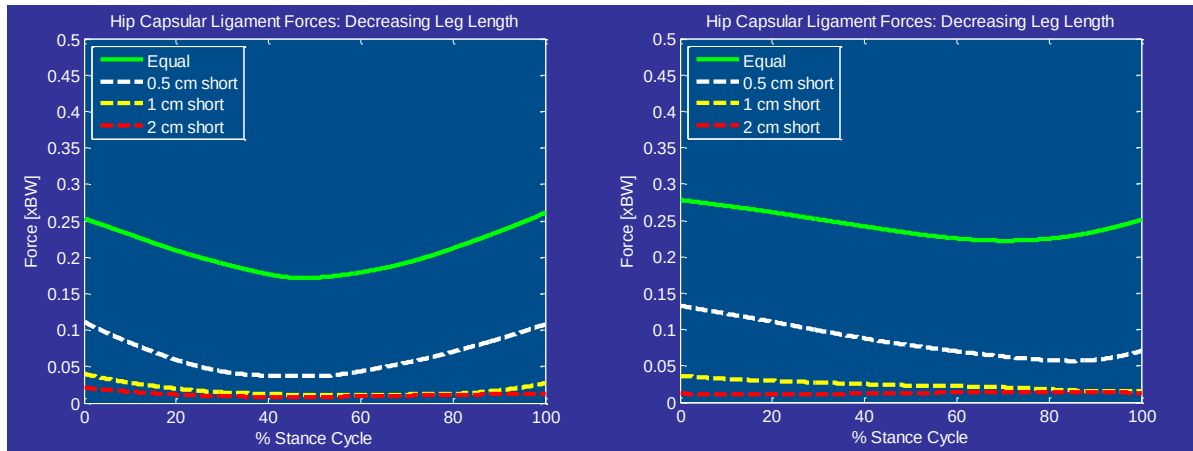
(b)



(c)

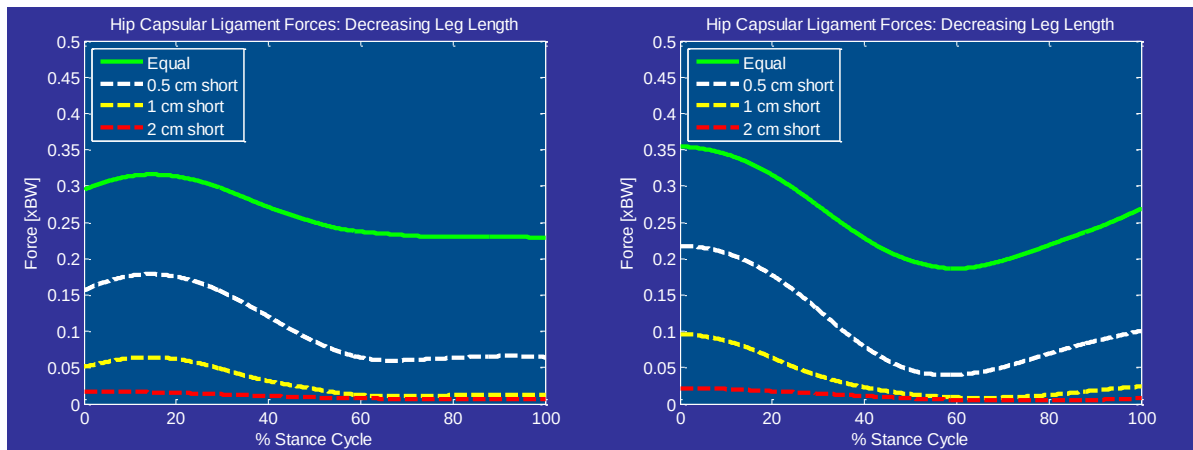
(d)

Figure 124: Leg length analysis, decreasing leg length, stance phase hip force effects for Patients (a) 1, (b) 2, (c) 3, and (d) 4.



(a)

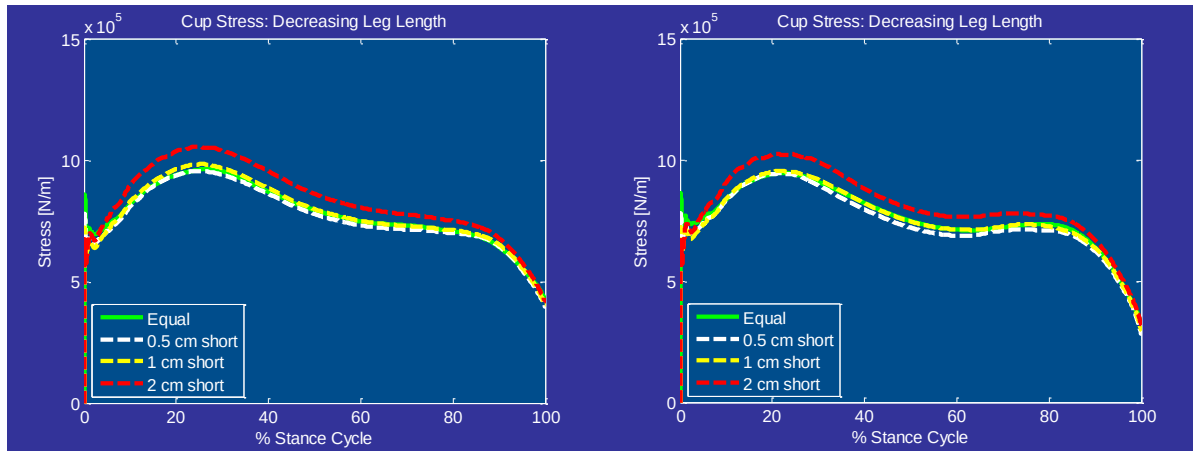
(b)



(c)

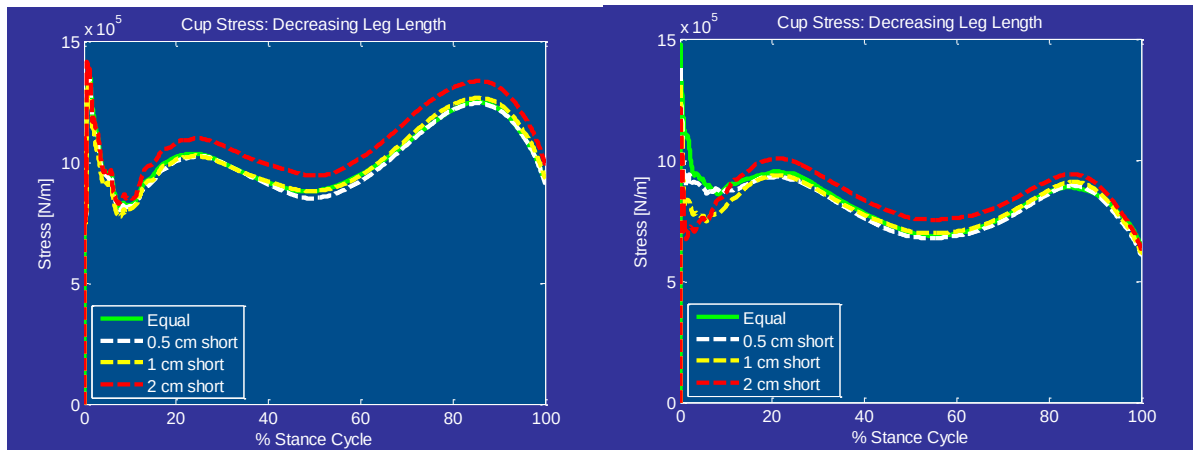
(d)

Figure 125: Leg length analysis, decreasing leg length, stance phase capsular ligament force effects for Patients (a) 1, (b) 2, (c) 3, and (d) 4.



(a)

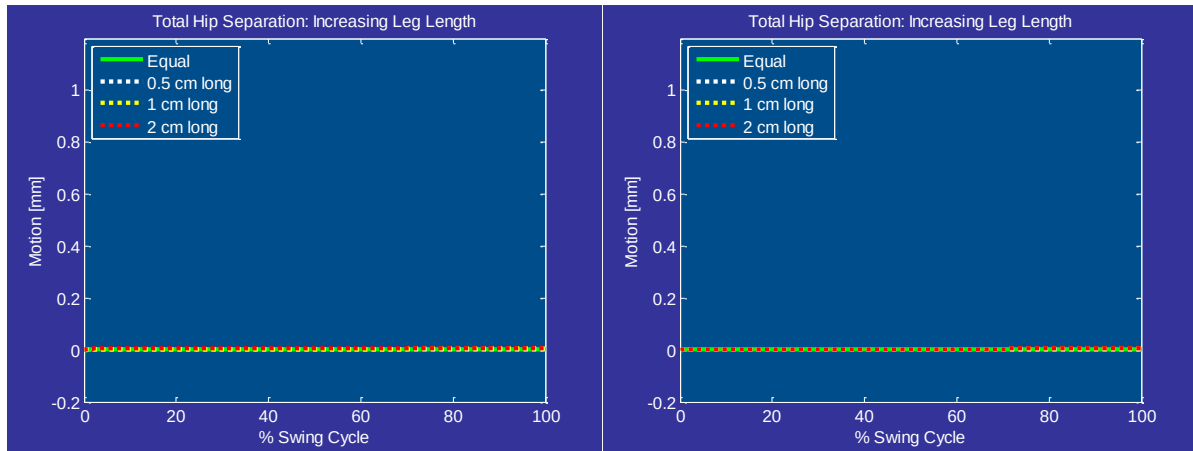
(b)



(c)

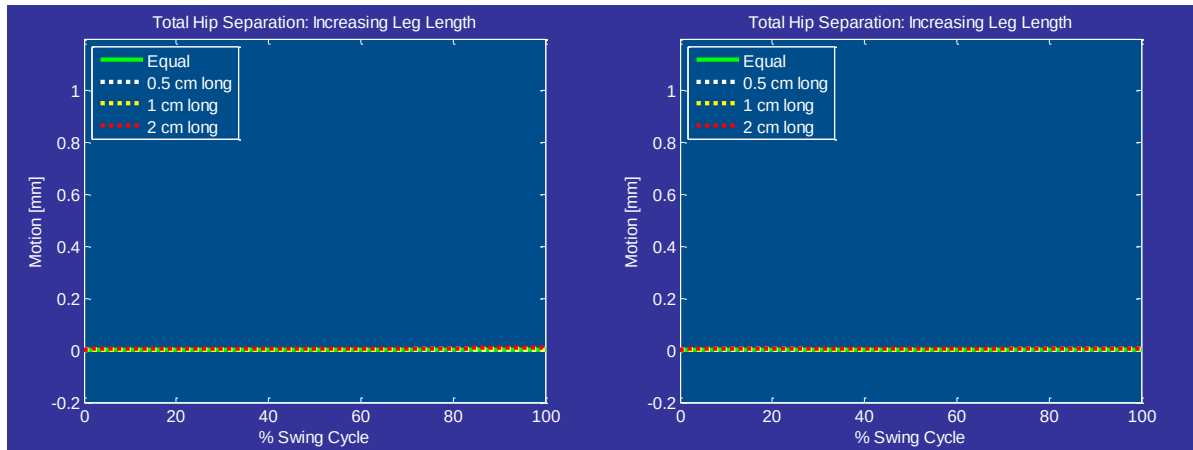
(d)

Figure 126: Leg length analysis, decreasing leg length, stance phase cup stress effects for Patients (a) 1, (b) 2, (c) 3, and (d) 4.



(a)

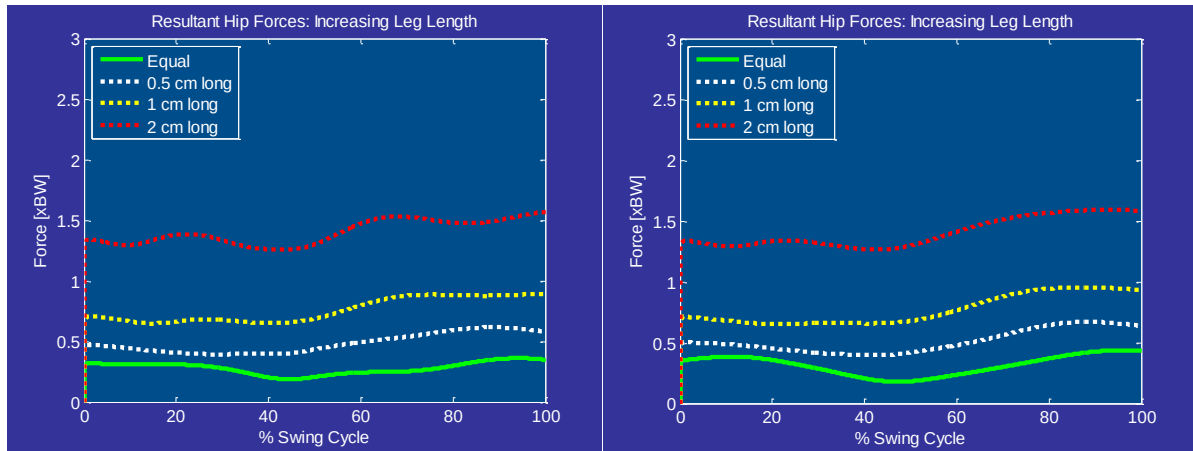
(b)



(c)

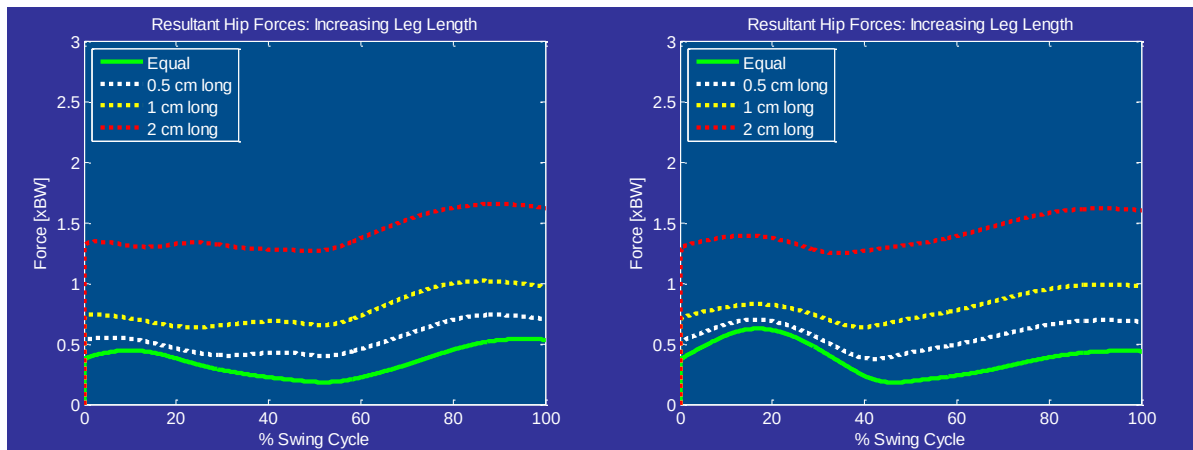
(d)

Figure 127: Leg length analysis, increasing leg length, swing phase hip separation effects for Patients (a) 1, (b) 2, (c) 3, and (d) 4.



(a)

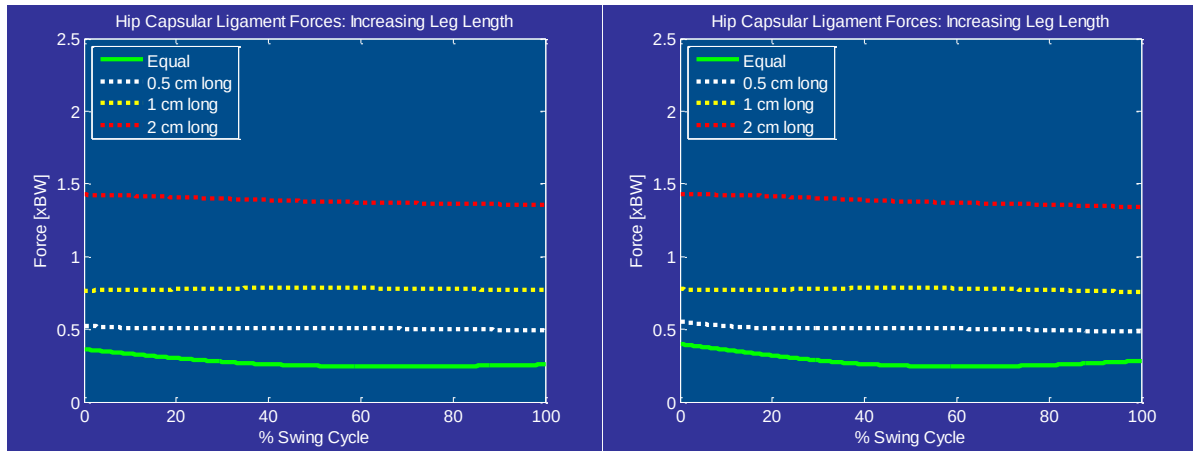
(b)



(c)

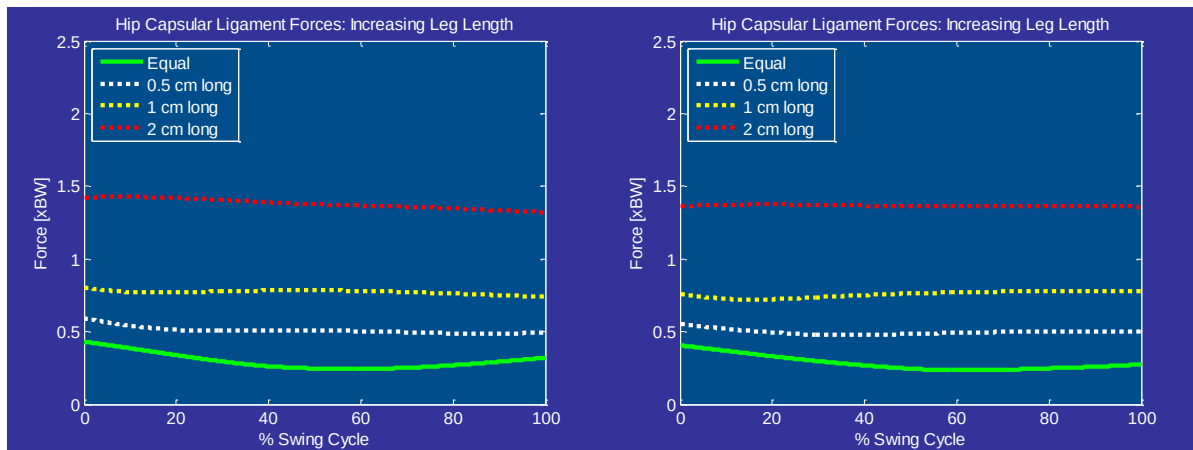
(d)

Figure 128: Leg length analysis, increasing leg length, swing phase hip force effects for Patients (a) 1, (b) 2, (c) 3, and (d) 4.



(a)

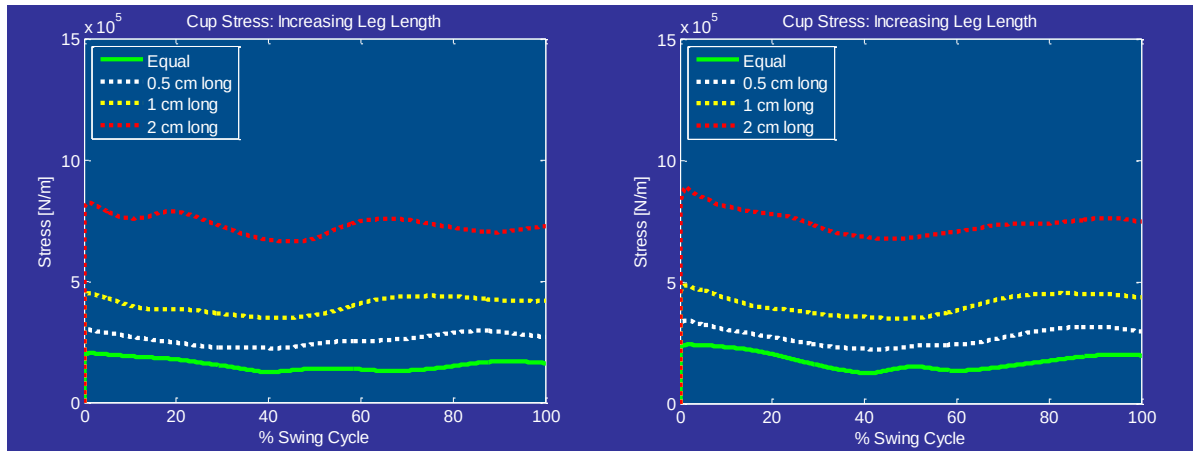
(b)



(c)

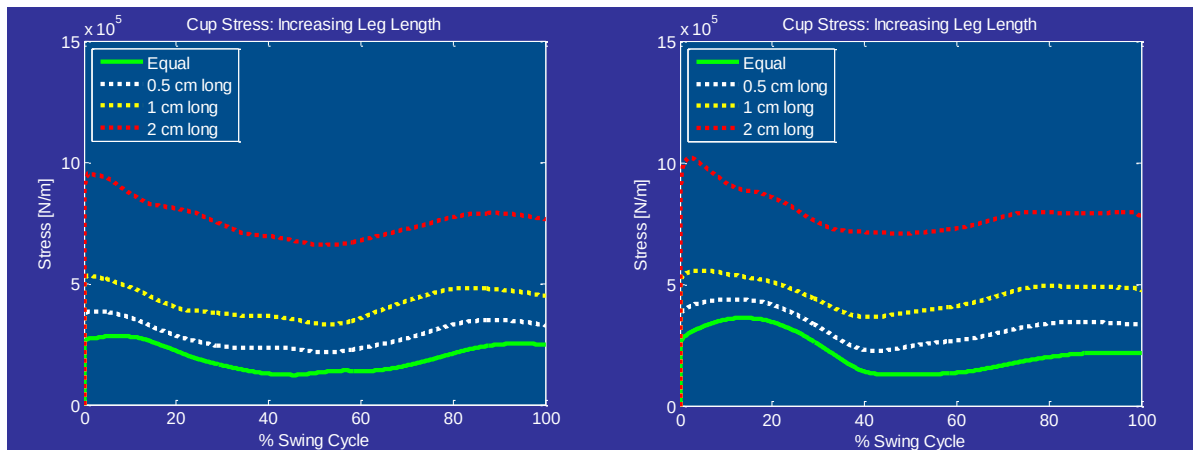
(d)

Figure 129: Leg length analysis, increasing leg length, swing phase capsular ligament force effects for Patients (a) 1, (b) 2, (c) 3, and (d) 4.



(a)

(b)



(c)

(d)

Figure 130: Leg length analysis, increasing leg length, swing phase cup stress effects for Patients (a) 1, (b) 2, (c) 3, and (d) 4.

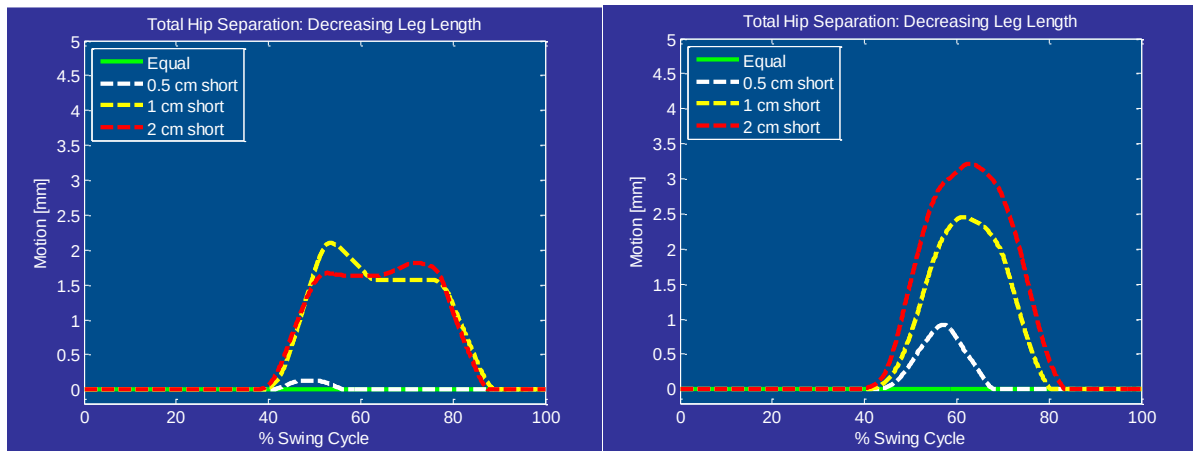
stride, so perhaps this is having a currently unknown effect. Figure 131, Figure 132, Figure 133, and Figure 134 show the effects of decreasing the length of the implanted leg by 5mm, 10mm, and 20mm for all four patients during swing phase.

7.2.4 Hip Offset Analysis

Much like leg length discrepancies, hip offset errors are also introduced by errors in femoral component placement. For this analysis, hip offset errors are introduced by theoretically selecting the wrong head or neck offset (too large, in all cases), and thus the femoral head center is shifted medially compared to the native femoral head center. A visual example can be seen in Figure 135. Note that the offset numbers in this analysis do not correspond to the actual stem model offset values/sizes. Rather, these offset values represent the displacement, due to surgical error, of the component femoral head compared to the native femoral head. The hip offset analysis was conducted on both swing and stance phase.

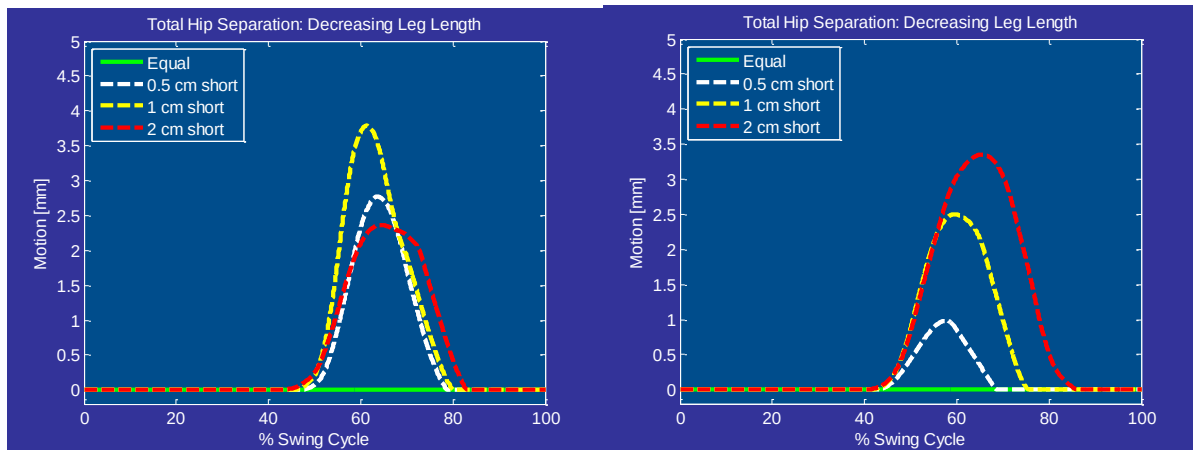
7.2.4.1 Stance Phase

The hip offset analyses behaved very similarly to the increased leg length analyses. During stance phase, an increase in the hip offset corresponded to a lengthened, and therefore tighter, hip capsule. Thus, minimal hip separation is observed, and this is accompanied by a minor increase in the forces and stresses present between the head and the cup. Figure 136, Figure 137, Figure 138, and Figure 139 show the effects of increasing the hip offset by 4mm, 6mm, and 8mm for all four patients during stance phase.



(a)

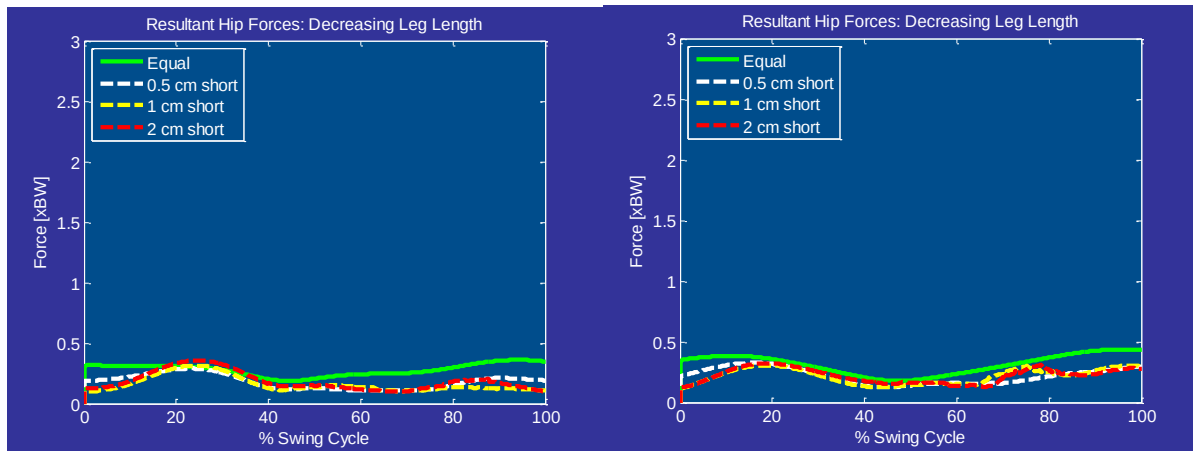
(b)



(c)

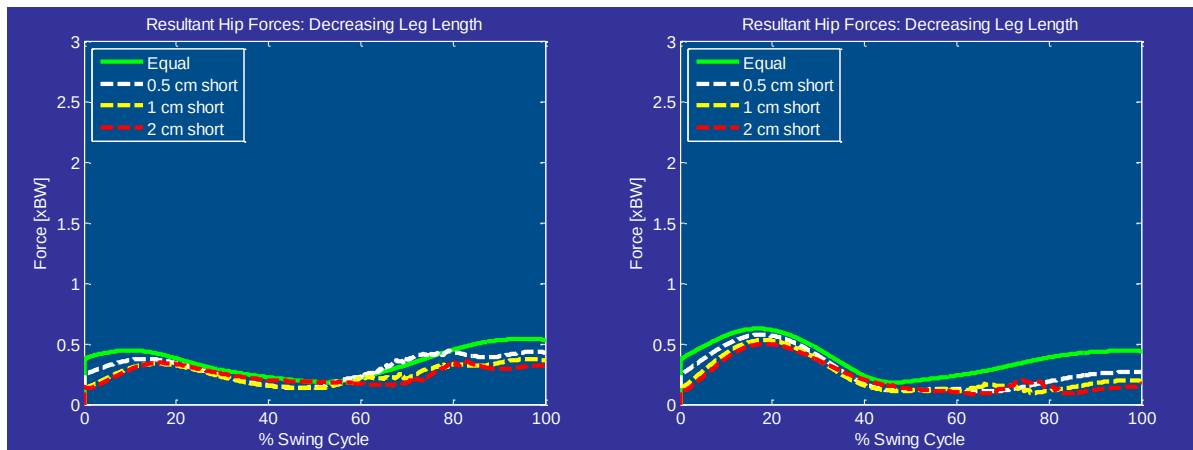
(d)

Figure 131: Leg length analysis, decreasing leg length, swing phase hip separation effects for Patients (a) 1, (b) 2, (c) 3, and (d) 4.



(a)

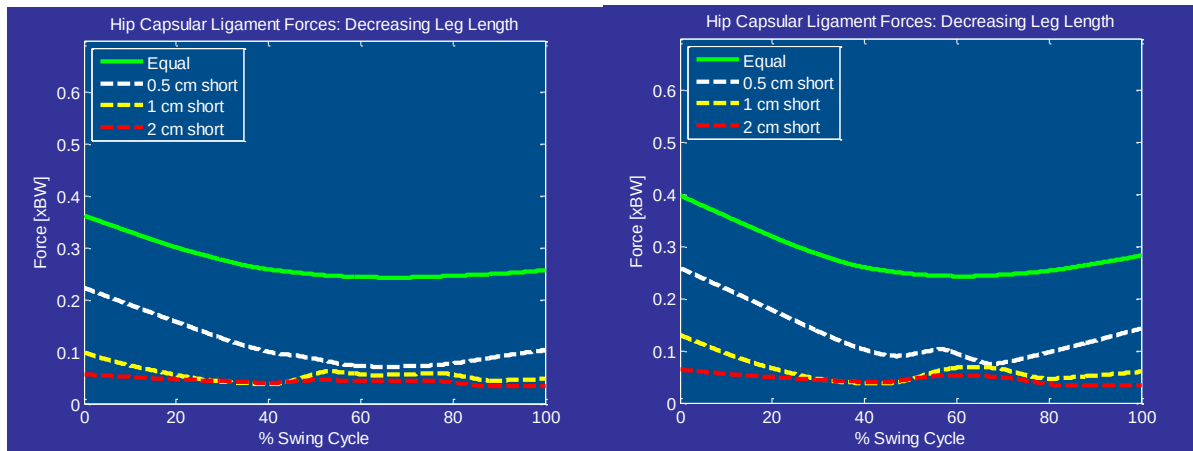
(b)



(c)

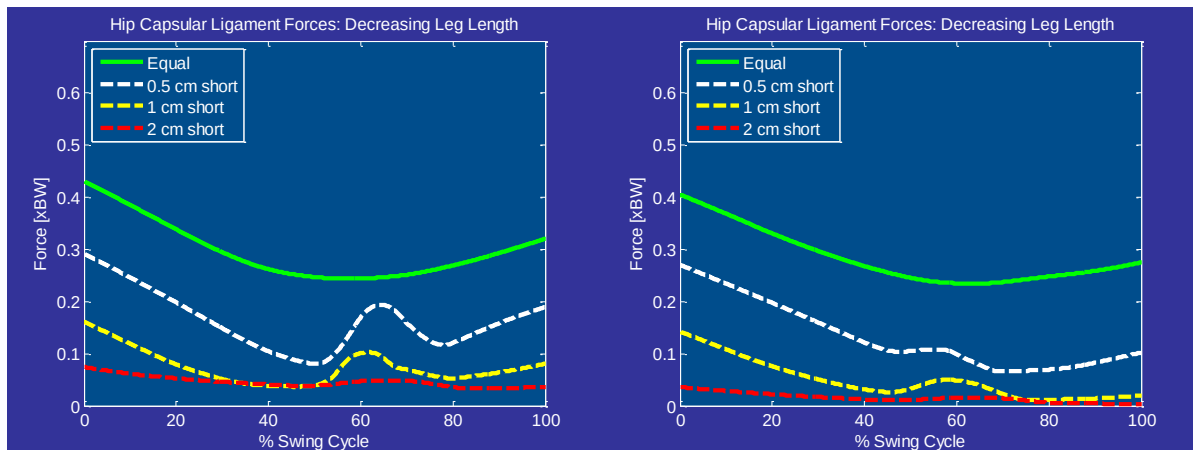
(d)

Figure 132: Leg length analysis, decreasing leg length, swing phase hip force effects for Patients (a) 1, (b) 2, (c) 3, and (d) 4.



(a)

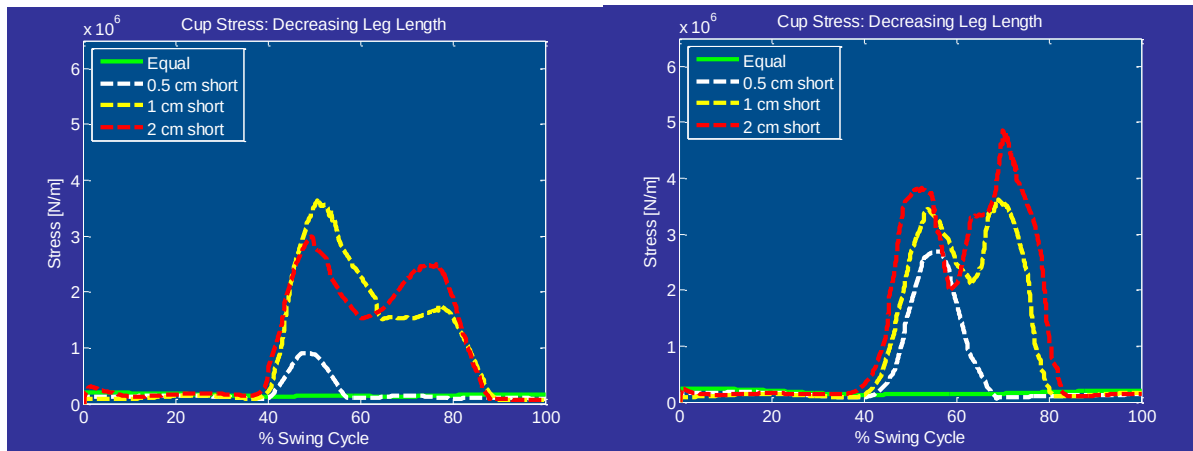
(b)



(c)

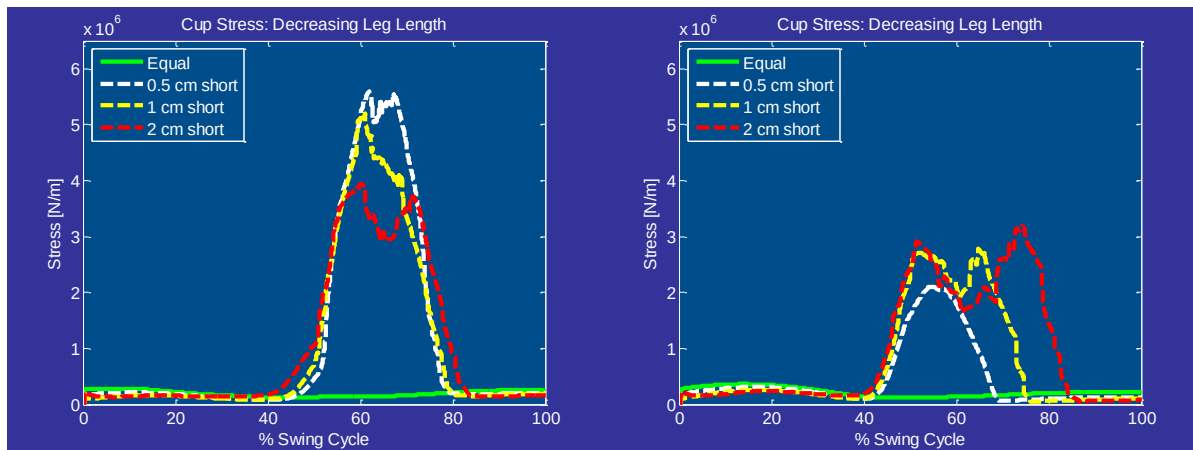
(d)

Figure 133: Leg length analysis, decreasing leg length, swing phase capsular ligament force effects for Patients (a) 1, (b) 2, (c) 3, and (d) 4.



(a)

(b)



(c)

(d)

Figure 134: Leg length analysis, decreasing leg length, swing phase cup stress effects for Patients (a) 1, (b) 2, (c) 3, and (d) 4.

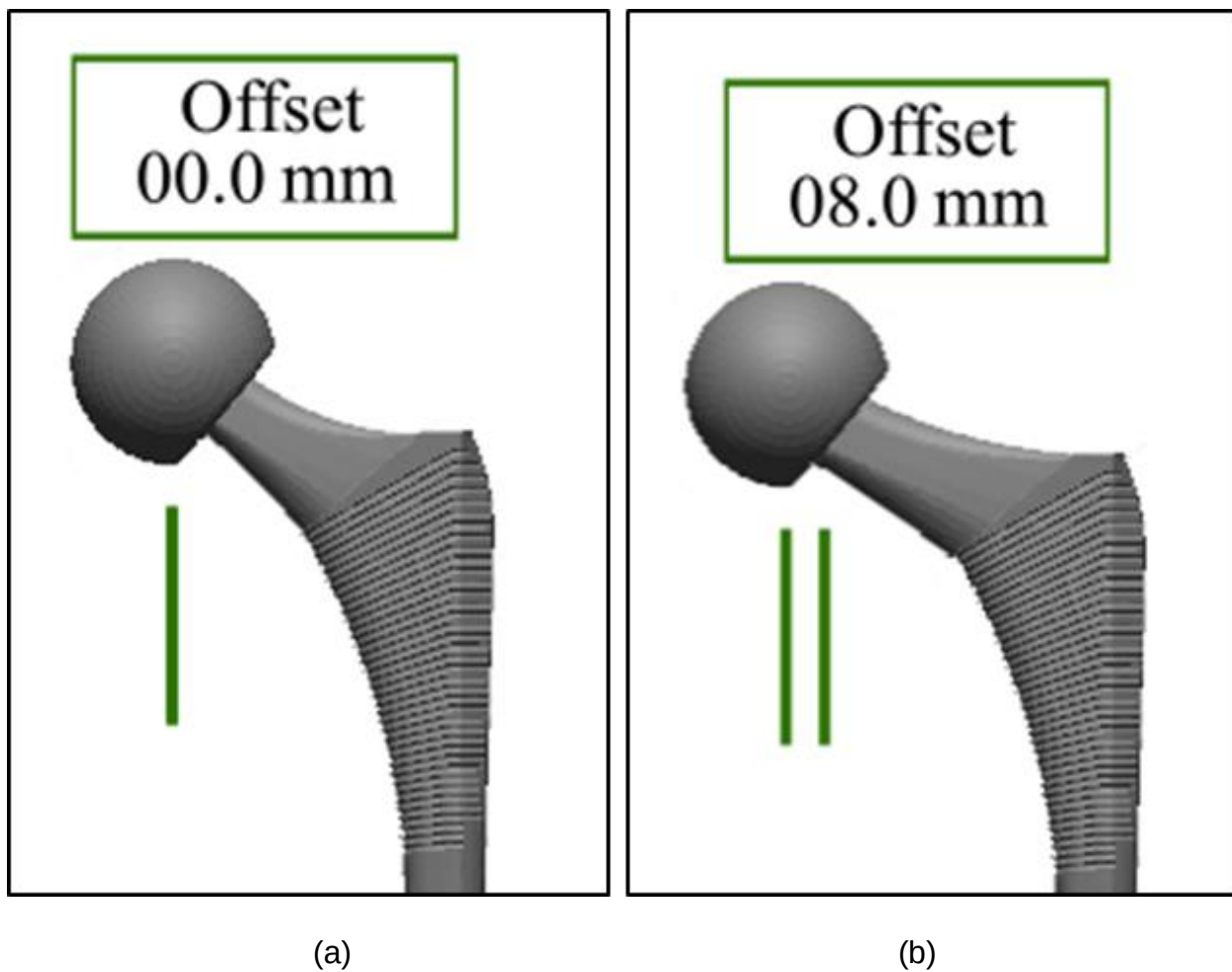
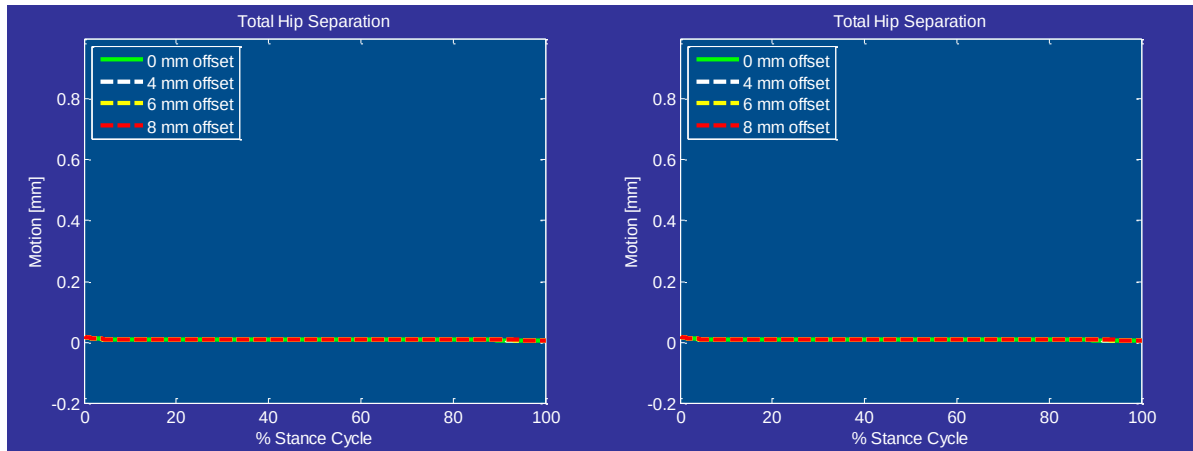
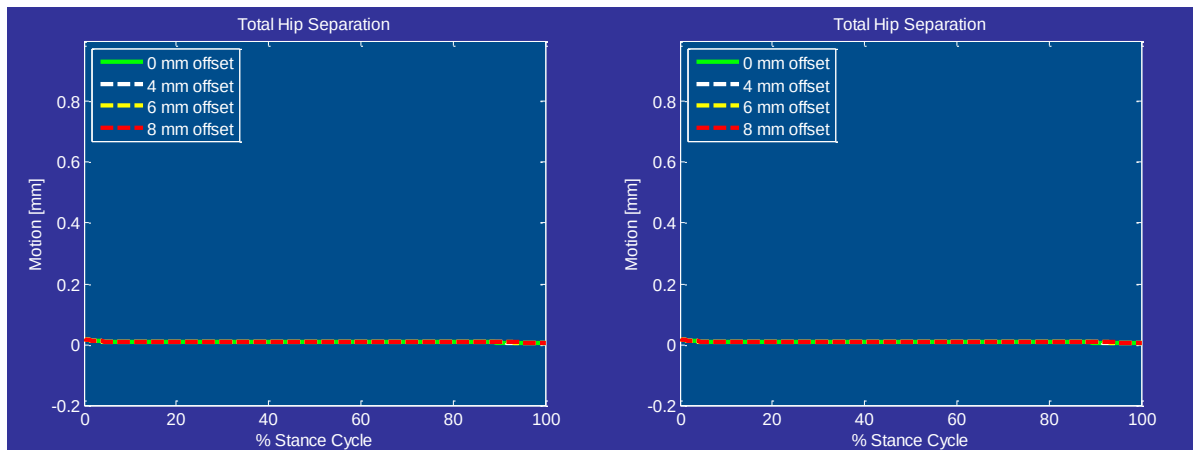


Figure 135: (a) aligned offset, and (b) medialized hip offset.



(a)

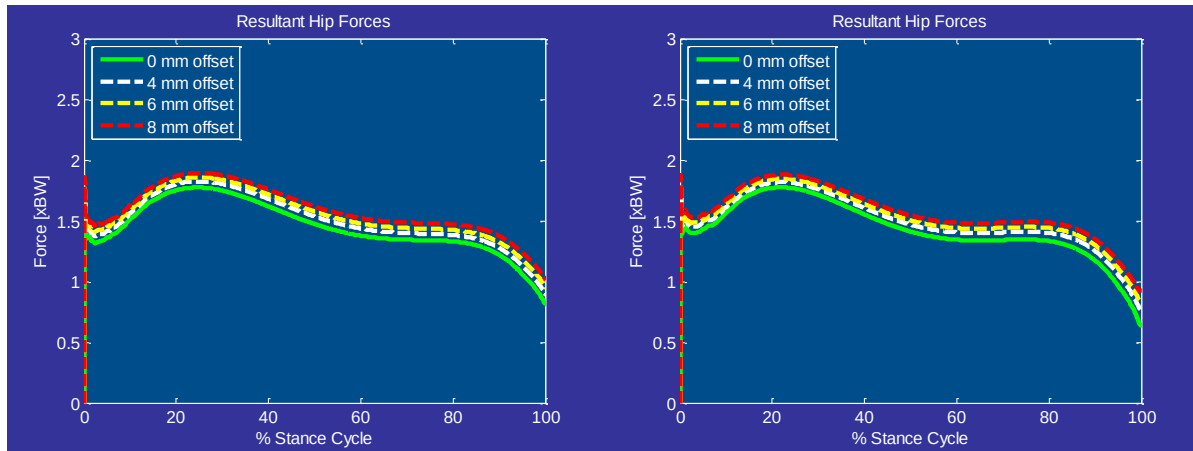
(b)



(c)

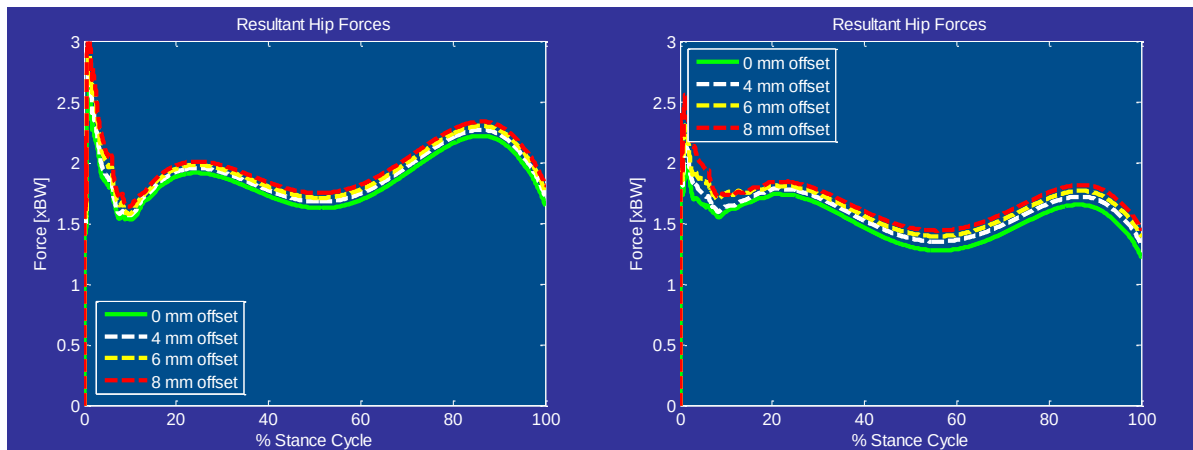
(d)

Figure 136: Hip offset analysis, stance phase hip separation effects for Patients (a) 1, (b) 2, (c) 3, and (d) 4.



(a)

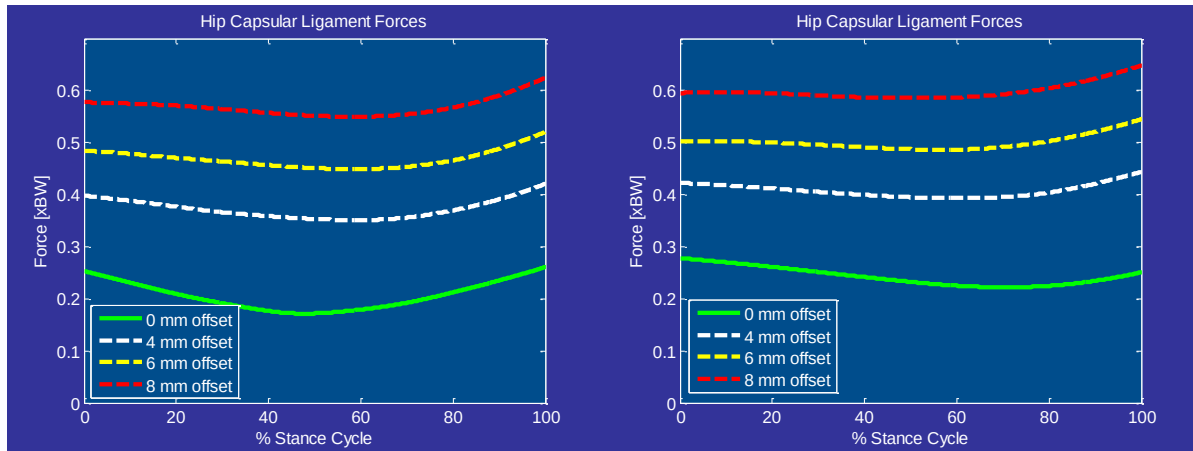
(b)



(c)

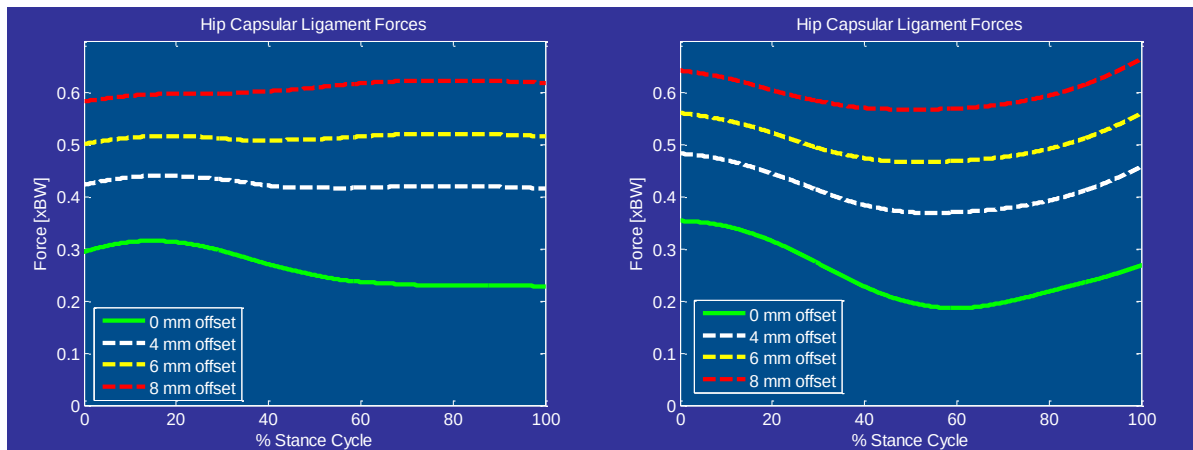
(d)

Figure 137: Hip offset analysis, stance phase hip force effects for Patients (a) 1, (b) 2, (c) 3, and (d) 4.



(a)

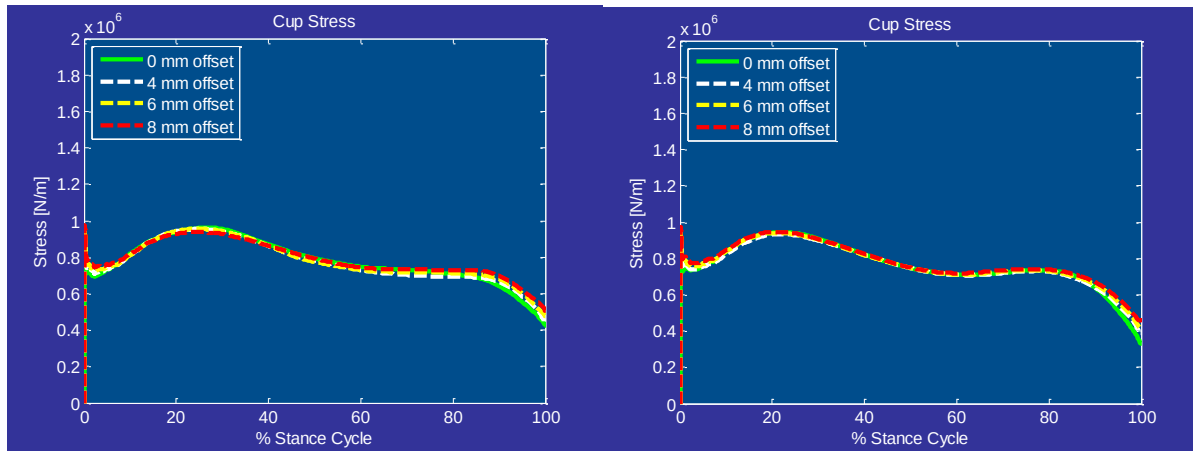
(b)



(c)

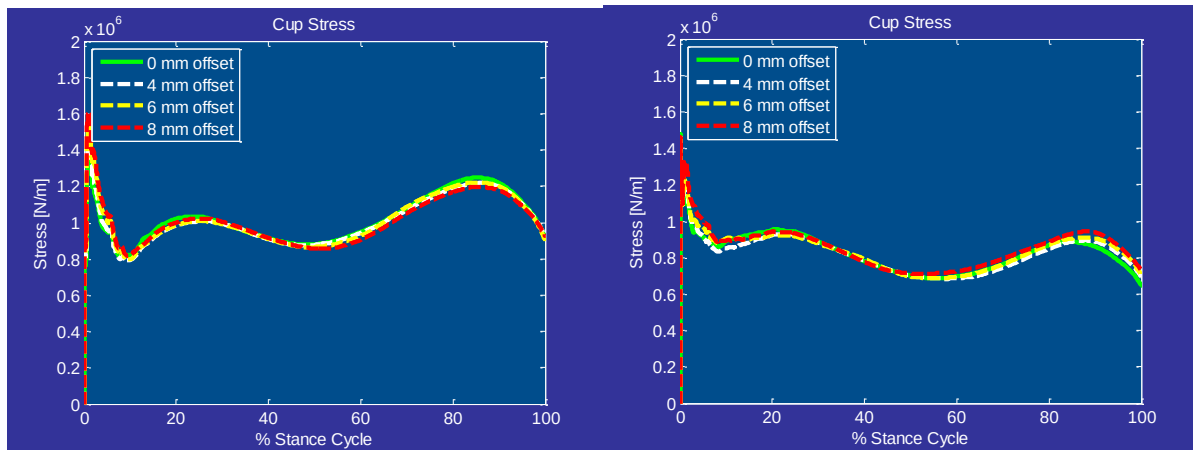
(d)

Figure 138: Hip offset analysis, stance phase capsular ligament force effects for Patients (a) 1, (b) 2, (c) 3, and (d) 4.



(a)

(b)



(c)

(d)

Figure 139: Hip offset analysis, stance phase cup stress effects for Patients (a) 1, (b) 2, (c) 3, and (d) 4.

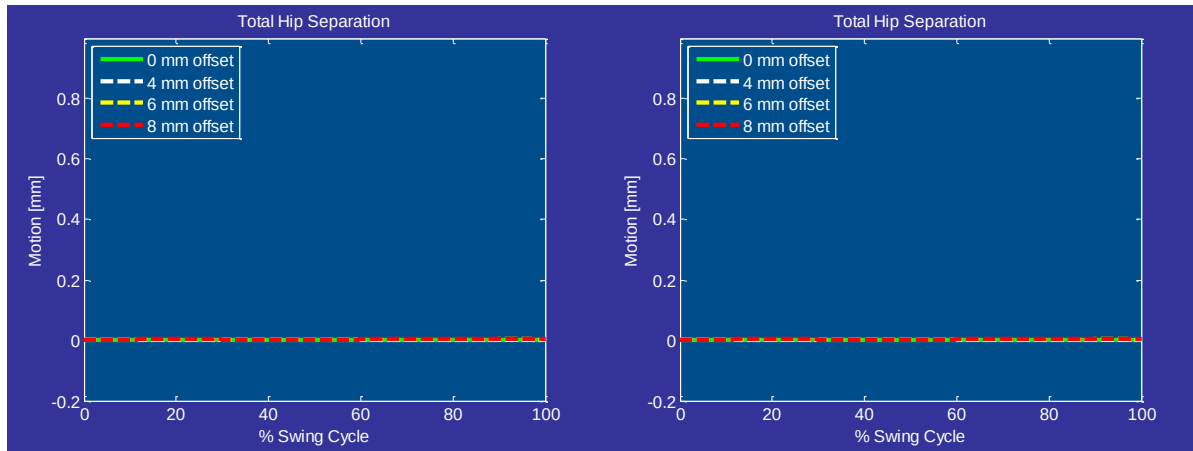
7.2.4.2 Swing Phase

Increasing the hip offset led to an increase in capsular ligament tension, and therefore hip separation was not observed during swing phase. However, these adjustments in offsets and ligament tensions again yielded an increase in cup stresses and resultant hip forces, which can be seen in Figure 140, Figure 141, Figure 142, and Figure 143, which show the effects of increasing the hip offset by 4mm, 6mm, and 8mm for all four patients during swing phase.

7.2.5 Femoral Component Version Analyses

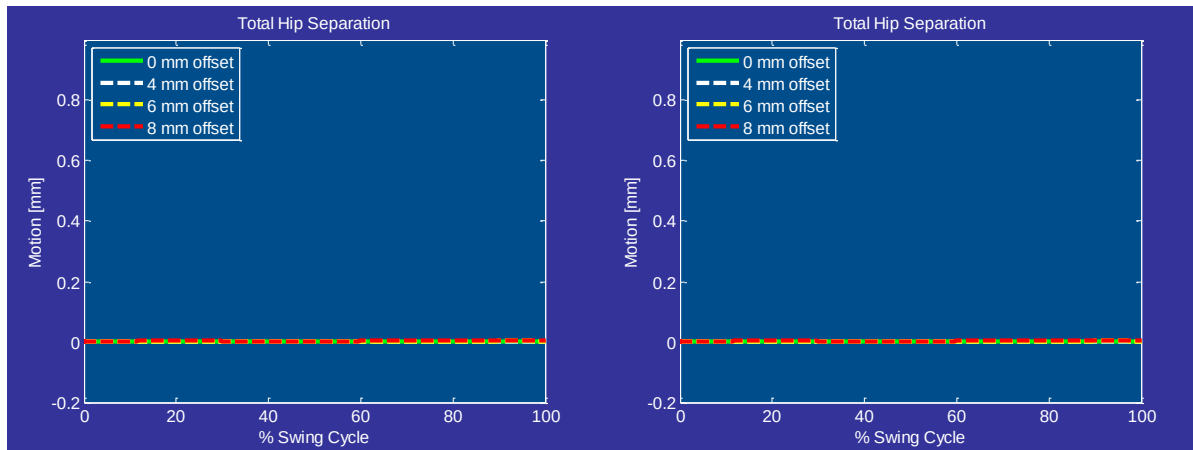
In the operating room, femoral component version errors are introduced if the surgeon mal-rotates the femoral component within the femoral canal. Certain implant designs incorporate modular methods of choosing femoral component version, while other implant types yield a fixed version angle depending on component placement. Femoral component version changes yield both rotational and translational errors. In other words, rotating the femoral component axially within the femoral canal will yield a translational shift of the component head center away from the anatomical center, as well as a corresponding internal or external rotation of the leg.

Figure 144 shows the orientation used to describe positive version change errors (anteversion errors) as well as the resulting leg rotation direction. Figure 145 shows the orientation used to describe negative version change errors (retroversion errors) as well as the resulting leg rotation direction. For reference, Figure 146 shows the orientation and foot rotation of a version error of 0°, indicating that the stem version accurately recreates the femur anatomical center. The following analysis was conducted on a single patient in



(a)

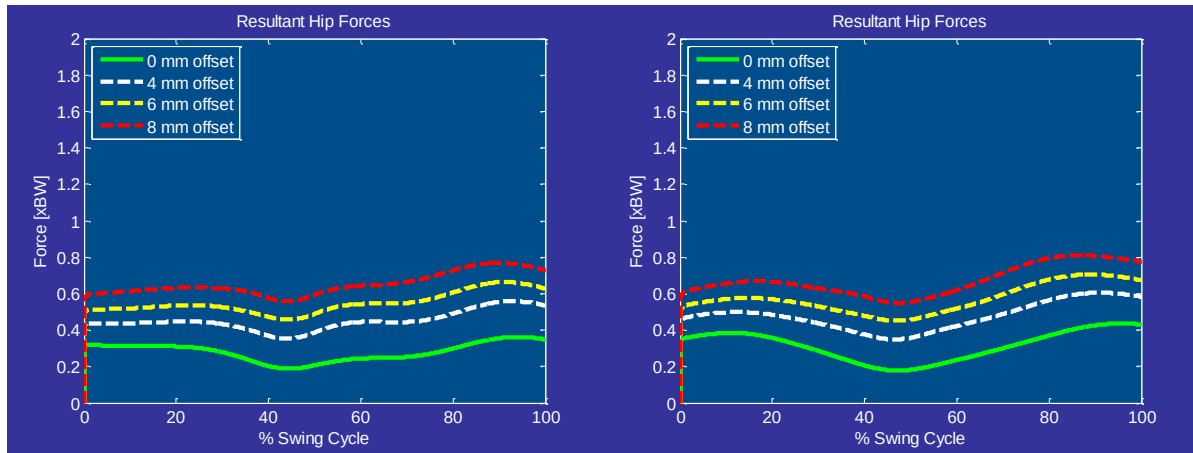
(b)



(c)

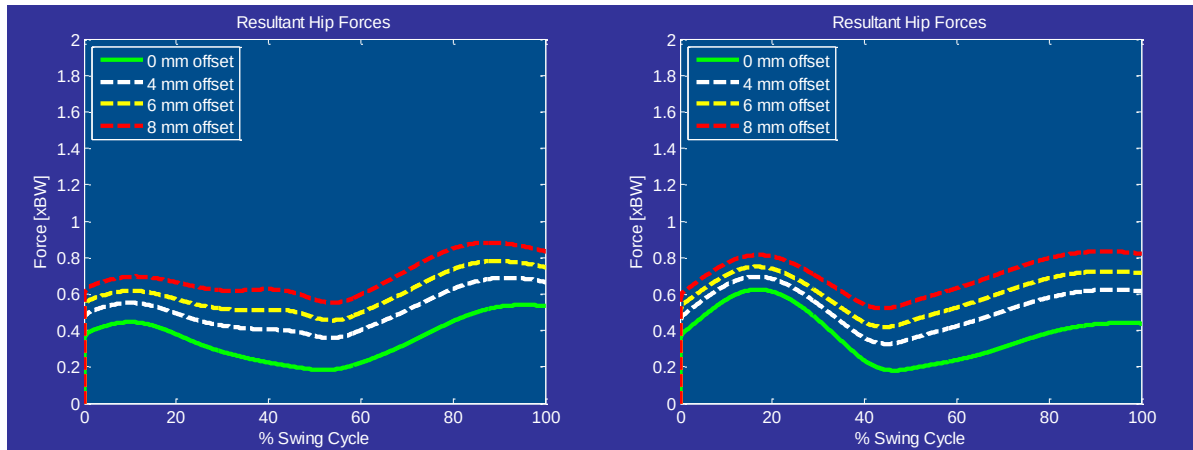
(d)

Figure 140: Hip offset analysis, swing phase hip separation effects for Patients (a) 1, (b) 2, (c) 3, and (d) 4.



(a)

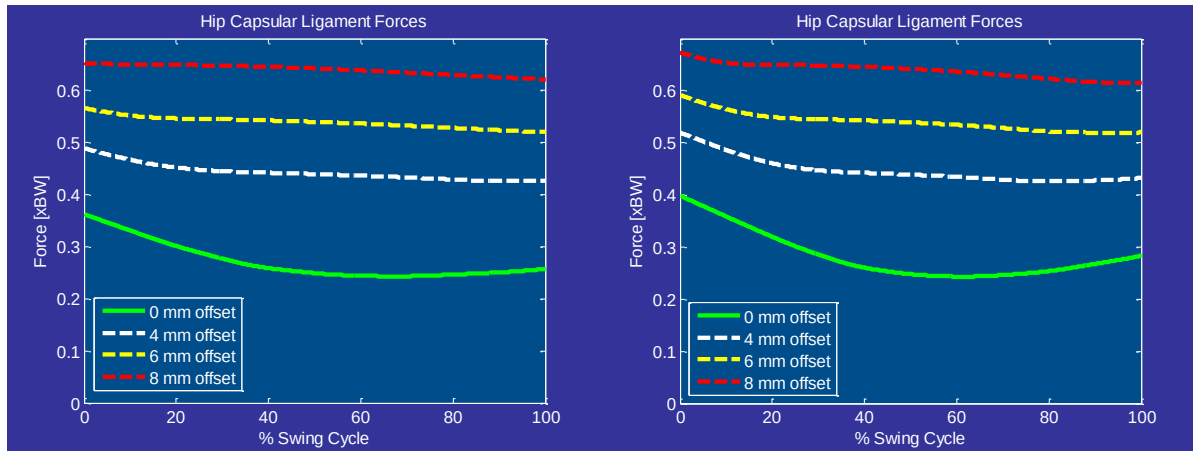
(b)



(c)

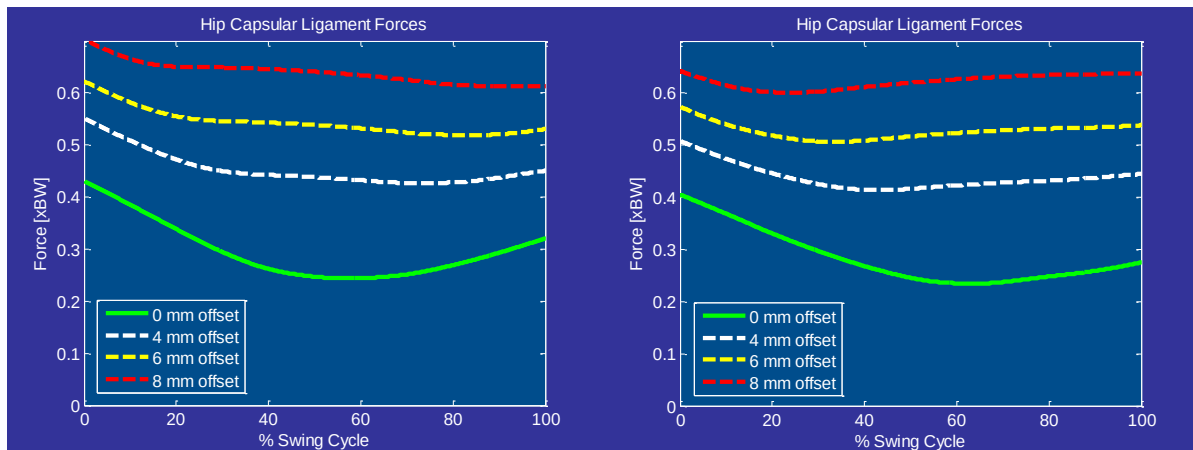
(d)

Figure 141: Hip offset analysis, swing phase hip force effects for Patients (a) 1, (b) 2, (c) 3, and (d) 4.



(a)

(b)

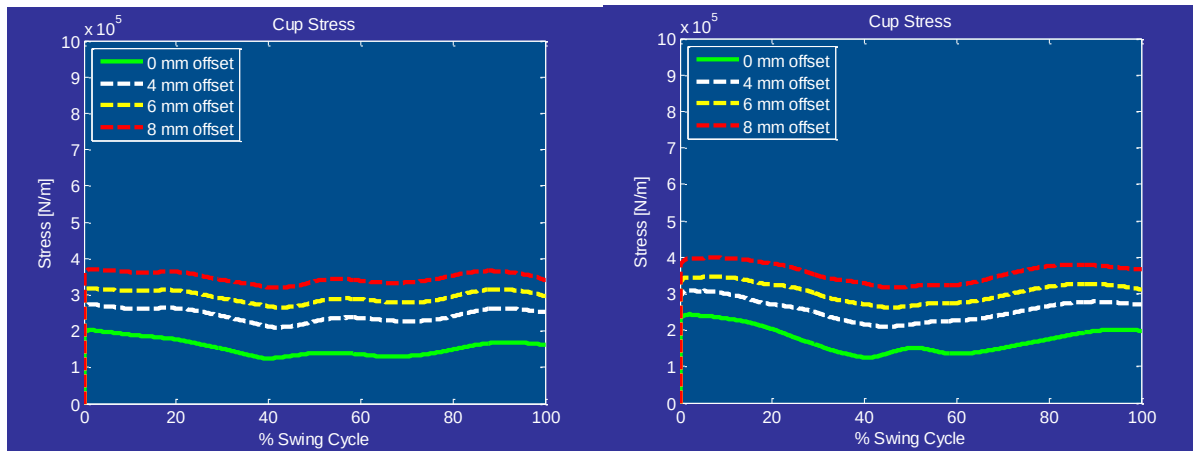


(c)

(d)

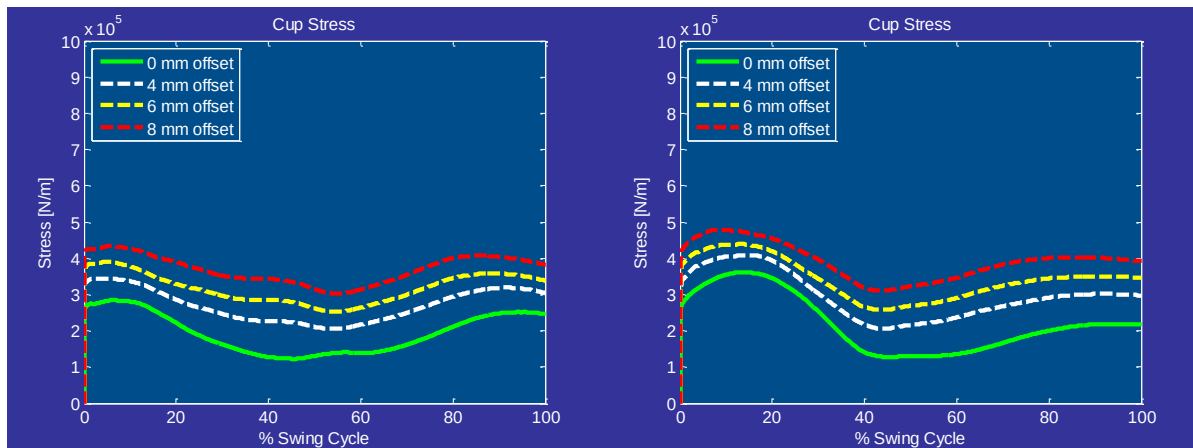
Figure 142: Hip offset analysis, swing phase capsular ligament force effects for Patients

(a) 1, (b) 2, (c) 3, and (d) 4.



(a)

(b)



(c)

(d)

Figure 143: Hip offset analysis, swing phase cup stress effects for Patients (a) 1, (b) 2, (c) 3, and (d) 4.

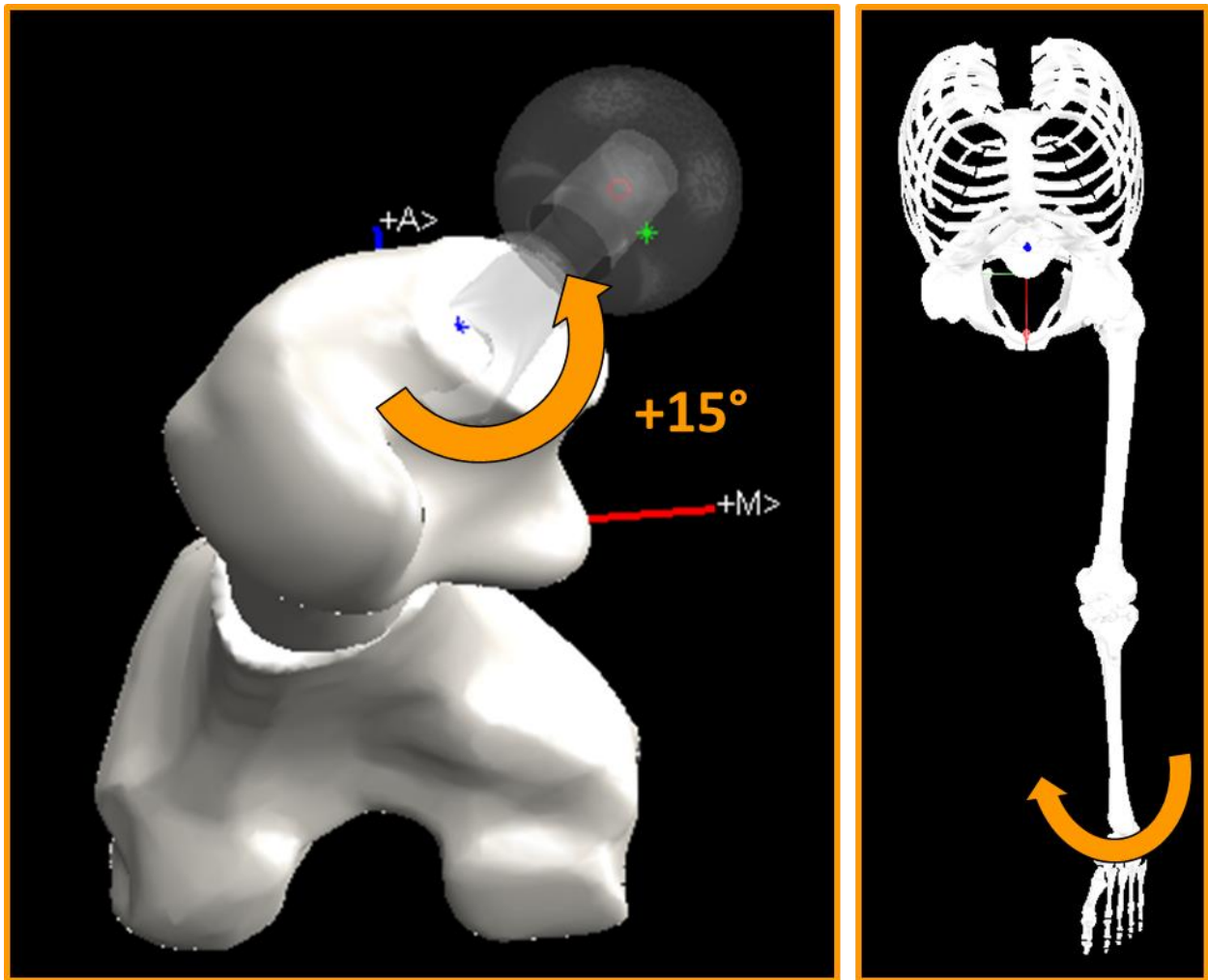


Figure 144: Positive version direction (anteversion) and accompanying internal rotation of the foot.

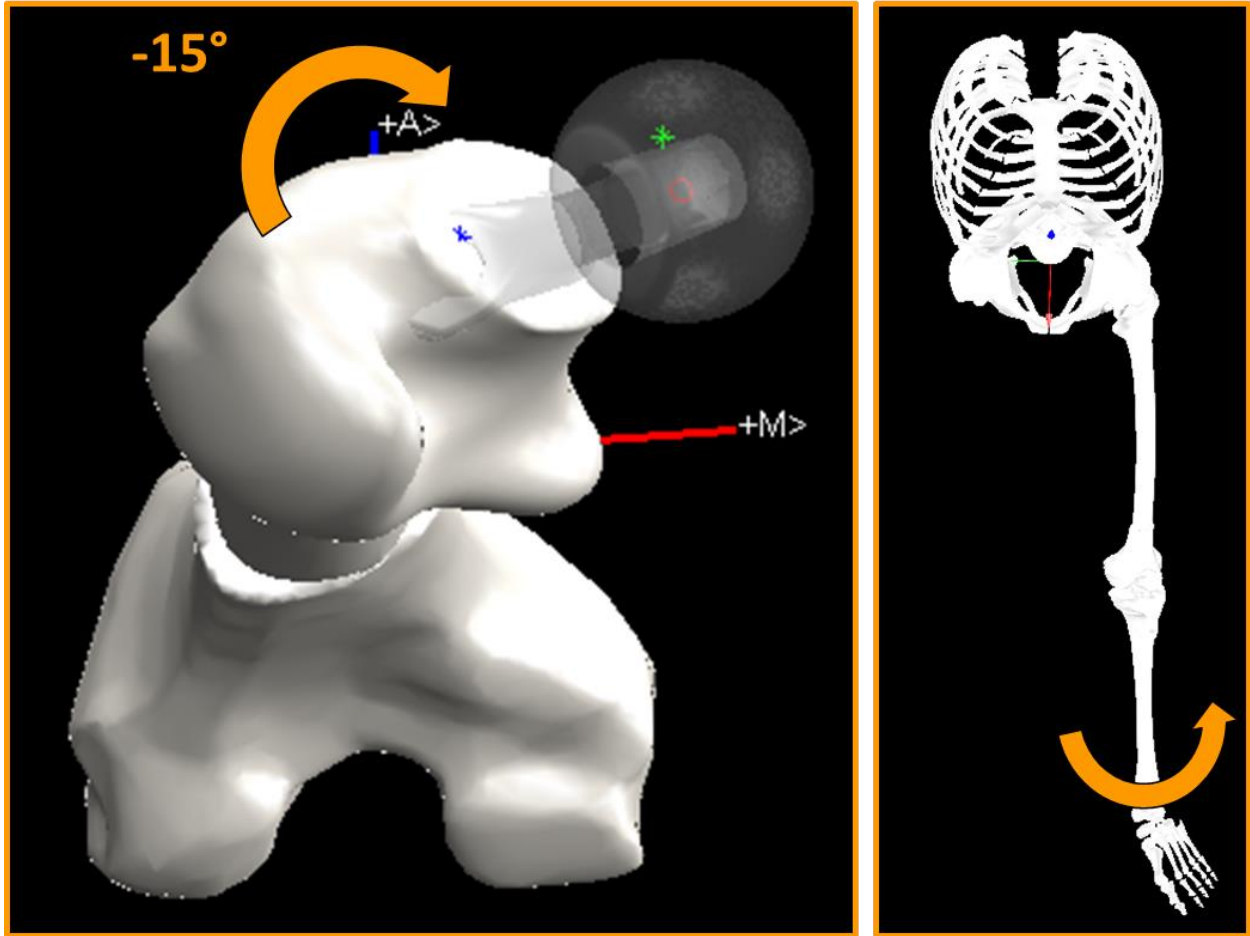


Figure 145: Negative version direction (retroversion) and accompanying external rotation of the foot.

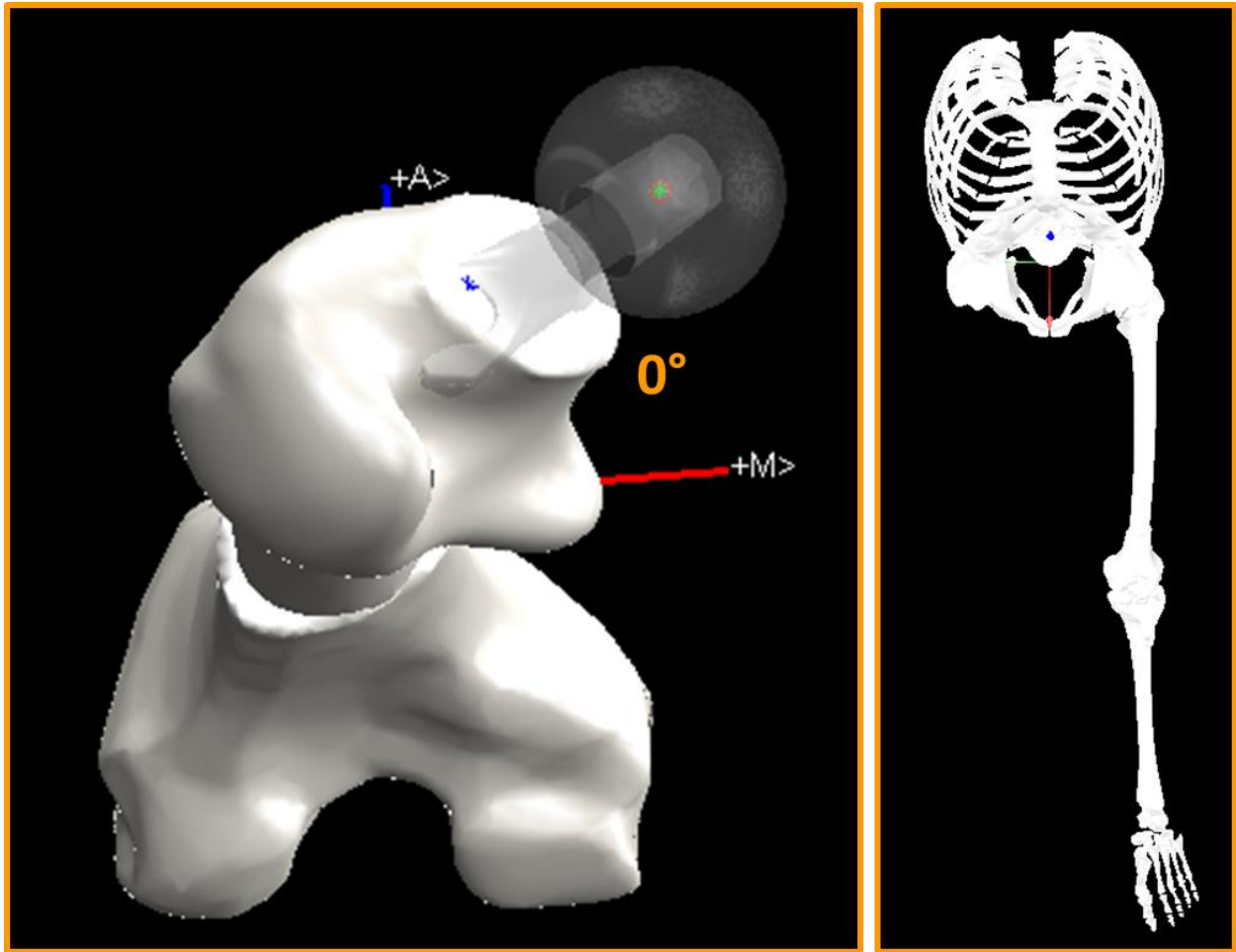
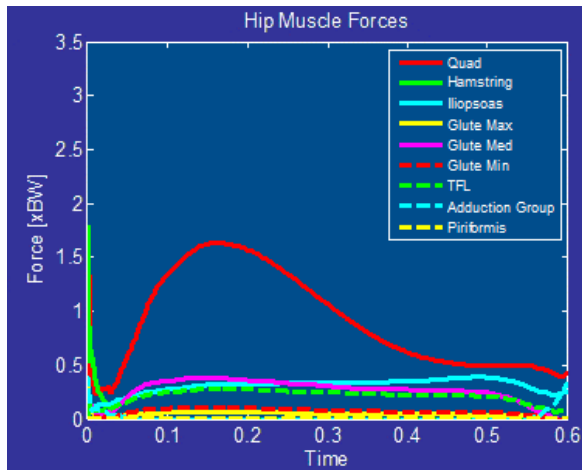


Figure 146: 0 degrees of version error of the femoral component.

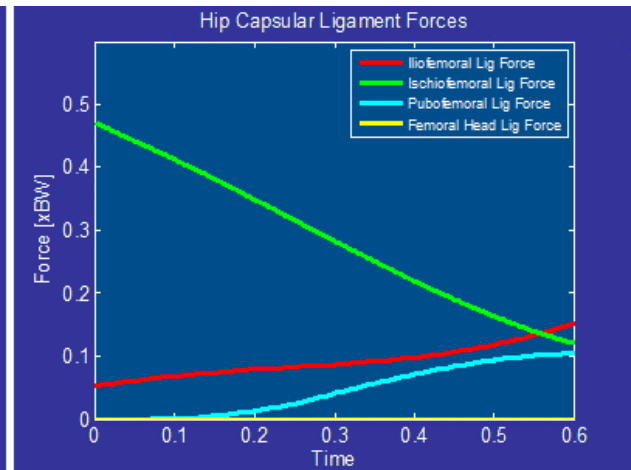
stance phase with a version error range of -15° to $+15^{\circ}$ in increments of 5° . Figure 147 shows the muscle forces, ligament forces, contact forces, and hip separation values for the control scenario of 0° of version alignment. Figure 148 also shows, for reference, the contact map location of the femoral head within the acetabular cup for a properly aligned femoral component.

In general, negative version errors (retroversion, yielding external rotation of the foot) did not appear to yield substantial hip separation (Figure 149). However, they did yield a posterior shift in the contact map of the femoral head, shown in Figure 150, indicating abnormal mechanics and potential joint instability. Accompanied by a misaligned acetabular cup, these abnormal mechanics could be exacerbated. Retroversion also yielded a decrease in the quadriceps muscle forces, an increase in the gluteus medius forces during loading phase, and a decrease in the gluteus medius forces during unloading phase, shown in Figure 151. Furthermore, negative version errors caused an increase in the tension of the anterior hip capsule and a decrease in the tension of the posterior hip capsule, shown in Figure 152. Most significantly, negative version changes appeared to yield an increase in the resultant joint reaction forces during loading phase and a decrease in the resultant joint reaction forces during unloading phase, shown in Figure 153.

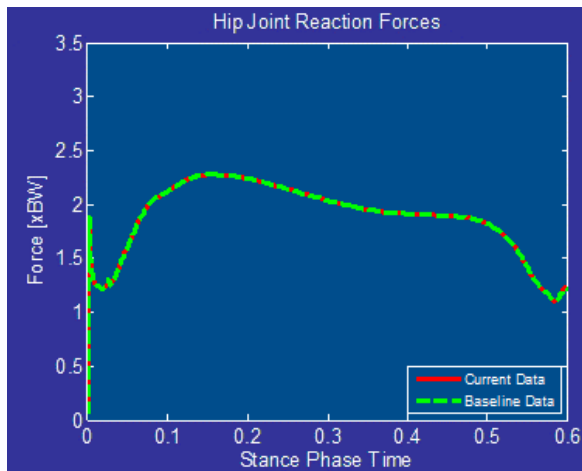
Positive version errors (anteversion, yielding internal rotation of the foot) also appeared to yield minimal hip separation ($<0.5\text{mm}$, Figure 154). Conversely to retroversion, these changes yielded an anterior shift in the contact map, shown in Figure 155. Once again, this indicates abnormal mechanics and potential joint instability, and these abnormal mechanics could be exacerbated by misaligned cups. Anteversion also



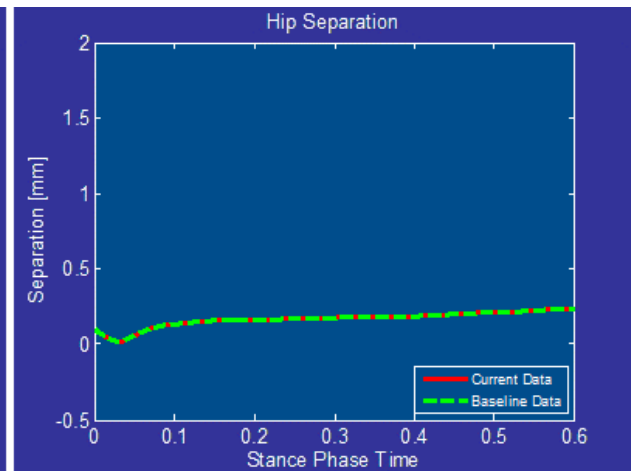
(a)



(b)



(c)



(d)

Figure 147: (a) hip muscle forces, (b) hip capsular ligament forces, (c) hip joint reaction forces, and (d) hip separation for 0 degrees of version error of the femoral component.

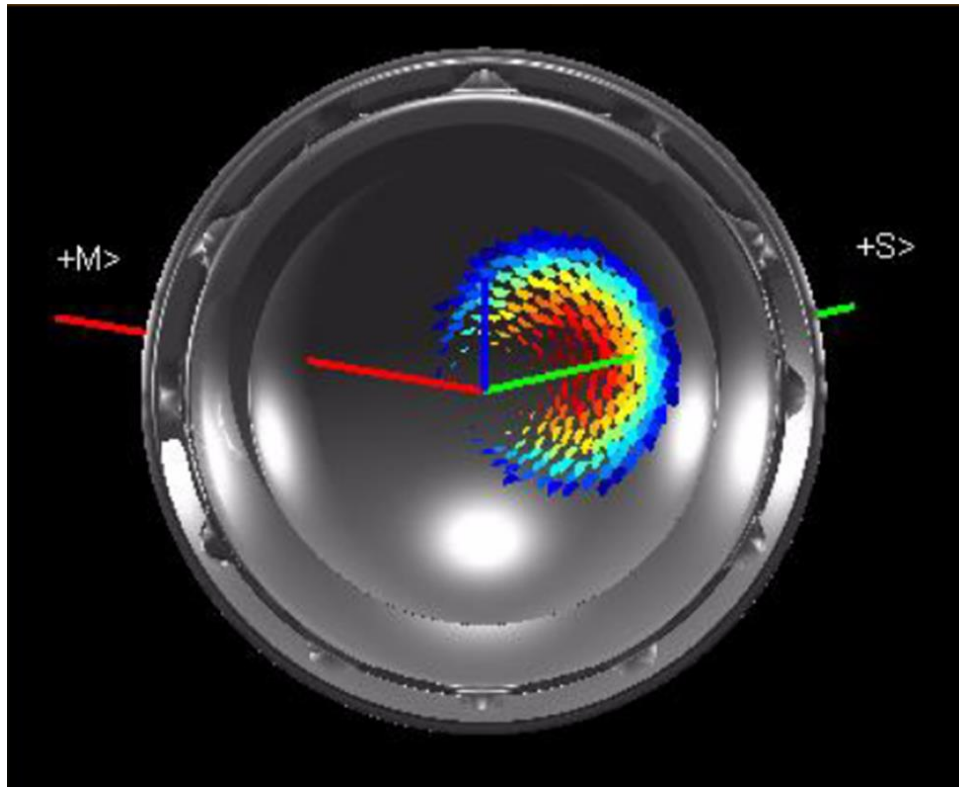
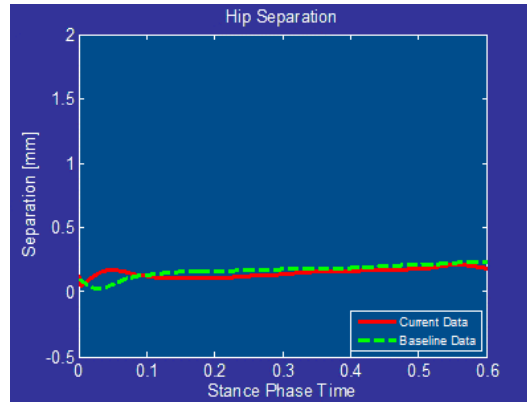
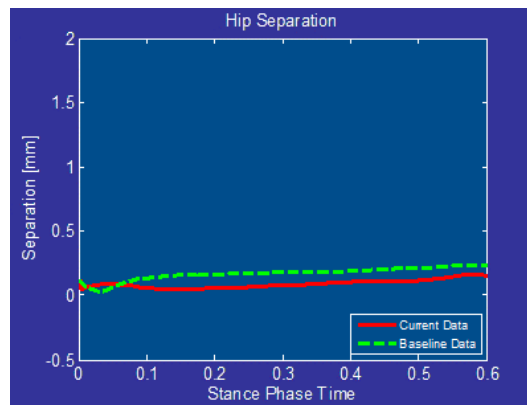


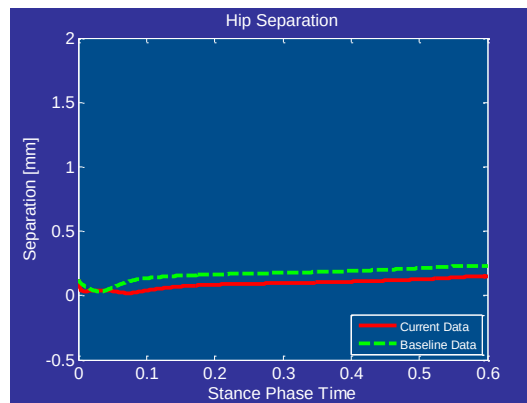
Figure 148: Contact map location for no version error scenario, showing a superior contact patch.



(a)



(b)



(c)

Figure 149: Hip separation profiles for (a) -15 degrees, (b) -10 degrees, and (c) -5 degrees of version error of the femoral component.

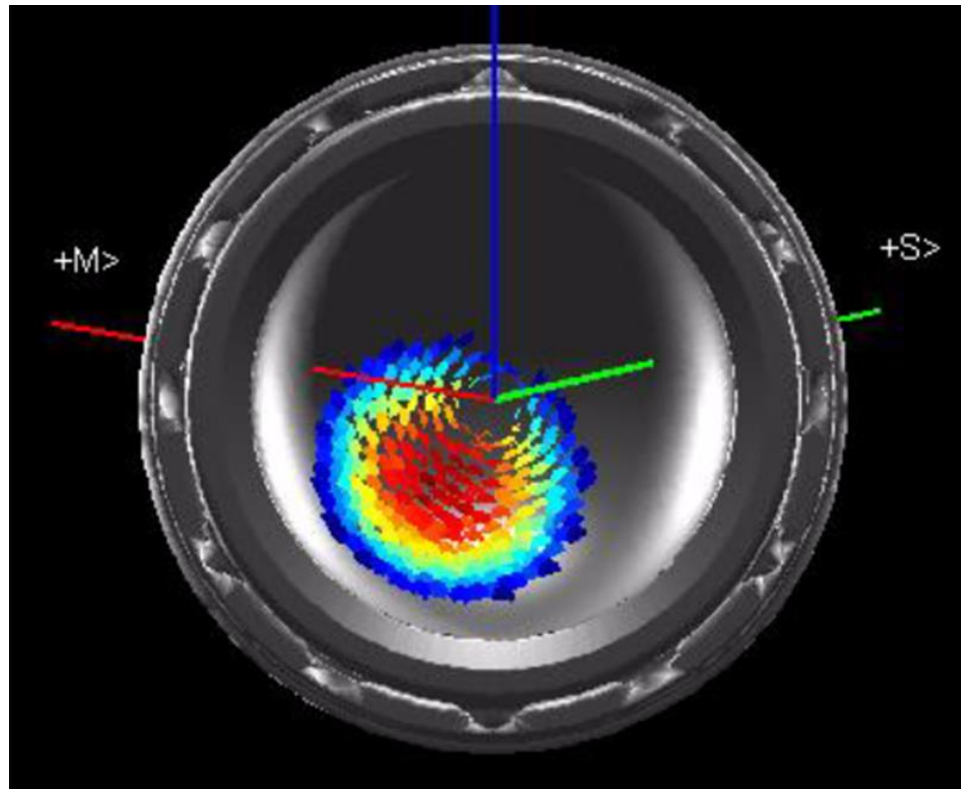
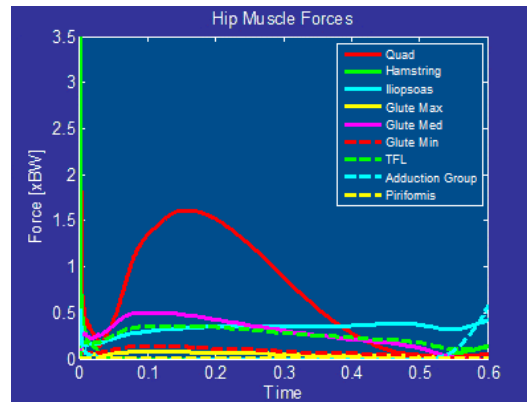
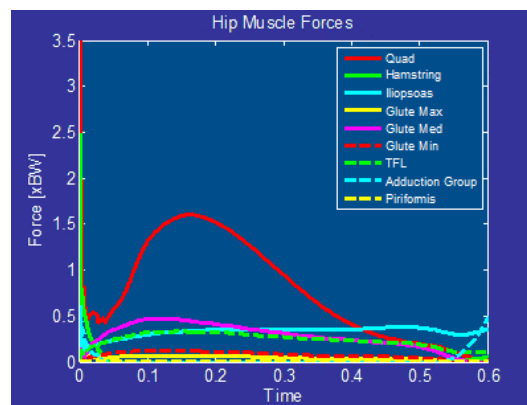


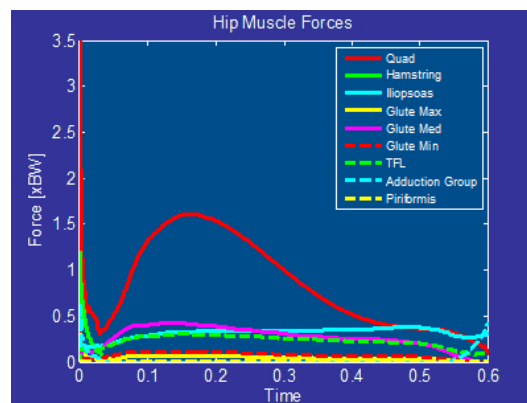
Figure 150: Contact map location for the -15 degree version error scenario, showing a posterior contact patch.



(a)

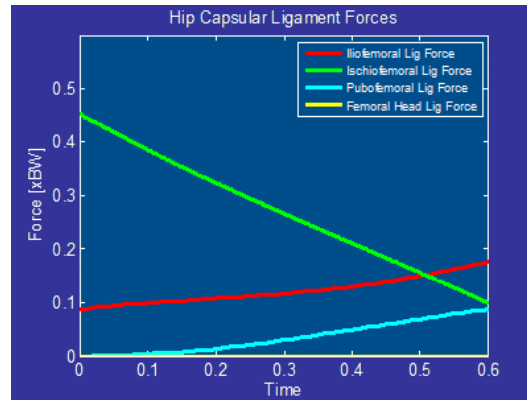


(b)

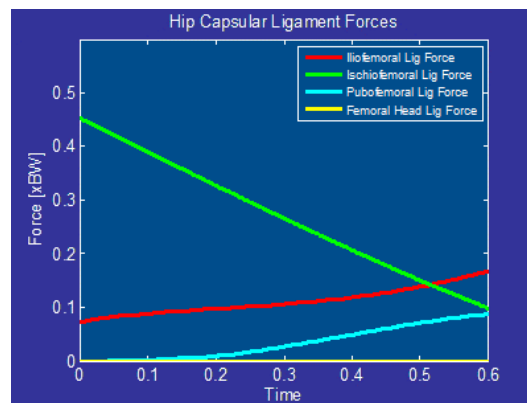


(c)

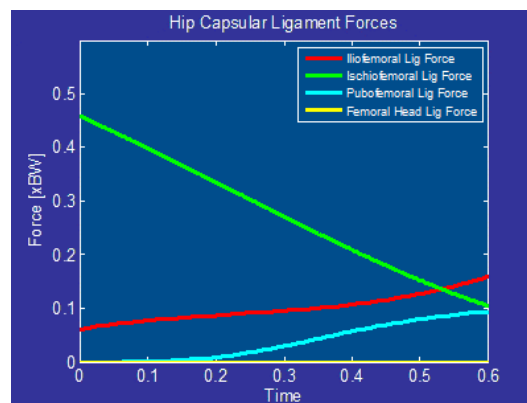
Figure 151: Hip muscle force profiles for (a) -15 degrees, (b) -10 degrees, and (c) -5 degrees of version error of the femoral component.



(a)

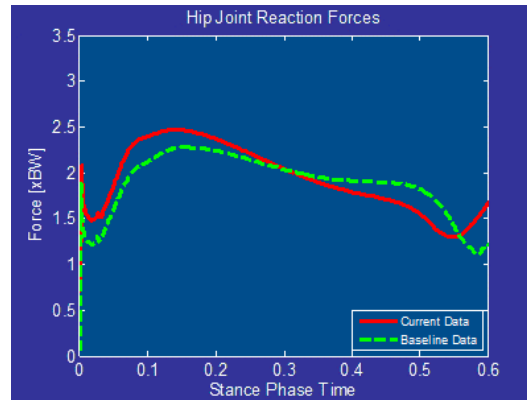


(b)

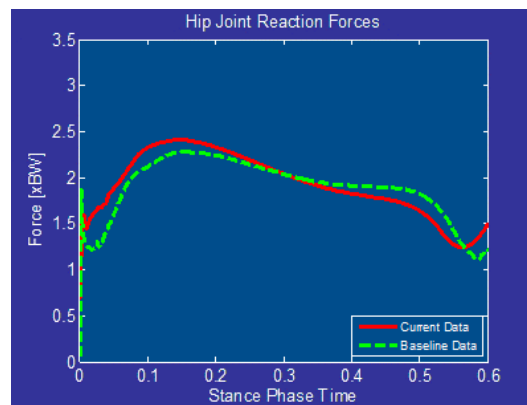


(c)

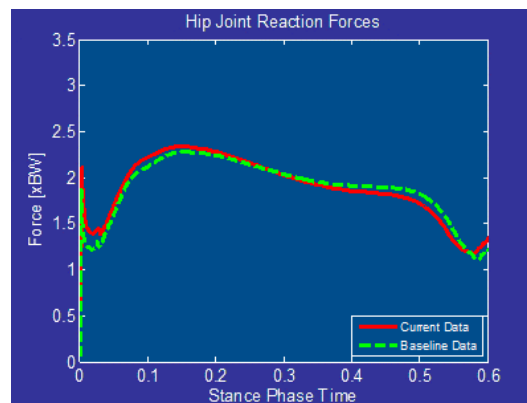
Figure 152: Hip capsular ligament force profiles for (a) -15 degrees, (b) -10 degrees, and (c) -5 degrees of version error of the femoral component.



(a)

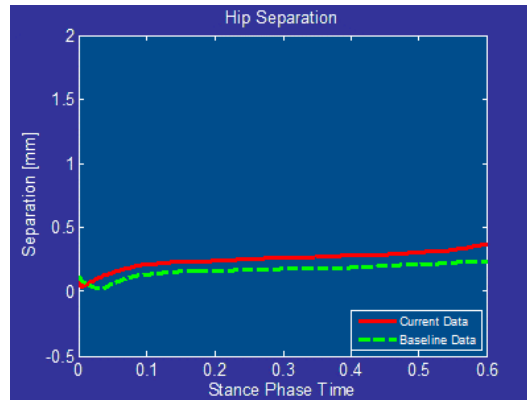


(b)

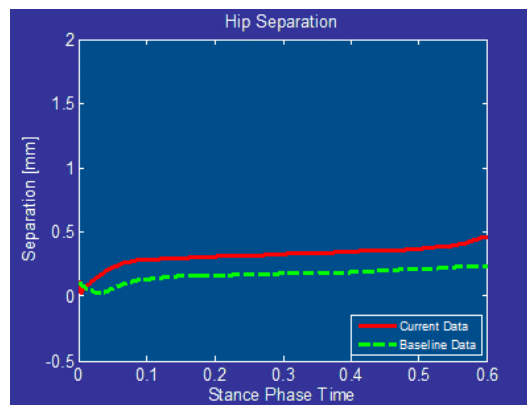


(c)

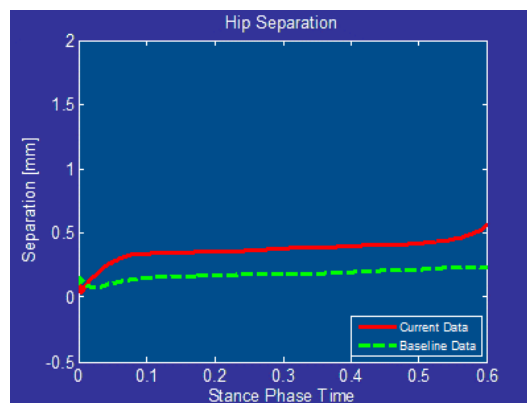
Figure 153: Hip joint reaction force profiles for (a) -15 degrees, (b) -10 degrees, and (c) -5 degrees of version error of the femoral component.



(a)



(b)



(c)

Figure 154: Hip separation profiles for (a) +5 degrees, (b) +10 degrees, and (c) +15 degrees of version error of the femoral component.

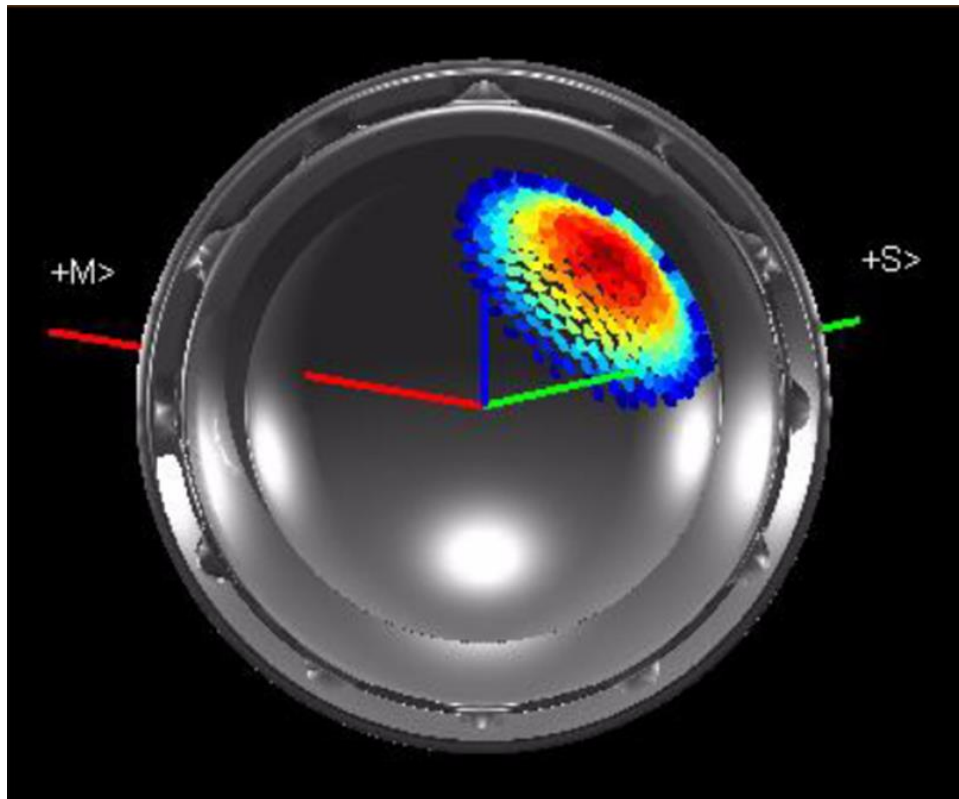
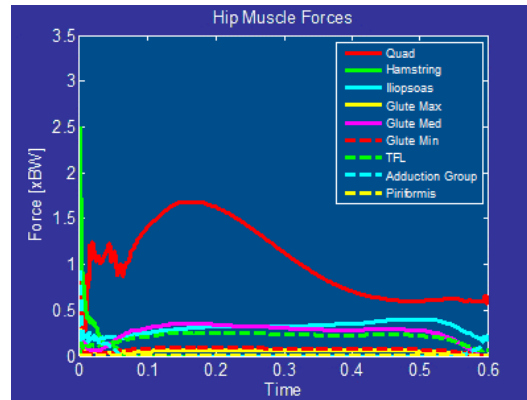


Figure 155: Contact map location for the +15 degree version error scenario, showing an anterior contact patch.

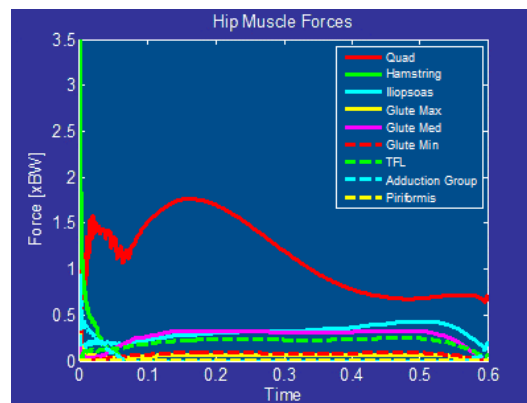
yielded an increase in the quadriceps muscle forces, a decrease in the gluteus medius forces during loading phase, and an increase in the gluteus medius forces during unloading phase (Figure 156). Furthermore, anteversion errors caused a decrease in the tension of the anterior hip capsule and an increase in the tension of the posterior hip capsule (Figure 157). Most significantly, positive version changes appeared to yield a decrease in the resultant joint reaction force during loading phase and an increase in the resultant joint reaction forces during unloading phase (Figure 158).

7.3 Component Design Analyses

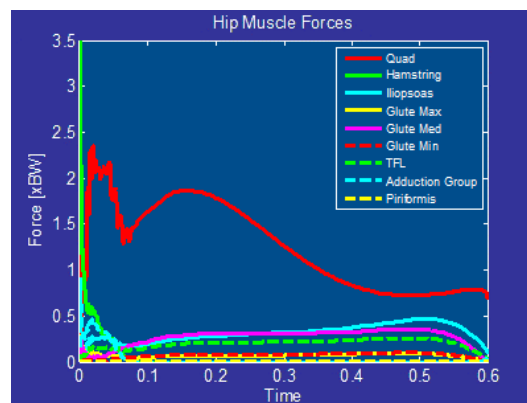
The following analyses are a comparison between three different liner designs: the Neutral liner, the Lipped liner, and the Augmented liner. Five separate analyses were done to compare these three liners: 1) an analysis of the range-of-motion (ROM) of the liners, 2) an analysis of the effects of internal/external (I/E) rotation of the liners, 3) an analysis of lateralization of the liners, 4) a brief analysis on the effects of cup inclination, and 5) an analysis A/P shifting of the acetabular cup. The purpose of these analyses was to determine whether the addition of a lateral lip to the acetabular liner would improve the stability of the hip, thereby decreasing hip separation. The first of the following subsections presents a geometric comparison of the three liners, and then the remaining five subsections each constitute one of the five aforementioned analyses. At the very end of this section, an analysis of a dual mobility system is included.



(a)

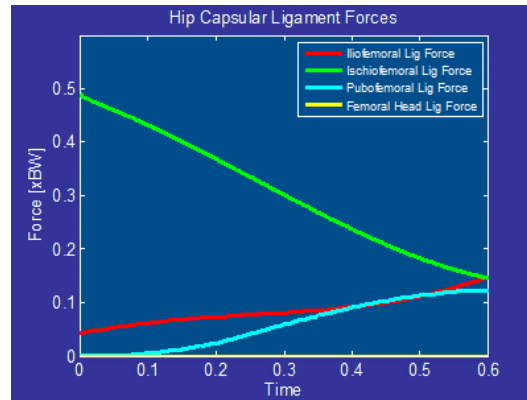


(b)

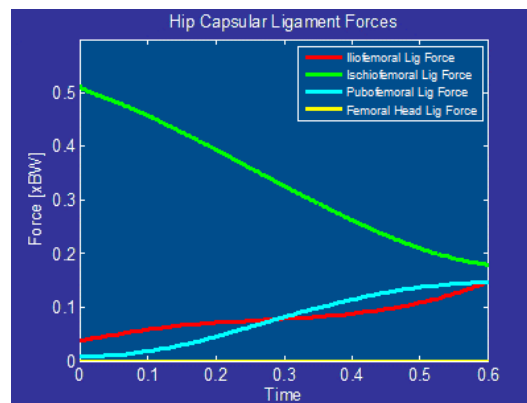


(c)

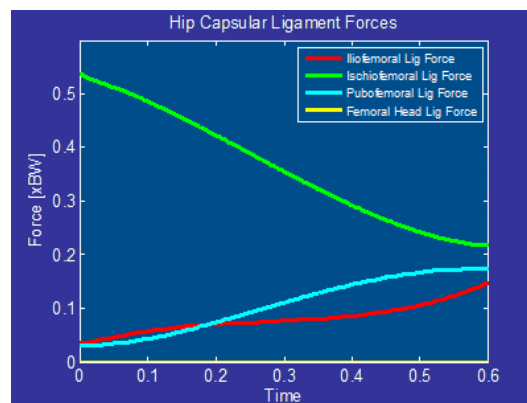
Figure 156: Hip muscle force profiles for (a) +5 degrees, (b) +10 degrees, and (c) +15 degrees of version error of the femoral component.



(a)

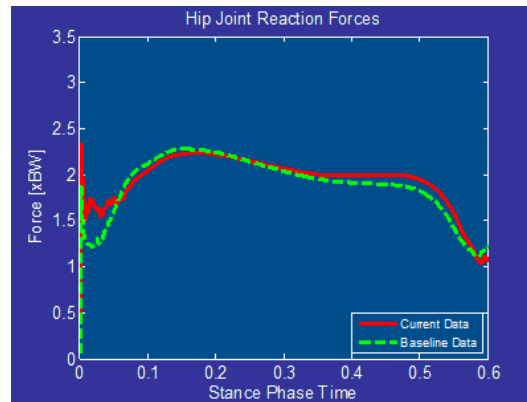


(b)

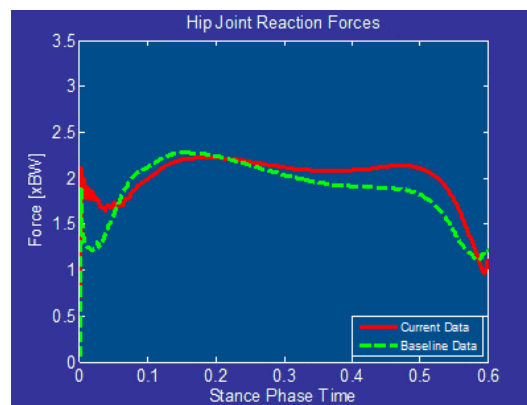


(c)

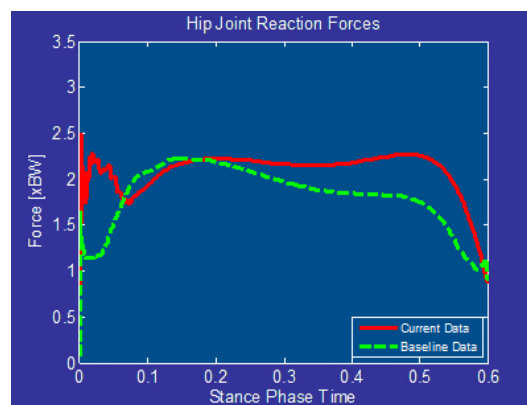
Figure 157: Hip capsular ligament force profiles for (a) +5 degrees, (b) +10 degrees, and (c) +15 degrees of version error of the femoral component.



(a)



(b)



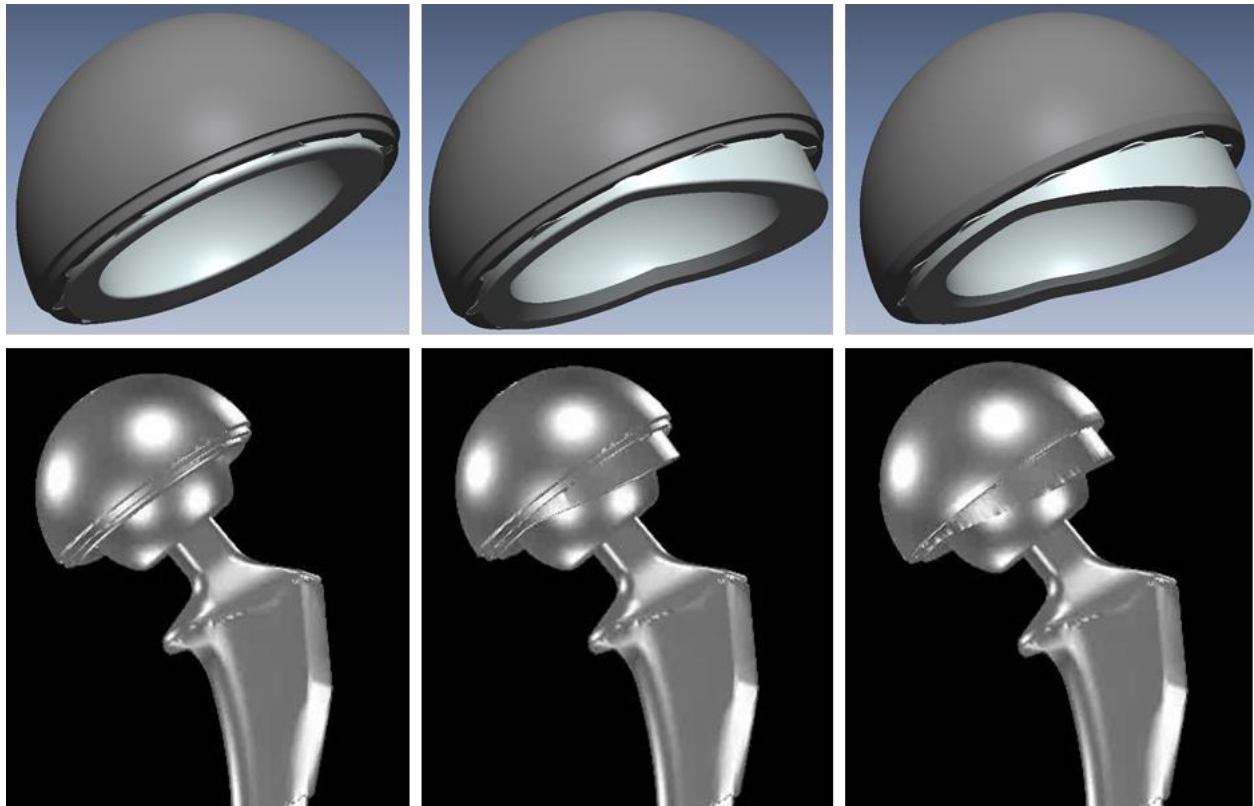
(c)

Figure 158: Hip joint reaction force profiles for (a) +5 degrees, (b) +10 degrees, and (c) +15 degrees of version error of the femoral component.

7.3.1 Liner Specifications

The three liners of interest can be seen in Figure 159. The Neutral liner (Figure 160) does not contain a lateral lip and is hence contains rotational symmetry about the liner's z axis (refer to Figure 40). Thus, this liner is resistant to internal/external rotational surgical errors. Conversely, both the Lipped (Figure 161) and Augment (Figure 162) liners contain a lip on the lateral side and thus do not contain rotational symmetry about the liner's z axis. Hence, these liners are prone to internal/external surgical misalignment, which affects kinematics and hip stability. Other key differences between the three liners include internal radius, liner lateralization, the Cylindrical Bore (Figure 163), and the Charnley Bore (for the Augmented liner).

Note in Figure 160, Figure 161, and Figure 162 the difference between the "Rotation Center" and "Liner Center" of the liners. The Rotation Center (RC) is based purely on the center of the articulating surface (inner surface) of the liner, thus negating the effects of liner lateralization. The Liner Center (LC) is based on the actual center of the liner CAD model, thus including the effects of liner lateralization. This differentiation will dramatically impact the analyses: aligning the acetabular component based on the Rotation Center will yield kinematic differences based solely on the design of the articulating surface (i.e. radius, Cylindrical and Charnley Bore depths, lip angle, rim shape, etc.), while aligning the acetabular component based on the Liner Center will introduce another variable to affect kinematics (lateralization), thus making it more difficult to differentiate what is causing the changes in kinematics (but simultaneously making the analyses more realistic). Both alignment definitions have been analyzed, and the proceeding analyses will clarify which alignment method is being used.



(a)

(b)

(c)

Figure 159: (a) Neutral, (b) Lipped, and (c) Augmented liners.

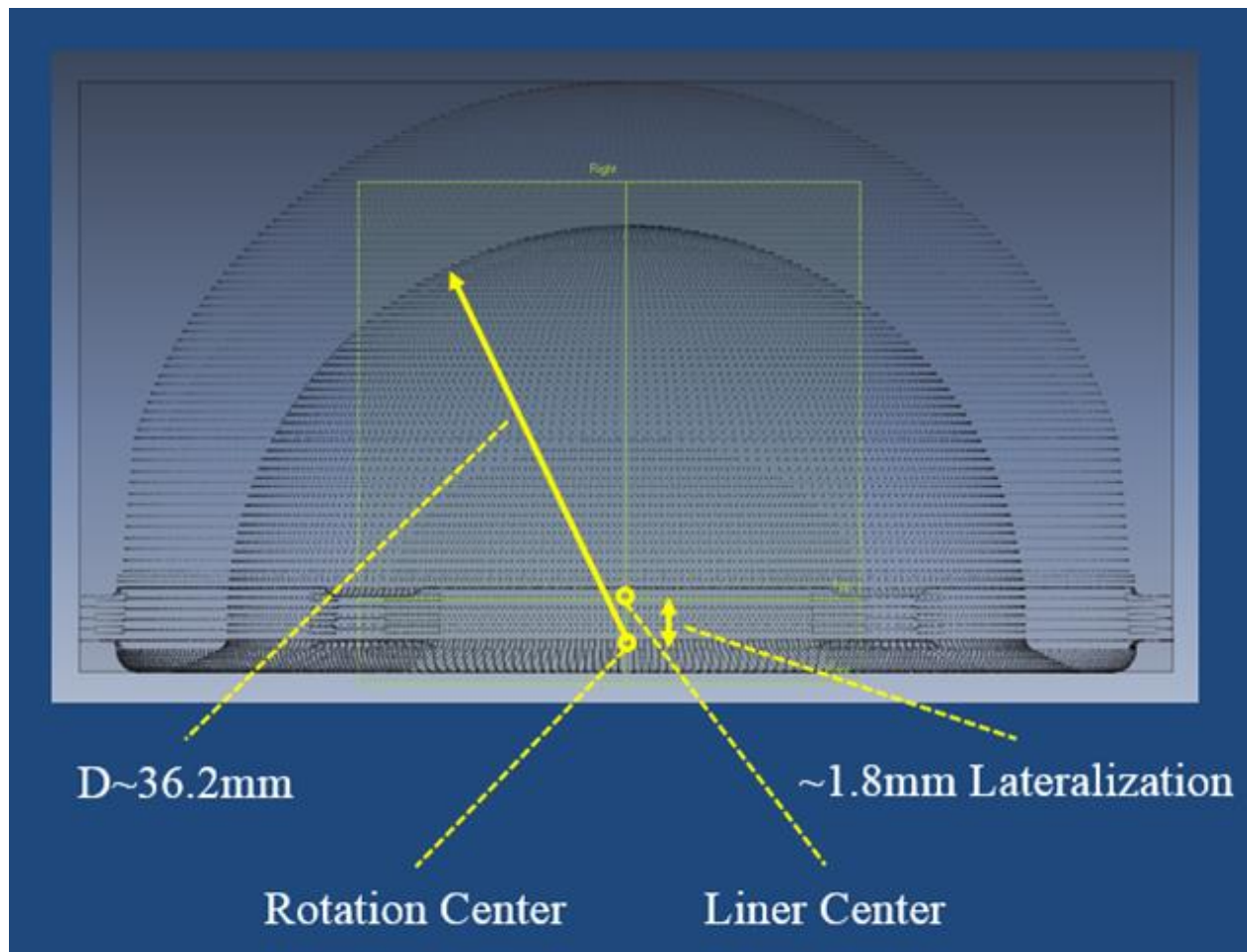


Figure 160: Neutral liner schematics.

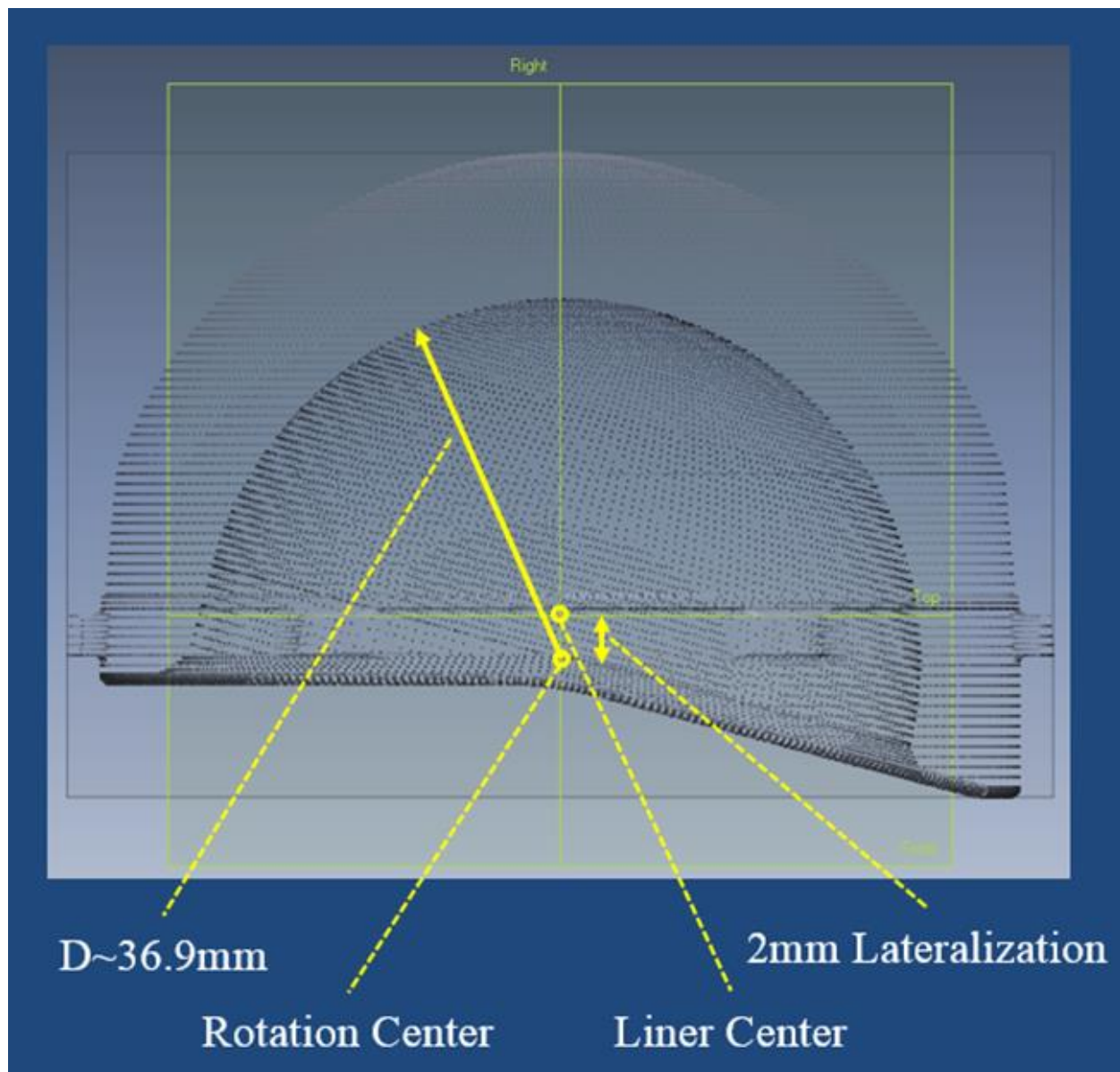


Figure 161: Lipped liner schematics.

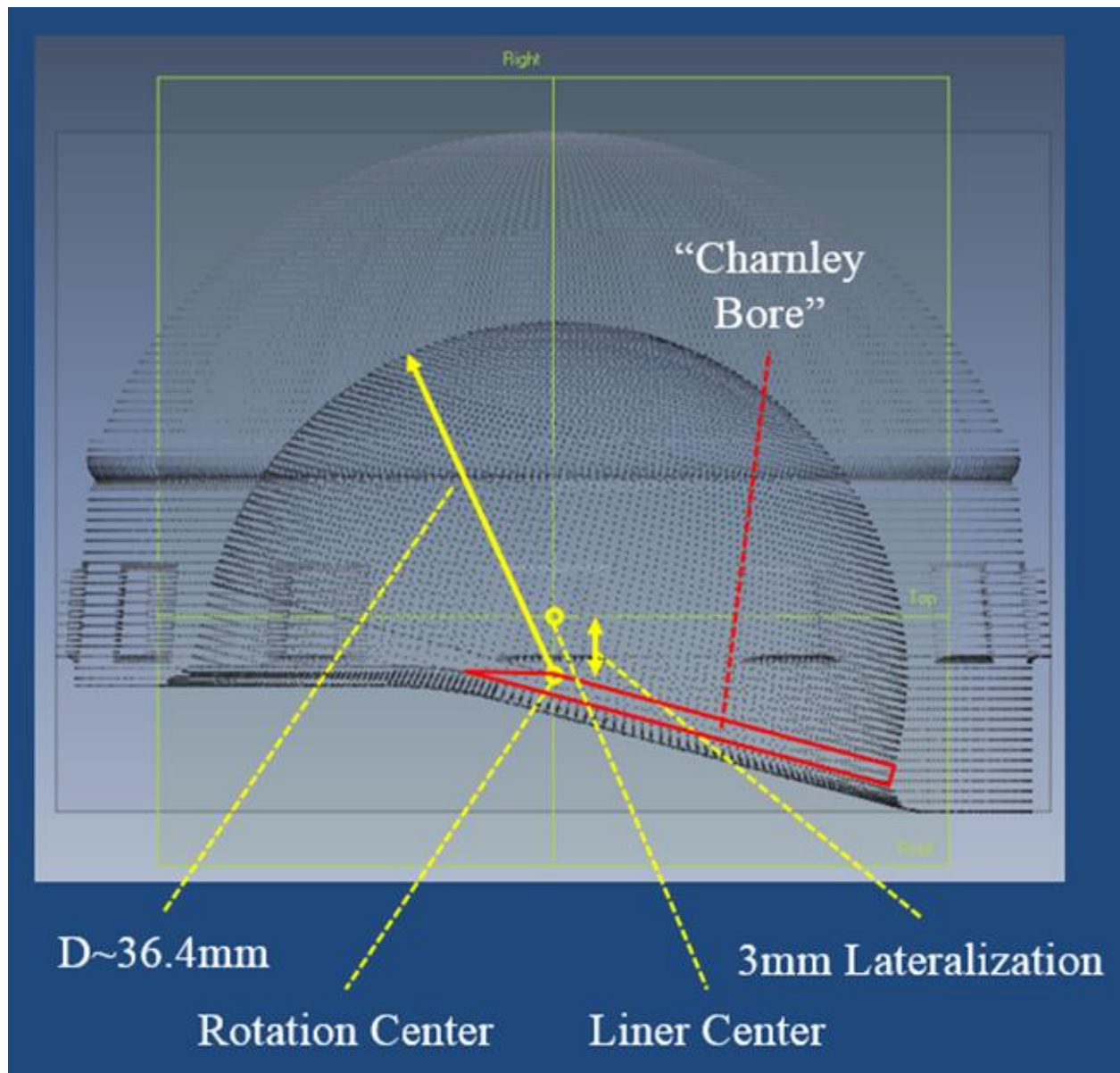


Figure 162: Augmented liner schematics.

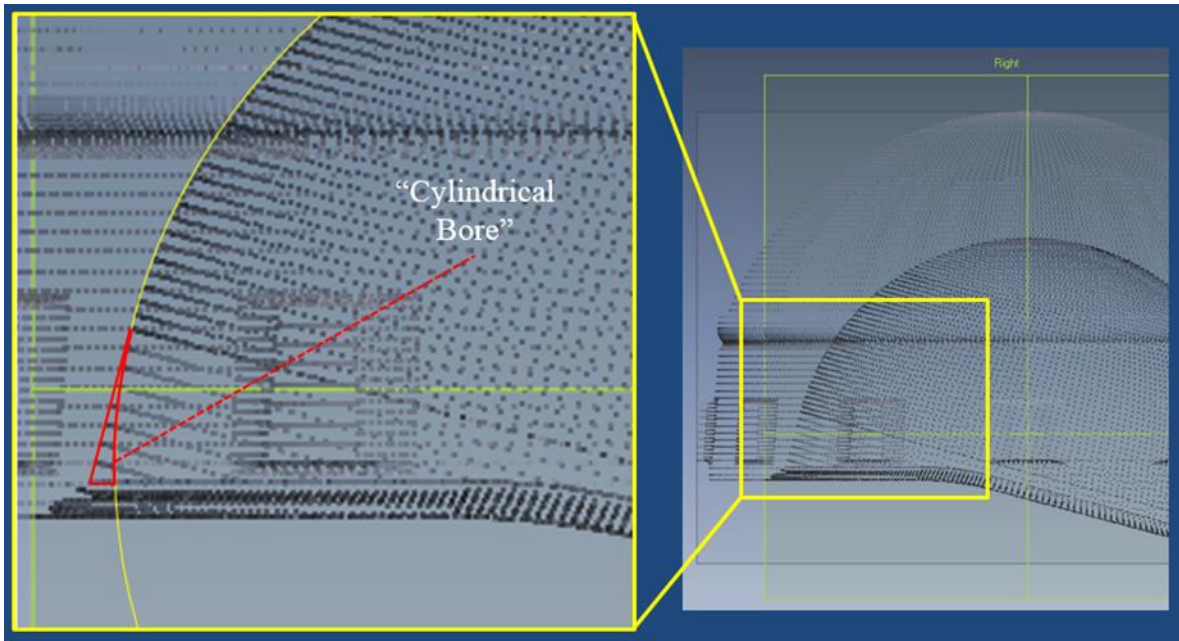


Figure 163: Cylindrical Bore, present on both Augmented and Lipped liners.

7.3.2 Liner Range of Motion Comparison

The first comparison conducted between the three liners was an analysis of their Range-of-Motion (ROM). The ROM analysis was conducted with the acetabular cup anatomically aligned, placed at 45° inclination and 10° anteversion, and the femoral stem at 10° valgus. Figure 164 shows the overall hip system in a neutral position.

Because the Lipped and Augmented liners are highly dependent on I/E rotation, the ROM analysis was conducted with each liner at six different angles of I/E rotation: 0°, +45°, +90°, +180°, -45°, and -90°. Figure 165a shows the acetabular cup implanted at 0°, and Figure 165b shows the positive axis of rotation for this analysis.

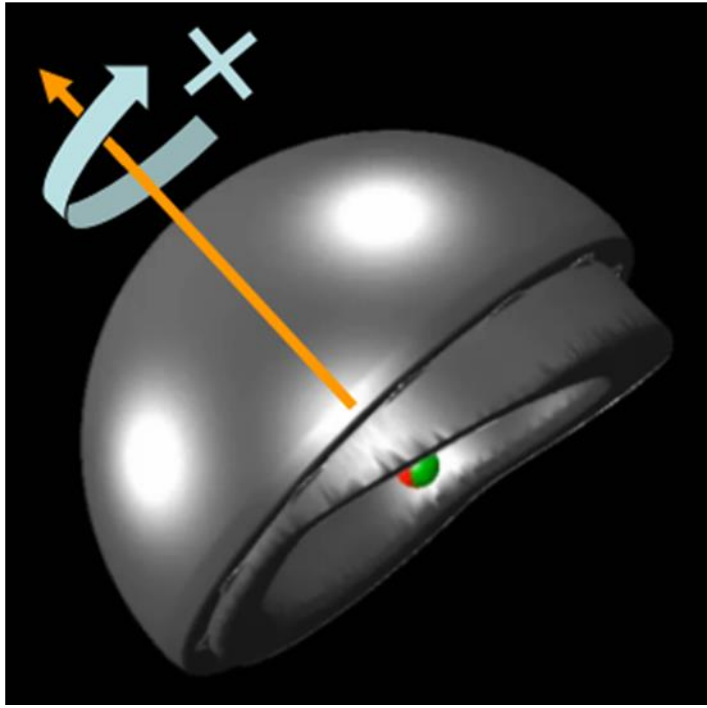
The ROM analysis was conducted to determine A/P, S/I, and M/L ROMs of the femoral component. Figure 166 shows positive A/P rotation of the femoral component, and Table 6 shows the A/P ROM results; Figure 167 shows positive S/I rotation of the femoral component, and Table 7 shows the S/I ROM results; Figure 168 shows positive M/L rotation of the femoral component, and Table 8 shows the M/L ROM results. Overall, it appears that the lips associated with the Lipped and Augmented liners have the greatest effect on S/I rotation of the femoral component (femur internal/external rotation), yielding up to $\pm 25^\circ$ of ROM changes compared to the Neutral liner (control). Changes to the M/L rotation of the femoral component (femur flexion/extension) are less than or equal to approximately -14° to $+18^\circ$, and changes to the A/P rotation of the femoral component (femur abduction/adduction) are less than or equal to approximately $\pm 8^\circ$. Moreover, whether the lipped liners increase or decrease the ROM appears to be highly depended on the I/E orientation of the acetabular cup, thus emphasizing the importance of proper surgical alignment.



Figure 164: Acetabular cup at 45° inclination and 10° anteversion; femoral stem at 10° valgus.



(a)



(b)

Figure 165: (a) acetabular cup with 0° of I/E rotation and (b) positive rotation of the cup.



Figure 166: Positive A/P rotation of the femoral component.

Table 6: A/P ROM Analysis.

Liner	Angle [deg]	Range	ROM
Neutral	0	-67.5 - 57.5	125
	45	-67.5 - 57.5	125
	90	-67.5 - 57.5	125
	180	-67.5 - 57.5	125
	-45	-67.5 - 57.5	125
	-90	-67.5 - 57.5	125
Lipped	0	-55 - 62.5	117.5
	45	-58 - 62	120
	90	-69 - 58.5	127.5
	180	-73 - 43.5	116.5
	-45	-60 - 62	122
	-90	-71.5 - 61	132.5
Augmented	0	-55.5 - 63.5	119
	45	-57.5 - 63	120.5
	90	-66.5 - 57.5	124
	180	-74 - 45.5	119.5
	-45	-60 - 63.5	123.5
	-90	-69.5 - 59	128.5

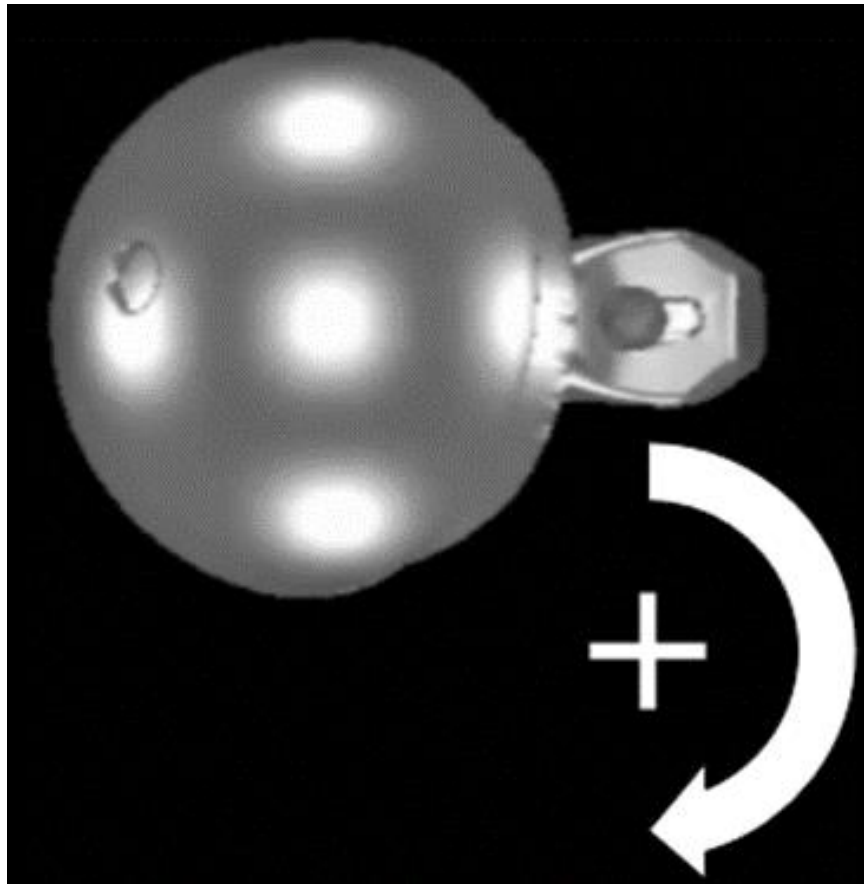


Figure 167: Positive S/I rotation of the femoral component.

Table 7: S/I ROM Analysis.

Liner	Angle [deg]	Range	ROM
Neutral	0	-88.5 - 116.5	205
	45	-88.5 - 116.5	205
	90	-88.5 - 116.5	205
	180	-88.5 - 116.5	205
	-45	-88.5 - 116.5	205
	-90	-88.5 - 116.5	205
Lipped	0	-97.5 - 126.5	224
	45	-86.5 - 127	213.5
	90	-68 - 125.5	193.5
	180	-79.5 - 101.5	181
	-45	-99 - 125.5	224.5
	-90	-98.5 - 101.5	200
Augmented	0	-100.5 - 129.5	230
	45	-84 - 129.5	213.5
	90	-69 - 128	197
	180	-78 - 102	180
	-45	-101 - 122.5	223.5
	-90	-101 - 100.5	201.5



Figure 168: Positive M/L rotation of the femoral component.

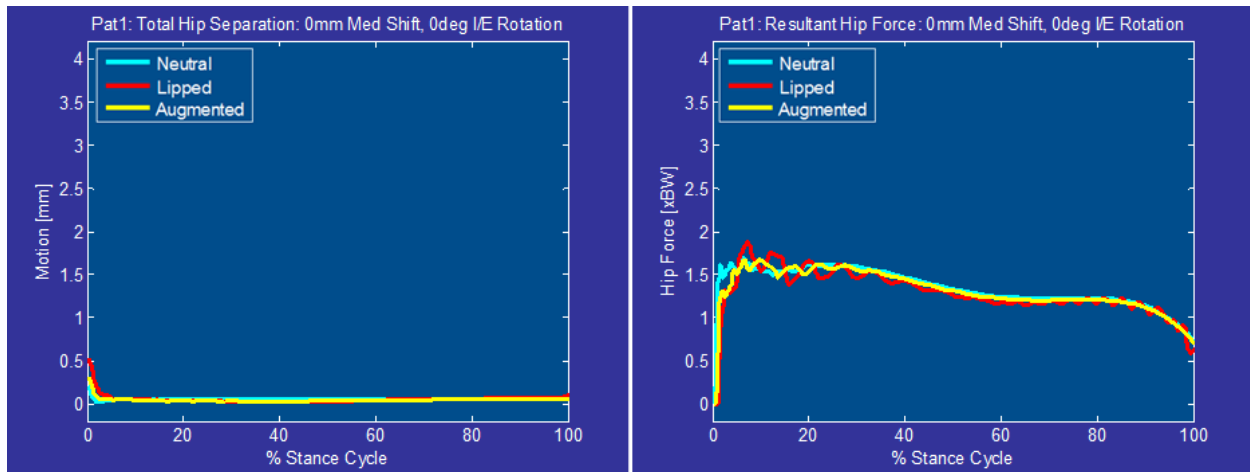
Table 8: M/L ROM Analysis.

Liner	Angle [deg]	Range	ROM
Neutral	0	-104 - 76	180
	45	-104 - 76	180
	90	-104 - 76	180
	180	-104 - 76	180
	-45	-104 - 76	180
	-90	-104 - 76	180
Lipped	0	-95 - 73.5	168.5
	45	-110.5 - 60.5	171
	90	-111.5 - 55.5	167
	180	-111 - 83	194
	-45	-85 - 82.5	167.5
	-90	-86 - 84	170
Augmented	0	-93.5 - 72.5	166
	45	-107.5 - 61.5	169
	90	-113.5 - 58	171.5
	180	-113.5 - 85	198.5
	-45	-86.5 - 85	171.5
	-90	-88 - 86	174

7.3.3 Cup Medialization Comparison with Rotation Center Alignment

A hip separation analysis was conducted to compare the effects of cup medialization for the three liners of interest. The analysis was conducted on the four “subjects” that have been used thus far. Each subject was implanted with 45° of cup inclination and 10° of cup anteversion, and the effects of liner I/E rotations were examined at the six aforementioned I/E rotation angles. A total of 5 different medial shift scenarios were examined: 0mm of intended medial cup shifting with 0mm, 2mm, 5mm, 8mm, and 10mm of actual cup medial shifting (refer to Figure 110a), yielding a total of 120 different scenarios. The analysis was conducted assuming Rotation Center alignment to compare the kinematic differences based purely on the articulating surface geometry. A Liner Center alignment comparison is in the next section.

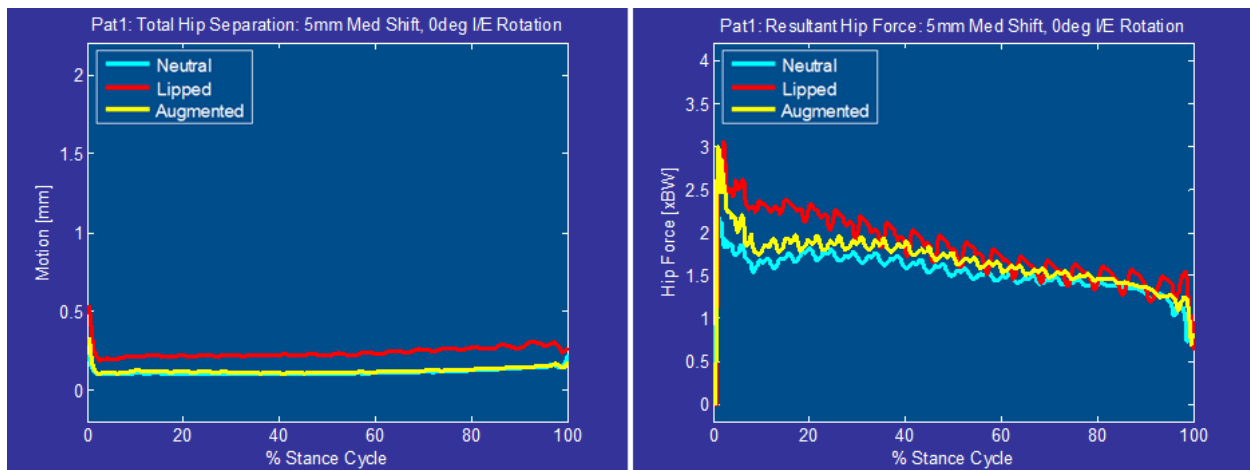
Due to the large nature of this analysis, the results will be briefly summarized here, including a few sample figures. Figures for all scenarios can be found in Appendix C, found in the Supplemental Document. As a control, properly aligned cups (0mm medial shift) yielded no separation, regardless of cup I/E orientation (Figure 169). In general, medial shifts of 5mm or less yielded minimal differences in hip separation. These smaller medial shifts generally yielded minor increases in hip forces as well as noticeable hip instability, which can be seen from the oscillatory nature of the hip forces in Figure 170b, but overall the large-scale hip separation magnitudes remained minimally affected. At 8mm and 10mm of medial shifting, the orientation of the lipped liners had a substantial influence on hip separation. Correctly oriented liners reduced hip separation (Figure 171) while poorly oriented liners increased hip separation (Figure 172). In general, if the liner was oriented within $\pm 45^\circ$ of proper I/E orientation, the lips were effective at reducing hip



(a)

(b)

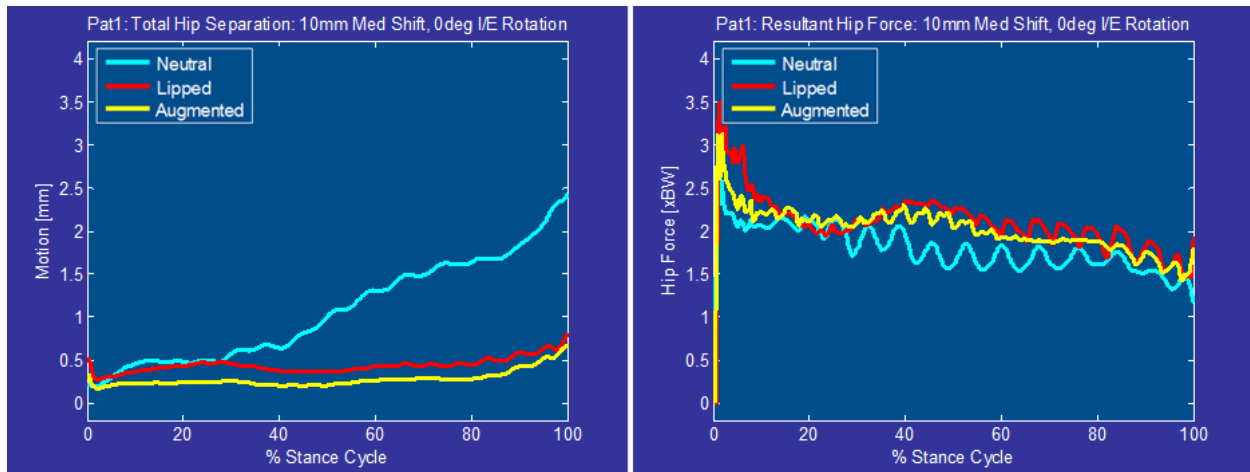
Figure 169: (a) hip separation and (b) resultant hip force liner comparison for Patient 1 with 0mm med shift, 0° I/E rotation with RC alignment.



(a)

(b)

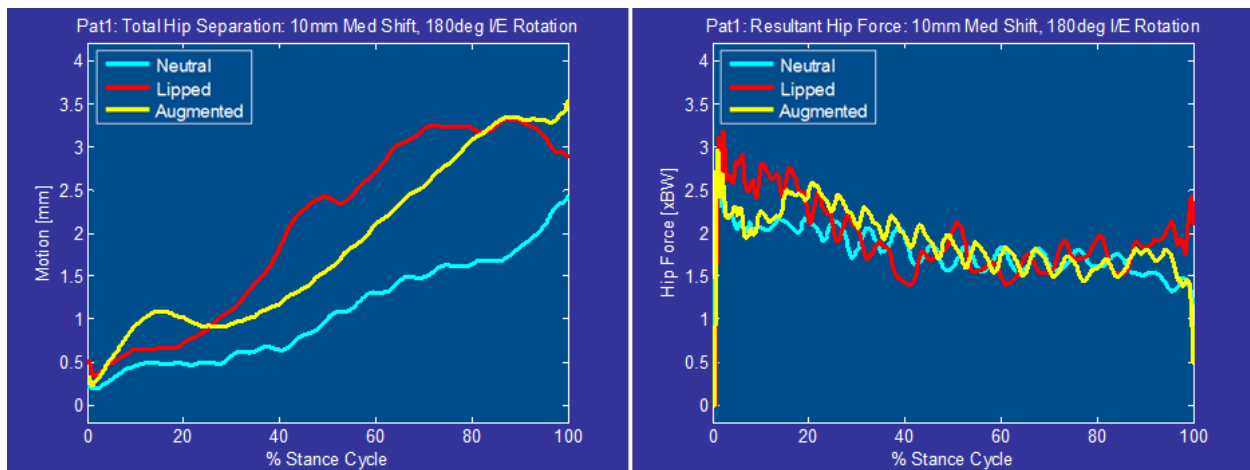
Figure 170: (a) hip separation and (b) resultant hip force liner comparison for Patient 1 with 5mm med shift, 0° I/E rotation with RC alignment.



(a)

(b)

Figure 171: (a) hip separation and (b) resultant hip force liner comparison for Patient 1 with 10mm med shift, 0° I/E rotation with RC alignment.



(a)

(b)

Figure 172: (a) hip separation and (b) resultant hip force liner comparison for Patient 1 with 10mm med shift, 180° I/E rotation with RC alignment.

separation. However, once the liners were oriented to $>\pm 90^\circ$, their effectiveness was reduced. Interestingly, if the lipped liners are oriented at 180° , hip separation is greater than the neutral liner. This is due to the lack of femoral head constraint induced by the Cylindrical Bore on the non-lipped side of the liner.

Overall, although minor, the Augmented liner generally appears to yield slightly less hip separation than the Lipped liner. It is hypothesized that this is primarily due to the articulating surface diameter difference. The larger diameter of the Lipped liner allows for more femoral head instability within the acetabular cup, thus yielding the potential for more hip separation.

7.3.4 Cup Medialization Comparison with Liner Center Alignment

The previous analysis was repeated for the same 120 scenarios, this time aligning the two lipped liners according to the Liner Centers. The neutral liner alignment remained as Rotational Center, to create a more generic neutral control with no lateralization. This will allow for a better understanding of the effects that liner lateralization alone has on hip separation, minimizing articulating surface design factors. Thus, the results from both the previous analysis and the current analysis should be considered when determining the optimal lateralization amount for novel articulating surfaces.

Once again, due to the large nature of this analysis, the results will be briefly summarized here, including a few sample figures. Figures for all scenarios can be found in Appendix D, found in the Supplemental Document. Lateralization of the cup can be both beneficial and detrimental, depending on the alignment. In scenarios where the cup is nearly anatomically aligned (0mm – 2mm), hip separation still does not occur, but

interestingly the lateralized liners cause an increase in hip forces (Figure 173, compare to Figure 169). This is due to the increase in capsular ligament tension from lateralizing the liner and not compensating by adjusting the cup position. Conversely, for larger amounts of cup medialization (8mm and 10mm), liner lateralization functions to further reduce hip separation. This is achieved by re-lateralizing a medialized cup, thus pushing the component rotation centers back towards the anatomical center. For correctly oriented liners (0°), lateralization will further reduce hip separation (Figure 174, compare to Figure 171). For incorrectly oriented liners (180°), the lateralization will essentially negate the separation caused by the incorrect orientation (Figure 175, compare to Figure 172), yielding comparable results to the control.

7.3.5 Cup Inclination Analysis with Rotational Center Alignment

A very brief analysis was conducted to begin to understand the benefits and detriments of adjusting cup inclination. Adding a lip to a liner is comparable to modifying the inclination a neutral acetabular cup by a fixed number of degrees (-15° , in this case; compare Figure 160 to Figure 161 and see Figure 176 for positive rotation direction), the primary difference being that a lesser-inclined neutral cup would not have the bony contact on the medial side that is present in the lipped liner because the neutral cup would consequently be overhanging the pelvic rim.

Nonetheless, if cup fixation is ignored, then reducing the inclination of the acetabular cup should have similar kinematic results to simply adding a lip. This is indeed the case. Figure 177 shows that, when decreasing the inclination of both (a) the Neutral and (b) the Augmented liners, hip separation seems to reduce.

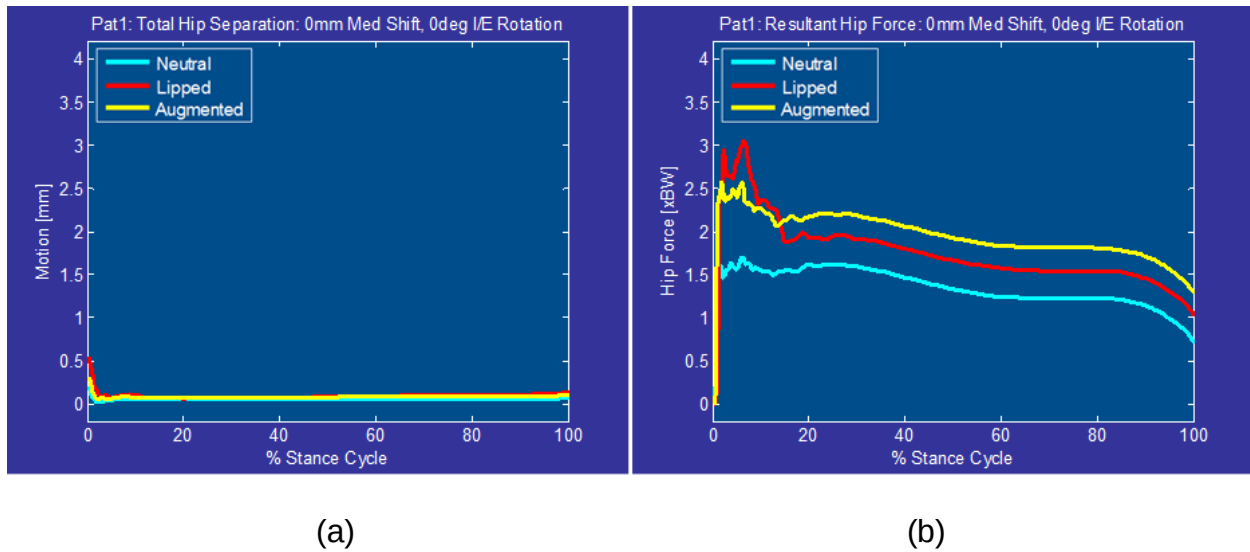


Figure 173: (a) hip separation and (b) resultant hip force liner comparison for Patient 1 with 0mm med shift, 0° I/E rotation with LC alignment.

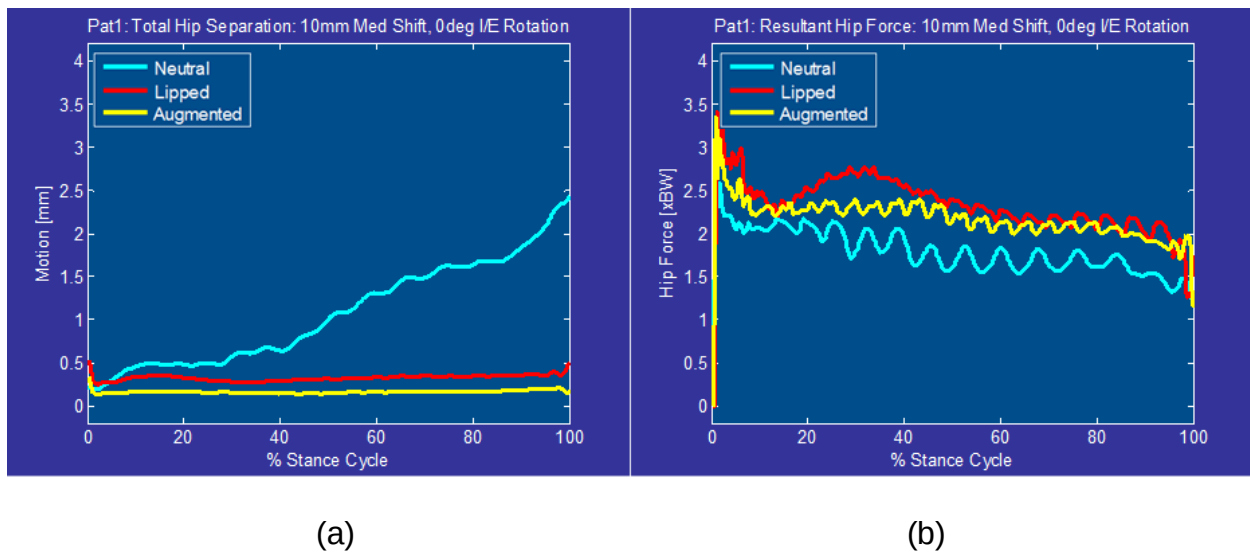
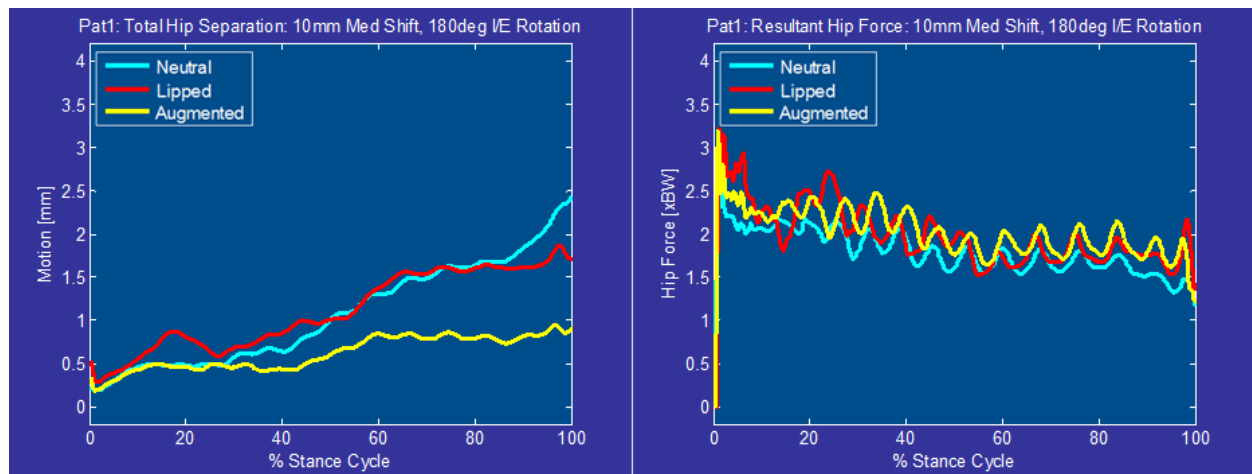


Figure 174: (a) hip separation and (b) resultant hip force liner comparison for Patient 1 with 10mm med shift, 0° I/E rotation with LC alignment.



(a)

(b)

Figure 175: (a) hip separation and (b) resultant hip force liner comparison for Patient 1 with 10mm med shift, 180° I/E rotation with LC alignment.

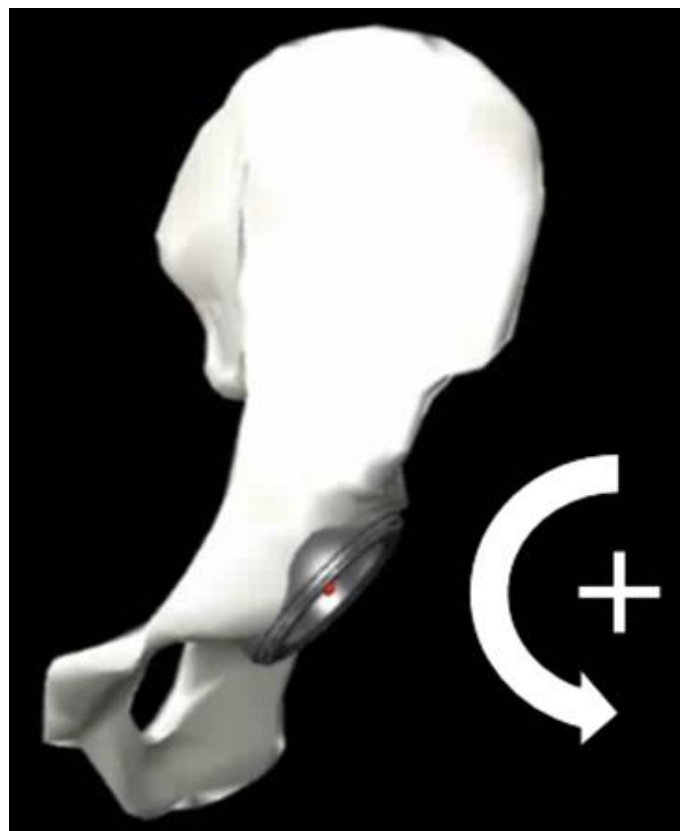


Figure 176: Positive inclination rotation of the cup.

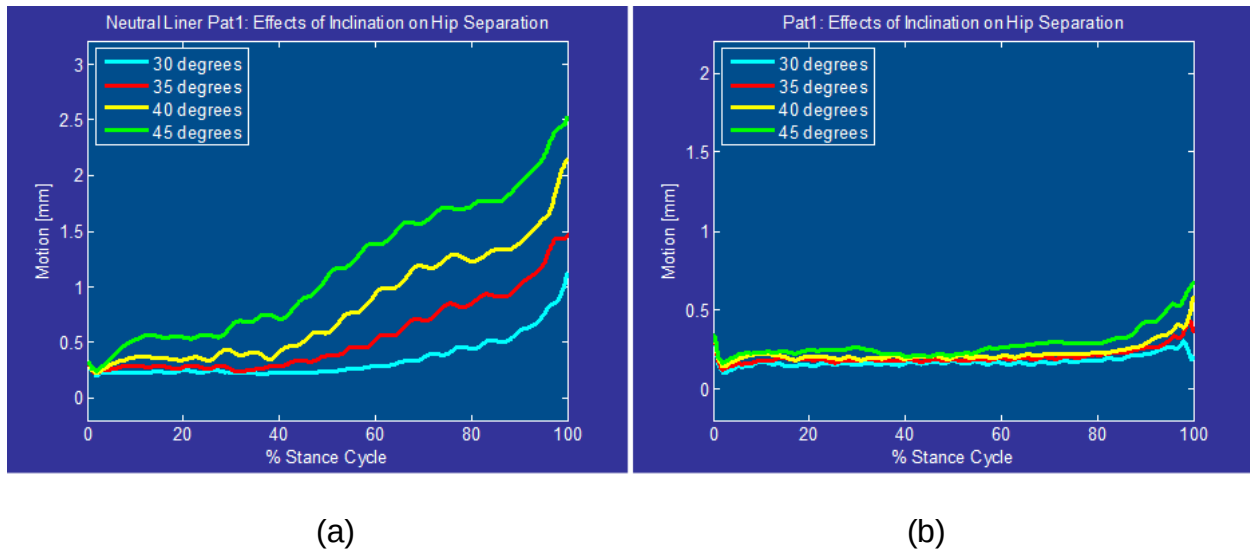


Figure 177: Effects of inclination modification on hip separation for (a) the Neutral liner and (b) the Augmented liner.

However, the brevity of this analysis constitutes discretion when analyzing the results. This analysis was only conducted on one patient under one extreme scenario: a 0mm intended medial shift but a 10mm actual medial shift. A lesser medial shift, differing patient kinematics, and different intended/unintended shifts in the S/I and A/P directions may yield different outcomes. Furthermore, the cup fixation issue, as well as consequential soft tissue rubbing from cup overhang, are problems that indeed cannot be ignored and will most certainly affect the postoperative kinematics for the patient. Hence, recommendations resulting from this analysis will be avoided until more detailed analyses can be conducted.

7.3.6 Anterior/Posterior Shifting Analysis with Rotational Center Alignment

The final analysis that was conducted comparing the three liner designs is an analysis to determine the effects of anterior (+) and posterior (-) shifting of the acetabular cup. The analysis was conducted at 7 A/P increments within the range of ± 9 mm. The analysis was conducted on all four patients, for all three liners, at all six degrees of cup I/E rotation. Because the Neutral liner is symmetrical, this liner was only analyzed at 0° of I/E rotation. Finally, the A/P shifts were coupled with either a 0mm medial shift or a 5mm medial shift, yielding a total of 728 scenarios. Figures of all scenarios can be found in Appendix E, found in the Supplemental Document.

When shifting a neutral cup in the A/P direction, it was determined that A/P shifting alone does not yield significant amounts of hip separation (Figure 178a). However, when accompanied with existing medial shifting, cup posterior shifting appears to exacerbate hip separation while anterior shifting continues to have minimal effects (Figure 178b).

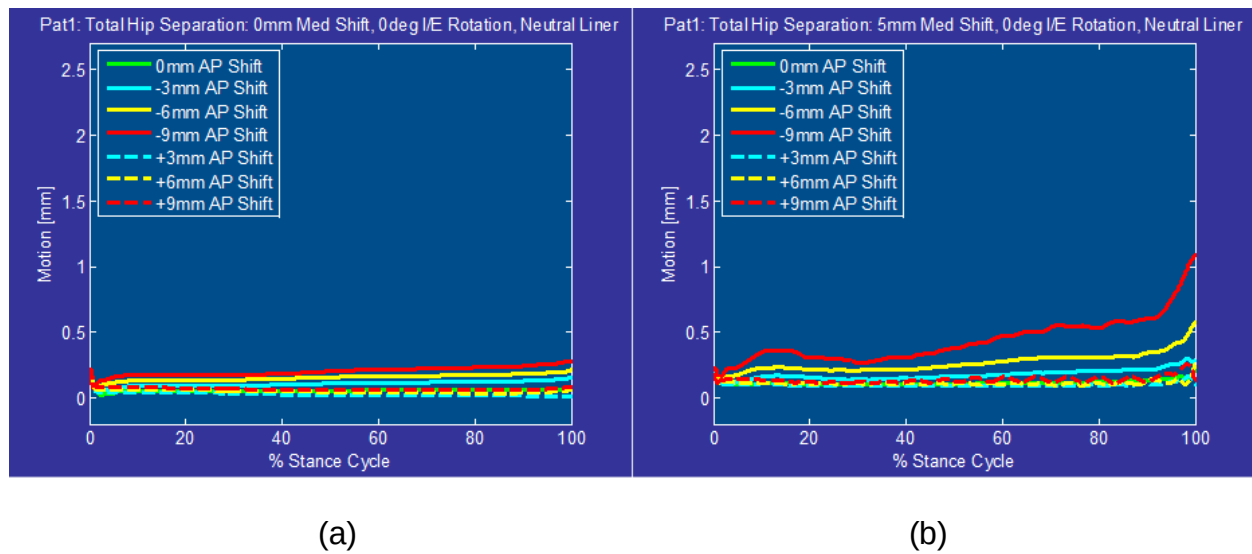
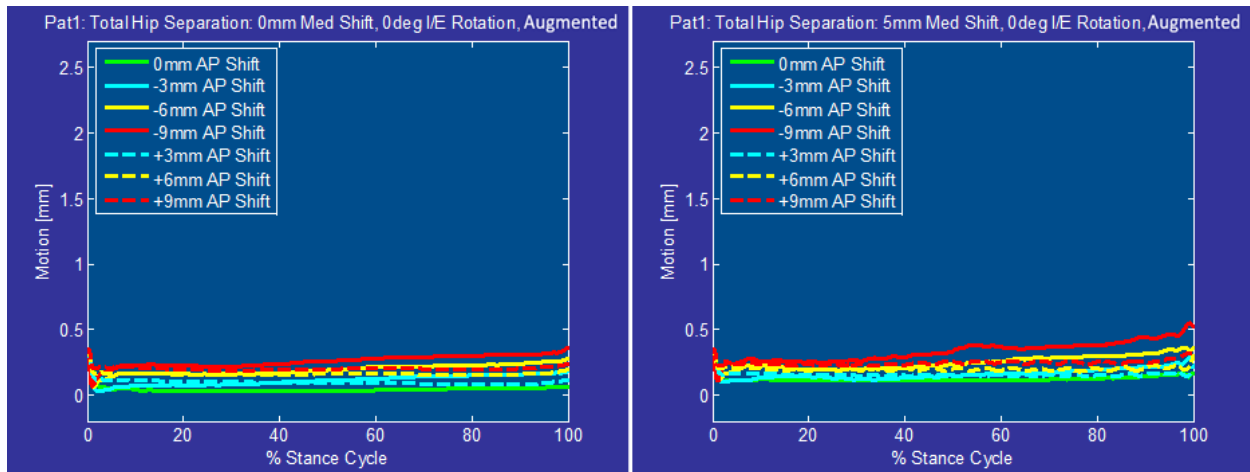


Figure 178: Patient 1 hip separation; Neutral Liner, 0° I/E Cup Rotation; Rotational Center (RC) alignment, for (a) 0mm and (b) 5mm medial shift.

The reasoning for this lies rooted in the fact that a properly anteverted cup opens anteriorly, which coincides with the native geometry of the pelvis. Specifically, because the proprioceptive hip separation hypothesis attempts to pull the femoral head back towards the anatomical center, the posterior shift of the acetabular component pulls the femoral head back anteriorly towards the anatomical center, generating hip separation in the anterior direction. Since the acetabular component is opening anteriorly, the femoral head is not as constrained by the bearing in that particular direction, thus allowing for anterior hip separation. Conversely, when shifting the cup anteriorly, the femoral head is being pulled posteriorly. However, the angle of the cup prevents the femoral head from separating in this direction.

With simply shifting the cup in the A/P direction, this proprioceptive hypothesis does not generate enough separation force to overcome the liner rim and the compressive hip force. However, when accompanied with additional medial shifting, the magnitude of this proprioceptive force is larger, and thus hip separation can occur.

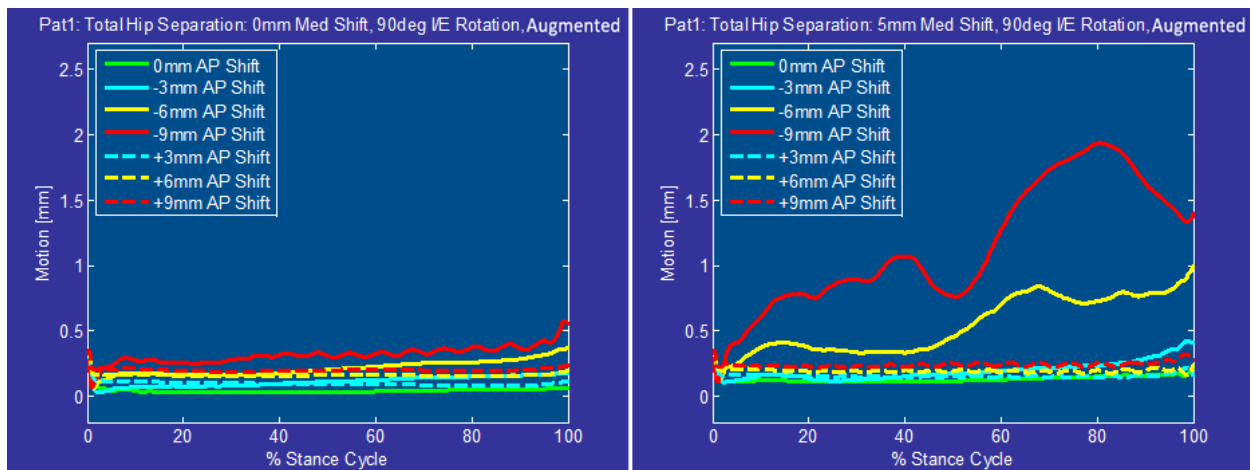
When properly aligned, lipped liners again do well to minimize hip separation (Figure 179). However, interestingly, this time the lipped liner I/E rotation error that yielded the most separation was 90° (as opposed to 180° in previous analyses). Figure 180 shows the separation associated with an Augmented liner at 90° of I/E rotation and Figure 181 shows the separation associated with an Augmented liner at 180° of I/E rotation. When the liner is rotated at 90°, the least constrained portion of the liner (the Cylindrical Bore) is the most anterior, thereby providing the least amount of femoral constraint on the anterior aspect of the liner. Conversely, when the cup is oriented at 180°, a portion of the lip is covering the anterior aspect of the rim, thus providing more anterior constraint at



(a)

(b)

Figure 179: Patient 1 hip separation; Augmented liner, 0° I/E Cup Rotation; Rotational Center (RC) alignment, for (a) 0mm and (b) 5mm medial shift.



(a)

(b)

Figure 180: Patient 1 hip separation; Augmented liner, 90° I/E Cup Rotation; Rotational Center (RC) alignment, for (a) 0mm and (b) 5mm medial shift.

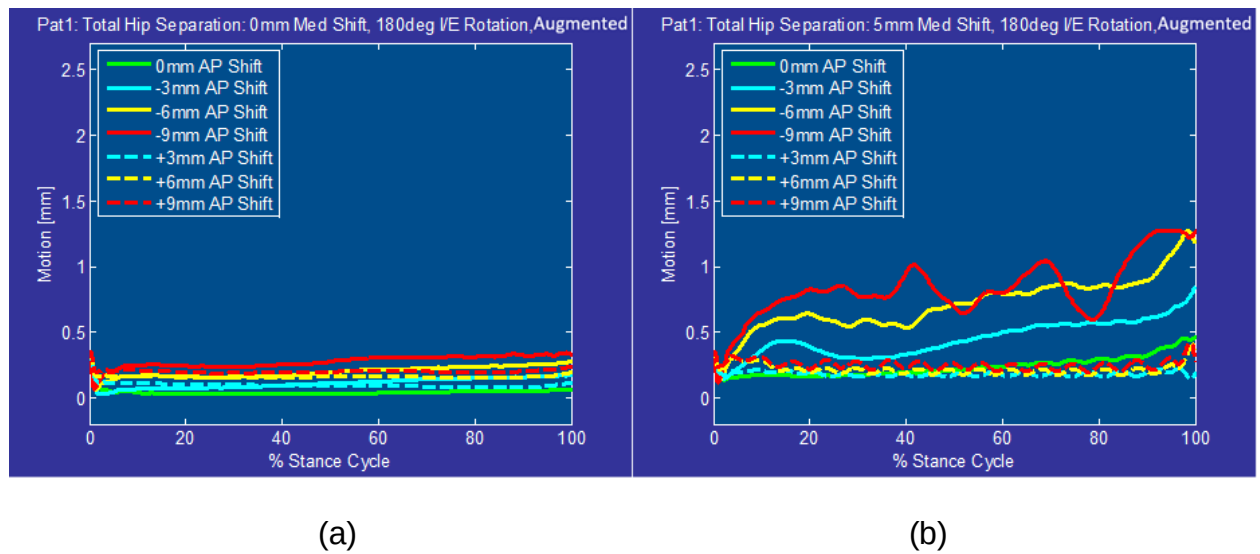


Figure 181: Patient 1 hip separation; Augmented liner, 180° I/E Cup Rotation; Rotational Center (RC) alignment, for (a) 0mm and (b) 5mm medial shift.

180° as opposed to 90°. Thus, because posterior shifting yields a pull on the femoral head in the anterior direction, the least constrained orientation of 90° yields the most hip separation.

7.3.7 Neutral versus Dual Mobility Analyses

7.3.7.1 Separation

The in vivo mechanics of the dual mobility design has been analyzed for both swing and stance phase. The dual mobility design in this analysis is not an actual implant cup available in the market, but rather a conceptual design used for this research study. Figure 182 shows the in vivo mechanics for a dual mobility implant during stance phase with proper alignment. Figure 183 shows the in vivo mechanics for a dual mobility implant during stance phase with a 10mm medial shift in the acetabular cup, yielding approximately 2mm of separation. To compare, Figure 184 shows the in vivo mechanics for a fixed bearing, neutral implant during stance phase with a 10mm medial shift in the cup, yielding approximately 1.5mm of separation. In this case, the dual mobility implant appears to yield slight increases in hip separation during stance phase. However, this is not due to the dual mobility nature of the design. Rather, this is due to the lack of a cylindrical bore on the acetabular cup (refer to Figure 43 if necessary). In other words, the cup that mated with the dual mobility bearing in this analysis was exactly a half-sphere, with no added cylindrical bore. In comparison, the neutral liner contains a ~1.5mm cylindrical bore, in addition to the half-sphere, that adds stability in cases that yield edge-loading. Thus, adding a cylindrical bore to the dual mobility design would yield decreases in hip separation during stance phase, but this would come at the cost of a reduce ROM.

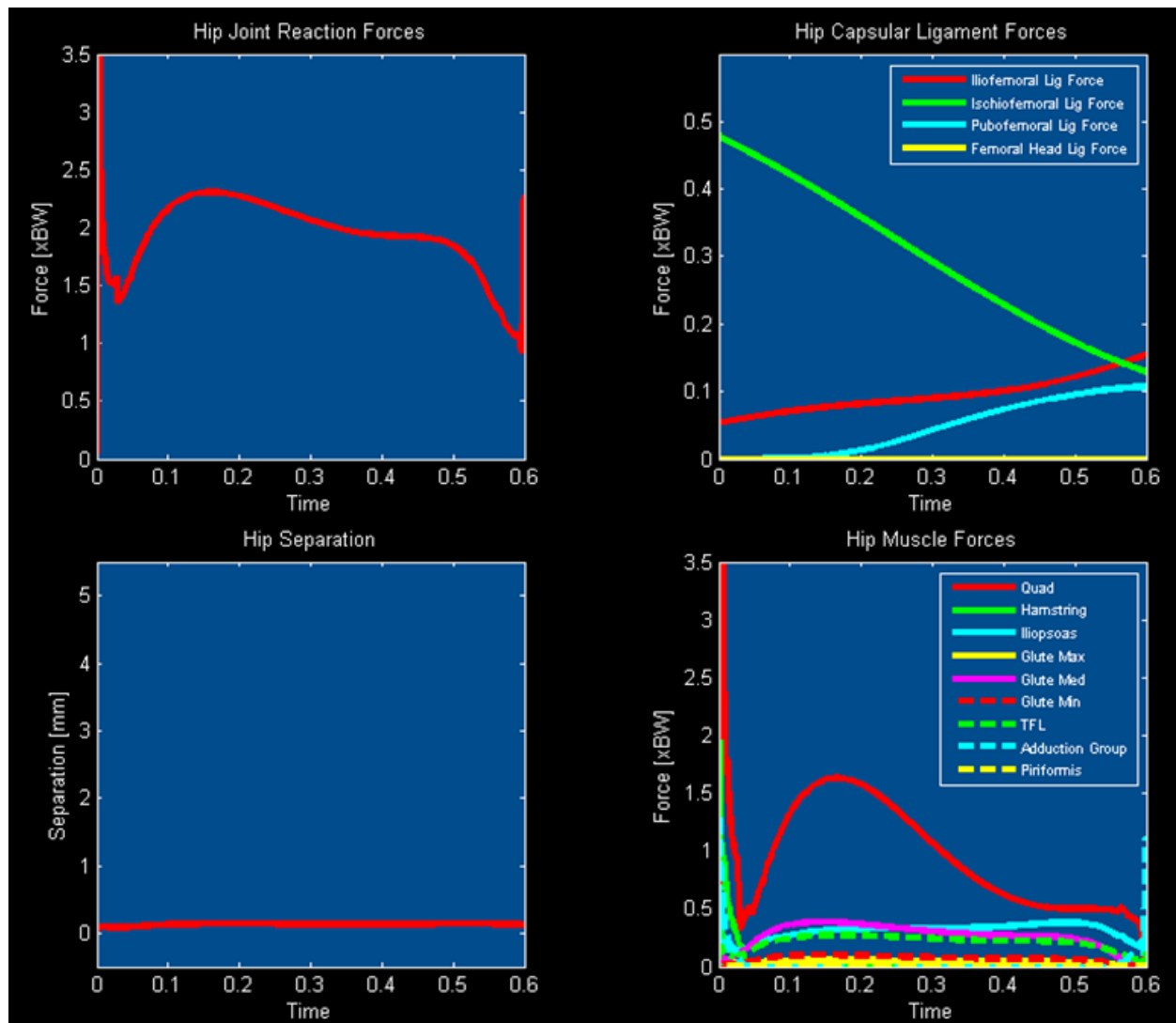


Figure 182: In vivo mechanics for a dual mobility design, stance phase, aligned components.

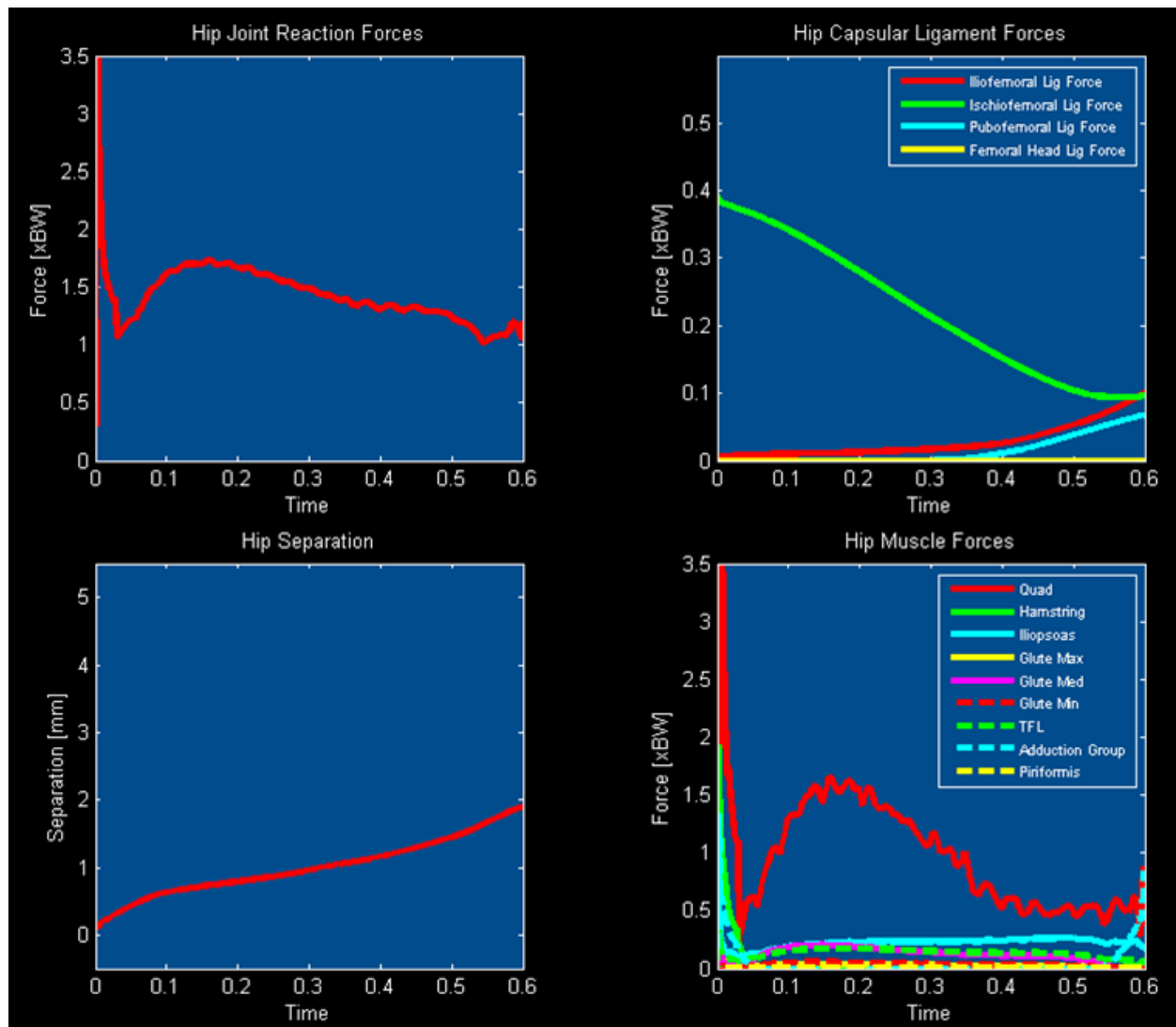


Figure 183: In vivo mechanics for a dual mobility design, stance phase, with a 10mm medial shift in the acetabular component. ~2mm of hip separation.

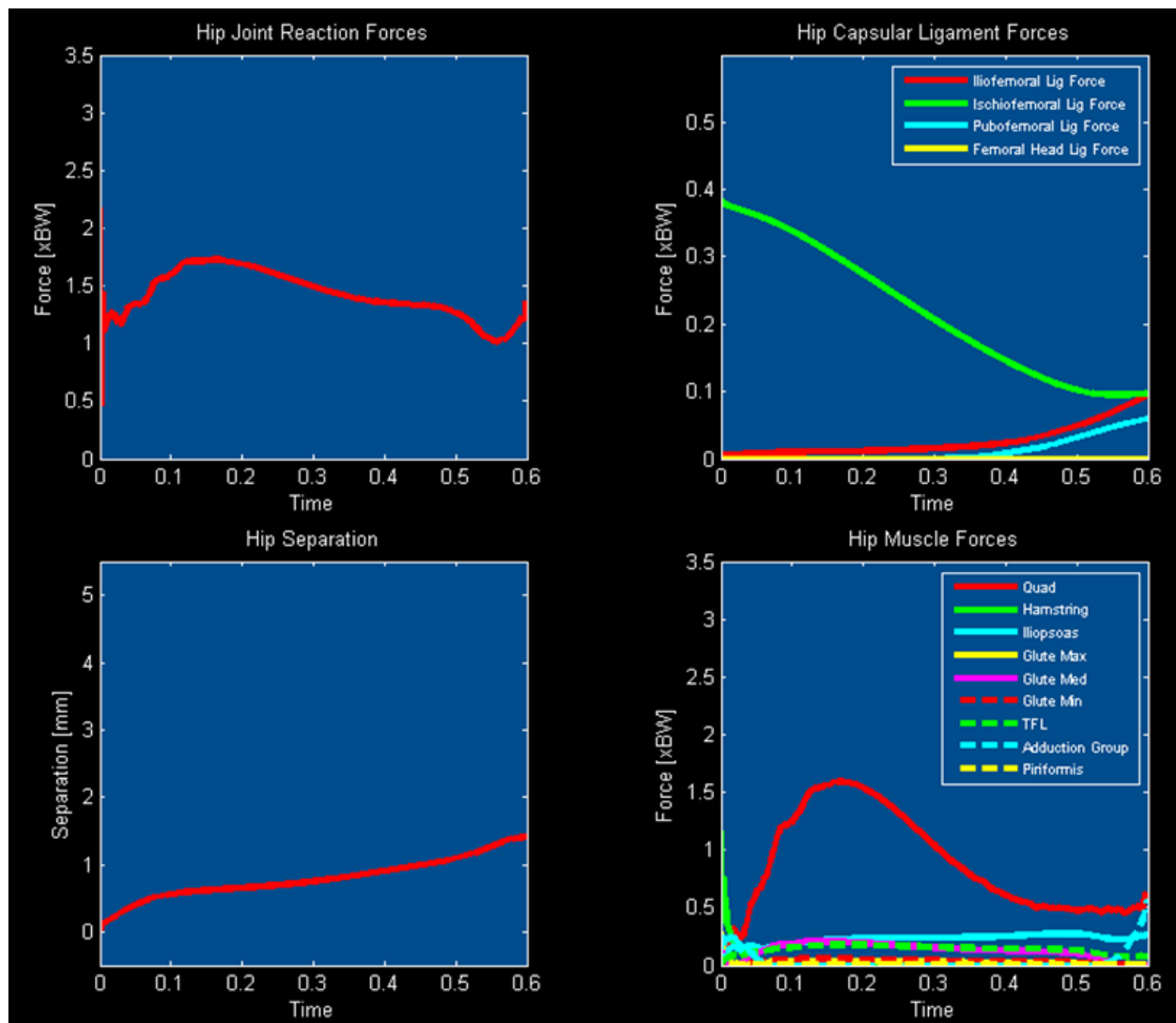


Figure 184: In vivo mechanics for a fixed, neutral design, stance phase, with a 10mm medial shift in the acetabular component. For comparison, showing ~1.5mm of hip separation.

Figure 185 shows the in vivo mechanics for a dual mobility implant during swing phase with proper alignment. Figure 186 shows a dual mobility implant with a 7mm superior shift in the acetabular cup, yielding approximately 2.3mm of hip separation. To compare, Figure 187 shows a 7mm superior shift in a neutral acetabular cup, again yielding approximately 2.3mm of hip separation. In this case, the cylindrical bore does not add stability to the system, as the type of separation that is experienced during swing phase is a pistoning motion, as opposed to sliding and edge-loading.

7.3.7.2 Contact Area

The following analysis was conducted on the stance phase of gait, comparing two specific implant types: a neutral liner with a 36mm internal diameter and a dual mobility cup with a 48mm internal diameter. Due to the increased diameter of the articulating surface of the cup, the model predicts an increase in contact area and therefore a decrease in contact stresses. Figure 188 shows visualizations of the contact maps of the articulating surfaces, and Figure 189 shows a comparison of the contact areas and contact stresses. It is clear from Figure 189 that the dual mobility implant yields an articulating contact area of nearly twice as much as the neutral implant. Logically, with an increase in contact area, there is a decrease in the contact stress at the cup's articulating surface. In general, this analysis goes deeper than just fixed bearing vs. dual mobility, as the analysis would yield similar results for fixed liners with larger internal diameters.

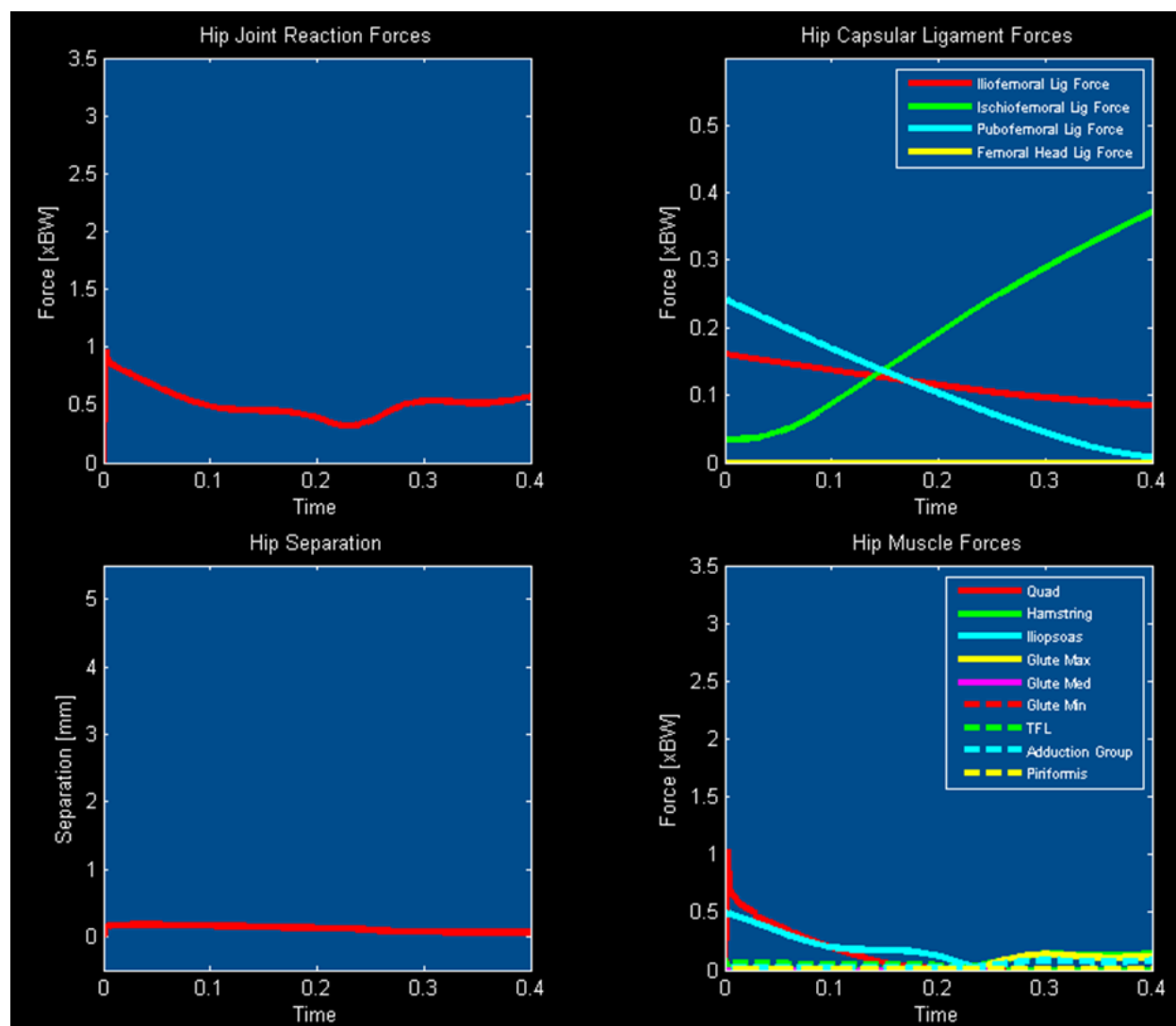


Figure 185: In vivo mechanics for a dual mobility design, swing phase, aligned components.

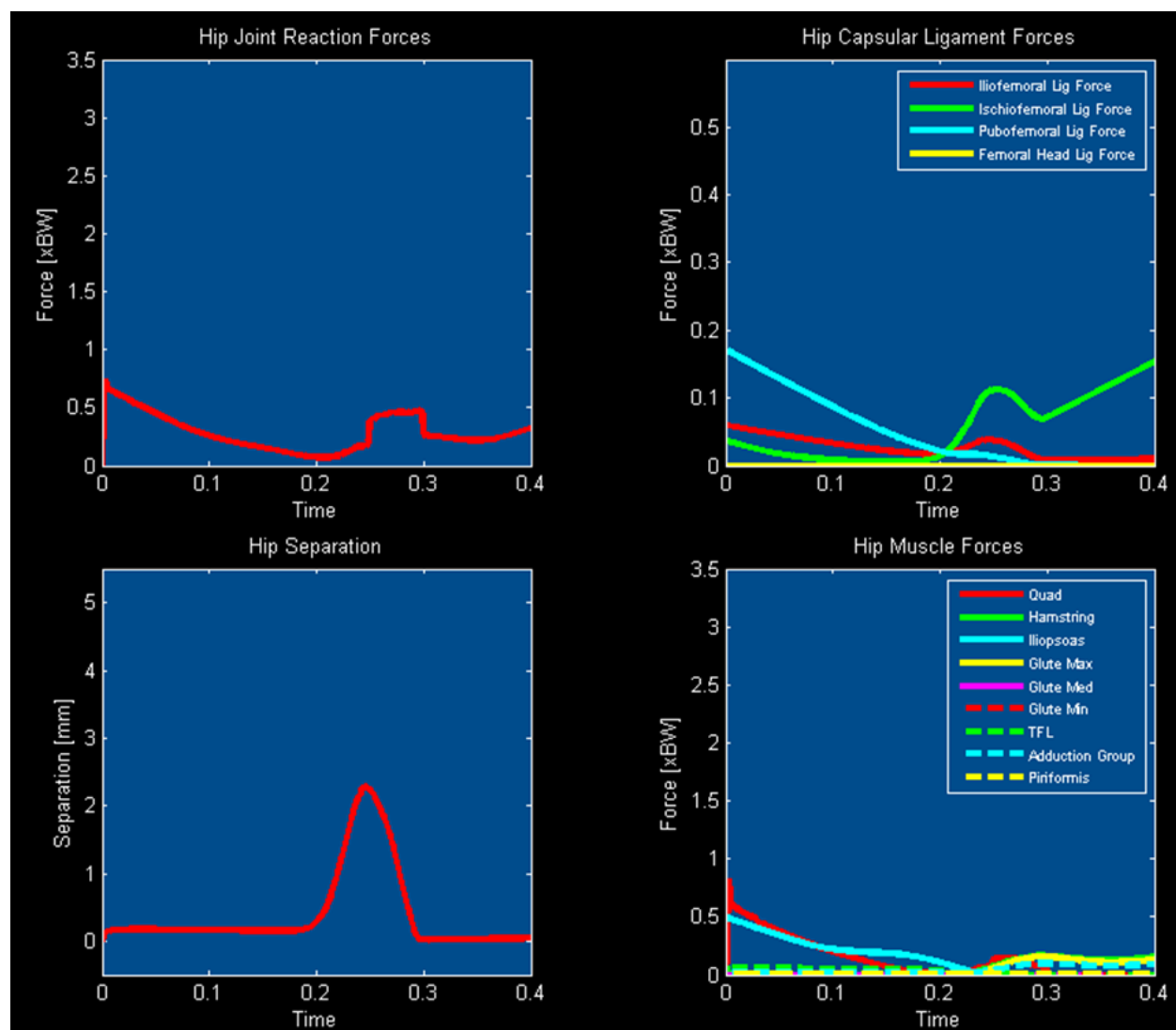


Figure 186: In vivo mechanics for a dual mobility design, swing phase, with a 7mm superior shift in the acetabular component. ~2.3mm of hip separation.

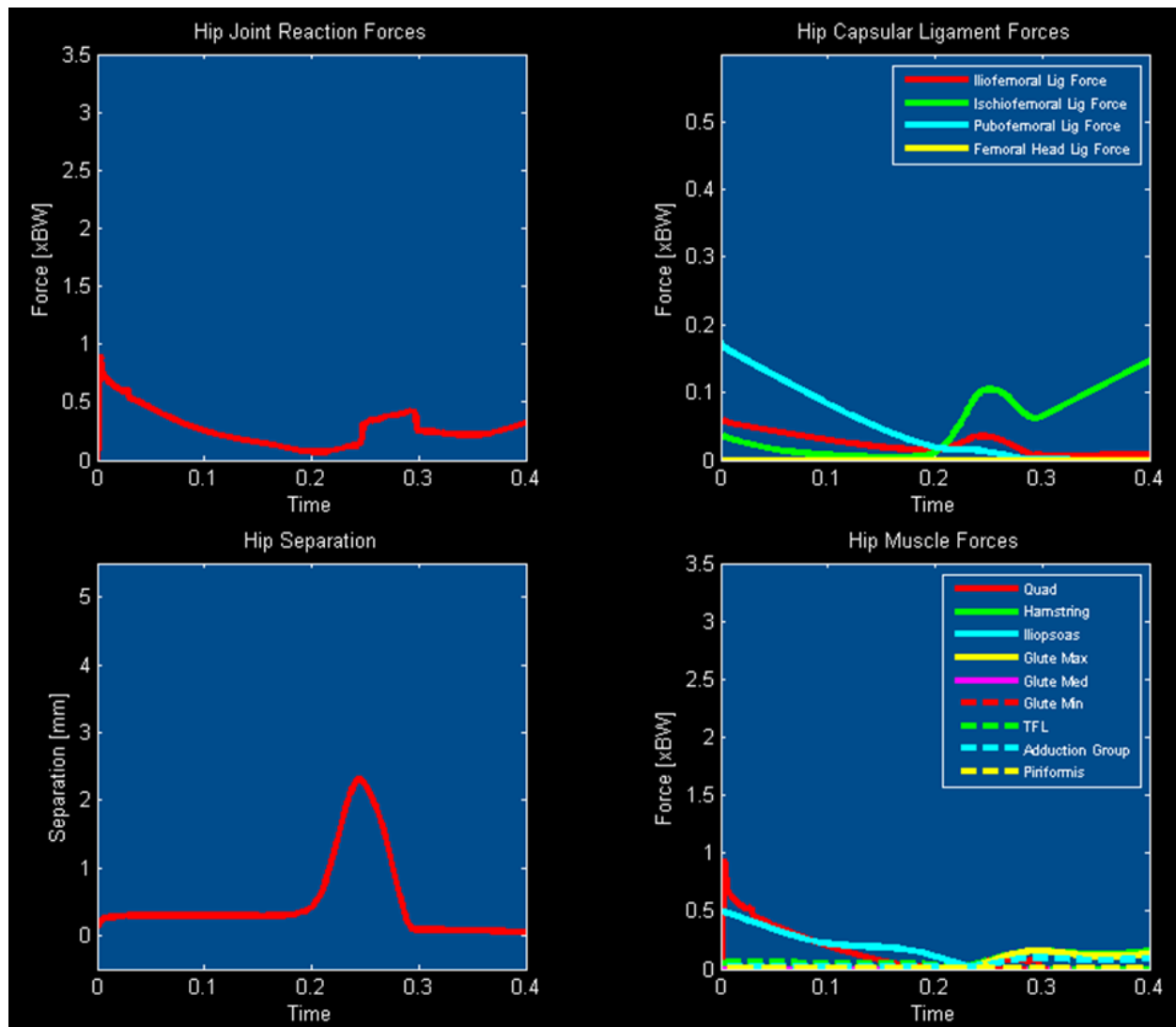


Figure 187: In vivo mechanics for a fixed, neutral design, swing phase, with a 7mm superior shift in the acetabular component. For comparison, showing ~2.3mm of hip separation.

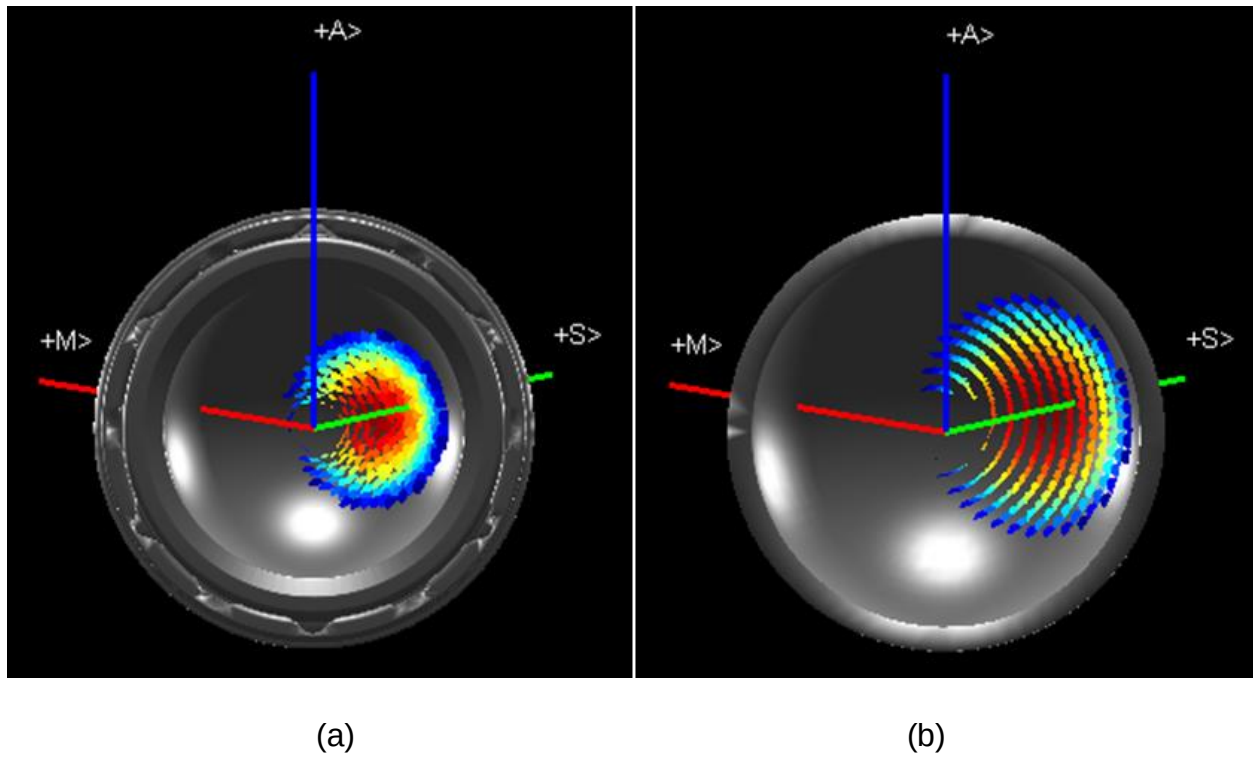


Figure 188: Contact map for (a) neutral liner and (b) dual mobility cup.

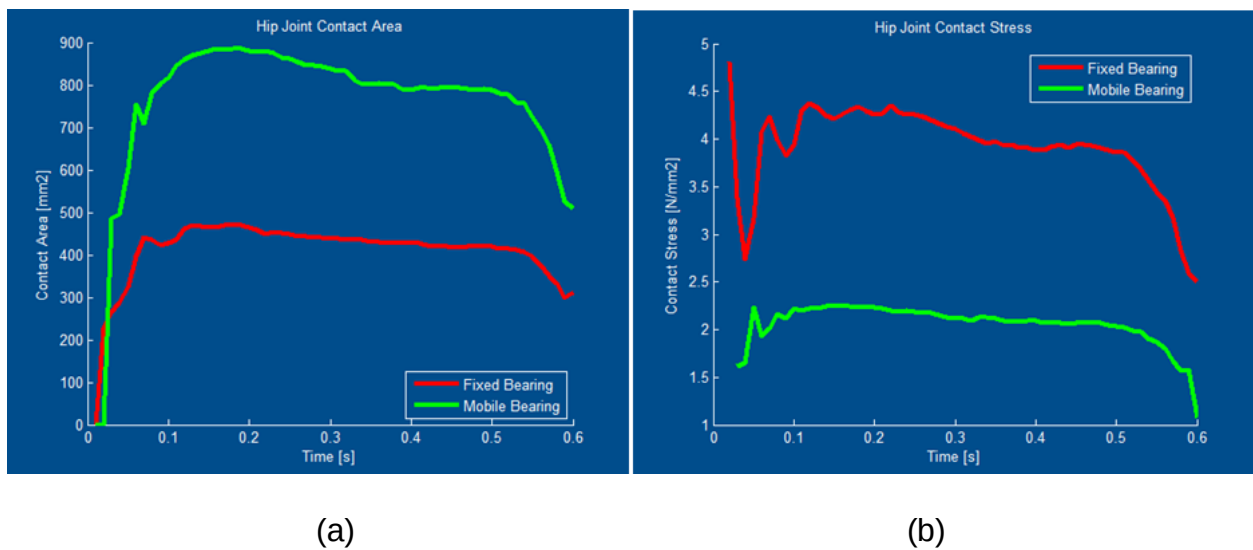


Figure 189: Hip joint contact areas and stresses for dual mobility vs fixed bearing stance phase.

CHAPTER 8: SUMMARY

8.1 Hip Separation

To summarize the entirety of the analyses previously described, it is necessary revisit Figure 60 and Figure 62 to understand how the model mathematically generates hip separation. Throughout this project, hundreds of simulations have been conducted, yielding countless occurrences and patterns of hip separation. During stance phase (consider Figure 60), the muscle forces are much higher, and hence the body's proprioceptive desire to rotate about the anatomical center yields forces that are large enough to slide the femoral head out of the acetabular cup, despite the system being in compression. During stance phase, shifts in the acetabular cup seem to be the primary factor causing hip separation, while shifts in the femoral component do not yield substantial separation but can still generate contact shifting and potential edge loading. Changes in the femoral component may also induce larger muscle forces, which could lead to higher forces and stresses on the liner surface.

Mathematically, the directions of component shifting that generate hip separation correspond to directions that would yield the proprioceptive separation force pulling the femoral head out of the socket: medial shifts, superior shifts, and posterior shifts of the joint center. These shifts would yield separation in the lateral, inferior, and anterior directions, respectively. Conversely, lateral, inferior, and anterior cup shifts attempt to generate hip separation in the medial, superior, and posterior direction, but hip separation does not occur because the liner is blocking the occurrence. However, these scenarios

do generate an increase in hip force because the proprioceptive separation force is essentially pushing the femoral head into the polyethylene liner.

Similarly, cup orientations will either lessen or exacerbate hip separation, depending on the specific direction of hip separation that the model is generating. Thus, it is incorrect to claim that excessive cup anteversion will always increase hip separation. There are scenarios where it may increase separation, if the unconstrained portion of the liner is in the same direction as the resulting separation, but there are also scenarios where it will decrease hip separation. This finding is the main reason why lipped liners implanted medially experience the most separation at 180° of I/E rotation, while lipped liners implanted posteriorly experience the most separation at 90° of I/E rotation.

During swing phase (consider Figure 62), muscle forces are substantially lower and the hip joint is in tension, thus it is predominantly the hip capsule that keeps the femoral head stable within the acetabular cup. Consequently, slack in the hip capsule (due to accidental component misalignment) is the primary cause of hip separation during swing phase. During swing phase, acetabular cup and femoral component errors both yield hip separation, as both types of errors can yield slack in the hip capsule. Certain shifting directions (i.e. cup medialization) will yield slack in the hip capsule and thus hip separation, while other shifting directions (i.e. cup lateralization) will not yield hip separation but will conversely yield greater ligament tension and thus greater hip forces.

On a practical, intraoperative note, separation has been generated through a variety of joint center positions throughout this research project. For example, one scenario pertained to the effects of medial shifting of the acetabular cup, coupled with a further shift of the cup in the superior direction. At this point, one might ask “how would a

surgeon first generate an acetabular medial shift and *then* generate an acetabular superior shift?” In reality, a surgeon does not shift the cup medially and then shift the cup superiorly. Instead, specific combinations of medial, superior, and posterior shifting can be achieved by unknowingly modifying the reaming axis intraoperatively. This occurrence is shown in Figure 190. If a surgeon is intending on reaming along axis A_1 but instead, due to unknown reasoning, reams along a different axis due to a reaming angle error (θ_E), the surgeon would thus induce cup shifting errors in multiple directions.

Overall, it is clear that larger magnitudes of cup shifting errors will generate larger instances of hip separation. Due to an increase in the abductor muscle moment arm [55], it is often the practice of surgeons to intentionally medialize the acetabular cup (and compensate by offsetting the femoral head) in an attempt to reduce muscle forces and consequently the interactive joint forces at the hip. However, care must be taken when intentionally medializing the cup: the safe-zone analysis conducted in Section 7.2 shows that the more the surgeon intends on shifting the acetabular cup, the less room for error the surgeon has before hip separation becomes a legitimate concern.

8.2 Hip Prosthesis Design

Designing a femoral or acetabular component that successfully reduces or eliminates hip separation is an extremely challenging task. In theory, the design must adequately constrain the joint in all implanted orientations, not just the “intended” orientation. This becomes difficult, however, due to the multitude of possible combinations of rotations and translations of both the femoral component and the acetabular cup. Unfortunately, it is an unwise design practice to assume that the surgeon will routinely

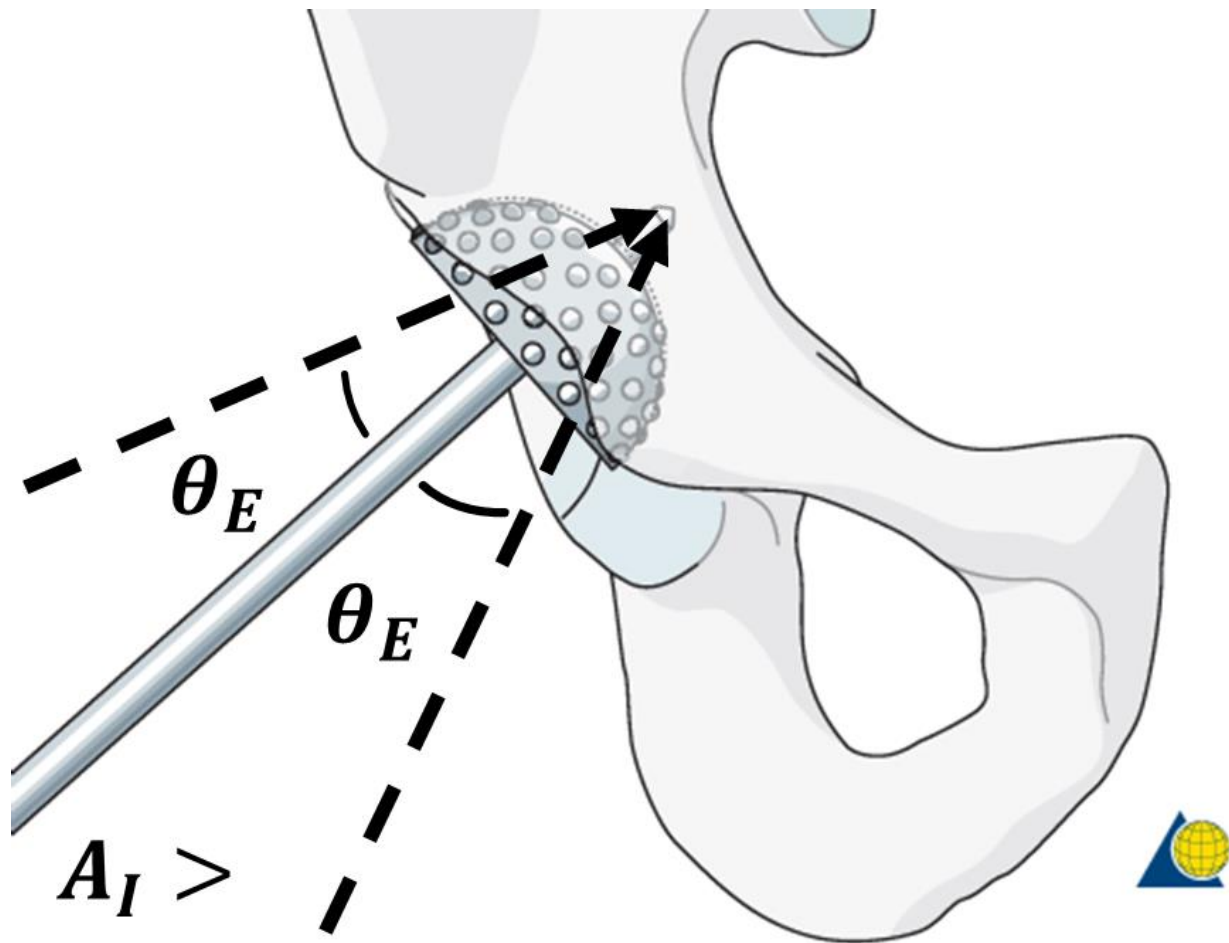


Figure 190: Intended reaming axis with potential reaming angle errors. Image modified from [54].

implant the cup perfectly, as substantial research has been done showing that this is not the case (although surgical navigation may change this) [57] [59] [60] [61] [62]. For the time being, however, it is necessary to assume that the surgeon will *not* orient the component properly, and therefore more specialized component designs each come with their own unique risks.

For example, the lipped designs analyzed in this study have accomplished a reduction in hip separation when they are properly oriented, as evidenced by Figure 171. However, these designs are only effective if they are properly rotated. Mal-rotations of these liners will shift the lip to an ineffective location, which can worsen hip separation, as evidenced by Figure 172. Thus, the lipped design may not be the ideal solution to eliminating hip separation. In addition, depending on the orientation of the liner, the ROM of the hip joint may be reduced (Table 6, Table 7, and Table 8). Thus, it is essential to weigh the benefits and detriments associated with lipped liner designs when it comes to increasing joint stability.

Offsets and lateralization of the stems and liners have often been designed to compensate for capsular laxity and to improve joint stability. The current research shows that, when implanted correctly, lateralized components can reduce hip separation. Section 7.3.4 discusses a multitude of scenarios where lateralized liners substantially reduce hip separation when compared to their non-lateralized counterparts. Unfortunately, lateralized liners can yield undesired increases in joint tension if the acetabulum is not reamed properly. In addition, because the “lateral” shift occurs in a fixed direction with respect to the acetabular cup (refer to Figure 160), mal-rotations of the acetabular cup can result in obscure and unexpected translational shifting directions

of the hip joint. In other words, if a surgeon intends on implanting a lateralized cup system at 40° of inclination and 15° of anteversion, he/she will compensate with the femur offset accordingly. However, if the surgeon unknowingly mal-rotates the cup, the lateralization shift will occur in an unexpected direction and the femur compensation will be incorrect. Femoral offsets are similar, in that improperly compensated offsets can yield undesired joint tension or unexpected/unplanned joint shifting.

Dual-mobility designs appear to yield very promising clinical results regarding reduced dislocation rates. Like large-diameter femoral heads, dual-mobility designs generally have a larger head-to-neck ratio, which results in a larger ROM for the stem before impingement and a larger clearance required for the hip to dislocate. However, most studies do not consider the micro-instability, separation, and edge loading possibilities associated with the design. As evidence by the current research, dual mobility designs that do not include a cylindrical bore may come with an increased risk of edge-loading, and dual-mobility designs that include a cylindrical bore come with a reduction in range-of-motion. On the other hand, dual-mobility systems (and any system with a large head-to-neck ratio) do appear to experience an increase in contact area and a corresponding decrease in contact stress when aligned properly. Thus, it is again important to weigh the pros and cons of dual mobility designs.

A wide array of component designs have been analyzed in the current research, yet none of them have adequately solved the problem of “hip separation.” While many designs appear promising, mal-orientations of the components are often counter-productive. Thus, a surgeon must fully weigh the pros, cons, and risks of each design before selecting an appropriate one for use. While it is beyond the scope of the current

research to design a new prosthesis, it is very evident from the current research that component design is extremely complex, and perhaps the ideal solution to eliminate hip separation truly lies elsewhere.

8.3 Component Positioning Conclusions

Among all simulations presented within this dissertation, one common theme was found to occur: implants that were properly oriented experienced minimal hip separation and desirable overall mechanics. Due to the complexity of implant design and the risk of detrimental postoperative mechanics due to mal-positioning, focus needs to instead be shifted to encourage surgeons to maintain anatomical alignment of the components to the best of their ability. Surgical technique has been shown to be of minimal consequence to joint stability, so the specific surgical technique that allows the surgeon the most accuracy should be considered the better technique. Moreover, as orthopaedic technology continues to advance and it becomes easier for surgeons to accurately implant the components, it is essential that the community embraces these advancements.

In both swing and stance phase, component positioning was the primary cause of hip instability. Whether lax capsules, tight capsules, proprioceptively non-anatomic mechanics, or component impingement, the position of the components has been shown to be the primary factor in joint stability. Specifically, from the current research, there are two primary conclusions that can be drawn: (1) anatomic alignment is essential for postoperative stability, and it should be the goal intraoperatively to recreate the anatomical joint center. Substantial research has been conducted within this dissertation

to show that shifting the joint center yields the potential for joint instability. (2) If it becomes necessary to shift the joint center away from the anatomical center, extra care should be taken. The allowable error becomes much smaller and therefore maximum accuracy becomes essential. Section 7.2.2 clearly demonstrates that as the intended shift magnitude increases, the margin for error dramatically decreases.

It is not the purpose of this dissertation to attempt to develop new alignment procedures or surgical techniques, nor is it the purpose to overturn existing beliefs regarding the benefits of joint shifting. Instead, the purpose of this dissertation is to conduct an in-depth theoretical study assessing all factors of THA implantation and to allow the results to be reviewed by the orthopaedic community, possibly leading to a new train of thought regarding implant position: translational alignment. Thus far, substantial research has been conducted regarding rotational safe zones, but minimal research has been conducted regarding translational safe zones. The findings from this study support the need for a change regarding this concept. Offset shifting serves its purpose, and there is no denying the benefits associated with it, but offset shifting can be counter-productive if it is inaccurate, which is unfortunately quite common [60]. Thus, if a surgeon is intending on shifting the hip joint, caution needs to be taken to ensure that the components are positioned exactly where they are supposed to be.

Most importantly, it is essential that the orthopaedic community adopt technology that allows for the highest level of accuracy regarding component positioning. Mal-positioning of the components is detrimental to the patient's postoperative success and to the lifetime of the implant, and thus it is imperative that we ensure the components are accurately placed exactly where intended.

8.4 Future Work

Continuation of this project has the potential to yield substantial gains for total hip replacements. While the model developed herein is certainly powerful, the assumptions and limitations outlined in Chapter 6 leave room for improvement. Additionally, future work on the GUI will yield a much more user-friendly, viable option for intraoperative use. Finally, further work should be done to analyze more THA components, more alignment scenarios, and non-implanted patients.

First and foremost, the model itself can be improved. Primary future developments revolve around improving the muscle algorithms, improving the contact detection algorithm (if desired), and improving the format for kinematic input. Improving the muscle algorithms will yield more optimized and accurate muscle forces. Improving the contact detection algorithm may or may not be an actual benefit, but this should be considered and weighed in future versions. Improving the kinematic input format will yield more accurate forces, as polynomial fits can inaccurately capture the accelerations of the bones.

Second, the model GUI should be advanced further, with the intention of developing a powerful intraoperative tool that can assist the surgeon and be of use in the operating room to aid in determining ideal component placement. The GUI that has been developed thus far aims for that goal, but it falls short in a number of ways, which are outlined in the limitations section. Improving the overall ease-of-use as well as real-time speed of the program will be beneficial to surgeons intraoperatively. Present day intraoperative software packages only give a surgeon feedback on component placement. The mathematical model discussed in this dissertation could give the surgeon

theoretical feedback regarding postoperative patient mechanics while the actual patient is in the operating room.

Lastly, although a substantial number of simulations were conducted for this research, there will always be more scenarios to analyze. Work should be done to further analyze possible intraoperative scenarios, including non-implanted hips as well as more obscure combinations of transformation errors. This will help represent more extreme complications and will further enhance our knowledge of hip instability, allowing us to further improve the implant designs and surgical techniques that are available today.

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APPENDICES

Appendix A: Cup Alignment during Swing Phase Figures

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Appendix B: Cup Alignment during Stance Phase Figures

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Appendix C: Cup Medialization Figures with Rotation Center Alignment

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Appendix D: Cup Medialization Figures with Liner Center

Alignment

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Appendix E: Anterior/Posterior Shifting Figures with Rotational Center Alignment

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VITA

Michael Thomas LaCour was born in Dallas, Texas on November 3, 1988 and moved to Peachtree City, Georgia when he was 8. Michael grew up in Peachtree City, gaining a love for mathematics and problem solving in high school. Michael began college in 2007 at Rhodes College, studying Physics for three semesters until he ultimately decided that engineering was his ideal field. Michael graduated from Clemson University in 2012 with a degree in Mechanical Engineering and began studying at the University of Tennessee immediately after. Throughout his time as UT, Michael quickly learned how to apply his mechanical engineering skills to the human body, and despite receiving his degree in Biomedical Engineering, Michael truly considers himself a “Mechanical Engineer of the human body.”

Outside of school, Michael is an avid outdoor enthusiast and rock climber. As a rock climber, Michael has learned how to push himself beyond his limits, both physically and mentally, even when the task seems daunting and impossible. Being blessed with a love for the outdoors has provided Michael with a way to escape the stresses of city life, finding peace in the forest, on the rocks, and with his friends.