FINITE ELEMENT ANALYSIS AND VALIDATION OF HIP JOINTS
WITH THE MAIN TYPES OF FEMOROACETABULAR IMPINGEMENT

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Validation of Hip Joints with the
Main Types of Femoroacetabular
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ABSTRACT

Finite Element Analysis and Validation of Hip Joints with the Main Types of Femoroacetabular Impingement

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Current research suggests that femoroacetabular impingement can be a cause of osteoarthritis. Femoroacetabular impingement is a condition that can affect both the femur and the pelvis of an individual. Femoroacetabular impingement can cause damage to the hip joint and its surrounding tissues. The articular cartilage and the labrum are both affected by this condition. A cam impingement is where a bony protrusion develops at the femoral head/neck junction. A pincer impingement is where a bony protrusion develops at the acetabular rim. Often, patients are seen with a combination of both impingements. The main goal of this study was to computationally model and analyze acetabular stresses in a healthy hip, a hip with a cam impingement, a hip with a pincer impingement, and a hip with a combination of the two impingements. The bone models were taken from CT scans. The impingements were created by using Autodesk Maya to modify the surfaces of the models. The hip models were set up to model the single-leg stance

phase of the walking cycle. For the most part, the impingements reduce the stress experienced by the femur. The only exception to this is that the cam femur paired with the pincer pelvis experienced the highest maximum principal stress in the proximomedial region. The pincer impingements increase both the maximum and minimum principal stresses experienced in the acetabulum. Overall, the two types of femoroacetabular impingement change the stress experienced by both the femur and the pelvis. The results of this study demonstrate that acetabular stresses can increase as a result of femoroacetabular impingements. These increased stresses can lead to damage in the hip joint which presents a clinical problem.

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TABLE OF CONTENTS

LIST OF FIGURES.....	ix
1. Introduction.....	1
1.1 Purpose.....	1
1.2 Bone Function and Structure.....	3
1.3 Bone Remodeling.....	8
1.4 Material Composition of Bone.....	10
1.5 Cartilage.....	10
1.6 The Synovial Joint.....	12
1.7 The Labrum.....	13
1.8 Osteoarthritis in the Hip.....	15
1.9 Femoroacetabular Impingement.....	15
1.10 Treatment Options.....	20
1.11 Abaqus.....	22
1.12 Autodesk Maya.....	25
1.13 Current Model.....	26
2. Methods.....	28
2.1 Model Creation of the Healthy Femur.....	28
2.2 Model Creation of the Healthy Hemi-Pelvis..	28
2.3 Model Creation of the Impingements.....	29
2.4 Model Material Properties.....	33
2.5 Load and Boundary Conditions.....	36
2.6 Mesh Convergence.....	38
2.7 Model Implementation.....	42

2.8 Femur Validation.....	42
2.9 Pelvis Validation.....	44
3. Results.....	46
3.1 Femur Validation.....	46
3.2 Femoral Stresses.....	47
3.3 Pelvis Validation.....	77
3.4 Acetabular Stresses.....	79
4. Discussion.....	94
BIBLIOGRAPHY.....	105

LIST OF FIGURES

Figure 1-1 The cycle of bone remodeling as a result of microdamage accumulated during daily activities [11].....	4
Figure 1-2 Structure of cortical bone [1].....	6
Figure 1-3 Cross-section of a femur demonstrating the cortical bone lining and the inner structure comprised of trabecular bone [1].....	7
Figure 1-4 Phases of bone remodeling in trabecular bone [10].....	9
Figure 1-5 Zones found within articular cartilage [1].....	11
Figure 1-6 A normal hip joint as an example of a synovial joint [24].....	13
Figure 1-7 Acetabular labrum [27].....	14
Figure 1-8 An illustration of the different types of femoroacetabular impingement. The top is a healthy hip. The middle shows a pincer impingement. The bottom shows a cam impingement [7].....	17
Figure 1-9 An example of a femur experiencing a cam impingement and how to measure the alpha angle [5].....	18
Figure 1-10 A bony protrusion on the acetabulum that may be associated with a pincer impingement [5].....	19
Figure 1-11 An example of how the CE angle is measured on the acetabulum [7].....	20
Figure 1-12 Ganz surgical procedure used to gain access to the hip joint [29].....	21
Figure 1-13 A process flowchart demonstrating the order of events for an Abaqus simulation [14].....	23
Figure 2-1 A single vertex being manipulated in Autodesk Maya.....	30

Figure 2-2 The vertices on the femoral head/neck junction being manipulated to create the cam impingement.....	31
Figure 2-3 Views of the bony protrusion on the femur.....	32
Figure 2-4 Views of the bony protrusion on the pelvis. This creates the overcoverage on the femoral head and neck.....	32
Figure 2-5 The outer elements were selected to represent the cortical bone. Inside, elements were selected to represent trabecular bone.....	34
Figure 2-6 The elements that were selected to represent the medullary canal.....	35
Figure 2-7 A view of each of the hip models with their boundary conditions and point load. (A) healthy hip (B) cam impingement hip (C) pincer impingement hip (D) combination hip.....	38
Figure 2-8 A graph showing the convergence of the healthy femur.....	39
Figure 2-9 A graph showing the convergence of the femur with a cam impingement.....	40
Figure 2-10 A graph showing the convergence of the healthy pelvis.....	41
Figure 2-11 A graph showing the convergence of the pelvis with a pincer impingement.....	41
Figure 2-12 Deuel's experimental set up [4].....	43
Figure 3-1 Strains on the femur of the healthy hip model compared to Deuel's experimental data.....	46
Figure 3-2 Maximum principal stresses on the femur of the healthy hip model compared to the cam femur paired with the healthy pelvis... ..	48
Figure 3-3 Minimum principal stresses on the femur of the healthy hip model compared to the cam femur paired with the healthy pelvis.....	48

Figure 3-4 Maximum principal stresses on the femur of the healthy hip model compared to the femur paired with the pincer pelvis.....	50
Figure 3-5 Minimum principal stresses on the femur of the healthy hip model compared to the femur paired with the pincer pelvis.....	50
Figure 3-6 Maximum principal stresses on the femur of the healthy hip model compared to the femur with a cam impingement paired with the pelvis with a pincer impingement.....	52
Figure 3-7 Minimum principal stresses on the femur of the healthy hip model compared to the femur with a cam impingement paired with the pelvis with a pincer impingement.....	52
Figure 3-8 Maximum principal stress in the proximomedial region of the healthy femur.....	54
Figure 3-9 Minimum principal stress in the proximomedial region of the healthy femur.....	55
Figure 3-10 Maximum principal stress in the proximolateral region of the healthy femur.....	56
Figure 3-11 Minimum principal stress in the proximolateral region of the healthy femur.....	57
Figure 3-12 Maximum principal stress in the neck region of the healthy femur.....	58
Figure 3-13 Minimum principal stress in the neck region of the healthy femur.....	58
Figure 3-14 Maximum principal stress in the proximomedial region of the cam femur.....	60
Figure 3-15 Minimum principal stress in the proximomedial region of the cam femur.....	61
Figure 3-16 Maximum principal stress in theproximolateral region of the cam femur.....	62
Figure 3-17 Minimum principal stress in the proximolateral region of the cam femur.....	63
Figure 3-18 Maximum principal stress in the neck region of the cam femur.....	64

Figure 3-19 Minimum principal stress in the neck region of the cam femur.....	64
Figure 3-20 Maximum principal stress in the proximomedial region of the cam femur that was paired with the pincer pelvis.....	66
Figure 3-21 Minimum principal stress in the proximomedial region of the cam femur that was paired with the pincer pelvis.....	67
Figure 3-22 Maximum principal stress in the proximolateral region of the cam femur that was paired with the pincer pelvis.....	68
Figure 3-23 Minimum principal stress in the proximolateral region of the cam femur that was paired with the pincer pelvis.....	69
Figure 3-24 Maximum principal stress in the neck region of the cam femur that was paired with the pincer pelvis.....	70
Figure 3-25 Minimum principal stress in the neck region of the cam femur that was paired with the pincer pelvis.....	70
Figure 3-26 Maximum principal stress in the proximomedial region of the healthy femur that was paired with the pincer pelvis.....	72
Figure 3-27 Minimum principal stress in the proximomedial region of the healthy femur that was paired with the pincer pelvis.....	73
Figure 3-28 Maximum principal stress in the proximolateral region of the healthy femur that was paired with the pincer pelvis.....	74
Figure 3-29 Minimum principal stress in the proximolateral region of the healthy femur that was paired with the pincer pelvis.....	75
Figure 3-30 Maximum principal stress in the neck region of the healthy femur that was paired with the pincer pelvis.....	76
Figure 3-31 Minimum principal stress in the neck region of the healthy femur that was paired with the pincer pelvis.....	76

Figure 3-32 Minimum principal stresses in the acetabulum of the healthy hip model compared to experimental data.....	78
Figure 3-33 Maximum principal stresses in the acetabulum of the healthy hip model compared to the pelvis paired with the cam femur.....	79
Figure 3-34 Minimum principal stresses in the acetabulum of the healthy hip model compared to the pelvis paired with the cam femur.....	80
Figure 3-35 Maximum principal stresses in the acetabulum of the healthy hip model compared to the pincer pelvis that was paired with a healthy femur.....	81
Figure 3-36 Minimum principal stresses in the acetabulum of the healthy hip model compared to the pincer pelvis that was paired with a healthy femur.....	81
Figure 3-37 Maximum principal stresses in the acetabulum of the healthy hip model compared to the pincer pelvis paired with the cam femur.....	82
Figure 3-38 Minimum principal stresses in the acetabulum of the healthy hip model compared to the pincer pelvis paired with the cam femur.....	83
Figure 3-39 Maximum principal stress in the acetabulum of the healthy pelvis that was paired with the healthy femur.....	84
Figure 3-40 Minimum principal stress in the acetabulum of the healthy pelvis that was paired with the healthy femur.....	85
Figure 3-41 Maximum principal stress in the acetabulum of the healthy pelvis that was paired with the cam femur.....	86
Figure 3-42 Minimum principal stress in the acetabulum of the healthy pelvis that was paired with the cam femur.....	87
Figure 3-43 Maximum principal stress in the acetabulum of the pincer pelvis that was paired with the cam femur.....	89

Figure 3-44 Minimum principal stress in the acetabulum of the pincer pelvis that was paired with the cam femur.....	90
Figure 3-45 Maximum principal stress in the acetabulum of the pincer pelvis that was paired with the healthy femur.....	91
Figure 3-46 Minimum principal stress in the acetabulum of the pincer pelvis that was paired with the healthy femur.....	92
Figure 4-1 A comparison of the femoral maximum principal stresses in all of the models.....	95
Figure 4-2 A comparison of all the femoral minimum principal stresses in all of the models.....	96
Figure 4-3 A comparison of the acetabular maximum principal stresses in all of the models.....	97
Figure 4-4 A comparison of the acetabular minimum principal stresses in all of the models.....	98

1. Introduction

1. 1 Purpose

Femoracetabular impingement is a cause of hip pain in all age groups and has been proposed as a link to the early development of osteoarthritis [5]. According to Arbabi et al, it is now recognized as the leading pathomechanism that causes "primary" osteoarthritis [13]. Femoracetabular impingement describes abnormal contact, or impingement, between the head of the femur and the acetabulum. The abnormalities are typically found at the junction between the anterior/superior femoral neck and head and the rim of the acetabulum. A cam impingement is associated with the abnormal junction between the anterior/superior femoral neck and head while a pincer impingement is associated with an abnormal acetabular rim [5].

Finite element analysis is a powerful computational tool used to analysis stresses and strains on two and three dimensional models. An organic shape such as that of a femur can be made into a solid part using CT scans. The solid part can be imported into Abaqus where stress/strain analysis can

be conducted. While there are many examples of finite element models consisting of the femur throughout the literature, there are few that represent the entire hip joint with both the femur and the hemi-pelvis. Furthermore, there are very few, if any, simulations that attempt to model the femoroacetabular impingement.

The goal of this study is to observe the acetabular stresses that develop due to the contact between the femoral head and the acetabulum. The main interest is comparing a healthy hip to hips experiencing femoroacetabular impingement using finite element analysis. Three different types of femoroacetabular impingement will be looked at: a pure cam impingement, a pure pincer impingement, and a combination of the cam and pincer impingement. The main objectives are: (1) create femoroacetabular impingement models using existing healthy bone models, (2) develop anatomically accurate hip models using the bones experiencing femoroacetabular impingement, (3) validate the models using experimental data found in literature, and (4) compare the stresses found in the acetabula of the models.

1.2 Bone Function and Structure

Bone is a living tissue that is often referred to as a dynamic tissue because it is constantly adapting to its mechanical environment as well as experiencing damage and responding to repair that damage [1].

Figure 1-1 shows how daily activities cause bones to constantly be in loading. As a result, microdamage accumulates, causing the activation or apoptosis of osteocytes and the formation of BMUs to remodel bone.

Figure 1-1 The cycle of bone remodeling as a result of microdamage accumulated during daily activities [11].

There are four types of cells that are responsible for the maintenance of bone: osteoblasts, osteoclasts, osteocytes, and bone lining cells. Osteoblasts are cells that form new bone. They secrete osteoid which contains collagen, noncollagenous proteins, proteoglycans, and water [2]. Osteoclasts are recruited to resorb the old bone and are capable of

accomplishing this at a rate of tens of microns per day [1,2]. They resorb bone by secreting acids and enzymes. Osteocytes are osteoblasts that become buried in the bone matrix as the bone is being formed. Lining cells line the surface of the bone tissue.

There are two main types of bone tissue. Cortical bone, also known as compact bone, is very dense with a porosity of 5%-10% and a modulus of 17 GPa [1]. The structure can be seen in figure 1-2. It can be found in shafts of long bones and forms the outer layer of other bones [2]. Cortical bone exists in layers known as lamellae that are about 5 μm thick each. In a layer, the fibers of collagen are organized in parallel with one another. Different layers contain fibers organized in different directions. Osteons are cylindrical structures made up of concentric lamellae

and surround the Haversian canal.

Figure 1-2 Structure of cortical bone [1].

Bone contains various canals and channels to deliver nutrients to the tissue. Haversian canals contain blood vessels and nerves while Volkmann's canals run perpendicular to the Haversian canals and

connect one another. They contain capillaries and nerves and are roughly 50 μm in diameter [2].

Figure 1-3 Cross-section of a femur demonstrating the cortical bone lining and the inner structure comprised of trabecular bone [1].

Trabecular bone, also known as cancellous or spongy bone, is very porous with a porosity of 75%-90%. Trabecular bone can be found in the ends of long bones such as the femur. Trabecular bone is organized

by a network of struts called trabeculae [1] as seen in figure 1-3. The struts are connected to one another and form the porous appearance of trabecular bone. These trabeculae are about 200 μm thick. These pores are filled with bone marrow and cells [28]. Trabecular bone is much weaker than cortical bone, however, it is lighter due to it being a less dense and more porous tissue.

1.3 Bone Remodeling

Basic multicellular units, or BMUs, are groups of cells functioning as organized units that are responsible for replacing old bone with new bone. BMUs are responsible for a remodeling sequence known as the A-R-F (Activation, Resorption, Formation) sequence which can be seen in figure 1-4.

Figure 1-4 Phases of bone remodeling in trabecular bone [10].

First, osteoclasts are activated in the BMUs. Osteoclasts resorb bone at a rate of about 40 $\mu\text{m}/\text{day}$ [2]. Osteoblasts line the recently resorbed cavities and begin to refill these cavities with new bone. The osteoblasts leave space for the Haversian canal in cortical bone. Mineral is deposited in the unmineralized bone matrix. Once these processes are complete, the osteoclasts disappear and the osteoblasts can become osteocytes, lining cells, or they can disappear.

1.4 Material Composition of Bone

Bone is made up of water, collagen, hydroxyapatite mineral, proteoglycans, and noncollagenous proteins. Collagen is a structural protein that exists in many different types. Type I collagen is the primary collagen found in bones and provides flexibility and tensile strength. The mineral in bone mostly consists of hydroxyapatite crystals. Bone contains proteoglycans such as decorin, biglycan, and osteocalcin [2].

1.5 Cartilage

Cartilage consists of collagen and proteoglycans which make up about 30% of the tissue. The other 70% is almost all water. The proteoglycans are hydrophilic, meaning they are attracted to water. Because of this, the cartilage tissue consists of large amounts of water that interacts with the proteoglycans. Cartilage is a viscoelastic material meaning that its elastic modulus varies based on the rate at which the tissue is loaded. At lower rates of loading, the modulus can measure at only a few MPa. At higher rates of loading that match physiologic conditions, the modulus can be as high as 500 MPa [2].

Hyaline cartilage is the most common type of cartilage in the body and is cartilage found in joints. Hyaline cartilage in joints is called articular cartilage. Articular cartilage contains type II collagen, proteoglycans, and other non-collagenous proteins. Articular cartilage typically does not have nerves or blood vessels. There are few cells in articular cartilage called chondrocytes. Chondrocytes make up roughly 5-10% of the total volume of the tissue [1].

Figure 1-5 Zones found within articular cartilage [1].

There are four identifiable zones in articular cartilage as seen in figure 1-5. The superficial zone makes up roughly 10-20% of the thickness of the

cartilage. This region is capable of retaining fluid within the collagen matrix and assists with the support of the compressive loads. The middle zone is roughly 40-60% of the thickness and contains more randomly oriented fibrils, less water, and a greater number of proteoglycans. The deep zone is roughly 20-30% of the thickness and contains fibers that are oriented perpendicularly to the surface of the bone and has the highest proteoglycan content of all the tissue. The calcified zone is the final layer where collagen fibers are mineralized and provide stability between the cartilage and the bone [1,2].

1.6 The Synovial Joint

Synovial joints are those that contain synovial fluid and are able to move freely (figure 1-6). The joint consists of two bones held together by the joint capsule. The surfaces of the bones are covered by articular cartilage. Articular cartilage serves as a lubricated surface for the bones and helps to prevent wear. Synovial fluid lubricates these surfaces and reduces the friction between cartilage surfaces [1].

Figure 1-6 A normal hip joint as an example of a synovial joint [24].

These lubricated surfaces are essentially frictionless, with friction coefficients as low as 0.001-0.005. It is important to note that cartilage does not serve as an absorber for the surrounding forces because of how thin the tissue is [2].

1.7 The Labrum

The labrum (figure 1-7) is a cartilagenous layer, composed mostly of Type I collagen, that lines the acetabulum of the hip [26]. The labrum increases both the surface area and the volume of the acetabulum [25].

Figure 1-7 Acetabular labrum [27].

The function of the labrum has been debated and is still subject to investigation. The labrum does not appear to sustain any substantial loading because its removal does not significantly alter the stress between the femoral head and the acetabulum [25].

1.8 Osteoarthritis in the Hip

The hip joint can be susceptible to osteoarthritis, or wear and tear arthritis. In the United States, roughly 3% of adults over the age of 30 have osteoarthritis [3]. People that are at risk for osteoarthritis include those that have a family history of the disease or those that are elderly, obese, or have had a hip injury. Non-surgical treatments for osteoarthritis of the hip include rest, physical therapy, nonsteroidal anti-inflammatory medications, and weight loss. In later stages of osteoarthritis, surgical treatments may be necessary. In advanced stages, patients may need a total hip replacement. Recently, femoroacetabular impingement has been suggested as being responsible for most cases of idiopathic hip osteoarthritis [9].

1.9 Femoroacetabular Impingement

A cam impingement and a pincer impingement are two types of femoroacetabular impingements [6]. While the two can be separate and with unique conditions, a combination of the two is usually the case for

patients. In a study conducted by Beck et al, 86% of the hips that were analyzed featured a combination of both impingement types [23]. A cam impingement is associated with the femur and appears as a bony protrusion on the anterior/superior femoral head-neck junction. This raised surface impinges on the acetabular labrum and articular cartilage. Typically, the anterior/superior head-neck junction has a convex shape [5]. In the case of the cam impingement, the shape becomes flat or concave, as seen in figure 1-8. Repeated contact between the raised surface of the femur and the acetabulum can result in tearing or detachment of the acetabular labrum [5]. Figure 1-8 shows how the impingements impede the movement of the hip.

Figure 1-8 An illustration of the different types of femoroacetabular impingement. The top is a healthy hip. The middle shows a pincer impingement. The bottom shows a cam impingement [7].

The alpha angle is a simple measurement of the sphericity of the femoral head and can be seen in figure 1-9. In a study conducted by Nötzle et al, patients with a cam impingement had at least an alpha

angle of 55° while patients without a cam impingement had an alpha angle less than 48° [33].

Figure 1-9 An example of a femur experiencing a cam impingement and how to measure the alpha angle [5].

This bony protrusion raises and contacts the acetabulum during daily activities, causing a delaminating effect on the articular cartilage. Additionally, the raising of the cam impingement causes damage to the labrum.

A pincer impingement is associated with the acetabulum. This condition can be due to overcoverage of the acetabulum on the femoral head and neck (figure 1-10). With the pincer impingement, the labrum becomes damaged because of overcoverage of the acetabulum contacting the femoral head/neck region.

Figure 1-10 A bony protrusion on the acetabulum that may be associated with a pincer impingement [5].

The femoral neck makes contact with the rim of the acetabulum as a result of this overcoverage. The constant contact damages the labrum and results in degeneration of the labrum and ossification of the acetabular rim. As a result, the apparent depth of the acetabulum can deepen and the overcoverage of the acetabulum on the femur can become worse.

The center edge angle can be used to quantify acetabular impingement. The center edge angle is the angle formed by a vertical line and a line connecting the center of the femoral head with the lateral edge of the acetabulum as seen in figure 1-11 [7]. Center edge angles in healthy individuals vary between 25° and 39° . A center edge angle greater than 40° has been proposed as evidence for a pincer impingement.

Figure 1-11 An example of how the CE angle is measured on the acetabulum [7].

1.10 Treatment Options

The Ganz surgical dislocation method (figure 1-12) is currently one of the best surgical methods for exposing the hip joint to allow for treatment [6, 29]. While invasive, the dislocation method allows for

osteochoondroplasty, the removal of excess bony abnormalities, which can be used to treat cam impingements and bony acetabular rim trimming to treat pincer impingements. Success rates range from 68-80% for open surgery treatments [30].

Figure 1-12 Ganz surgical procedure used to gain access to the hip joint [29].

Arthroscopy has been proposed as a non-invasive technique to treat both cam and pincer impingements. However, arthroscopy provides limited visualization of the areas of interest which makes it difficult to determine how much bone is being removed from the surgical site. Osteochondroplasty of the cam lesions and rim trimming of the acetabulum can both be performed using arthroscopy. Because of the difficulty associated with acetabular trimming, treatment of pincer lesions using arthroscopy is less common than treatment of cam lesions. Success ranges from 90-100% for arthroscopic treatment [30].

1.11 Abaqus

Abaqus is powerful software used for finite element analysis. It contains a library of elements that enable the user to model virtually any geometry. This is especially useful when attempting to model an organic shape such as bone. Additionally, material properties can be assigned and selected to properly model bone.

Figure 1-13 A process flowchart demonstrating the order of events for an Abaqus simulation [14].

Creating a model in Abaqus can be divided into three distinct stages (figure 1-13). Preprocessing involves defining the characteristics of the model. This can be done graphically through Abaqus. An input file is made that is a text representation of the created model. The input file is run as a job and the simulation takes place. Abaqus solves the problem numerically and stores the results to be read in the post processing stage. The output of a simulation can include, but is not limited to, stresses, strains, forces, and displacements. The simulation creates output files that can be viewed in the post processing. Post processing is used to evaluate the results once the simulation has completed. The results

can be displayed in a variety of ways such as deformed shape plots and X-Y plots.

The structure of what is being modeled can be imported (from software such as SolidWorks) or created in Abaqus. Elements and nodes make up the structure of what is being modeled. A model can be made up of many different elements, each representing a discrete portion of the structure. These elements are connected by nodes. A mesh is the collection of all the elements and nodes that make up the model. The number of elements within the model can vary greatly which means that the mesh density can vary. A denser mesh will provide a more accurate solution, but it will also require more computational time. There are different element types, shapes, and locations that also affect the results of a simulation.

Material properties must be defined for the physical structure being modeled. Loads must be given to distort the structure. The loads that are commonly used are point loads, pressure loads, body forces, and thermal loads. Boundary conditions are used to constrain the model from moving in specific

directions. Portions of a model can be pinned and will neither translate nor rotate in any direction [14].

1.12 Autodesk Maya

Autodesk Maya, usually referred to as Maya, is an interactive 3D computer graphics software that can be used to create models for video games, animated films, or visual effects. While Maya is typically used to create 3D models, it can also serve as a useful tool for manipulating existing surfaces. The file type .iges, a common engineering software part file, can be imported into Maya. This is useful if the part being modified is organic in shape (for example, a femur).

Polygons are the most basic geometry type and are applied directly to the surface of a part. Polygonal meshes are made up of faces. These faces are lined with edges. The edges of different faces meet at a point called the control vertex. The control vertex is the point on the polygonal mesh that can be manipulated to change the surface of the mesh. Clusters can exist around a vertex of a model and, together, can be manipulated to change the surface of a model [12].

1.13 Current Model

A hip model consisting of a femur and a hemi-pelvis was created and developed using SolidWorks and Abaqus. Computational modeling serves as a non-invasive alternative to analyzing the stresses and strains in the human hip. Few finite element models have been developed to model the femoroacetabular impingement. Experimental data from multiple sources was used to validate this model. First, an experiment involving strains in the femur [4] was used to validate the femur model. Second, experimental data involving the stress distribution inside of the acetabulum [31] was used to validate the hemi-pelvis. Combined, the two sets of data were used to validate the healthy hip model.

Autodesk Maya was used to modify the surface geometry of the bones models to create the bony protrusions necessary for modeling the femoral acetabular impingement. Autodesk Maya was chosen because it provides a simple interface that can easily modify the geometry of a .iges file while maintaining the integrity of the solid part. Because organic shapes are being used, traditional engineering

software, such as SolidWorks, would be difficult to use to modify the shapes of the bones. A cam impingement was added to the femur and a pincer impingement was added to the hemi-pelvis. Three additional Abaqus models were created: a pure cam impingement model, a pure pincer impingement model, and a model that combines the two impingements. All models were compared to the previously validated healthy model.

The advantage to this method is that the models are geometrically identical with the exception of the bony protrusions used to model the impingement conditions. Therefore, any differences in stress and strain can be attributed to the impingements. It would be more difficult to draw conclusions if bone models were used from two different patients. This is because the stress distributions within the acetabula will be different from patient to patient due to the differences in the shape of the bones [8]. By eliminating that variability, this study demonstrates the direct effects of the two different types of femoroacetabular impingement.

2. Methods

2.1 Model Creation of the Healthy Femur

The femur model was taken from the Biomed Town website [16]. In SolidWorks, the femur was scaled to a length of 528 mm and an osteotomy was performed to remove the condyles. The modified femur was imported into Abaqus as an .iges file at a scale of 1:1. It then went through a series of translations and rotations to position the femoral head against the surface of the acetabulum. For the healthy femur model, a seed size of 8 was used which resulted in 27441 elements and 41480 nodes. The part was meshed using quadratic tetrahedral elements (C3D10). Quadratic tetrahedral elements have been used in models and yielded results that correlated well with experimental data [18].

2.2 Model Creation of the Healthy Hemi-Pelvis

The pelvis model was obtained from the 3D Content Central website [22]. In SolidWorks, the right hemi-pelvis was removed, leaving the left hemi-pelvis. The hemi-pelvis was imported into Abaqus as an .iges file at a scale of 1:1. For the healthy hemi-pelvis model,

a seed size of 4 was used which resulted in 70,054 elements and 108,100 nodes. To create the hip assembly, the hemi-pelvis was left alone while the femur underwent a series of translations and rotations in order for the head of the femur to be positioned appropriately inside of the acetabulum of the hemi-pelvis.

2.3 Model Creation of the Impingements

Both the femur and hemi-pelvis models were imported into Autodesk Maya as .iges files. To manipulate the surface of the part, Vertex mode is used to select an individual vertex or a group of vertices. In figure 2-1, a single vertex, indicated by the arrow, has been selected and raised to demonstrate how vertices are manipulated.

Figure 2-1 A single vertex being manipulated in Autodesk Maya.

In figure 2-2, a group of vertices is selected between the femoral head/neck junction. By manipulating a group of vertices and fine tuning by selecting individual vertices at a time, the desired shape was constructed to represent a cam impingement on the femur. The same was done on the pelvis model to create the pincer impingement.

Figure 2-2 The vertices on the femoral head/neck junction being manipulated to create the cam impingement.

The models were imported into Abaqus to create the impingement models. For the cam femur, a seed size of 7 was used which resulted in 30,923 elements and 46,478 nodes. For the pincer pelvis, a seed size of 6 was used which resulted in 37,176 elements and 59,436 nodes. The healthy hip model was used as a guide to ensure that the bones with the impingement features were aligned the same way as the healthy bones.

Figure 2-3 Views of the bony protrusion on the femur.

Figure 2-3 shows the bony protrusion that was added to the femur to create the cam impingement.

Figure 2-4 Views of the bony protrusion on the pelvis. This creates the overcoverage on the femoral head and neck.

Figure 2-4 shows the bony protrusion that was added to the pelvis to create the pincer impingement.

2.4 Model Material Properties

The healthy femur, healthy hemi-pelvis, cam femur, and pincer hemi-pelvis were all given cortical and trabecular bone properties (figure 2-5). The healthy femur and the cam femur were both given medullary canals. To simulate cortical bone, the surface elements on all of the bone models were selected by creating an element set and selecting all elements at a 20 degree angle. By creating cortical element sets with this method, cortical bone is modeled with a cortical thickness of approximately 4 mm.

Figure 2-5 The outer elements were selected to represent the cortical bone. Inside, elements were selected to represent trabecular bone.

This cortical elements set was hidden using the display group tool. For the two hemi-pelvis models, the remaining elements were added to a new set to represent the trabecular bone (figure 2-5).

Figure 2-6 The elements that were selected to represent the medullary canal.

For the two femur models, the elements distal to the lesser trochanter were added to a new set to represent the medullary canal (figure 2-6). The remaining bone was then added to a set to represent the trabecular bone. To assign a section to each of these sets, the mesh had to first be deleted. Once deleted, the sets could be selected and assigned to their respective sections and the part could then be re-meshed.

The labrum was not considered in the model based on research suggesting that it does not appear to support any major loads across the tissue as a result of contact within the joint [25]. Cartilage was not included in the CT scans and was therefore not included in the model. Cartilage does not serve as a shock absorber and should have no overall effect on the maximum stresses experienced in the acetabulum [2]. However, cartilage is expected to change the stress distribution within the acetabulum. The joint was modeled as a frictionless surface which will be described further in the next section.

2.5 Load and Boundary Conditions

For the two femur models, the boundary condition used was a pinned constraint at the distal end [20]. For the hemi-pelvis models (figure 2-7), the boundary conditions were pinned constraints at the sacroiliac joint and the pubic symphysis [21].

Contact was established in the model by creating two surfaces at the head of the femur and the acetabulum. For the head of the femur, elements were selected on the surface to represent where contact would be. Similarly, elements were selected within the

acetabulum to represent where contact would be. A frictionless surface was defined to represent the articulation of the two surfaces.

In a previous model by Vyas et al, a point load was applied at the sacroiliac joint [19]. However, because the sacroiliac joint was constrained in the current model, a point load was applied at a node on the ischium (figure 2-7). The load was set to 700 N to represent the weight force of a person that is approximately 70 kg in the negative z-direction to represent the force due to the weight of the body.

Figure 2-7 A view of each of the hip models with their boundary conditions and point load. (A) healthy hip (B) cam impingement hip (C) pincer impingement hip (D) combination hip.

2.6 Mesh Convergence

Convergence tests were performed for each of the four bone models (figure 2-8 - 2-11). A total of five

different mesh densities were used to test for convergence.

Figure 2-8 A graph showing the convergence of the healthy femur.

For the femur models, a 700 N load in the negative z-direction was placed on the medial side of the femoral neck. The strain was measured on the lateral side of the femoral neck.

Figure 2-9 A graph showing the convergence of the femur with a cam impingement.

The modified femur with the cam impingement was loaded similarly and convergence was achieved (figure 2-9). The strain was measured on the lateral side of the femoral neck.

Figure 2-10 A graph showing the convergence of the healthy pelvis.

For the hemi-pelvis models, a 700 N load in the negative z-direction was placed on the ischium. The strain was measured in the posterior/superior region.

Figure 2-11 A graph showing the convergence of the pelvis with a pincer impingement.

The modified pelvis with the pincer impingement was loaded similarly and convergence was achieved (figure 2-11). The strain was measured in the posterior/superior region.

2.7 Model Implementation

Abaqus models were run on two custom built workstations each with Intel Core i7-950 3.06 GHz quad-core processors, ASUS P6x58D premium motherboards, and 12 GB RAM.

Maya models were created on a custom built workstation with Intel Core i7-930 2.80 Ghz quad-core processor with 6 GB RAM.

2.8 Femur Validation

The femoral components of the hip models were validated by comparing strain data from the simulations to experimental data obtained by Deuel et al (figure 2-12) [4]. In the study by Deuel et al, femora were screened for abnormalities. Femoral strains were measured using one three-element, 45° stacked rosette strain gauge (model 060WR-350, Vishay Micro-Measurements, Raleigh, North Carolina) and three

uniaxial strain gauges (model 125UN-350, Vishay Micro-Measurements, Raleigh, North Carolina), which were attached to the surface of the femur. The strain gauges were used to measure strain on the proximolateral, proximomedial, distolateral, and distomedial surfaces.

Figure 2-12 Deuel's experimental set up [4].

To simulate loading conditions experienced during the single-leg stance phase, a custom designed fixture was used. The results of interest were the proximolateral and proximomedial strains of the femoral neck. For the proximolateral region, the average maximum principal strain was 637 $\mu\epsilon$. For the proximomedial region, the average minimum principal strain was -621 $\mu\epsilon$. For validation, the hip model was used in attempt to replicate the loading conditions in Deuel's experiment. As seen in figure 2-12, at (a), a 600 N load was applied to the test fixture. Therefore, a 600 N point load was applied at the ischium to represent the 600 N compressive load applied in Deuel's fixture. This was the only instance where a 600 N load was used.

2.9 Pelvis Validation

The acetabula of the models were validated by comparing the stresses from the simulations to experimental data obtained by Afoke et al [31]. To measure the stresses within the joint, pressure sensitive film (Fujifilm Prescale [Low Grade]) was used. The pressure sensitive film was placed in between the acetabulum and the femoral head to measure

the contact stress. The pressure range that the film could record was 1 to 10 MPa. The film was calibrated using a known force to produce a range of pressures. The experiment measured stress during three different phases of the walking cycle. The phase of interest is the single-leg stance phase. The Abaqus models attempted to replicate the single leg-stance phase of the walking cycle. In this study, the hip joint of an approximately 70 kg patient was studied and the maximum stress recorded was 8.6 MPa. Therefore, a 700 N load for the body weight was used to represent the approximately 70 kg patient in the experimental study.

3. Results

3.1 Femur Validation

The femur of the hip model was validated by comparing the strain results in the simulation to experimental data of a cadaveric femur (figure 3-1). The hip models were set up to reproduce loading conditions during the single-leg stance phase of walking [4].

Figure 3-1 Strains on the femur of the healthy hip model compared to Deuel's experimental data.

For the proximalmedial and proximalateral sites on the femur, the mean principal strains were $-621 \mu\epsilon$ and

637 $\mu\epsilon$, respectively. For the healthy hip model, the resulting femoral strains were -638 $\mu\epsilon$ and 606 $\mu\epsilon$ in the proximomedial and proximolateral regions, respectively. Because this hip model did not have any noticeable abnormalities, it was compared to Deuel's experimental data to validate the hip set up.

3.2 Femoral Stresses

The values generated in the Abaqus hip model compare similarly to those found in the experimental data. This helps to validate the femur portion of the hip model. The proximomedial and proximolateral strains in the impinged models were then compared to those found in the healthy hip model.

Figure 3-2 Maximum principal stresses on the femur of the healthy hip model compared to the cam femur paired with the healthy pelvis.

Figure 3-3 Minimum principal stresses on the femur of the healthy hip model compared to the cam femur paired with the healthy pelvis.

Figures 3-2 and 3-3 show the maximum and minimum principal stresses, respectively. These measurements were taken in the proximomedial and proximolateral regions of the femur. The comparison is between the femur in the healthy hip model and the cam femur. For the healthy femur, the maximum principal stresses were 1.76 MPa in the proximomedial region and 10.81 MPa in the proximolateral region. For the cam femur, the maximum principal stresses were 0.98 MPa in the proximomedial region and 10.16 MPa in the proximolateral region. For the healthy femur, the minimum principal stresses were -30.10 MPa in the proximomedial region and -1.02 MPa in the proximolateral region. For the cam femur, the minimum principal stresses were -21.48 MPa in the proximomedial region and -0.38 MPa in the proximolateral region.

Figure 3-4 Maximum principal stresses on the femur of the healthy hip model compared to the femur paired with the pincer pelvis.

Figure 3-5 Minimum principal stresses on the femur of the healthy hip model compared to the femur paired with the pincer pelvis.

Figures 3-4 and 3-5 show the maximum and minimum principal stresses, respectively. These measurements were taken in the proximomedial and proximolateral regions of the femur. The comparison is between the femur in the healthy hip model and the healthy femur paired with the pincer pelvis. For the healthy femur with the pincer pelvis, the maximum principal stresses were 3.22 MPa in the proximomedial region and 8.99 MPa in the proximolateral region and the minimum principal stresses were -21.05 MPa in the proximomedial region and -0.68 MPa in the proximolateral region.

Figure 3-6 Maximum principal stresses on the femur of the healthy hip model compared to the femur with a cam impingement paired with the pelvis with a pincer impingement.

Figure 3-7 Minimum principal stresses on the femur of the healthy hip model compared to the femur with a cam impingement paired with the pelvis with a pincer impingement.

Figures 3-6 and 3-7 show the maximum and minimum principal stresses, respectively. These measurements were taken in the proximomedial and proximolateral regions of the femur. The comparison is between the femur in the healthy hip model and the cam femur paired with the pincer pelvis. For the cam femur with the pincer pelvis, the maximum principal stresses were 1.00 MPa in the proximomedial region and 8.96 MPa in the proximolateral region and the minimum principal stresses were -25.42 MPa in the proximomedial region and -0.29 MPa in the proximolateral region.

Figure 3-8 Maximum principal stress in the proximomedial region of the healthy femur.

Figure 3-9 Minimum principal stress in the proximomedial region of the healthy femur.

Figure 3-10 Maximum principal stress in the proximalateral region of the healthy femur.

Figure 3-11 Minimum principal stress in the proximolateral region of the healthy femur.

Figure 3-12 Maximum principal stress in the neck region of the healthy femur.

Figure 3-13 Minimum principal stress in the neck region of the healthy femur.

Figures 3-8 and 3-9 show contour plots for the proximomedial region below the lesser trochanter on the healthy femur that was paired with the healthy pelvis. Figure 3-8 shows maximum principal stresses while figure 3-9 shows minimum principal stresses. For the maximum principal stresses, the stress in the

region appears to be fairly uniform. However, for minimum principal stresses, the area just below the lesser trochanter appears to have the highest minimum principal stresses. This is expected because this area of the bone is in compression. Figures 3-10 and 3-11 show contour plots for the proximolateral region below the greater trochanter. Figure 3-10 shows maximum principal stresses while figure 3-11 shows minimum principal stresses. Figure 3-10 shows the area of highest maximum principal stresses where the bone is in tension. Figure 3-11 shows lower minimum principal stresses as a result of less compression. Figures 3-12 and 3-13 provide a top view of the femoral neck. Figure 3-12 shows maximum principal stresses while figure 3-13 shows minimum principal stresses. This is another area of high maximum principal stress as a result of the femoral neck being in tension.

Figure 2-14 Maximum principal stress in the proximomedial region of the cam femur.

Figure 3-15 Minimum principal stress in the proximomedial region of the cam femur.

Figure 3-16 Maximum principal stress in the proximolateral region of the cam femur.

Figure 3-17 Minimum principal stress in the proximolateral region of the cam femur.

Figure 3-18 Maximum principal stress in the neck region of the cam femur.

Figure 3-19 Minimum principal stress in the neck region of the cam femur.

Figures 3-14 and 3-15 show contour plots for the proximomedial region below the lesser trochanter on the cam femur that was paired with the healthy pelvis. Figure 3-14 shows maximum principal stresses while figure 3-15 shows minimum principal stresses. For the maximum principal stresses, the region appears to be fairly uniform. For minimum principal stresses, the area just below the lesser trochanter appears to have

the highest minimum principal stresses. However, the stress in the region is not as high as it was in the same region of the healthy femur. Figures 3-16 and 3-17 show contour plots for the proximolateral region below the greater trochanter. Figure 3-16 shows maximum principal stresses while figure 3-17 shows minimum principal stresses. Figure 3-16 shows the area of highest maximum principal stresses where the bone is in tension. The region has fewer high stress areas compared to the healthy femur. Figure 3-17 shows lower minimum principal stresses as a result of less compression. Figures 3-18 and 3-19 provide a top view of the femoral neck. Figure 3-18 shows maximum principal stresses while figure 3-19 shows minimum principal stresses. This is another area of high maximum principal stress as a result of the femoral neck being in tension in this location. For the cam femur, the neck region does not contain stresses as high as those seen in the healthy femur. The higher stresses are concentrated closer to the bony protrusion that causes the cam impingement.

Figure 3-20 Maximum principal stress in the proximomedial region of the cam femur that was paired with the pincer pelvis.

Figure 3-21 Minimum principal stress in the proximomedial region of the cam femur that was paired with the pincer pelvis.

Figure 3-22 Maximum principal stress in the proximolateral region of the cam femur that was paired with the pincer pelvis.

Figure 3-23 Minimum principal stress in the proximolateral region of the cam femur that was paired with the pincer pelvis.

Figure 3-24 Maximum principal stress in the neck region of the cam femur that was paired with the pincer pelvis.

Figure 3-25 Minimum principal stress in the neck region of the cam femur that was paired with the pincer pelvis.

Figures 3-20 and 3-21 show contour plots for the proximomedial region below the lesser trochanter on the cam femur that was paired with the pincer pelvis. Figure 3-20 shows maximum principal stresses while figure 3-21 shows minimum principal stresses. For the maximum principal stresses, the region appears to be

fairly uniform. For minimum principal stresses, the area just below the lesser trochanter appears to have the highest minimum principal stresses. Similar to the cam femur paired with the healthy pelvis, the stress in the region is not as high as it was in the same region of the healthy femur. Figures 3-22 and 3-23 show contour plots for the proximolateral region below the greater trochanter. Figure 3-22 shows maximum principal stresses while figure 3-23 shows minimum principal stresses. Figure 3-22 shows the area of highest maximum principal stresses where the bone is in tension. Similar to the cam femur paired with the healthy pelvis, the region has fewer high stress areas compared to the healthy femur. Figure 3-23 shows lower minimum principal stresses as a result of less compression. Figures 3-24 and 3-25 provide a top view of the femoral neck. Figure 3-24 shows maximum principal stresses while figure 3-25 shows minimum principal stresses. This is another area of high maximum principal stress as a result of the femoral neck being in tension in this location. For the cam femur, the neck region does not contain stresses as high as those seen in the healthy femur. The higher

stresses are concentrated closer to the bony protrusion that causes the cam impingement.

Figure 3-26 Maximum principal stress in the proximomedial region of the healthy femur that was paired with the pincer pelvis.

Figure 3-27 Minimum principal stress in the proximomedial region of the healthy femur that was paired with the pincer pelvis.

Figure 3-28 Maximum principal stress in the proximolateral region of the healthy femur that was paired with the pincer pelvis.

Figure 3-29 Minimum principal stress in the proximolateral region of the healthy femur that was paired with the pincer pelvis.

Figure 3-30 Maximum principal stress in the neck region of the healthy femur that was paired with the pincer pelvis.

Figure 3-31 Minimum principal stress in the neck region of the healthy femur that was paired with the pincer pelvis.

Figures 3-26 and 3-27 show contour plots for the proximomedial region below the lesser trochanter on the cam femur that was paired with the healthy pelvis. Figure 3-26 shows maximum principal stresses while figure 3-27 shows minimum principal stresses. For the maximum principal stresses, the region appears to be

fairly uniform. For minimum principal stresses, the area just below the lesser trochanter appears to have the highest minimum principal stresses. This appears to be similar to the healthy femur paired with the healthy pelvis. Figures 3-28 and 3-29 show contour plots for the proximolateral region below the greater trochanter. Figure 3-28 shows maximum principal stresses while figure 3-29 shows minimum principal stresses. Figure 3-28 shows the area of highest maximum principal stresses where the bone is in tension. Figure 3-29 shows lower minimum principal stresses as a result of less compression. Figures 3-30 and 3-31 provide a top view of the femoral neck. Figure 3-30 shows maximum principal stresses while figure 3-31 shows minimum principal stresses. This is another area of high maximum principal stress as a result of the femoral neck being in tension in this location. The distribution is similar to that seen in the healthy femur paired with the healthy pelvis.

3.3 Pelvis Validation

The pelvis of the hip model was validated by comparing the stress results in the simulation to experimental data of a cadaveric pelvis (figure 3-32).

From experimental data, a maximum pressure of 8.6 MPa was recorded for an individual that weighed 73 kg [31]. This pressure was used for validation because it approximately corresponds to the body weight used for the model. This pressure is assumed to represent a minimum principal stress based on Afoke et al's experimental set up. In the experimental set up, pressure sensitive film was placed in between the femoral head and the acetabulum. The measured stresses would be compressive.

Figure 3-32 Minimum principal stresses in the acetabulum of the healthy hip model compared to experimental data.

Figure 3-32 shows the minimum principal stress that was taken from the superior region of the acetabulum and represents an average of values. The

comparison is between the pelvis in the healthy hip model and the acetabulum from Afoke at al's experiment [31]. For the healthy pelvis, the minimum principal stress was -7.16 MPa. From Afoke's experimental data, the minimum principal stress was -8.6 MPa [31]. The simulated value is very close to the experimental value, indicating that the model is both reliable and accurate.

3.4 Acetabular Stresses

The values generated by the Abaqus model vary slightly from the experimental data, indicating that the model can be used.

Figure 3-33 Maximum principal stresses in the acetabulum of the healthy hip model compared to the pelvis paired with the cam femur.

Figure 3-34 Minimum principal stresses in the acetabulum of the healthy hip model compared to the pelvis paired with the cam femur.

Figures 3-33 and 3-34 show the maximum and minimum principal stresses that were taken from the superior region of the acetabulum. The comparison is between the pelvis in the healthy hip model and the healthy pelvis paired with the cam femur. For the healthy pelvis paired with the cam femur, the maximum principal stress was 9.97 MPa and the minimum principal stress was -7.16 MPa. For the healthy pelvis paired with the cam femur, the maximum principal stress was 7.45 MPa and the minimum principal stress was -4.01 MPa.

Figure 3-35 Maximum principal stresses in the acetabulum of the healthy hip model compared to the pincer pelvis that was paired with a healthy femur.

Figure 3-36 Minimum principal stresses in the acetabulum of the healthy hip model compared to the pincer pelvis that was paired with a healthy femur.

Figures 3-35 and 3-36 show the maximum and minimum principal stresses that were taken from the superior region of the acetabulum. The comparison is between the pelvis in the healthy hip model and the pincer pelvis paired with the healthy femur. For the pincer pelvis paired with the healthy femur, the maximum principal stress was 17.42 MPa and the minimum principal stress was -9.89 MPa.

Figure 3-37 Maximum principal stresses in the acetabulum of the healthy hip model compared to the pincer pelvis paired with the cam femur.

Figure 3-38 Minimum principal stresses in the acetabulum of the healthy hip model compared to the pincer pelvis paired with the cam femur.

Figures 3-37 and 3-38 show the maximum and minimum principal stresses that were taken from the superior region of the acetabulum and represent an average of values. The comparison is between the pelvis in the healthy hip model and the pincer pelvis paired with the cam femur. For the pincer pelvis paired with the cam femur, the maximum principal stress was 17.72 MPa and the minimum principal stress was -7.76 MPa.

Figure 3-39 Maximum principal stress in the acetabulum of the healthy pelvis that was paired with the healthy femur.

Figure 3-40 Minimum principal stress in the acetabulum of the healthy pelvis that was paired with the healthy femur.

Figures 3-39 and 3-40 show contour plots of the acetabulum from the healthy pelvis that was paired with the healthy femur. Figure 3-39 shows a contour plot of the maximum principal stresses while figure 3-

40 shows a contour plot of the minimum stresses. In both figures, the greater maximum and minimum principal stresses occur in the superior region of the acetabulum and extend into the inferior region.

Figure 3-41 Maximum principal stress in the acetabulum of the healthy pelvis that was paired with the cam femur.

Figure 3-42 Minimum principal stress in the acetabulum of the healthy pelvis that was paired with the cam femur.

Figures 3-41 and 3-42 show contour plots of the acetabulum from the healthy pelvis that was paired with the cam femur. Figure 3-41 shows a contour plot of the maximum principal stresses while figure 3-42

shows a contour plot of the minimum stresses. In figure 3-41, the greatest maximum principal stresses occur in the superior region. In figure 3-42, the greatest minimum principal stresses occur in the anterior-superior region and extend into the inferior region.

Figure 3-43 Maximum principal stress in the acetabulum of the pincer pelvis that was paired with the cam femur.

Figure 3-44 Minimum principal stress in the acetabulum of the pincer pelvis that was paired with the cam femur.

Figures 3-43 and 3-44 show contour plots of the acetabulum from the pincer pelvis that was paired with the cam femur. Figure 3-43 shows a contour plot of the maximum principal stresses while figure 3-44 shows a contour plot of the minimum stresses. In both figures,

the greatest maximum and minimum principal stresses occur in the superior regions along with the anterior-superior regions.

Figure 3-45 Maximum principal stress in the acetabulum of the pincer pelvis that was paired with the healthy femur.

Figure 3-46 Minimum principal stress in the acetabulum of the pincer pelvis that was paired with the healthy femur.

Figures 3-45 and 3-46 show contour plots of the acetabulum from the pincer pelvis that was paired with the cam femur. Figure 3-45 shows a contour plot of the maximum principal stresses while figure 3-46 shows a contour plot of the minimum stresses. In both figures,

the greatest maximum and minimum principal stresses occur in the superior regions.

4. Discussion

A healthy hip model consisting of a femur and a hemi-pelvis was constructed and validated using experimental data for femoral strains and acetabular stresses. The validated healthy hip model was used to compare models containing both cam and pincer impingements. The femoroacetabular impingements were created using the 3D modeling software Autodesk Maya and finite element simulations were created using Abaqus. Acetabular stress was the main area of interest in this study. Femoral strains in the healthy hip model closely matched those found in Deuel's experiment. The acetabular stress in the healthy hip model was slightly lower than the stress presented by Afoke [31]. Overall, the healthy hip model compared reasonably well to the experimental data, with a difference of about 1.44 MPa. The acetabular stresses in both of the models containing a pincer pelvis were greater than the healthy model and the model containing the cam impingement.

Acetabular stresses in the models containing the pincer pelvis were expected to be higher due to the overcoverage of the pincer impingement. Pincer

impingements are associated with labral tearing [6]. It has been suggested that labral tearing in dysplastic hips is due to increased loading in the hip [Hanek 2011]. For the model containing a cam femur and healthy hemi-pelvis, stresses were not expected to be significantly different than those seen in the healthy model. This is because the cam impingement is not making contact with the acetabulum during the single-leg stance phase. All models were positioned the same and contained the same boundary and loading conditions. Any differences in stress would be due to the shape of the femora and acetabula.

Figure 4-1 A comparison of the femoral maximum principal stresses in all of the models.

Figure 4-2 A comparison of all the femoral minimum principal stresses in all of the models.

Figure 4-1 shows a comparison of the femoral maximum principal stresses in all of the models. From the figure, the healthy hip experienced the highest maximum principal stress in the proximolateral region and the second highest in the proximomedial region and this was closely followed by the cam femur that was paired with the healthy pelvis. The results indicate that the pincer impingement actually decreases the stress on the proximomedial and proximolateral regions of the femur while the cam femur paired with the healthy pelvis maintains stress values that are comparable to a healthy hip.

Figure 4-2 shows a comparison of the femoral minimum principal femoral in all of the models. The healthy femur with the pincer pelvis experienced the highest maximum principal stress in the proximomedial region. For the minimum principal stresses, the healthy femur experienced the highest values for both the proximomedial and proximolateral regions. These minimum principal stress values indicate that both impingement types may reduce the stress experienced by the femur.

Figure 4-3 A comparison of the acetabular maximum principal stresses in all of the models.

Figure 4-4 A comparison of the acetabular minimum principal stresses in all of the models.

Figure 4-3 shows a comparison of the acetabular maximum principal stresses in all of the models, including the experimental model. The pelvis from the healthy hip model compared very well to the experimental maximum pressure of 8.6 MPa. The healthy pelvis that was paired with the cam femur actually registered lower maximum principal stresses compared to the healthy hip model. This indicates that the cam impingement does not increase acetabular stresses during the single-leg stance phase of the walking cycle. Both models containing the pincer pelvis showed

significantly greater maximum principal stresses than the models containing the healthy pelvis.

Figure 4-4 shows a comparison of the acetabular minimum principal stresses for all of the models. For minimum principal acetabular stress, the healthy pelvis paired with the cam femur registered the lowest stress. Both models containing the pincer pelvis registered the highest stresses out of the models. These results suggest that the cam femur may actually reduce the stress in the acetabulum during this moment of the walking cycle, but that the pincer pelvis increases the acetabular stress.

The pincer impingements are responsible for increasing both maximum and minimum principal stresses when compared to a healthy hip. Interestingly, the pincer impingement decreases the stress experienced by the femur while the cam impingement decreases the stress experienced by the pelvis. In combination, however, the stress experienced by the femur decreases while the stress experienced by the pelvis increases. The bony protrusion responsible for the cam impingement appears to aid the femur in resisting bending, therefore reducing compression and tension in

the proximomedial and proximolateral regions, respectively. With reduced stress in these regions, the bone may become less dense and, therefore, weaker as a result. Furthermore, the bony protrusion extending from the acetabular rim increases the bending experienced in the acetabulum. As a result, this increases the maximum principal stresses in the region, as seen in figure 4-3. The increased stress in the acetabulum may increase the pain felt by the patient and may also place the patient at greater risk for acetabular fracture.

The models compared very well to experimental data, however, there are slight differences between the experimental results and simulation results. The femur and the hemi-pelvis came from two separate sources and the femur had to be resized to fit the acetabulum of the hemi-pelvis. As Harris et al [8] demonstrated in a finite element study involving hips from several different patients, the stress distributions within the acetabulum will vary greatly from person to person. While the range of the stress distribution is predictable based on data found in literature, it is difficult to predict the stress

distribution for a particular patient. Because the joint did not contain acetabular cartilage, the stress distribution will not be physiologically accurate. Without cartilage, the two bony surfaces are not compliant and may cause pressure points where peak pressures appear to be higher than they should. In a study conducted by Hao et al, cartilage was not included in their model and they suggested that this may influence the stress distribution around the acetabulum [15]. Afoke et al, however, report that their experiment does not corroborate the idea that cartilage does not distribute the applied evenly, which would be consistent with the results found in the simulation that is presented here [31].

The model did not use muscle forces on either of the bones to simplify the model. The experiment described by Afoke et al [31] did not consider muscle forces when acetabular stresses were being recorded. However, Deuel et al [4] replicated the abductor in their experiment. A study conducted by Correa et al concluded that muscle forces contribute significantly to the resultant contact force within the hip in the normal gait cycle [32]. Therefore, muscle forces

should be considered in the future to make the model more physiologically accurate.

The cortical thickness was approximately 4 mm throughout each of the bones, which is inconsistent with the varying cortical thickness found in native bones. Cartilage was not modeled on the femoral head or within the acetabulum. Instead, surfaces were assigned for the femoral head and the horse shoe-shaped articular cartilage of the acetabulum in Abaqus. Contact was established between these two surfaces and the contact was assumed to be frictionless. Cartilage serves as a lubricating, nearly frictionless surface, but it is too thin to provide any considerable amount of stress absorption [2]. The interaction between the two articular cartilage surfaces can modify the transmission of forces between the femoral head and the acetabulum [34]. For the sake of this model, the goal was to compare average stress between the acetabula of each model. If the stress distribution of the articular cartilage surface of the acetabulum was being studied, inclusion of cartilage in the model would be necessary.

The model only considers the single-leg stance phase of the walking cycle and did not model the full gait cycle. Future models should consider the entire gait cycle. Different effects of the impingements can be seen by looking at different points in the gait cycle. The cam impingement, for example, would be expected to affect the hip the greatest when the impingement abuts against the acetabular rim. Different points in the gait cycle are associated with different amounts of stress seen in the acetabulum [17, 31]. Furthermore, the impingement features would affect the joint differently depending on how the femoral head is oriented in relation to the acetabulum.

This study demonstrated that the 3D modeling software Autodesk Maya can be used to modify the geometry of bone models that were created using CT scans. This is an advantageous method because it allows for the modification of organic shapes, such as bones, and does not affect how the solid part behaves in Abaqus. These altered models can be used to create finite element models to analyze the effects of bone defects. Specifically, the two main types of

femoracetabular impingement were created and analyzed in a simplified walking model. The results of the simulations demonstrated that femoracetabular impingements, in general, decreases stress on the femur while a pincer impingement increases the stresses in the acetabulum during single-leg stance phase of the walking cycle.

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