

The Mechanics of Cam-type Femoroacetabular Impingement

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Cam-type Femoro-Acetabular Impingement (FAI) is a common cause of hip osteoarthritis (OA). In this condition a bony abnormality at the head-neck junction of the femoral head, called the “cam”, abuts against the acetabulum causing labral damage and articular cartilage delamination, which in turn may lead to progressive degeneration and OA. The understanding of the damage mechanism is currently at a conceptual level. The aim of the thesis is to develop a more detailed understanding of the underlying mechanism so as to improve methods of detection and treatment of cam-type FAI and thus to help prevent hip OA.

A geometric-kinematic model combining hip joint motion and hip joint geometry was created to determine what motions, activities or cam shapes give rise to cam-type impingement, which was quantified by the proximity of the acetabular and femoral bony surfaces. Five normal subjects and five symptomatic cam-type FAI patients were modelled. The FAI patients experienced early impingement during the impingement test but did not have impingement during common functional activities. The early impingement was possibly due to the larger coverage and protrusion of their cams and the smaller overall proximity in their hip joints.

A 2D finite element (FE) model was created to simulate cam-type FAI. As idealised 2D rectangular and circular geometries did not reproduce the damage seen clinically, subject-specific geometry, loads, and motions were introduced. Under some circumstances, as the cam entered the hip joint, large shear strains developed near the cartilage-bone interface of the acetabulum which would result in cartilage delamination.

In vitro experiments were undertaken to validate the FE model and verify the damage mechanism by which cam-type FAI leads to cartilage delamination. Porcine cartilage-bone samples were loaded under conditions similar to those generated by a cam (shear and compression). A validation FE model was created that used the same material and contact representations and analysis framework as the impingement FE model but mimicked the experimental setup. The cartilage shear strains assessed with a video-based method were similar to predicted FE results. *In vitro* damage experiments demonstrated that delamination can be caused by repetitive shear and compressive loading that lead to large shear strains near the cartilage-bone interface.

The impingement FE model was used to further explore the effect of cam anatomy. In hips with low clearance, cams with large protrusions (75% hip joint clearance) would not enter into the hip joint, but caused high shear strains in the labrum, which would result in labral tears. A narrower cam caused damage to the labral tip, whereas a wider cam caused damage to the labral-bone junction. In contrast, cams with small protrusion (25% hip joint clearance) were able to enter the joint and caused damage at the articular cartilage-bone interface, which would result in cartilage delamination. The wider the cam, the further into the hip joint the damage was initiated.

The FE model was used to explore the effect of different labral anatomy and of reshaping surgery. A labrum connected to the articular cartilage resulted in shear strains of up to five times greater in the articular cartilage and labrum compared to an unconnected labrum and was more likely to cause articular cartilage delamination. For a cam that damages the articular cartilage, surgical removal of the cam reduced shear strains. For a cam that abuts the labrum, surgical removal of the cam eliminated labral abutment and increased the range of motion of the hip, but resulted in greater shear strains in the articular cartilage. It is not known whether these shear strains are normal or could possibly be damaging. Also, reshaping the head to be spherical resulted in slightly reduced shear strains in the articular cartilage compared to the current surgical practice of cutting deeper into the femoral head when removing the cam.

This study has, for the first time, using a validated FE model demonstrated the mechanism by which a cam can cause articular cartilage delamination and labral tearing. Further analysis using the geometric and FE model should help identify cam deformities that would be likely to cause OA and the best way to treat them surgically so as to prevent OA.

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Chapter 1

Introduction

1.1 Motivation

The hip joint is essential for normal biomechanical function, supporting high loads which may be many times body weight and facilitating smooth articulation of the lower limbs [2, 3, 4]. It is a ball-and-socket joint consisting of the head of the femur bone and the acetabulum of the pelvis. The surfaces of the femoral head and acetabulum are covered by articular cartilage, a highly specialised tissue with unique mechanical properties which provide a low-friction, wear-resistant surface for articulation and the distribution of loads in the hip joint. Articular cartilage is avascular, which contributes to its limited capacity for self-healing and repair. It is aneural, which leads to inconsistencies when it comes to the relationship between patient pain and damaged or diseased cartilage. Articular cartilage is the key biomechanical structure that enables the hip joint to endure a lifetime of repetitive loading. Unfortunately, damage to articular cartilage is difficult to repair, is challenging to detect through non-invasive imaging modalities (such as Magnetic Resonance Imaging (MRI) and Magnetic Resonance Arthrography (MRA)), and is not clearly related to patient symptoms such as pain. Osteoarthritis (OA) is a joint disease that primarily affects the articular cartilage which reduces the mobility and biomechanical functioning of the hip joint, and it is not fully understood and difficult to treat.

OA was ranked the 6th most disabling condition for all age groups by the World Health Organization (WHO) in 2004 [5]. It is characterised by the loss of articular cartilage and eburnation of the underlying subchondral bone. It predominantly affects the older population, however, is now more often seen in younger patients. A possible cause is thought

to be cam-type femoroacetabular impingement (FAI), which involves a slightly abnormally shaped femur and acetabular articular cartilage damage. In particular, an extra bony portion, known as the “cam” or “bump”, at the junction of the femoral head and neck is thought to cause damage to the articular cartilage of the acetabulum when it is forced into the hip joint [3, 4, 6, 7, 8, 9, 10, 11, 12, 13, 14, 15, 16, 17, 18, 19]. It has been suggested that this kind of damage is due to the wedging of the cam into the hip joint causing excessive compressive or shear forces [12, 13, 19]. During keyhole surgery of FAI patients, known as hip arthroscopy, the acetabular articular cartilage is reported to have delaminated from the underlying bone presumably resulting from damage at the cartilage-bone interface [20, 21, 22, 23]. Eventually, the delaminated cartilage is torn off, leaving exposed bone. This in turn is thought to lead to progressive degeneration and OA of the hip.

The current strategy to treat cam-type FAI entails removing the cam deformity. This strategy has yet to be shown to prevent OA in the long-term. There is also a lack of guidelines on how to contour the head-neck junction of the femoral head to best prevent further progression of disease. In addition, there is evidence that surgical outcomes are less favorable in patients with more advanced hip pathology and sometimes require a conversion to a total hip replacement (THR) [24, 25]. Therefore, there is a need for early detection of disease, careful patient selection for surgical treatment, and pre-operative planning tools for re-contouring surgical treatments. To approach these three current needs, it is essential to understand the mechanism by which disease is initiated and progresses. This thesis aims to simulate the disease initiation and propagation mechanism of cam-type FAI in order to understand how the various factors involved in this condition, focusing on hip morphology, affect the disease mechanism. Understanding how these factors come into play in the development of hip pathology can give valuable insight into who might be at greater risk of developing disease, which patients are suitable for surgical treatment, and how a person’s hip morphology should be contoured to best treat the condition.

1.2 Scope of Project

Current trends in research revolving around cam-type FAI generally focus on cam-type related hip pain, the etiology and prevalence of the cam deformity, and the etiology, prevalence, diagnosis, progression, and treatment of the disease. Some studies have presented a comprehensive assessment of common daily activities that lead to pain in cam-type FAI patients [26, 27]. Some studies have presented possible compensation mechanisms of the hip joint as a result of hip deformity that may lead to hip pain [28], and numerous studies have associated hip pain with signs of having a cam deformity in the young population [21, 24].

Research centred on the possible etiologies of the cam deformity focus on its possible development involving pre-disposing hip conditions and trauma-related, iatrogenic, or biomechanical causes of the formation of the bump at the head-neck junction of the femoral head [6, 29]. Many studies have been published on the conceptual etiology of cam- and pincer-type FAI, however, these studies remain very general and lack depth of understanding of the underlying mechanism by which disease is initiated and progressed [3, 4, 6, 7, 8, 9, 10, 11, 12, 13, 14, 15, 16, 17, 18, 19]. Understanding the underlying mechanism of cam-type FAI is the area of research that this thesis aims to address.

A few large cross-sectional or population-based studies and many smaller studies have been conducted to determine the prevalence of the cam deformity among asymptomatic individuals [11, 23, 30, 31, 32, 33]. Many studies have also been conducted to determine the prevalence of the cam deformity or of cam-type FAI among the younger population and among athletes involved in high impact sports such as football or basketball [34, 35]. Some studies have also focused on the prevalence of soft tissue damage in the hip joint among asymptomatic [36] or symptomatic individuals [20, 24].

Research on the diagnosis of cam-type FAI revolves around the alpha angle (p.21), 3D characterisation of femoral morphology, and imaging of the cam deformity and of soft tissue damage. The validity of the alpha angle as an assessment tool for the presence of a cam deformity or cam-type FAI has been much debated [27, 37], numerous measurements of the normal range [38, 39] of an alpha angle and of alpha angle thresholds for pathogenic cam deformities have been presented [40, 41, 42, 43], and a few studies have assessed which radiographic view should

be used to obtain accurate measurement of an alpha angle [44]. Various techniques for 3D characterisation of the morphology of the femoral head and cam deformity have been explored [19, 45, 46, 47]. Many studies have presented guidelines for MRI and MRA imaging of lesions [22, 48, 49] and have assessed the diagnostic sensitivity and accuracy of using MRI and MRA for the detection of labral tears [50, 51, 52] or cartilage delamination [20, 52].

Many studies have demonstrated the possible progression of cam-type FAI to hip OA through the high prevalence of signs of cam-type FAI in hip OA patients [3, 4, 53]. However, very few have demonstrated the natural progression of cam-type FAI into hip OA in actual patients.

Finally, numerous studies have presented short-term outcomes of hip arthroscopy and open surgery as treatments for cam-type FAI [24, 25, 54, 55], and a few studies have presented the use of computer-aided modelling of patient-specific hip joint geometry for surgical pre-planning and post-operative analysis [56, 57]. However, a lack of long-term studies have left one of the biggest questions in the field unanswered – can early intervention or surgical treatment of cam-type FAI stop the natural progression of disease?

Ultimately, this thesis aims to investigate the complicated mechanism by which cam-type FAI leads to damage in the hip joint. This investigation may possibly lead to insights for characterisation of the cam deformity and hip joint morphology, early intervention and patient selection for treatment based on pathogenic cam morphology, and possibly lead to guidelines for reshaping surgery.

1.3 Thesis Aims

The overall aim of this research project is to simulate the disease mechanism of cam-type FAI through computational models and *in vitro* experiments. These models and experiments will be used to gain insight into the effect that various anatomical factors have on the initiation and progression of hip joint damage with the overall goal of improving methods of detection and treatment of cam-type FAI. To fulfill this objective, the following specific aims have been generated:

1. To produce a computational model which determines the physical relationship of the cam deformity to the acetabulum during key activities such as the impingement test and the functional activities of walking, sit-to-stand, and stand-to-sit.
 - a. To determine the proximity of the cam deformity to various regions of the acetabulum during functional activities.
 - b. To determine the presence of cam-type impingement during various motions and activities to understand when and why it occurs.
2. To simulate the disease mechanism of cam-type FAI.
 - a. To analyse the shear strains that develop in the soft tissues of the hip joint, focusing on the anterior acetabular articular cartilage and the labrum.
 - b. To simulate the initiation and propagation of damage in the acetabular articular cartilage and the labrum.
 - c. To investigate the disease mechanism's sensitivity to varying hip joint anatomy.
3. To develop an *in vitro* system which can simulate the model created for Aim 2.
 - a. To validate the model through the comparison of measured and calculated strain results.
 - b. To verify the delamination process of articular cartilage.

Chapter 2

Background

2.1 Introduction

This chapter begins with an overview of the anatomy of the hip joint focusing on the bony anatomy and soft tissues including the labrum and articular cartilage. This is followed by a description of the hip joint pathology, involving the degeneration of cartilage, due to femoroacetabular impingement (FAI) and more specifically cam-type FAI. This leads to a description of the damage initiation and propagation of articular cartilage due to large strain. Mathematical models used to quantify damage, and finally, Finite Element Analysis (FEA) and FE methods for modelling damage and failure are also described.

2.2 Anatomical Orientations

It is important to be able to describe the orientation of the different parts of the human body for the purposes of studying human motion or kinematics. The movement of the human body parts and their relative position to each other can be described using three anatomical planes – coronal (frontal), transverse (axial), and sagittal (Figure 2.1). The coronal plane splits the body into its front and back parts, where parts positioned towards the front (belly) of the body are described as anterior or ventral, and parts positioned towards the back of the body are described as posterior or dorsal. The transverse plane splits the body into its upper and lower parts, where parts positioned towards the upper part of the body are described as superior, and parts positioned towards the lower part of the body are described as inferior. The sagittal plane splits the body into its left and right halves, where parts positioned towards the midline of the

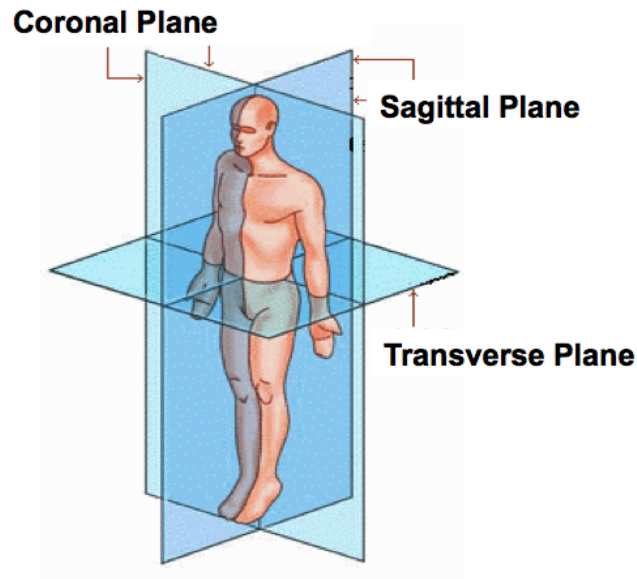


Figure 2.1: The three anatomical planes – coronal, sagittal, and transverse – which describe movement of the human body parts and their relative positions to each other [58].

body are described as medial, and parts positioned towards the side of the body are described as lateral. Additionally, the term cranial can be used to describe body parts positioned towards the head, and the term caudal can be used to describe body parts towards the lower end of the spine. The term distal can be used to describe body parts positioned away from the origin, and the term proximal can be used to describe body parts towards the origin. These terms are essential when it comes to describing the kinematics of joints, which form where two or more bones meet.

2.3 The Hip Joint

The hip joint is a synovial ball-and-socket joint (Figure 2.2). Its structure gives it its load-carrying capacity and ability to provide smooth articulation for large ranges of motion (ROM). It is formed from the proximal end of the long femur bone and the acetabulum of the pelvis. The spherical femoral head of the proximal femur lies within the concavity of the socket-like acetabulum and articulates during motion. An incomplete ring of fibrocartilagenous tissue, called the labrum, is attached to the rim of the acetabulum, deepening the socket of the joint. The femoral head and acetabulum are encased in a highly innervated, but avascular and fibrous articular capsule. The inner layer of the articular capsule is the synovial membrane which secretes lubricating and nourishing synovial fluid. Secreted synovial fluid lies in the spaces between the adjacent bones, known as the synovial cavity. The articular capsule and labrum

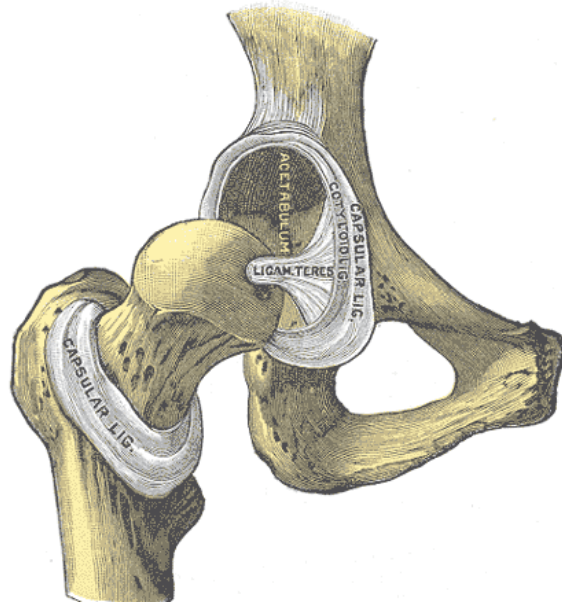


Figure 2.2: The anatomy of the hip joint is presented including the acetabular socket of the pelvis and the femoral head of the femur which are attached by the ligament of teres [59].

structure provide the hip joint with stability as it articulates through large ROM.

The articulating surfaces of the femoral head and acetabulum are covered by a thin layer of articular cartilage ranging from 0.8 to 4 mm in thickness [60, 61, 62, 63]. The articulating surfaces of the femoral head consists of all surfaces surrounding the fovea capitis femoris, a small area located medially for attachment of the ligament of the head of the femur. The articulating surfaces of the acetabulum, called the lunate surface, consists of a horseshoe-shaped region. This region surrounds the anterior, superior, and posterior surfaces of the acetabular fossa, a rough depression in the central inferior part of the acetabulum. The acetabular fossa is where the ligament (the ligament of teres) of the head of the femur attaches, and the transverse acetabular ligament lies at the base of the acetabular fossa, completing the ring of the labrum.

2.3.1 Pelvic and Femoral Bone

The pelvis is a ring-like structure which holds and protects digestive and reproductive organs, and connects the spine of the upper body to the lower limbs through the hip joint. It consists of two hip bones and the sacrum. The hip bone is formed from three separate ossification centres – the ilium, ischium, and pubis – which start out as three separate bones in childhood. During adolescence, these three bones unite at the acetabulum to form the y-shaped tri-radiate

cartilage which fuses usually by the age of 18 [64]. The ilium is the uppermost and largest bone of the pelvis. The anterior superior iliac spine is the portion of the ilium that protrudes most anterior superiorly. The socket-like opening of the acetabulum is oriented about 45° caudally and is anteverted or oriented anteriorly 5 to 15° [65, 66, 67].

The femur is the longest bone in the human body. Proximally, it consists of the femoral head, the neck of the femur, and lateral bony protrusions called the greater and lesser trochanters. More distally, it consists of the diaphysis, or shaft of the long bone, and the femoral medial and lateral condyles which form part of the knee joint. The angle between the femoral neck and shaft is about 130° , and the femoral head and neck is anteverted about 14° relative to the femoral condyles [67].

The structure of the pelvic and femoral bone consists of a dense, stiff shell of cortical (compact) bone filled with less dense trabecular (spongy) bone. The shell of cortical bone is 0.7 to 3.2 mm thick [68] in the pelvis and 5.13 to 7 mm thick in the femoral head and femoral diaphysis [69, 70]. In the areas with articular cartilage, a thin layer of subchondral bone lies sandwiched in between the cortical bone and articular cartilage. The transition from articular cartilage to subchondral bone is referred to as the cartilage-bone interface or osteochondral junction.

2.3.2 Labrum

The labrum is a very tough connective tissue which is important in the lubrication of the hip joint. It also provides stability in the hip and distributes pressures in the hip [71, 72, 73]. It is made up of type I collagen fibre bundles oriented parallel to the acetabular rim. It is firmly attached to the non-articulating side of the acetabular rim and to the calcified articular cartilage through a transition zone of 1 to 2 mm [74, 75]. The margins of the joint capsule are attached to the anterior, superior, and posterior base of the labrum. The anterior superior sections of the labrum are thought to be the most innervated, consisting of free nerve endings and sensory nerve end organs.

The labrum is 2 to 3 mm in thickness, and its cross-section is usually triangular in shape [16]. The incidence of this triangular shape decreases in older subjects, whereas the incidence of rounded and irregularly shaped labra increases with age [49, 74, 75]. Its cross-section is thinner and extends further from the rim anteriorly and is thicker posteriorly [74, 75, 76].

Sulci, or depressions in the labrum, can occur at its articular junction to the acetabular rim. These depressions are also known as sub-labral recessions. They have been determined to be normal variants, occurring in 17.4 to 30% of the population and have been found in the anterior, superior, and posterior regions of the labrum [49, 77, 78]. When viewed using imaging modalities with intra-articular contrast, a linear shape of contrast material will be positioned at the chondrolabral junction, resulting in partial separation of the labrum. They may look similar to a chondrolabral tear, however, they should not be mistaken for a tear. A tear would result in full thickness separation of the labral base and would be associated with perilabral abnormalities.

Due to its circular structure, the labrum serves as a seal to help keep synovial fluid within the articular cartilage. This creates fluid pressurisation which carries some of the transmitted loads and prevents direct contact between the acetabular and femoral articular cartilage [72, 74, 76]. Because it extends the rim of the acetabulum, it deepens the joint by about 20%, providing stability in the joint and resisting the translation of the femoral head. Also, it increases the surface contact area of the acetabulum by about 30%, distributing pressures over a larger area and thus reducing stress in the hip joint [74, 76, 79]. In previous studies a hip joint without a labrum increased contact stress up to about 90% [72, 76].

2.3.3 Articular Cartilage

Articular cartilage is a highly specialised tissue that forms a thin covering over all articulating surfaces of synovial joints in the human body; in the hip joint, it forms a thin 0.8 to 4 mm covering which is thickest in the anteromedial and superior regions [60, 61, 62, 63]. It plays a vital role in the biomechanical functioning of synovial joints. It allows nearly frictionless contact between articular surfaces, distributes transmitted loads over a larger contact area, reducing contact stress, and dissipates part of the energy associated with loads. It is an avascular and aneural tissue, making it difficult for self-repair, and yet this tissue manages to carry out its function in most people for their entire lifetime.

Articular cartilage's biomechanical function and durability can be attributed to its non-homogenous composition, anisotropic structure, and time-dependent strain material properties. Its composition consists of mainly water, chondrocytes, and an extracellular matrix (ECM) made up of stiff collagen fibres and a gel of proteoglycan (PG), a negatively charged protein.

The composition and organisation of these components differ through the thickness of articular cartilage. More specifically, articular cartilage can be separated functionally and structurally into four layered zones – superficial, middle, deep, and calcified. The unique composition, organisation, and structure of articular cartilage through these layers gives rise to the unique interaction between the ECM, composed of PG and collagen fibres, with water, giving articular cartilage its unique mechanical and material properties.

Proteoglycan (PG)

PG's are composed of a long protein backbone with protein side chains covalently attached called glycoaminoglycan (GAG). GAG's are highly negatively charged, causing the formation of a high cation concentration inside the tissue compared to the surrounding synovial fluid. This causes a pressure difference which gives rise to the swelling of the tissue and makes it resilient to compressive stresses [80]. The repulsion of closely organised negatively charged GAG's also contribute to the swelling of articular cartilage [81]. This swelling is restricted by the dense network of stiff collagen fibres which are strong in tension [82].

Collagen Fibres

Collagen fibres are fibrous proteins formed from smaller elongated fibrils. The stiffness of the collagen network depends on the density of cross-linking between collagen fibrils. Cross-linking is based on fibril interactions which can arise from either the entwinement of fibrils [83] or direct fibril-fibril interactions [84]. The stiffness of the collagen network gives articular cartilage its resilience to tensile and shear stresses.

Interaction of ECM with Water

Water is attracted to the cation concentration built up by the GAG side chains of PG's and contributes to the swelling and osmotic pressure in the tissue. When articular cartilage is compressed, the initial load is carried by its water content. This is followed by a consolidation process where water gradually exudes out of the tissue through its superficial membrane, the tissue volume decreases, and collagen fibres realign in the direction of stress in the tissue [85]. The rate of exudation is dependent on the permeability of the PG-collagen matrix which is fairly low and becomes even lower during compression when the pores of the ECM become

smaller. This results in articular cartilage's time-dependent strain behaviour, meaning articular cartilage will deform according to how much strain is applied and at what strain-rate deformation is applied. It also results in high interstitial fluid pressurisation which contributes more than 90% of the load transmission function of cartilage [86, 87, 88, 89]. This unique load-support mechanism protects the ECM from excessive stresses and reduces friction at the articular surfaces. Therefore, the composition and structure of the ECM and the interaction of the ECM with water gives articular cartilage its low-friction, wear-resistant, and strain-rate dependent material properties.

Material Properties

Articular cartilage is a viscoelastic material due to its matrix consolidation process. It can be modelled as a biphasic material where a liquid phase flows under the influence of pressure gradients produced by deformation of the tissue. It can also be modelled as a poroelastic material where fluid flows based on the permeability of the tissue. These model representations enable viscoelastic properties such as strain-rate dependence, creep, stress-relaxation, and hysteresis to be accounted for. Articular cartilage's strain-rate dependence results in the increase in its stiffness for increasing strain-rates. Creep is where a material exhibits time-dependent deformation under constant stress, stress-relaxation is where a material exhibits time-dependent stress relief under constant strain, and hysteresis is where there is a difference under loading and unloading (stress-strain) curves of force-deformation tests. The difference under the curves is the area of hysteresis which represents the energy lost due to internal friction within the material.

During dynamic, rapid, or impact loading (< 1 second) when articular cartilage is subjected to fast strain rates, these material properties are not exhibited because there is insufficient time for water to exude out of the tissue and thus tissue consolidation does not occur. During such loads, the volume of articular cartilage remains about the same, the tissue deforms laterally, its material stiffness becomes insensitive to strain-rate, and hysteresis effects become negligible resulting in more elastic behaviour [90, 91, 92, 93]. In a study by Oloyede *et al* where strain-rates ranging from $1.0\text{E-}5$ to $1.0\text{E}3 \text{ s}^{-1}$ were applied to articular cartilage, strain rates below 0.1 s^{-1} resulted in increasing stiffness for increasing strain-rates, whereas strain-rates above 0.1 s^{-1} resulted in approximately constant stiffness. These results indicate that articular cartilage behaves in a similar fashion to a linear elastic material at high strain-rates above 0.1 s^{-1} [93].

Zonal Properties

Understanding the varying composition and structure of articular cartilage in all four zones of articular cartilage provides further important details on the highly specialised structure and function of articular cartilage (Figure 2.3).

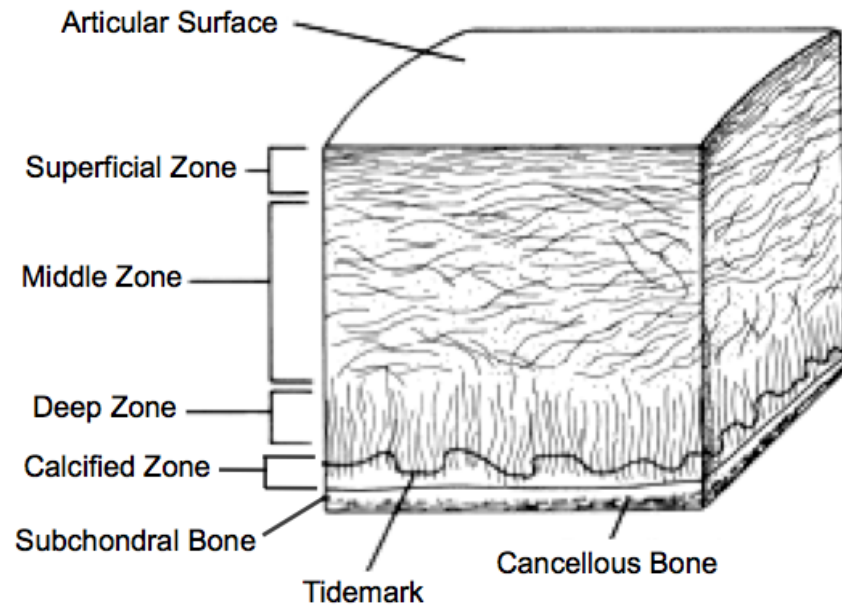


Figure 2.3: The zonal layers of articular cartilage including the superficial, middle, deep, and calcified zones [94].

Superficial Zone

The superficial zone accounts for 10 to 20% of the thickness of articular cartilage [95]. This zone has a high water and collagen content, but low PG content. Since PG's are associated with compressive stiffness, this zone has a low compressive modulus. However, collagen fibers are densely packed and oriented parallel to the articular surface. During compression and articulation, the tissue deforms laterally, increasing the load bearing contact area and more evenly distributing the transmitted load to the underlying bone [96]. Collagen fibres in the superficial layer are also stretched in tension which is the orientation for which they are strongest, so they are able to resist compressive forces on the tissue and frictional and shear forces at the articular surface.

Middle Zone

The transitional zone accounts for 40 to 60% of the volume of articular cartilage [95]. The water and collagen content is lower in this zone than the superficial zone, but there is a high PG content resulting in a high compressive modulus. Collagen fibres have a larger diameter and are randomly oriented [97]. This zone serves as the first line of resistance to compressive forces.

Deep Zone

The deep zone accounts for 30% of the volume of articular cartilage [95]. The water content and collagen content is lowest in this zone, and the PG content is highest in this zone [98]. Also, collagen fibres have the largest diameter in this zone and are oriented radially to the articular surface in bundles. When the tissue deforms laterally during compression and articulation, radially oriented bundles of collagen fibres in this zone tend to bend and orient obliquely or parallel to the articular surface. They resist tension in these directions, therefore, they are responsible for resisting shear strains that develop during compression and articulation.

Calcified Zone

The calcified zone accounts for 3 to 8% of the thickness of articular cartilage [99]. It is a transitional layer from articular cartilage to subchondral bone. Collagen fibres from the deep zone enter the calcified zone where it becomes mineralised or calcified and transitions into subchondral bone. This calcified layer of cartilage is separated from the cartilage of the deep zone by the tidemark and is separated from the underlying subchondral bone by the cartilage-bone interface or osteochondral junction. This zone is important for the anchoring of collagen fibres to the subchondral bone and providing resistance to shear stresses [99, 100, 101, 102].

2.4 Hip Joint Pathology

2.4.1 Osteoarthritis (OA)

Osteoarthritis (OA) is the most common type of arthritis. It affects nearly 40 million people or about 18% of the population in the United States and has a prevalence of 9 to 11% in the

European population [103, 104, 105]. It is commonly known to be a condition that affects the older population, but it is becoming more frequently seen in younger patients [106]. It was ranked the 6th most disabling condition for all age groups in a report by the World Health Organization (WHO) in 2004 [5]. In end-stage disease, it is characterised by joint pain, stiffness, dysfunction, and radiographic signs of loss of articular cartilage, and therefore, joint space narrowing, with the eburnation of the underlying subchondral bone and formation of osteophytes [107, 108].

OA affects the hip joint through the degeneration of articular cartilage. It is thought that the fluid-support mechanism of degenerated articular cartilage may be disturbed, making it vulnerable to further structural ECM changes upon normal physiological loading, progressing the development of OA [106]. The progressive nature of OA is of great concern because there is no currently known treatment to reverse or permanently stop the degeneration of articular cartilage. Additionally, the common treatment for hip OA is a total hip replacement (THR) which has survival rate at 20 years of about 90% [109]. Regardless of its high survival rate, it is still not an ideal option for younger patients. Therefore, particularly in the younger population, there is a need for early detection and prevention of hip OA.

OA in the hip was once commonly thought to be primary or idiopathic, meaning that the cause was unknown. About four decades ago, radiographic observations and retrospective studies provided evidence that minor anatomical abnormalities could be the etiology behind previously considered idiopathic hip OA [4]. This condition is known as femoroacetabular impingement (FAI), encompassing minor femoral or acetabular morphological abnormalities that may be disruptive to the biomechanical functioning of the hip joint [9, 10, 11]. However, it was not until the last 15 years that such etiologies have come to be more established as a possible cause of early hip OA. This recent occurrence can be attributed to the development of the technique of safely dislocating the hip and improvements in magnetic resonance imaging (MRI). Safe dislocation of the hip allowed for a 360° view of the minor anatomical abnormality and visualisation of the possible biomechanical mechanism leading to disease. Improvements in MRI techniques allowed for analysis of the 3-dimensional (3D) morphology of the bony surfaces and soft tissues in the hip joint.

2.4.2 Femoroacetabular Impingement (FAI)

FAI is a mechanical disorder of the hip joint involving the abutment of bony abnormalities in the hip joint during motion, leading to articular cartilage and labral pathology. It has been found in recent studies that radiographic features associated with FAI is prevalent in up to 66% of healthy and asymptomatic cohorts [39, 110, 111]. It is characterised by hip or groin pain and restricted ROM. Due to hip pain and restricted hip ROM, it has been reported that FAI patients have moderate difficulty to inability to perform repetitive functional activities. In particular, 55% of patients have reported moderate difficulty to inability to walk for 15 minutes or more, 60% for squatting, 45% for rising from a sitting position, about 25% for standing or sitting for 15 minutes, and about 88% for high demanding sports involving sprinting or cutting [26].

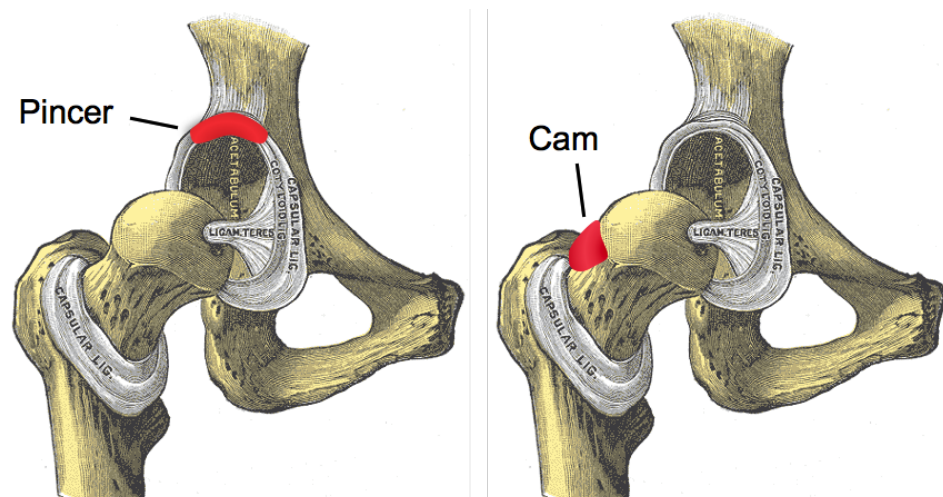


Figure 2.4: The pincer-type (left) and cam-type (right) FAI deformities are depicted.

There are two types of FAI, namely pincer- and cam-type (Figure 2.4). Pincer-type FAI predominantly affects middle-aged women (M:F 1:3) and is associated with acetabular bony deformities [7, 10]. Cam-type FAI predominantly affects younger males (M:F 14:1) and is associated with femoral bony deformities [7, 10]. Because cam-type FAI affects the younger population and can cause damage to acetabular articular cartilage leading to OA, it will be the focus of this work.

2.4.3 Cam-type FAI

Cam-type FAI involves abnormal bony morphology on the femoral side of the hip joint. The contours of these minor abnormalities can be described as a bump, or being hooked, flattened,

or pistol-griped shaped (Figure 2.5) [12]. The classical cam-type bony morphology is a pistol-grip which is named after its resemblance to the grip of a pistol when viewed in an anterior-posterior (AP) radiograph. These abnormal contours make the femoral head of the proximal femur non-spherical and create a broader head-neck junction. Overall, the head-neck junction of the femur loses its concavity which is essential for the articulation of the femoral head within the acetabulum for large ROM. The cam deformity and, therefore, loss in femoral head-neck concavity, usually lies in the anterior-superior region of the femoral head. This means that when a clock face is overlapped on the femoral head with the 3 o'clock pointing anteriorly, the cam deformity can range from the 11 o'clock position to the 4 o'clock position. The largest, or most protruding portion of the cam deformity is usually positioned at the 1:00 to 1:30 o'clock position [7, 112].



Figure 2.5: The presence of a cam lesion in an AP radiograph is indicated by a flattened femoral head-neck junction; this shape is comparable to the grip of a pistol, thus this deformity is known as the “pistol grip deformity” [12].

The development of a cam deformity can be due to a number of reasons ranging from idiopathic to developmental, childhood orthopaedic conditions, trauma-related, and iatrogenic reasons [6, 12, 29, 48]. Proposed developmental reasons include having a non-spherical femoral head or having coxa vara, a deformity of the hip whereby the angle between the femoral head and shaft of the femur is less than 120° (an angle of 120 to 135° is considered normal). Childhood conditions that affect the capital femoral epiphysis (bone growth plate at the head of the femur) can lead to a cam deformity such as Legg-Calve-Perthes Disease which may occur because the

blood supply to the capital femoral epiphysis becomes inadequate leading to breaking down of the bone and Slipped Capital Femoral Epiphysis (SCFE), in which fracturing of the growth plate may occur resulting in slippage of the epiphysis and altered bone growth. Trauma which results in a mal-united neck fracture or retroversion (tilting backwards) of the femoral head can also lead to a cam deformity. Finally, surgical treatments such as femoral osteotomy, where the femoral bone is cut to be shortened, lengthen, and/or be re-aligned, can lead to an iatrogenic-produced cam deformity.

Cam-type impingement has been associated with the motions of flexion, abduction, internal rotation during flexion, and during the combination of flexion, adduction, and internal rotation [7, 10, 12, 26, 112, 113, 114, 115]. It has been determined that different parts of the cam deformity comes into contact with, or impinge on the anterior superior area of the acetabulum during these motions. During flexion, the medial portions of the cam deformity are impingement-engaging. During abduction, the lateral portions of the cam deformity are impingement-engaging. During internal rotation at 90° of flexion, the centromedial portion of the cam lesion are impingement-engaging [113, 114].

It has been determined that the ROM is restricted in hips with a cam deformity compared to normal hips, with the deficit, greatest in internal rotation. Flexion has been determined to range from 120 to 125° in the normal hip and 102 to 114° in a cam-impinged hip [7, 21, 26, 113]. Abduction has been determined to range from 39 to 63° in the normal hip and 35 to 47° in a cam-impinged hip [7, 21, 26, 113]. Adduction has been determined to range from 20 to 30° in the normal hip and 10 to 26° in a cam-impinged hip [7, 21, 26]. Internal rotation has been determined to range from 30 to 40° in the normal hip and 10 to 27° in a cam-impinged hip [7, 21, 26, 113]. Internal rotation at 90° of flexion has been determined to be about 20 to 28° in the normal hip and 12 to 17° in a cam-impinged hip [113, 114].

Mechanism

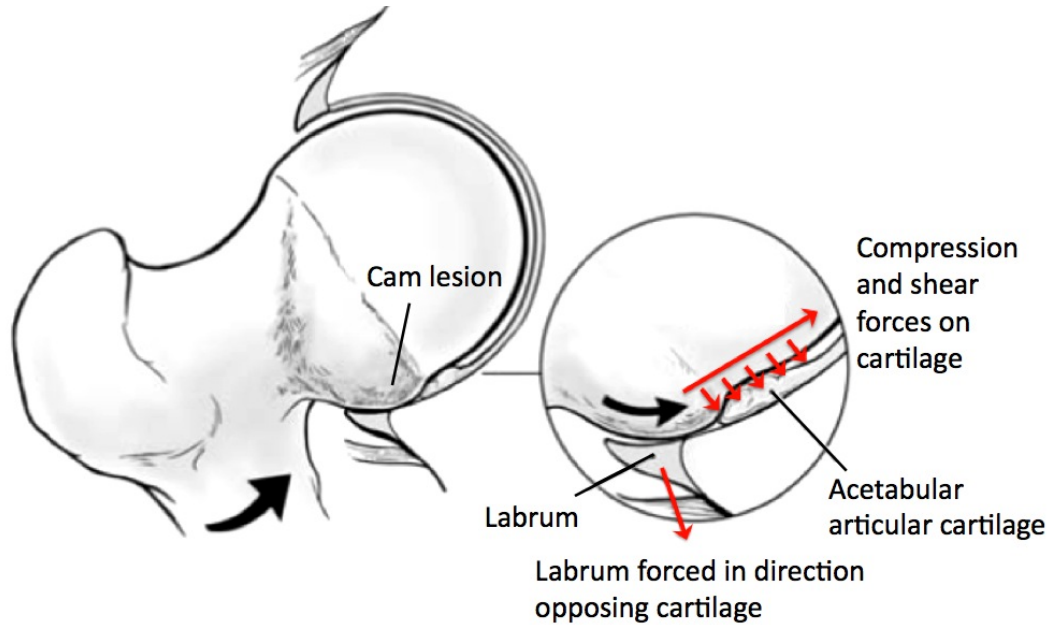


Figure 2.6: The currently proposed cam-type FAI mechanism; when the hip is flexed and internally rotated, the cam deformity enters the acetabulum, pushing the labrum outwards and the acetabular articular cartilage inwards [25].

During cam-type impingement, the area affected in the acetabulum is the anterior superior area. The articular cartilage and labrum in this area is where pathology commonly develops in this condition. It is also the area that the cam deformity contacts during impingement-engaging motions. It is thought that during impingement-engaging motions, the cam deformity can come into contact with the rim of the acetabulum. If larger ROM's are applied, it has been suggested that the cam deformity can be forced into the acetabulum, leading to structural damage in the articular cartilage and labrum [3, 4, 6, 7, 8, 9, 10, 11, 12, 13, 14, 15, 16, 17, 18, 19]. It has been suggested that this kind of damage is due to the wedging of the cam into the hip joint causing excessive compressive or shear forces [12, 13, 19]. Once articular cartilage is damaged, its biomechanical function may be disrupted, making it vulnerable to further damage. Therefore, the cam-type FAI mechanism can be very destructive to cartilage. However, aside from the hypothesis that this mechanism can cause excessive forces on articular cartilage, leading to articular cartilage damage, there is very little known about this disease mechanism. For example, it is unknown how much strain can develop in acetabular articular cartilage due to this mechanism. It is also unknown how factors such as varying size and contour of cam deformities would affect the strain in acetabular articular cartilage. What is known is that this mechanism can lead to tissue damage depending on the combination of hip morphology

and participation in impingement-engaging activities. Therefore, this mechanism depends on various factors including a hip morphology, activity level, and tissue durability.

Diagnosis

The diagnosis of cam-type FAI is threefold including:

1. The presence of hip pain and restricted ROM
2. A positive anterior impingement (FADIR) test
3. Radiographic imaging data confirming cam-type deformity

Cam-impingement Test

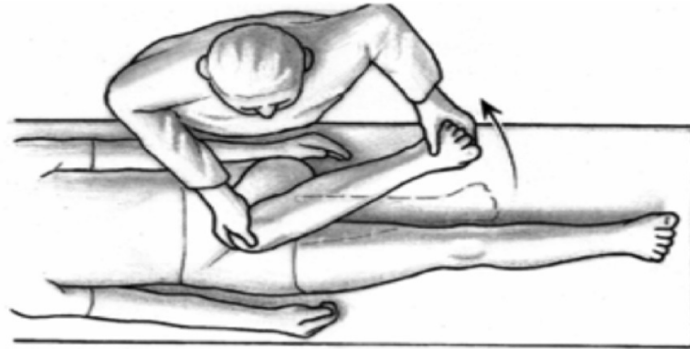


Figure 2.7: The anterior impingement test for cam-type FAI where the hip is flexed 90°, adducted, and internally rotated is demonstrated; a positive test is one that results in groin pain and/or limited hip ROM [15, 26, 27, 48].

During clinical examination of a patient with hip pain, careful history taking is conducted, the FADIR test is performed, and both a standard AP plain radiograph of the pelvis and a cross-table lateral radiograph of the affected hip are taken [15, 17, 26, 27, 48]. During history taking, it is important to understand the severity and onset of symptoms in the hip such as chronic or acute pain, stiffness, and instability and mechanical symptoms such as clicking, locking, and sharp stabbing pain. During an anterior impingement test, the patient is placed in the supine position, and the examiner passively flexes the patient's hip to 90° and forces the hip in adduction and internal rotation (Figure 2.7). This test does not simulate a functional motion. Rather, it directly brings the anterior superior areas of the femoral head neck junction and acetabulum into contact. The presence of hip pain during this manipulation consequences a positive test. This test is nearly always positive in all FAI patients (up to 99%) [10, 26].

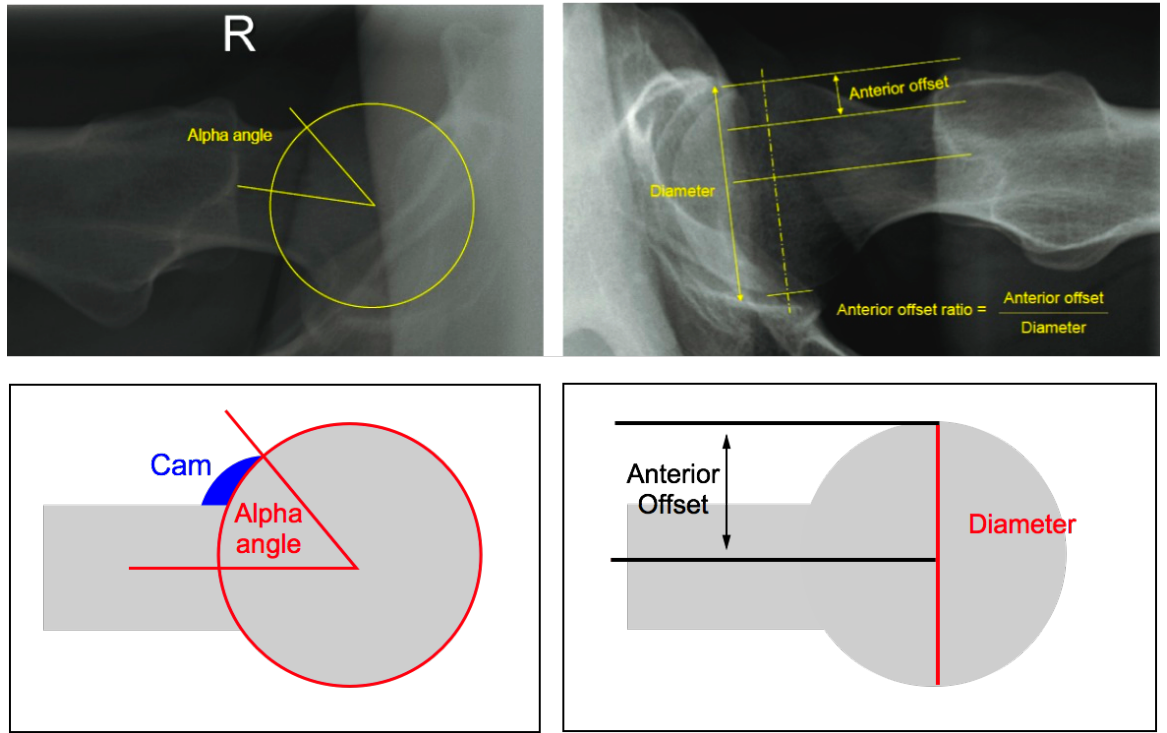


Figure 2.8: The measurement of the alpha angle (left) and anterior offset ratio (AOR = Anterior offset / femoral head diameter) (right) in a cross-table lateral radiograph are shown; these measurements are used to characterise the cam deformity [39, 40, 116, 117].

The presence of lateral cam deformities can be detected in an AP radiograph, whereas anterior cam deformities can be detected in cross-table lateral radiographs. The measurement of the alpha angle and anterior offset ratio (AOR) can be used to quantify the presence of a cam deformity [40, 116, 117] (Figure 2.8). The alpha angle is the angle that is formed when a line is drawn from the centre of the femoral neck to the centre of the femoral head and then to the point on the anterior part of the femoral head that exceeds the radius of the femoral head [40]. A study determined that the 95% confidence interval for the the alpha angle was 32 to 62° for the normal population [39]. Alpha angles greater than 55° [40] and 60° [43] have been proposed to define the presence of a cam deformity. The threshold of 65° has been proposed to indicate increased risk for cartilage damage [41], and a threshold of 78° has been proposed to indicate a pathological cam deformity [43]. The AOR is the ratio of the anterior offset to the diameter of the femoral head. The anterior offset is the distance between two parallel lines, one adjacent to the anterior aspect of the femoral neck and the other touching the most anterior part of the femoral head and both parallel to the femoral neck axis. An AOR of less than 0.15 on a cross-table lateral radiograph is an indication of a cam deformity.

Other imaging modalities such as computed tomography (CT) scans, MR imaging, and MR arthrography (MRA) can be used to view the hip joint in 3D. They are ideal imaging modalities in the diagnosis of cam deformities, but they are not a part of standard procedure when clinically examining patients who present hip pain. CT scans are useful in viewing the bony surfaces of the hip joint. MR imaging and MRA are useful in detecting the presence of articular cartilage or labral pathology. MRA is the preferred modality for investigation if labral pathology is suspected. It has also been useful in pre-operative planning for surgical treatment to detect articular cartilage and labral pathology [13, 118, 119]. It uses intra-articular contrast material, such as gadolinium, to distend the joint, separate the intra-articular structures, and provide internal contrast to delineate the edges of the labrum and articular cartilage [14]. However, even with MRA, articular cartilage delamination is difficult to detect because contrast medium will seldom penetrate beneath the delaminating articular cartilage [119, 120].

Natural Progression of Pathology

The natural progression of the hip joint pathology due to cam-type FAI is difficult to assess for two main reasons. First, there is no clear indicator of who is at risk of disease. The development of the disease depends on a person's hip morphology, activity level, and tissue durability, so simply having a cam deformity is not predictive of the development of pathology [121, 122]. Therefore, early screening of people with cam-type FAI for hip pathology is currently not possible. Secondly, patients with cam-type FAI usually seek help after the onset of symptoms such as pain. So, the state of the tissue in a patient's hip before they seek help and undergo clinical examination is unknown.

Current knowledge about the progression of hip joint pathology due to cam-type FAI has been pieced together from imaging examinations and observations during surgical operations. Early pathology includes tears within the articular cartilage, within the labrum, at the osteochondral junction, and/or at the chondrolabral junction. These soft tissue tears can be associated with various levels of fibrillation, which is when the ECM of articular cartilage splits and frays. Fraying occurs in the direction of the principal collagen fibrils [123, 124]. In a study by McCarthy *et al*, isolated labral tears were found more often in younger patients, whereas a labral tear in conjunction with chondral lesions was found more often in older patients [125]. This indicates that the development of chondral lesions may follow the development of labral tears. Also,

chondral and labral lesions often occur adjacent to one another [20, 125].

When tears develop at the osteochondral junction, articular cartilage is considered to be delaminated. The appearance of delaminated articular cartilage has been described using the “wave” or “carpet” sign [10]. It has also often been described as being “soft”, however, this description is not to be confused with the degenerative properties of cartilage softening due to a reduced stiffness [21]. If tearing of delaminated articular cartilage progresses, it can detach from the subchondral bone at one end to form a flap (Figure 2.9). This flap can detach from the subchondral bone completely, exposing the underlying bone [20]. The detached cartilage can form into intra-articular loose bodies which remain within the joint space and can become caught within the hip during movements.

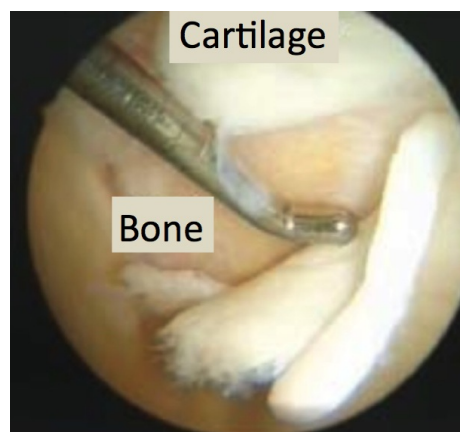


Figure 2.9: Cam-type FAI can cause the delamination of acetabular articular cartilage. The partial detachment of delaminated cartilage can form a flap, exposing the underlying subchondral bone, which is shown [126].

Overall, delamination of articular cartilage and labral lesions are the most characteristic and commonly found signs of pathology in cam-type FAI patients [20, 21, 22, 23, 24, 50]. Cartilage delamination can lead to the detachment of cartilage, exposing underlying bone, so it is thought to lead to hip OA and is therefore of greater concern.

Treatment

Reshaping the femoral head, using surgical approaches such as hip arthroscopy or open hip dislocation, is a common treatment for symptomatic FAI patients (Figure 2.10). These operations aim to increase hip ROM, prevent further tissue damage, and to reduce hip pain in these patients. Hip arthroscopy is key-hole surgery on the hip joint where procedures are performed

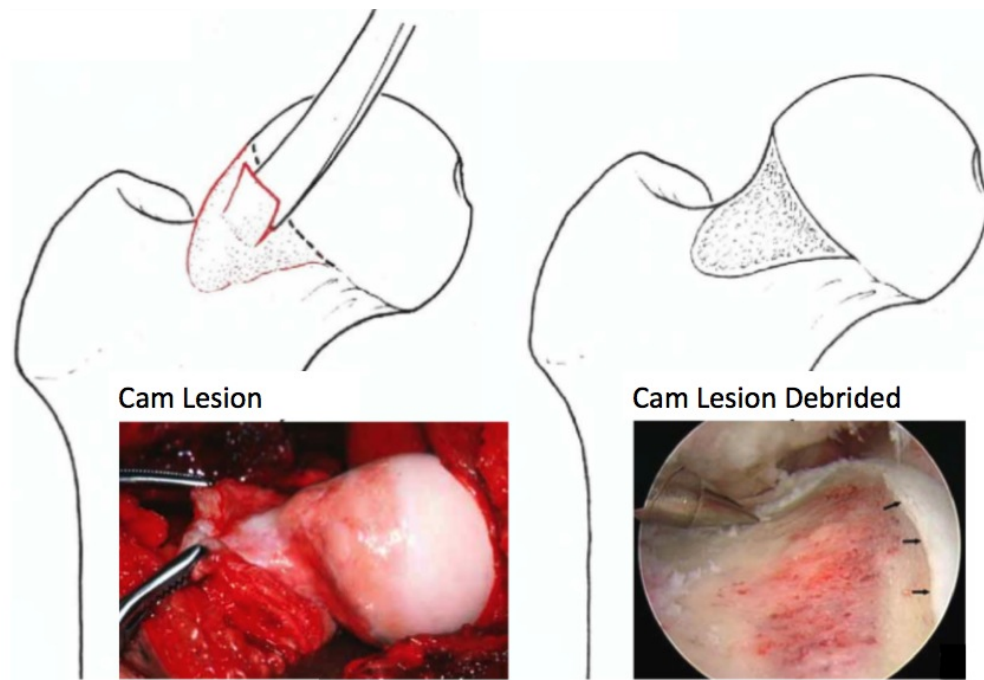


Figure 2.10: The femoral head with a cam lesion (left) and debrided cam lesion (right) during open surgery are shown [25].

through a small incision in the hip, and a camera is used to visualise the inside of the joint. Open hip dislocation is open surgery where the hip is dislocated, exposing articular hip joint surfaces.

For treatment of cam-type FAI patients, femoral osteochondroplasty is a common procedure during these operations. Femoral osteochondroplasty aims to remove the cam deformity. Upon removal of the cam deformity, a deeper cut is made into the femoral head so that additional underlying bone is also removed, creating greater clearance for the acetabular rim during end-stage ROM. Other various procedures can be conducted during these operations depending on the patient's specific pathology. These procedures include debridement of a labral tear, repair of a torn labrum, chondroplasty of acetabular lesions, drilling or microfracture of acetabular lesions, acetabular rim excision, debridement of acetabular chondral delamination, and capsular excision [6].

Since the concept of FAI has only been well established in the last 15 years, there are no long-term outcomes of surgical treatment available. However, these operations have been reported to have resulted in positive short-term outcomes. In a systemic review [24] of surgical outcomes for FAI patients, hip arthroscopy produced good to excellent outcomes in a range of 67 to 93% of patients (n=271) with a mean follow-up of 2.2 years. About 1.5 years after surgery, 93%

of patients were able to return to sport. Procedure failure, based on patient dissatisfaction or the conversion to total hip replacement, occurred in 1 to 33% of patients. Open surgical procedures were considered to have produced good to excellent outcomes in a range of 65 to 94% of patients (n=194) with a mean follow-up of 3.3 years. Procedure failure occurred in 3 to 40% of patients.

The study with the longest follow-up to date was conducted by Byrd *et al* [25]. A 10-year follow-up in 29 patients (31 hips) that underwent arthroscopic resection of a labral tear was taken. At 10 years, 83% of patients without arthritis at the time of surgery had an 18-point or greater improvement in the Harris Hip Score [25]. A large portion of patients (88%) who had arthritis at the time of surgery had to undergo a THR after a mean duration of 5.25 years [25]. This shows that surgical outcomes were less favorable in patients with more developed pathology.

There are also other studies which show less favorable surgical outcomes due to more advanced hip pathology. For example, patients with signs of OA of grade II or above according to the Outerbridge classification [127], or patients with chondral tears or delamination have shown less favorable outcomes [54, 55].

Overall, positive short-term results of less than 10 years for FAI patients with limited OA and chondral pathology are encouraging. However, because the natural history of OA often takes place well over the course of 10 years, more extensive follow-ups are required to determine the effectiveness of hip arthroscopy or open hip dislocation as long-term treatments [25].

2.5 Articular Cartilage Strain and Damage

Articular cartilage's specialised structure has given rise to its durability to repetitive loading and resilience to compressive, tensile, and shear stresses. Its superficial, middle, and deeper layers each have unique composition, structure, and material properties to resist different types of loading and stresses. However, it has been suggested that damage to any part of articular cartilage's structure can disrupt its load carrying and distribution mechanisms, at which point, normal loading can possibly lead to further damage within its structure [128, 129].

Damage such as tears or cracks in articular cartilage have been found in all layers of articular

cartilage – the superficial and middle layers [93, 130] and deep layers at the cartilage-bone interface or tidemark [130, 131, 132]. Such damage has been attributed to the rapid loading of articular cartilage at fast strain-rates and large tensile or shear strains [92, 93, 130, 131, 132]. It has been determined that at rapid loading and fast strain-rates, articular cartilage behaves more similarly to a linear elastic material [90, 91, 92, 93]. Also, damage such as cracks represent the brittle-like failure mechanism of elastic materials. Therefore, the material behaviour of articular cartilage at fast strain-rates is consistent with its possible brittle-like failure mechanism at fast strain-rates.

Since the structure of each layer of articular cartilage varies, its resistance to tensile and shear strains varies in each layer as well. Tensile and shear strains are associated with the tensile and shear strains of articular cartilage’s collagen network and collagen fibrils. Thus, the failure mechanism and resistance to failure in each layer of articular cartilage is dependent on the structure of its collagen fibres. In cam-type FAI patients, cracks at the tidemark or at the cartilage-bone interface are particularly common [8, 9, 118, 120, 133]. Collagen fibres are aligned radially to the cartilage-bone interface, anchored at the tidemark, and resist shear stresses when articular cartilage is loaded in compression or shear.

There is evidence that the earliest change in the collagen network due to mechanical overloading is loosening or breaking of the links between collagen fibrils due to shear strains along the collagen fibrils [128, 129]. These links can be entwinements of the collagen fibrils or nonentwinement-based interactions. Breaking these links in the lateral direction (links that provide interconnectivity between fibrils in the direction parallel to the tidemark) has been shown to result in more fibrillar cartilage structure with bundles of fibrils that are relatively unentwined and, therefore, strongly aligned in the radial direction and closely packed [134]. Normal cartilage, in contrast, consists of less discrete bundles of fibrils that are more randomly structured and have higher interconnectivity laterally [134].

There is also evidence that further straining of the collagen network can lead to more severe degenerative changes, where the collagen fibres themselves break due to fibril strains [128, 129]. Mathematical calculations of strains in a rapidly loaded cartilage-bone layer has determined that principle stresses and strains are greatest in planes oriented 45° to the cartilage-bone interface [135, 136]. Using elastic fracture mechanics, it has been suggested that cracks are most likely

to occur near the cartilage-bone interface oriented 45° to the interface where shear strains are greatest [135, 136]. Finally, there is evidence that a crack oriented 45° to the interface is likely to propagate in the direction along the interface [135, 136]. Finite element analyses (FEA) computing the distribution of maximal shear stresses have confirmed that isotropic models of articular cartilage can explain the formation of cracks along the tidemark [129, 137]. Based on current evidence and knowledge about the loading behaviour of articular cartilage the following series of events may be responsible for the initiation and propagation of a crack at the tidemark or cartilage-bone interface:

1. Rapid loading of articular cartilage
2. Alignment of collagen fibres along direction of maximum principle shear strains
3. Failure of interactions and entwinements between collagen fibres due to shear strains along the fibres
4. Further rapid loading of articular cartilage
5. Alignment of collagen fibres along direction of maximum principle shear strains
6. Failure of less entwined collagen fibrils due to increased strains along fibrils
7. Further rapid loading of articular cartilage
8. Propagation of failure of collagen fibrils in direction along interface

2.6 Modelling Damage and Failure

In order to simulate cartilage delamination, it is important to understand some of the available methods used to model the damage and failure of materials. Damage of a material in this context refers to the higher propensity of the material to fail. Failure of a material refers to the formation of a crack where there is a separation in the material where new surfaces have been created. Two common methods to model the initiation and propagation of damage and failure of materials are Cohesive Zone Modelling (CZM) and Linear Elastic Fracture Mechanics (LEFM).

2.6.1 Cohesive Zone Modelling (CZM)

Cohesive Zone Modelling (CZM) is a technique to simulate crack initiation and crack growth which uses cohesive laws. CZM models fracture as a gradual phenomenon where separation takes place across an extended crack tip (cohesive zone) and is resisted by cohesive tractions [138]. In order to model crack initiation and crack growth, traction-separation laws are used to describe the opening up of the cohesive elements. These laws describe tractions as a function of separations and determine the constitutive behaviour of CZMs. There are two main constitutive behaviours in CZMs. First, as cohesive surfaces separate, the traction between the surfaces increase until a maximum is reached. Secondly, once this maximum is reached, the traction decreases to zero which results in complete local separation of the cohesive surfaces. These two behaviours can be described by different functions (curves). The area under these traction-separation curves correspond to the energy needed for separation[139].

2.6.2 Linear Elastic Fracture Mechanics (LEFM)

Fracture mechanics is the study of the propagation of cracks in materials. It involves using analytical solid mechanics to calculate the driving force on a crack and experimental solid mechanics to characterise the material's resistance to fracture. The field of fracture mechanics can be divided into several categories based on a materials toughness including linear elastic fracture mechanics (LEFM), elastic plastic fracture mechanics (EPFM) and, plastic fracture mechanics (PFM). The field of interest in fracture mechanics for this study is LEFM.

Although it is not fully understood exactly how the crack surfaces are formed during cartilage delamination, it is understood that cartilage behaves similar to linear elastical material during rapid loading. Assuming that delamination occurs during rapid loading, the concepts of LEFM are probably the most appropriate field of fracture mechanics to describe delamination.

LEFM is typically used to analyse fracture in isotropic linear elastic materials with low toughness that exhibit brittle fracture. A brittle fracture is associated with fracture where very little plastic deformation is involved, which means the energy around a crack tip goes into creating new crack surfaces rather than deformation of the material around the crack tip. LEFM also assumes that a material is isotropic and linear elastic. Isotropic materials by definition have materials properties that are independent of direction. These materials can be described using

two elastic constants – the Poisson’s ratio and the Young’s modulus (the bulk modulus or shear modulus can alternatively be used). Linear elastic materials deform linearly which means their Young’s modulus (ratio of stress over strain) is constant over a range of strains, obeying Hooke’s law ($F = kx$). [140]

Based on the assumptions of an isotropic and linear elastic material, the stress field near a crack tip can be calculated using the theory of elasticity. So, when stresses near the crack tip exceed the material’s fracture toughness, the crack will grow. A material’s fracture toughness can be represented by stress intensity factors which are a function of stress and crack length. The stress intensity at a crack tip is equivalent to the strain energy release rate (SERR), which is the energy dissipated or released during fracture per unit of newly created crack surface area. This means the energy used to form a crack must be balanced by the energy dissipated from the formation of crack surfaces. When a critical stress intensity (fracture toughness) is reached at a crack tip, the crack will propagate [140].

2.7 Finite Element Analysis (FEA)

Finite element analysis (FEA) is a numerical method for predicting how real world objects (solid or fluid) will react to externally applied physical phenomena such as mechanical, thermal, dynamic, or electrical loads. These analyses for simple cases can be directly solved using differential equations. However, for real world applications, these analyses may involve complicated geometries and boundary conditions where multiple differential equations are involved and cannot be directly solved. The finite element method addresses this complication through its key concept in breaking down objects into smaller pieces. In this way, simple equations can be solved for each smaller piece and combined to approximate complex differential equations that describe the behaviour of the entire object’s geometry.

2.7.1 Finite Element Method (FEM) Overview

Overall, the finite element method (FEM) consists of the following main steps:

1. Discretisation of objects
2. Application of boundary conditions
3. Solution analysis

4. Calculation of additional field values

Step 1: Discretisation of objects

Through FEA, a geometry can be broken down into 1-dimensional (1D), 2-dimensional (2D), or 3D shapes such as bars in 1D, triangles and quadrilaterals in 2D, and tetrahedrals and hexahedrals in 3D. These shapes are called elements, and the vertices at which elements are connected are called nodes. The combination of elements and nodes creates a mesh where the density of a mesh can highly affect the accuracy of the analysis. Typically, the refinement of a mesh (increasing of the density of the mesh) can increase the accuracy of the analysis up until a certain point where any further refinement will lead to negligible changes in the results. A study where the mesh of an object is incrementally changed to detect the threshold mesh density for which the analysis results will reach an equilibrium is called a mesh refinement study. [1]

Step 2: Application of boundary conditions

Boundary conditions include prescribed loads, displacements, and rotations in a model. It is important to have an inherent understanding of the problem being modelled in order to accurately apply such boundary conditions. They are important to define how loading is applied and resisted by constraining displacements and rotations of certain nodes. These displacements and rotations are referred to as degrees of freedom (DOF). A node in 1D has one DOF (one direction of displacement), three DOF in 2D (two directions of displacement and one direction of rotation), and six DOF in 3D (three directions of displacement and three directions of rotation) [1].

Step 3: Solution analysis

After a model is setup, a solution is performed to calculate field values, such as displacements (x), at the nodes. As an overview, the general solution approach is to calculate field values at integration (Gauss) points located within each element through numerical integration. Each element type has uniquely specified equations, called shape functions, which define how field values vary across the element. Therefore, shape functions are used to extrapolate field values calculated at integration points to the nodes of the elements. Since many elements are connected

at each node, the field values are averaged at the nodes. [1]

There are two main approaches used in FE to solve these solutions – implicit (or standard) or explicit. These solution approaches are made to analyse different types of models. Their differences have been summarised in Table 2.1.

	Implicit Analysis	Explicit Analysis
Used for Modelling	Static, low-speed dynamic, or steady-state transport analyses	Brief transient dynamic events, or severely nonlinear behaviour (e.g. contact)
Equation Solved	$F = kx$	$F = ma$
Solution Method	Direct-integration by Hilber-hughes-taylor operator	Direct-integration by Central-difference operator
Solution Steps	1. k is calculated/inverted to solve for x ($x = k^{-1} * F$)	1. a is solved for ($a = m^{-1} * F$)
	2. Convergence of solution is checked (e.g. forces, displacements)	2. $v_{i+1/2}$ is calculated using a_i (assuming a_i is constant over whole increment)
	3. If solution hasn't converged, new iteration conducted with new x	3. x_{i+1} is calculated using $v_{i+1/2}$ (assuming v_i is constant over whole increment)
Time Increments	Many iterations performed over a few increments	Many increments used to step through analysis step
Solution Time	Determined by size of k (based on the number of elements in the model)	Determined by minimum characteristic increment size ($\Delta t = \min(L_e/c_d)$) (based on element size L_e and wave propagation speed through the material c_d)
Solution Stability	Inherently Stable (convergence checked at each increment)	Conditionally stable (based on increment size, no checks at increments)

Table 2.1: Comparison of implicit and explicit analyses for solving solutions of FE models [1].

Step 4: Calculation of additional field values

Once the displacement at every node has been calculated, additional field values such as strain and stress can be calculated.

2.7.2 Modelling Damage and Failure using FEA

The damage and failure material modelling methods of CZM and LEFM can be implemented in FEA. CZM can be implemented through cohesive elements, cohesive surfaces, or through the extended finite element method (XFEM) with cohesive segments. LEFM can also be implemented in FEA through XFEM to define the initiation and propagation of cracks.

Cohesive Elements and Surfaces

Cohesive elements are elements that do not represent any physical material; they are placed between continuum elements to describe the cohesive forces which occur when material elements are being pulled apart [139]. Cohesive surfaces are surfaces where cohesive-based surface behaviour has been applied [139]. Therefore, cohesive elements and surfaces have to be created and defined in the geometry pre-processing step of the FEM [139]. This means it must be understood what elements and surfaces in the model can possibly separate and form a crack before the solution of the model can be undertaken.

Cohesive elements can be implemented in FEA by defining the initiation of damage and evolution of damage in the bulk material definition (Abaqus 6.11, Dassault Systems; Velizy-Villacoublay Cedex, France). Cohesive surfaces can be implemented in FEA through the same material bulk definition in addition to defining a cohesive behaviour surface interactions. The initiation of damage can be defined using a number of criterion based on stress and strain such as maximum principle strain. This criterion defines the maximum traction reached before the traction between cohesive surfaces begins to decrease. The evolution of damage defines the rate at which the cohesive stiffnesses degrade once the initiation criterion has been reached. This rate is described by a scalar damage variable (D) which represents the overall damage at the intersection between cracked surfaces and edges of cracked elements. This value is zero until the maximum traction is reached, at which point it can evolve to one, which means complete failure has been reached. [1]

Extended Finite Element Method (XFEM)

The extended Finite Element Method (XFEM) models discontinuities such as a crack as an enriched feature. This means enrichment (phantom) nodes are added to the model and are described by additional functions called enrichment functions. If the phantom nodes are near a crack tip, the enrichment functions consist of asymptotic functions that capture singularity around the crack tip. If the phantom nodes are across crack surfaces, the enrichment functions consist of discontinuous functions that represent a jump in displacement. [1]

XFEM makes modelling cracks easier and more accurate by allowing cracks to be modelled independent of the mesh and by allowing the simulation of the initiation and propagation of discrete cracks along an arbitrary, solution-dependent path. This means very refined meshes around cracks are not required, and neither is remeshing the model when the crack grows, which saves substantial time in model pre-processing and solution computation. XFEM does this by not considering near-tip asymptotic singularity, so a crack has to propagate across an entire element at a time to avoid the need to model the stress singularity. [1]

There are two types of damage modelling within an XFEM framework – a cohesive segments approach and a LEFM-based approach. The cohesive segment approach uses traction-separation laws of CZM and follows the general framework for surface-based cohesive behaviour to define damage properties where damage initiation is defined by an initiation criterion which can be based on energy, strain, or stress. The evolution of damage can be defined by either the material's fracture energy (area under traction-separation curve) or displacement (maximum separation at which the material fails and a crack forms). Once a crack forms, when it is closed, a pressure-closure contact relationship governs its behaviour. If the crack is open, cohesive behaviour contributes to the contact normal stress.

LEFM-based XFEM uses the laws of LEFM. A damage initiation property is defined in the bulk material definition, and the fracture criterion based on a virtual crack closure technique (VCCT) is assigned through an interaction property. Once the damage initiation criterion has been met, the fracture criterion is activated where VCCT calculates the critical strain energy release rate (SERR) at the crack tip. If the SERR calculated at the crack tip exceeds this critical SERR value, the crack will propagate.

2.8 Current Methods for Characterisation of the Cam Deformity

Characterisation of the cam deformity has not ventured very far from a radiographic measurement of the alpha angle. The measurement of the alpha angle is typically taken from a single radiograph, however it has been shown that measuring the alpha angle in 3D across a range of anatomical planes can lead to measurements that are greater than radiographic measurements [39, 40, 41, 43, 46]. Two unique methods have been used to characterise the cam deformity. One method consists of segmenting raw DICOM data to determine the distance from the cam lesion's surface points to the centroid of the femoral head in order to calculate mean bump height (cam protrusion), volume, and location [45]. This method was used to estimate the range of cam protrusion in 5 subjects to be 0.78 to 2.06 mm with a mean of 1.54 mm [45]. Another method consists of statistical shape modelling (SSM) which enables the comparison of 3D geometries without the need for assuming ideal geometry [47]. This method has been used to estimate the mean protrusion of a cam deformity above the control mean to be a maximum of 3.3 mm with sustained protrusions of 2.5 to 3.0 mm along the anterolateral head-neck junction and distal anterior neck [47].

2.9 Previous Modelling of Cam-type FAI

Very few FE models of cam-type FAI have been published. A 3D FE model with idealised cam shape utilising a range of alpha angles focused on calculating peak stresses in the hip joint and showed that peak stresses formed in clinically affected areas of the hip [141]. Two studies have created 3D FE models which utilised subject-specific geometry [142, 143]. One FE model was used to examine maximum contact pressure at four chosen nodes in the hip joint during flexion at 90°, 24° of internal rotation, and internal rotation at 30° of flexion [142]. Another study used FE models to calculate peak maximum shear stress during standing and squatting in the cartilage and bone of two cam-type FAI patients and two normal subjects. The bone in the cam-type patients produced peak maximum shear stresses that were 3 to 4 times larger than the bone of the normal subjects during the motion of squatting. All other calculated values were comparable between the normal subjects and cam-type patients [143]. All of these studies focused on comparing measured stresses to that of calculated stresses in normal patients and aimed to show that stresses were greater in cam patients in the anterolateral area of the

hip joint during motions of large flexion or internal rotation. None of them have introduced possible detailed mechanisms for which labral tears or cartilage damage may occur or look at shear strain, confirming the need for the development of this thesis.

Chapter 3

Development of Geometric-kinematic Hip Joint Models: A Study of How Motion and Geometry Affect Cam-type FAI

3.1 Introduction

The development of femoroacetabular impingement (FAI) is highly dependent on a patient's specific hip joint motion and activity level, hip joint surface geometries, and tissue durability [7, 10, 12, 26, 112, 113, 114, 115]. This is supported by the lower prevalence of FAI compared to the prevalence of anatomical features that lead to FAI in the population. The prevalence of deformities related to cam- and pincer-type FAI has been found to be up to 66%¹ [110, 111], and the prevalence of cam-type deformities has been found to be between 24 to 29% in men [30, 31, 32, 33] and 6 to 12% in women [30, 32, 33], whereas the prevalence of FAI (cam- and pincer-type) in the population is only about 10% [11]. The portion of the asymptomatic population with cam deformities that may never develop cam-type FAI in the future, could possibly remain symptom-free because they have highly durable tissue properties, do not subject their hip joints to extreme impingement-inducing ROM, or their specific acetabular surface geometry allows for clearance of their cam deformity at impingement-inducing ROM. The lattermost case suggests that an individual's hip joint motion and hip joint geometry may have a complicated coupling effect which may determine the presence or absence of cam-type FAI.

¹It is important to note that the 66% prevalence of FAI-related deformities was based on the finding of one abnormal radiological measurement. This prevalence has been found to dramatically decrease to about 29% when considering two or more abnormal radiological measurements [111].

This hypothesis has led to the creation of geometric-kinematic models which combine hip joint motion and geometry in order to study their effect on cam-type FAI.

In this chapter, geometric-kinematic hip joint models which applied motion to a set of hip joint geometries from cam-type FAI patients and normal subjects were created. The motions of the impingement test and the functional activities of walking, sitting, and standing were chosen because they have been demonstrated to lead to pain for cam-type FAI patients. The impingement test involves internal rotation at 90° flexion, walking involves normal walking at a normal pace, sitting involves the motion of sitting down from a standing position, and standing involves the motion of standing up from a sitting position. The motion of the impingement test involves a very large hip ROM, which causes the cam deformity to come into closer proximity with the acetabulum and has shown to cause pain in up to 99% of patients [10, 26]. The functional activities of walking, sitting, and standing involve much smaller ROM's, so it is debatable whether impingement occurs during these activities. However, patients often complain about groin pain being exacerbated during such activities, so they are an important part of this investigation.

To analyse the output of the geometric-kinematic hip joint models, the proximity between the surfaces of the acetabulum and femoral head were calculated. The motivation of using proximity analysis stemmed from the underlying mechanism of cam-type FAI – the cam deformity is thought to affect the soft tissues in the hip joint by being forced into the hip joint, causing excessive shear forces. Thus, the cam-type FAI mechanism requires the cam deformity to be in greater proximity with the surfaces of the acetabulum compared to the proximity of the femoral head and acetabulum in normal subjects.

Overall, these geometric-kinematic hip joint models will give insights into how the presence of a cam deformity may affect the proximity between the surfaces of the acetabulum and femoral head in the hip joint. In particular, insights into which properties of the cam deformity may potentially be most damaging to soft tissues in the hip joint will assist in the development of future simulations and parametrised models of the cam-type FAI mechanism. Finally, for the purposes of consistency, the motions and geometries of the left hip (versus the right hip) were arbitrarily chosen to be acquired and analysed using geometric-kinematic models.

3.2 Methods

3.2.1 Processing of Motion Data

The motion or kinematics of the hip joint can be described by the individual rotations of the pelvis and femur over time. These rotations are referred to as cardan angles when they are described by the three anatomical planes – coronal, sagittal, and transverse (Figure 3.1). The three cardan angles for the pelvis and hip are described as pelvic obliquity, tilt, and rotation and hip abduction/adduction, flexion/extension, and internal/external rotation, respectively [144] (Figure 3.1).

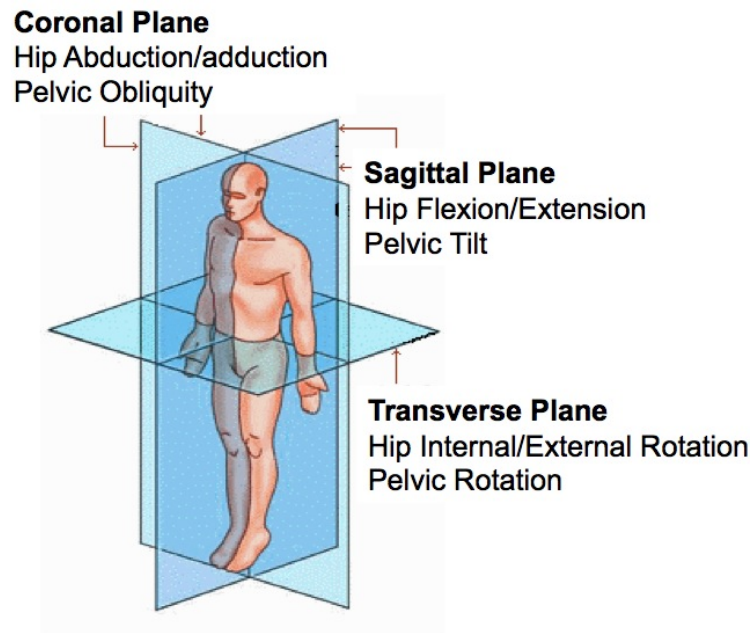


Figure 3.1: The pelvic and hip cardan angles which lie in the three anatomical planes of coronal, sagittal, and transverse are presented [58].

The kinematics of the hip joint were captured using motion analysis at the Oxford Gait Laboratory (Figure 3.2), and the above described cardan angles were calculated based on the acquired motion analysis. During motion analysis, the spatial position of the pelvis and femur were recorded using numerous markers. Three markers were chosen to track the pelvis and femur each (Figure 3.2). The acetabulum is rigidly fixed to the pelvis, and the femoral head is rigidly fixed to the femur, thus, tracking the pelvis and femur directly translated into tracking the acetabulum and femoral head, respectively.

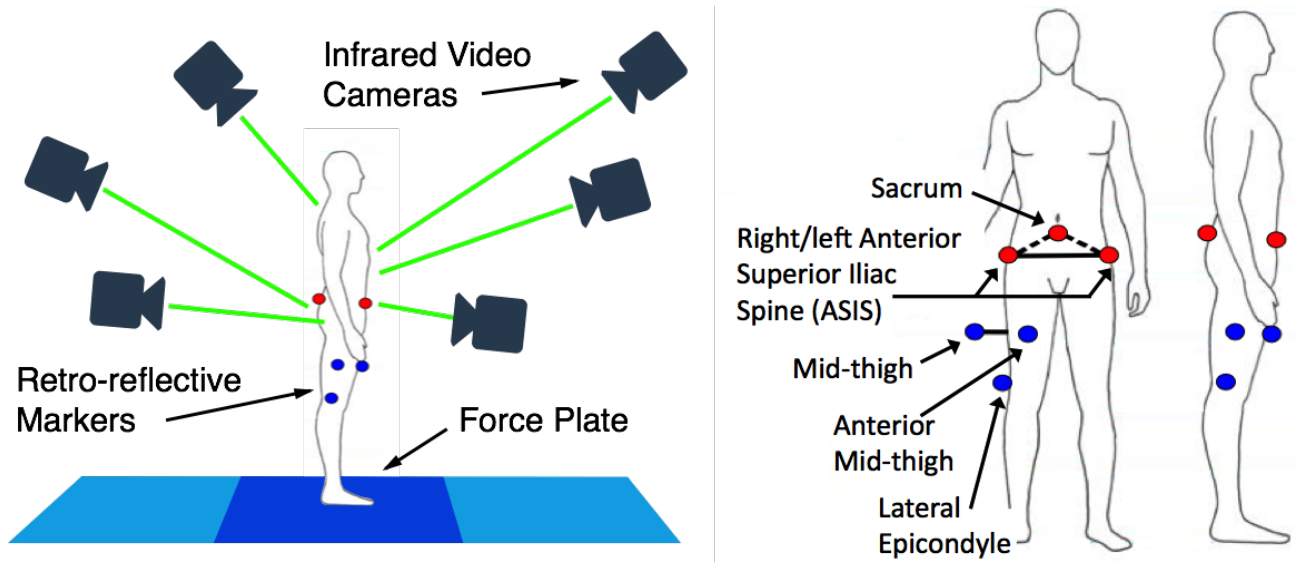


Figure 3.2: A schematic of the gait laboratory setup (left) and the marker distribution used to track the pelvis and femur indicated by three red and three blue dots, respectively (right), are depicted [145].

Impingement Test

The motion of the impingement test was chosen to range in flexion from 0 to 120°, internal rotation from 0 to 30°, and adduction. These ranges were chosen based on the hip ROM for normals [7, 26, 113, 114]. It is important to note that the impingement test usually only involves flexion of up to 90° and may involve internal rotations with ranges less than 30°, however, extreme ROM's were applied to investigate the resulting effect on the proximity of surfaces in the hip joint.

Walking

Walking data were acquired from 20 normal female and 20 normal male subjects with mean ages of 24.7 (3.3) and 28.2 (7.2) years and mean body mass of 61.4 (8.0) and 71.8 (12.6) kg, respectively, using a state-of-the-art motion analysis system (Vicon 612, Vicon Motion Systems Ltd., Oxford, UK).² The average and standard deviation of the three hip angles, flexion/extension, ad/abduction, and internal/external rotation, were calculated and automatically filtered using a Woltring quintic spline filter [146] (AnyBody Version 3.0, AnyBody Technology A/S, Aalborg, Denmark) (Figure 3.3).

²Data was acquired at the Oxford Gait Laboratory.

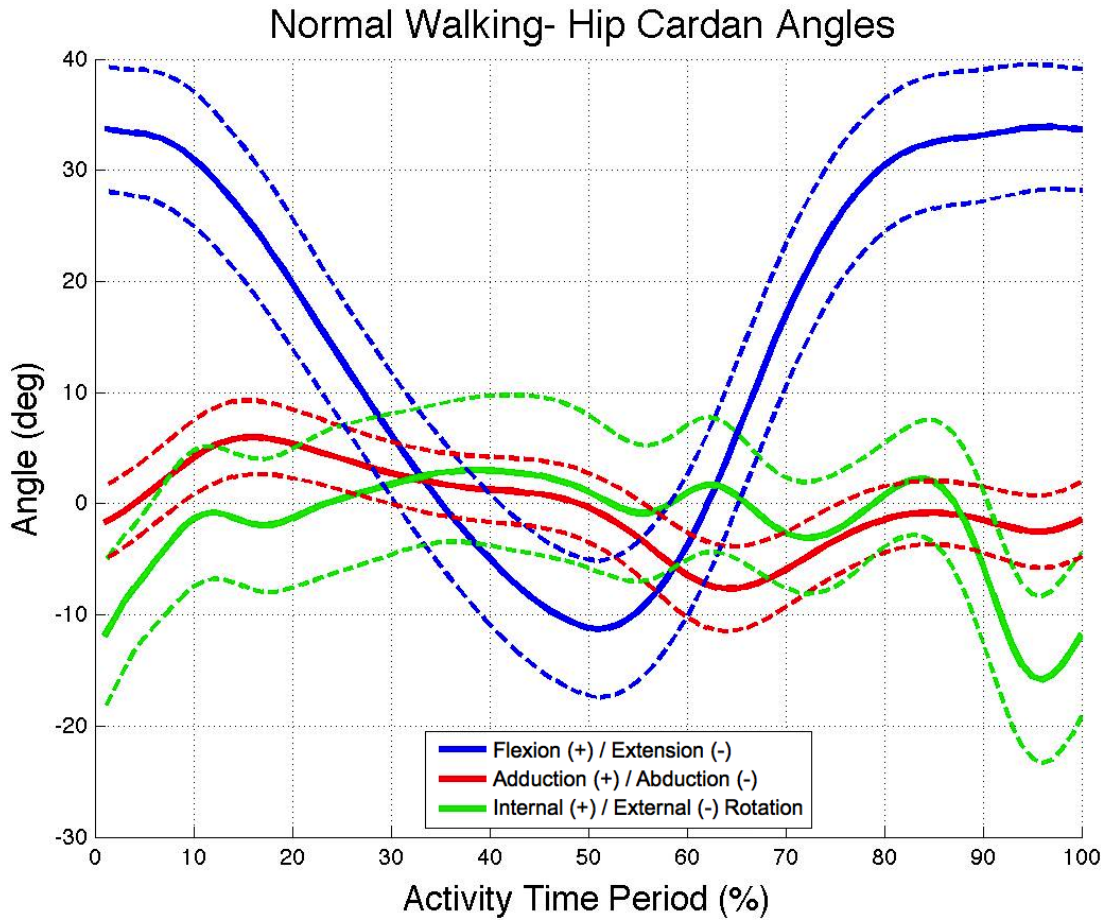


Figure 3.3: The average and standard deviations of hip angles of 20 normal female and 20 normal male subjects during walking were calculated.

Sitting and Standing

Sitting and standing data were acquired from two female hip resurfacing patients and three male hip resurfacing patients with mean ages of 46.0 (24.0) and 58.0 years and mean body mass of 55.4 (3.0) and 78.0 (3.7) kg, respectively (age data was missing for one female and one male patient).³ These patients had unilateral hip resurfacing. The use of their hip kinematics to model normal sitting and standing motion was thought to be suitable since the dominating angle of movement, flexion, during sitting and standing appeared normal in these patients. Patients with left unilateral hip resurfacing were chosen because more accurate kinematic information could be acquired using the implant's coordinate system determined by CT scanning.

The average and standard deviation of the three hip angles were determined through transformation calculations using the femoral, pelvic, and implant coordinate systems (Matlab R2010a, Math Works, Inc., Natick, MA) (Figure 3.4). Flexion/extension was calculated based on the

³Data was acquired at the Oxford Gait Laboratory.

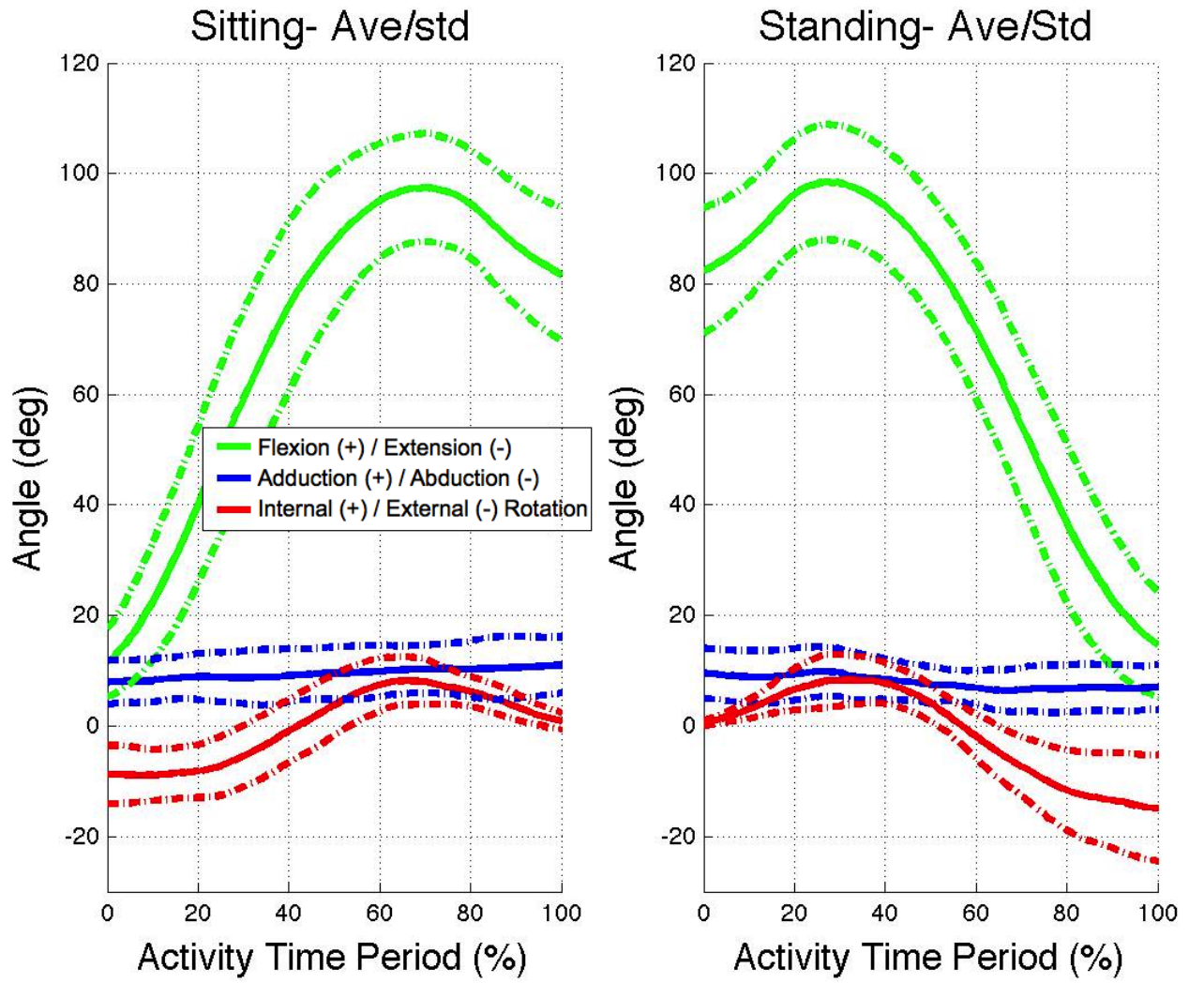


Figure 3.4: The average and standard deviations of hip angles of five hip resurfacing patients during sitting (left) and standing (right) were calculated.

rotational transformation from the pelvic coordinate system to the femur coordinate system about the y-axis. Ad/abduction was calculated based on the rotational transformation from the pelvic coordinate system to the femur coordinate system about the x-axis. Internal/external rotation was calculated based on a rotational transformation from the implant's cup coordinate system to the implant's ball coordinate system. The implant's coordinate system was used for hip rotation instead of using a rotational transformation between the pelvic and femur coordinate systems because hip rotation occurs about the long axis of the femur which has the greatest potential for measurement error. The implant was implanted at a different orientation in each subject, so the implant's cup and ball coordinate systems were not consistently defined across all subjects. Therefore, the calculation of internal/external hip rotation based on rotational transformations between the implant's cup and ball coordinate systems was subject-specific. To choose the correct rotational transformation for each subject, all the rotational transfor-

mations between the three axes of the implant’s cup coordinate systems to the three axes of the implant’s ball coordinate system were conducted (total of nine). The motion of flexion was easily identifiable during the activities of sitting and standing, so the transformation that represented flexion was first identified. This led to the understanding of the relative orientation of the implant’s cup coordinate system to the implant’s ball coordinate system and enabled the correct rotational transformation that represented internal/external rotation to be identified and used.

Hip angles from three trials for each subject for each activity were calculated and standardised by converting their activity time periods into percentages. Flexion dominated both of these activities, so it was important to standardise the hip angles by the peak flexion angle to prevent losing information after averaging across subjects and their trials. Therefore, the activity time period was adjusted so that the peak flexion angle occurred at 70% and 30% of the activity time period for sitting and standing, respectively.

3.2.2 Processing of Geometric Data

Bony hip joint geometry was acquired from five normal subjects and five cam-type patients using a 6-slice CT scan (LightSpeed VCT, GE Healthcare, Milwaukee, Wisconsin) with a 0.79 mm x 0.79 mm pixel size.⁴ Density-based automated segmentation was performed on these geometries (Mimics 12.5, Materialise NV, Leuven, Belgium) [114].

The CT scans were taken with the subject in the supine position and with their lower limbs in a neutral position with no hip flexion/extension, ad/abduction, or internal/external rotation. However, errors in the positioning of the subjects’ lower limbs may have caused the hip to vary from the neutral hip position. Also, the orientation of the subject with respect to the scanner may have been angled out of alignment with the neutral pelvic position. Therefore, the pelvic and hip orientation of each subject was manually adjusted into neutral position using anatomical landmarks and background knowledge of the anatomy of the pelvis and hip (Geomagic Studio 11, Geomagic, Inc; Research Triangle Park, North Carolina, USA). When the geometries were imported into Geomagic, the hip joint centre (HJC) was determined by fitting a sphere onto the femoral head and using the centre of the fitted sphere to be the HJC (Figure 3.5). The hip was translated so that the HJC was at the origin of the coordinate system.

⁴Data was acquired at the University of Ghent at the Faculty of Medicine and Health Sciences.

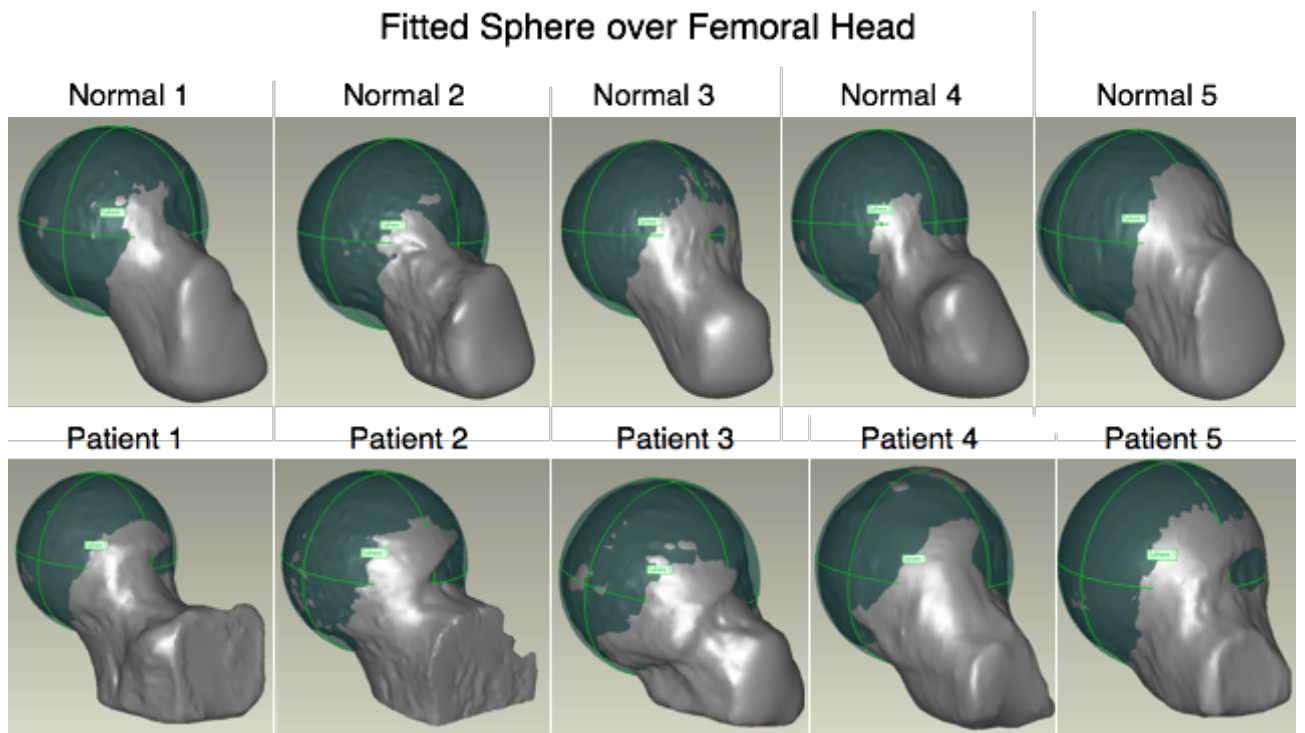


Figure 3.5: Spheres were fitted over the femoral heads of normal subjects (top) and cam-type FAI patients (bottom) to determine the HJC of the normal subjects and patients which was estimated to be the centre of the fitted sphere (Geomagic Studio 11).

Then, the following protocol was followed to orient the pelvis and hip into neutral position:

Orienting the pelvis to obtain a neutral pelvic position:

1. The pelvic obliquity was oriented so that the iliac crest of the left and right hip were in the same transverse plane.
2. The pelvic rotation was oriented by centring the pelvic bone and sacrum with respect to the same sagittal plane.
3. The pelvic tilt was oriented so that the anterior superior iliac spines and the pubic tubercles were in the same frontal plane.

Orienting the femur to obtain a neutral hip position:

1. Hip rotation was oriented so that the line drawn through the centres of the left and right femoral condyles lay in the frontal plane.
2. Hip flexion was oriented so that the line drawn from the HJC through the centre of the shaft of the femur was aligned with the frontal plane.

3. Hip adduction was oriented so that the line drawn from the HJC to the centre of the femoral condyles was aligned with the sagittal plane.

The geometry of the pelvis and femur were cropped around the left acetabulum and femoral head (Mimics 12.5) and edited further to reduce the surfaces to only the acetabulum and femoral head (Geomagic Studio 11). The rotational adjustment of the geometries into the neutral positions previously determined were applied to the edited geometries. Finally, these geometries were exported as Virtual Reality Modelling Language (VRML) files, a standard file format for representing a three-dimensional polygon by its vertices and edges (described by the triangulation of the vertices) to later be used in the creation of geometric-kinematic hip joint models.

The surfaces of the femoral heads were also analysed to study the presence of a cam deformity and the coverage and protrusion of the deformity (Figure 3.6). This was done by calculating the radial distance of the surfaces of the femoral head from the HJC. This radial distance was represented in terms of the femoral head's radius and the protrusion of the surface beyond this radius. Protrusions of up to 1 mm at the femoral head-neck junction seemed to be consistent among normal subjects and cam-type FAI patients. Protrusions of more than 2 mm showed more coverage at the femoral head-neck junctions of Patients 1 to 5 and Normal Subject 5. The coverage in Normal Subject 5 was more lateral, while the coverage among Patients 1 to 5 was more anterior.

Overall, Patients 1, 2, 4, and 5 had cam deformities that lay more anteriorly on the femoral head-neck junction and also had spreading coverage superiorly onto the femoral head, while Patient 3 had a very large cam deformity, with its greatest proximity located anteriorly on the femoral head-neck junction, that did not spread as superiorly onto the femoral head as in the other patients (Figure 3.6). Normal Subjects 1, 2, and 4 had small protrusions of less than 1 mm that covered a small area anteriorly on the femoral head-neck junction. Normal Subject 3 had a wider spreading protrusion that ran anteriorly to laterally, and Normal Subject 5 had a larger protrusion of up to 2 mm that lay laterally.

Radial Distance of Femoral Head Surfaces from Hip Joint Centre

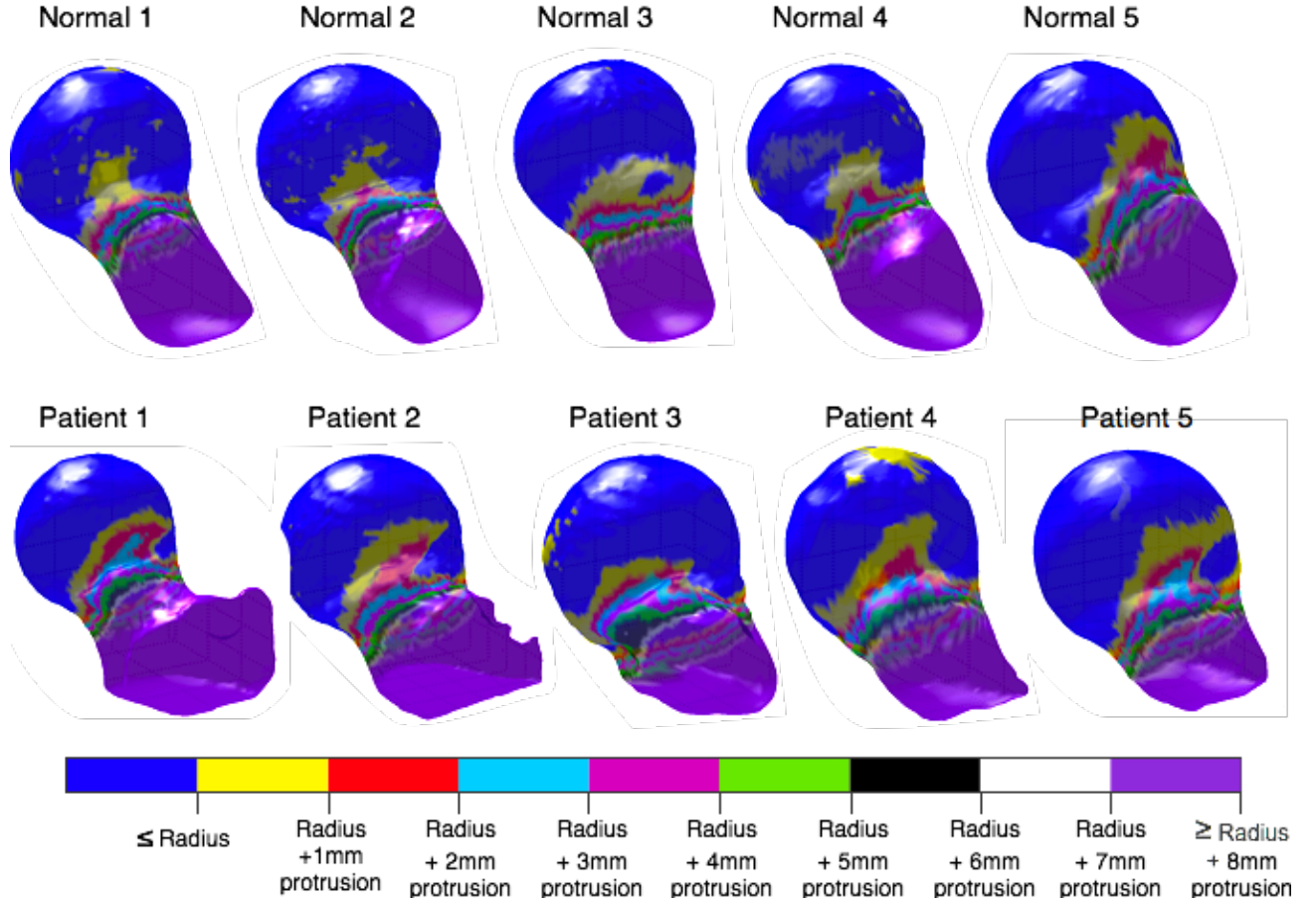


Figure 3.6: The radial distance of the surfaces of the femoral head from the HJC were represented in terms of the femoral head's radius and the protrusion of the surface beyond this radius. Protrusions of up to 1 mm at the femoral head-neck junction seemed to be consistent among normal subjects (top) and cam-type FAI patients (bottom). Protrusions of more than 2 mm showed more coverage at the femoral head-neck junctions of the patients compared to the normal subjects, except for Normal Subject 5. The coverage in Normal Subject 5 was more lateral, while the coverage among the patients were more anterior.

3.2.3 Creation of Geometric-kinematic Hip Joint Models Combining Motion and Geometry and Calculation of Proximity

In order to combine motion and geometry data to create geometric-kinematic hip joint models, the geometry data was manipulated by the motion data (Matlab R2010a). More specifically, the surface geometry data of the hip joint was described in the form of vertices and the triangulation of the vertices. Each vertex was transformed, thereby leading to an overall transformation of the entire hip joint. For the purposes of applying motion to the hip joint, described by a series of three hip cardan angles, the acetabulum was held rigid as the femoral head was transformed by the hip joint angles.

To calculate the proximity of the bony surfaces in the hip joint models, a proximity algorithm was used to compare each vertex on the femoral head with each vertex on the acetabulum, and vice versa. This algorithm determined the distance between two points based on the Euclidean distance (d) between them where:

$$d = \sqrt{(p_1 - q_1)^2 + (p_2 - q_2)^2 + (p_3 - q_3)^2} \quad (3.1)$$

and in cartesian coordinates, $p = (p_1, p_2, p_3)$ and $q = (q_1, q_2, q_3)$ are two points in Euclidean three-dimensional space.

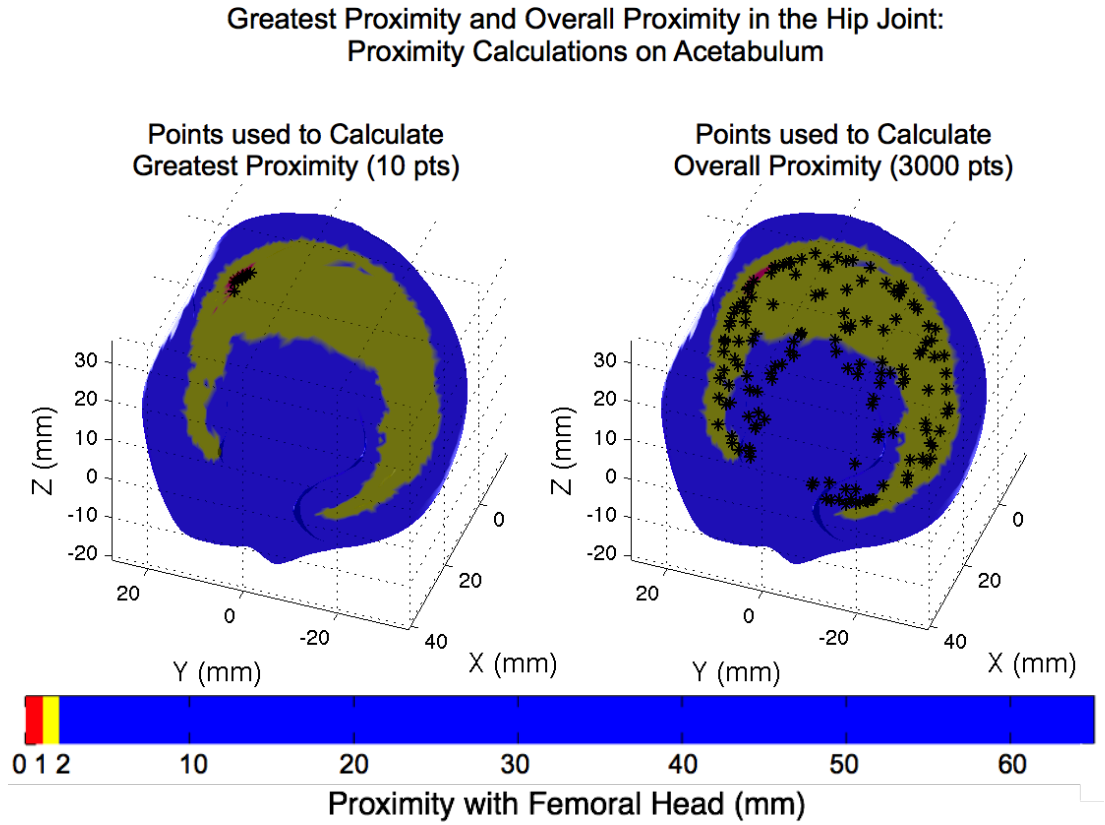


Figure 3.7: The acetabulum of Patient 1 during the impingement test at 90° flexion, 0° internal rotation, and 20° adduction is shown. The ten most proximal points (left) used to calculate the greatest proximity in the hip joint and the 3000 most proximal points (right) used to calculate the overall proximity in the hip joint are shown.

The proximity calculations determined the distance between points, which were all considered to be positive values, but was not able to detect when intersection occurred between the surfaces for which the points lay on. As a result, the intersection of any surfaces was detected visually through observation, and the proximity calculation for the points on the intersected surfaces were adjusted to be negative proximities. To gain a quantifiable understanding of the proximity calculations, the average of the ten most proximal points on the femoral head and acetabulum

were determined, as well as the average of the 3000 most proximal points to give an overall measure of proximity in the hip joint (Figure 3.7).

By conducting these transformations of the femur and these proximity calculations, geometric-kinematic hip joint models were created and analysed. They were created for the hip joint geometries of each of the five normal subjects and the five cam-type FAI patients and at selected time points during various motions including the impingement test and the functional activities of walking, sitting, and standing.

3.3 Intra- and Inter-Observer Study for Determining the Hip Joint Centre

3.3.1 Introduction

Hip joint angles were directly applied to the hip joint centre (HJC) to simulate motion in the geometric-kinematic hip joint model. The HJC was directly determined by the fitting of a sphere over the femoral head. Fitting a sphere over the femoral head was not an entirely automatic process. It required the user to select the spherical portions of the femoral head to generate a sphere, however, discretion was needed to fine-tune the sphere's centre and diameter to best fit the sphere to the femoral head. Thus, it was important to understand the variability of an individual fitting a sphere on a femoral head over different days and the effect of different observers on fitting a sphere on a femoral head. An intra-observer study was conducted with one observer performing fitting over five days, and an inter-observer study was conducted using five observers.

3.3.2 Methods

Participants in the study were shown how to fit a sphere on the femoral head using the femur of cam-type FAI Patient 2. The femur of a patient was used for demonstration since it required more user discretion in fitting a sphere over its femoral head than required for a normal subject, therefore better preparing the participants to perform the task themselves. Participants were introduced to basic maneuvering functions in Geomagic (e.g. how to zoom in and out, how to rotate the femur, how to select surfaces, etc.). The participants were then provided with a

set of instructions (Appendix A.1) to guide them in fitting a sphere on the femur of Normal Subject 1 and Patient 1. The patient’s femur used in the study differed from the femur used for the initial demonstration. The mean and standard deviations of the sphere’s diameter and the three coordinates of the sphere’s centre were calculated for the intra- and inter-observer studies.

3.3.3 Results

Standard deviations for intra-observer results were within ± 0.22 mm, and inter-observer results were within ± 0.77 mm (Table 3.1). This meant that variations in fitting a sphere on a femoral head were within 0.44 mm (95% CI) if done by the same person on different days or 1.54 mm (95% CI) if done by different people.

	Intra-Observer		Inter-Observer	
	Av (mm)	Stdev (mm)	Av (mm)	Stdev (mm)
Normal 1				
X	97.82	0.15	97.55	0.77
Y	-1.78	0.03	-1.81	0.04
Z	-74.46	0.22	-74.47	0.18
Diameter	43.70	0.15	43.65	0.18
Patient 1				
X	277.54	0.08	277.6	0.45
Y	169.61	0.10	169.67	0.14
Z	1222.17	0.07	1222.14	0.16
Diameter	50.45	0.16	50.56	0.39

Table 3.1: Intra- and Inter-observer study results for the diameter and x-, y-, and z-coordinates of the fitted sphere on the femoral head of Normal Subject 1 and Patient 1 are presented.

3.3.4 Discussion

The variations in fitting a sphere on the femoral head were greater for the inter-observer study compared to the intra-observer study, which was as expected since the inter-observer study included the results of different subjects who were expected to have varying discretion in the fitting. The greatest variations in fitting occurred for the fitting of the x-coordinate of a sphere on the femoral head of a normal subject and the fitting of the x-coordinate and diameter of a sphere on the femoral head of a cam-type FAI patient. The errors in fitting the x-coordinate of a sphere may have been due to the need for user discretion in deciding how much of the femoral head-neck junction should have been included in the sphere. Including more of the femoral

head-neck into the sphere would have resulted in the x-coordinate to have been greater, while less inclusion would have resulted in the x-coordinate to have been smaller. The errors in fitting the diameter of a sphere for the cam-type FAI patient could have been due to user discretion in how much of the cam deformity should have been included in the sphere. Including more of the cam deformity into the sphere would have resulted in the diameter to have been greater, while less inclusion would have resulted in the diameter to have been smaller.

The smallest variations were for fitting the y-coordinate of a sphere on the femoral head of a normal subject. This could have resulted because the anterior and posterior femoral head surfaces along the y-axis were very straight forward to detect. However, larger variations occurred for fitting the y-coordinate of a sphere for the cam-type FAI patient. This could have been due to the need for user discretion in deciding how much of the anterior portion of the cam deformity should have been included in the sphere. More inclusion of the cam's anterior portion into the sphere would have resulted in the y-coordinate to have been greater, while less inclusion would have resulted in the y-coordinate to have been smaller.

3.3.5 Conclusion

Overall, the intra- and inter-observer studies both showed good repeatability of the fitting of a sphere on a femoral head. However, it is important to take into account possible variability in fitting a sphere on a femoral head when fitting it across different days (± 0.44 mm; 95% CI) or when being fit by different people (± 1.54 mm; 95% CI).

3.4 Results

3.4.1 Proximity Results

The average of the most proximal points and overall proximity between the femoral head and acetabulum for the impingement test at 20° adduction were compared between all normal subjects and cam-type FAI patients (Figure 3.8).

On average, the surfaces of the acetabulum and femoral head began to intersect soon after the hip was flexed at 90° for the cam-type FAI patients, while the surfaces did not intersect among the normal subjects until the hip was flexed at 90° and internally rotated to 20°. Also,

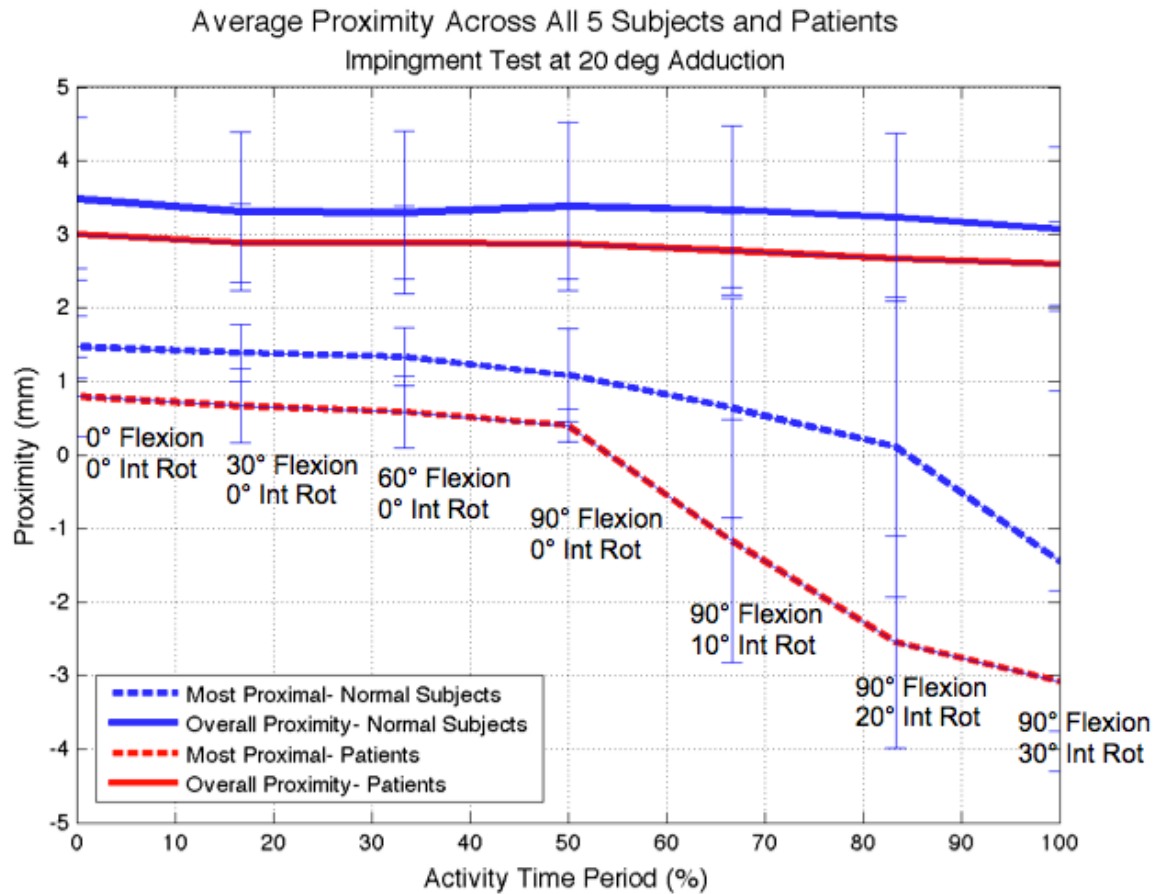


Figure 3.8: The proximity results averaged across all five normal subjects and patients for the impingement test at 20° adduction are presented; most proximal results are an average of the ten most proximal points; overall proximity is considered between the bony surfaces of the acetabulum and femoral head. Early impingement occurred in patients compared to normal subjects. Error bars represent one standard deviation.

once the hips were flexed at 90° and internal rotation was applied, the proximity in the hip joints of the patients became much closer at an accelerated rate compared to the proximity in the hip joints of the normal subjects, showing that the proximity in the patients' hip joints was much more sensitive to internal rotation compared to the normal subjects. This overall result is consistent with what was expected. The most significant finding was the occurrence of the expected result irregardless of the fact that Normal Subjects 3 and 5 seemed to have cam deformities as well. To investigate these results further, the individual results of Normal Subjects 1 to 5 and Patients 1 to 5 for the most proximal points and overall proximity between the femoral head and acetabulum for the impingement test at 20° adduction were compared (Figure 3.9).

As expected, there were common trends among the patients where the greatest proximity in the hip joint became much smaller (resulting in the intersection of joint surfaces) once internal

Proximity Between the Acetabulum and Femoral Head- Impingement Test at 20 deg Adduction Normal Subjects Patients

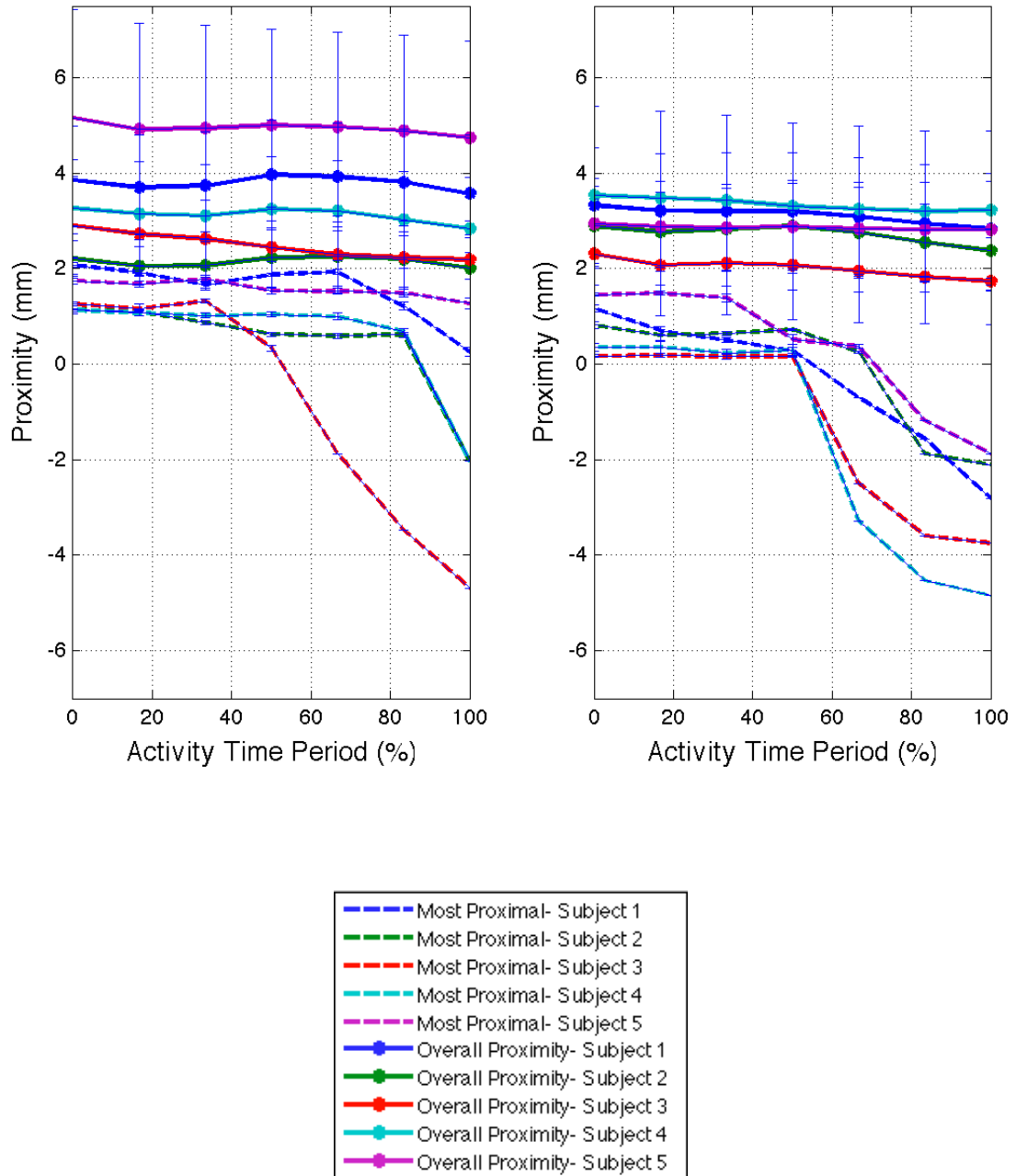


Figure 3.9: The individual proximity results for Normal Subjects 1 to 5 and cam-type FAI Patients 1 to 5 are presented for the impingement test at 20° adduction; most proximal results are an average of the ten most proximal points; overall proximity is considered between the bony surfaces of the acetabulum and femoral head. Early impingement began in Normal Subject 3 and in Patients 1, 3, and 4 at 50% the activity time period when internal rotation began to be applied. Early impingement began in Patients 2 and 5 when about 12° of internal rotation was applied. Error bars represent one standard deviation.

rotation was applied. Among the normal subjects, however, it was clear that Normal Subject 3 was very sensitive to internal rotation as well, whereas all other normal subjects, including Normal Subject 5, were not affected by internal rotation until about 20° of internal rotation was applied (80% activity time period). This may have been due to the protrusion of the bone

at the femoral head-neck junction of Normal Subject 3 being wider spread, covering anterior to lateral areas. The overall proximity in the hip joint of Normal Subject 3 was also among one of the lowest (meaning the hip joint surfaces were closer to each other) when compared to all normal subjects, which could have also made its measurement of greatest proximity more sensitive to orientation changes. As for Normal Subject 5, who had the greatest protrusion of bone at the femoral head-neck junction among the normal subjects, this hip seemed to be the least affected by internal rotation. This may have been due to the fact that Normal Subject 5 had the greatest overall proximity (meaning the hip joint surfaces were farther apart) or due to the fact that the protrusion lay more laterally.

The most proximal points and overall proximity between the femoral head and acetabulum for internal rotation at 20° adduction and at 1) 0° flexion (Figure 3.4.1), 2) 30° flexion (Figure 3.11), and 3) 60° flexion (Figure 3.11), averaged across all normal subjects and cam-type FAI patients, are presented. These plots, in contrast to the impingement test (internal rotation at 90° of flexion), showed that there was not much sensitivity to internal rotation at these degrees of flexion and adduction for normal subjects or for patients. The proximity results for individual normal subjects and cam-type FAI patients for these activities are given in Appendix A.2.

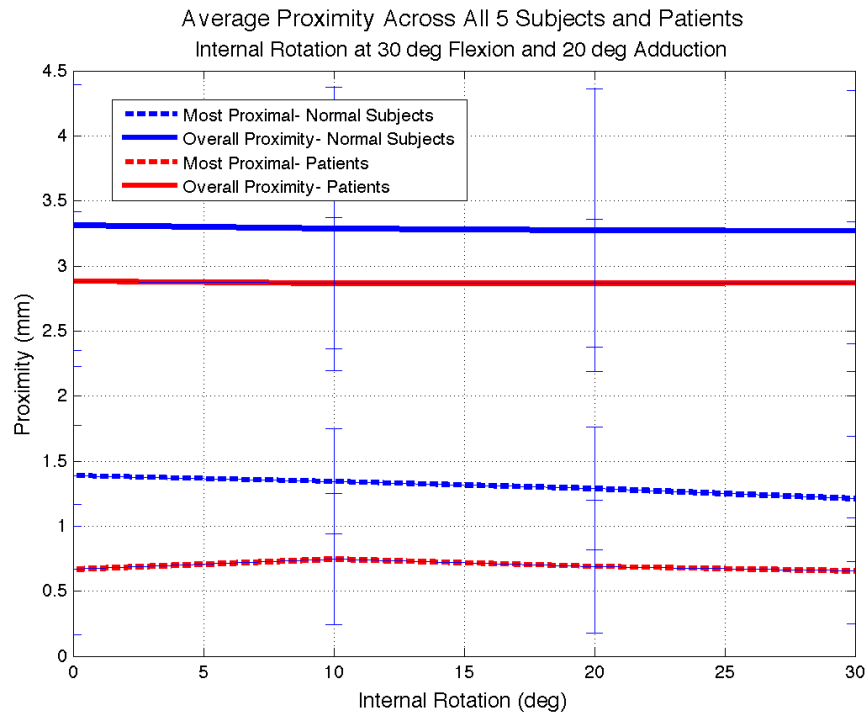
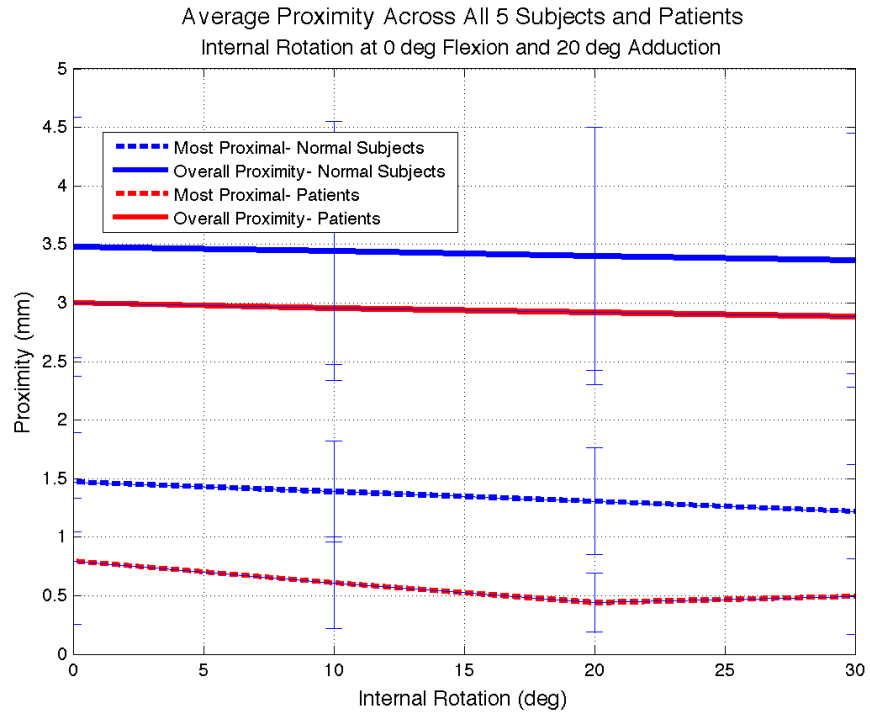


Figure 3.10: The average proximity results across all 5 normal subjects and patients for the motion of internal rotation at 20° adduction and 0° (top) and 30° (bottom) flexion are presented; most proximal results are an average of the ten most proximal points; overall proximity is considered between the bony surfaces of the acetabulum and femoral head. Impingement did not occur. Error bars represent one standard deviation.

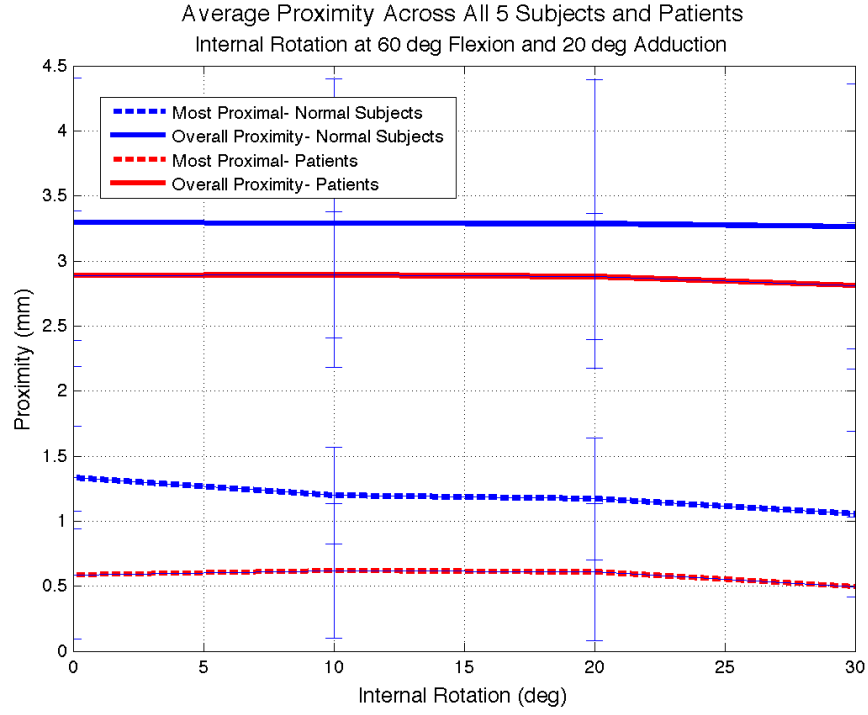


Figure 3.11: The average proximity results across all 5 normal subjects and patients for the motion of internal rotation at 20° adduction and 60° flexion are presented; most proximal results are an average of the ten most proximal points; overall proximity is considered between the bony surfaces of the acetabulum and femoral head. Impingement did not occur. Error bars represent one standard deviation.

The most proximal points and overall proximity between the femoral head and acetabulum for the functional activities of walking, sitting, and standing, averaged across all normal subjects and cam-type FAI patients, are presented (Figure 3.12). Overall, there was closer proximity in the hip joint among the patients compared to the normal subjects. However, the presence of a cam deformity among the patients did not seem to affect the trends in the proximity in the hip joint during the functional activities. The proximity results for functional activities for individual normal subjects and patients are given in Appendix A.2.

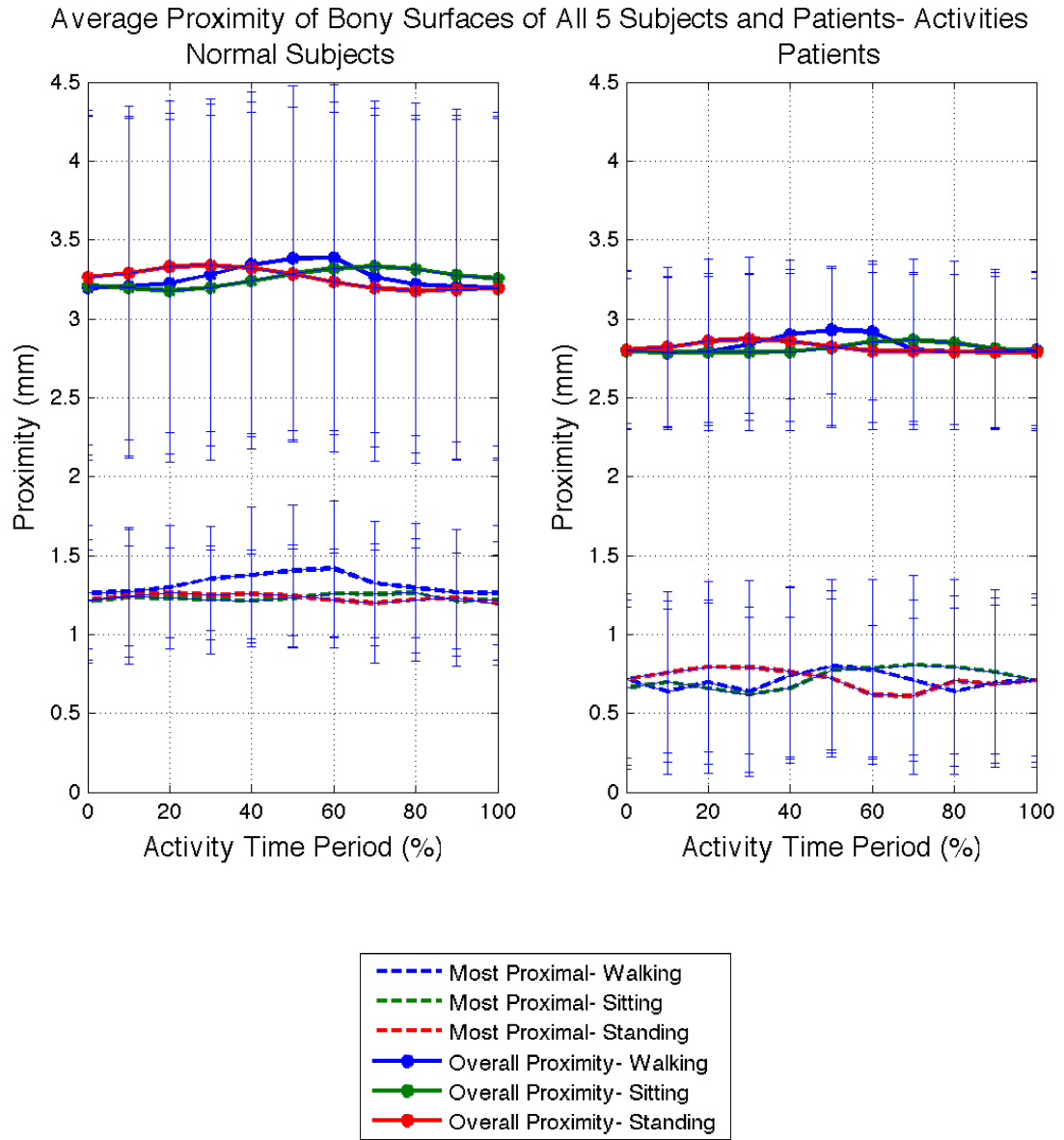


Figure 3.12: The average proximity results across all five normal subjects and patients for the functional activities of walking, sitting, and standing are presented; most proximal results are an average of the ten most proximal points; overall proximity is considered between the bony surfaces of the acetabulum and femoral head. Impingement did not occur. There was closer proximity between acetabular and femoral surfaces among patients compared with normal subjects. Error bars represent one standard deviation.

3.4.2 Geometric-kinematic Hip Joint Model Results

Measuring the proximity between the hip joint surfaces is important to provide a quantifiable comparison between the patients and normal subjects. However, to determine where the greatest proximity occurs on the hip joint surfaces, visualisation of the geometric-kinematic hip joint models was required. Also, the femoral head and acetabulum must be presented separately to view the proximity results of their surfaces. For demonstration of what the geometric-kinematic

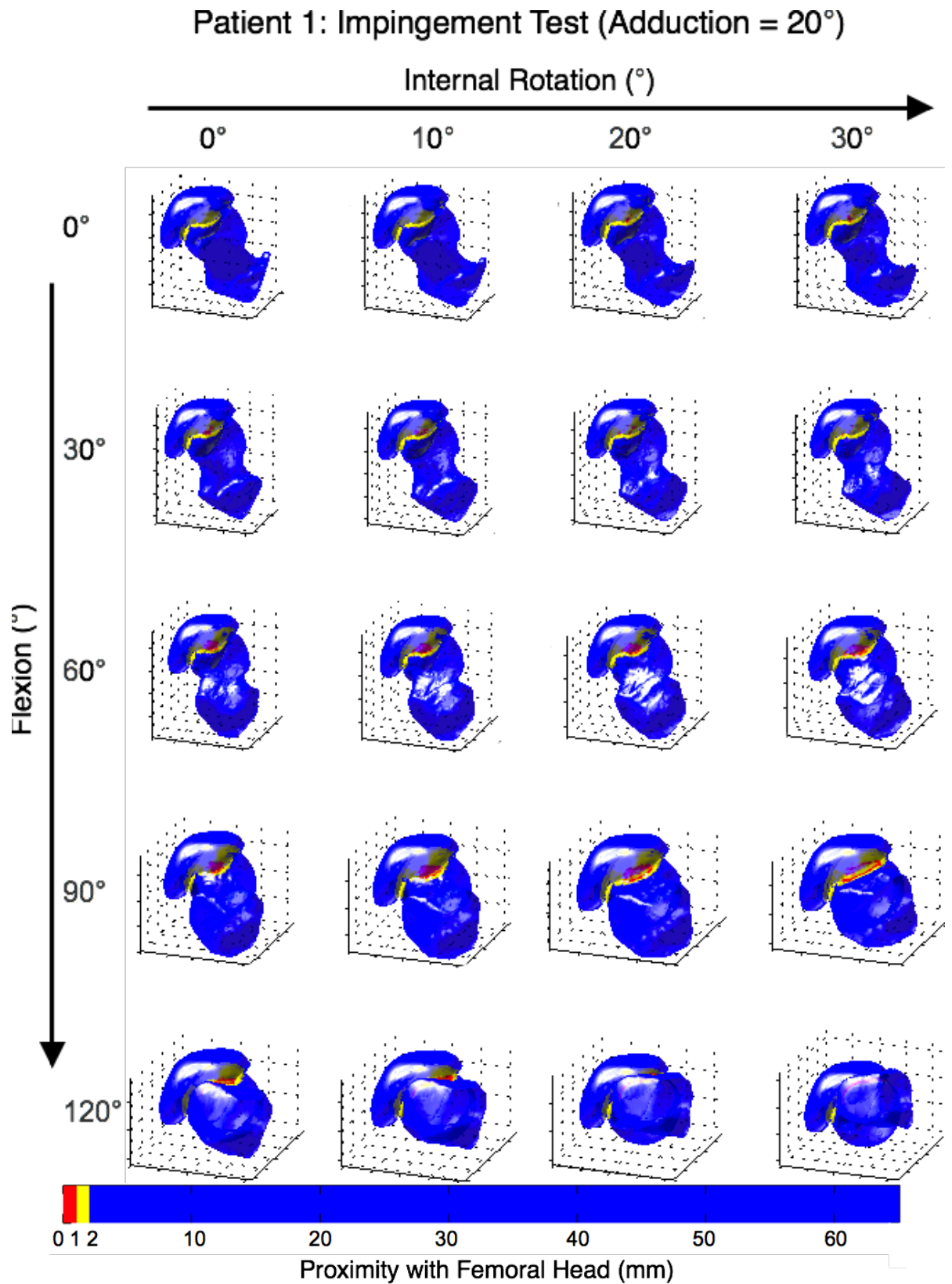


Figure 3.13: The geometric-kinematic simulation of internal rotation at varying ranges of flexion at 20° of adduction for cam-type FAI Patient 1 is presented.

hip joint model output looked like with the femoral head and acetabulum presented together, the geometric-kinematic model for cam-type FAI Patient 1 is shown where the hip joint is ori-

ented at 10° increments of internal rotation and 30° increments of flexion at a constant 20° of adduction (Figure 3.13).

The geometric-kinematic hip joint models of the femoral head and acetabulum for Normal Subjects 1 to 5 and cam-type FAI Patients 1 to 5 have been presented at the following hip joint orientations:

1. 90° of flexion, 0° of internal rotation, and 20° of adduction (Figure 3.14)
2. 90° of flexion, 20° of internal rotation, and 20° of adduction (Figure 3.15)

These hip joint orientations correlate with the orientations for which, on average, the patients' and normal subjects' hip joint surfaces began to intersect, respectively, due to sensitivity to internal rotation.

At 90° of flexion, 0° of internal rotation, and 20° of adduction (Figure 3.14) the models showed that generally among the patients, the greatest proximity in the hip joint lay in the anterior part of the femoral head-neck junction and the anterior superior part of the acetabulum as expected. This trend also occurred among Normal Subjects 3 and 5. At 90° of flexion, 20° of internal rotation, and 20° of adduction (Figure 3.15), this trend generally occurred among all the patients as well as all the normal subjects which was an expected finding, since substantial internal rotation was applied.

The visualisation of these geometric-kinematic models for Normal Subject 3 confirmed that the anterior to lateral bony protrusion on the femoral head-neck junction led to the greatest proximity in the hip joint. As for Normal Subject 5, these models showed that the anterior portion of the femoral head led to the greatest proximity, confirming that the large lateral cam deformity did not contribute to the greatest proximal points in the hip joint. Thus, regardless of the large size of the cam deformity in Normal Subject 5, because of its lateral location, it did not contribute to any increased sensitivity of the hip joint to internal rotation.

Impingement Test: 90° Flexion, 0° Internal Rotation, 20° Adduction

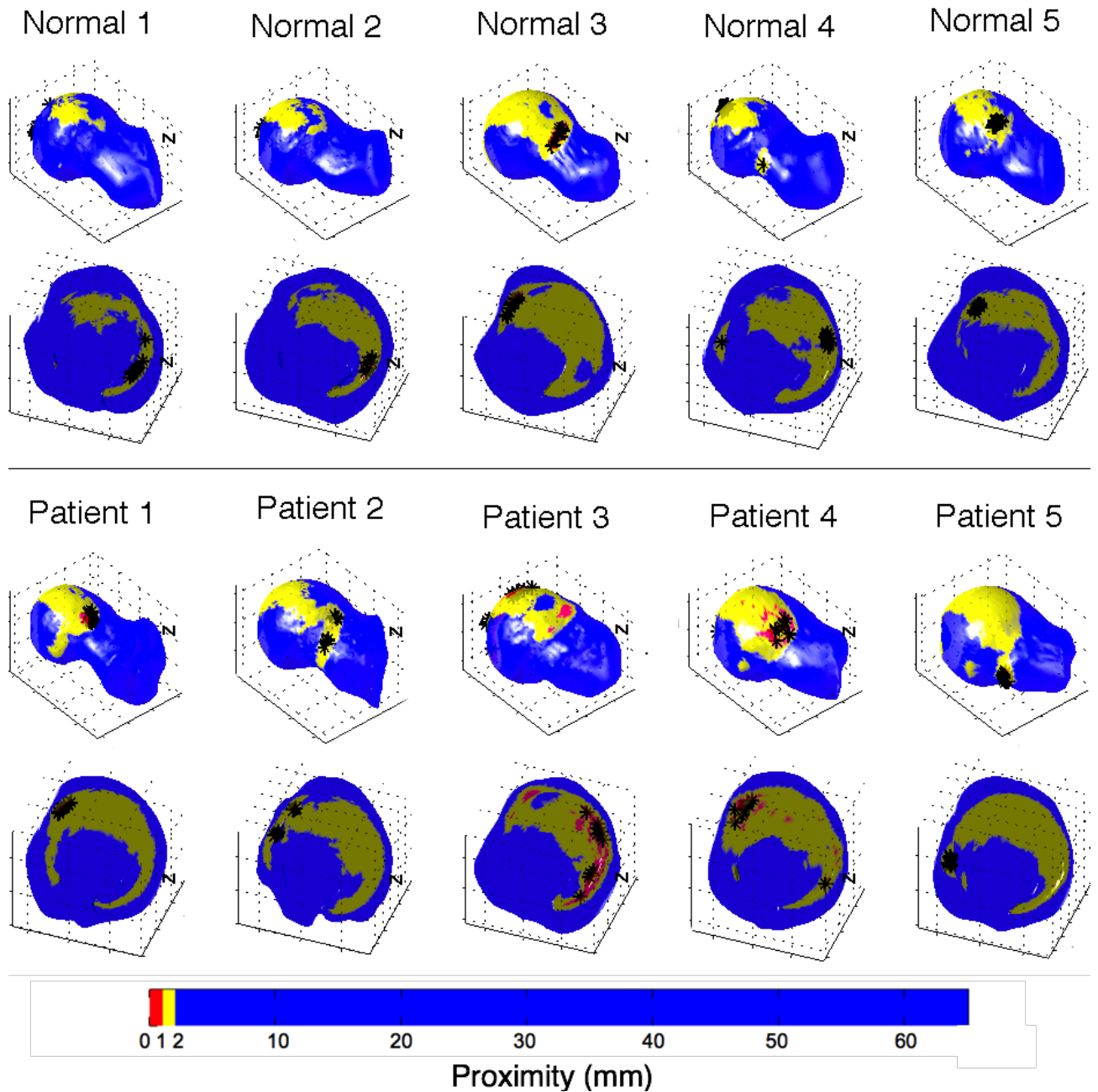


Figure 3.14: The femoral head and acetabulum are displayed from the geometric-kinematic simulation of the impingement test on normal subjects (top) and patients (bottom) at 90° of flexion, 0° of internal rotation, and 20° of adduction. The black points indicate the ten most proximal points that lie between the femoral and acetabular hip joint surfaces. The greatest proximity lies in the anterior part of the femoral head-neck junction and the anterior-superior region of the acetabulum in Patients 1 to 4 and Normal Subjects 3 and 5. There is generally greater proximity among the patients compared to the normal subjects. Please note that for Patient 3, the intersection of the surfaces occurred in the anterior area of the femoral head-neck junction and the anterior-superior region of the acetabulum; this is the area of greatest proximity, however, the black points do not indicate this.

Impingement Test: 90° Flexion, 20° Internal Rotation, 20° Adduction

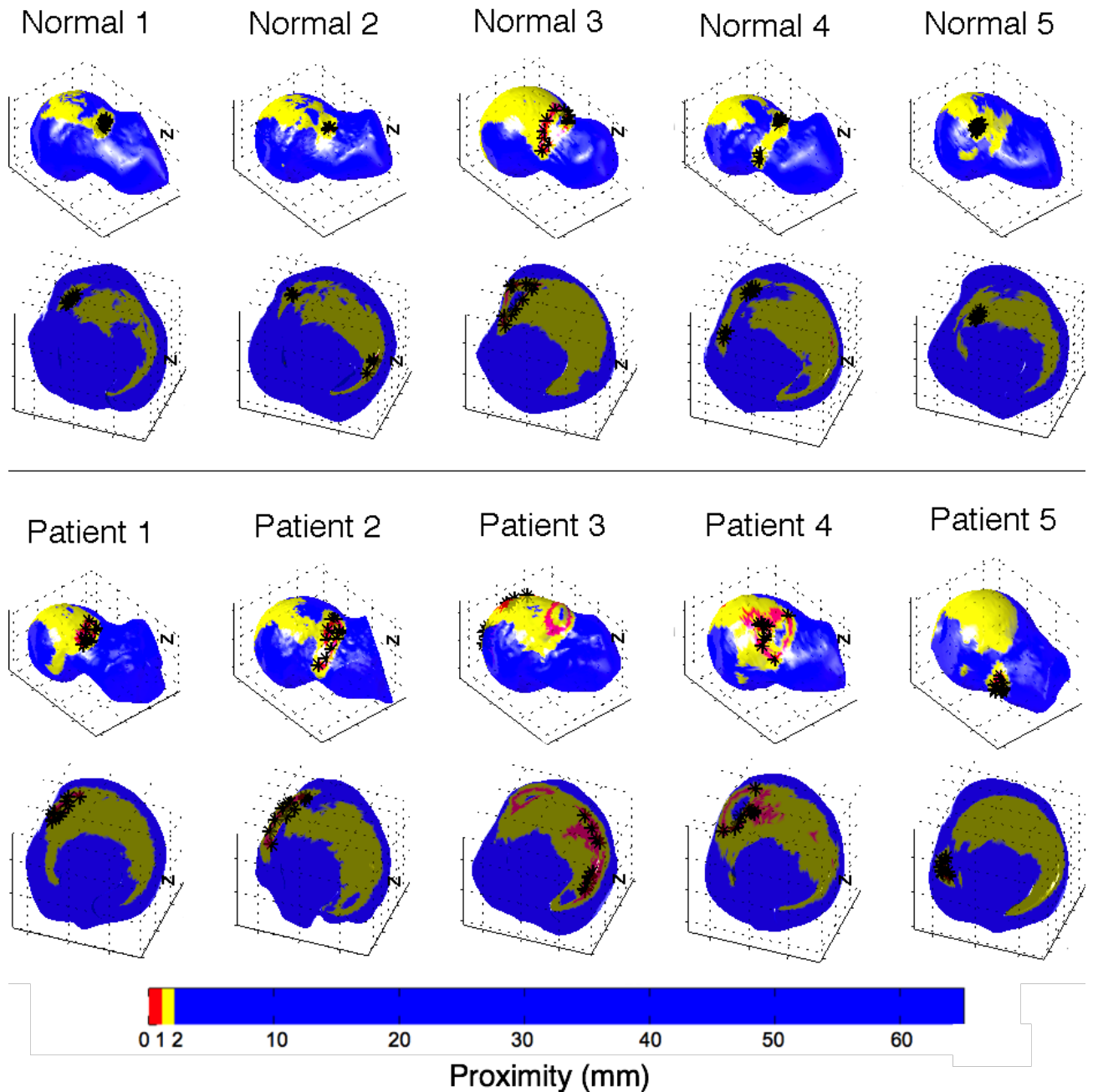


Figure 3.15: The femoral head and acetabulum are displayed from the geometric-kinematic simulation of the impingement test on normal subjects (top) and patients (bottom) at 90° of flexion, 20° of internal rotation, and 20° of adduction. The black points indicate the ten most proximal points that lie between the femoral and acetabular hip joint surfaces. The greatest proximity lies in the anterior part of the femoral head-neck junction and the anterior-superior region of the acetabulum in Patients 1 to 4 and Normal Subjects 1 to 5. There is generally greater proximity among the patients compared to the normal subjects. Please note that for Patient 3, the intersection of the surfaces occurred in the anterior area of the femoral head-neck junction and the anterior-superior region of the acetabulum; this is the area of greatest proximity, however, the black points do not indicate this.

3.5 Discussion

When the bony surfaces of the femoral head and acetabulum were simulated to have intersected each other, this did not represent a realistic event. The intersecting of the surfaces was probably due to the application of unrealistic ROM's or because the HJC was restricted from any translation.

The geometric-kinematic hip joint models and proximity results in this chapter showed that during the impingement test (internal rotation at 90° of flexion), the cam deformity impinged on the anterior superior area of the acetabulum which is the area clinically affected by cam-type FAI. Also, it was determined that cam-type FAI patients experienced early impingement during the motions of the impingement test, and regardless of the fact that cam-type patients experience pain due to functional activities such as walking, sitting, and standing, geometric-kinematic models and proximity results indicated that impingement does not occur during these activities. The early impingement of cam-type patients was possibly due to the larger coverage and protrusion of their cam deformities and the smaller overall proximity in their hip joints.

The results also showed that it is possible that normal subjects can have cam deformities, and it is possible for them to experience cam-type FAI if their hip is maneuvered into certain positions. Normal Subject 3 had only a slight cam deformity, however, this hip showed smaller overall clearance than most of the other normal hips and clearly demonstrated the presence of hip impingement for internal rotation at 90° of flexion (Figure 3.9). It is possible that individuals such as Normal Subject 3 avoid impingement-producing ROM because they are not very active or their femur is normally in a more retroverted position, requiring much greater ranges of internal rotation for impingement to occur. It is also possible that they presently do have cam-type FAI, but have yet to experience painful symptoms or to show any signs of damage.

For Normal Subject 5, the presence of a lateral cam deformity led to only a slight change in greatest proximity in the hip joint during internal rotation at 90° of flexion (Figure 3.9). This may have been due to the fact that this subject's hip joint had the greatest overall proximity of about 5 mm, allowing for clearance of the cam deformity, whereas all other individuals' overall proximity ranged from 2 to 4 mm. It is also possible that greater proximity due to the presence

of the cam deformity could have occurred if the deformity was located more anteriorly versus laterally on the femoral head. This indicates that measuring the overall proximity between the acetabulum and femoral head is sometimes essential to understanding the presence or absence of impingement, and that the location of the cam deformity can greatly affect the presence or absence of impingement.

Finally, Patients 1, 2, 4, and 5 had cam deformities that generally covered similar areas on the femoral head, and Patients 1 and 5 had cam deformities that protruded more than Patients 2 and 4, however, Patients 3 and 4 presented the greatest amount of impingement compared to other patients. For Patient 3, this could have possibly been the case because their overall hip joint proximity was the smallest among all the patients, averaging around 2 mm, whereas other patients had overall proximities ranging from about 3 to 3.5 mm. However, it is not clear why Patient 4 experienced greater impingement compared to Patients 1 and 5. It is possible that variations in cam shape and size and the relative clearance and shape of the acetabulum could have account for these results, however, more sophisticated modelling would be required to fully determine this. Therefore, it is possible that slight variations in the coverage and protrusion of cam deformities, hip joint clearance, and specific hip joint geometry can result in variations in greatest proximity calculations.

3.6 Conclusion

Geometric-kinematic hip joint models were successfully created for five normal subjects and five cam-type FAI patients which combined hip motion and hip joint geometry. A method based on proximity calculations was also successfully created and implemented to detect hip impingement in these ten individuals during various activities with large and small ROM's.

Proximity results showed that early impingement occurred in the cam-type FAI patients during the motion of the impingement test, however, impingement did not occur during functional activities such as walking, sitting, and standing. The early impingement of cam-type patients was possibly due to the larger coverage and protrusion of their cam deformities and the smaller overall proximity in their hip joints. However, deeper analyses of these geometric factors would be required to understand their full effect on impingement in cam-type FAI patients. It was also shown that cam deformities can be found in normal subjects. Possible reasons for the symptom-

free state of these subjects include having large hip joint clearance, high tissue durability, or a lifestyle and/or hip kinematics and geometry that prevent their cam deformity from causing any harm.

Overall, the study conducted in this chapter showed that it is essential to measure and characterise an individual's hip joint clearance, cam location, cam protrusion, and cam coverage or width to understand in greater depth when impingement might occur, why it occurs, and help determine if the individual is susceptible to cam-type FAI based on their subject-specific activity level, hip joint geometry, and tissue durability. Since the measurement of an alpha angle does not provide clear information about the above listed geometric hip joint parameters, it can also be concluded that the clinical standard of taking a single measurement of a person's alpha angle in an AP or cross-table lateral radiograph is an insufficient measurement for determining the presence of a cam deformity that may cause or lead to cam-type FAI.

Chapter 4

Preliminary FE Models for Simulating Cam-type FAI

4.1 Introduction

The aim of this chapter was to develop an FE model which could simulate the two main types of damage clinically seen in cam-type FAI patients, labral tears and cartilage delamination. An additional aim was to develop the FE model to enable the investigation of the following parameters which were identified in the previous chapter to be essential to characterise in order to gain a deeper understanding of the cam-type impingement mechanism:

1. Cam protrusion
2. Cam width
3. Hip joint clearance

To develop such an FE model, a series of preliminary models were created which started with simplified 2D geometries that were rectangular in shape. Later models consisted of circular geometries and then subject-specific geometries created from CT scans of human hip joints. The series of models that were created can be described in five stages:

- Stage 1. Idealised rectangular geometry, linear motion
- Stage 2. Subject-specific rectangular geometry, linear motion
- Stage 3. Idealised circular geometry, rotational motion

- Stage 4. Subject-specific geometry, subject-specific motion and load
- Stage 5. Utilising a subject-specific FE model for parameter studies

The setup of the models at each stage including geometry creation, defining material properties, mesh generation, defining interactions, assigning boundary conditions, and defining analysis formulations are described in the rest of this chapter.

4.2 Stage 1. Idealised Rectangular Geometry, Linear Motion

This stage consisted of the very simple setup of using idealised rectangular geometry, linear motion, and implicit analysis to simulate cam-type impingement.

4.2.1 Model Setup

Creating Geometry and Defining Analysis Steps

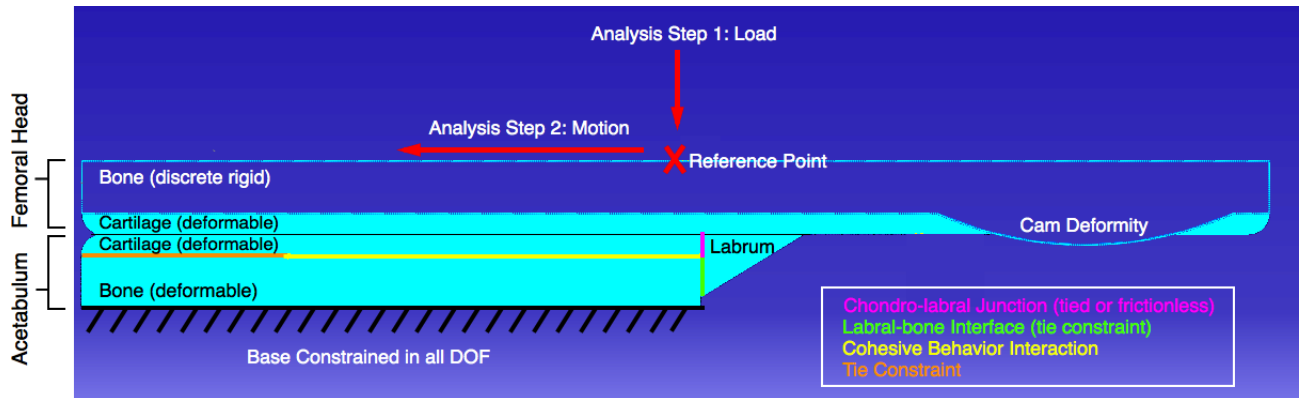


Figure 4.1: The Stage 1 2D FE model with idealised rectangular geometry and linear motion is presented. Two rectangular geometries represented the femoral head and acetabulum which each have articular cartilage and bone structures; the triangular structure on the outer edge of the acetabular articular cartilage represented the labrum; a discrete rigid, bony, semi-circle protrusion on the femoral joint surface represented the cam deformity. A load was applied to simulate body weight, and the femoral head was translated at a certain velocity to simulate impingement.

The cartilage and bone of the acetabulum and femoral head were modelled as rectangular geometries (Abaqus 6.11, Dassault Systems; Velizy-Villacoublay Cedex, France), while the cam deformity was modelled as a circular segment protruding from the cartilage surface of the acetabulum (Figure 4.1). The labrum was modelled as a triangular structure on the outer edge of the acetabular articular cartilage.

Defining Material Properties

The deformable cartilage ($E = 13.5$ MPa, $\nu = 0.45$), labrum ($E = 22$ MPa, $\nu = 0.45$), and bone ($E = 17$ GPa, $\nu = 0.3$) were modelled as linear elastic isotropic materials [68]. The bone of the femoral head was not given any material properties as it was modelled as a rigid body.

Mesh Generation

Eight-node biquadratic, plain strain, solid continuum, reduced integration (CPE8R) elements with a size of 0.1 mm were used to model the deformable cartilage, labrum, and bone components. Two-node, linear link, plain strain (R2D2) elements with a size of 0.3 mm were used to model the discrete rigid bone of the femoral head.

Defining Interactions between Structures

Tie constraints were applied to the surfaces of the articular cartilage and bone of the femoral head, to the surfaces of the labrum and acetabular bone at the labral-bone interface, and to the surfaces of the articular cartilage and bone of the acetabulum that lied towards the inside the hip joint. Either a tie constraint or frictionless behaviour (penalty contact with tangential friction of 0) was applied to the surfaces of the acetabular articular cartilage and labrum at the chondrolabral junction. A surface-to-surface contact interaction modelling frictionless behaviour (penalty contact with tangential friction of 0) was applied between the articulating surfaces of the femoral head and acetabulum including the surfaces of the femoral and acetabular articular cartilages, the labrum, and the cam deformity. A rigid body constraint was applied to the femoral bone which made it rigid to a single reference point. Cohesive behaviour was applied between the outer portion of the acetabular cartilage and bone with cohesive behaviour coefficients (normal and shear stress-intensity factors) $k_{nn} = 1.83 \text{ MPa}(mm)^{0.5}$ and $k_{ss} = 0.7 \text{ MPa}(mm)^{0.5}$, creating a cohesive zone modelling (CZM) traction-separation behaviour between the surfaces [147].

Assigning Boundary Conditions and Defining Analysis Steps

Two analysis steps were applied (Figure 4.1). The base of the acetabular bone was constrained in all degrees of freedom (DOF) during both analysis steps. A concentrated load of 10 N

(pressure of 0.5 MPa) was applied at the reference point of the rigid femoral bone during Analysis Step 1 to simulate body weight. This was based on using a 2D slice thickness of 1 mm. The femoral bone was constrained in every DOF except for the direction in which the load was applied. During Analysis Step 2, a constant body weight was continuously applied as well as a velocity of 8 *mm/s* which translated the femoral head across the acetabulum to simulate the motion of impingement. The femoral bone was constrained from rotating in the 2D plane.

4.2.2 Results

For the applied cohesive properties, this model showed that the entering of the cam deformity caused enough traction between the acetabular articular cartilage and bone to create separation between the surfaces. The greatest separation between the cartilage and bone typically occurred ahead of the cam deformity towards the inner portion of the hip joint. This model also showed the behaviour of the cam deformity pushing the labrum outwards from the hip joint as it entered the hip joint. This behaviour occurred whether the chondrolabral junction was modelled with a tie constraint (connected) or with a frictionless contact interaction (unconnected) (Figure 4.2).

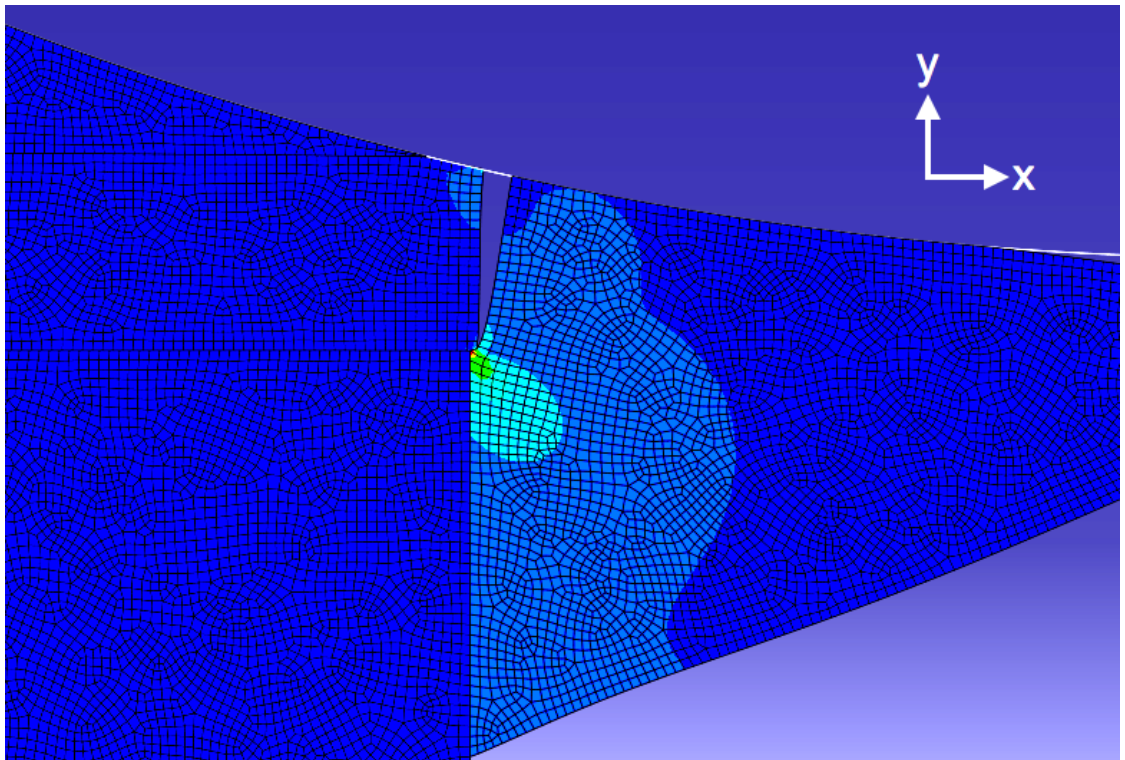


Figure 4.2: 2D FE Model of impingement where the cam deformity pushes the labrum outwards from the hip joint upon entering the hip joint.

4.2.3 Discussion and Conclusion

The anatomy of the labrum and acetabular bone and cartilage at the chondrolabral junction and labral-bone interface involved an unrealistic point load which was avoided in later models. Since cohesive behaviour has traditionally been used to describe the delamination between very stiff materials compared to articular cartilage, guidelines indicating the appropriate range of cohesive coefficients was not available. An experimentally calculated value for the poroelastic fracture toughness of the superficial layer of cartilage was used for the normal stress-intensity factor, whereas the shear stress-intensity factor was arbitrarily chosen. Therefore, these coefficients have not been validated for this model, making the amount of separation that occurred between the cohesive surfaces difficult to interpret. The simulation of the separation of the cohesive surfaces did, however, provide some insight into the possible interaction between hip joint surfaces in the case that cartilage has been delaminated. Overall, this model was not able to realistically simulate cam-type FAI.

4.3 Stage 2. Subject-specific Rectangular Geometry, Linear Motion

After developing an implicit FE model with simplified and idealised rectangular geometry, an implicit model with more realistic geometry and a more realistic damage model was created.

4.3.1 Model Setup

Creating Geometry

Subject-specific rectangular geometry was created by capturing anatomy from a Magnetic Resonance Imaging (MRI) scan through the following steps (Figure 4.3):

1. An image from an MRI scan of a patient with a cam-type abnormality was chosen where the cam deformity, bony, and soft tissue structures were all clearly visible.
2. The structures of the hip joint were segmented including the acetabular subchondral bone, acetabular articular cartilage, labrum, femoral articular cartilage, and femoral bone (Adobe Photoshop CS4, Adobe Systems Incorporated, San Jose, California).
3. The edges of the segmented structures were detected (Matlab R2010a).

4. A circle was manually fitted over the femoral subchondral bone to determine the hip joint centre (HJC) (Matlab R2010a).
5. A rectangular-to-polar coordinate system transformation was used to transform the edges of the circular structures into rectangular structures (Matlab R2010a).
6. The shape and proportionality of the structures were recreated for FE modelling (Abaqus 6.11).

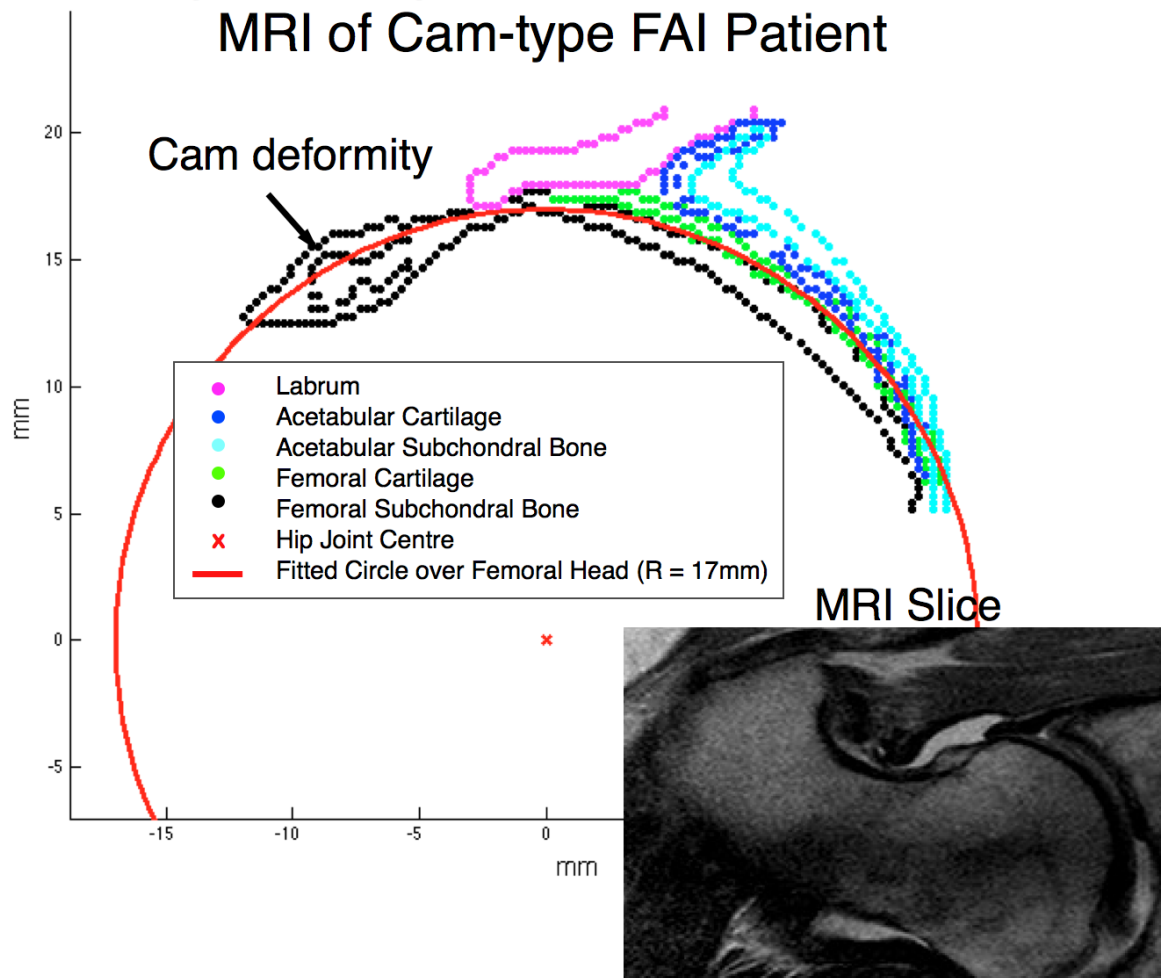
Mesh Generation, Assigning Boundary Conditions, and Defining Material Properties, Interactions, and Analysis Steps

Similar material properties mesh generation, boundary conditions, and analysis steps were applied in this model as they were applied to the FE model in Stage 1 described in Section 4.2.1. Tie constraints were applied to the surfaces of the articular cartilage and bone of the femoral head, to the surfaces of the acetabular articular cartilage and bone, to the cohesive surfaces of the acetabular cartilage-bone interface, and to the surfaces of the labrum and acetabular articular cartilage at the chondrolabral junction. A surface-to-surface contact interaction modelling frictionless behaviour (penalty contact with tangential friction of 0) was applied between the articulating surfaces of the femoral head and acetabulum including the surfaces of the femoral and acetabular articular cartilages and the labrum. Frictionless behaviour was also applied to the surfaces of the labrum and acetabular articular cartilage, which were not attached to each other.

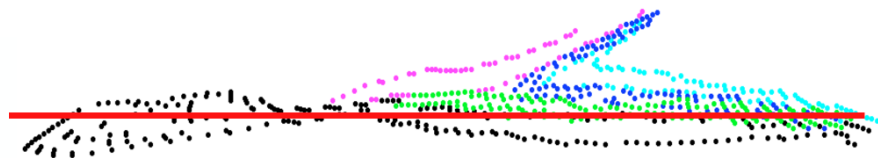
Defining Damage

A single row of cohesive elements were created at the acetabular cartilage-bone interface with cohesive behaviour coefficients similar to those defined in Stage 1 in Section 4.2.1 (Figure 4.4). An XFEM formulation based on a cohesive segments approach was applied to the acetabular articular cartilage (Figure 4.4).

Edges of Segmented Structures from MRI of Cam-type FAI Patient



Edges of Segmented Structures Transformed into Rectangular Structures



2D FE Model with Subject-specific Rectangular Geometry

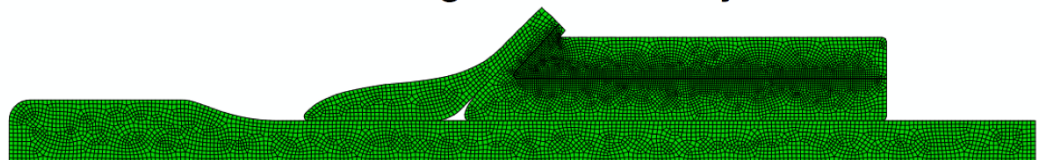


Figure 4.3: The creation of subject-specific rectangular geometry (bottom) by segmenting anatomical structures, detecting the edges of the segmented structures (top), and transforming the structures into rectangular structures (middle) is depicted.

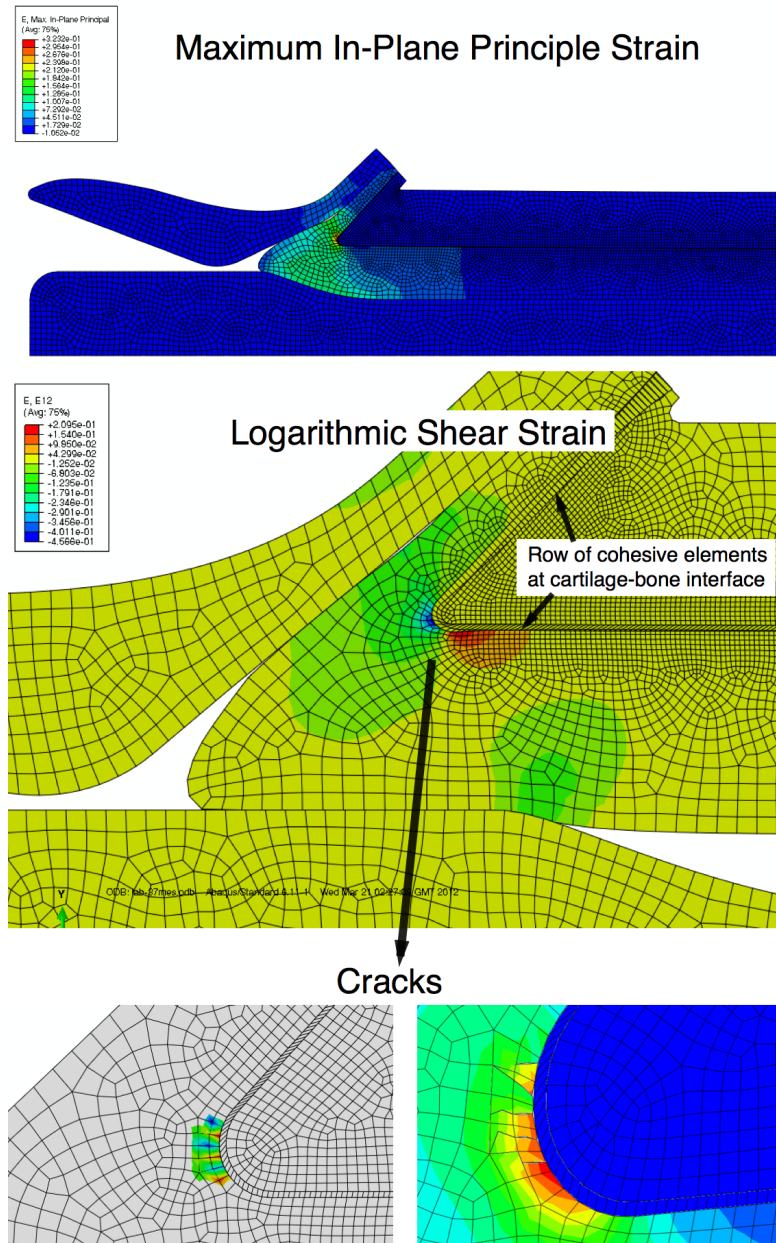


Figure 4.4: The results of the FE model with subject-specific geometry and linear motion is displayed. The maximum “maximum in-plane principal strain” (red) was 0.32; the minimum “maximum in-plane principal strain” (blue) was -0.01 (top). The maximum logarithmic shear strain (red) was 0.21; the minimum logarithmic shear strain (blue) was -0.46 (middle). Cracks formed in the acetabular articular cartilage close to the cartilage-bone interface due to simulated cam-type impingement motion (bottom).

Results

The labrum seemed unaffected by the cam deformity in this model because when the acetabular articular cartilage was deformed by the cam deformity, it caused the labrum to move away from the cam deformity (Figure 4.4). The row of cohesive elements gave similar results to the cohesive surface in Stage 1 where separation of interfacial surfaces due to traction between the interfacial surfaces occurred along the entire interface where cohesive behaviour was applied.

Multiple cartilage cracks initiated near the cartilage-bone interface. These cracks initiated at the same time increment, otherwise, only one crack would have been able to form because once one or multiple cracks have been initiated in a part, no more cracks are allowed to form.

4.3.2 Discussion and Conclusion

The cohesive behaviour definition was not able to detect any local damage that could possibly lead to or result in damage such as a tear. This could possibly be due to the use of coefficients that were not determined for cohesive behaviour at the interface between cartilage and bone. As previously mentioned, definitive cohesive behaviour parameters for cartilage-bone interfacial cracking is not available, therefore, cohesive behaviour was determined to not be a suitable damage representation.

The movement of the labrum away from the cam deformity seemed to be unrealistic behaviour, so it was determined that a rectangular model with linear motion was not sophisticated enough to model cam-type impingement. However, the damage formulation using XFEM showed potential in providing a quantitative and visual representation of soft tissue tears or cartilage delamination.

4.4 Stage 3. Idealised Circular Geometry, Rotational Motion

4.4.1 Model Setup

An implicit FE model with circular geometry and rotational motion was created based on modelling the femoral head to be a perfect circle (Figure 4.5). For simplicity, only the cartilage and bone structures were modelled initially. Similar material properties and mesh generation were applied in this model as they were applied in Stage 1 in Section 4.2.1. Tie constraints were applied to the surfaces of the articular cartilage and bone of the femoral head and acetabulum. A surface-to-surface contact interaction modelling frictionless behaviour (penalty contact with tangential friction of 0) was applied between the articulating articular cartilage surfaces of the femoral head and acetabulum. A rigid body constraint was applied to the femoral bone which made it rigid to a single reference point placed at the HJC. A rigid body constraint was also applied to the acetabular bone which made it rigid to a single reference point placed in the centre of the component.

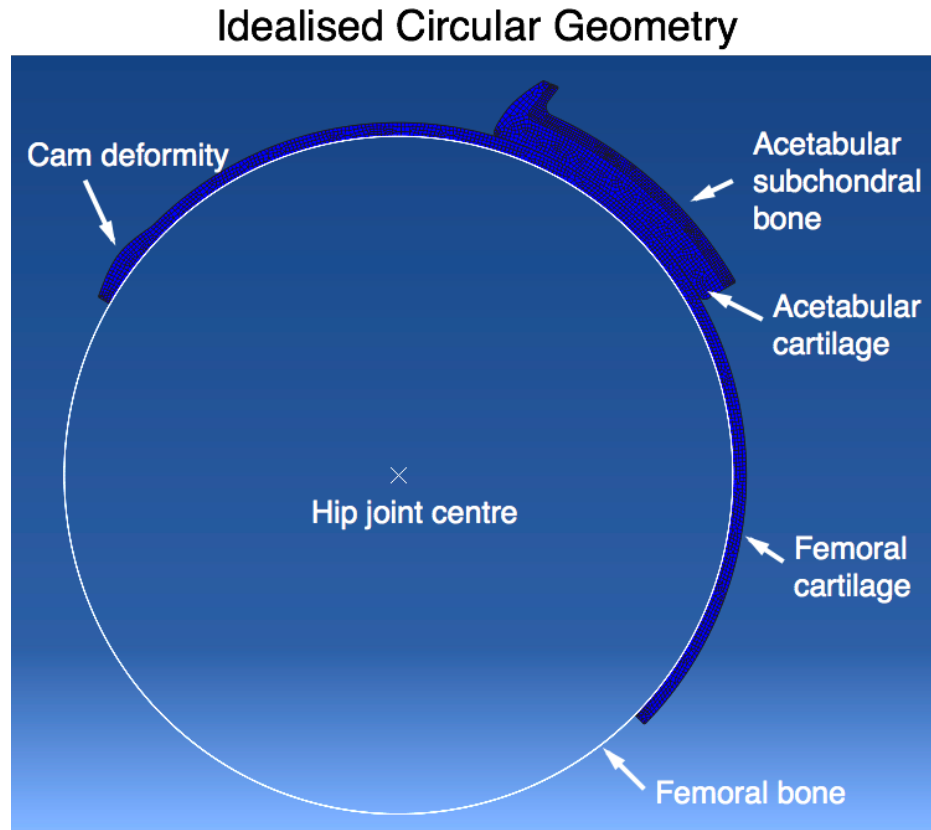


Figure 4.5: The Stage 3 2D FE model with idealised circular geometry is presented. The femoral and acetabular bone and articular cartilage components are represented. Rotational motion was applied at the HJC to simulate cam-type impingement.

Two analysis steps were applied. A concentrated load of 10 N was applied at a reference point on the acetabular bone during Analysis Step 1 to simulate body weight on the 2D slice. The direction in which body weight was applied was directed from the center of the acetabular bone part towards the HJC. The acetabular bone was constrained in every DOF except for the direction in which the load was applied. This allowed the translation of the acetabulum so that hip joint dislocation during motion could be simulated. During Analysis Step 2, body weight was continuously applied, the acetabulum was still allowed to translate in the direction of body weight, and a rotational displacement was applied at the HJC which was constrained in all DOF except for rotation.

4.4.2 Results

The cam deformity entered the hip joint, since the acetabulum was allowed to move in the direction in which body weight was applied, giving clearance for the cam. The entering of the cam deformity caused shear strain to occur at the surface of the acetabular articular cartilage (Figure 4.6).

Idealised Circular Geometry Model: Logarithmic Shear Strain

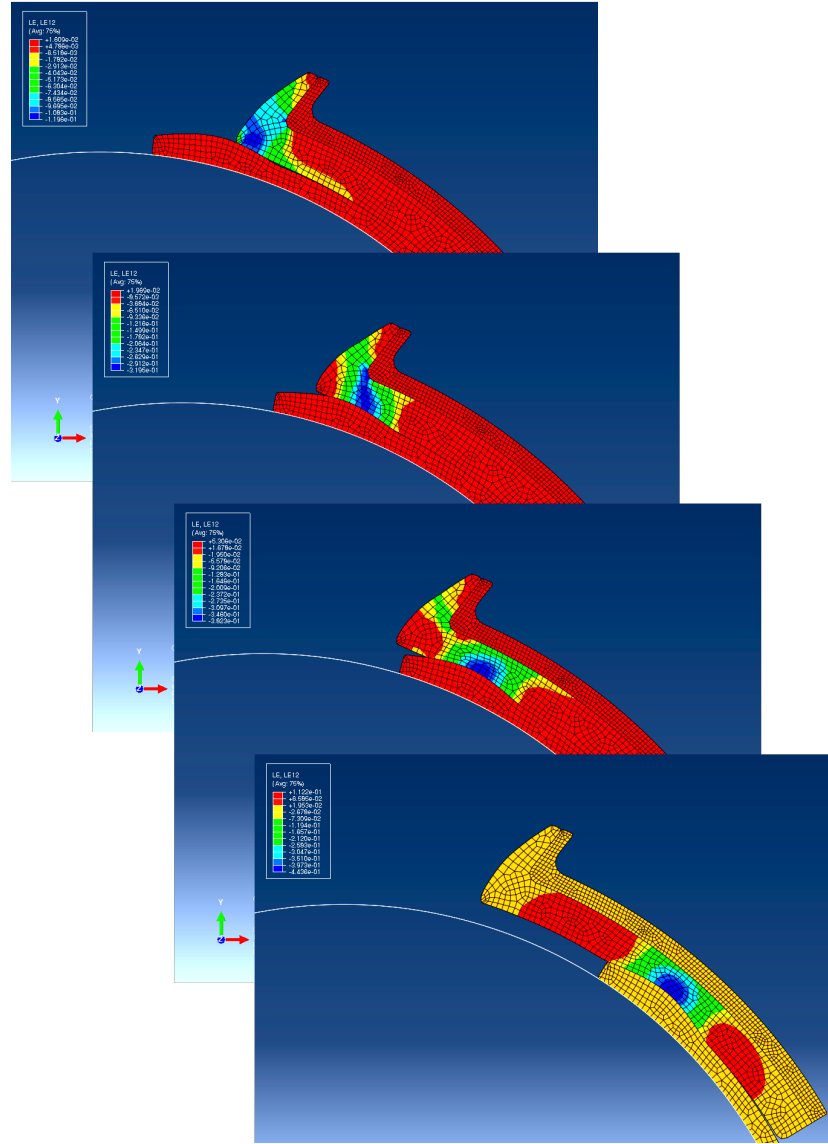


Figure 4.6: The logarithmic shear strain results of the Stage 3 FE model with idealised circular geometry and rotational motion is presented. Body weight was applied from the centre of the acetabular bone to the HJC. The acetabular bone was allowed to move in the direction in which body weight was applied, thus, the cam deformity was able to enter the hip joint. Upon full entry (bottom), the maximum logarithmic shear strain (red) was 0.11; the minimum logarithmic shear strain (blue) was -0.44.

4.4.3 Discussion and Conclusion

There were limitations in setting realistic boundary conditions for this model due to its idealised geometry. Constraining the acetabulum in all DOFs would have caused the cam deformity to abut the acetabular cartilage and not enter the hip joint. Therefore, the acetabulum was allowed to move in the direction in which body weight was applied, allowing the cam deformity to enter the hip joint. However, applying these boundary conditions did not result in any

potential damage or shear strain at the cartilage-bone interface of the acetabulum. It was also uncertain which boundary condition(s) would have led to a realistic and characteristic simulation of cam-type impingement. Therefore, this model was not able to simulate the cam-type FAI mechanism realistically, since the idealised circular geometry did not provide enough information about the possible constraints and boundary conditions in the hip joint. It was concluded that subject-specific geometry, motion, and load would likely provide this insight.

4.5 Stage 4. Subject-specific Geometry, Subject-specific Motion and Load

The aim of this stage of modelling was to create the most realistic model possible based on realistic hip joint geometry, motion, and load. To achieve this, subject-specific motion and load data [148, 149] and bony geometry from an actual cam-type FAI patient¹ were used where the motion and load data were obtained from one patient and the geometry data was obtained from a different patient. Obtaining subject-specific geometry, motion, and load from the same subject would have been ideal, however, this set of subject-specific data is not common and was not available. Implicit analysis was also used at this stage.

4.5.1 Model Setup

Mesh Generation

Four-node bilinear, quadrilateral, plain strain, solid continuum, reduced integration (CPE4R) elements with a size of 0.35 mm were used to model the deformable cartilage, labrum, and subchondral bone components. Similar elements of size 1.4 mm were used to model the deformable cortical and trabecular bone. Two-node, linear link, plain strain (R2D2) elements with a size of 0.35 mm were used to model the discrete rigid bone of the femoral head.

Defining Material Properties

Similar to the material properties that were applied to the models in Stage 1, the deformable cartilage ($E = 13.5$ MPa, $\nu = 0.45$) and labrum ($E = 22$ MPa, $\nu = 0.45$) were modelled as linear elastic isotropic materials [68]. Additional material properties were defined including

¹CT data was acquired at the University of Ghent using a 6-slice CT scan (LightSpeed VCT).

Structure	Young's Modulus	Poisson's Ratio (ν)
Articular Cartilage	13.5 MPa	0.45
Labrum	22 MPa	0.45
Subchondral Bone	2 GPa	0.3
Cortical Bone	17 GPa	0.3
Trabecular Bone	70 MPa	0.2

Table 4.1: The material properties used to represent the deformable structures in the Stage 4 2D FE model are presented.

acetabular subchondral bone ($E = 2$ GPa, $\nu = 0.3$), cortical bone ($E = 17$ GPa, $\nu = 0.3$), and trabecular bone ($E = 70$ MPa, $\nu = 0.2$) bone which were all defined to be linear elastic isotropic [68]. These material properties that were used are summarised in Table 4.1 above. The subchondral bone was a 1 mm-thick layer of bone directly underlying articular cartilage [68]. The cortical bone was a 1.5 to 3 mm-thick layer of bone underlying subchondral bone and surfaces that were absent of articular cartilage [68]. The trabecular bone was present at the innermost portion of the bone segments [68]. Femoral articular cartilage was modelled to have a uniform thickness of 1.5 mm. Acetabular articular cartilage was modelled to maximise its congruity with the femoral articular cartilage when the hip joint was in a neutral position (no internal rotation). It was also modelled to have a thickness ranging from 1.4 to 3 mm with its anterior medial and superior regions having the thickest cartilage [60, 61, 62, 63] (Figure 4.7).

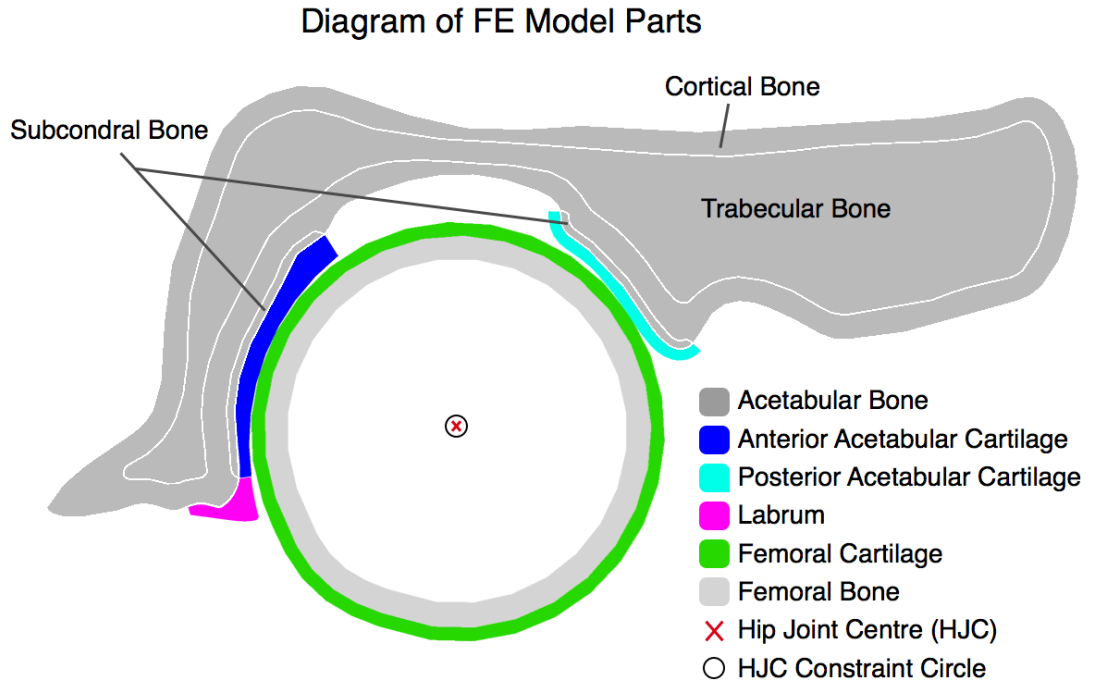


Figure 4.7: The acetabular and femoral parts modelled in the Stage 4 2D FE model are depicted.

Defining Interaction between structures

Tie constraints were applied to the surfaces of the articular cartilage and bone of the acetabulum and femoral head, to the surfaces of the labrum and acetabular bone at the labral-bone interface, and to the surfaces of the labrum and acetabular articular cartilage at the chondrolabral junction. A surface-to-surface contact interaction modelling contact friction with a coefficient of friction of 0.006 was applied between the articulating surfaces of the acetabular and femoral soft tissues. The static and kinetic coefficients of friction in human synovial joints has been determined to be 0.01 and 0.003, respectively [150, 151]. A coefficient of 0.006 was chosen because it was mid-range between these coefficients of friction. The femoral articular cartilage surfaces were defined to be the master surface, while the acetabular articular cartilage and labral surfaces were defined to be the slave surface. A rigid body constraint was applied to the femoral bone which made it rigid to a single reference point placed at the HJC.

Motion and Load, Assigning Boundary Conditions, and Creating Geometry

Motion and load data was obtained from a study where hip joint contact forces and synchronous kinematic data were recorded from patients with instrumented hip implants performing daily activities [148]. There were four patients in this study; however, three displayed abnormal hip kinematics, so only the motion and load data of a single patient (HSW) was chosen to be used. Also, all motion and load data that was used was an average of four trials.

Two different 2D models were created to simulate cam-type impingement due to either flexion or internal rotation at flexion. Both of these models utilised the motion and load data from the activity of sitting (the motion of going from a standing to sitting position), an activity with a large range of motion (ROM) of hip flexion, but relatively insignificant ROM of ab/abduction and external/internal rotation. In both of these models, load data and motion data were applied at the HJC which was constrained by a rigid circle with a radii of 2 to 4 mm. Also, the surfaces of the acetabular bone were constrained in all DOF. Details about the geometry, motion, and load data used for these two models are described.

Flexion Model

To simulate cam-type impingement due to flexion in a 2D model, the hip flexion angles and

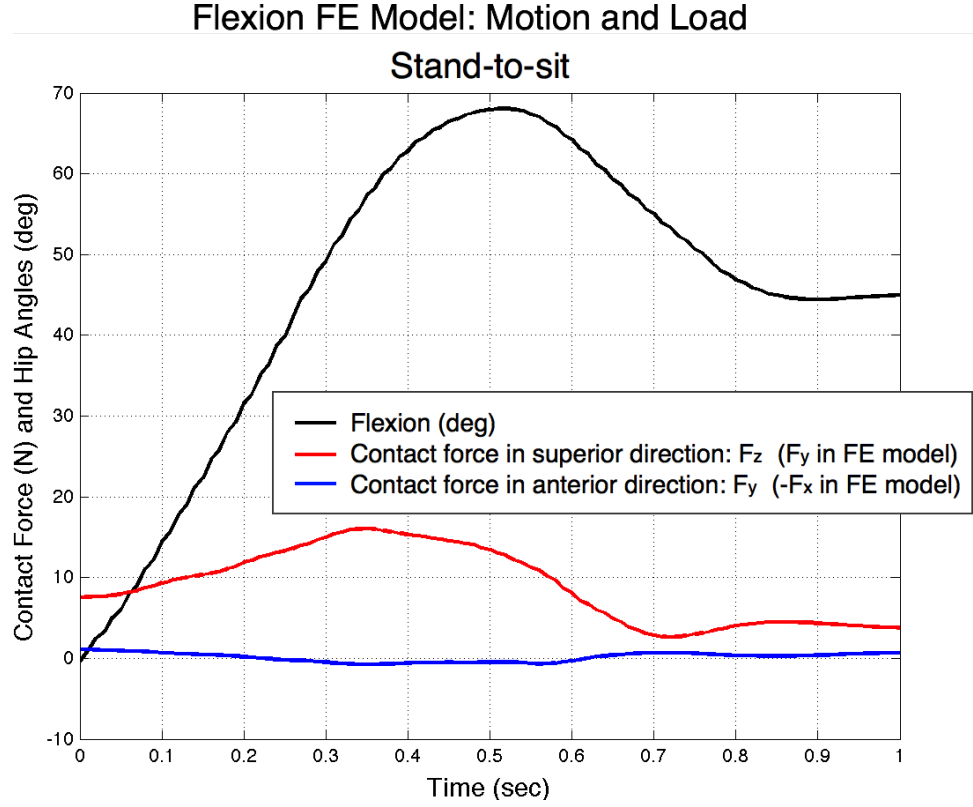


Figure 4.8: The average of four trials of hip flexion and hip contact forces in the sagittal plane in the superior direction (F_z) and anterior direction (F_y) for the motion of sitting are presented for patient HSW who has instrumented hip implants [149].

hip contact forces in the sagittal plane (plane of flexion) of patient HSW performing the sitting motion were used (Figure 4.8) [149]. The hip contact force corresponded to the contact area over the entire hip joint, however the contact area in a 2D model is only a portion of the entire contact area. Assuming that the contact area for sitting is similar to the contact area during walking and stair descent, the average contact area during sitting is about 354 mm^2 [60] and can be estimated to be rectangular in shape with its longer sides roughly four times the length of its shorter sides. Assuming that the long side of the rectangle is aligned with the sagittal plane and that a 2D model consists of a 1 mm-thick slice of the hip joint, the entire hip contact area (A) can be expressed as

$$A = 4L * L = 4L^2, \quad (4.1)$$

where L is the length of the short sides of the rectangular contact area. Solving this expression for L leads to the expression:

$$L = \sqrt{\frac{A}{4}} = \sqrt{\frac{354}{4}} = 9.4 \text{ mm} \quad (4.2)$$

The hip contact force (F_{2D}) for the 2D Internal Rotation at Flexion model can be expressed as

$$F_{2D} = \frac{F_{3D}}{L} = 0.1064F_{3D} \quad (4.3)$$

Therefore, the hip contact force used for this 2D Internal Rotation at Flexion model was 10.64% of the entire hip contact force.

A 2D slice of bony geometry from a CT scan of the hip joint through the plane of flexion (sagittal plane) was taken (Figure 4.10). The soft tissue structures including the labrum and articular cartilage were constructed based on literature values of the shape and size of the structures and by observing MRI's of the structures (Figure 4.10) (Abaqus 6.11) [16, 49, 61, 62, 63, 74, 75, 76].

Hip Contact Force Vector Projected onto Transverse (XY) Plane: Stand-to-sit

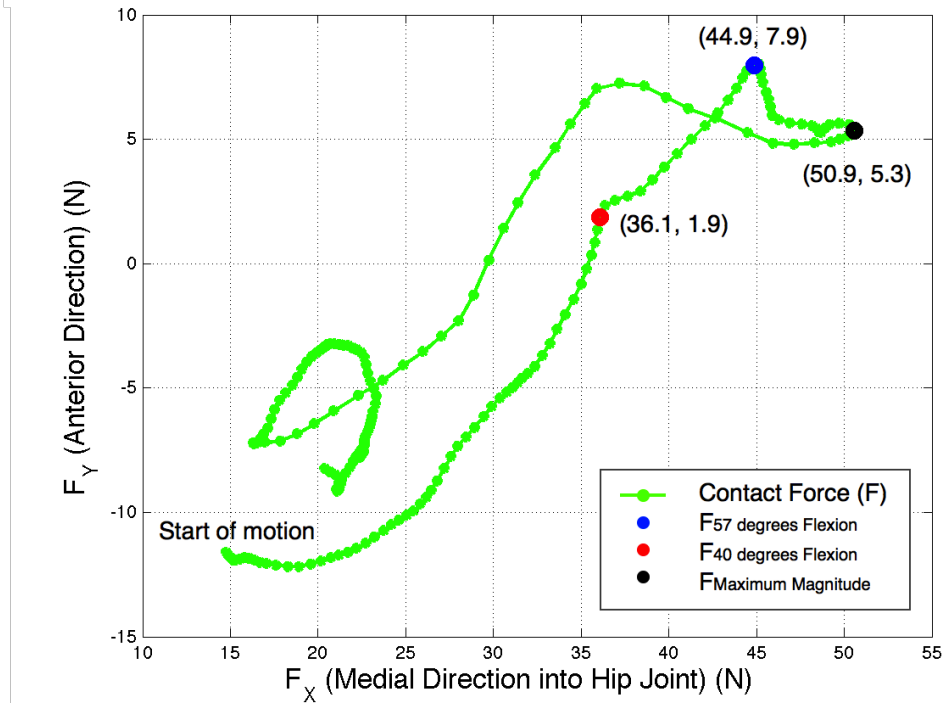


Figure 4.9: The hip contact force vector projected onto the transverse plane (plane of internal rotation) during sitting (stand-to-sit activity) is presented.

Internal Rotation at Flexion Model

To simulate cam-type impingement due to internal rotation at flexion, internal rotation was simulated during sitting. This encompassed choosing a time point during sitting where internal rotation would be applied as the hip contact forces remained constant. A time point was initially chosen based on when the hip contact force was greatest in the anterior direction. This time point was chosen because damage due to cam-type impingement typically develops in the

Subject-specific Hip Joint Geometry: Flexion Model

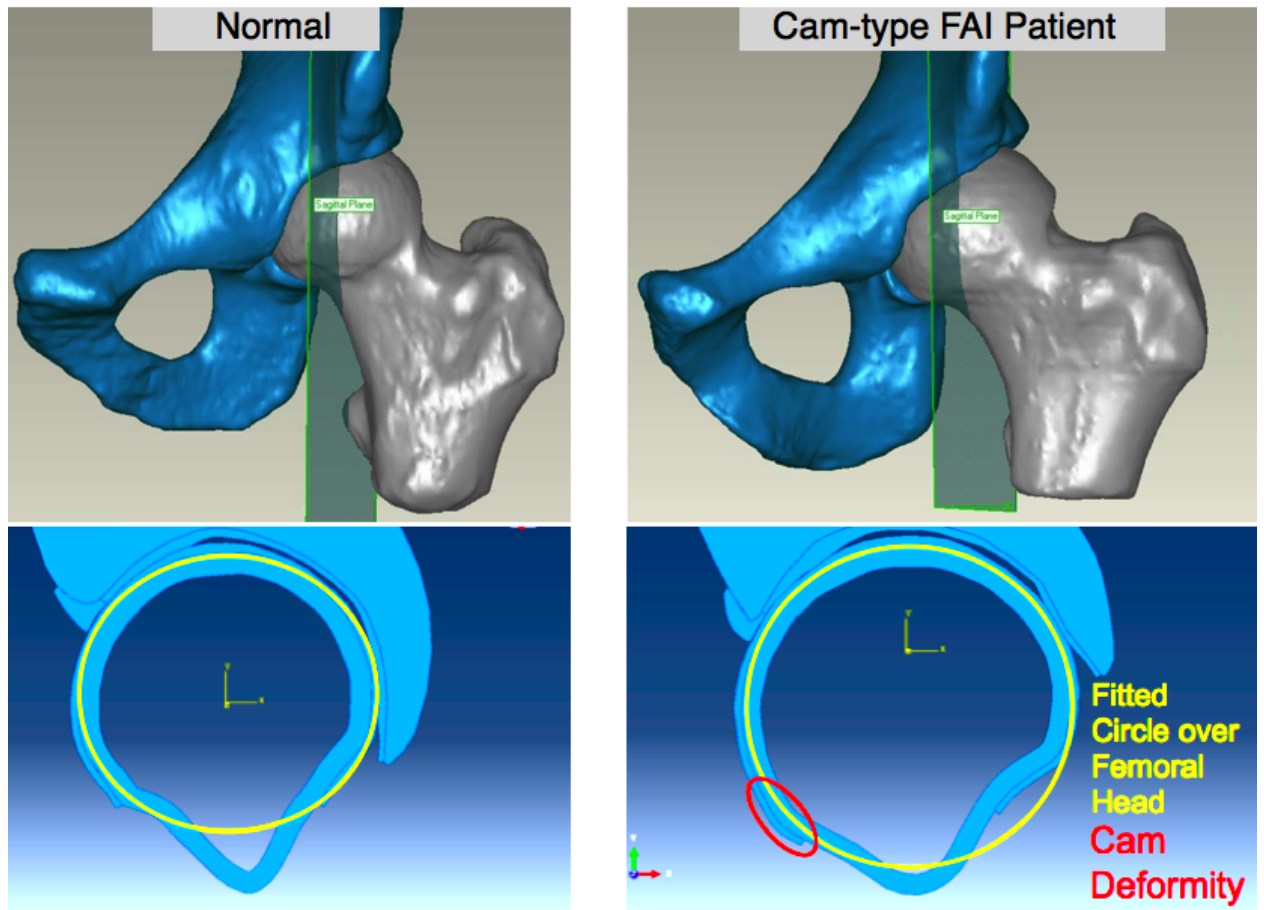


Figure 4.10: The hip joint geometry from CT scans of a normal subject (top left) and cam-type FAI patient (top right) are presented. Bony geometry was obtained from the sagittal plane (plane of flexion) from the CT data, and soft tissues were reconstructed (bottom).

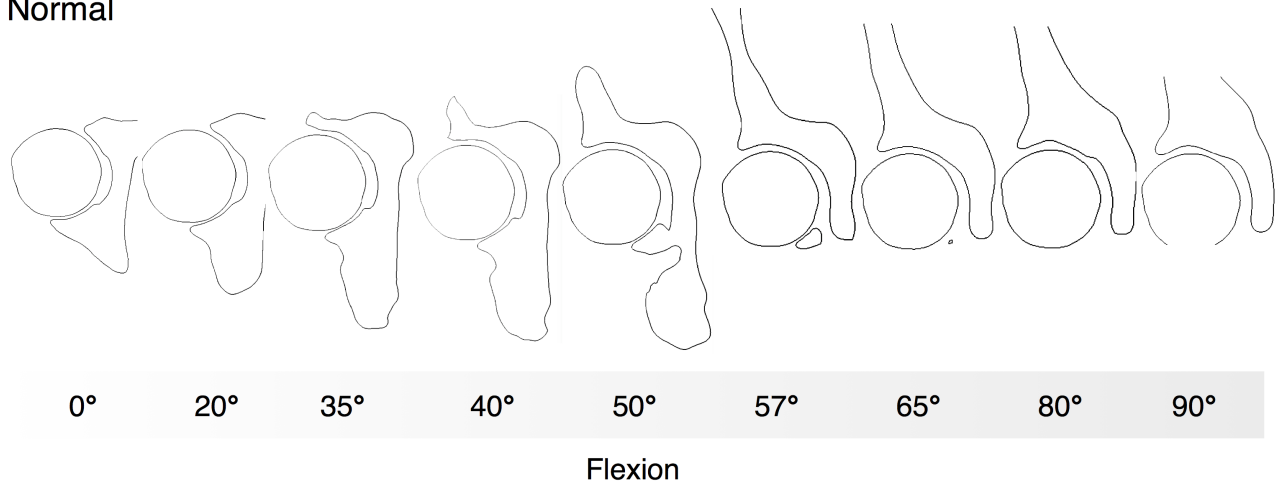
anterior superior area of the acetabulum. It was thought that greater loading could lead to greater damage, so choosing a time point where loading was greatest in the anterior direction would be more consistent with the characteristic cam-type FAI mechanism. Projecting the hip contact force vector onto the plane of internal rotation (transverse plane) showed that the time point when the hip contact force was greatest in the anterior direction was when the hip was flexed at 57° (Figure 4.9). A 2D slice of a hip joint flexed at 57° was taken from a CT scan of a cam-type FAI patient. However, at 57° , a 2D slice of the hip joint did not include a significant portion of the posterior acetabular wall. The posterior acetabular wall would help to realistically and accurately constrain the femoral head without having to define additional constraints or boundary conditions. Based on an investigation of the hip joint anatomy of a normal subject and cam-type FAI patient at various increments of hip flexion (Figure 4.11), the amount of flexion at which a significant portion of the posterior wall appeared was at 40° of flexion. Thus, the time point at which the hip was flexed at 40° was chosen (Figure 4.12). The

soft tissue structures including the labrum and articular cartilage were constructed based on literature values of the shape and size of the structures and by observing MRI's of the structures (Figure 4.13) (Abaqus 6.11) [16, 49, 61, 62, 63, 74, 75, 76].

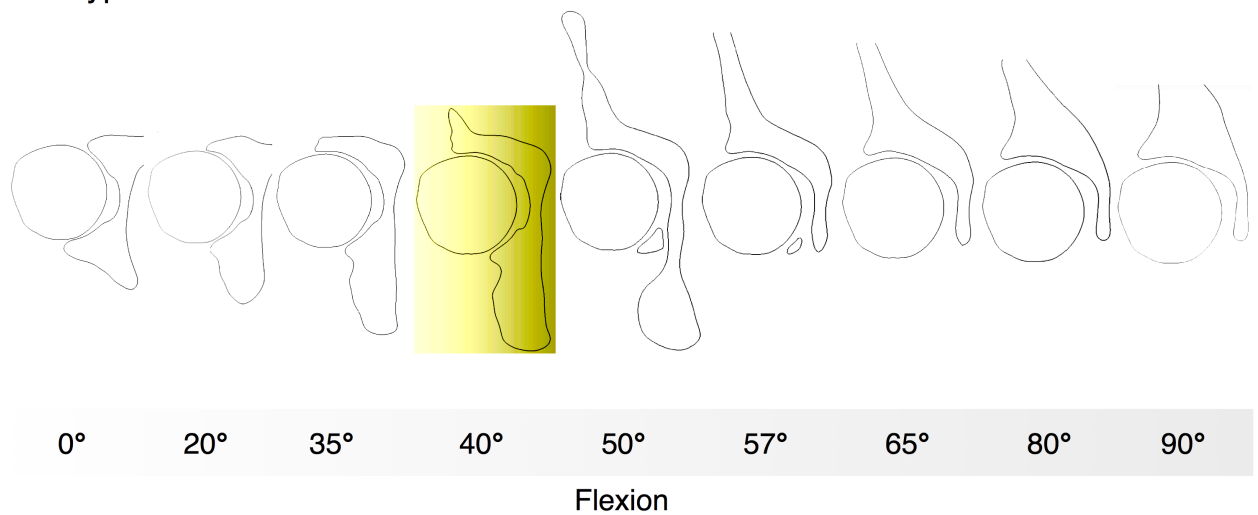
At the time point chosen the hip contact force was 1.9 N in the medial direction and 36.1 N in the anterior direction (Figure 4.9). As explained for the Flexion model setup, the hip contact

Slices of the Left Hip Joint through Transverse (XY) Plane

Normal



Cam-type FAI Patient



Lateral $\longleftrightarrow$ Medial

Anterior
 $\updownarrow$
 Posterior

Figure 4.11: 2D slices of the left hip joint of a normal (top) and cam-type FAI patient (bottom) through the transverse plane (plane of internal rotation) from CT scans at increments of hip flexion are presented.

Subject-specific Hip Angles and Hip Contact Force: Stand-to-sit

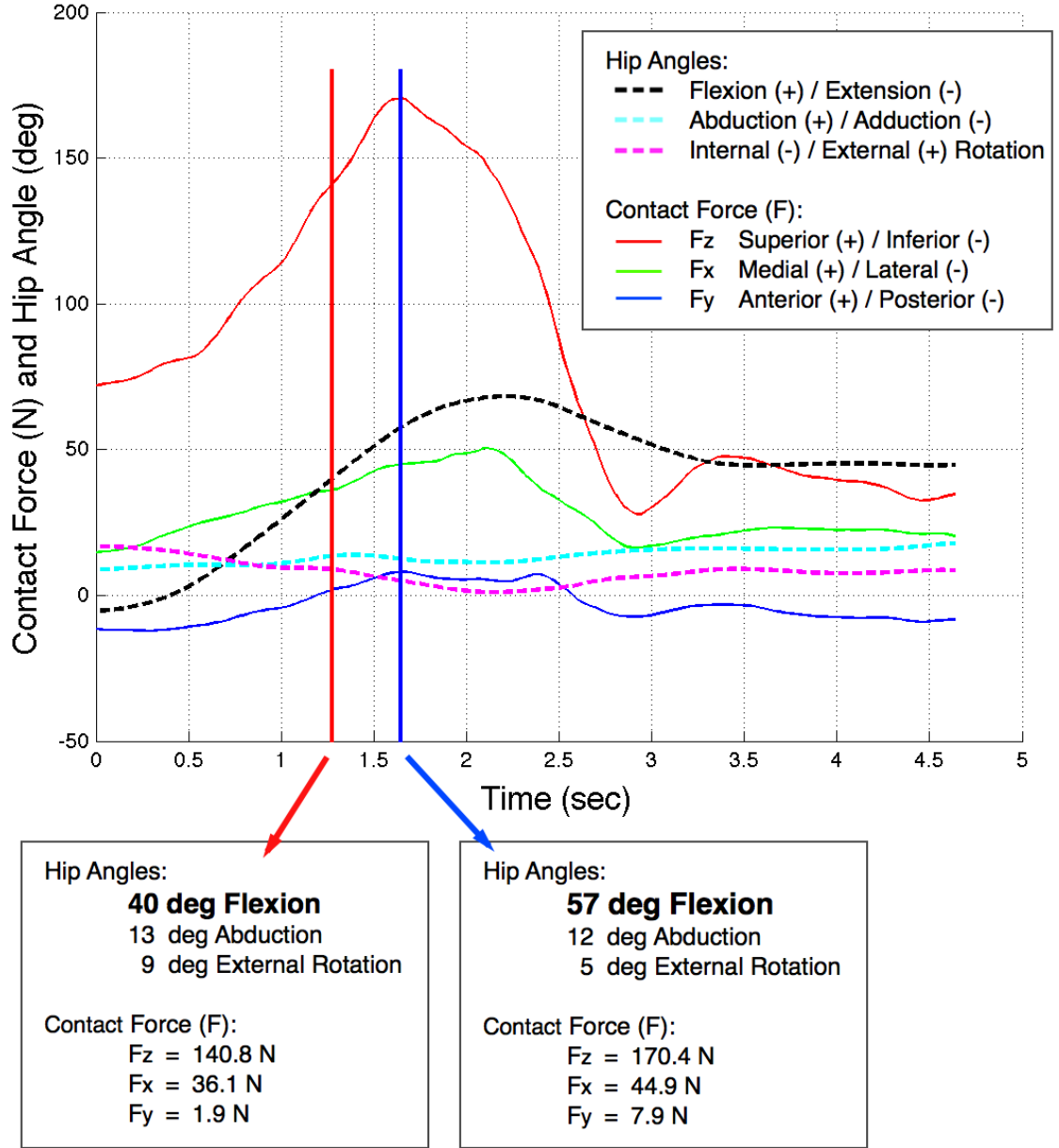


Figure 4.12: The average of four trials of hip angles and hip contact forces for the motion of sitting are presented for patient HSW who had instrumented hip implants [149].

force corresponds to the contact area over the entire hip joint, however the contact area in a 2D model is only a portion of the entire contact area. Also, assuming that the contact area for sitting is similar to the contact area during walking and stair descent, the average contact area during sitting is about 354 mm^2 [60]. Assuming that the contact area for internal rotation during sitting covers an area that is a perfect square and that a 2D model consists of a 1 mm-thick slice of the hip joint, the entire hip contact area (A) can be expressed as

$$A = L^2, \quad (4.4)$$

**Subject-specific Hip Joint Geometry:
Internal Rotation at Flexion Model
Cam-type FAI Patient**

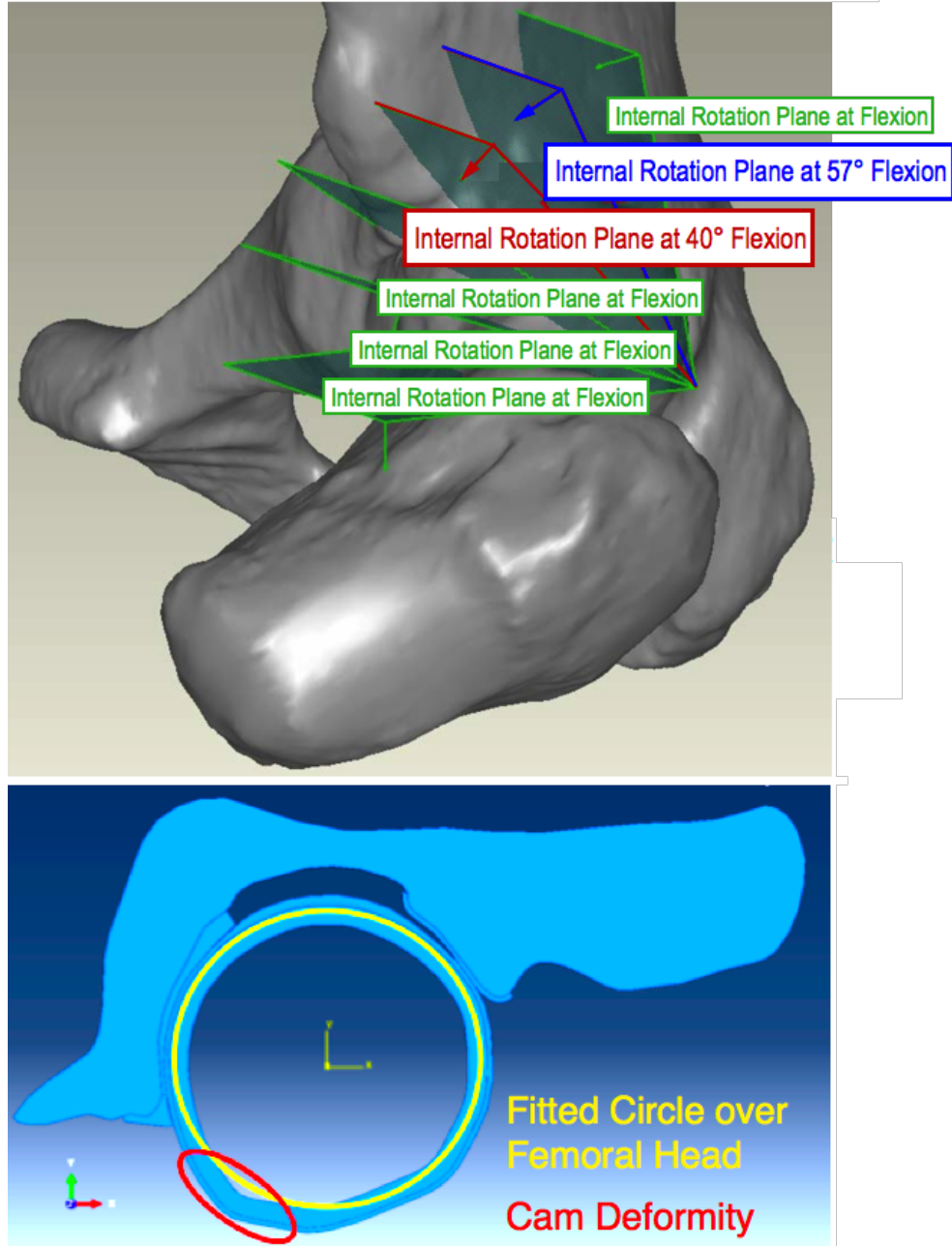


Figure 4.13: The hip joint geometry from CT scans of a cam-type FAI patient (top) is presented. Bony geometry was obtained from the transverse plane (plane of internal rotation) at 40° of flexion from the CT data, and soft tissues were reconstructed (bottom).

where L is the length of one of the sides of the square contact area. Solving this expression for L leads to the expression:

$$L = \sqrt{A} = \sqrt{354} = 18.8 \text{ mm} \quad (4.5)$$

The hip contact force (F_{2D}) of the 2D Internal Rotation at Flexion model can be expressed as

$$F_{2D} = \frac{F_{3D}}{L} = 0.0532F_{3D} \quad (4.6)$$

Therefore, the hip contact force used for this 2D Internal Rotation at Flexion model was 5% of the entire hip contact force which is 0.1 N medially and 1.92 N anteriorly.

Defining Damage

An XFEM formulation based on a cohesive segments approach was applied to the labrum and acetabular articular cartilage. A maximum strain criterion was used where the maximum strain was defined to be 0.1 in all directions. There is very little information regarding the strain at which cartilage fails due to its strain-rate dependent behaviour, so a strain value of 0.1 was arbitrarily chosen. This value of strain is under 20% strain which has been shown to be the yield strain of cartilage when loaded at $8.6\% s^{-1}$ and $12.4\% s^{-1}$ strain rates which are within physiological range [152]. For comparison, strain rates ranging from $500-1000\% s^{-1}$ would likely be comparable to those experienced in skeletal injuries associated with unrestrained motorcar impacts [92]. So 10% strain can be considered to be within physiological range but is still a significant amount of strain [152].

The evolution of damage was defined by the fracture energy in two directions – parallel and perpendicular to the interface. The fracture energy of $0.009 \text{ MPa}(mm)^{0.5}$ in the direction parallel to the interface was used which was the energy that was needed to be overcome in order for a crack to form at a 45° angle to the interface due to shear strain along the interface [135, 136]. The fracture energy of $0.09 \text{ MPa}(mm)^{0.5}$ in the direction perpendicular to the interface was used which was the energy that was needed to be overcome in order for a crack to form at a 45° angle to the interface due to shear strain perpendicular to the interface [135, 136]. Comparing these two values shows that it is ten times more difficult for a crack to form at a 45° angle to the interface due to shear strains perpendicular to the interface compared to shear strains along the interface.

It would take about 23 times more energy for a crack to form parallel to the interface compared to 45° to the interface due to shear strains along the interface [135, 136]. Since the formation of a parallel crack is much more difficult to form, a 45° angle crack was thought to be more likely

to form. It was also assumed that the progression of a crack could possibly be the combination of multiple neighbouring 45° angled cracks (resulting in delamination). Therefore, the fracture energy due to shear strains along the interface was chosen to be the fracture energy for a crack to form at an angle of 45° to the interface.

Simulating Labral Degeneration

The range of cartilage material stiffness reduction due to cam-type FAI-related damage is unknown; however, cartilage's material stiffness has been determined to reduce by about 84% due to collagenase digestion [153]. This shows that cartilage's material stiffness can reduce significantly due to a disturbed collagen structure. Damage such as a labral tear is a result of a highly disturbed collagen structure in the labrum. Assuming that the amount of collagen structure damage due to a labral tear is a fraction of the amount of collagen structure damage due to collagenase digestion, the degeneration of the labrum was simulated by reducing its material stiffness to 85% and 80% of its original stiffness, correlating to a 15% and 20% stiffness reduction, respectively. For these two values of labrum stiffness, the same modelling procedure was repeated.

4.5.2 Results

Flexion Model Results

During the activity of sitting, the hip joint of the cam-type FAI patient was flexed to 63° prior to fully sitting and then extended slightly to 55° of flexion at full-sit (Figure 4.14). The HJC was also constrained by a circle with a radius of 4 mm. Comparing shear strain results for the normal subject and cam-type FAI patient during the activity sitting, the labrum of the cam-type FAI patient was affected by the cam deformity, while the labrum in the normal subject remained unaffected (Figure 4.15). Looking at the shear strain at 15° and 23° of flexion (Figure 4.15), it was clear that the cam deformity in the cam-type FAI patient abutted the labrum in a way which produced greater shear strains close to the labral-bone interface. In the normal subject, the greatest shear strains occurred near the cartilage-bone interfaces, leaving the labral-bone interface unaffected.

A 15% reduction in material stiffness resulted in a small crack near the labral-bone interface

spanning two elements (Figure 4.16). A 20% reduction in material stiffness resulted in a larger crack near the labral-bone interface spanning eight elements.

Subject-specific Hip Joint Geometry: Cam-type FAI Patient Flexion Model

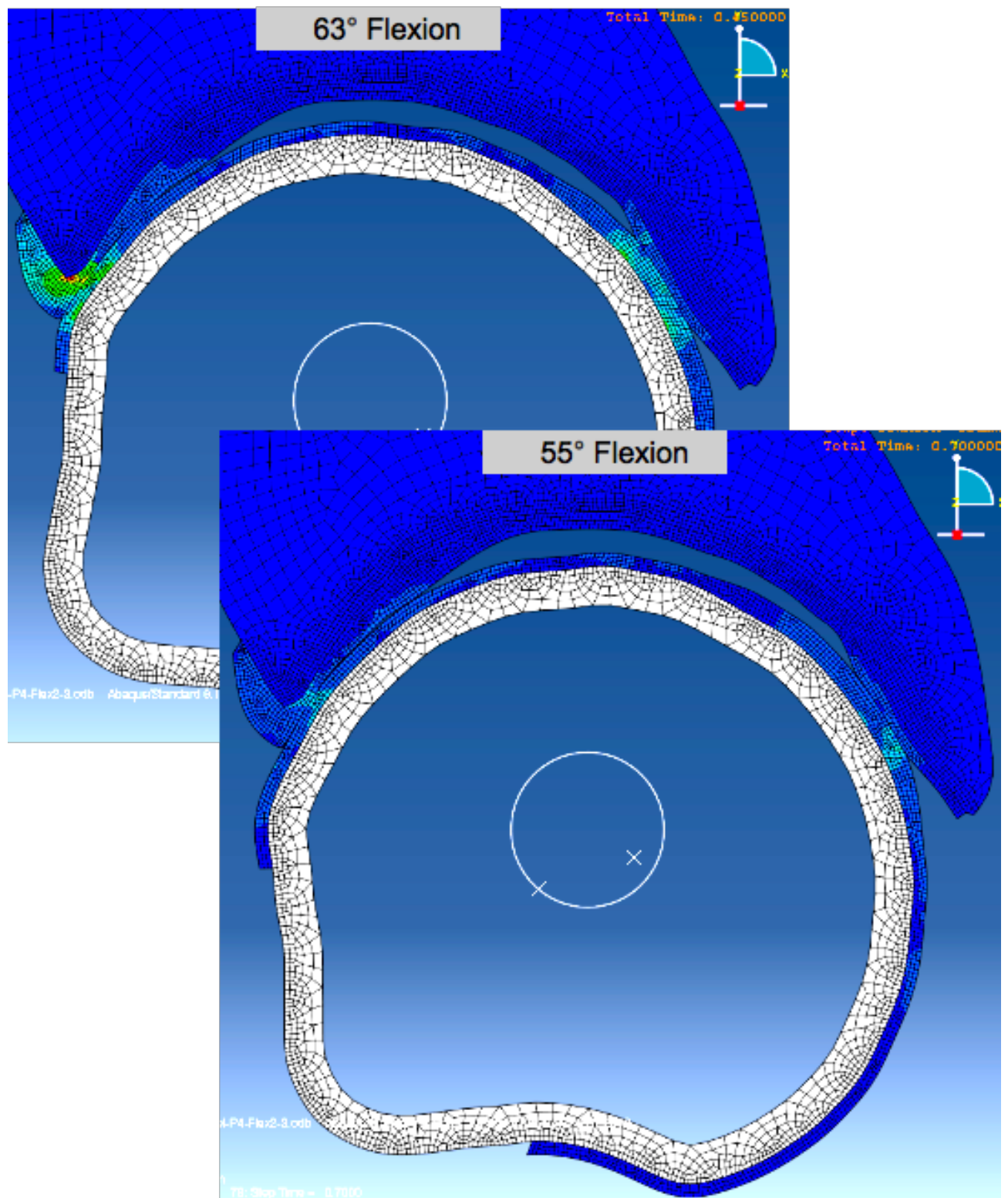


Figure 4.14: The maximum in-plane principal strain results in the hip joint of a cam-type FAI patient undergoing hip flexion during sitting are presented. The maximum “maximum in-plane principal strain” (red) was 0.15; the minimum “maximum in-plane principal strain” (blue) was -0.01. The hip flexed to 63° as the patient prepared to fully sit. Upon fully sitting, the hip extended slightly until it was flexed at 55°.

Subject-specific Flexion Model: Logarithmic Shear Strain

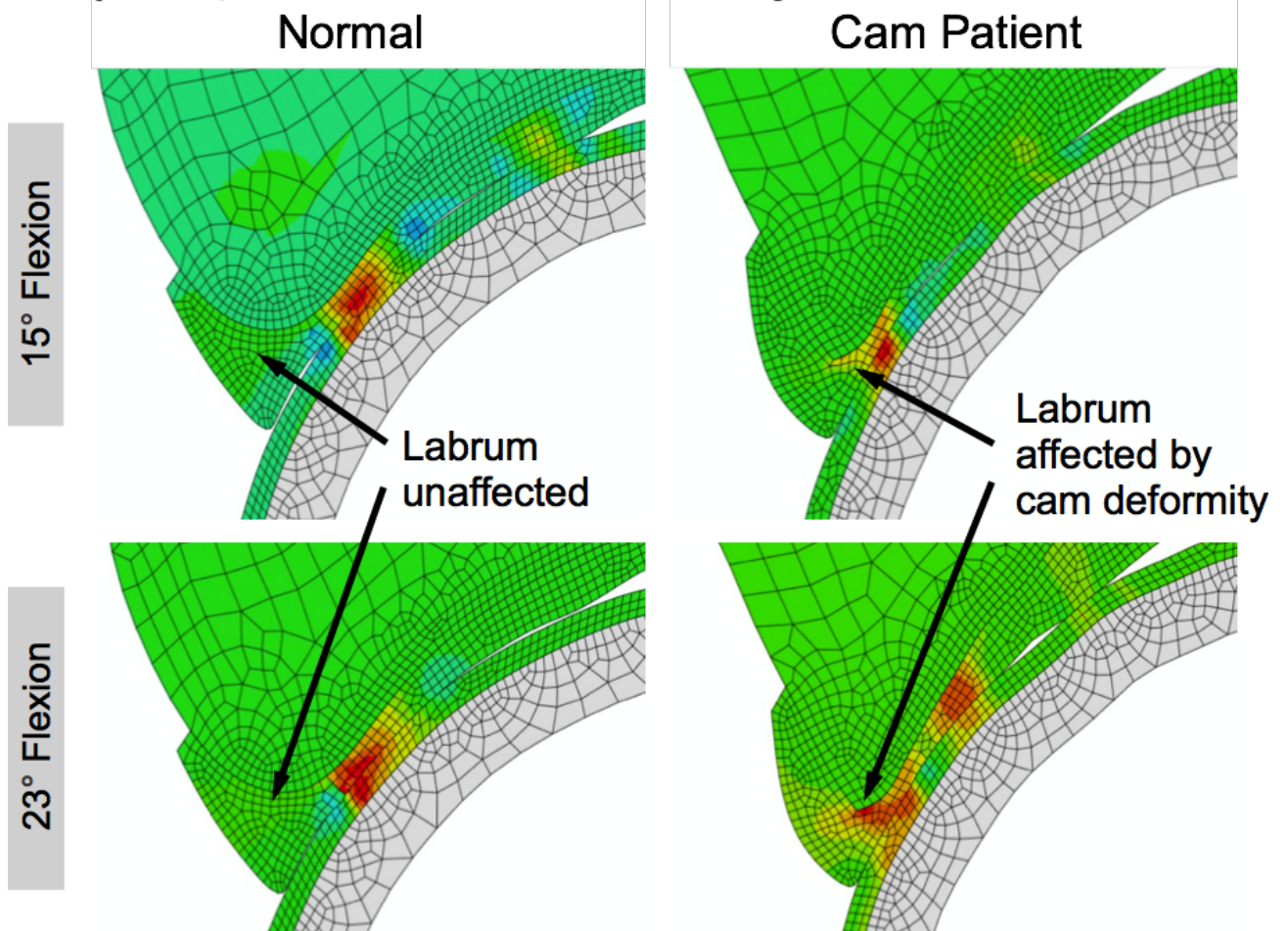


Figure 4.15: The logarithmic shear strain for a normal subject (left) and cam-type FAI patient (right) during the activity sitting at 15° (top) and 23° of flexion (bottom) are presented. Greater shear strains occurred in the labrum of the cam-type FAI patient with a maximum logarithmic shear strain (red) of 0.18 and a minimum logarithmic shear strain (blue) of -0.13 (right). A maximum logarithmic shear strain (red) of 0.11 and a minimum logarithmic shear strain (blue) of -0.11 was reached in the normal subject (left).

Subject-specific Flexion Model: Cam-type FAI Patient Damage Analysis for Labral Degeneration

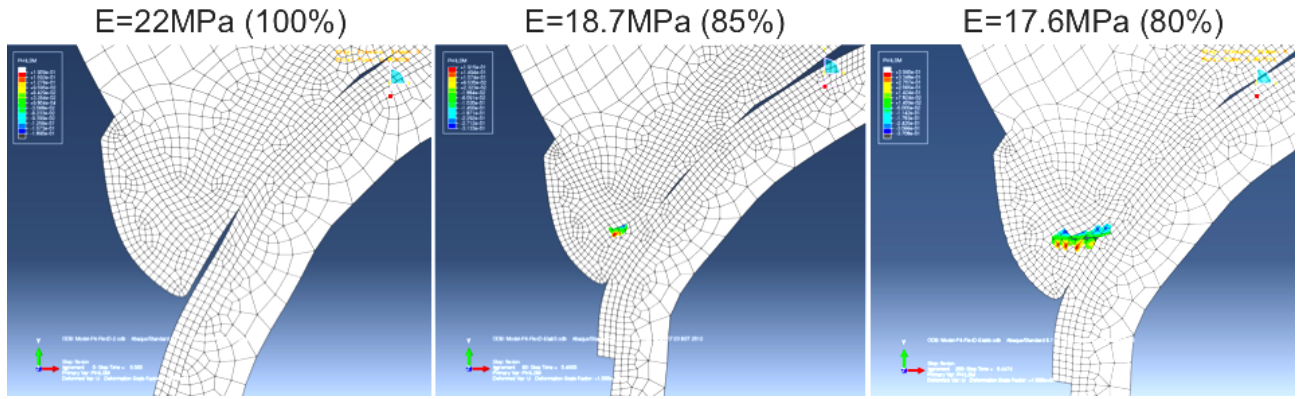


Figure 4.16: The XFEM results for labral degeneration simulation are presented. A 15% reduction in stiffness resulted in a small crack near the labral-bone interface spanning two elements (middle). A 20% reduction in stiffness resulted in a large crack near the labral-bone interface spanning eight elements (right).

Internal Rotation at Flexion Model Results

In the model where the hip joint was constrained by a circle with a radius of 2 mm, the hip joint reached a maximum of 23° of internal rotation (Figure 4.17). At 0° of internal rotation, the articular cartilage surfaces of the femoral head and anterior acetabulum were perfectly congruent. At 13° of internal rotation, the articular cartilage surfaces were still relatively congruent, however the cam deformity began to abut the labrum, causing greater strains to develop in the labrum. Difficulty in the cam deformity entering the hip joint also caused the surfaces of the femoral head and anterior acetabulum to slightly separate from each other. At 23° of internal rotation, the cam deformity caused even greater strains to develop in the labrum, and the articular cartilage surfaces of the femoral head and anterior acetabulum were clearly separated, since the cam deformity could not enter the hip joint.

Further investigation into the shear strain showed that the greatest shear strains occurred near the labral-bone interface (Figure 4.18).

Subject-specific Hip Joint Geometry: Cam-type FAI Patient

Internal Rotation at Flexion Model

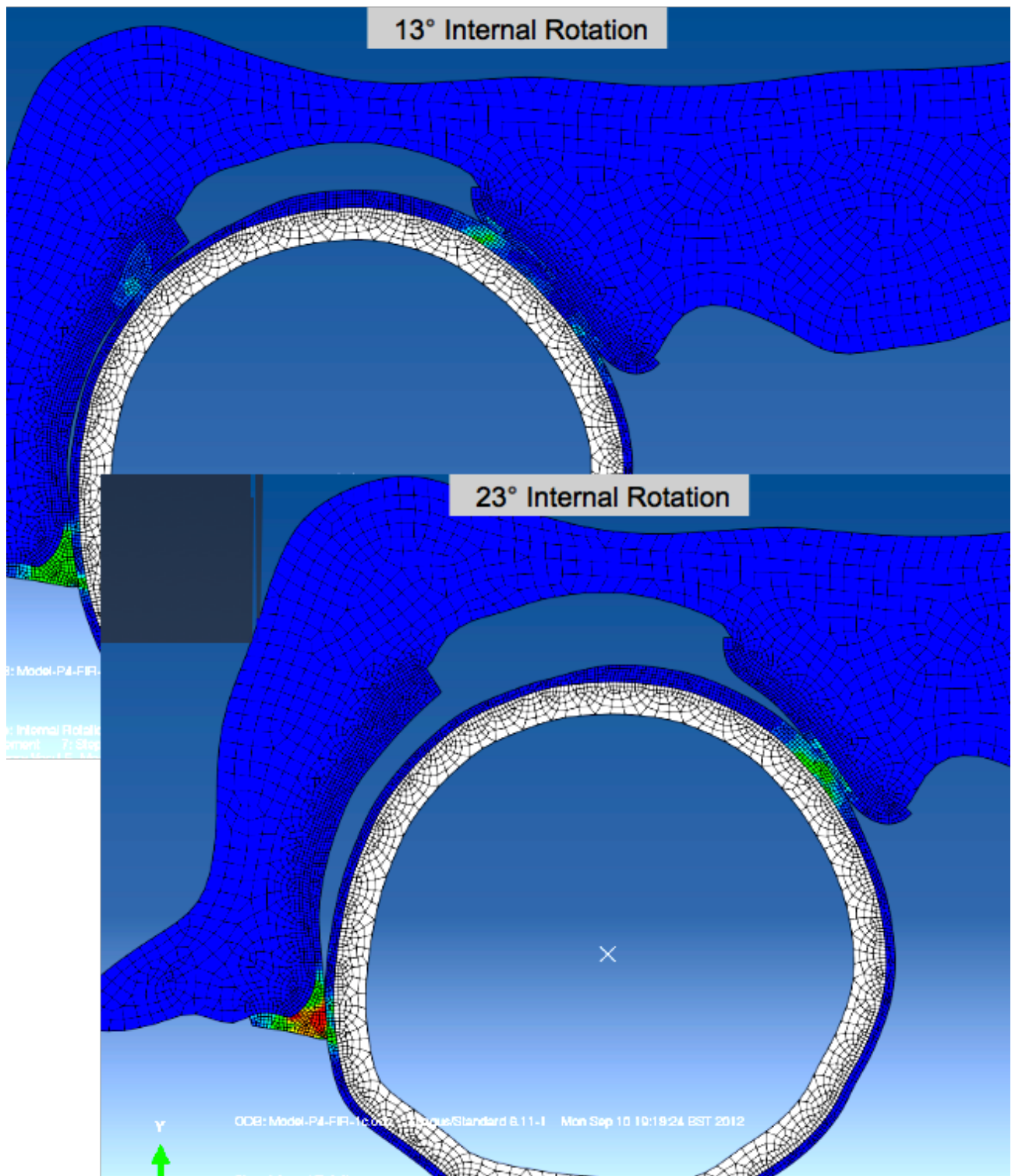


Figure 4.17: The maximum in-plane principal strain in hip joint of cam-type FAI patient undergoing internal rotation at 40° hip flexion during sitting is presented. Abutment of the cam deformity resulted in greater strains in the labrum and separation of the surfaces of the femoral head and anterior acetabulum. The maximum “maximum in-plane principal strain” (red) was 0.1; the minimum “maximum in-plane principal strain” (blue) was -0.003.

Subject-Specific Internal Rotation at Flexion Model: Cam-type FAI Patient

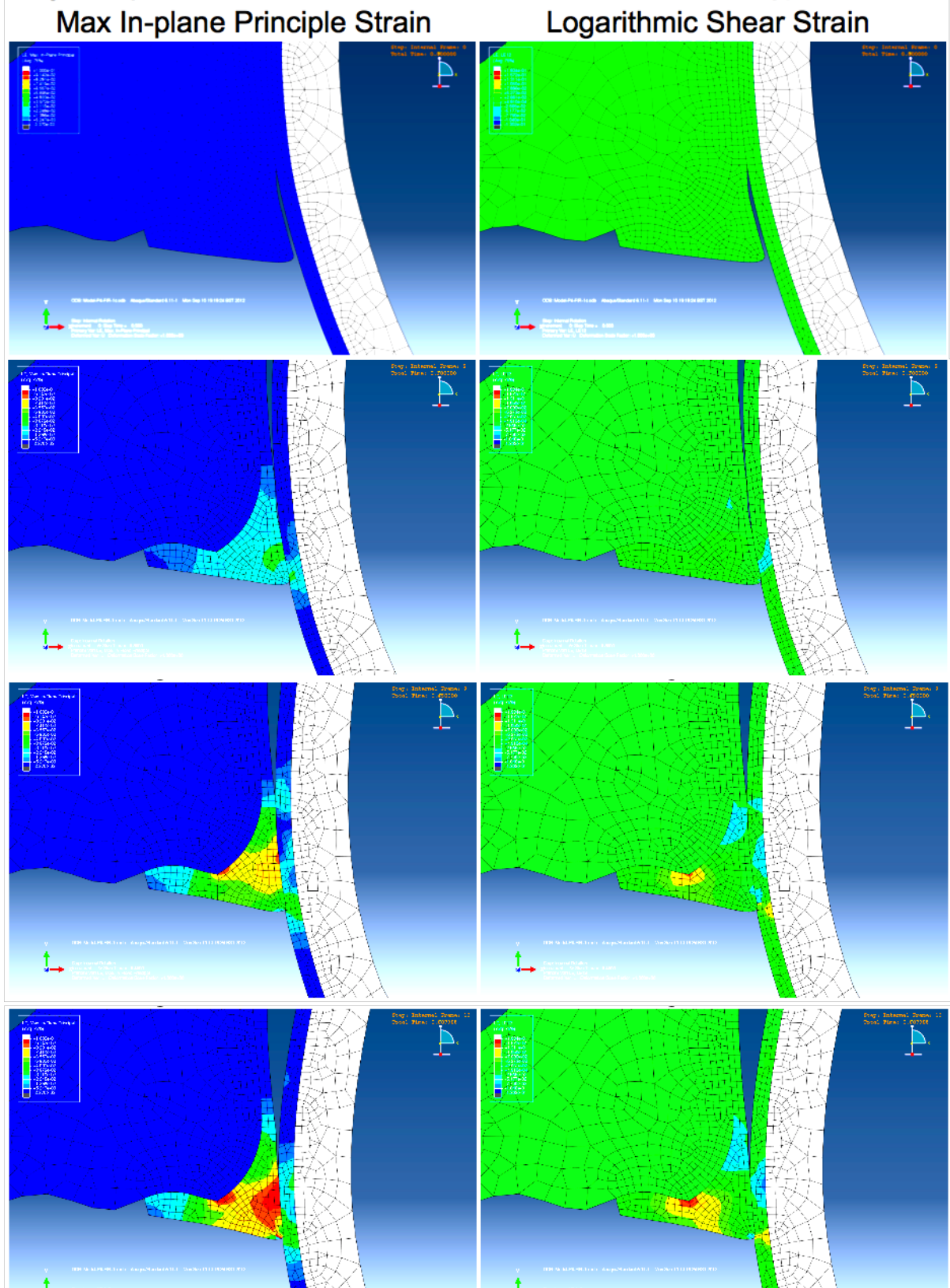


Figure 4.18: The maximum in-plane principal strain (left) and logarithmic shear strain (right) in hip joint of a cam-type FAI patient undergoing internal rotation at 40° hip flexion during sitting are presented. Greatest logarithmic shear strains occurred near the labral-bone interface. The maximum “maximum in-plane principal strain” (red) was 0.1; the minimum “maximum in-plane principal strain” (blue) was -0.003 (left). The maximum logarithmic shear strain (red) was 0.18; the minimum logarithmic shear strain (blue) was -0.13 (right).

4.5.3 Discussion

The shear strains that developed in the labrum of the cam-type FAI patient occurred due to the wedging of the cam deformity into the hip joint. However, it was also observed that the labral-bone junction in the cam-type FAI patient in the Flexion and Internal Rotation at Flexion models were not perfectly smooth. There was a less rounded edge at the labral-bone junction which was an area that was more susceptible to developing greater shear strains. To ensure that this sharper edge would not give way to point load-like results, this sharper edge was slightly rounded out in future models.

In the Internal Rotation at Flexion model, the HJC of the femoral head translated outwards from the hip joint as the cam deformity entered the hip joint. There is little information about the normal ranges of translation of the HJC during extreme or daily functional activities. It is, therefore, unknown whether the amount of translation of the HJC was realistic or not. An arbitrarily chosen, but conservative estimate of realistic HJC translation of 1 mm was applied for subsequent models by constraining the motion of the HJC by a circle with a radius of 1 mm.

Also, it is important to note that the materials in this model have been defined to be linear elastic, meaning that the results are only valid for activities that involve dynamic, rapid, or impact loading. This involves any activity where cartilage is subjected to fast strain rates ($> 0.1 \text{ s}^{-1}$) and loading occurs under a timeframe of less than 1 s [90, 91, 92, 93]. Under such guidelines, most functional activities such as the ones used in these models, are considered to involve rapid loading.

4.5.4 Conclusion

Labral tears are commonly seen clinically in cam-type FAI patients. Such damage was reproduced in a cam-type FAI patient, whereas the labrum of a normal subject was virtually unaffected by shear strains and free from abutment by the femoral head. More specifically, the motions of flexion and internal rotation at flexion caused large shear strains to develop in the labrum of a cam-type FAI patient, and when the labrum was increasingly degenerated through material stiffness reduction, increasingly larger cracks occurred near the labral-bone junction. Therefore, it is believed that the level of sophistication of this model including subject-specific

geometry and subject-specific motion and load data is sufficient for simulating cam-type impingement.

4.6 Stage 5. Utilising a Subject-specific FE model for Parameter Studies

After successfully developing FE models that realistically simulated cam-type impingement, the Internal Rotation at Flexion FE model was chosen to be altered to simulate not only labral tearing, but cartilage delamination as well. It was also chosen to be altered to allow parameter studies to be performed which investigated various geometric parameters. The Internal Rotation at Flexion model will be the focus of future models since internal rotation at flexion is the most characteristic damaging motion associated with cam-type FAI.

By altering the femoral geometry, whilst leaving the acetabular geometry unaltered, parameters such as cam protrusion and width and hip joint clearance were able to be idealised. A pilot study which parametrised cam protrusion was performed to establish what an appropriate range for small and large protrusion might be and to examine if altering cam protrusion could possibly lead to different types of damage.

4.6.1 Methods

Defining Geometry

The geometry created for the Internal Rotation at Flexion model in Stage 4 in Section 4.5.1 was used. The acetabular geometry was unaltered, whilst the femoral geometry was modified so that geometric parameters such as cam protrusion and width and hip joint clearance were able to be idealised.

Defining Geometric Parameters

The geometric parameters under investigation included hip joint clearance, cam protrusion, and cam width. Hip joint clearance is referred to as the shortest distance from the acetabular bone at the labral-bone junction to the femoral bone (Figure 4.19). If a fitted circle is drawn over the bone of the femoral head, the cam protrusion can be measured to be the radial distance

from the edge of the fitted circle to the peak of the cam deformity (Figure 4.19). The cam protrusion in addition to the radius of the femoral head (radius of the fitted circle) makes up the cam radius (Eqn 4.7), and the cam protrusion can be calculated to be a percentage (P%) of the hip joint clearance (Eqn 4.8).

$$\text{Cam Radius} = \text{Femoral Head Radius} + \text{Cam Protrusion}, \quad (4.7)$$

where

$$\text{Cam Protrusion} = \text{Hip Joint Clearance} * P\%, \quad (4.8)$$

where P is some chosen percentage.

The cam width in a particular slice of the hip joint was defined to be an angle measurement. When a circle is fitted over the bone of the femoral head, two lines can be drawn, each from the HJC, to the two junctions at which the cam deformity meets the edge of the circle. The angle that these two lines make is defined to be the cam width (Figure 4.19).

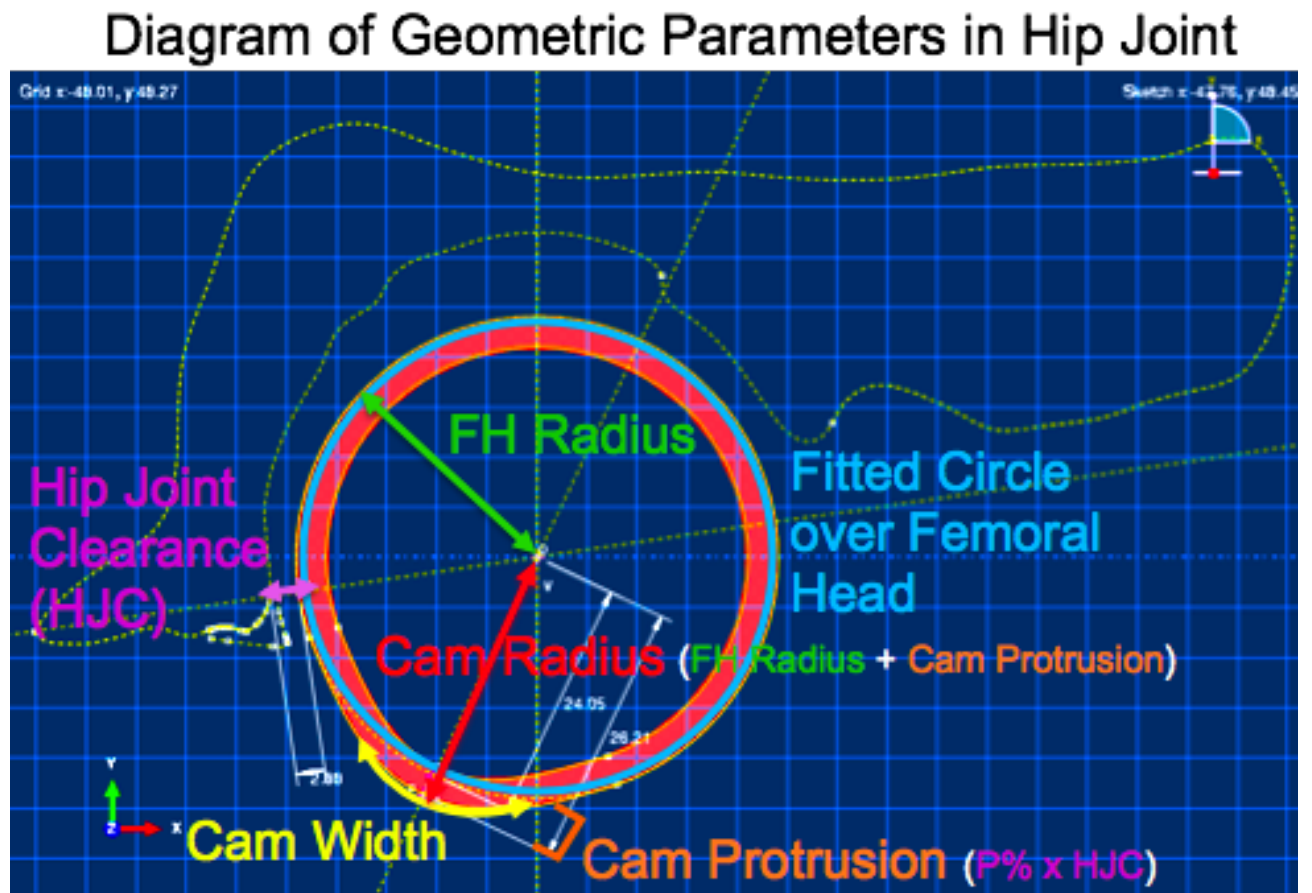


Figure 4.19: The parameters of the hip joint to be investigated using FE modelling are depicted.

Mesh Generation, Assigning Boundary Conditions, and Defining Interactions, Analysis Steps, and Damage

Similar mesh generation, boundary conditions, interactions, analysis steps, and damage definitions were applied in this model as they were applied in the Internal Rotation at Flexion model in Stage 4 in Section 4.5.1.

Simulating Soft Tissue Degeneration

The degeneration of the soft tissues in the hip joint were simulated by reducing the material stiffness of the articular cartilage and labrum. The material stiffness of the articular cartilage was reduced to 50% of its original stiffness, while the material stiffness of the labrum was reduced to 60% of its original stiffness.

4.6.2 Pilot Study

Two models were created for this pilot study with differently sized cam protrusions with the goal of developing models which would each lead to a different type of damage – labral or articular cartilage damage. First, a small hip joint clearance was established by creating a layer of articular cartilage on top of bone which was in normal range and which also optimised the congruency of the hip joint surfaces when the hip was in the neutral position. This led to cartilage thicknesses ranging from 0.8 to 2.7 mm on the acetabulum and 0.8 to 1.6 mm on the femoral head, and a hip joint clearance of 2.88 mm.

The cam deformity of the subject-specific hip joint geometry in Stage 4 had already shown that it abutted the labrum during internal rotation at flexion and may possibly lead to labral damage. Therefore, its shape and size served as a guideline for a cam deformity that might lead to labral damage. Using it as a guideline led to creating an idealised model of a hip joint with a cam protrusion equivalent to 75% ($P=75$) of the hip joint clearance which was 2.16 mm and a cam width of 90° .

To create a model that could possibly lead to articular cartilage damage, it was hypothesised that the cam deformity would have to be relatively small to enable it to enter the hip joint, but big enough to be destructive to the soft tissue. Therefore, a second model was created also

with a small hip joint clearance of 2.88 mm and a cam width of 90°, but with a cam protrusion equivalent to 25% (P=25) of the hip joint clearance which was 0.72 mm.

4.6.3 Results

The results of the model with a larger cam protrusion (Figure 4.20) led to expected results (similar to the model in Stage 4) where the cam deformity abutted against the labrum during internal rotation at flexion and caused shear strains to develop near the labral-bone junction. XFEM results led to a small crack spanning 1 element near the labral-bone interface in the labrum (Figure 4.21). Simulating soft tissue degeneration resulted in further cracking of the labral tissue near the labral-bone interface, creating a crack that spanned 8 elements.

The results of the model with a smaller cam protrusion (Figure 4.22) led to the results that were hypothesised where the cam deformity was small enough to enter the hip joint, but was large enough to cause shear strains to develop near the cartilage-bone interface as the cam deformity was wedged into the hip joint. XFEM results did not lead to any crack formation in the soft tissues, however, simulating tissue degeneration led to a crack near the cartilage-bone interface in the anterior acetabular articular cartilage that spanned 3 elements (Figure 4.23).

Large Cam Deformity (P=75%)

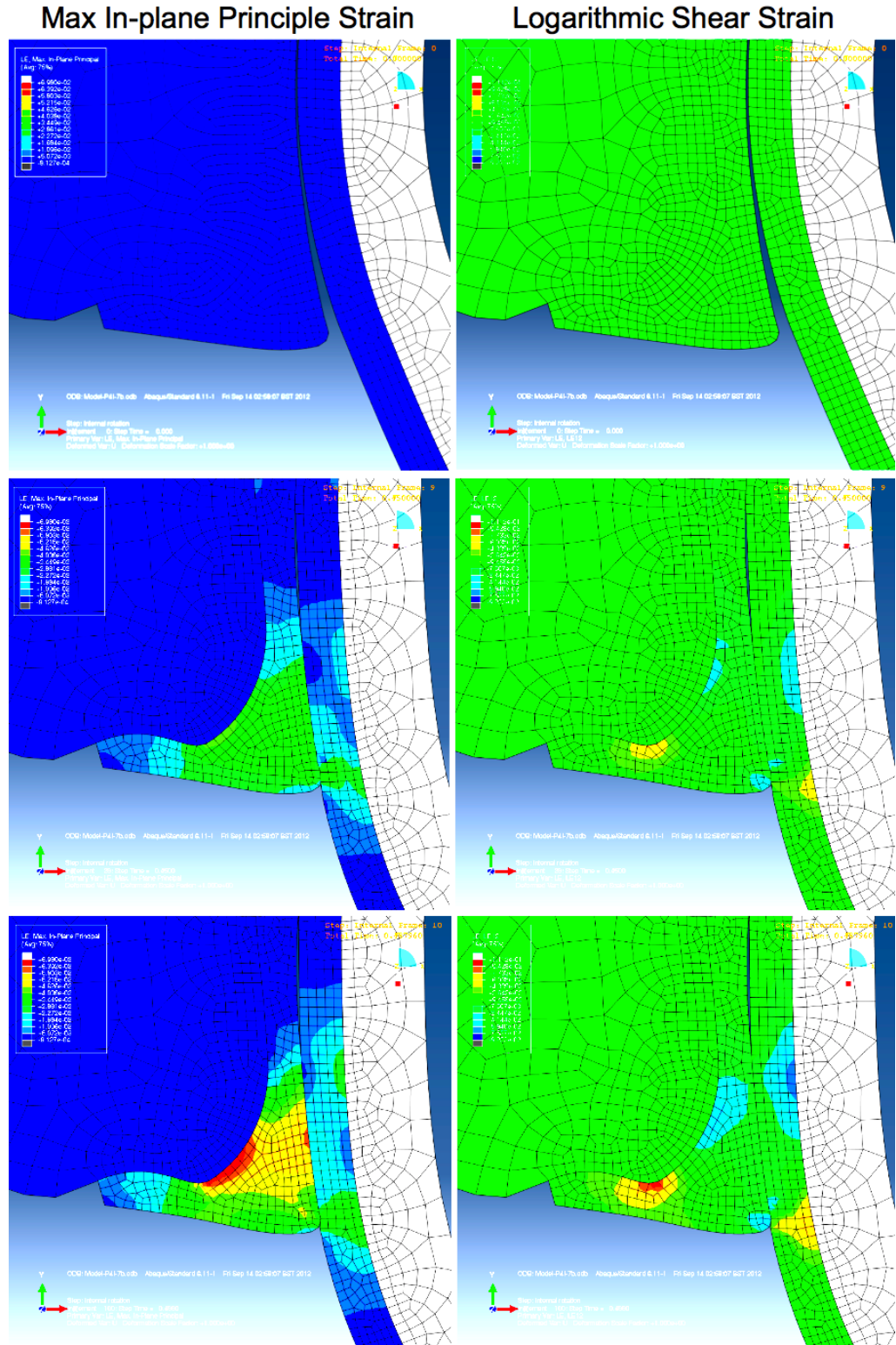


Figure 4.20: The maximum in-plane principal strain (left) and logarithmic shear strain results (right) for a hip with an idealised large cam deformity undergoing internal rotation at flexion are presented. The hip joint was modelled to have small hip joint clearance, a relatively large cam protrusion (75% hip joint clearance), and a cam width of 90°. As expected, the cam deformity abutted against the labrum during internal rotation at flexion and caused shear strains to develop near the labral-bone junction. The maximum “maximum in-plane principal strain” (red) was 0.07; the minimum “maximum in-plane principal strain” (blue) was -0.0008 (left). The maximum logarithmic shear strain (red) was 0.11; the minimum logarithmic shear strain (blue) was -0.09 (right).

Large Cam- Internal Rotation Model Damage Analysis for Labral Stiffness Reduction

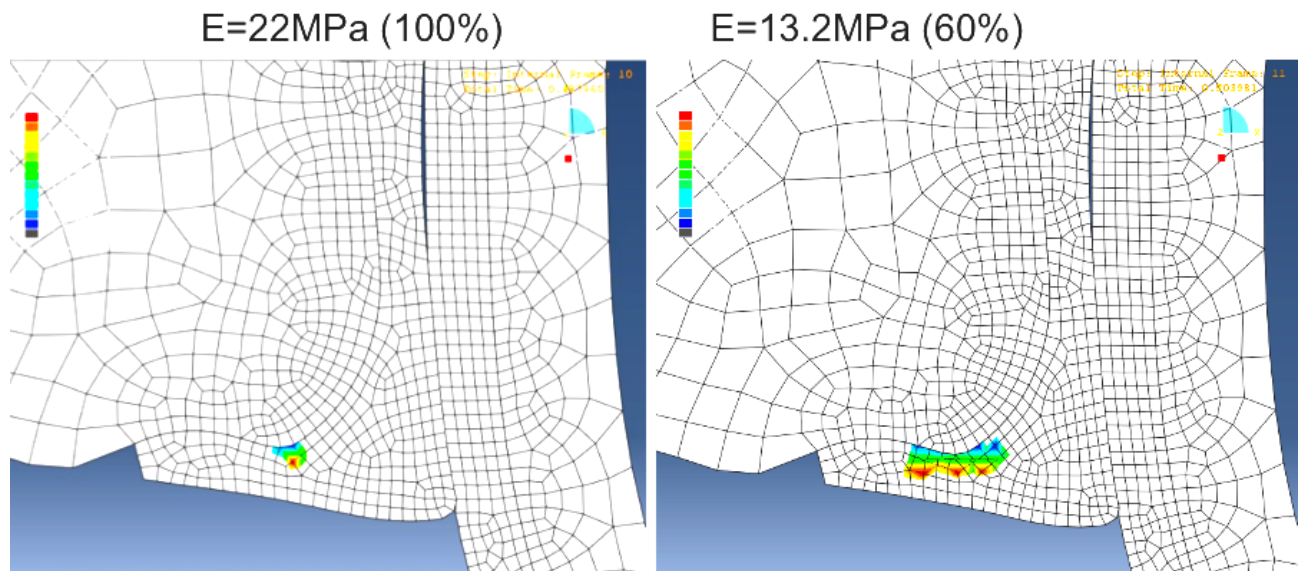


Figure 4.21: The XFEM results for labral degeneration of a hip joint with a large cam deformity undergoing internal rotation at flexion are presented. Decreasing the labral stiffness by 40% resulted in a larger XFEM crack near the labral-bone interface due to a large cam deformity abutting against the labrum during internal rotation at flexion.

Small Cam Deformity (P=25%)

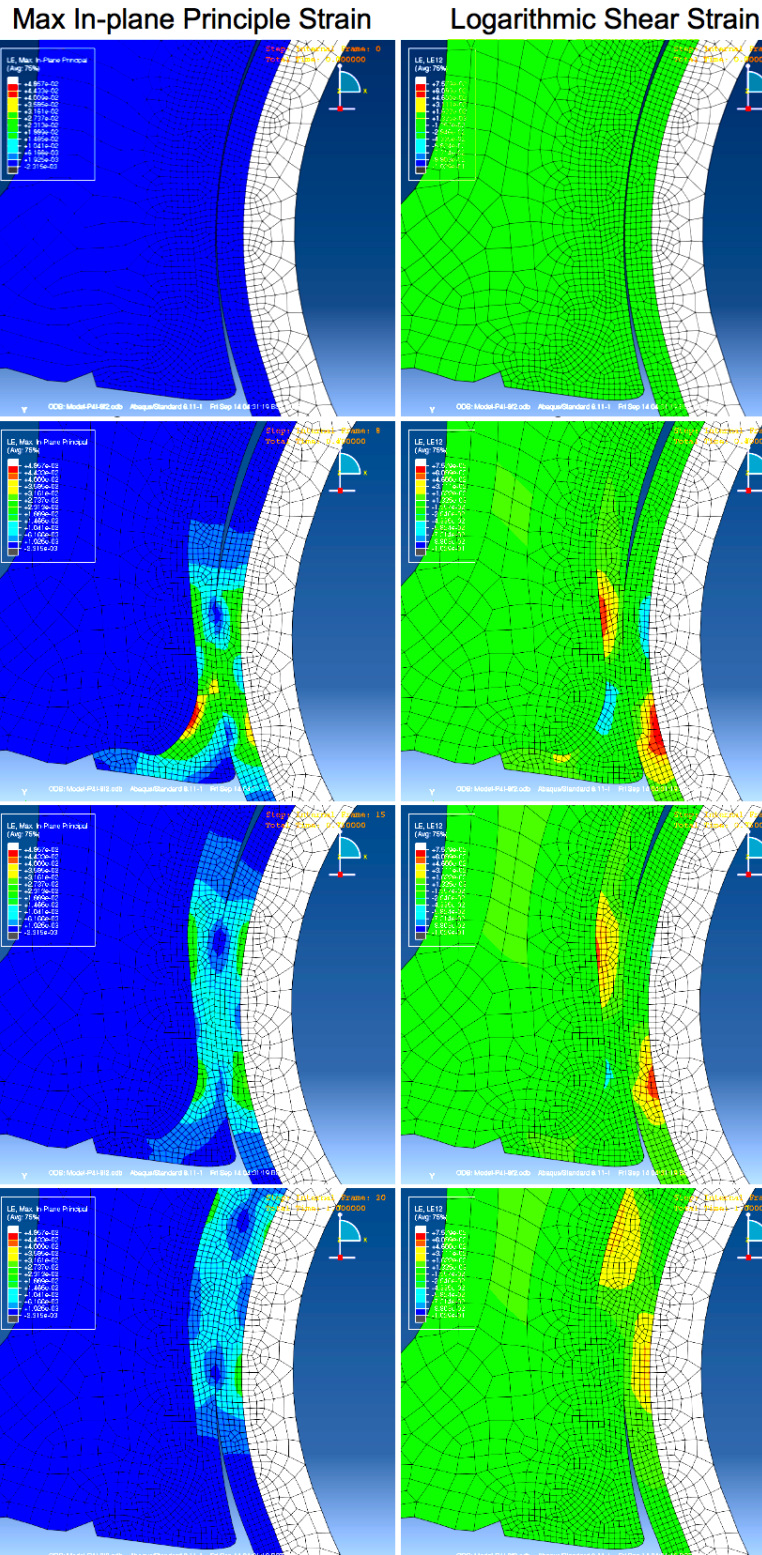


Figure 4.22: The maximum in-plane principal strain (left) and logarithmic shear strain results (right) for a hip with an idealised small cam deformity undergoing internal rotation at flexion are presented. The hip joint was modelled to have small hip joint clearance, a relatively small cam protrusion (25% hip joint clearance), and a cam width of 90°. The cam deformity entered the hip joint, causing shear strains to develop near the cartilage-bone interface. The maximum “maximum in-plane principal strain” (red) was 0.05; the minimum “maximum in-plane principal strain” (blue) was -0.002 (left). The maximum logarithmic shear strain (red) was 0.08; the minimum logarithmic shear strain (blue) was -0.10 (right).

Small Cam- Internal Rotation Model

Damage Analysis for Cartilage Stiffness Reduction

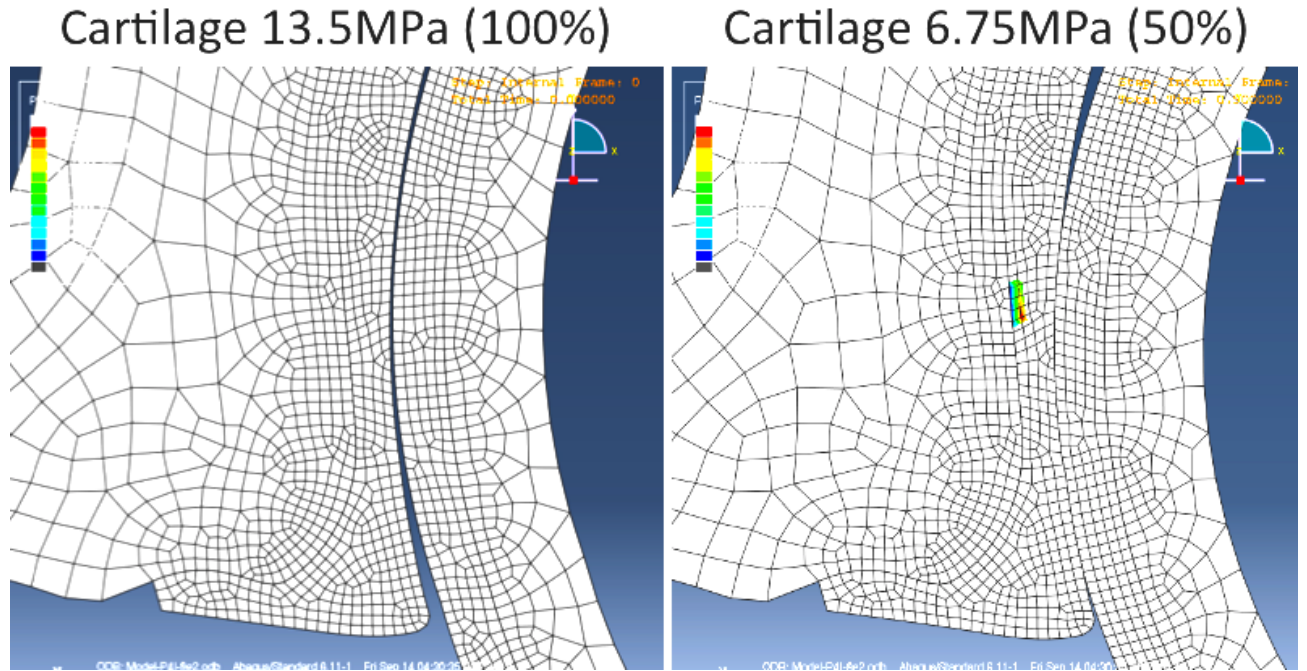


Figure 4.23: The XFEM results for articular cartilage degeneration of a hip joint with a small cam deformity undergoing internal rotation at flexion are presented. Decreasing the articular cartilage stiffness by 50% resulted in an XFEM crack near the labral-bone interface due to a small cam deformity being wedged into the hip joint during internal rotation at flexion.

4.6.4 Conclusion

The chosen parameterisation of cam protrusion of 25% ($P=25$) and 75% ($P=75$) of the hip joint clearance successfully led to the two main types of damage caused by cam-type FAI. A cam deformity with a small cam protrusion ($P=25$) was able to enter the joint, causing large shear strains and damage in the region of the articular cartilage-bone interface, which could result in cartilage delamination if the damage progressed further. A cam deformity with a large cam protrusion ($P=75$) abutted the labrum, causing large shear strains and damage at the labral-bone junction, which was representative of a labral tear.

4.7 Overall Chapter Conclusion

An FE model has been created which can realistically simulate hip impingement. Idealised 2D rectangular and circular geometries were found to not be sophisticated enough to simulate hip impingement realistically. By introducing subject-specific geometry, load, and motion in

a 2D model, impingement was able to be simulated realistically. Large shear strains in the labrum and labral cracking as a result of labral degeneration were able to be produced for a cam-type FAI patient, whereas, the labrum of a normal subject remained virtually free from any abutment or shear strains.

By altering the femoral head geometry, cam protrusion was idealised and parametrised, and damage to the cartilage and labrum was able to be simulated under different parameters. The model with a small cam protrusion was able to simulate the initiation of cartilage delamination, whereas the model with a large cam protrusion was able to simulate labral tearing. A model able to demonstrate these two types of damage has never been previously presented. It has great potential in providing further insights into how cartilage delamination may be initiated and progressed through more refined parameter studies which will be conducted in later chapters. However, the shear strain results of the impingement FE model must first be validated in order for it to provide meaningful results and conclusions. Therefore, an *in vitro* study was conducted and will be described in the next chapter to validate the shear strain results of the impingement FE model. *In vitro* damage tests were also conducted and will be described in the next chapter to confirm whether cartilage can be delaminated by being subjected to repetitive shear strains and compressive and shear loading as predicted by the impingement FE model developed in this chapter.

Chapter 5

Validation Experiment of 2D FE Model Results

5.1 Introduction

Validation of the impingement FE model developed in Stage 5 of Chapter 4 was essential in ensuring the value and accuracy of its produced results. It was also important to verify that the loading conditions in the hip joint which cause shear strains to develop at the acetabular articular cartilage-bone interface can indeed cause cartilage to delaminate from its underlying bone – a commonly suggested hypothesis which has never previously been demonstrated *in vitro*.

To validate the impingement FE model, *in vitro* experiments which loaded cartilage-bone samples under similar conditions experienced within the hip joint were conducted. Validation FE models were created to mimic the setup of the experimental tests, and experimental and FE results were compared based on shear strain distribution and value. The validation FE models were created to have the same material and contact representation and analysis framework to the impingement FE model described in Stage 5 of Chapter 4. In this way, the key model parameters of the impingement FE model could be tested for validity. A mesh convergence test was also performed for the validation FE models in this chapter.

To investigate the mechanism by which cam-type FAI causes shear strains that lead to cartilage delamination, repetitive compressive and shear loading at fast physiological strain rates were applied to cartilage-bone samples until damage occurred.

The sample preparation for the experiments consisted of preparing slices of cartilage-bone

samples from porcine knee specimen and embedding them in bone cement in a sample holder. A speckle pattern was applied to the cut cartilage surfaces of the samples used for the validation experiments. This allowed digital images of the speckled sample surface to be captured prior to loading and incrementally during loading so that a digital image correlation (DIC) method could be used to calculate the deformation in the tissues and thus, strain in the tissues. This could then be compared to the FE calculated values.

5.2 Methods

5.2.1 Sample Preparation

Cartilage-bone samples used in this study were obtained from two porcine knee specimen obtained from a local supplier. The knees were immediately dissected to remove all muscle, tendons, and ligaments to reveal the cartilage surfaces of the femoral condyles and tibial plateau. The femoral shaft of the femur was clamped while 1 cm-thick slices of cartilage-bone samples were carefully cut using a saw (Figure 5.1). The surfaces of the anterior and posterior femoral condyles had relatively consistent curvature along the sagittal plane, so samples were cut along these directions. The samples were also cut so that the saw was as close to perpendicular to the surface of the cartilage as possible to extract planar slices of cartilage-bone where the cartilage thickness and curvature of the cartilage-bone interface were as uniform as possible through its thickness. This produced samples that appeared as if they had been cut from a thicker part of a cylindrical shape with a cartilage layer around its circumference. Samples from the tibial plateau were not used because the surface of the tibial plateau was slightly warped, which would not have led to ideal loading for the experiments conducted in this study.

After the samples were cut, they were wrapped and soaked in saline solution to keep them hydrated and stored in petri dishes. The samples that were not immediately used were kept in a refrigerator at 4°C for no longer than 48 hours. Samples prior to testing were trimmed to fit into a sample holder, and bone cement was poured into the sample holder to fix the samples into place. Before the hardening of the bone cement, samples were held using forceps and slightly adjusted to eliminate any unwanted tilting of the sample. The sample was positioned so its cartilage layer formed a perfect arc from the front aspect of the embedded sample and so the cartilage surface of the apex of the sample from the side aspect lay as flat or horizontally as

Femoral Condyle of Porcine Knee

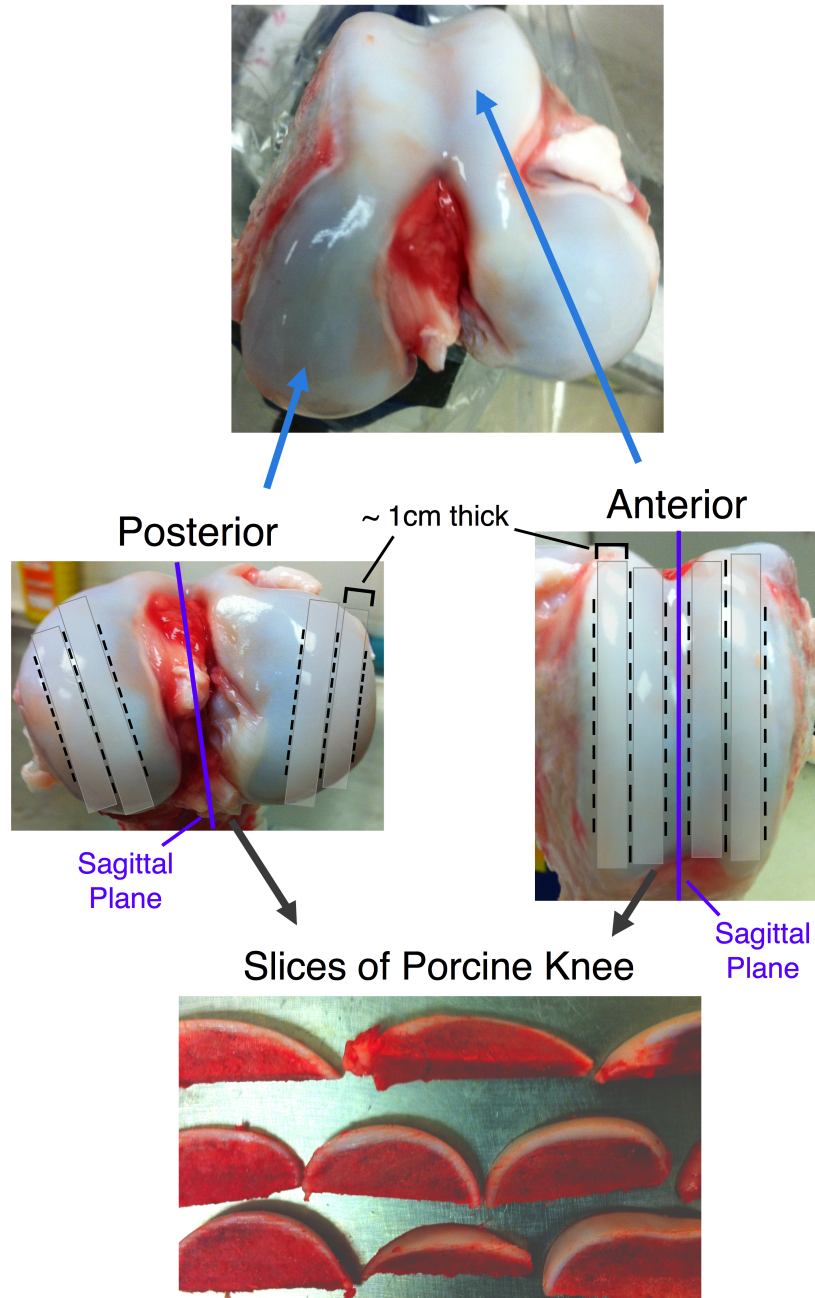


Figure 5.1: Cartilage-bone samples were cut from the anterior and posterior regions of the femoral condyle of porcine knees. 1 cm-thick slices of cartilage-bone were cut so the cartilage thickness and curvature of the cartilage-bone interface were as consistent as possible through the sample's thickness.

possible. Whenever the sample was not being tested, it was kept hydrated with saline solution.

5.2.2 Application of Speckle Pattern

A speckle pattern was applied to the cut cartilage surface so that the speckle pattern was distributed across the superficial, middle, and deep layers of cartilage and the subchondral bone (Figure 5.2). Printer toner was used to form the speckles of the speckle pattern. A plastic

dropper was used to extract some printer toner from a dish and dispense the printer toner onto samples by finger-tapping the tip of the dropper so that the toner would fall out onto the sample. The target speckle pattern consisted of speckles that densely covered the sample surface, while smudging, clumping, or too much overlapping of speckles was avoided (Figure 5.3).

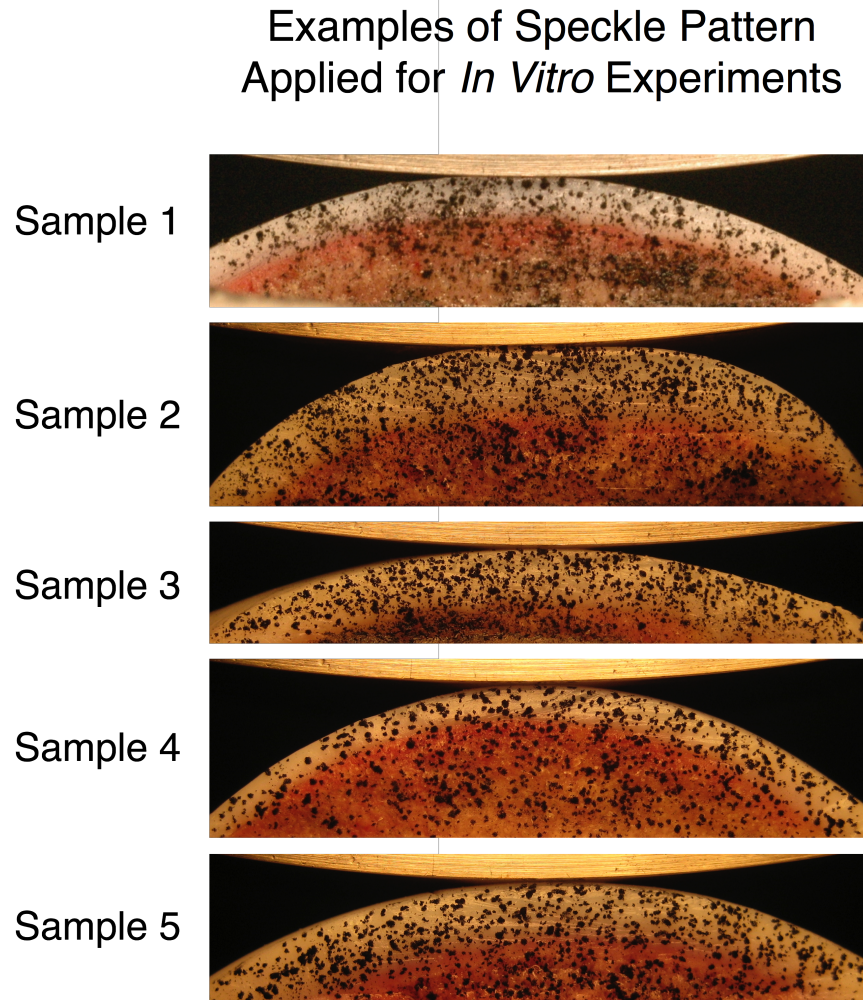


Figure 5.2: Examples of speckle patterns applied to cartilage-bone Samples 1 to 5 for *in vitro* validation experiments are shown.

Examples of Poorly Applied Speckle Patterns

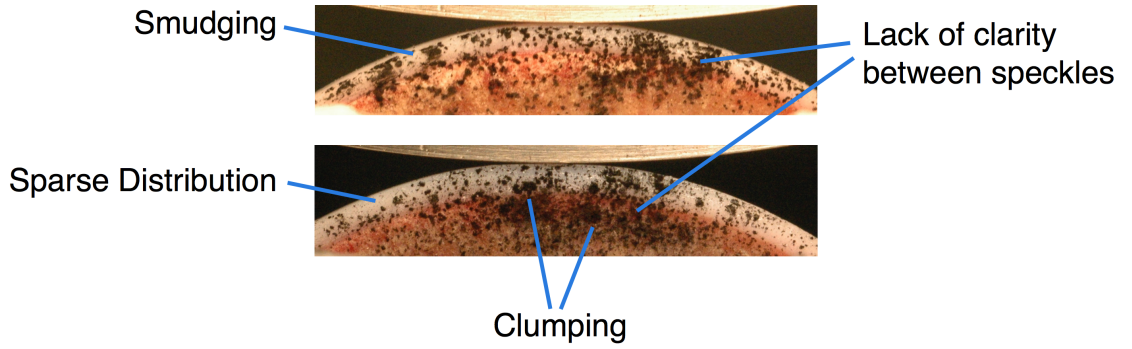


Figure 5.3: Examples of poorly applied speckle patterns are presented which show the smudging and clumping of speckles, a sparse distribution of speckles, and a lack of clarity between speckles.

5.2.3 *In Vitro* Validation Experiments

Five cartilage-bone samples were chosen for the validation experimental tests. These five samples were chosen based on the uniformity of the thickness of their cartilage layers and the consistency of the curvature of their cartilage-bone interface. The cartilage thickness (at the apex of the sample) of the five test specimen that were used was on average 2.8 mm (± 0.9 mm) ranging from 1.8 to 4 mm (Table 5.1). The curvature of the cartilage-bone interface of the samples were measured by fitting a circle to the curvature of the cartilage-bone interface and calculating curvature to be

$$\text{Curvature} = \frac{1}{r} \quad (5.1)$$

where r was the radius of the fitted circle. The curvature of the cartilage-bone interface of the five test specimen that were used was on average 0.031 mm^{-1} ($\pm 0.006 \text{ mm}^{-1}$) ranging from 0.026 to 0.042 mm^{-1} (Table 5.1).

Cartilage in the hip joint is subjected to a combination of compression and shear, so these two types of loading were simulated *in vitro*. The Whole Articular Surface Indentation Machine (WASIM), a novel 4-degrees of freedom (DOF) testing machine, was used to hold, orient, and load the samples.¹ The WASIM apparatus is comprised of a stage which allows for three DOF (translation in the $\pm X$ and $\pm Y$ directions and pitch), a sample holder embedded into the stage

¹WASIM was designed by a collaboration between Professor Richie Gill and Tiab Ltd (Oxfordshire, UK).

Sample	Cartilage Thickness (mm)	Fitted Circle Radius (mm)	Curvature of Interface (mm^{-1})
1	2.0	33	0.030
2	4.0	33	0.030
3	1.8	39	0.026
4	3.0	24	0.042
5	3.0	35	0.029
Average	2.8	33	0.031
Std	0.9	5.5	0.006

Table 5.1: The cartilage thickness at the apex of Samples 1 to 5 are listed along with the radius of a circle fitted to the cartilage-bone interface of the samples and the curvature (Eqn 5.1) of the cartilage-bone interfaces.

with one DOF (yaw), and a stepper motor which provides yet another DOF (motion along the vertical Z-axis) (Figure 5.4). The testing machine was operated using a propriety controller and software developed by Tiab Ltd (Middleton Cheney, United Kingdom).

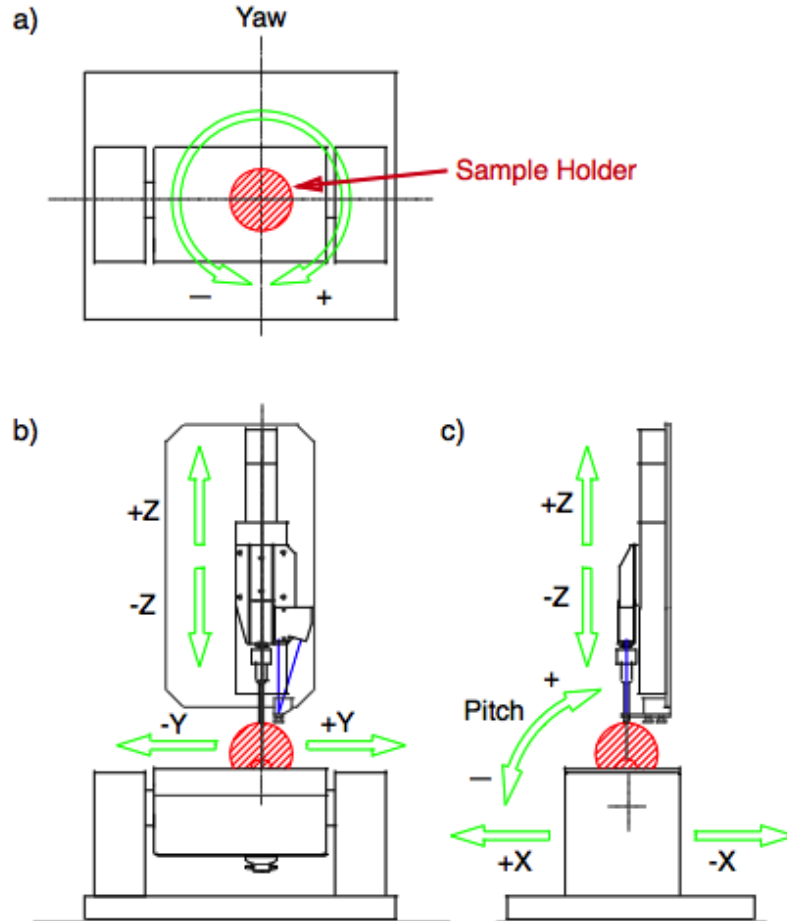


Figure 5.4: The WASIM Apparatus is shown from different views to depict its DOF [154]. a) The top view shows the yaw movement of the sample holder. b) The front view shows the possible movement on the Z and Y-axes. c) The lateral view shows the pitch movement of the stage and the possible movement on the X-axes.

To enable the capturing of digital images of the speckle pattern on the samples, a camera was attached to the stage holding the sample, and a light fixture was adjusted before capturing each image to ensure good lighting of the sample and its speckle pattern (Figure 5.5). An indenter was specially designed and made for the loading of cartilage samples during validation experiments. The indenter (Figure B.1) comprised of an aluminum body which attached to the Z-axis of the stepper motor of the WASIM apparatus and a curved stainless steel end piece which formed the indenting surface. The stainless steel indenting end was a segment of a stainless steel cylinder.

Experimental Setup for Validation Tests

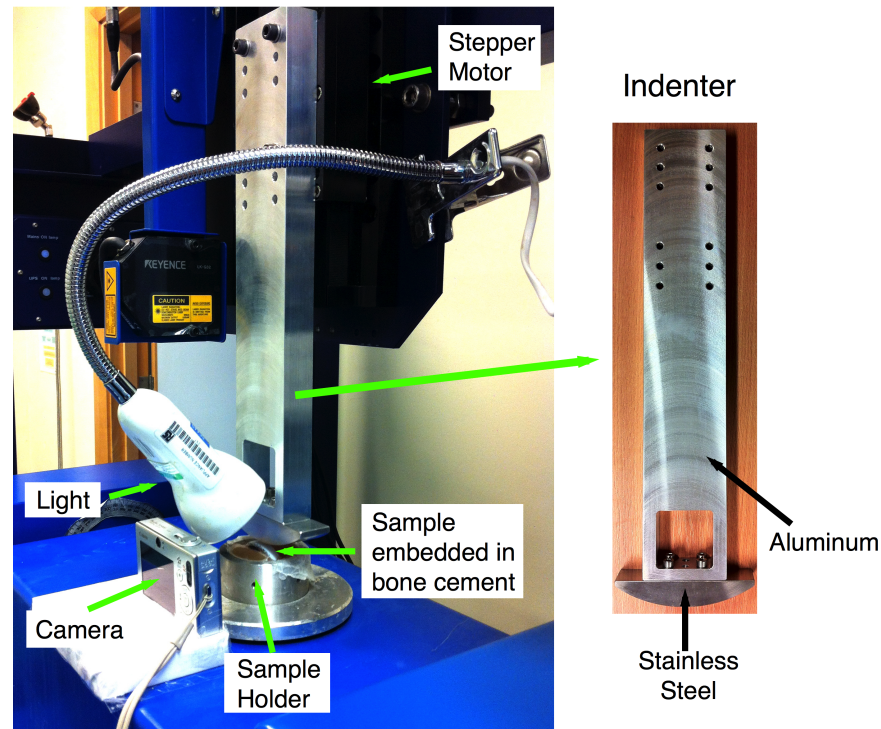


Figure 5.5: The experimental setup for *in vitro* FE validation experiments are shown (left) including the sample holder to hold samples embedded in bone cement, a camera to capture digital images of the sample's speckle pattern, a light fixture to provide good lighting of the sample, and an indenter (right) with an aluminum body and curved steel end piece.

Prior to conducting the experimental tests, the samples were placed in the sample holder and oriented so the surface with the speckle pattern was in the capturing view of the camera. They were also aligned with the front edge of the indenter to prevent the indenter from creating any shadows on the sample. Also, the top of the cartilage surface, when viewed laterally, was oriented parallel to the indenting surface of the indenter. Orienting the samples accordingly entailed moving the yaw so that the speckle patterned surface faced the camera, moving the pitch so the top surface of the cartilage was aligned with the bottom of the indenter, and

moving the stage along the X-axis so the speckled plane of the sample aligned with the front edge of the indenter.

During the compression validation experiments, the apex of the cartilage sample was aligned with the apex of the indenter, and the indenter was lowered so its apex came into contact with the surface of the superficial layer of the cartilage without applying any load (Figure 5.6). The indenter was lowered to apply compression to the sample in displacement increments equivalent to 5% of the thickness of the cartilage at its apex until the indenter was lowered to 50% of the thickness of the cartilage. A digital image of the cartilage sample and its speckle pattern was captured prior to loading and at each increment of loading. Since planar slices of cartilage-bone samples were cut so the curvature of the interface and thickness of the cartilage layer was as uniform as possible through the sample's thickness, compression applied by the cylindric indenter simulated Hertzian contact between two cylinders where one cylinder was layered with cartilage.

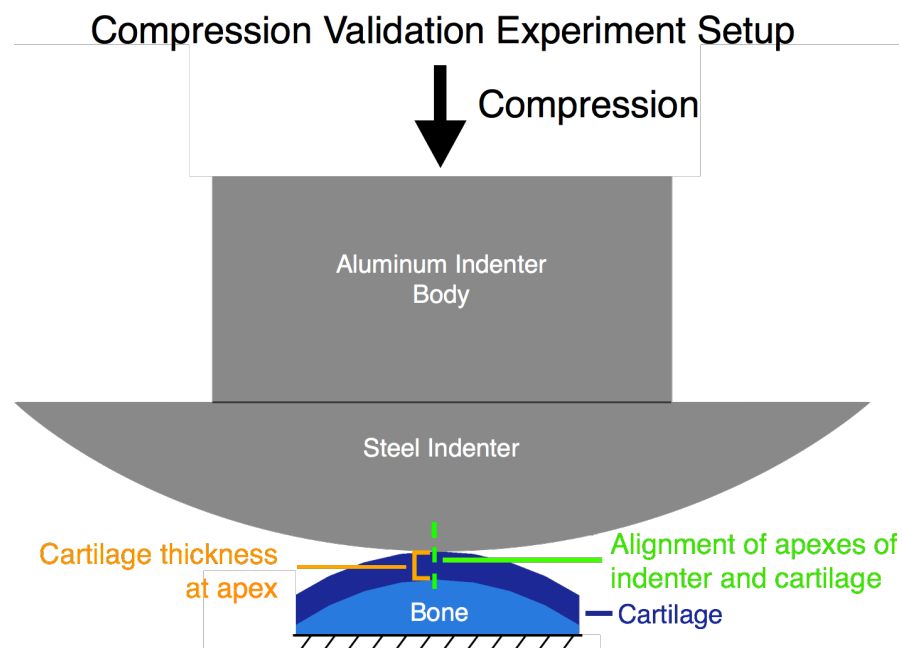


Figure 5.6: The experiment setup used for compression tests for FE validation is depicted. The apexes of the indenter and cartilage were aligned. The indenter was lowered to 50% of the thickness of the cartilage in increments equivalent to 5% of the thickness of the cartilage.

During the shear validation experiments, the apexes of the cartilage and indenter were also aligned, and the indenter was lowered so its apex came into contact with the surface of the superficial layer of the cartilage without applying any load. The platform holding the sample was then displaced 20 mm in the -Y direction, and the indenter was lowered to either 25% or

40% of the thickness of the cartilage at its apex (Figure 5.7). This amount of lowering was referred to as 25% or 40% overlap, respectively. The platform holding the sample was translated in increments of 5 mm in the +Y direction so that the indenter passed across the surface of the cartilage, causing compression and shear on the sample. A digital image of the cartilage sample and its speckle pattern was captured prior to contact between the sample and the indenter and at every increment, thereafter, until the sample fully passed across the indenter.

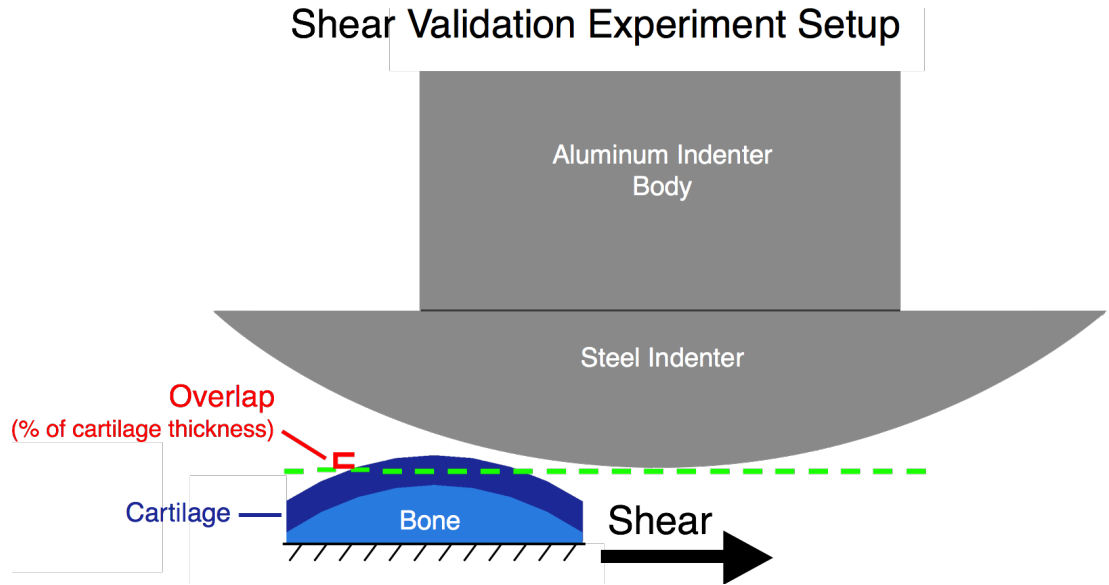


Figure 5.7: The experiment setup used for shear tests for FE validation is depicted. The apexes of the indenter and cartilage were offset from each other, the indenter was lowered past the apex of the cartilage sample by a certain amount of overlap (25% or 40% of the cartilage thickness), and the cartilage-bone sample was translated so it passed under the indenting surface.

Three trials of compression tests, shear tests at 25% overlap, and shear tests at 40% overlap were conducted for each sample. Each sample was allowed to rehydrate for 40 to 60 minutes after each trial before the next trial.

5.2.4 Using Digital Image Correlation (DIC) for Strain Calculation

Digital Image Correlation (DIC) software called MatchID 2D (Katholieke Hogeschool Sint-Lieven (KOHOSL), Gent, Belgium) was used to analyse the digital images taken during experiment tests [155]. This DIC software allowed the determination of displacement and strain fields with sub-pixel accuracy on specimen not subject to out-of-plane deformation with a single camera aligned perpendicular to the specimen surface.

Each trial produced up to eleven images for the compression tests and up to eight images for the

shear tests. The series of images for each experimental trial were analysed using the following software settings: 1) a subset between 20 to 50, 2) a step of 5, 3) a strain window within 10 to 20, 4) bicubic interpolation, and 5) the zero-normalized sum of squared differences correlation coefficient (ZNSSD). The subset and strain window were adjusted for each series of results to obtain results that were able to detect shear strain in the cartilage. Bicubic interpolation was used because it was determined to be the most accurate compared to other interpolation methods [156]. ZNSSD was used because it is very robust in variations due to offset and scaling in lighting conditions. The cartilage and bone of the samples were selected as the region of interest (ROI) in each image. This ROI in each image was compared with the ROI of the initial image which represented the unloaded sample. To obtain plots of the shear strain distribution for each series, each image in a series was extracted with the same colourbar which was based on thresholds of the maximum and minimum range of shear strain in the cartilage across the entire series. To obtain quantitative results, the colourbar thresholds were adjusted for each image in a series to determine the maximum and minimum shear strain values in the cartilage for that image.

5.2.5 Modelling *In Vitro* Validation Experiments using FE Models

FE models were created to mimic the experimental setup, which importantly used the same contact and material representations as those used in the impingement FE model. Two sets of implicit FE models for all simulations of the validation experiments were created – one with 2 mm-thick cartilage, and the other with 4 mm-thick cartilage. The geometry of a cartilage-bone sample was created by modelling either a 2 mm- or 4 mm-thick cartilage layer tied to bone at an interface with a curvature of 0.031 mm^{-1} . The geometry of the indenter was created to be identical to the actual profile of the indenter. The curvature of the indenter head was 0.011 mm^{-1} . The steel head and aluminum body were created as a single part but with a partition to enable the defining of different material properties for the two sections.

The cartilage ($E=3.4 \text{ MPa}$, $\nu=0.45$ [157]), bone ($E = 17 \text{ GPa}$, $\nu = 0.3$) [68], steel end of the indenter ($E = 180 \text{ GPa}$, $\nu = 0.3$ [158]), and aluminum body of the indenter ($E = 69 \text{ GPa}$, $\nu = 0.33$ [158]) were defined to be isotropic linear elastic in the same way that materials were defined in the impingement FE model. However, the cartilage stiffness and Poisson’s ratio for porcine cartilage were used instead of those for human cartilage.

Four-node biquadratic, plain strain, solid continuum, reduced integration (CPE4R) elements were used to model all parts. The use of a plain strain element formulation results in the calculation of stress and strain, assuming the 2D geometries are slices of a thicker slice with consistent shape through its thickness. This resulted in Hertzian contact between two cylinders where one cylinder was layered (with cartilage), simulating the experimental setup and mimicking the impingement FE model's element formulation. An element size of 0.1 mm was used for the cartilage and bone parts and the indenting edge of the indenter. An element size of 0.5 mm was used for non-indenting surfaces of the indenter, and an element size of 1.0 mm was used for the aluminum body of the indenter.

Similar boundary conditions were assigned, and similar constraints and interactions were defined as those in the impingement FE model. The base of the bone surface was constrained in all DOF. Tie constraints were applied to the surfaces of the cartilage and bone at the cartilage-bone interface. A surface-to-surface contact interaction modelling contact friction with a coefficient of friction of 0.006 (penalty contact) was applied between the articulating surfaces of the acetabular articular cartilage and the indenter. The indenting surface of the indenter was defined to be the master surface, while the superficial surface of the cartilage was defined to be the slave surface.

To simulate the compression validation experiment, the edge of the aluminum body that was furthest from the steel head was constrained in all DOF except for the direction of compression (Figure 5.8). The incremental lowering of the indenter was applied through displacement boundary conditions. To simulate the shear validation experiment, the edge of the aluminum body that was furthest from the steel head was constrained in all DOF except the direction of translation which allowed the indenter to move across the sample (Figure 5.9). This setup differed from the experimental test where the platform holding the sample was translated instead of the indenter. However, for the simulation, it did not matter which object was stationary and which was moving, the same results would be obtained had the situation been reversed.

The shear strain results were extracted from the Abaqus output database files and analysed to determine the maximum and minimum shear strain values. The top five maximum and minimum values were averaged and used as the calculated values for the FE model.

Validation FE Model for Compression Experiments

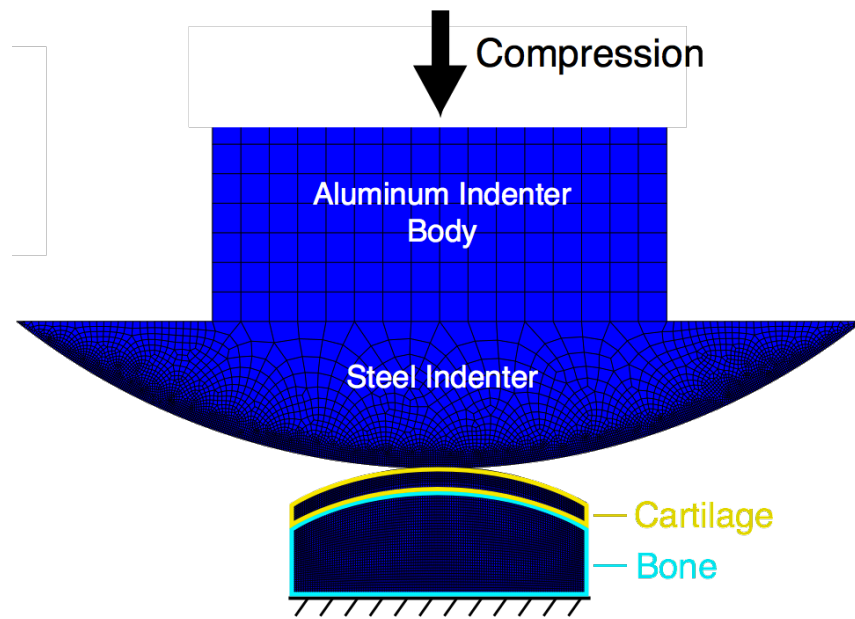


Figure 5.8: The FE model setup used to simulate *in vitro* compression validation experiments is depicted. The apices of the indenter and cartilage were aligned. The indenter was lowered to 50% of the thickness of the cartilage in increments equivalent to 5% of the thickness of the cartilage.

Validation FE Model for Shear Experiments

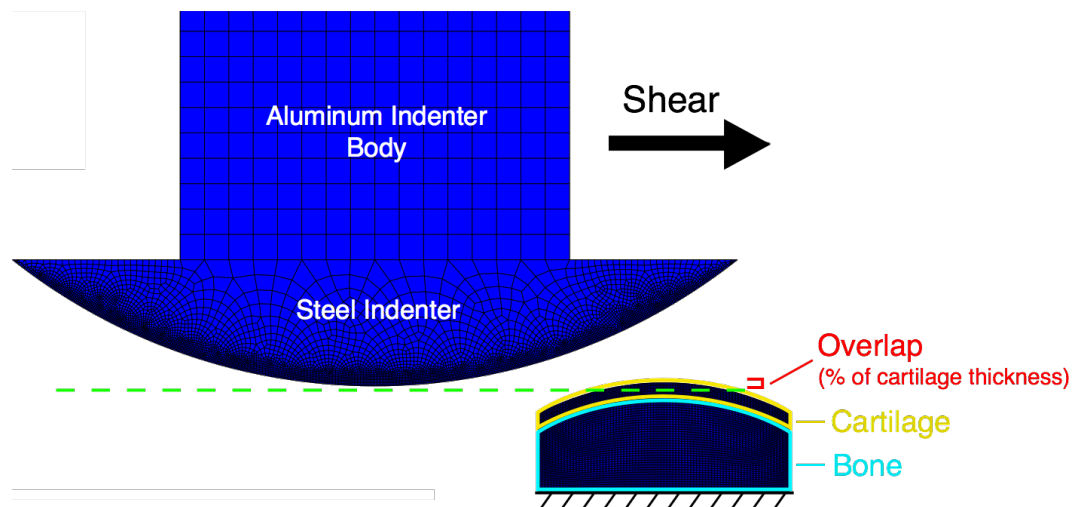


Figure 5.9: The FE model setup used to simulate *in vitro* shear validation experiments is depicted. The apices of the indenter and cartilage were offset from each other, the indenter was lowered past the apex of the cartilage sample by a certain amount of overlap (25% or 40% of the cartilage thickness), and the indenter was translated so it passed over the cartilage sample's surface.

5.2.6 *In Vitro* Damage Experiments

Damage experiments were conducted on the five samples that were used for the FE validation experiments in Section 5.2.3 and on four additional samples. The four additional samples,

In Vitro Damage Experiment Samples

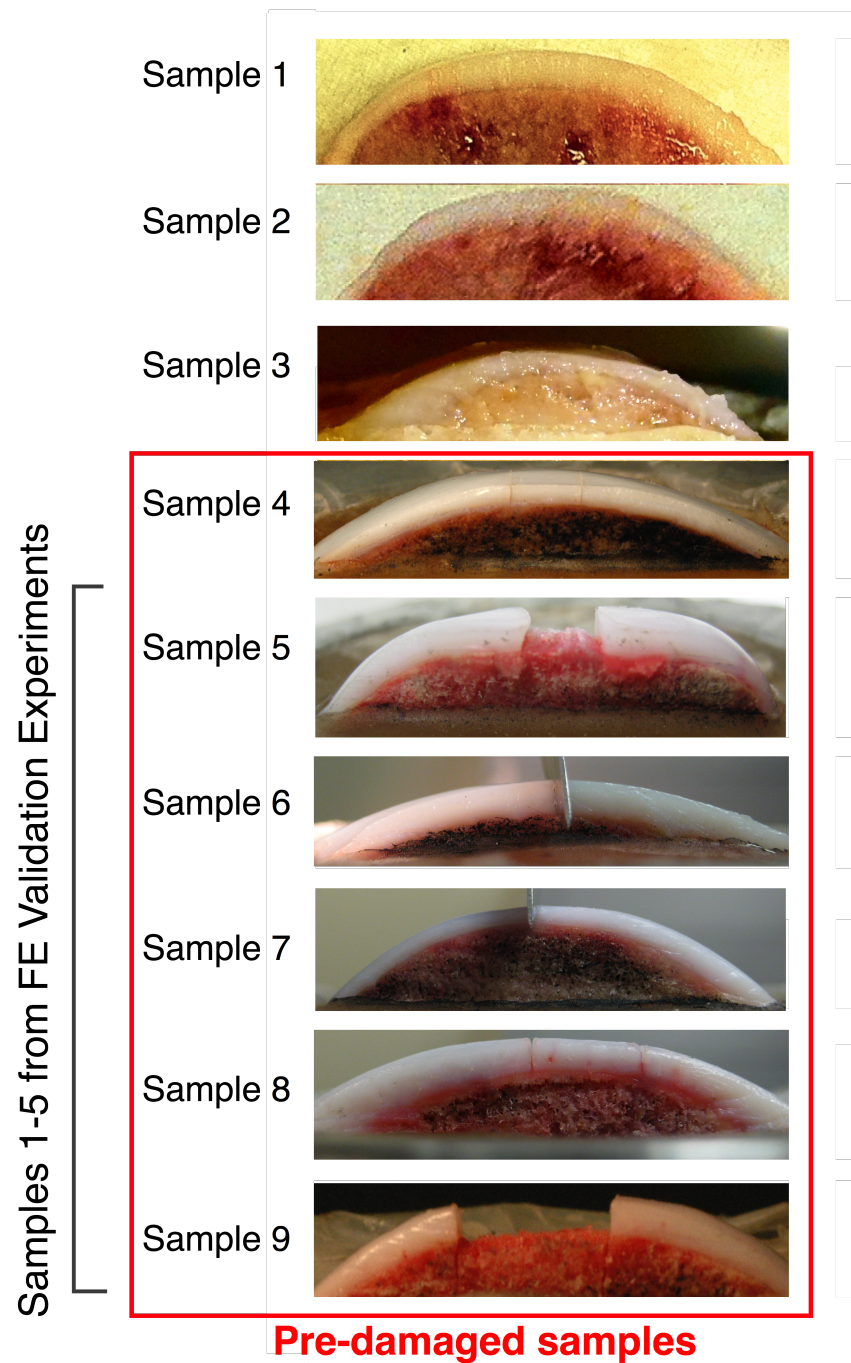


Figure 5.10: The samples used for *in vitro* damage experiments are shown. Samples 5 to 9 were also used in the *in vitro* FE validation experiments. Samples 4 to 9 were pre-damaged to create an edge so that they mimicked the anatomy of acetabular articular cartilage at the acetabular rim. Sample 4 and 8 were pre-damaged by vertically cutting through the thickness of the cartilage in two places near the apex of the sample. Sample 5 was pre-damaged by removing the mid-section of the cartilage along with the subchondral bone. Samples 6 and 7 were pre-damaged by vertically cutting through the thickness of the cartilage at the apex of the sample. Sample 9 was pre-damaged by removing the mid-section of the entire cartilage layer.

hereafter, are referred to as Damage Samples 1 to 4, and the samples also used for the FE validation experiments, hereafter, are referred to as Damage Samples 5 to 9. All samples were

prepared according to the sample preparation methods for the validation experiments in Section 5.2.1. The cartilage in Damage Samples 4 to 9 were pre-damaged so that they represented the condition of the acetabular articular cartilage at the acetabular rim where the articular cartilage forms an edge that may be grazed by a cam deformity (Figure 5.10). Sample 4 and 8 were pre-damaged by vertically cutting through the thickness of the cartilage in two places near the apex of the sample. Sample 5 was pre-damaged by removing the mid-section of the cartilage along with the subchondral bone. Samples 6 and 7 were pre-damaged by vertically cutting through the thickness of the cartilage at the apex of the sample. Sample 9 was pre-damaged by removing the mid-section of the entire cartilage layer.

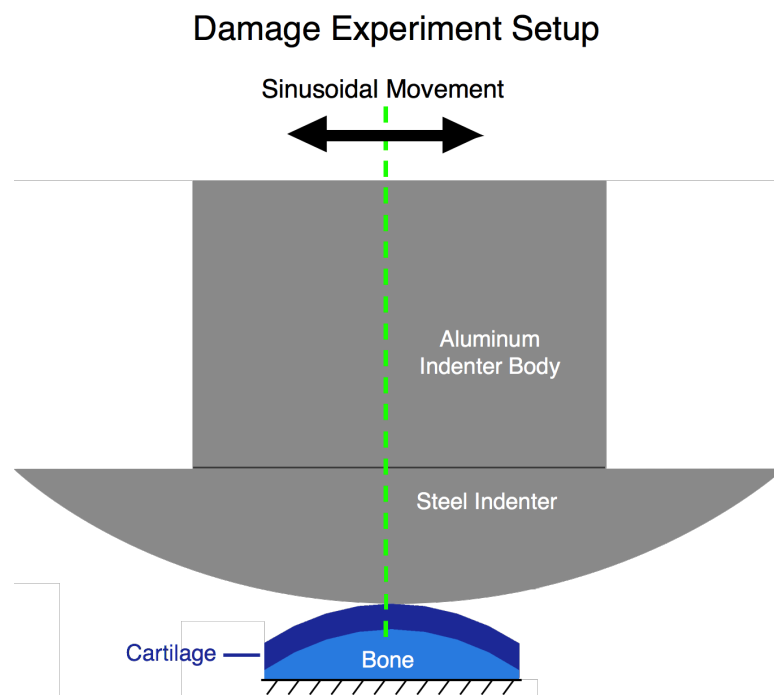


Figure 5.11: The damage experiment setup is depicted. The apices of the indenter and cartilage were aligned. The indenter was moved sinusoidally across the cartilage and was slowly lowered in 0.1 mm increments when the water had exuded out of the portion of cartilage that the indenter came into shear contact with.

During the damage experiments, the apices of the cartilage samples were aligned with the apex of the indenter, and the indenter was lowered so its apex came into contact with the surface of the superficial layer of the cartilage without applying any load. The indenter was moved with a sinusoidal motion along the Y-axis with an amplitude of ± 20 mm at a frequency of 0.8 Hz (Figure 5.11). The indenter was also lowered along the Z-axis in 0.1 mm increments. The timing of the lowering of the indenter depended on whether it was observed that the water had exuded out of the portion of cartilage that the indenter came into shear contact with. The

progression of the lowering of the indenter varied between samples, but they all started out similarly where the indenter was lowered 0.1 mm after every 20 cycles of sinusoidal motion. Over time, the indenter was lowered 0.1 mm only after 50 cycles, and then 100 cycles, so on and so forth, up to about 1000 cycles. The sinusoidal movement of the indenter was periodically paused so that the samples could be inspected for any damage.

5.3 Mesh Convergence Study for Validation FE Models

A mesh convergence study was performed for the compression and shear validation FE models, and for the 2 mm- and 4 mm-thick cartilage models to ensure that the FE results simulating the experimental tests were meaningful and accurate. The element sizes for the cartilage and the bone components were varied as follows: 1.0, 0.8, 0.6, 0.4, 0.3, 0.25, 0.2, 0.15, and 0.1 mm (Figure 5.12, 5.13). The shear strain results were extracted from the Abaqus output database files and analysed to determine the maximum and minimum shear strain values. The average of the top five maximum and minimum values were used as the calculated values to test for convergence in the cartilage and bone parts.

The convergence of shear strain of the compression model was investigated at the time point when maximum compression was applied; this occurred at the FE solution increment of 50 (Figure 5.12). For the shear models, the translation of the indenter across the sample in one direction and the return direction were both analysed in FE, however, the convergence was investigated only for the initial direction and when shear strain reached its apex. This occurred at the FE solution increment of 10 for the shear model at 25% overlap, and at the FE solution increment of 72 for the shear model at 40% overlap (Figure 5.13). The model was considered to have fully converged if for three consecutive models, the difference in the results for shear strain was less than 5% [159]. This meant that convergence was met for a particular result if:

$$\frac{(\epsilon_n - \epsilon_{n-1})}{\epsilon_n * 100} < 5\% \quad (5.2)$$

where ϵ_n was the shear strain of the model with a more refined mesh and ϵ_{n-1} was the shear strain of the model with a courser mesh, and three consecutive models met these criteria.

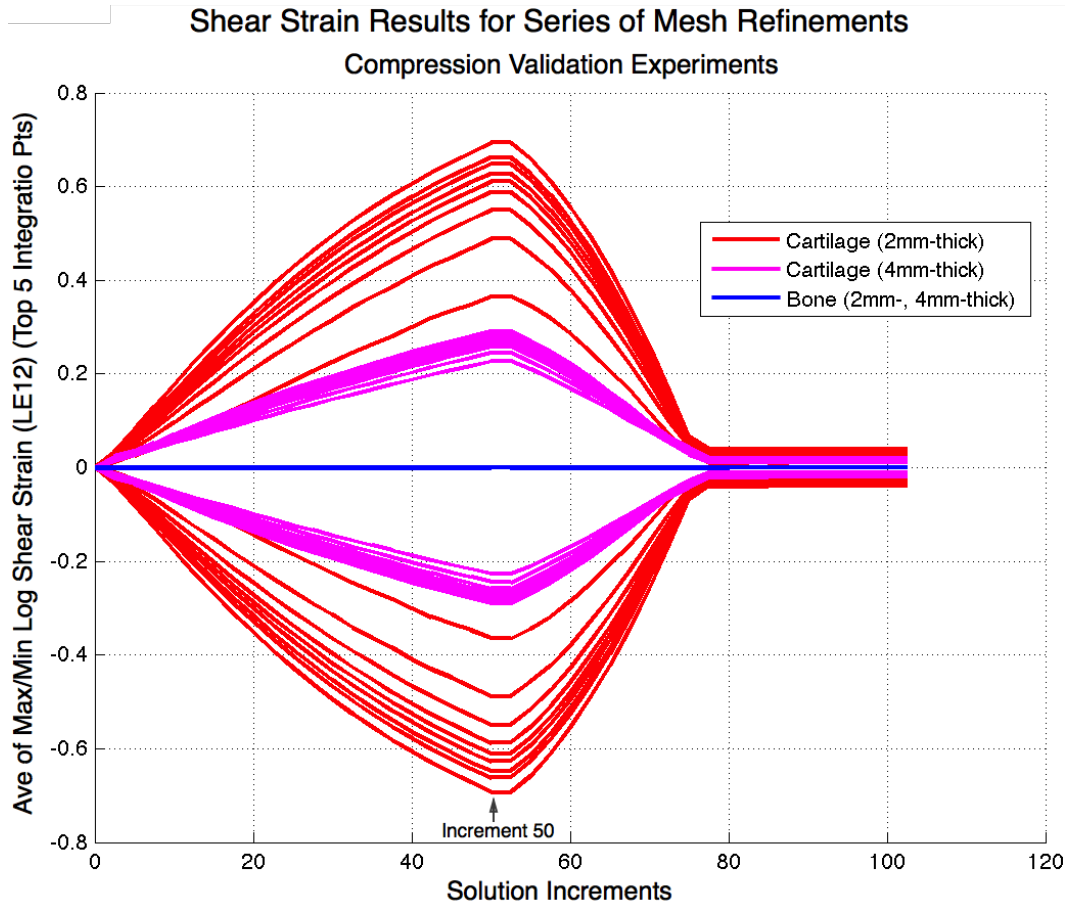


Figure 5.12: The shear strain results for validation FE models simulating compression experiments using a series of mesh refinements are presented. Cartilage was meshed with a series of element sizes including 1.0, 0.8, 0.6, 0.4, 0.3, 0.25, 0.2, 0.15, and 0.1 mm. Each shear strain result has a positive and negative component. A mesh convergence study was chosen to be analysed when maximum compression was applied, which occurred at the FE solution increment of 50.

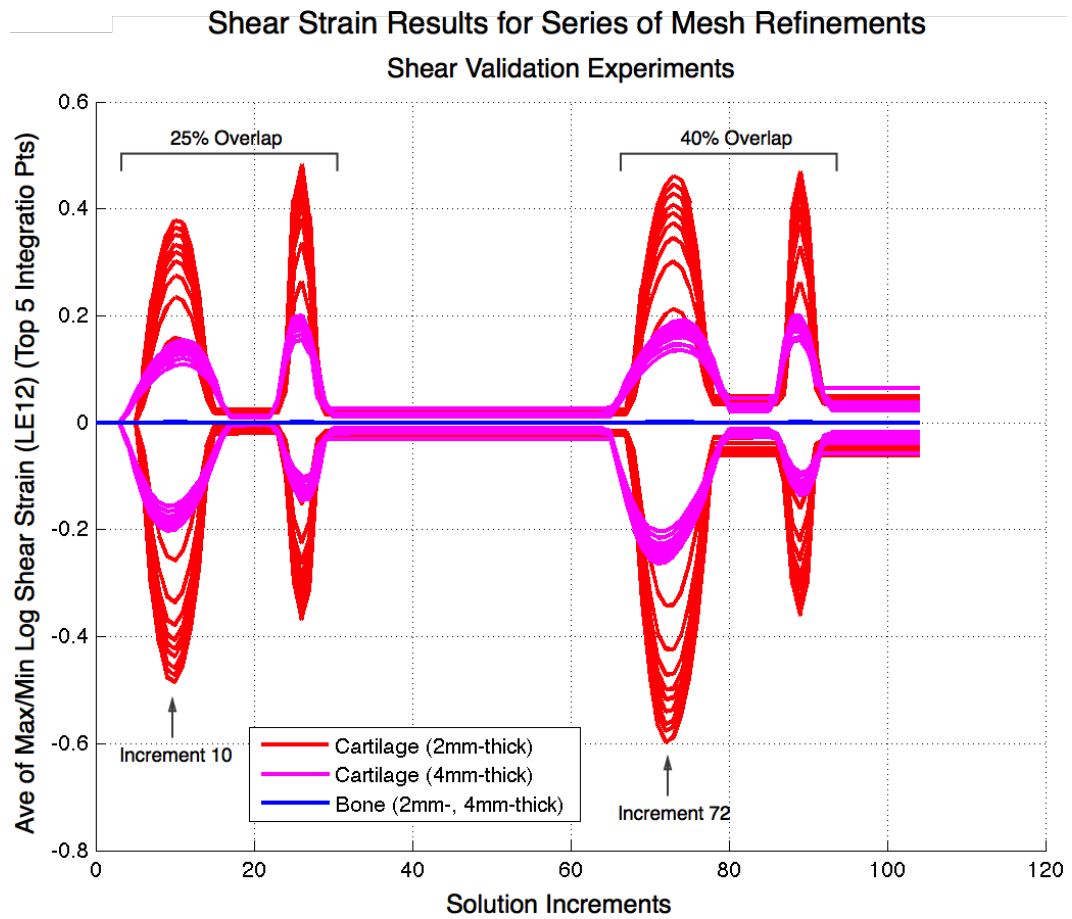


Figure 5.13: The shear strain results for validation FE models simulating shear experiments using a series of mesh refinements are presented. Cartilage was meshed with a series of element sizes including 1.0, 0.8, 0.6, 0.4, 0.3, 0.25, 0.2, 0.15, and 0.1 mm. Each shear strain result has a positive and negative component. A mesh convergence study was chosen to be analysed when the maximum shear strain occurred which occurred at the FE solution increments of 10 and 72 for shear models at 25% and 40% overlap, respectively.

An arrow was used to indicate that the shear strain results for a model, and the models with a more refined mesh, were within 5% of the shear strain results of the consecutive model with a more refined mesh (Figures 5.14, 5.15). A circle indicated the model with the element size of 0.1 mm which was chosen to be used for the final models (Figures 5.14, 5.15). The models fully converged for element sizes that were larger than 0.1 mm, so they could have been chosen, however, the computational time required to solve the model with an element size of 0.1 mm was only about 1 to 2 hours. This timeframe was considered to be easily manageable.

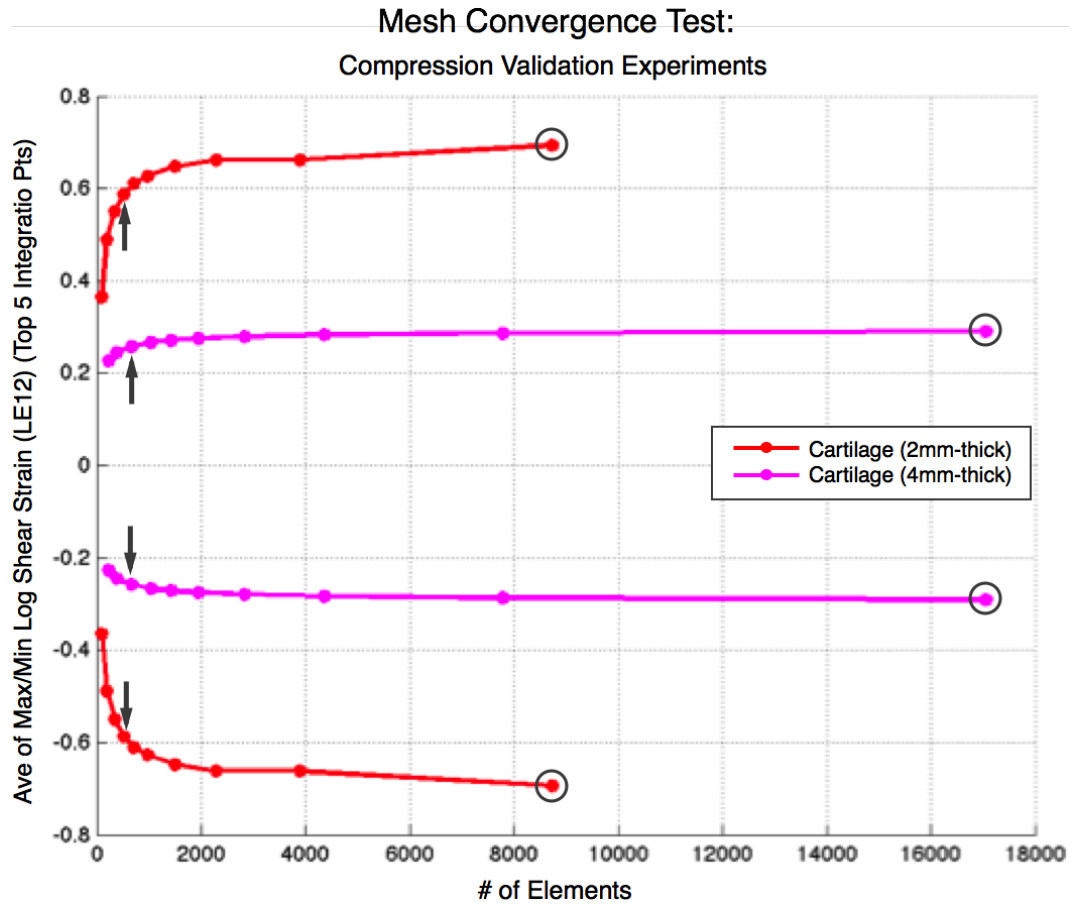


Figure 5.14: The mesh convergence test results for the compression validation FE model is presented. An arrow indicates convergence was met for that element size and for all more refined mesh sizes. A circle indicates the model with the element size of 0.1 mm which was chosen to be used to run final results.

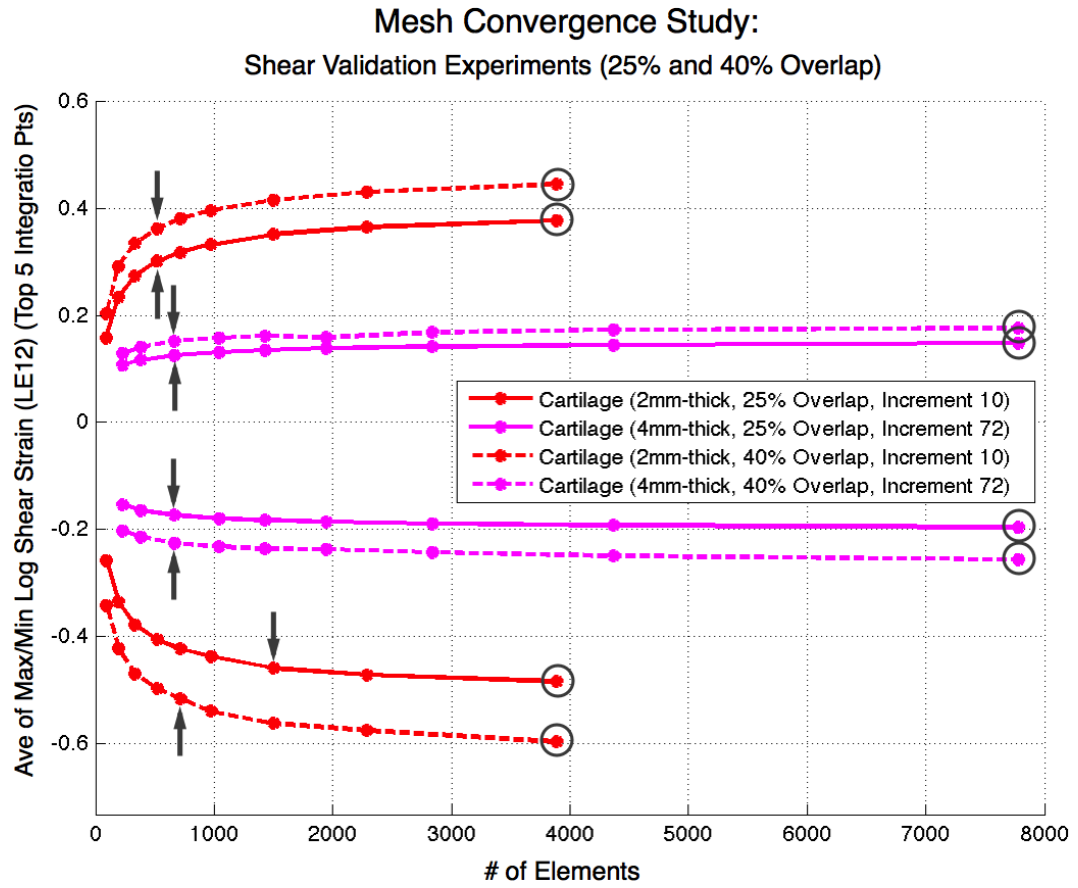


Figure 5.15: The mesh convergence test results for the shear validation FE model is presented. An arrow indicates convergence was met for that element size and for all more refined mesh sizes. A circle indicates the model with the element size of 0.1 mm which was chosen to be used to run final results.

The maximum and minimum shear strain results for 2 mm-thick cartilage were greater in magnitude than those for 4 mm-thick cartilage. It was expected that different cartilage thicknesses would lead to different results. The thickness of a layer covering a curved surface has been shown to affect the resulting maximum contact pressure between two curved surfaces, and thus the stresses and strains experienced [160, 161]. The type of curvature involved (spherical or cylindric), radius of the curvatures, magnitude of contact force applied, and the elastic modulus and the Poisson's ratio of the contact surfaces also affect the resulting maximum contact pressure between two curved surfaces, however, these parameters were the same in all models with different cartilage thicknesses, so they did not contribute to the shear strain differences.

All models converged for a mesh size of 0.2 mm or smaller. Some models even converged for an element size of 0.3 mm. Choosing an element size of 0.1 mm for the validation FE models was, therefore, a valid choice that would ensure accurate results.

5.4 Results

5.4.1 *In Vitro* Validation Experimental Results

Shear strain results are presented through series of plots for Samples 1 to 5 for the compression validation experiments (Figure 5.16) and shear validation experiments for 25% (Figure 5.17) and 40% overlap (Figure 5.18). Shear strains in the counterclockwise (CCW) direction were represented as positive shear strains (red), and shear strains in the clockwise (CW) direction were represented as negative shear strains (blue). The full distribution of shear strain across the samples was not able to be calculated in all tests since the samples often exuded water during loading which disturbed the integrity of the speckle pattern. This was especially apparent in Sample 5 for all experiments and during all validation shear experiments for 40% overlap (Figure 5.18). It also caused the results for Sample 1 to be unusable, so they were left out. However, the overall distribution of shear strain was still able to be determined in the presented results (Figures 5.16, 5.17, 5.18).

For all models, maximum and minimum shear strains developed near the cartilage-bone interface in the cartilage at the location closest to the location of indentation. In some cases, shear strain was also calculated at the superficial layer of cartilage, but they were much less dominant than the shear strains that developed near the cartilage-bone interface. This occurred for Samples 1 and 4 in the compression experiment, for Samples 2 and 4 in the shear (25% overlap) experiment, and for Samples 2 to 4 in the shear (40% overlap) experiment. In some cases, shear strain was calculated in only the superficial layer of cartilage prior to any development of shear strain in the deep layer when the indenter first made contact with the tissue and did not apply much compression or shear. This occurred for Samples 1 and 2 in the compression experiments and for Samples 4 and 5 in the shear (40% overlap) experiments.

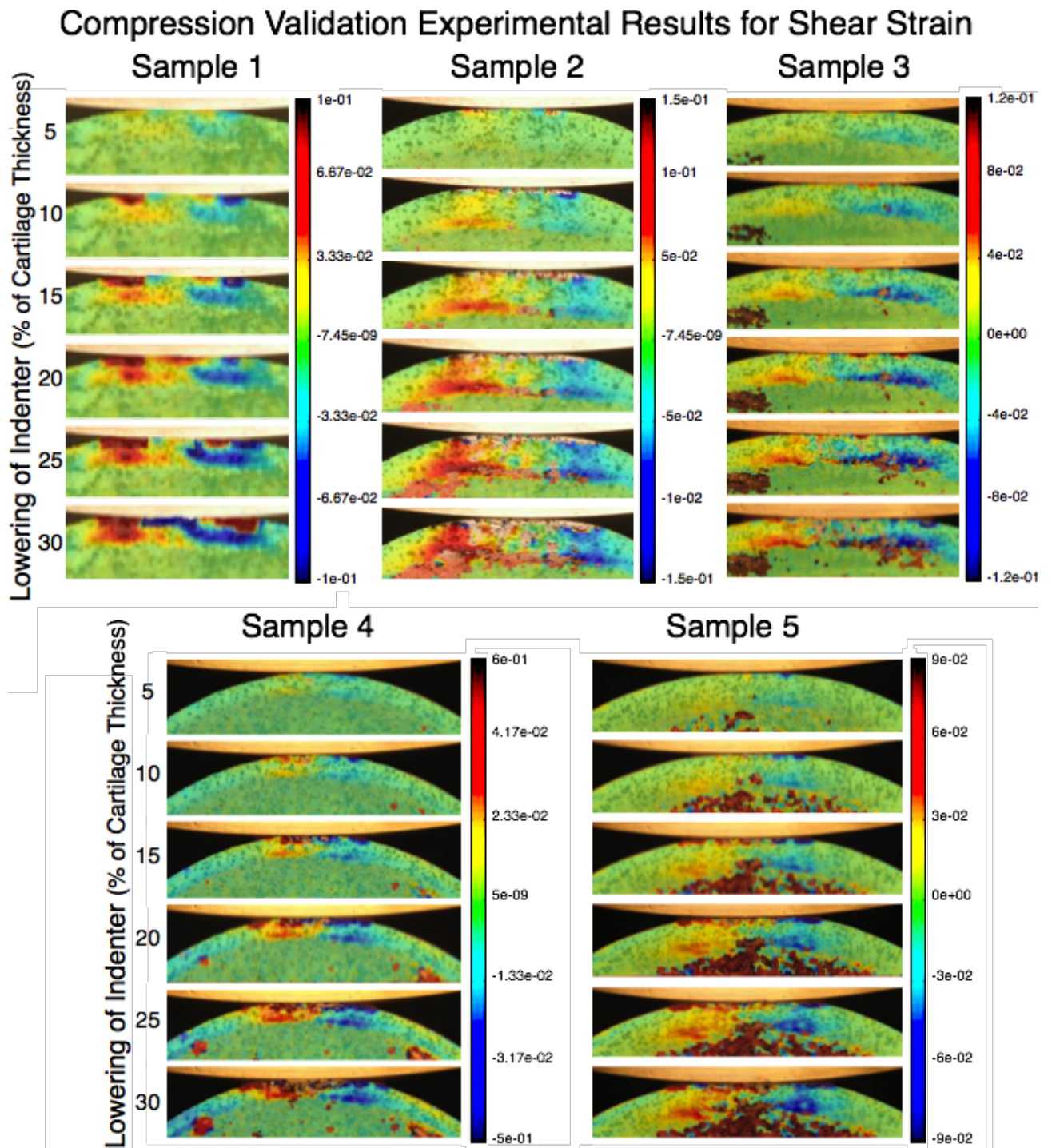


Figure 5.16: The shear strain results for the *in vitro* compression validation experiments are presented for Samples 1 to 5. Shear strains developed near the cartilage-bone interface in the cartilage for all models and at the superficial layer of cartilage for Samples 1 and 4.

Shear Validation Experimental Results (25% Overlap) for Shear Strain

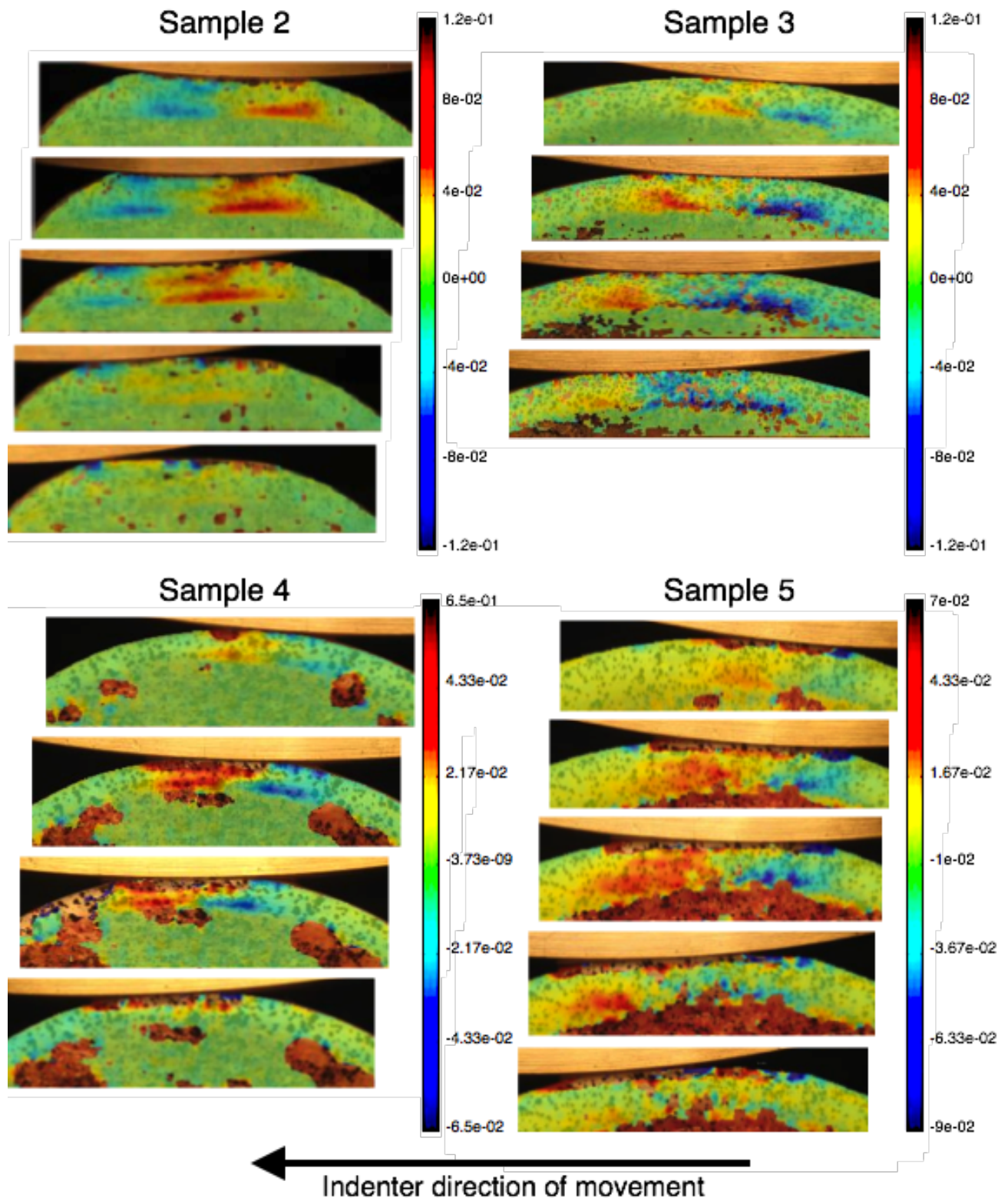


Figure 5.17: The shear strain results for the *in vitro* shear (25% overlap) validation experiments are presented for Samples 2 to 5. Shear strains developed near the cartilage-bone interface in the cartilage for all models and at the superficial layer of cartilage for Samples 2 and 4.

Shear Validation Experimental Results (40% Overlap) for Shear Strain

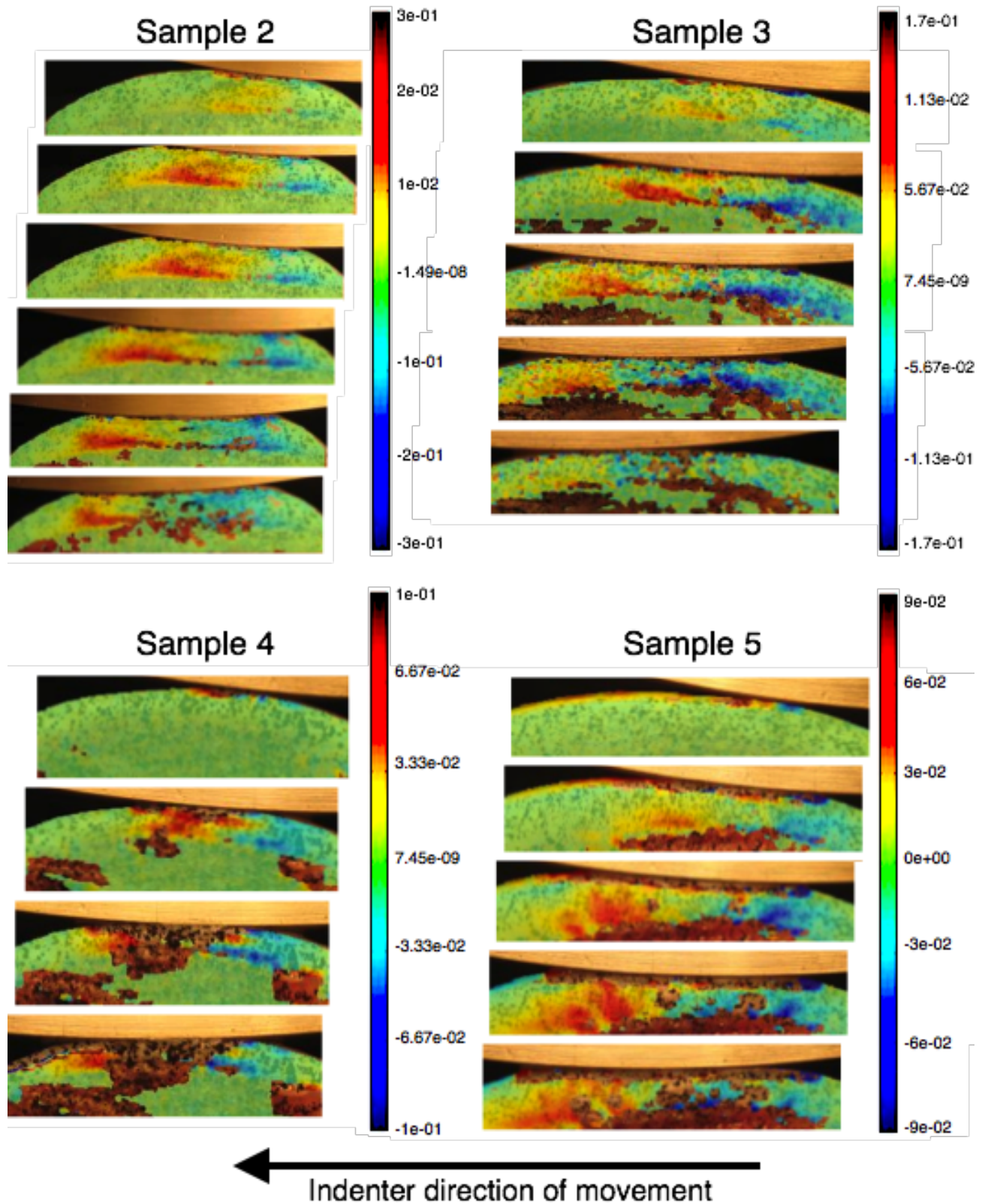


Figure 5.18: The shear strain results for the *in vitro* shear (40% overlap) validation experiments are presented for Samples 2 to 5. Shear strains developed near the cartilage-bone interface and at the superficial layer of the cartilage for all models, but were more dominant near the cartilage-bone interface.

5.4.2 FE Results

Logarithmic shear strain results are presented as a series of plots for 2 mm- and 4 mm-thick cartilage for the compression validation FE model (Figure 5.19) and shear validation FE models for 25% (Figure 5.20) and 40% overlap (Figure 5.21). Shear strains in the CCW direction were represented as negative shear strains (blue), and shear strains in the CW direction were represented as positive shear strains (red). This convention was opposite from the convention used to represent shear strains in the validation experiment where shear strains in the CCW direction were positive (red) and shear strains in the CW direction were negative (blue).

For all models, positive and negative shear strains developed near the cartilage-bone interface in the cartilage at the location closest to the apex of indenter. Shear strain, therefore, resulted from compression of the cartilage, and during shear loading, the shear strain translated across the cartilage-bone interface as the indenter translated across the cartilage sample. For the models with 2 mm-thick cartilage, the shear strain concentrations tended to spread wider along the cartilage-bone interface compared to its spread through the thickness of the cartilage. In contrast, for the models with 4 mm-thick cartilage, the shear strain concentrations would sometimes spread more through the thickness of the cartilage compared to its spread along the cartilage-bone interface.

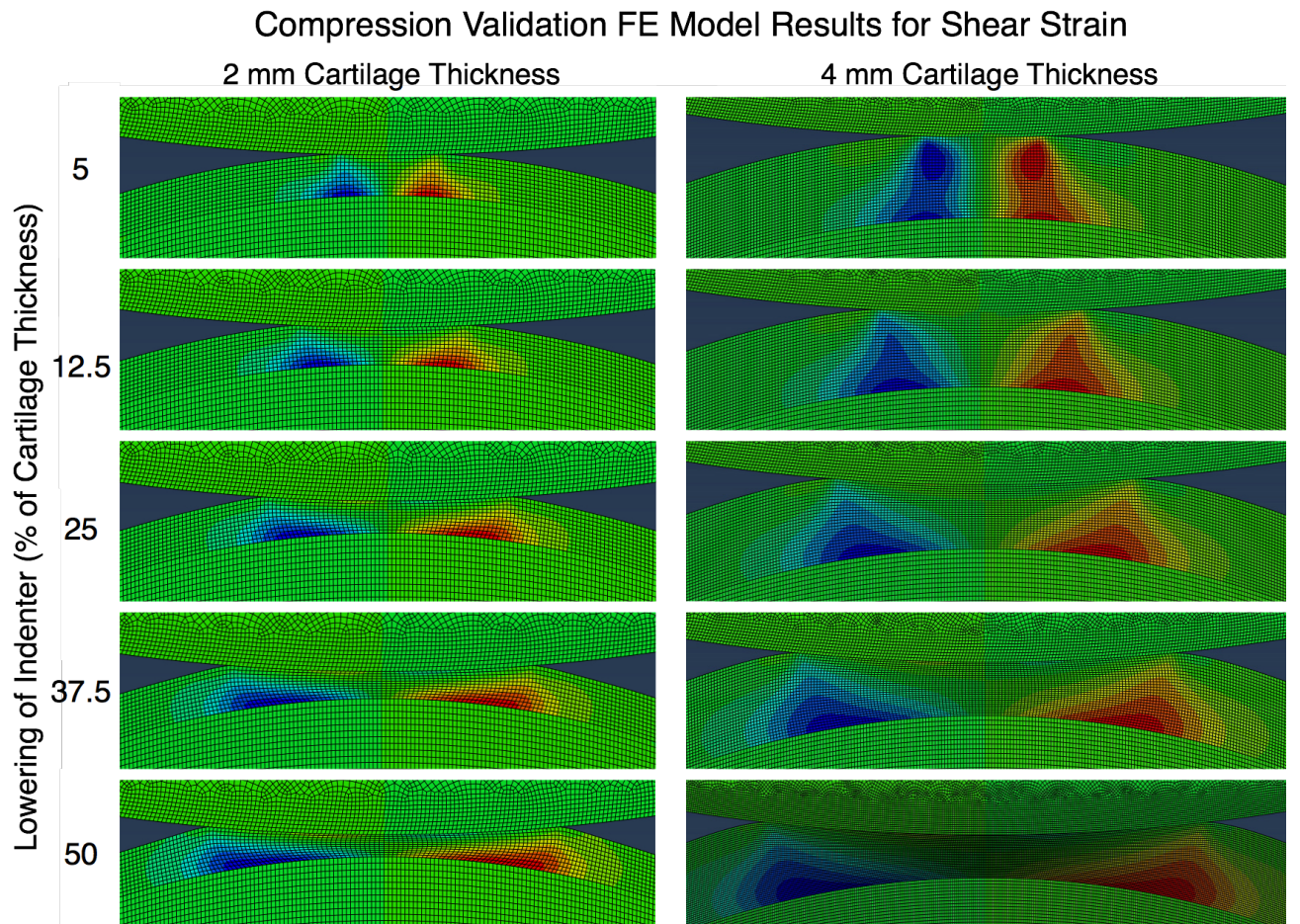


Figure 5.19: The shear strain results for the validation FE models simulating compression validation experiments are presented for 2 mm-thick (left) and 4 mm-thick cartilage (right). For 2 mm-thick cartilage, the maximum (red) and minimum (blue) logarithmic shear strains were 0.70 and -0.70, respectively. For 4 mm-thick cartilage, the maximum (red) and minimum (blue) logarithmic shear strains were 0.29 and -0.29, respectively. For all models, shear strains developed near the cartilage-bone interface in the cartilage.

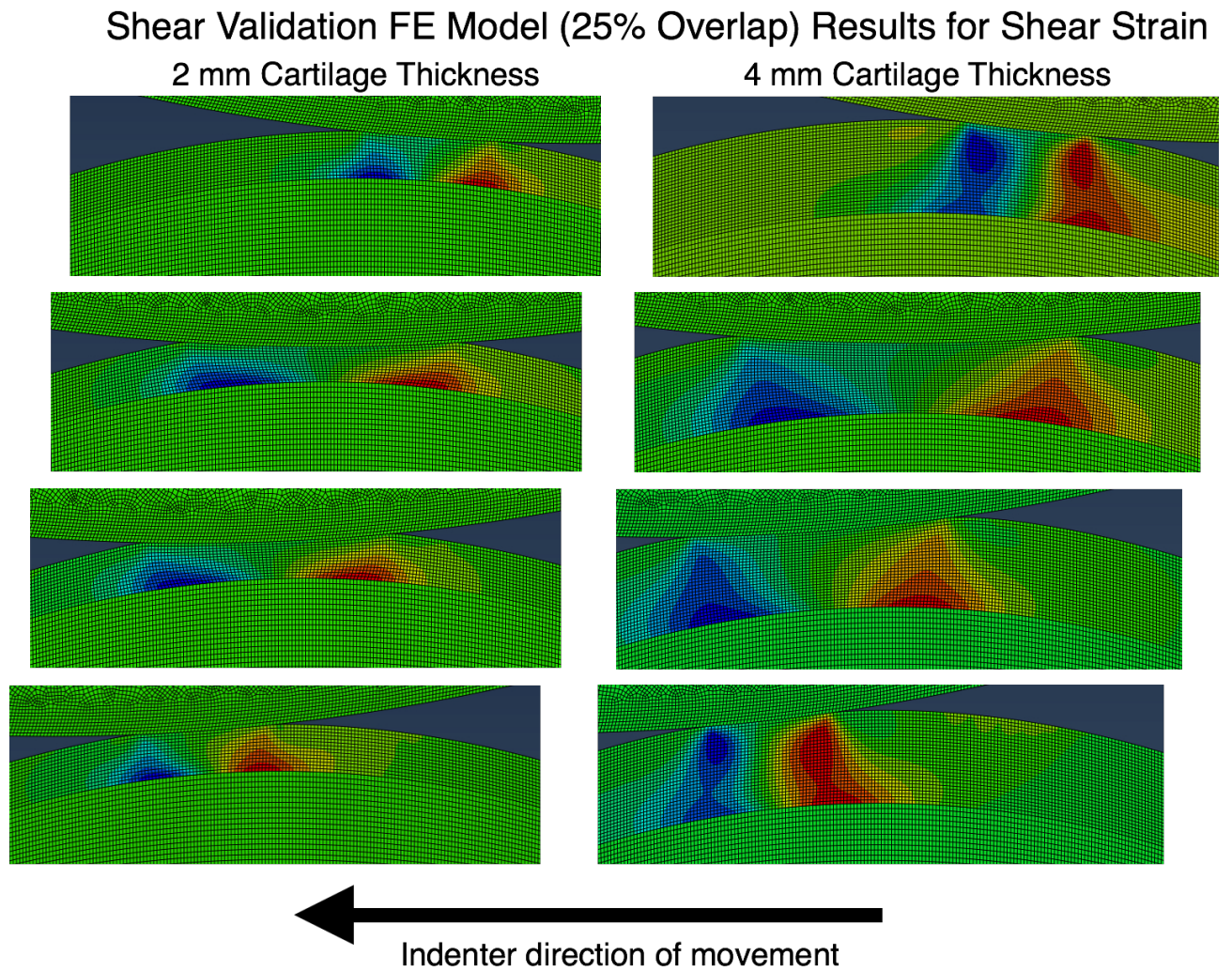


Figure 5.20: The shear strain results for the validation FE models simulating shear validation experiments are presented for 25% overlap and 2 mm-thick (left) and 4 mm-thick cartilage (right). For 2 mm-thick cartilage, the maximum (red) and minimum (blue) logarithmic shear strains were 0.38 and -0.49, respectively. For 4 mm-thick cartilage, the maximum (red) and minimum (blue) logarithmic shear strains were 0.16 and -0.21, respectively. For all models, shear strains developed near the cartilage-bone interface in the cartilage and translated along the interface as the indenter translated across the cartilage surface.

Shear Validation FE Model (40% Overlap) Results for Shear Strain

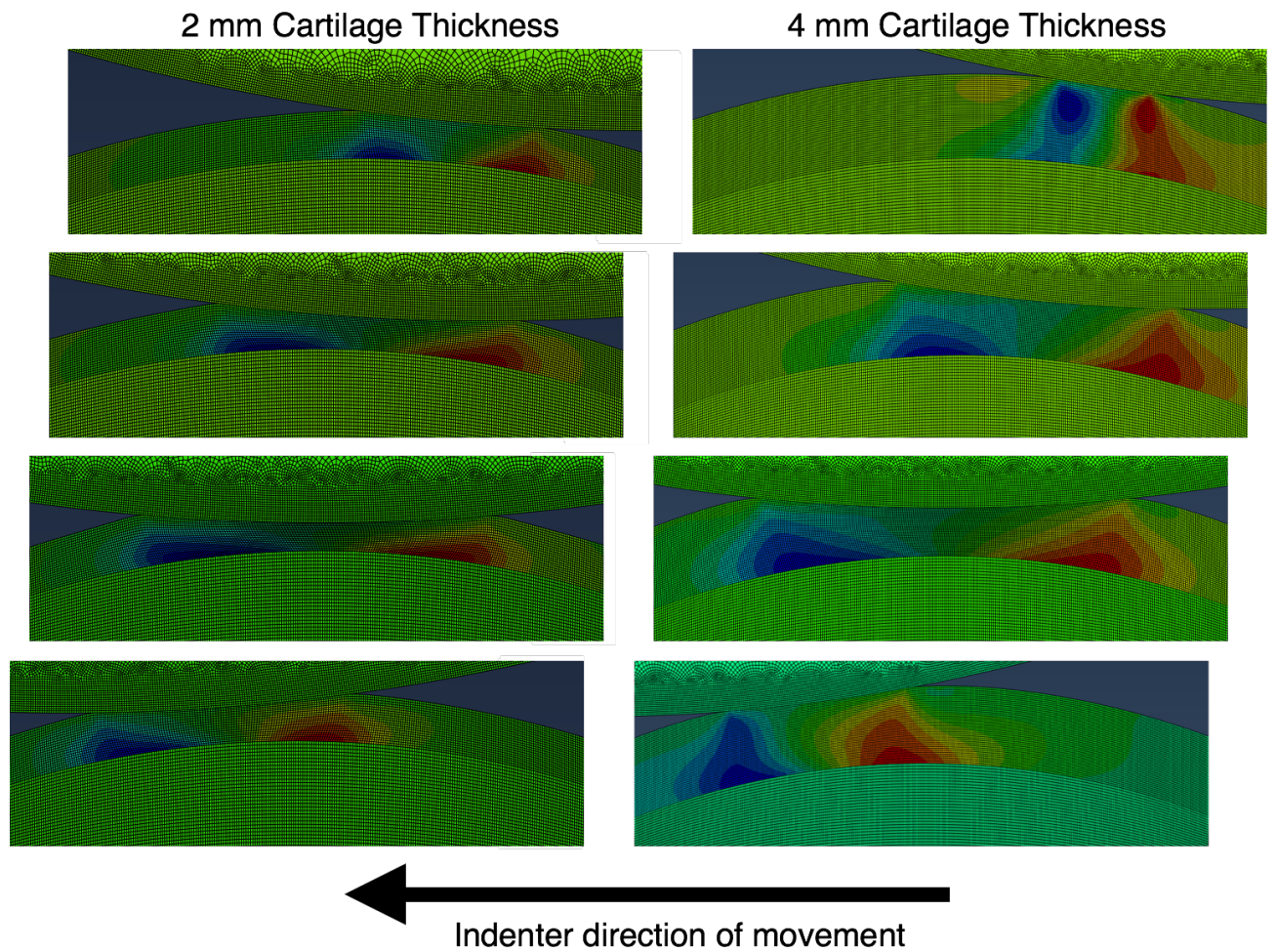


Figure 5.21: The shear strain results for the validation FE models simulating shear validation experiments are presented for 40% overlap and 2 mm-thick (left) and 4 mm-thick cartilage (right). For 2 mm-thick cartilage, the maximum (red) and minimum (blue) logarithmic shear strains were 0.44 and -0.6, respectively. For 4 mm-thick cartilage, the maximum (red) and minimum (blue) logarithmic shear strains were 0.20 and -0.28, respectively. For all models, shear strains developed near the cartilage-bone interface in the cartilage and translated along the interface as the indenter translated across the cartilage surface.

5.4.3 Comparison of Quantitative Experimental and FE Results

The validation FE results for 2 mm- and 4 mm-thick cartilage were compared with the average experimental results across Samples 1 to 5 for the compression (Figure 5.22) and shear (Figure 5.24) experiments as well as the individual experimental results for each sample for the compression (Figure 5.23) and shear (Figure 5.25) experiments. For the compression experiments, no experimental results were useful for the lowering of the indenter beyond 35% of the cartilage thickness because the cartilage often exuded water which disturbed the integrity of the speckle pattern.

Overall, the average experimental results were similar to the calculated FE results. Trends in shear strain matched in both sets of data. The average experimental results for the compression validation experiments lay close to the compression FE results for 2 mm and 4 mm-thick cartilage (Figure 5.22). The average experimental results for 25% overlap lay more or less within the FE results for 2 mm and 4 mm-thick cartilage, while the average experimental results for 40% overlap were slightly more positive than the FE results (Figure 5.24). The variation among the experimental sample results was much greater for the shear experiments compared to the compression experiments.

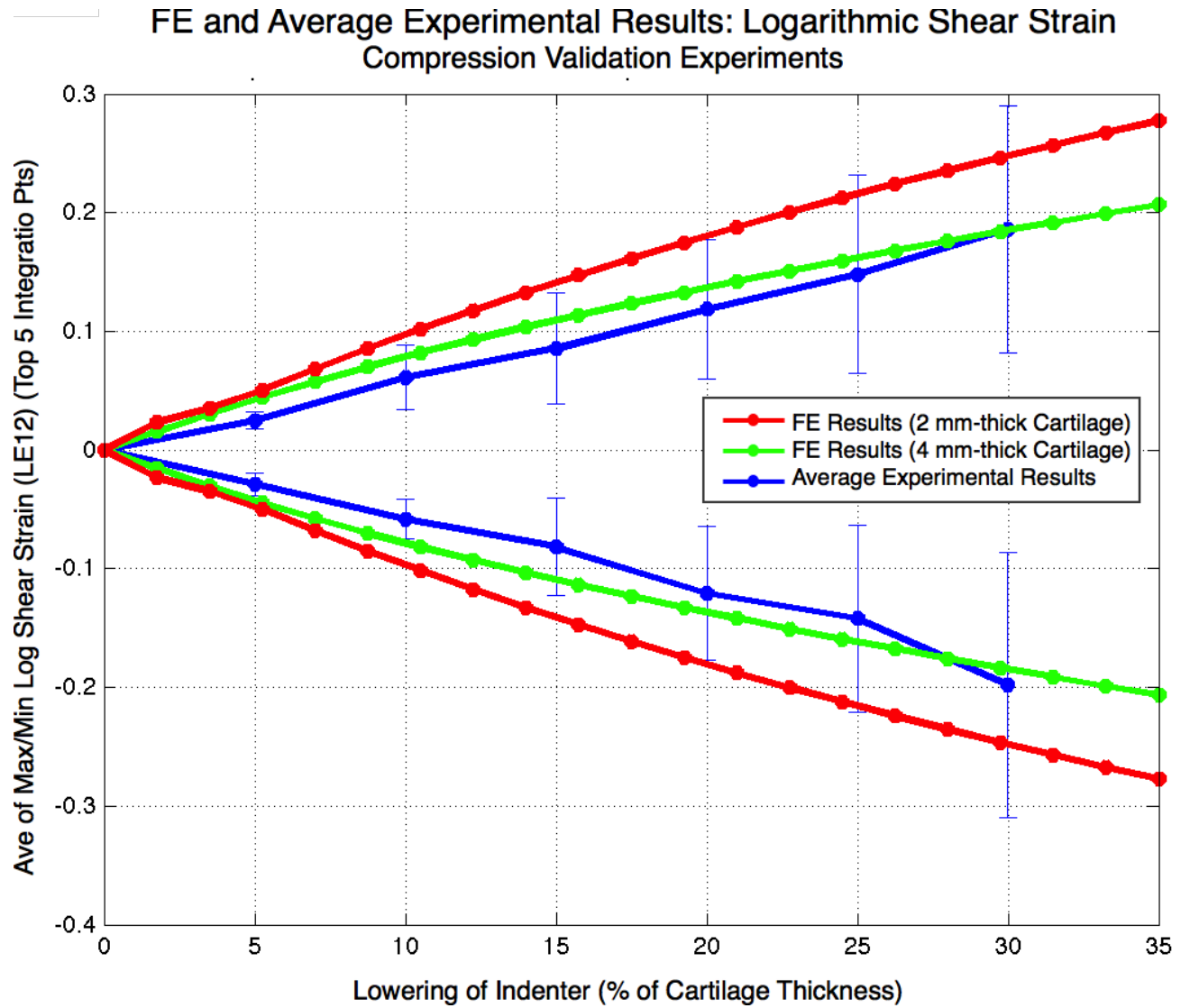


Figure 5.22: The quantitative shear strain results for the *in vitro* compression validation experiments averaged across Samples 1 to 5 are presented. The average of the experimental results lay close to the FE results for 4 mm-thick cartilage. Error bars represent one standard deviation.

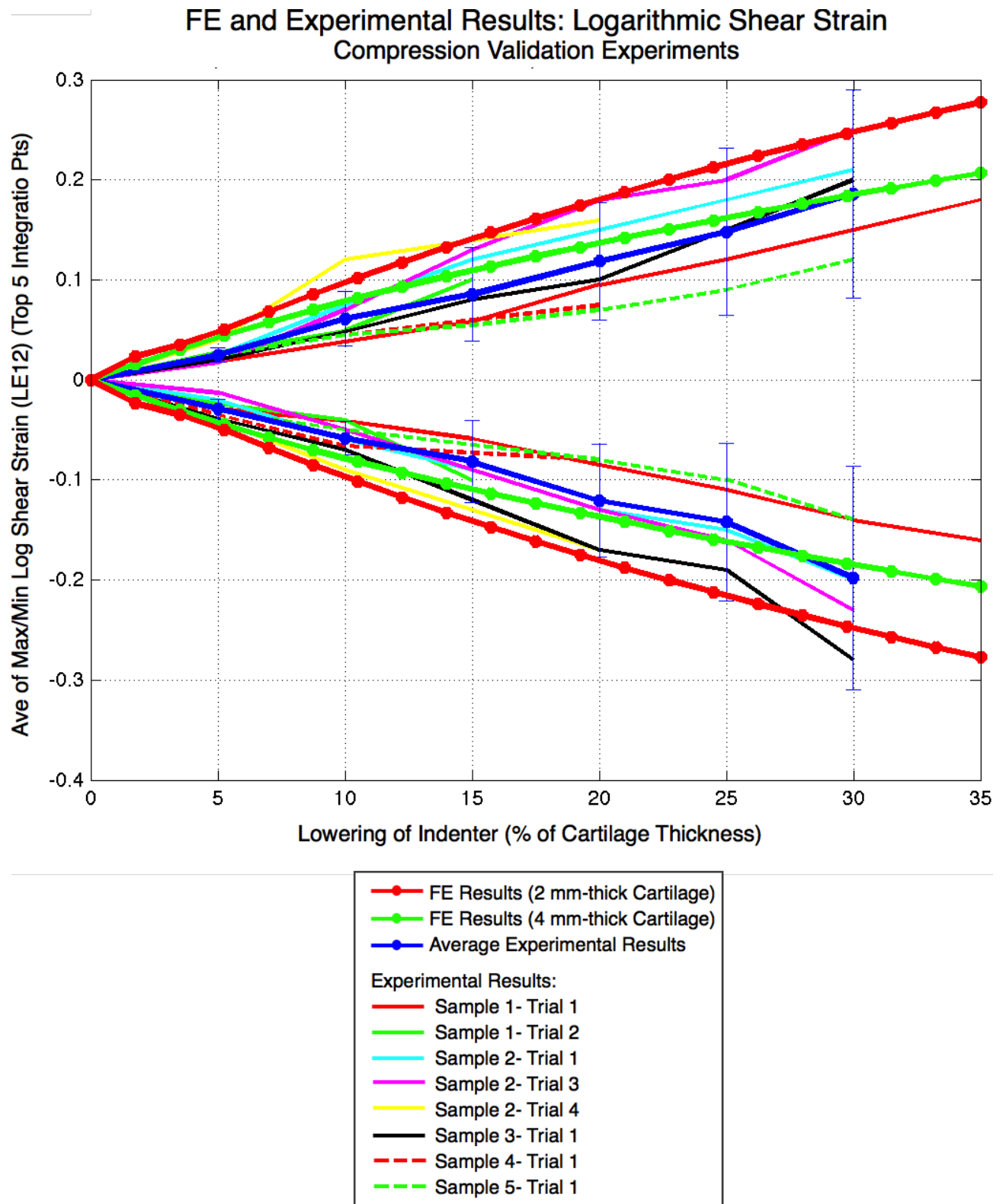


Figure 5.23: The quantitative shear strain results for the *in vitro* compression validation experiments for Samples 1 to 5 are presented. Error bars represent one standard deviation.

FE and Average Experimental Results: Logarithmic Shear Strain Shear Validation Experiments

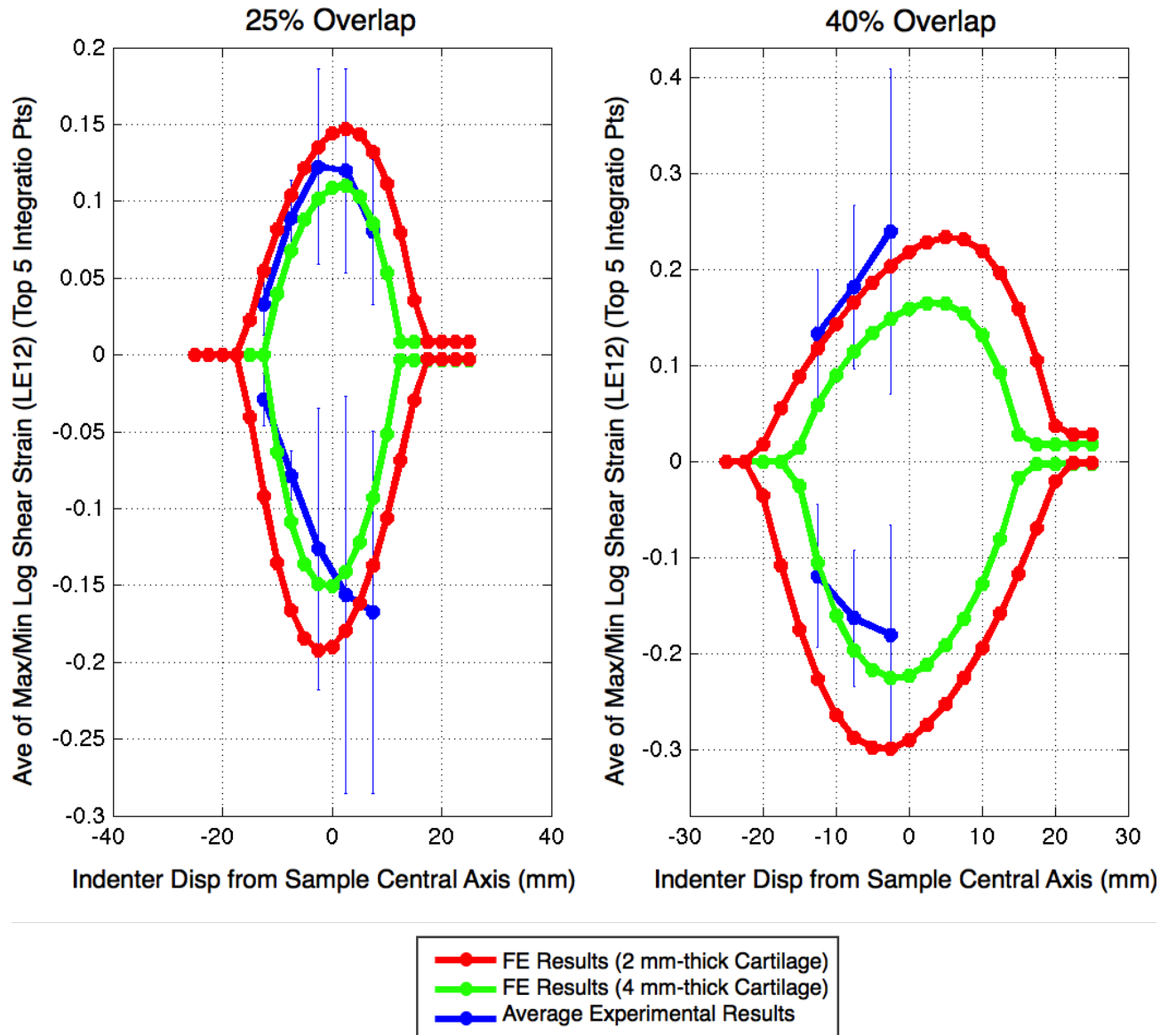


Figure 5.24: The quantitative shear strain results for the *in vitro* shear validation experiments averaged across Samples 1 to 5 are presented for 25% (left) and 40% overlap (right). The average of the experimental results for 25% overlap lies more or less within the FE results for 2 mm and 4 mm-thick cartilage, while the average of the experimental results for 40% overlap are slightly more positive than the FE results. Error bars represent one standard deviation.

FE and Experimental Results: Logarithmic Shear Strain Shear Validation Experiments

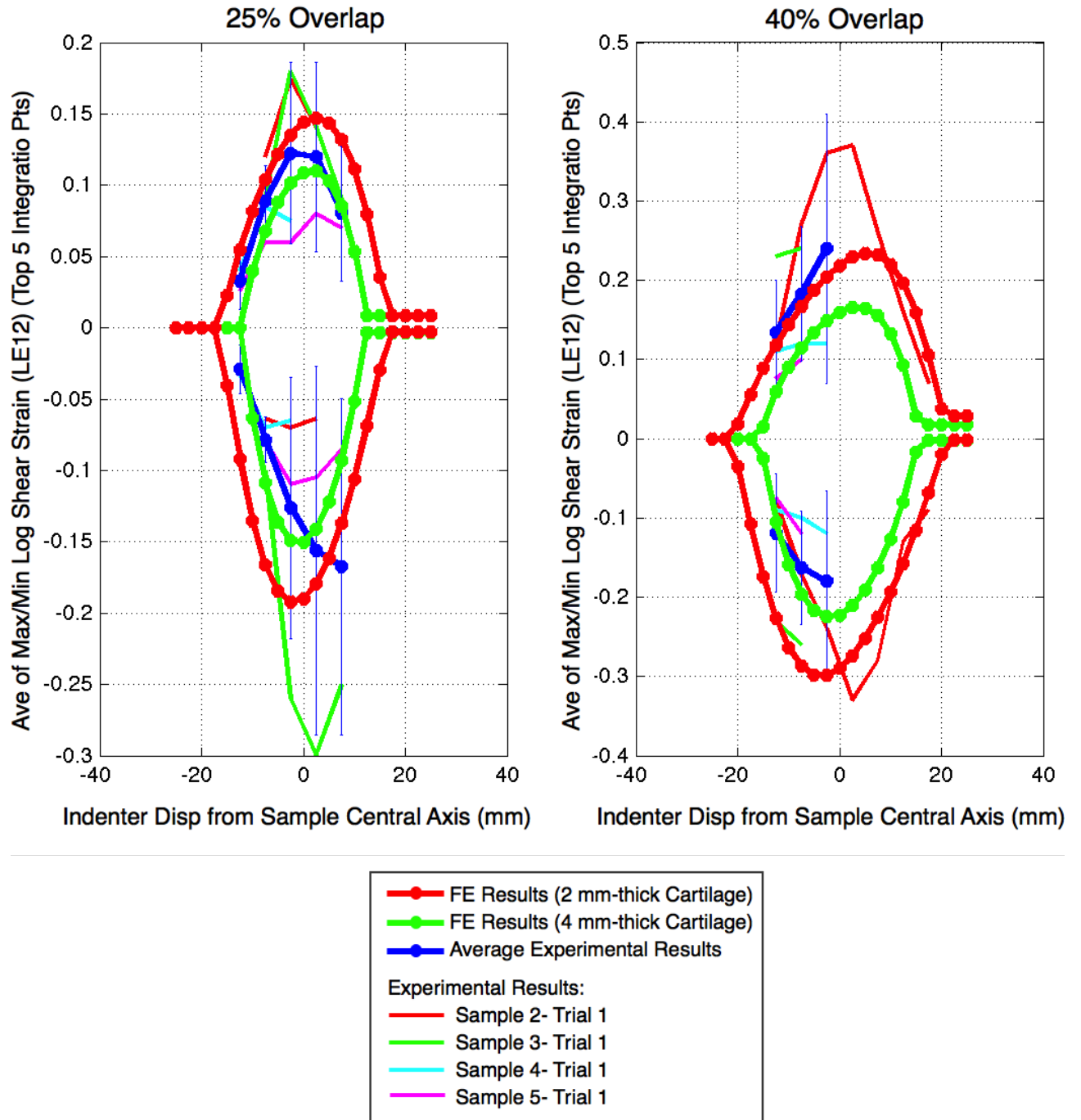


Figure 5.25: The quantitative shear strain results for the *in vitro* shear validation experiments for Samples 1 to 5 are presented for 25% (left) and 40% overlap (right). The variation in the results among the samples was much larger than the variation in the results for the compression validation experiments. Error bars represent one standard deviation.

5.4.4 *In Vitro* Damage Experiment Results

The samples before and after the damage experiments were performed are presented in figure 5.26. The images of the samples post-experiment were taken after the samples had been rehydrated so the resulting damage that occurred was more apparent. Samples 2 to 9 all demonstrated cartilage delamination, while a cartilage crack through the thickness of the cartilage also occurred in Sample 2, and a crack at the subchondral bone occurred in Sample 1. In Sample 4, the mid-section of the cartilage layer between the two pre-damaging cuts delaminated and fully detached from the bone. In Sample 8, only one side of the mid-section of the cartilage layer between the two pre-damaging cuts delaminated. For Samples 6 and 7, the edges of the pre-damaging centre cut delaminated. For Samples 5 and 9, both overhanging edges of the cartilage delaminated. In Sample 5, some subchondral bone was present prior to the damage test. It seems the shearing motion of the indenter during the damage experiment caused the damaged edge of the subchondral bone to detach from the tissue.

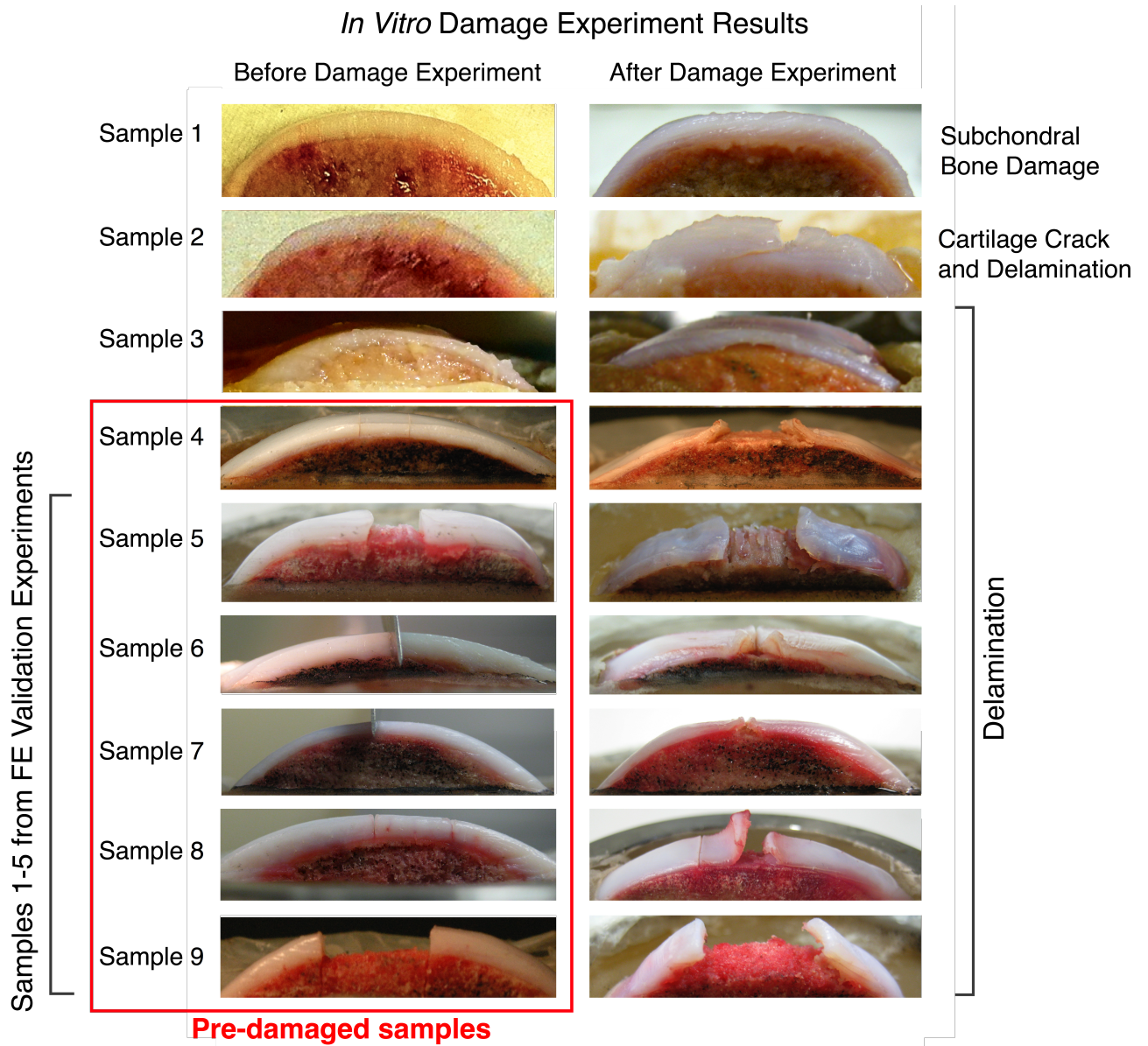


Figure 5.26: The samples before and after the damage experiment are shown. Samples 2 to 9 all demonstrated cartilage delamination, while a cartilage crack through the thickness of the cartilage also occurred in Sample 2, and a crack at the subchondral bone occurred in Sample 1.

5.5 Discussion

The use of human cartilage for the experimental tests would have been ideal, but this was not practical. However, the collagen structure of cartilage in porcine knees has been shown to be the closest to that of human compared with other commonly used animal models [162]. The elastic modulus and Poisson's ratio of porcine cartilage was also used in the validation FE model instead of human properties, to allow direct comparison of the experimental and FE results.

Using a video-based method for strain calculation utilising DIC was a novel way to directly measure strain in a soft tissue such as articular cartilage. For the loading conditions that were applied to the cartilage-bone samples, it was suitable for measuring shear strains in the cartilage. However, this method has limitations since deformation measurements using DIC is highly dependent on the preservation of a speckle pattern. However, soft tissues such as articular cartilage need to be kept hydrated to maintain their normal material properties, but can exude water when loaded, making the preservation of a speckle pattern on its surface challenging. Future work should consider non-water soluble dye for the speckle pattern.

Greater variation occurred among the experimental samples in the shear experimental results compared to the compression experimental results possibly due to higher water exudation during shear loading, lowering the integrity of the speckle pattern, and, therefore, creating errors and variation in shear strain measurements.

In general, variation in the measured shear strain results among the experimental samples was expected to occur due to differences in cartilage thickness and cartilage-bone interface curvature and inconsistencies in the curvature of the cartilage-bone interfaces. Additional variations could have occurred due to differences in material stiffness among the cartilage samples since the samples were taken from different areas of the femoral condyle of porcine knees. It has been reported that different material properties can occur as a function of location in the porcine knee [163]. Also, it was observed that upon loading, the cartilage would sometimes deform in planes other than the plane in which the speckle pattern was applied. This indicated that the cartilage in the plane where shear strain was measured was not as constrained as it would have been if it was not cut. This could have reduced the deformation of the cartilage in the measurement plane. Furthermore, the deformation of the speckle pattern in other planes could have made individual speckles appear to be larger in the view of the camera, causing artifactual measurements of deformation and, therefore, strain. Errors in orienting the sample could possibly have increased this effect. For instance, if the sample was oriented so it tilted towards the camera (decreased pitch), any type of compression would have caused more deformation to be out of the plane of strain measurement on the surface with the speckle pattern.

There were some experiments where shear strain was measured at the superficial layer of cartilage, while no shear strains were calculated using FE near the superficial layer of cartilage.

These experimentally measured shear strains could have possibly developed at the superficial layer of cartilage due to contact friction with the stainless steel indenting surface, especially if the articulating cartilage surface was not well hydrated. Also, these results could possibly indicate that the isotropic linear elastic material representation of the cartilage in the FE model was not suitable for measuring shear strains near the superficial layer of the cartilage. However, the shear strains measured near the cartilage-bone interface were much more dominant than those measured at the superficial layer of cartilage. Also, for the investigation of shear strains that commonly lead to damage at the cartilage-bone interface, small shear strains that occur near the superficial layer of cartilage are not of concern. Therefore, especially for the purpose of simulating cam-type FAI and the mechanism by which it causes damage such as cartilage delamination, an isotropic linear elastic material representation is considered to be appropriate.

Shear strains measured experimentally and calculated using FE were both found to be greatest near the cartilage-bone interface of the cartilage when loaded in compression or shear. For the purpose of validating the impingement FE model, finding that the region of shear strain development was similar in the experimental and FE calculated results was the most important criteria. It was even more important than finding that the quantitative shear strain results were comparable. This was the case because current hypotheses about how cam-type FAI leads to damage such as cartilage delamination revolve around large shear strains causing damage near the cartilage-bone interface. These hypotheses, therefore, hinge on that fact that large shear strains may occur near the cartilage-bone interface under loading conditions created by cam-type FAI. The experimental and FE shear strain results, therefore, not only validated the impingement FE model, but also verified that loading conditions similar to those experienced due to cam-type FAI (compression and shear) can lead to shear strains near the cartilage-bone interface of articular cartilage.

During the damage experiments, it was observed, for the first time, that cartilage could be delaminated by repeatedly applying shear loads which slowly increase in compression, verifying that shear and compression can indeed eventually lead to damage such as cartilage delamination. The progressive, but slow increase in compression ensured that water was squeezed out of the cartilage matrix, reducing the resilience of the cartilage, and allowing damage to occur at its interface with the underlying bone. It was also observed that if the resilience of the cartilage was not sufficiently compromised before additional compression was applied, other modes of

damage such as a superficial cartilage crack or a subchondral bone crack could possibly form. But even when other modes of damage such as a superficial cartilage crack occurred, it was still possible for delamination to occur as well. For example, Sample 2 for the *in vitro* damage experiments resulted in a cartilage crack which progressed towards the cartilage-bone interface where delamination of the cartilage begun to occur.

The pre-damaging cuts that were made to Samples 4 to 9 for the *in vitro* damage experiments created an edge in the cartilage which made it more similar to the anatomy of articular cartilage in the acetabulum near the acetabular rim. In all of these samples, cartilage delamination occurred starting at the cut edges of the cartilage. In cam-type FAI patients, cartilage commonly delaminates in the anterior acetabular articular cartilage near the chondrolabral junction which lies at the acetabular rim. In areas where the acetabular articular cartilage at the chondrolabral junction is not connected to the labrum at the chondrolabral junction, the anatomy of the acetabular articular cartilage in this region is similar to an edge. If loaded repetitively in shear and compression, similarly to the pre-damaged samples in the damage experiments, it can be hypothesised that the articular cartilage in this region may be highly susceptible to delamination compared to other areas of the hip joint. Furthermore, damage in this area due to cam-type FAI has been found clinically using MRI [164].

Sample 1 for the *in vitro* damage experiments was the only sample that did not result in cartilage delamination; it resulted in a crack between the subchondral and trabecular bone. For this sample, it is possible that the indenter was lowered too quickly before more water could have been squeezed out from the cartilage, allowing the deepest layer of cartilage to still be resilient to shear strain. It is also possible that the indenter was lowered too far so that it reached slightly beyond the cartilage-bone interface, resulting in damage to the subchondral bone.

5.6 Conclusion

A video-based method for strain calculation utilising DIC was successfully created to calculate shear strain in cartilage-bone samples loaded under compression and shear *in vitro*. The experimentally measured shear strains were comparable with the results calculated using the validation FE models based on location and quantitatively which gave confidence in the 2D

FE models and, therefore, validated the contact and material representations used in the impingement FE model developed in Stage 5 of Chapter 4. Therefore, shear strain results using the impingement FE results can be expected to be realistic. Also, the use of an isotropic linear elastic material representation for bone and articular cartilage has been validated for the purpose of strain analysis when modelling fast loading which occurs during functional activities, especially when high shear and compressive loading is expected.

The shear strains measured experimentally and calculated using FE were consistently focused near the cartilage-bone interface in the articular cartilage, confirming that loading conditions similar to those experienced due to cam-type FAI (compression and shear) can lead to shear strains in the region where cartilage often becomes damaged and delaminates from the underlying bone. Furthermore, the damage experiments showed that applying these loading conditions repetitively and more aggressively could reliably cause cartilage to delaminate, especially when an edge in the morphology of the articular cartilage was present (similar to its condition in the hip joint near the acetabular rim). Assuming that having an impinging cam deformity can lead to greater compression and shear loading in the hip joint during motion, these experiments support the hypothesis that such loading conditions can eventually lead to damage such as cartilage delamination. Therefore, the validation and damage experiments, in conjunction, have verified the hypothesised (general) damage mechanism by which cam-type FAI leads to cartilage delamination. Further research, however, would need to be conducted to verify the hypothesised detailed damage mechanism described in Section 2.5 by which large shear strains can cause the propagation of “micro-damage” (the breaking of the entwinements between collagen fibers) and “micro-cracking” (the failure of individual collagen fibers) until a crack forms at the cartilage-bone interface, initiating cartilage delamination.

Further insights have also been made suggesting that articular cartilage is most vulnerable to delamination when most of its water has exuded out from the tissue due to compression. Also, the acetabular articular cartilage in the region of the chondrolabral junction, where it is not connected to the labrum, may be a region where the initiation of cartilage delamination is highly probable.

Chapter 6

2D FE Model Simulating Cam-type FAI: Investigating Cam Protrusion, Cam Width, and Hip Joint Clearance

6.1 Introduction

In this chapter, the FE model developed in Stage 5 of Chapter 4 was used for parameter studies which investigated how varying hip joint clearance, cam width, and cam protrusion may lead to varying types of damage in the hip joint and varying locations of damage initiation. These parameters were described and defined in Section 4.6.2.

The ranges for a small and large cam protrusion were developed in the pilot study in Section 4.6. A preliminary parameter study is reported in this chapter which examines appropriate ranges for hip joint clearance and cam width and confirms that the cam protrusion ranges used in Section 4.6 are appropriate as well. After the exploration of the chosen geometric parameters, a mesh convergence study was conducted. Finally, the main parameter study investigating cam protrusion, cam width, and hip joint clearance was conducted using FE models which incorporated appropriate element sizes based on the mesh convergence study.

6.2 Preliminary Parameter Study

A preliminary parameter study was conducted to confirm the ranges of hip joint clearance and cam protrusion and width which should be used in the main parameter study presented

later in this chapter for strain analysis (Section 6.4). FE models were created with a range of various geometric parameters. Damage analysis using XFEM was conducted as well as the degeneration of the acetabular soft tissues. The geometric parameter ranges were assessed on whether they could produce the formation of different types of damage (labral and cartilage) and different damage initiation locations that relate to clinically seen pathology.

6.2.1 Methods

Geometric Parameters

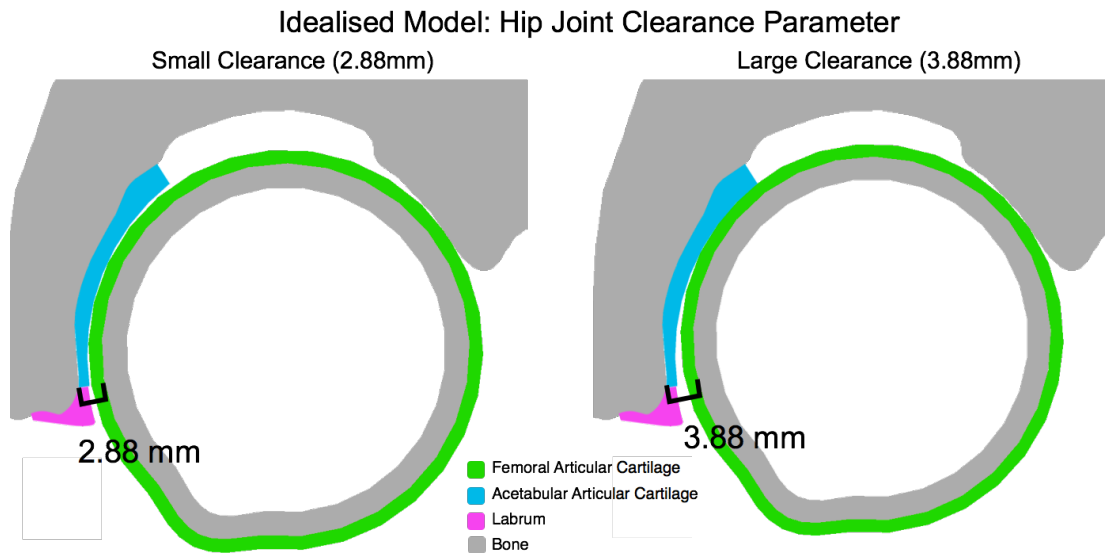


Figure 6.1: FE models with small and large hip joint clearance are presented for comparison.

Preliminary parameter studies were performed for two sets of models – one with a small hip joint clearance and one with a large hip joint clearance (Figure 6.1). Hip joint clearance was described to be the shortest distance from the acetabular bone at the labral-junction to the femoral bone in Section 4.6.1. The small hip joint clearance of 2.88 mm was used similarly to the hip joint clearance of the impingement FE model described in Section 4.6. The cam protrusion of the small cam used in the impingement FE model described in Section 4.6 was under 1 mm. A hip joint clearance that was 1 mm larger than a small hip joint clearance, was therefore chosen to represent a large hip joint clearance (3.88 mm). The cam protrusion sizes consisted of small and large protrusions which were calculated to be 25% and 75% of the hip joint clearance, respectively. This formulation was already verified to be able to lead to different kinds of damage (cartilage and labral) in the pilot study described in Section 4.6 and resulted in cam protrusions of 0.72 mm (small) and 2.16 mm (large) for the small hip joint

clearance model (Figure 6.2) and 0.97 mm (small) and 2.91 mm (large) for the large hip joint clearance mode (Figure 6.3). The three cam width sizes for the small hip joint set consisted of small (30°), medium (60°), and large (90°), whereas only the small (30°) and large (90°) cam width sizes were used for the large hip joint clearance set. The geometric parameters used are summarised in Table 6.1 below.

Hip Joint Clearance Model	Chosen Cam Protrusion Parameters	Chosen Cam Width Parameters
Small (2.88 mm)	1) Small (0.72 mm) 2) Large (2.13 mm)	1) Small (30°) 2) Medium (60°) 3) large (90°)
Large (3.88 mm)	1) Small (0.97 mm) 2) Large (2.91 mm)	1) Small (30°) 2) Large (90°)

Table 6.1: The cam protrusion and width parameters that were chosen to be investigated for models with small and large hip joint clearance are listed.

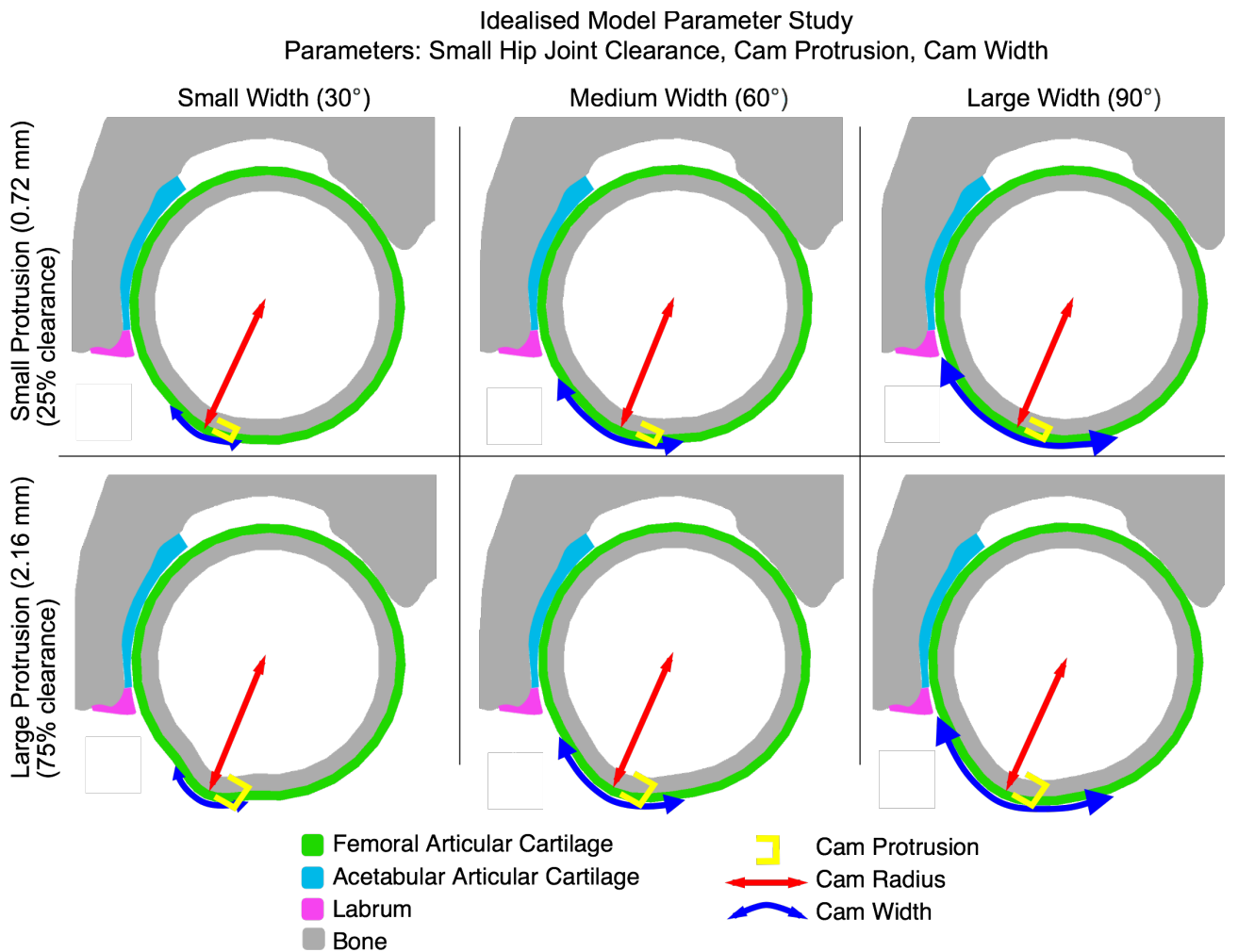


Figure 6.2: FE models with small hip joint clearance and parametrised cam protrusion and width are presented for comparison.

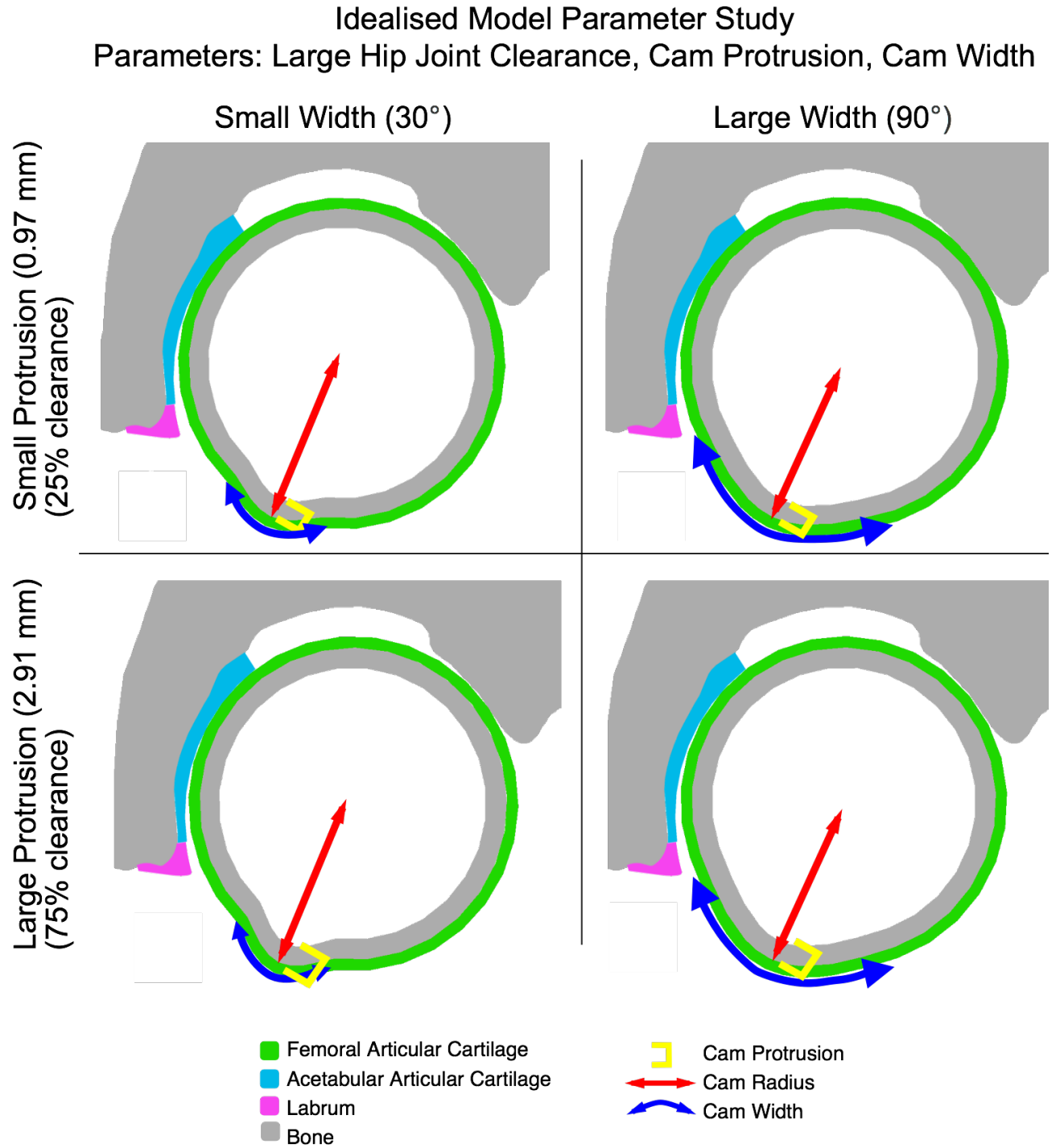


Figure 6.3: FE models with large hip joint clearance and parametrised cam protrusion and width are presented for comparison.

Mesh Generation, Assigning Boundary Conditions, and Defining Material Properties, Interactions, and Analysis Steps

Similar mesh generation, boundary conditions, material properties, interactions, and analysis steps were applied in the FE models as they were applied in the Internal Rotation at Flexion model in Stage 4 in Section 4.5.1. These analysis formulations created a model which simulated the internal rotation of the femoral head during the activity of sitting when the hip was flexed

at 40° (with a joint load of 0.1 N medially and 1.92 N anteriorly). In the presence of a cam deformity, these FE models, therefore, simulated cam-type FAI during the motion of internal rotation at flexion.

Defining Damage

A similar damage definition as defined for the model in Section 4.5.1 was used. This consisted of using XFEM to model damage such as tears or cracks using a strain criterion of 0.1 in all directions for damage initiation. A maximum strain criterion was used where the maximum strain was defined to be 0.1 in all directions. The evolution of damage was defined by a fracture energy of $0.009 \text{ MPa}(mm)^{0.5}$ in the direction parallel to the labral-bone and cartilage-bone interface for the labrum and anterior acetabular articular cartilage, respectively; a fracture energy of $0.09 \text{ MPa}(mm)^{0.5}$ was defined in the direction perpendicular to the labral-bone and cartilage-bone interface for the labrum and anterior acetabular articular cartilage, respectively.

Simulating Soft Tissue Degeneration

The degeneration of the acetabular soft tissues in the hip joint were simulated by reducing their material stiffness to 30% of the original stiffness. Anterior acetabular articular cartilage degeneration was simulated in models with cartilage-damaging cams (those with small protrusion) as demonstrated in Section 4.6.3. Labral degeneration was simulated in models with labrum-damaging cams (those with large protrusion) as demonstrated in Section 4.6.3.

6.2.2 Results

For the small hip joint clearance models, no cracks resulted unless tissue degeneration was defined. For the models with small cam protrusion and anterior acetabular articular cartilage degeneration, cracks formed along the cartilage-bone interface in the cartilage (Figure 6.4). For the models with large cam protrusion and labral degeneration, cracks formed either at the labral tip or near the labral-bone junction in the labrum (Figure 6.4).

For the large hip joint clearance models, cracks resulted prior to tissue degeneration, however, there was no apparent trend in the formation of cracks; XFEM cracks either did not initiate, or they initiated in only the labrum, or they initiated in both the cartilage and labrum (Figure 6.5).

Tissue Degeneration and Damage Analysis Small Hip Joint Clearance Model

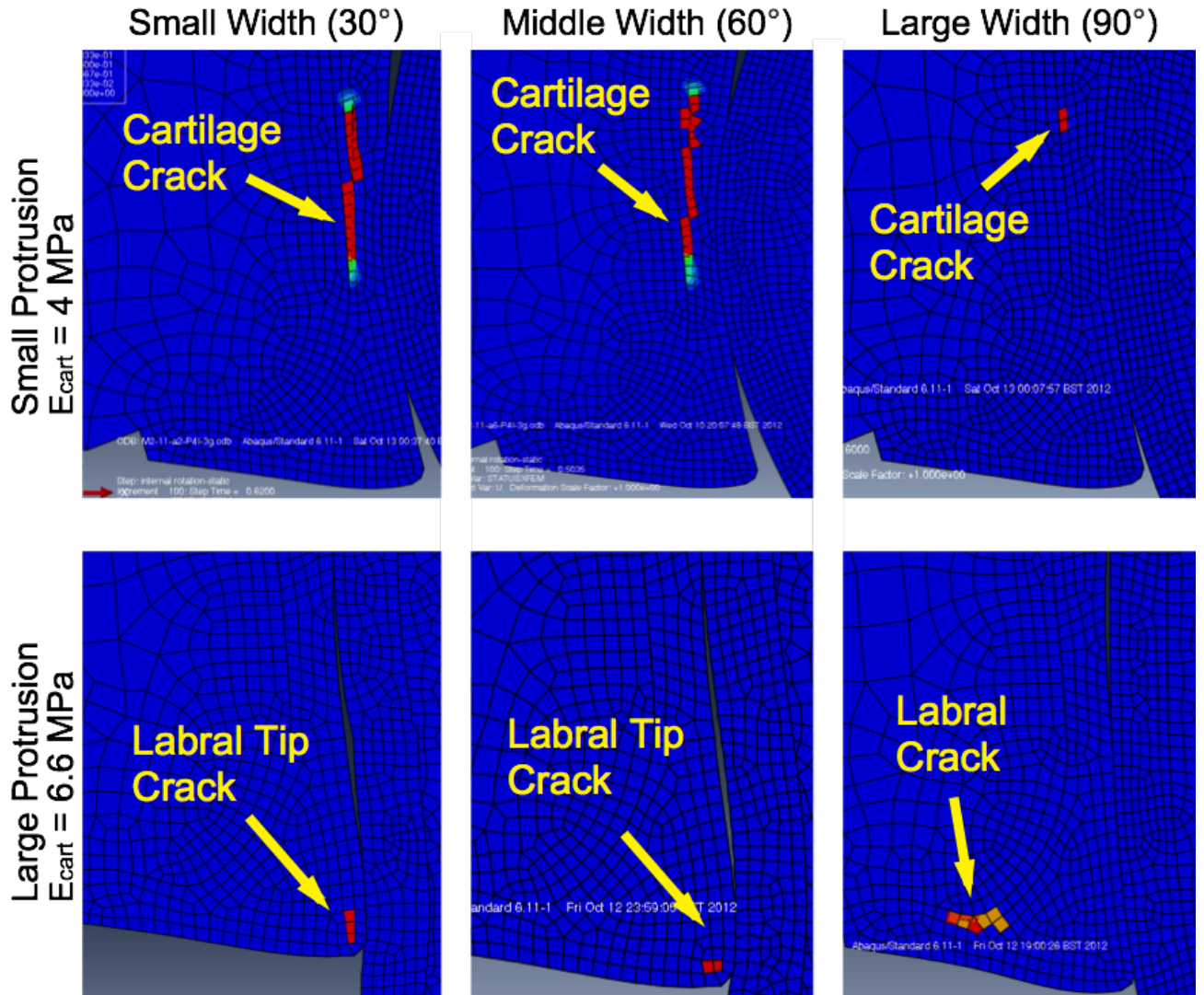


Figure 6.4: The XFEM results for a preliminary FE study parametrising cam protrusion and width for a hip with small clearance are presented. Tissue degeneration was simulated by decreasing the stiffness of the anterior acetabular articular cartilage or labrum to 30% of their original stiffness. XFEM cracks were initiated at different locations along the cartilage-bone interface or in the labrum for the models with small and large cam protrusion, respectively.

After tissue degeneration was defined in the anterior acetabular articular cartilage for the models with small cam protrusion and in the labrum for the models with large cam protrusion, cracks occurred in all of the soft tissues where degeneration was simulated (Figure 6.5). This trend was continued when further investigation was conducted where the labrum was degenerated for the model with a small cam protrusion and width; a labral crack near its articulating surface resulted. However, when the cartilage was degenerated in the model with a large cam protrusion and small cam width, crack initiation occurred near the labral-bone junction, not the cartilage. Thus, no clear trend was observed for damage initiation in the large hip joint clearance models.

Tissue Degeneration and Damage Analysis Large Hip Joint Clearance Model

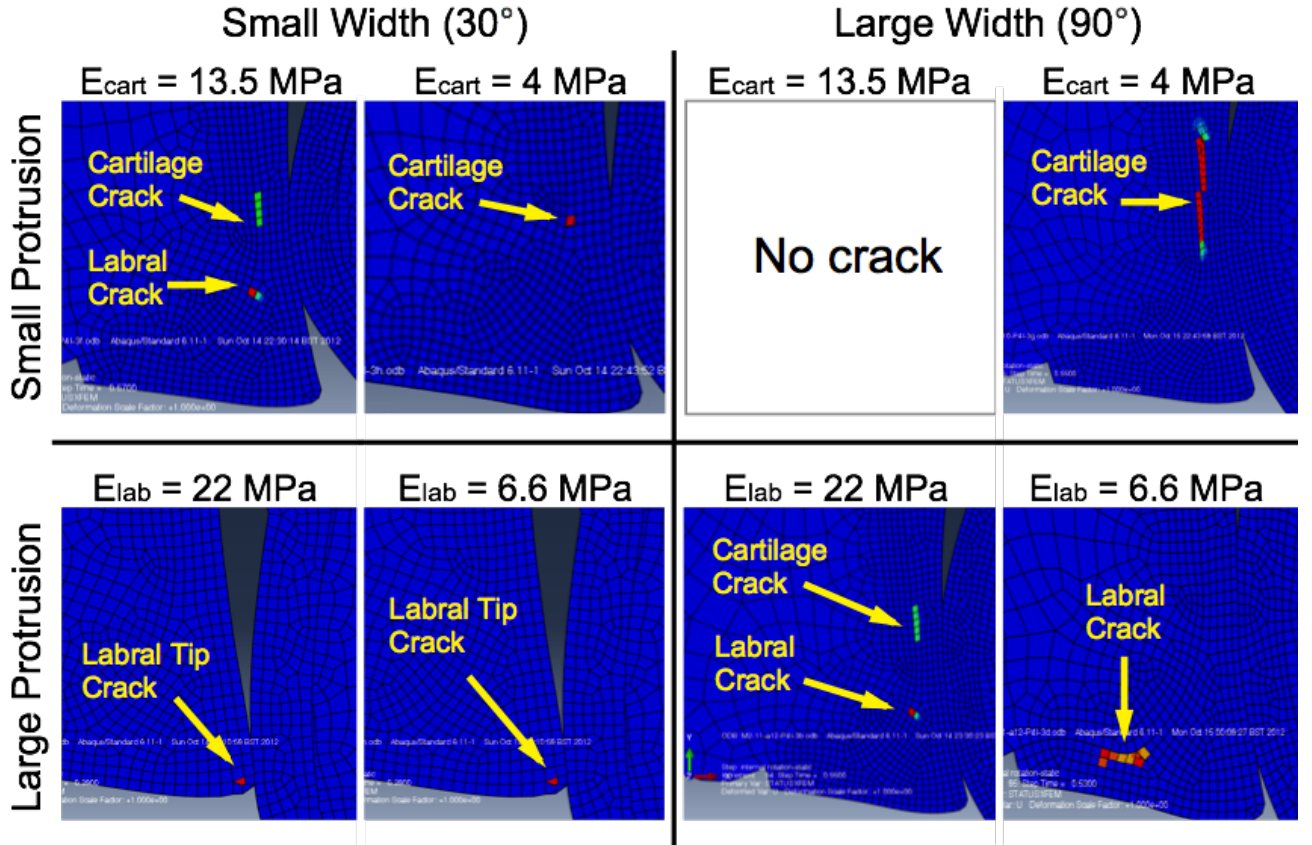


Figure 6.5: The XFEM results for a preliminary FE study parametrising cam protrusion and width for a hip with large clearance are presented. For normal stiffness, XFEM cracks either did not initiate, or they initiated in only the labrum, or they initiated in both the cartilage and labrum. When tissue degeneration was simulated by decreasing the stiffness of the articular cartilage or labrum, a cartilage or labral crack initiated if the cartilage or labrum was degenerated, respectively.

The damage initiation location was plotted for models with small hip joint clearance and varying cam protrusion and width (Figure 6.6). For models with small cam protrusion, the larger the cam width was, the further damage initiation occurred from the chondrolabral junction. For models with large cam protrusion, a large cam width resulted in damage initiating near the labral-bone interface, while a small or middle cam width resulted in damage at the labral tip.

6.2.3 Discussion

For all the models, cracks that were simulated in the anterior acetabular articular cartilage always lay near the cartilage-bone interface and often progressed along the interface towards the inner regions of the hip joint. Such damage represents the initiation and progression of

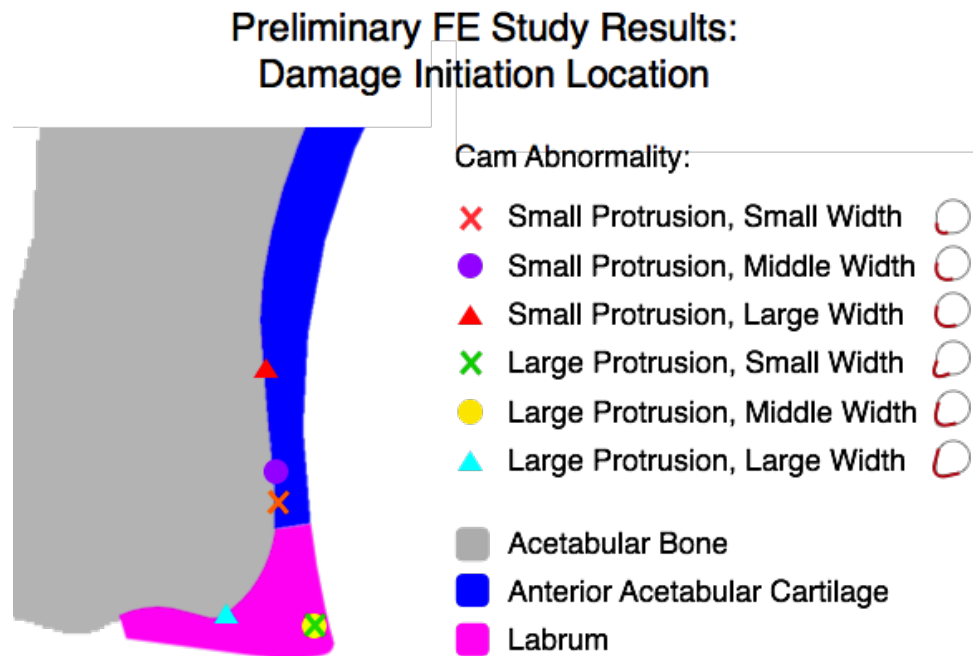


Figure 6.6: The damage initiation locations based on XFEM for parameterised cam protrusion and width for a hip with small clearance is depicted. A strain criterion for damage initiation of 0.1 was defined in all directions. For models with small cam protrusion, the larger the cam width, the further the damage initiation occurred from the chondrolabral junction along the cartilage-bone interface. For models with large cam protrusion, a large cam width resulted in damage initiating near the labral-bone interface, while a small or middle cam width resulted in damage at the labral tip.

cartilage delamination which is commonly seen in cam-type FAI patients in the anterior superior acetabular region of articular cartilage [20, 21, 22, 23]. Cracks that were simulated in the labrum lay either near the labral-bone interface or at the labral tip. Such damage represented the initiation of the partial detachment of the labrum from the acetabular rim (which could possibly progress to the full detachment of the labrum) and of a crack within the labrum, respectively, which are both commonly seen in cam-type FAI patients [20, 21, 22, 23, 24, 50]. For some of the large hip joint models, a crack initiating in the labrum and acetabular articular cartilage near the chondrolabral junction occurred. The formation of neighbouring damage in the labrum and articular cartilage has also been clinically observed [20, 125, 164]. The damage results, therefore, simulated damage commonly clinically associated with cam-type FAI patients.

Cracks occurred in the large hip joint clearance models even prior to tissue degeneration, whereas prior to tissue degeneration, no cracks occurred in the small hip joint clearance models. These results are counterintuitive since having a larger clearance in the hip joint was hypothesised to be less damaging to the acetabular soft tissues with respect to cam-type FAI.

However, the small and large cam protrusions that were used were a certain percentage of the hip joint clearance, so the relative cam protrusions that the small and large hip joint clearance models had for their cam deformities were similar. It is probable that having a larger hip joint clearance would be less damaging compared to having a small hip joint clearance if the cam protrusion in the models were the same. Overall, these results suggest that having a larger hip joint clearance may actually be more damaging compared to having a smaller hip joint clearance, if the size of the cam protrusion is proportional to the hip joint clearance.

6.2.4 Conclusion

The geometric parameters investigated in this study appeared to be involved in the main types of damage reported in cam-type FAI, leading to damage that formed along the cartilage-bone interface, at the labral tip, and near the labral-bone interface. Therefore, the geometric parameters developed in this study were concluded to be suitable and realistic for the main parameter study conducted later in this chapter.

The results in this preliminary study also showed that in a hip with small clearance, a small-protruding cam is likely to cause damage leading to cartilage delamination, while a large-protruding cam is likely to cause labral damage. Additionally, for a small-protruding cam, a larger cam width is likely to cause damage to initiate deeper within the hip joint. These trends indicate that having a smaller hip joint clearance may enable easier predictions of possible damage that might occur due to a cam of a certain shape and size. Finally, having a larger hip joint clearance may possibly be more damaging than having a smaller hip joint clearance, if the size of the cam protrusion is proportional to the hip joint clearance.

6.3 Mesh Convergence Study

6.3.1 Introduction

A mesh convergence study was conducted on the Internal Rotation at Flexion impingement FE model developed in Section 4.6. Since the shear strain results of the acetabular articular cartilage and labrum obtained from the impingement FE model were of interest, a mesh convergence study was conducted on the anterior and posterior acetabular articular cartilage and labrum components of the model. Meshes which led to full convergence in these three parts

were considered to be appropriate for obtaining meaningful results.

6.3.2 Methods

The soft tissue parts including the anterior and posterior acetabular articular cartilage, femoral head articular cartilage, and the labrum were meshed with elements of the following sizes: 0.8, 0.6, 0.4, 0.3, 0.25, 0.2, 0.15, and 0.1 mm. The bone surfaces that were tied to the labral and cartilage surfaces were meshed with the same element sizes, but all other bone surfaces were meshed with element sizes that were equivalent to four times the size of the elements used for the soft tissues. A mesh convergence study was run for two models, both with small hip joint clearance and a middle cam widths, but one with a small cam protrusion, and one with a large cam protrusion.

The shear strain results were extracted from the Abaqus output database files and analysed to determine the maximum and minimum shear strain values since each shear strain result has a positive and negative component. An average of the top five maximum and minimum values were used as the calculated values to test for convergence in the anterior and posterior acetabular cartilage components. An average of the top three maximum and minimum values were used as the calculated values to test for convergence in the labrum. An average of fewer values were used for the labrum because the maximum and minimum shear strains tended to affect fewer elements (cover less area) in the labrum.

The convergence of shear strain of the parameter model with a small cam protrusion was investigated when the shear strains were calculated to be the greatest in the labrum and anterior acetabular articular cartilage during internal rotation at flexion; this occurred when 22.5° and 45.0° of internal rotation was applied, respectively, and when 85.5° of internal rotation was applied for both of the soft tissues (Figure 6.7). For the parameter model with a large cam protrusion, the calculated shear strains increased for increasing internal rotation, so convergence was investigated at when the greatest amount of internal rotation was applied which was 31.5° (Figure 6.8). The models were considered to have fully converged if for three consecutive models, the difference in the results for shear strain between consecutive models was less than 5% [159] (Eqn 5.2).

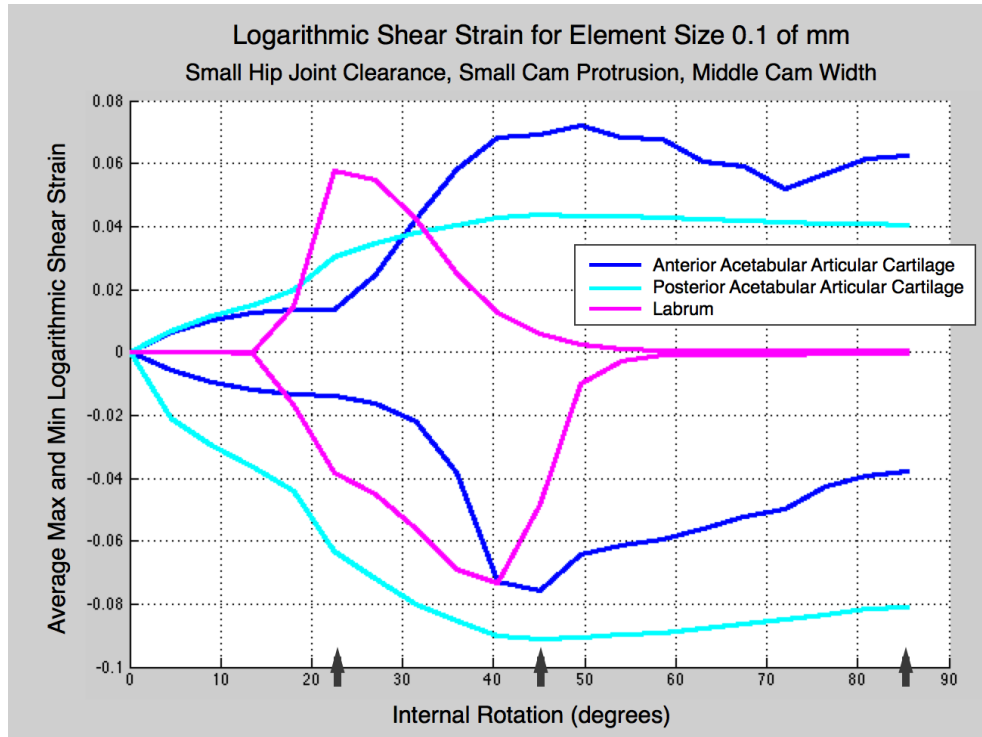


Figure 6.7: The shear strain for the FE model with small hip joint clearance, small cam protrusion, and middle cam width that was meshed with a minimum element size of 0.1 mm is presented. A mesh convergence study was chosen to be conducted at 22.5°, 45.0°, and 85.5° of internal rotation (indicated by black arrows).

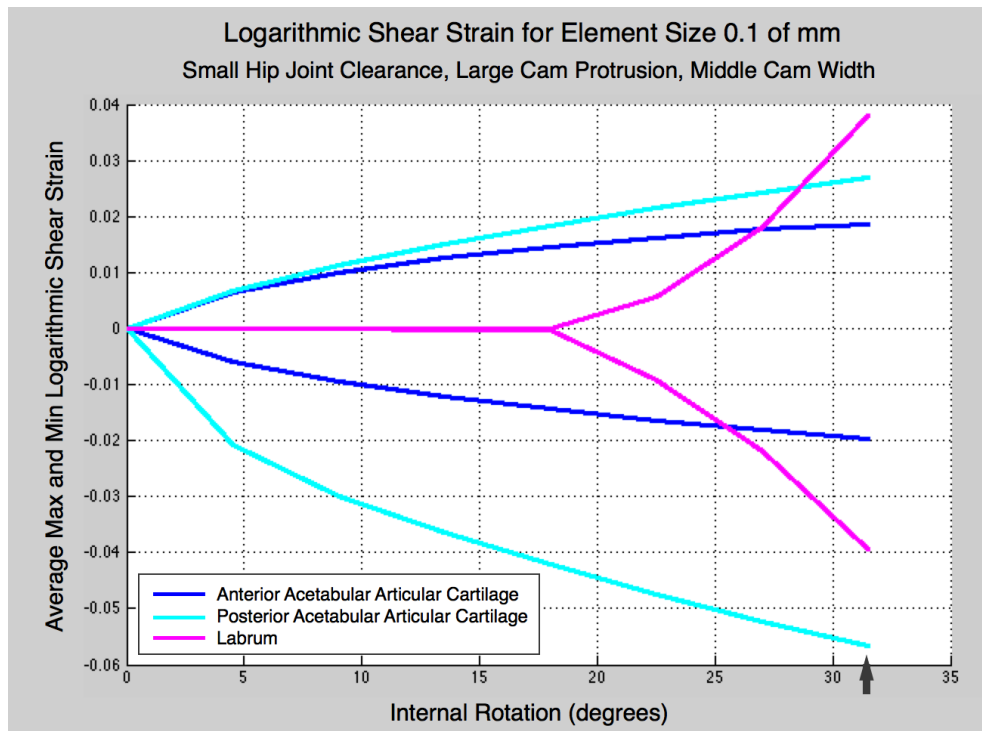


Figure 6.8: The shear strain for the FE model with small hip joint clearance, large cam protrusion, and middle cam width that was meshed with a minimum element size of 0.1 mm is presented. A mesh convergence study was chosen to be conducted at 31.5° of internal rotation (indicated by black arrow).

6.3.3 Results

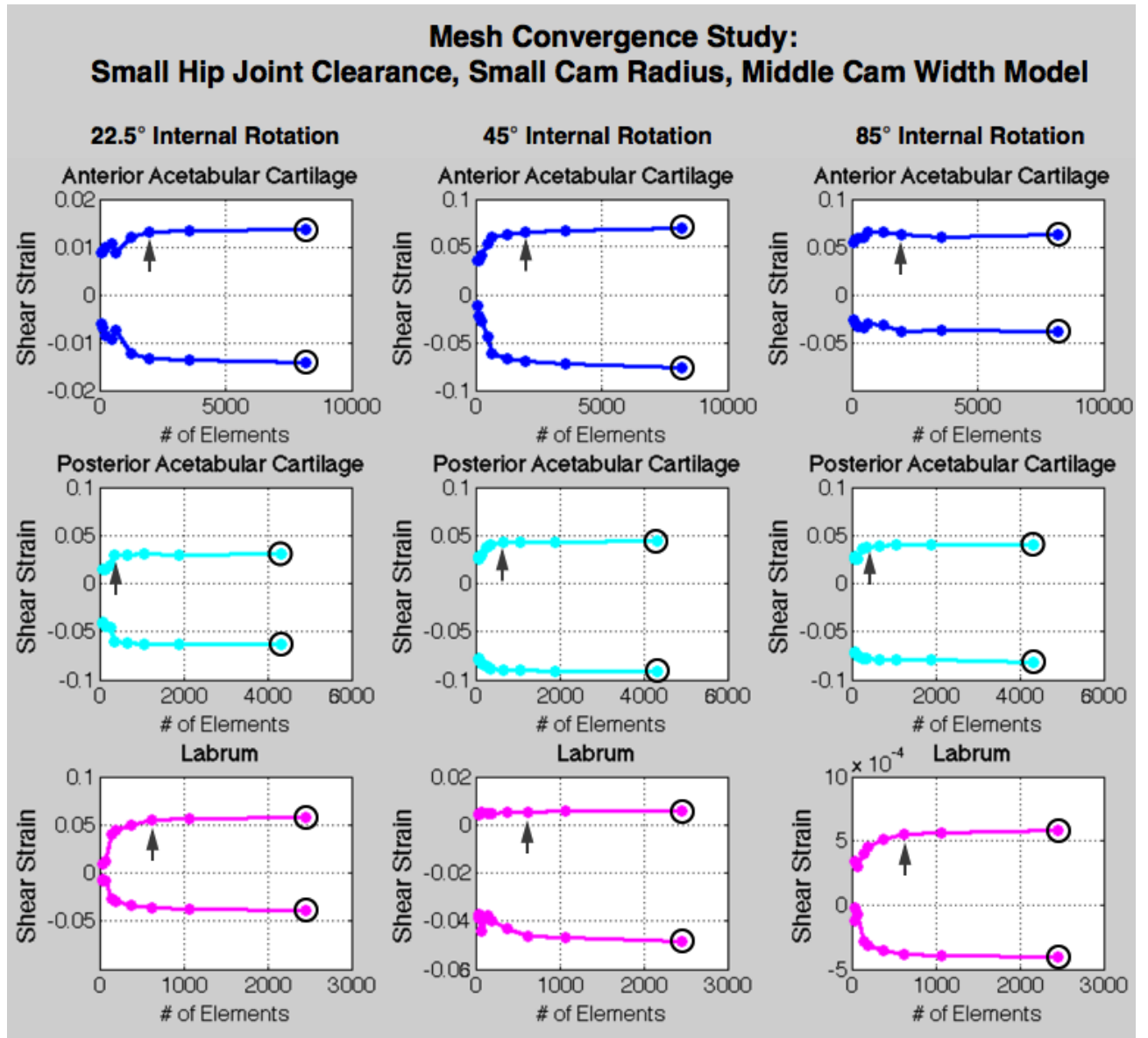


Figure 6.9: The mesh convergence study results are presented for the FE model with small hip joint clearance, small cam protrusion and middle cam width. Mesh convergence was investigated at 22.5°, 45.0°, and 85.5° of internal rotation. An arrow indicates full convergence was met for that model and for all models with more refined mesh sizes. A circle indicates the model with the element size of 0.1 mm which was chosen to be used to run final results.

The mesh convergence study results are presented for the model with small (Figure 6.9) and large cam protrusion (Figure 6.10). An arrow was used to indicate that the shear strain results for a model, and the models with a more refined mesh, were within 5% of the shear strain results of the consecutive model with a more refined mesh (Figures 6.9, 6.10). For the small hip joint model, full convergence was met for the posterior acetabular articular cartilage at the three time points investigated for an element size of 0.15 or smaller. For the large hip joint model, full convergence was met for the posterior acetabular articular cartilage at the

time point investigated for an element size of 0.25 or smaller. For anterior acetabular articular cartilage and labrum in the small and large hip joint clearance models, full convergence was reached for an element size of 0.1 mm. Models with this element size are indicated with a circle (Figures 6.9, 6.10).

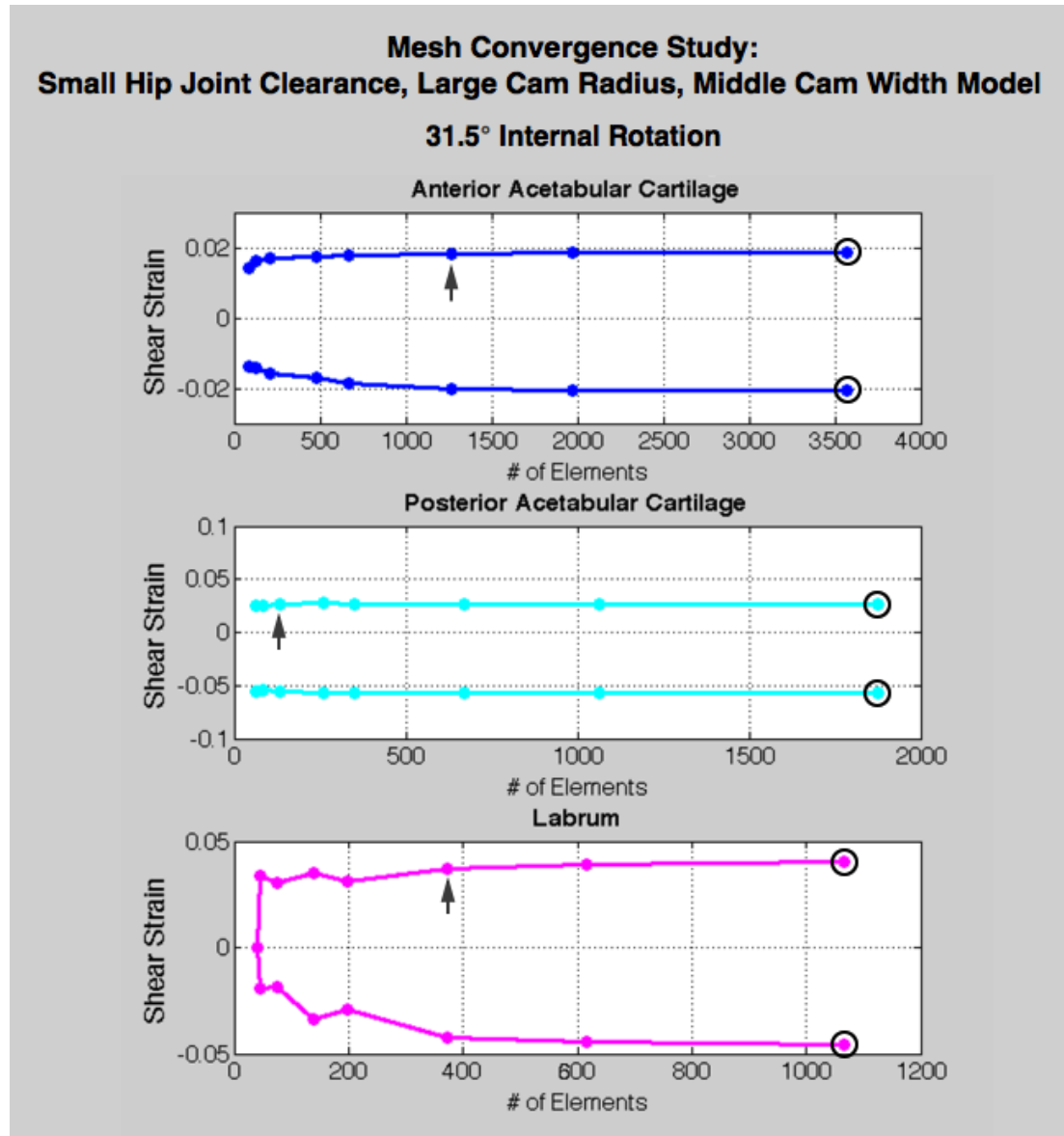


Figure 6.10: The mesh convergence study results are presented for the FE model with small hip joint clearance, large cam protrusion and middle cam width. Mesh convergence was investigated at 31.5° internal rotation. An arrow indicates full convergence was met for that model and for all models with more refined mesh sizes. A circle indicates the model with the element size of 0.1 mm which was chosen to be used to run final results.

6.3.4 Discussion

A modelling error was found in the tie constraint between the articular cartilage and bone of the femoral head in the models used to conduct the mesh convergence studies. This error resulted

from imperfect matching of the tied surfaces of the cartilage and bone, and thus, caused the surfaces that did not match perfectly to not be tied. The cartilage-bone interface in the area of the cam deformity was affected by this error. This resulted in a slight overlap of the femoral head articular cartilage with the femoral head bone during impingement. In effect, the simulations of impingement were results of cams with a smaller protrusion. Since the mechanism that was simulated in the models was still considered to be characteristic of cam-type FAI, the mesh convergence study results were considered to be valid. Also, the mesh convergence study did not require shear strains on the femoral articular cartilage to be investigated. Only the shear strains in the acetabular soft tissues were of interest because the acetabular soft tissues are where damage usually occurs in cam-type FAI patients. Therefore, it was determined that the modelling error did not affect the mesh convergence study results. The tie constraint error was corrected for all models, hereafter, so all final results were produced using corrected models.

6.3.5 Conclusion

The mesh convergence study ensured the validity of using an element size of 0.1 mm for all soft tissue components and an element size of 0.4 mm for all bone components of the FE models.

6.4 Main Parameter Study

6.4.1 Introduction

This parameter study was conducted to investigate the effect of varying geometric parameters of the hip joint including hip joint clearance, cam protrusion, and cam width, with the aim of understanding the underlying damage mechanism of cam-type FAI for varying cam and hip joint geometry. These parameters were defined in Section 4.6.1, and the chosen parameter ranges (Table 6.1) were verified in the preliminary parameter study in Section 6.2 to lead to the main types of damage that commonly occur in cam-type FAI patients. The models used in the preliminary parameter study were used in this main study, but with more refined meshes based on the mesh convergence study conducted in Section 6.3 to ensure the accuracy of the results. Since the mechanism by which cam-type FAI causes damage such as cartilage delamination was verified to be caused by large shear strains by the *in vitro* experiments in Chapter 5, the shear strain results for the FE models in this study were analysed. Shear strain

results for all models are presented in the form of plots and quantitatively so that the locations of greatest shear strain could be identified and the development of shear strains over time, as a result of cam-type impingement, could also be analysed.

6.4.2 Methods

Mesh Generation, Assigning Boundary Conditions, and Defining Material Properties, Interactions, and Analysis Steps

Similar mesh generation, boundary conditions, material properties, interactions, and analysis steps were applied in this model as they were applied in the Internal Rotation at Flexion model in Stage 4 in Section 4.5.1 and the preliminary parameter study in Section 6.2.1. However, a more refined mesh was used with an element size of 0.1 mm for all soft tissue components and interfaces between soft tissue and bone and an element size of 0.4 mm for all bone components. These analysis formulations created an FE model which simulated the internal rotation of the femoral head during the activity of sitting when the hip was flexed at 40° (with a joint load of 0.1 N medially and 1.92 N anteriorly). In the presence of a cam deformity, these FE models, therefore, simulated cam-type FAI during the motion of internal rotation at flexion.

6.4.3 Results

Shear Strain

Logarithmic shear strain FE results for small hip joint clearance (Figures 6.11, 6.12) and large hip joint clearance (Figures 6.13, 6.14) are presented by a matrix of plots. Results for small clearance and small cam protrusion (Figure 6.11), small clearance and large cam protrusion (Figure 6.12), large clearance and small cam protrusion (Figure 6.13), and large clearance and large cam protrusion (Figure 6.14) are each presented by displaying the progression of internal rotation down each column, while across the columns, plots for models for different cam widths are given. The greatest negative shear strains (shear strains in the clockwise (CW) direction) are represented in blue, and the greatest positive shear strains (shear strains in the counter-clockwise (CCW) direction) are represented in red.

For the models with small hip joint clearance and small cam protrusion (Figure 6.11), the cam

deformity was able to enter into the hip joint. Initial entry of the cam deformity into the hip joint caused shear strains to occur near the labral-bone interface towards the outer portion of the acetabular rim. Further progression of the cam deformity into the hip joint caused shear strains to develop near the anterior acetabular articular cartilage-bone interface, which produced Hertzian-like contact between the acetabular and femoral articular cartilages. The cam deformity with a large width began entering into the hip joint with the smallest amount of internal rotation, causing shear strains near the cartilage-bone interface at 18° of internal rotation. Shear strains developed near the cartilage-bone interface for the model with a middle cam width at 41° of internal rotation, followed by the model with a small cam width at 45° of internal rotation. Clear separation of the anterior acetabular and femoral joint surfaces was apparent in the models with large and middle cam widths for internal rotations greater than 27° and 45° , respectively. This separation was allowed to occur due to the smooth entry of the cam deformity and displacement of the hip joint centre (HJC). The HJC of these models were constrained by a circle with a 1 mm radius, so the acetabular and femoral head surfaces were not allowed to separate by more than 1 mm. In contrast, the hip joint surfaces remained more or less in congruent contact with only slight separation in the model with a small cam width before the cam deformity was inhibited from entering any further into the hip joint.

For the models with small hip joint clearance and large cam protrusion (Figure 6.12), the cam deformity was not able to enter the hip joint beyond 36° of internal rotation. At 23° of internal rotation, a cam deformity with a large width began entering into the hip joint and caused shear strains to develop near the cartilage-bone and labral-bone interfaces near the chondrolabral junction, near the labral-bone interface towards the outer portion of the acetabular rim, and at the labral tip. Shear strains affecting these similar regions occurred for a middle cam width at 32° of internal rotation. For a small cam width, shear strains only developed near the cartilage-bone and labral-bone interfaces near the chondrolabral junction and at the labral tip and only developed in these regions at 36° of internal rotation. Clear separation of the anterior acetabular and femoral joint surfaces was apparent at 32° and 36° of internal rotation for a large and middle cam width, respectively.

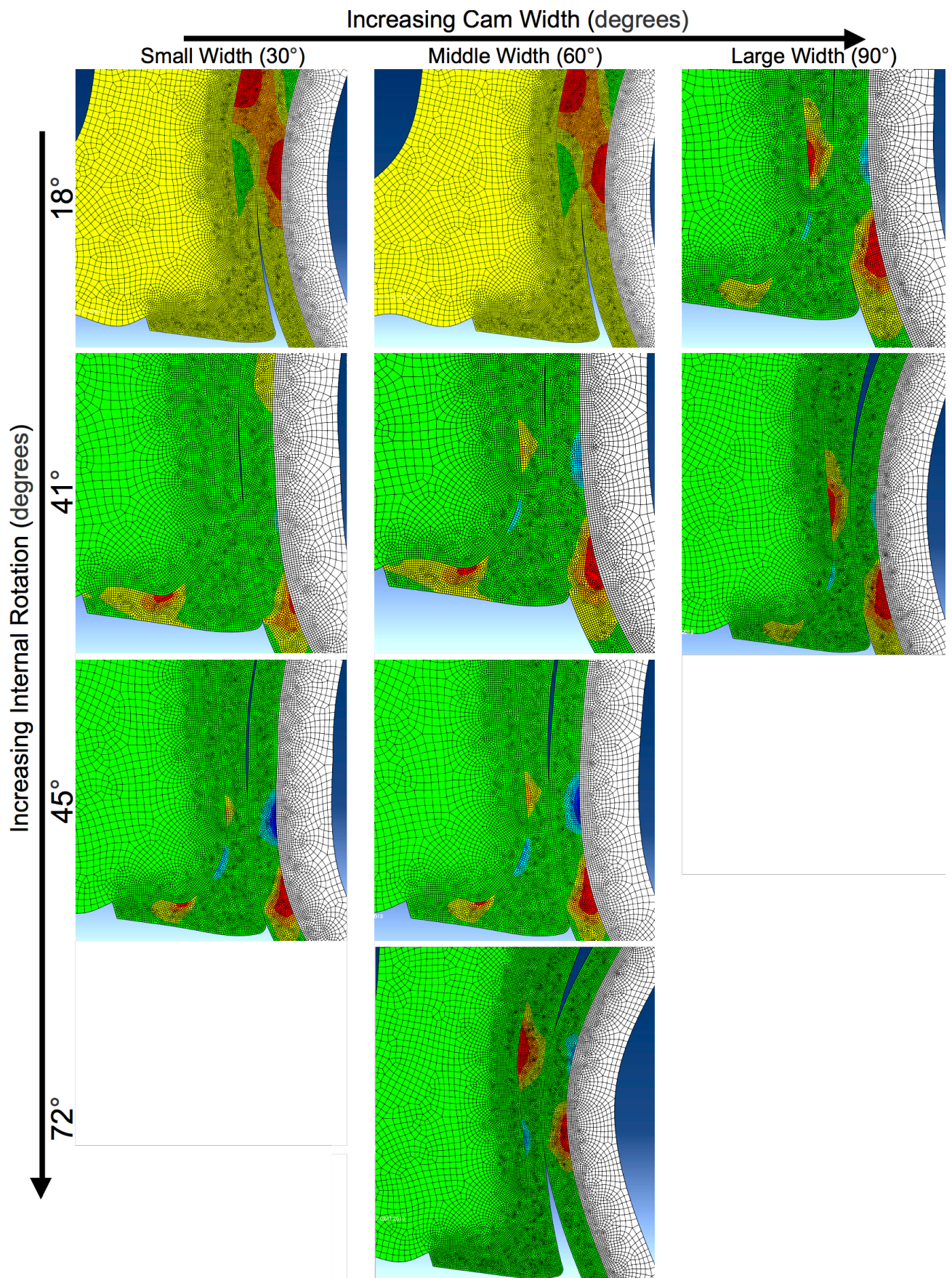


Figure 6.11: The logarithmic shear strain FE results are plotted for a hip with small clearance and small cam protrusion. The entry of the cam into the joint caused shear strains to develop near the labral-bone interface which progressed to the acetabular cartilage-bone interface.

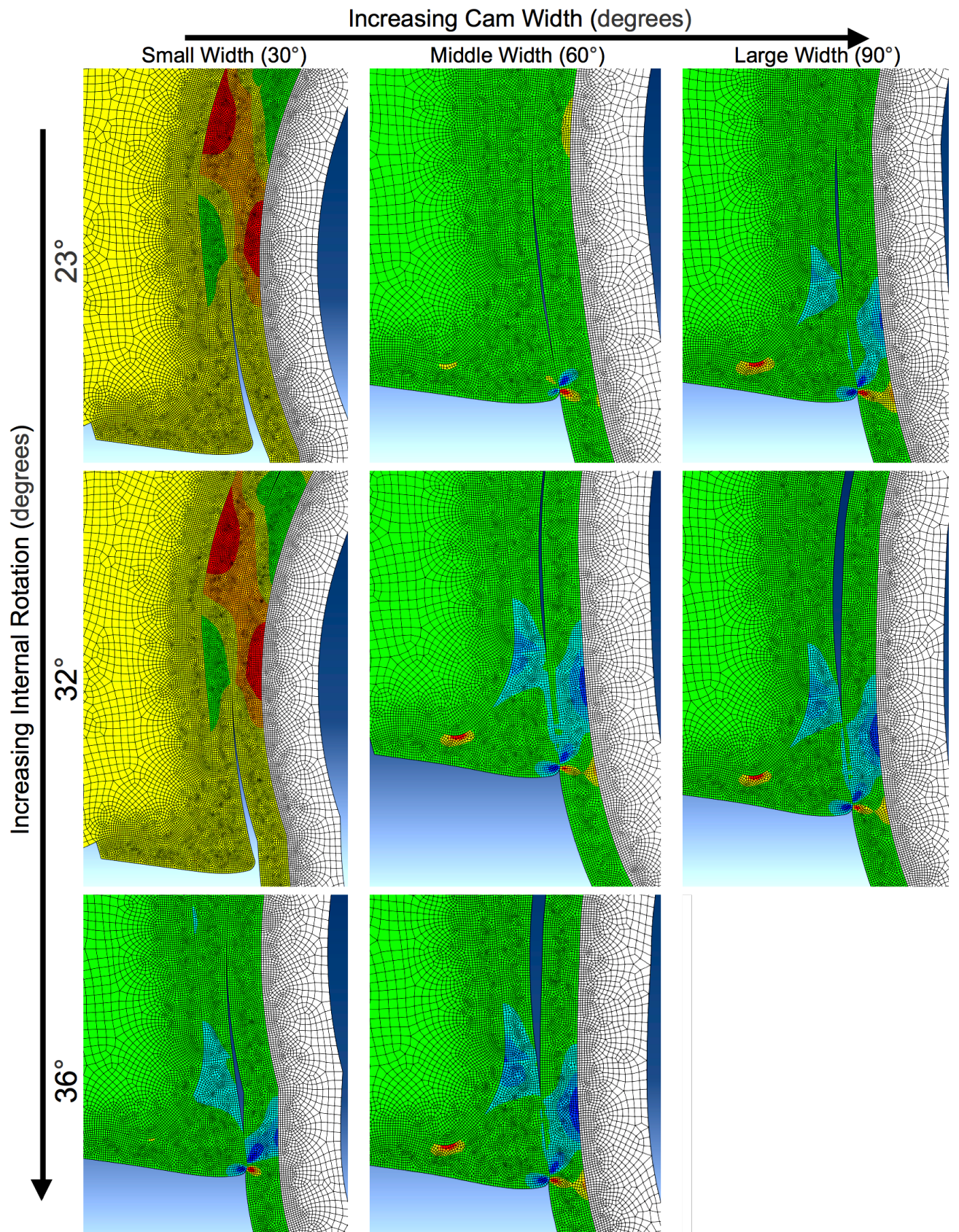


Figure 6.12: The logarithmic shear strain FE results are plotted for a hip with small clearance and large cam protrusion. The entry of the cam into the joint caused shear strains to develop near the labral tip and cartilage-bone and labral-bone interfaces near the chondrolabral junction, for all models, but only near the outer labral-bone interface for cams with middle and large width.

For the models with large hip joint clearance (Figures 6.13, 6.14), contact between the cam deformity and the acetabular joint surfaces did not occur earlier than 32° of internal rotation and occurred as late as 41° of internal rotation for a cam deformity with a small protrusion and width. Upon entry into the hip joint, the cam deformities with a small cam protrusion (Figure 6.13) caused shear strains to initially develop near the labral-bone interface towards the outer portion of the acetabular rim. Upon further entry into the hip joint, the model with a small cam protrusion and large cam width caused shear strains to develop near the acetabular cartilage-bone interface. In contrast, the models with a large cam protrusion (Figure 6.14) caused shear strains to develop near the labral-bone interface towards the outer portion of the acetabular rim and near the cartilage-bone and labral-bone interfaces in the region of the chondrolabral junction, simultaneously.

Parameter Study: Logarithmic Shear Strain Parametrised Cam Width, Large Hip Joint Clearance, Small Cam Protrusion

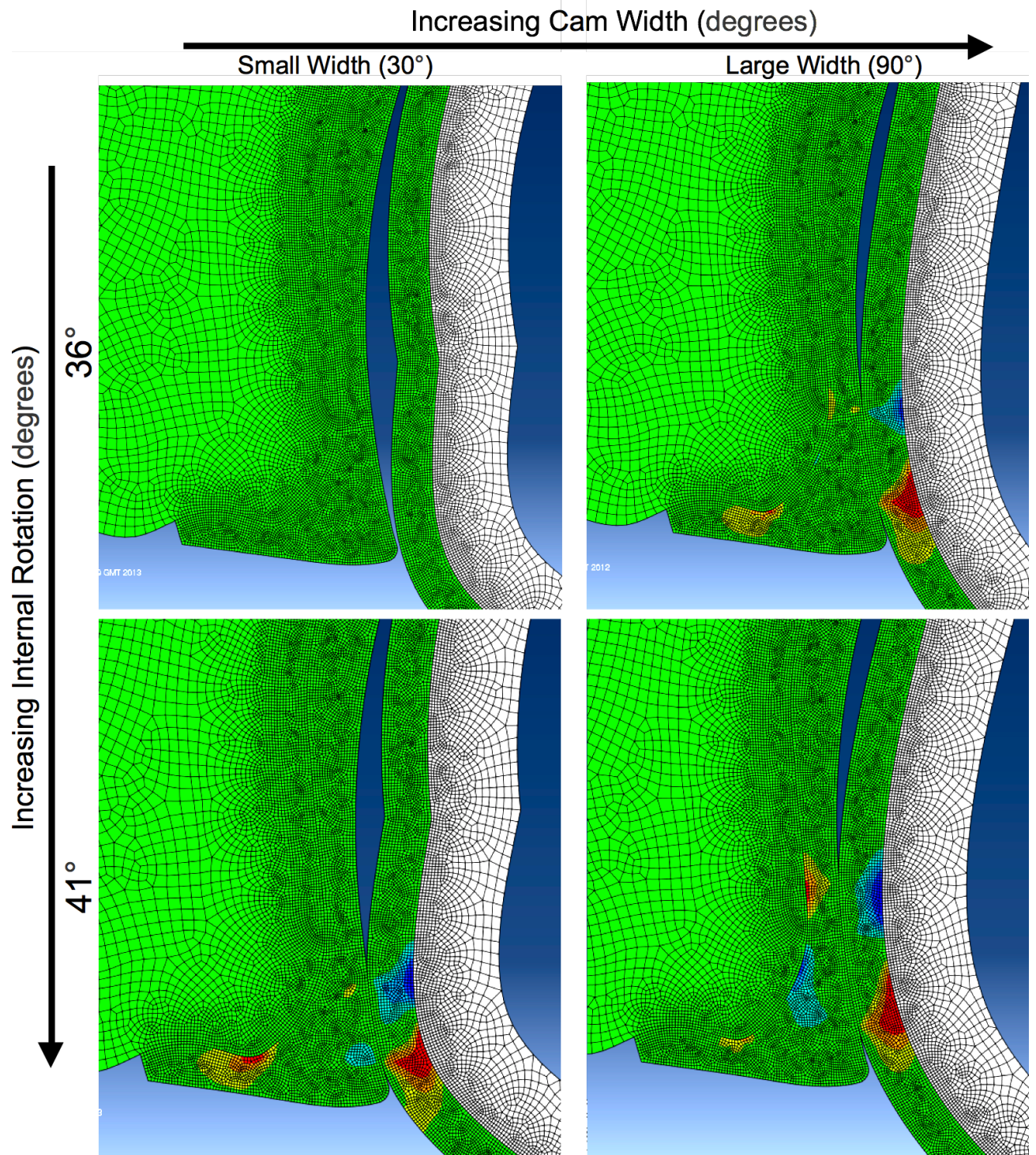


Figure 6.13: The logarithmic shear strain FE results are plotted for a hip with large clearance and small cam protrusion. The entering of the cam deformity into the hip joint caused shear strains to develop near the labral-bone junction, followed by the acetabular cartilage-bone interface.

Parameter Study: Logarithmic Shear Strain
 Parametrised Cam Width, Large Hip Joint Clearance, Large Cam Protrusion

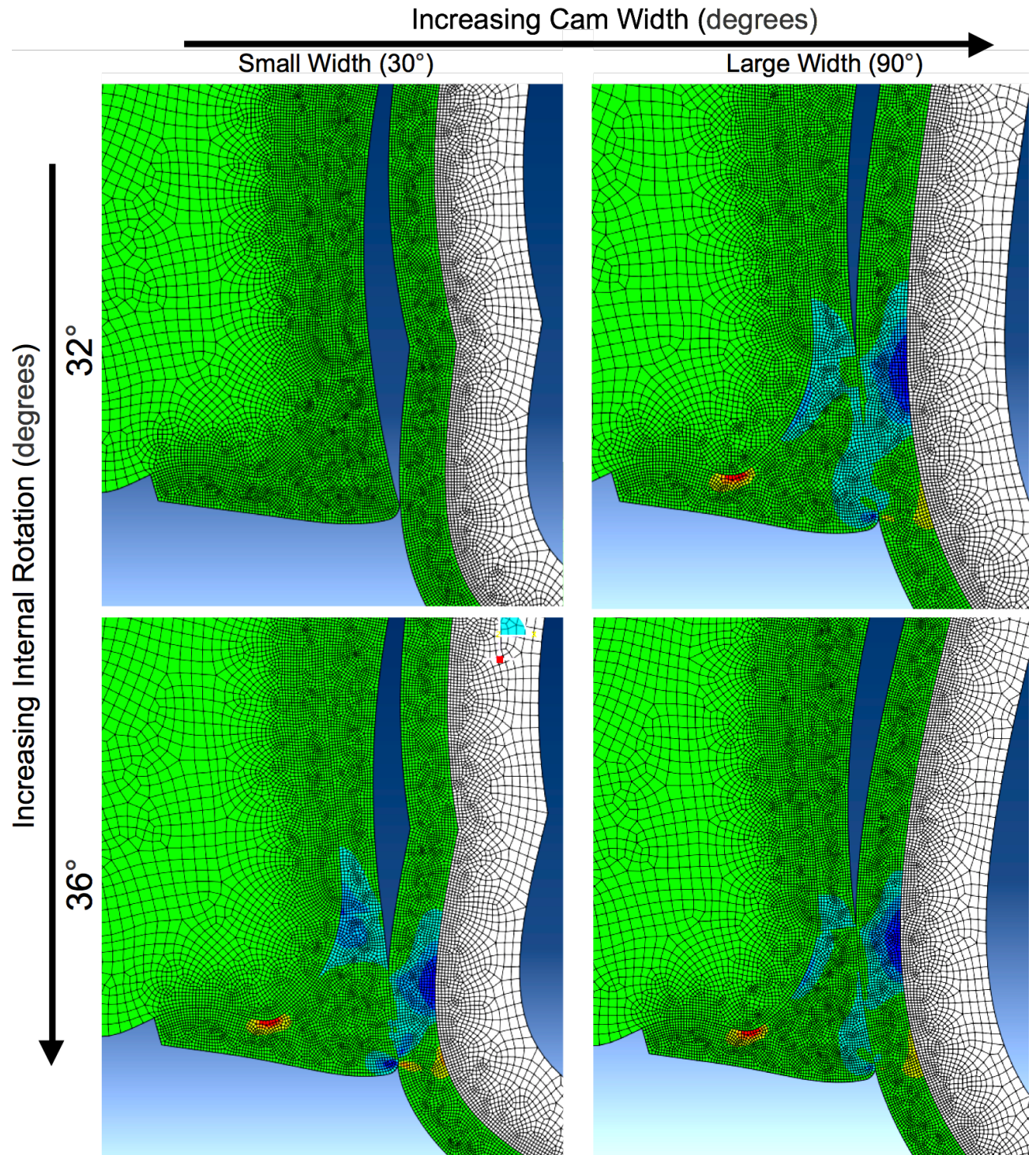


Figure 6.14: The logarithmic shear strain FE results are plotted for a hip with large clearance and large cam protrusion. The entering of the cam deformity into the hip joint caused shear strains to develop simultaneously near the labral-bone interface towards the outer portion of the acetabular rim and near the labral-bone and cartilage-bone interfaces near the chondrolabral junction.

Quantitative Results

Quantitative logarithmic shear strain results for small hip joint clearance models with small cam protrusion (Figure 6.15), large cam protrusion (Figure 6.16), and small and large cam protrusion combined (Figure 6.17) are presented. The shear strain results for large hip joint clearance models with small cam protrusion (Figure 6.18) and large cam protrusion (Figure 6.19) are also compared with results for small clearance. The results for varying cam widths for soft tissues including the anterior and posterior acetabular cartilage, the labrum, and femoral head articular cartilage are plotted for comparison.

For models with small cam protrusion (25% of small hip joint clearance) (Figure 6.15), a cam deformity with a larger cam width resulted in an earlier rise in shear strain in all of the soft tissues. A cam deformity with a smaller cam width resulted in the rapid increase of shear strains in the labrum and femoral cartilage, representing the sudden abutment of the labrum. Compared to a small cam width, the shear strains for the model with a middle cam width increased more gradually in all the soft tissues. The cam deformity with a middle cam was able to fully enter into the hip joint and resulted in the greatest amount of shear strain in the anterior acetabular articular cartilage. The negative shear strains (shear strains in the CW direction) that developed in the labrum were nearly twice as large as the positive shear strains (shear strains in the CCW direction) that developed. The negative shear strains affected the labral-bone interface near the chondrolabral junction, while the positive shear strains affected the labral-bone interface towards the acetabulum's outer rim. This meant that the labrum was dominated by shear strains that pushed it outwards from the hip joint in the region of the labral-bone interface closer to the chondrolabral junction rather than by shear strains that pulled it inwards into the hip joint in the region of the labral-bone junction towards the outer rim of the acetabulum. The positive and negative shear strains in the anterior acetabular and femoral articular cartilages were more or less similar in amplitude but slightly more negative when their maximum shear strains were reached at about 60° of internal rotation. This meant that, for the most part, their Hertzian-like contact resulted in the pushing of cartilage inwards into the hip joint and outwards from the hip joint in both cartilage components with neither effect dominating the other. However, when the femur was internally rotated 60° , the femoral articular cartilage is more dominated by shear strains that push it into the hip joint, while

the anterior acetabular articular cartilage was more dominated by shear strains that pushed it outwards from the hip joint.

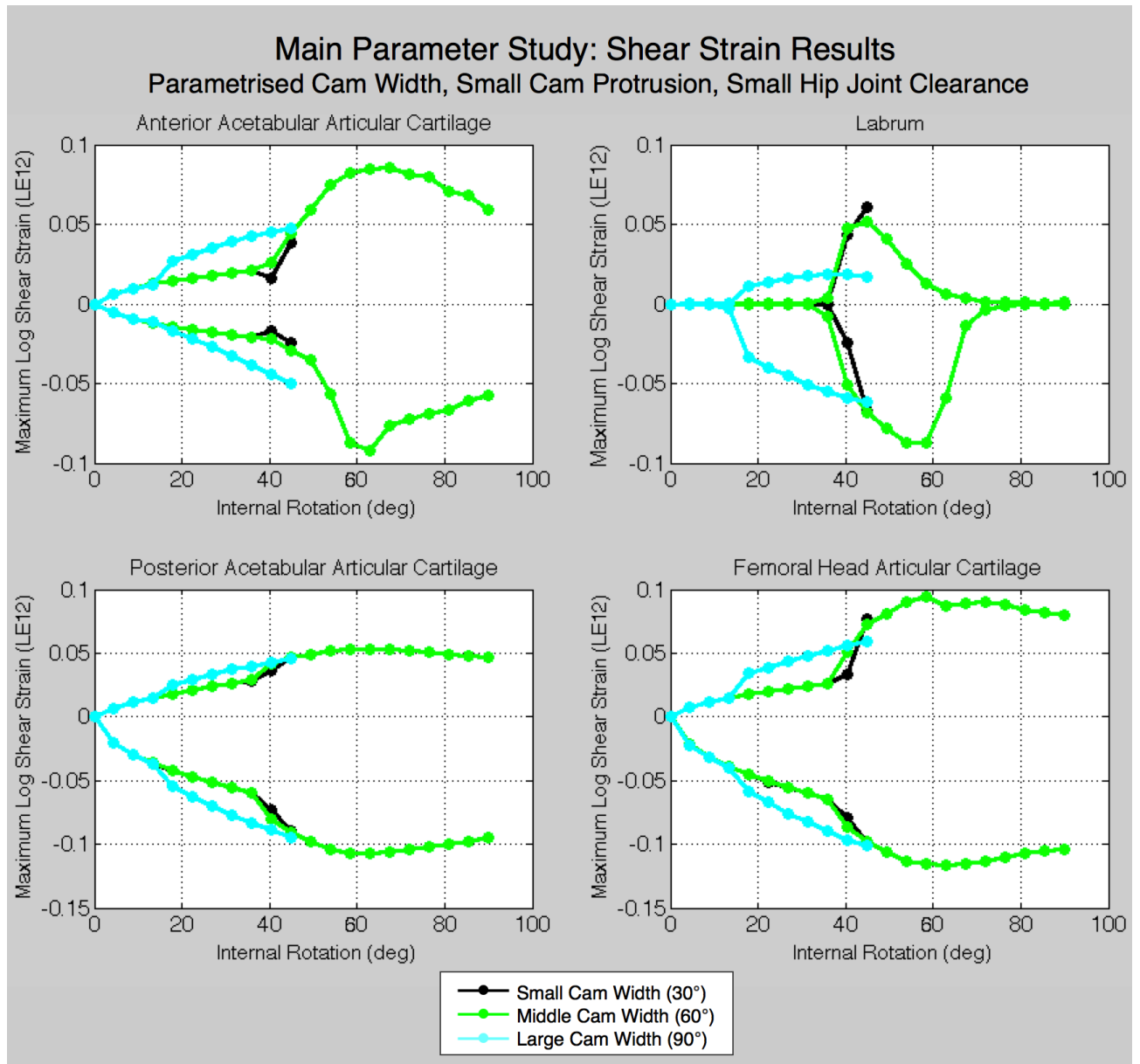


Figure 6.15: The quantitative shear strain FE results for the soft tissues in a hip joint with small clearance, small cam protrusion and parameterised cam width are presented. A wide cam led to more gradual increasing shear strain in the soft tissues. A middle cam width resulted in the greatest shear strains in the cartilage parts because it was allowed to fully enter into the hip joint. A small cam width led to an abrupt increase in shear strain in the anterior acetabular and femoral articular cartilages and the labrum due to the sudden abutment of the cam against the acetabulum.

For models with large cam protrusion (75% of small hip joint clearance) (Figure 6.16), a cam deformity with a larger cam width resulted in greater shear strains in the labrum and femoral articular cartilage compared to the anterior and posterior acetabular articular cartilages. The shear strains in the labrum and femoral articular cartilage reached beyond 0.1, while the shear strains remained less than 0.1 in the acetabular articular cartilages. A large cam width resulted in an earlier rise (less internal rotation) in shear strain in the anterior and posterior acetabular articular cartilages. Large and middle cam widths resulted in earlier rises in shear strain in the labrum and femoral articular cartilage. They also resulted in greater positive shear strains in the labrum. The positive shear strains occurred at the labral-bone interface, while the negative shear strains occurred at the labral tip. This meant that upon contact with the cam deformity, the labral tip was pushed outwards from the hip joint, while the labrum at the labral-bone interface was being pushed inwards into the hip joint. Also, shear strains in the labrum were more dominated by the pulling of the labrum inwards into the hip joint in the region of the labral-bone interface. The shear strains that developed in the femoral cartilage were also greater in the positive direction, meaning that for all cam widths the femoral cartilage was more dominated by shear strains that pushed it outwards from the hip joint. The shear strains that developed in the anterior acetabular articular cartilage were greater in the negative direction, meaning that the anterior acetabular articular cartilage was also more dominated by shear strains that pushed it outwards from the hip joint.

Comparing the shear strain results for small and large cam protrusion models showed that the maximum and minimum shear strains were generally greater in the anterior acetabular articular cartilage and labrum for the small and large cam protrusion models, respectively, as expected (Figure 6.17). It was also apparent that greater shear strains developed in the femoral head articular cartilage for models with a large cam protrusion. Additionally, a large cam protrusion led to an earlier rise in shear strain in all the soft tissues compared to a small cam protrusion.

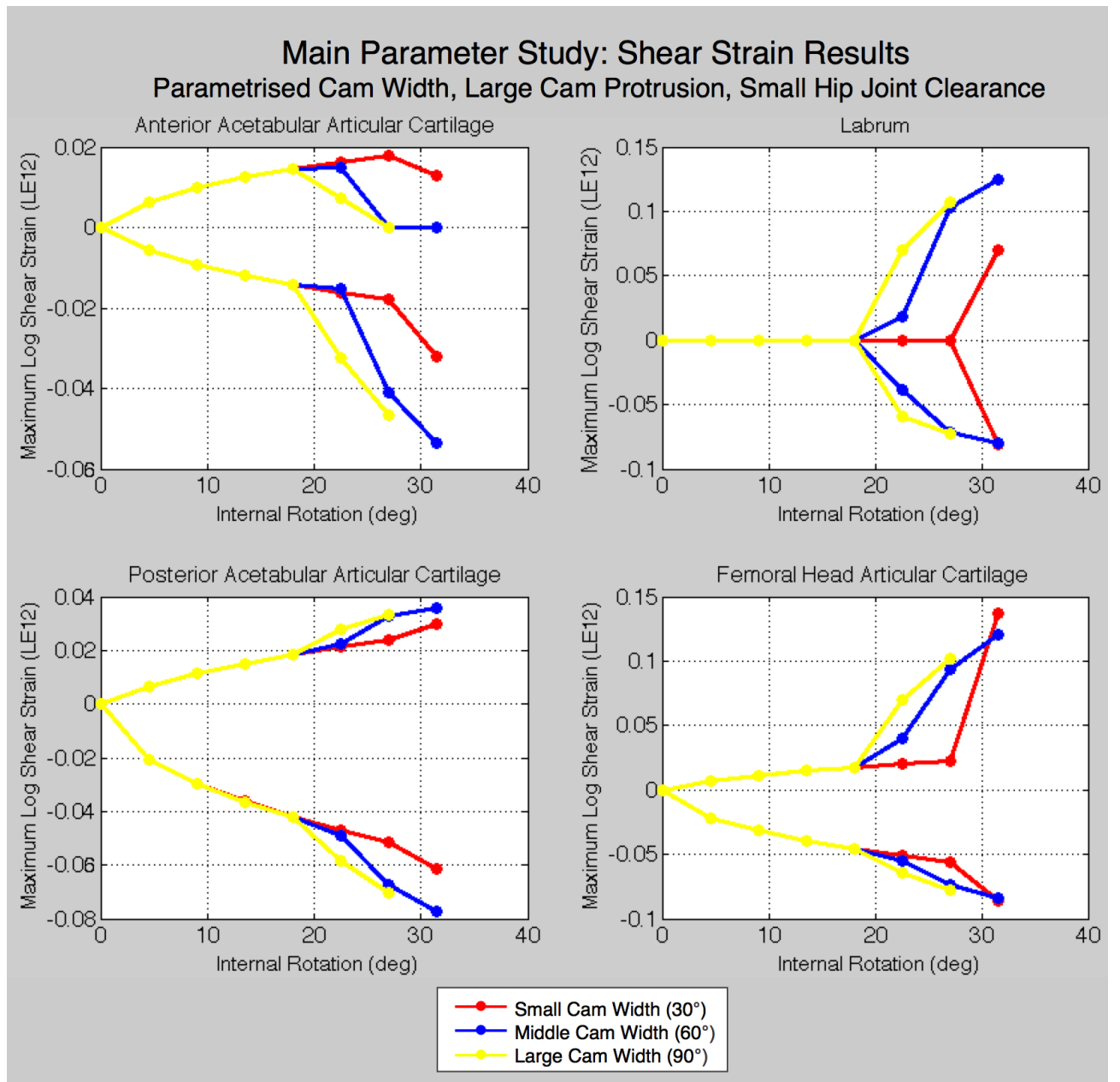


Figure 6.16: The quantitative shear strain FE results for the soft tissues in a hip joint with small clearance, large cam protrusion and parameterised cam width are presented. A wider cam led to the earlier rise in shear strain in the soft tissues. The shear strains that developed in the femoral cartilage were greater in the positive direction, meaning that the femoral cartilage was more dominated by shear strains that pushed it outwards from the hip joint. The shear strains that developed in the anterior acetabular articular cartilage were greater in the negative direction, meaning that the anterior acetabular articular cartilage was also more dominated by shear strains that pushed it outwards from the hip joint.

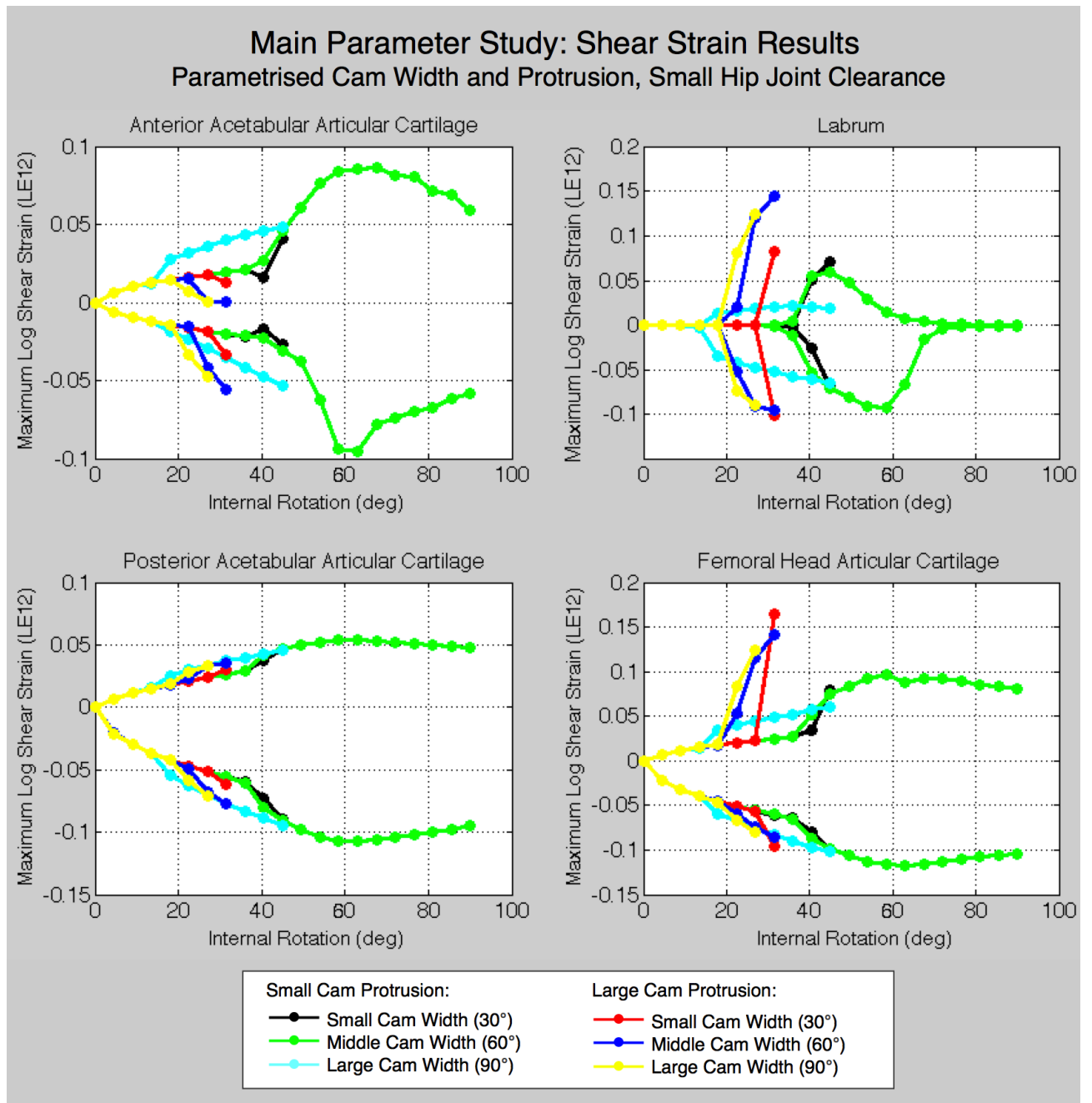


Figure 6.17: The quantitative shear strain FE results for small and large cam protrusion models with small clearance are presented. The maximum and minimum shear strains were generally greater in the anterior acetabular articular cartilage and labrum for the small and large cam protrusion models, respectively. The femoral articular cartilage also showed greater shear strain results for large cam protrusion models.

Quantitative logarithmic shear strain results for models with large hip joint clearance are compared with the results for models with small hip joint clearance (Figures 6.18, 6.19). For the models with large hip joint clearance and a small cam protrusion proportional to the large clearance (25% of clearance) (Figure 6.18), reduced shear strains in all the soft tissues except for the anterior acetabular articular cartilage for internal rotations above 40° resulted. A small cam width resulted in similar results to a hip with small clearance, small protrusion, and small

width, however, the shear strains were reduced, except for the positive shear strain in the anterior acetabular articular cartilage for internal rotations above 40°. A large cam width resulted in shear strains that were similar to those that developed in models with small clearance, small cam protrusion, and middle cam width (Figure 6.15) where the cam fully entered the hip joint caused large shear strains to develop in all soft tissues.

For the models with large hip joint clearance and a large cam protrusion proportional to the large clearance (75% of clearance) (Figure 6.19), a small cam width resulted in reduced shear strains in the posterior acetabular and femoral articular cartilages. It also produced reduced shear strains in the anterior acetabular articular cartilage, but only for internal rotations under 32°. The large clearance model was able to run past 32° internal rotation, which allowed for much greater shear strains to develop in the anterior acetabular articular cartilage and labrum. It also had similar results to the model with small hip joint clearance, large protrusion, and large width, but with delayed results. A large cam width resulted in shear strains similar to those that developed in models with small clearance, large cam protrusion, and middle cam width (Figure 6.16), with the exception that the rise in shear strains was delayed. Overall, the models with large clearance often had similar results to the models with small clearance for similar cam protrusion and width parameters, but with slight variations. These variations are summarised in Table 6.2 below.

Comparison of Large Clearance (LC) Model Results to those for Small Clearance (SC) Models with Similar Cam Protrusion and Cam Width Parameters

	Small Cam Protrusion (25% Hip Joint Clearance)	Large Cam Protrusion (75% Hip Joint Clearance)
Small Cam Width (30°)	LC similar to SC, but with reduced shear strains in all tissues, except AAAC for internal rotations > 40°	LC similar to SC's large cam width model, but with delayed results
Large Cam Width (90°)	LC similar to SL's middle cam width model	LC similar to SC's middle cam width model, but with slightly reduced shear strains and delayed results

Table 6.2: The results for large hip joint clearance (LC) models are compared to small hip joint clearance (SC) models with similar cam protrusion and cam width parameters. The anterior acetabular articular cartilage is abbreviated as AAAC.

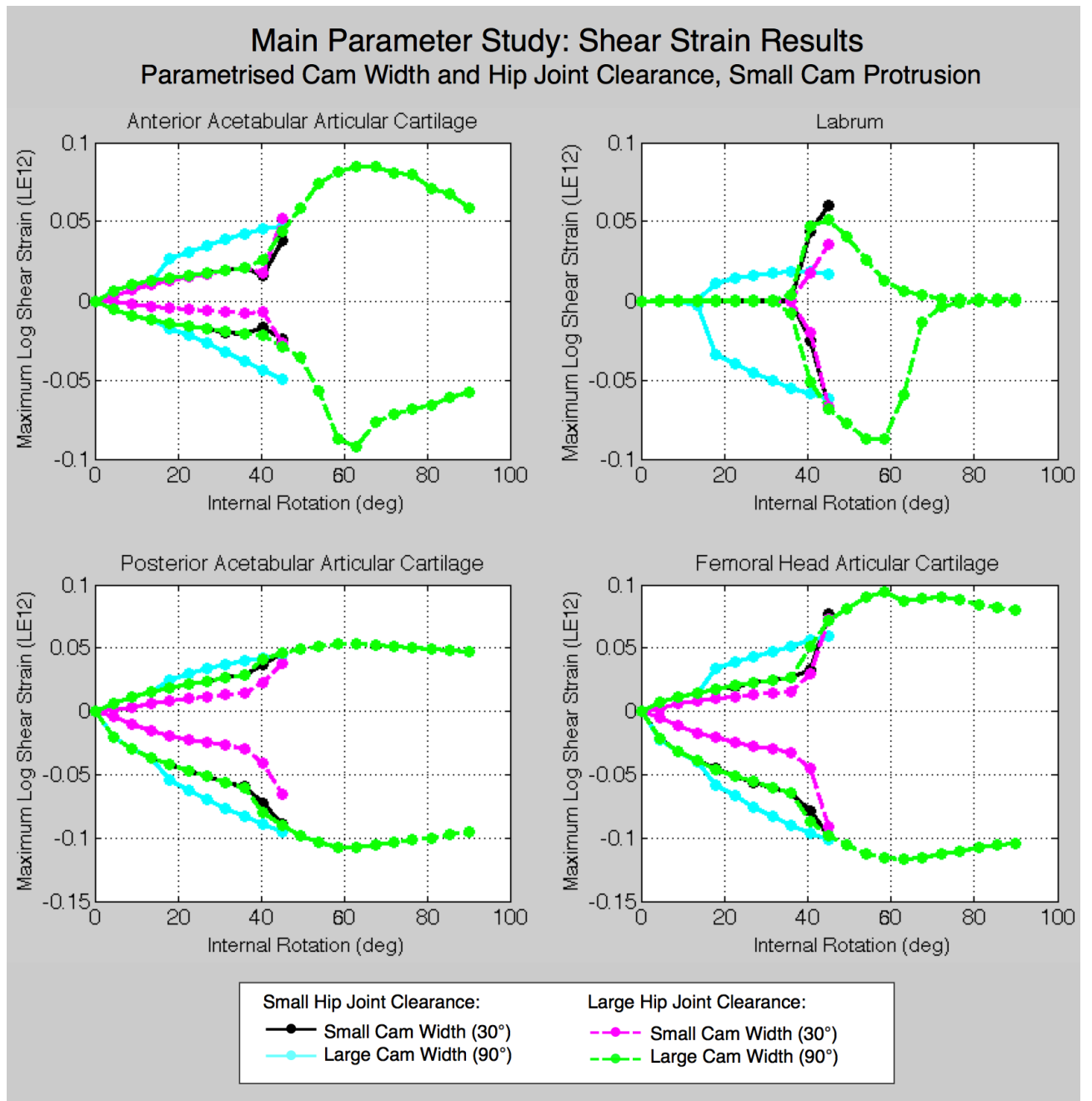


Figure 6.18: The quantitative shear strain results for small and large hip joint clearance models with small cam protrusion are presented. A small cam width in the large clearance models resulted in similar results to a hip with small clearance, small protrusion, and small width, however, the shear strains were reduced, except for the positive shear strain in the anterior acetabular articular cartilage for internal rotations above 40°. A large cam width in the large clearance models resulted in shear strains that were similar to those that developed in models with small clearance, small cam protrusion, and middle cam width (Figure 6.15).

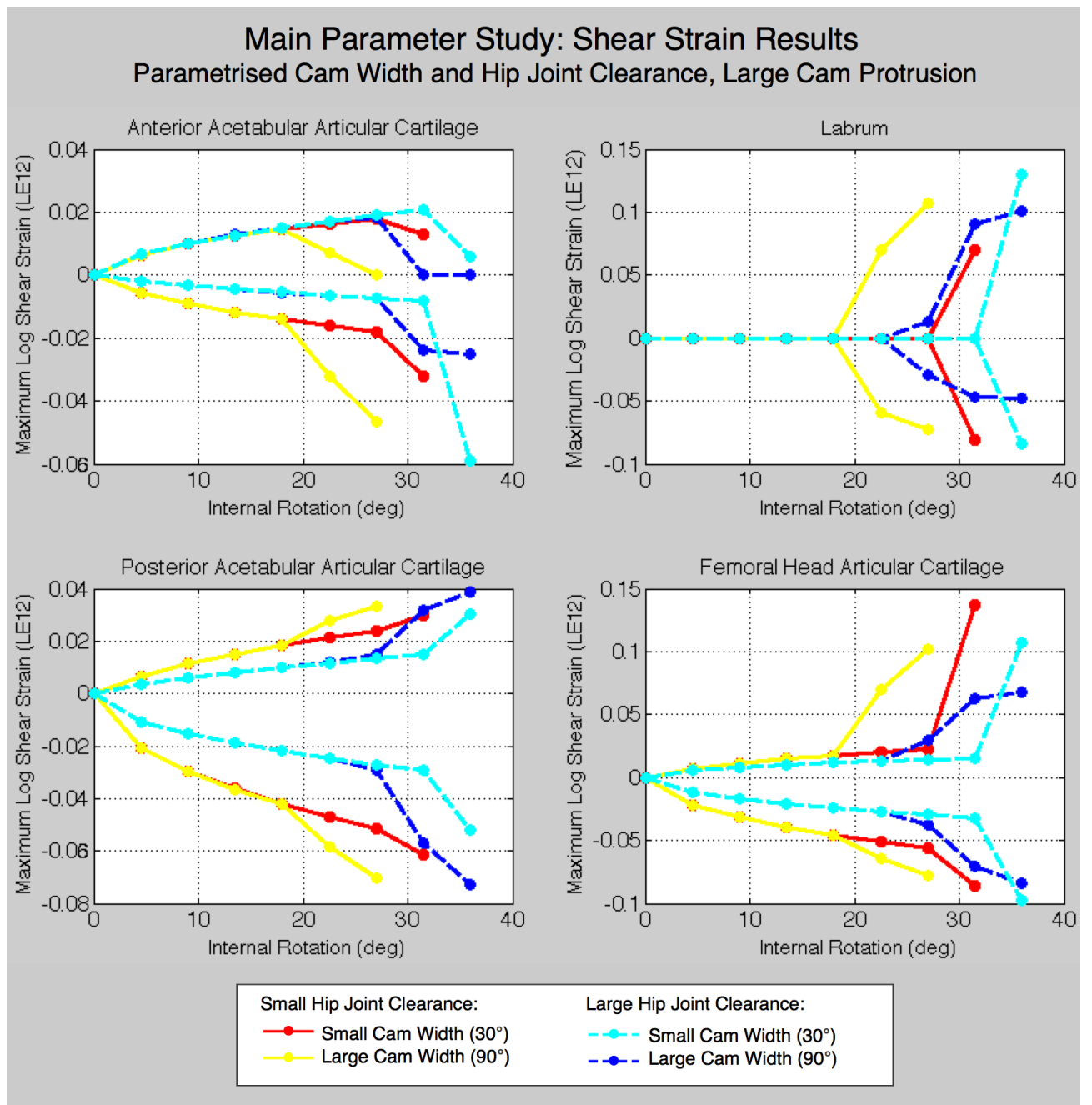


Figure 6.19: The quantitative shear strain FE results for small and large hip joint clearance models with large cam protrusion are presented. A small cam width in the large clearance models had similar results to the model with small hip joint clearance, large protrusion, and large width, but with delayed results. A large cam width resulted in shear strains similar to those that developed in models with small clearance, large cam protrusion, and middle cam width (Figure 6.16), but with delayed results.

6.4.4 Discussion

The shear strains that developed in the femoral head articular cartilage were comparable to the shear strains that developed in the anterior acetabular articular cartilage for the models with small hip joint clearance and small cam protrusion. However, for cam-type FAI patients, the

acetabular soft tissues are commonly damaged, while the femoral head articular cartilage is very rarely damaged [20, 21, 22, 23, 24, 50]. It is typically the anterior acetabular articular cartilage that is damaged through delamination, while the femoral head articular cartilage is thought to become damaged once delaminated acetabular cartilage tears off, leaving exposed bone to be in contact with the femoral cartilage [20]. This leads to the hypothesis that damaging cam deformities might not be covered by cartilage or at least not normal cartilage; if they consisted of exposed bone or very thin cartilage, they would be more destructive to the cartilage on the acetabular side and, therefore, lead to the pathology clinically seen in cam-type FAI patients.

The model with small hip joint clearance, small cam protrusion, and middle cam width was the only model where the cam deformity was able to fully enter into the hip joint and go further into the joint cavity. The cam deformity with a small width was not able to enter fully because it abutted the labrum too suddenly, inhibiting the hip joint surfaces in the anterior superior region to separate and provide clearance for the cam deformity, whereas the cam deformity with a large width was probably too wide, so its larger volume was not able to enter further into the hip joint. However, it is possible that if the cam with a large width was made slightly narrower or if it had a slightly smaller protrusion, it could have possibly been able to fully enter the joint cavity as well. Additionally, if the HJC was allowed to displace more than 1 mm, it is possible that greater clearance could have been made for the wide cam to enter fully. These results suggest that the volume of the cam structure and allowed HJC translation may play important roles in the cam-type FAI mechanism.

To depict the cam-type FAI mechanism of small and large cam protrusion in hips with small clearance, the shear strain directions discussed in Section 6.4.3 are depicted with black arrows, while red arrows were used to indicate the greatest shear strains that developed in the soft tissues, and a red circle was used to indicate the dominant direction of shear strain in a tissue (Figure 6.20).

For models with small clearance and small cam protrusion, the cam was able to enter further into the hip joint compared with large cam protrusion models, and the labrum was more dominated by shear strains that pushed it outwards from the hip joint in the region of the labral-bone interface closer to the chondrolabral junction. Also, the femoral articular cartilage was more dominated by shear strains that pushed it into the hip joint, while the anterior

acetabular articular cartilage was more dominated by shear strains that pushed it outwards from the hip joint. In contrast, for models with small clearance and large cam protrusion, the cam abutted against the labrum, pushing the labral tip outwards from the hip joint which pushed the labrum at the labral-bone interface (near the acetabulum's outer rim) inwards into the hip joint. Also, the anterior acetabular and femoral cartilages were more dominated by shear strains that pushed them outwards from the hip joint.

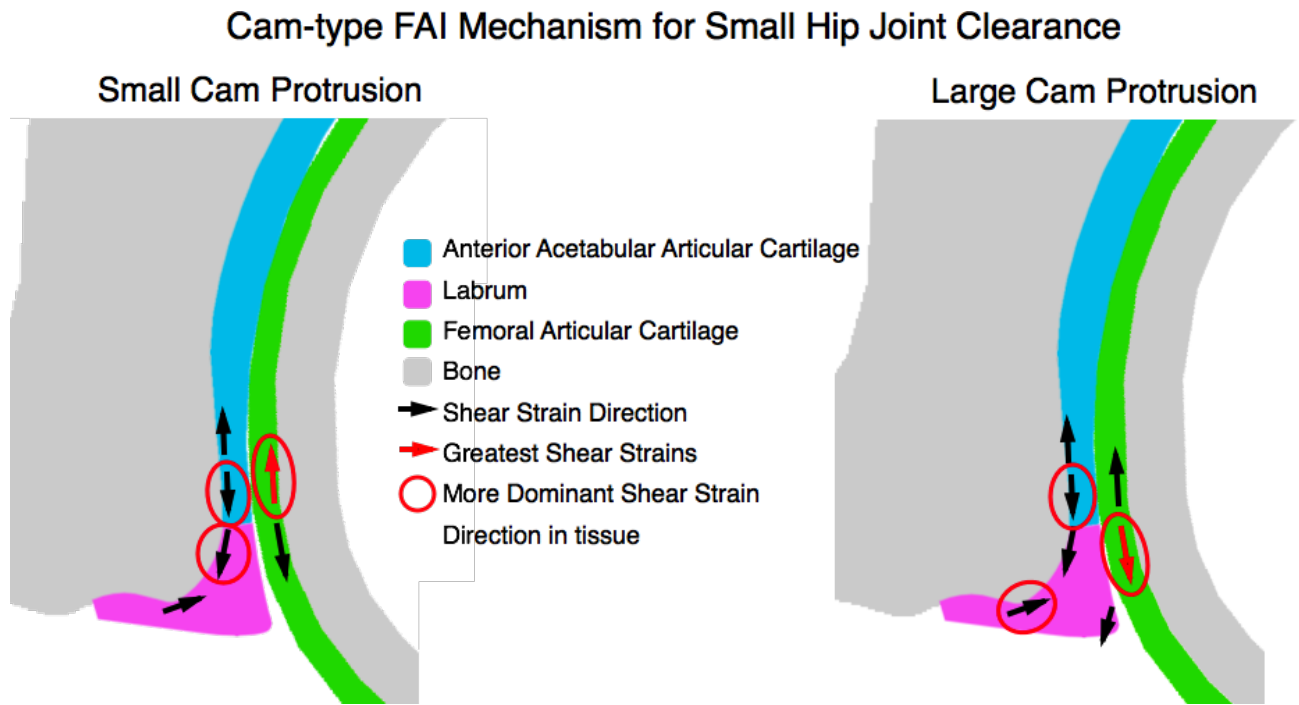


Figure 6.20: The cam-type FAI mechanisms for hips with small clearance and small and large cam protrusion are presented. Positive and negative shear strain directions for soft tissues in the anterior part of the hip joint are depicted with black arrows, while the greatest shear strains are depicted with red arrows. The dominant direction of shear strain in a tissue is indicated with a red circle. Cam-type FAI for a small cam protrusion was generally dominated by the pushing of the femoral articular cartilage into the hip joint, the anterior acetabular articular cartilage outwards from the hip joint, and the labrum (near the chondrolabral junction) outwards from the joint. For a large cam protrusion, cam-type FAI was generally dominated by the pushing of the anterior acetabular and femoral articular cartilages outwards from the hip joint and the labrum (near the outer portion of the acetabular rim) inwards into the joint.

In general, having a larger hip joint clearance resulted in reduced shear strains for smaller ROMs of internal rotation. These results support the hypothesis made in Section 3.5 that having a large hip joint clearance may give clearance for a cam deformity and prevent cam-type FAI. But for larger ROMs of internal rotation, increased shear strains occurred, usually because the cam was able to enter further into the hip joint. This suggests that the hip joint clearance may be a predictive factor in the development of cam-type FAI which is dependent on the

activity level of a subject. Also, the results for large hip joint clearance models often presented results that were similar for a model with small hip joint clearance with some similar and some different cam parameters. This shows that having a large hip joint clearance may produce similar mechanisms to hips with small hip joint clearance, given that the cam protrusions in these hips are proportional to the hip joint clearances in these models. For instance, for small (25% hip joint clearance) and large cam protrusions (75% hip joint clearance), a large hip joint clearance model with a large cam width resulted in similar results to models with small hip joint clearance and middle cam widths. A small cam width for a large hip joint clearance and large cam protrusion model resulted in similar results to a model with small hip joint clearance and a large cam width, but with delayed results.

6.5 Conclusion

This chapter described the use of a 2D model to simulate the cam-type impingement mechanism and the parameterisation of the hip joint geometry. A mesh convergence study allowed the selection of appropriate element sizes for the FE models in the main study, ensuring accurate shear strain results for the acetabular soft tissues.

The results of the parameter studies in this chapter demonstrated how varying hip joint clearance, cam protrusion, and cam width can lead to variations in the cam-type FAI mechanism and the initiation of different types of damage in different locations. The geometric parameterisation of hip joint clearance (ranging between 2.88 to 3.88 mm), cam protrusion (ranging between 25 to 75% of the hip joint clearance), and cam width (ranging between 30 to 90°) proved to be able to cause large shear strains to develop in the acetabular articular cartilage and labrum which may lead to cartilage delamination and labral tearing, respectively.

In general, for small hip joint clearance, a small cam protrusion allowed the entry of the cam deformity into the hip joint so that large shear strains developed in the acetabular articular cartilage which could cause delamination, while a large cam protrusion abutted against the labrum so that large shear strains developed in the labrum which could cause labral tears to occur at the labral tip or near the labral-bone interface. Wider cam deformities allowed for the gradual and slight displacement of the HJC, allowing the hip surfaces to separate, whereas a narrow cam deformity resulted in the sudden abutment of the cam with the labrum.

Cam-type FAI for a small cam protrusion was generally dominated by the pushing of the femoral articular cartilage into the hip joint, the anterior acetabular articular cartilage outwards from the hip joint, and the labrum (near the chondrolabral junction) outwards from the hip joint. For a large cam protrusion, cam-type FAI was generally dominated by the pushing of the anterior acetabular and femoral articular cartilages outwards from the hip joint, the labral tip outwards from the hip joint, and the labrum (near the acetabulum's outer rim) inwards into the joint.

Furthermore, for a hip with small clearance and a small-protruding cam deformity that primarily affects the cartilage, a wider cam deformity caused damage to initiate further in the hip joint and at smaller ranges of internal rotation. In contrast, for a large-protruding cam that primarily affects the labrum, a narrower cam deformity caused damage to initiate suddenly at the tip of the labrum upon abutment, whereas a wider cam deformity caused damage to initiate near the labral-bone junction.

For a hip with large clearance and small (25% hip joint clearance) or large cam protrusions (75% hip joint clearance), a large cam width resulted in similar results to models with small hip joint clearance and middle cam widths. A small cam width for a large hip joint clearance and large cam protrusion model resulted in similar results to a model with small hip joint clearance and a large cam width, but with delayed results. Therefore, having a large hip joint clearance may produce similar mechanisms to hips with small hip joint clearance, given that the cam protrusion in these hips are proportional to the hip joint clearance in these models.

Overall, the parameter studies investigating cam protrusion, cam width, and hip joint clearance and their effect on damage initiation or shear strain in soft tissues in the hip joint gave rise to deeper understanding of the cam-type impingement mechanism and have given rise to other important parameters such as cam volume and allowed HJC translation. However, further investigation of these geometric parameters along with new parameters would indeed be valuable to better characterise cam deformities by how damaging they might be and how they might initiate and progress damage.

Chapter 7

2D FE Model Simulating Cam-type FAI: Investigating Labral Anatomy and Surgery

7.1 Introduction

Cam-type FAI is dependent on geometry and motion. Variations in geometry due to parameters such as hip joint clearance and cam shape and size were studied in Chapter 6. These parameters showed that variations in hip joint and cam anatomy can lead to the main types of damage involved in cam-type FAI:

1. Damage to the anterior acetabular articular cartilage-bone interface leading to cartilage delamination [20, 21, 22, 23]
2. Damage to the labrum leading to tears within the structure or at its attachment site to the acetabular rim [20, 21, 22, 23, 24, 50]
3. Damage in the region of the chondrolabral junction affecting the neighboring anterior acetabular articular cartilage and labrum [20, 125, 164]

In this chapter, other conditions including labral anatomy and reshaping surgery were investigated. Labral anatomy can vary in shape and size, and can vary where it attaches to the acetabular rim. Also, the labrum can possibly be connected to articular cartilage or have sub-labral recessions at the chondrolabral junction leaving the labrum and articular cartilage unconnected [49, 77, 78]. Reshaping surgery can dramatically change the anatomy of the hip joint. During reshaping surgery, an orthopedic surgeon makes an experienced judgement as to

how to remove parts of bone which ideally improve the clearance in the hip joint, relieving the hip from any impingement at the extremes of ROM. The cut that is usually made to remove parts of bone is made towards the HJC on the femoral head and usually goes approximately 5 mm deeper than the spherical portion of the femoral head, rather than simply removing the protrusion of the cam deformity so it lies flush with the rest of the femoral head surface.

Since labral anatomy and reshaping surgery affect the geometry of the hip joint, they may also possibly affect the cam-type FAI mechanism as well. They were, therefore, investigated in this chapter with the following aims:

1. To study how variations in labral anatomy could affect the damage mechanism of cam-type FAI and the development of different types of damage such as those listed above
2. To study the affect of reshaping surgery and to assess the ability of such operations to treat the mechanism by which cam-type FAI causes damage within the hip joint

7.2 Labrum Parameter Study

The behavior of three variations in labral anatomy were compared as a result of internal rotation at flexion. The labral anatomy variations were studied in a hip joint model with a small hip joint clearance, large cam protrusion, and large cam width which was studied in Chapter 6 and was able to produce large shear strains at the labral tip and in the region of the chondrolabral junction, affecting the neighboring anterior acetabular articular cartilage and labrum (Figure 6.12).

7.2.1 Methods

Labral Anatomy Variations

Due to a lack of detailed information as to how the labrum may vary in the hip joint, MRA images of hip joints (proton-density sequence with fat saturation) were studied in ten cam-type FAI patients and ten subjects without cam-type FAI. After examining the labral anatomy of actual subjects, three variations of labral anatomy were chosen to be investigated. The original labral anatomy used in Chapter 6 was analysed, along with two variations of the original labral anatomy, both of which consisted of a more elongated shape, one with a connected chondrolabral

junction and one with an unconnected chondrolabral junction (Figure 7.1). Elongated labral anatomy was created by extending the anatomy of the original labrum used in Chapter 6 along the articular hip joint surface, so as to deepen the acetabular socket. The altered elongated labrums were also modelled to have a more outwardly directed attachment site to the acetabular rim. This meant that their chondrolabral junction was located more outwards from the hip joint towards the rim of the acetabulum.

Curved corners were modelled in the labrum and acetabular articular cartilage at the chondrolabral junction of an elongated unconnected labrum to prevent point loads from occurring. The corners of the labrum and acetabular articular cartilage at the chondrolabral junction of an elongated connected labrum were modelled to be curved so that it could be directly compared with the unconnected model. This chondrolabral junction anatomy differed from the anatomy of the chondrolabral junction in the model with the original labrum where the articulating surfaces of the labrum and anterior acetabular articular cartilage lie flush with each other.

Labrum Parameter Study Geometry

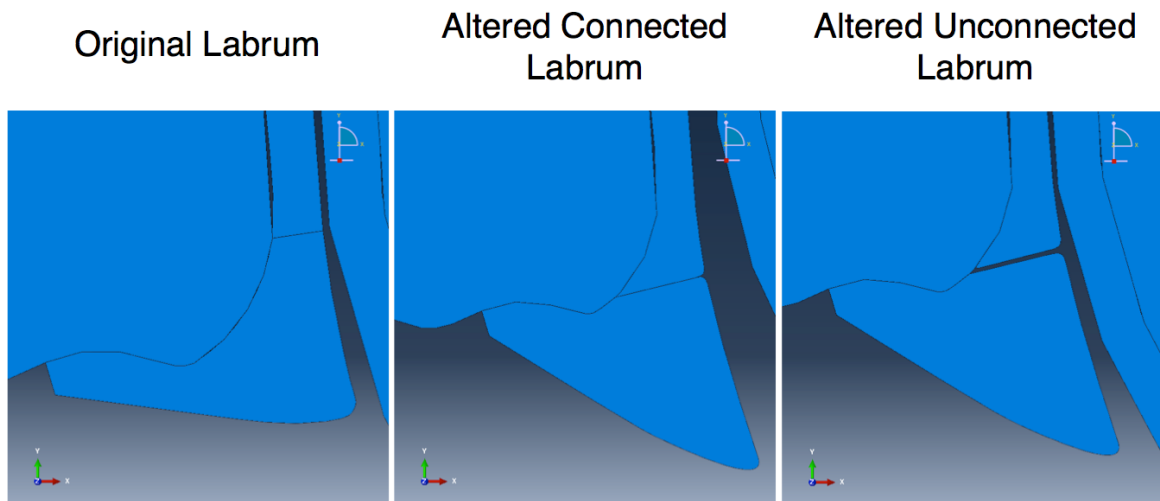


Figure 7.1: The variations of labral anatomy studied in the labrum parameter study for strain behavior with respect to cam-type FAI are displayed.

Geometry, Mesh Generation, Assigning Boundary Conditions, and Defining Material Properties, Interactions, and Analysis Steps

The geometry of the hip joint model with a small hip joint clearance, a large cam protrusion, and a large cam width that was developed in Chapter 6 was used. Similar mesh generation, boundary conditions, material properties, interactions, and analysis steps were applied in the

FE models as they were applied in the Internal Rotation at Flexion model in Stage 4 in Section 4.5.1. In the model with an unconnected chondrolabral junction, a surface-to-surface contact interaction modelling frictionless behavior (penalty contact with tangential friction of 0) was applied the surfaces of the labrum and acetabular articular cartilage at the chondrolabral junction. This differed from the models with a connected chondrolabral junction where tie constraints were applied to the surfaces of the labrum and acetabular articular cartilage at the chondrolabral junction.

These analysis formulations created a model which simulated the internal rotation of the femoral head during the activity of sitting when the hip was flexed at 40° (with a joint load of 0.1 N medially and 1.92 N anteriorly). In the presence of a cam deformity, these FE models, therefore, simulated cam-type FAI during the motion of internal rotation at flexion.

7.2.2 Results

Shear Strain

Logarithmic shear strain FE results for the labral parameter study for a model with small clearance and large cam protrusion and width are presented by displaying the progression of internal rotation down each column, while across the columns, the models with different labral anatomy are given (Figure 7.2). The greatest negative shear strains (shear strains in the CW direction) are represented in blue, and the greatest positive shear strains (shear strains in the CCW direction) are represented in red.

In all the models, the HJC gradually displaced, allowing the hip joint surfaces to separate upon the entering of the cam deformity. The HJCs of these models were constrained by a circle with a 1 mm radius, so the acetabular and femoral head surfaces did not separate by a displacement of more than 1 mm. Also, the entry of the cam deformity into the hip joint caused large positive shear strains (CCW direction) to develop in the labrum near the labral-bone junction towards the acetabulum's outer rim, which meant that the the labrum was pushed inwards into the hip joint in this region.

An elongated labrum resulted in earlier contact (less internal rotation) with the cam deformity compared to the original labrum. Rather than abutting the tip of the labrum (as in the original

labrum model), the cam deformity abutted the mid-section of the elongated labrums' articulating surfaces, causing greater shear strains to occur in this region. For the original labrum and elongated connected labrum models, large shear strains developed in the anterior acetabular articular cartilage near the cartilage-bone interface close to the chondrolabral junction; such shear strains did not occur in the elongated unconnected labrum model.

More complicated shear strain patterns were produced in the altered labrum models near the chondrolabral junction. For the elongated connected labrum, positive shear strain developed at the corner of the anterior acetabular articular cartilage (close to the chondrolabral junction and the articulating surface), meaning that it was pushed inwards into the hip joint (CCW direction) in this region. For the elongated unconnected labrum, the cam deformity was able to enter slightly further into the joint as the anterior acetabular articular cartilage and labrum separated at the chondrolabral junction. This produced negative shear strains at the corner of the anterior acetabular articular cartilage (close to the chondrolabral junction and the articulating surface), while positive shear strain developed at the corner of the labrum close to the chondrolabral junction. This shear strain pattern showed that these regions were pushed inwards into the chondrolabral junction. The femoral articular cartilage in contact with the chondrolabral junction area displayed positive and negative shear strains which showed that it was being pushed into the chondrolabral junction as well.

Labrum Parameter Study: Logarithmic Shear Strain

Small Hip Joint Clearance, Large Cam Protrusion, and Large Cam Width

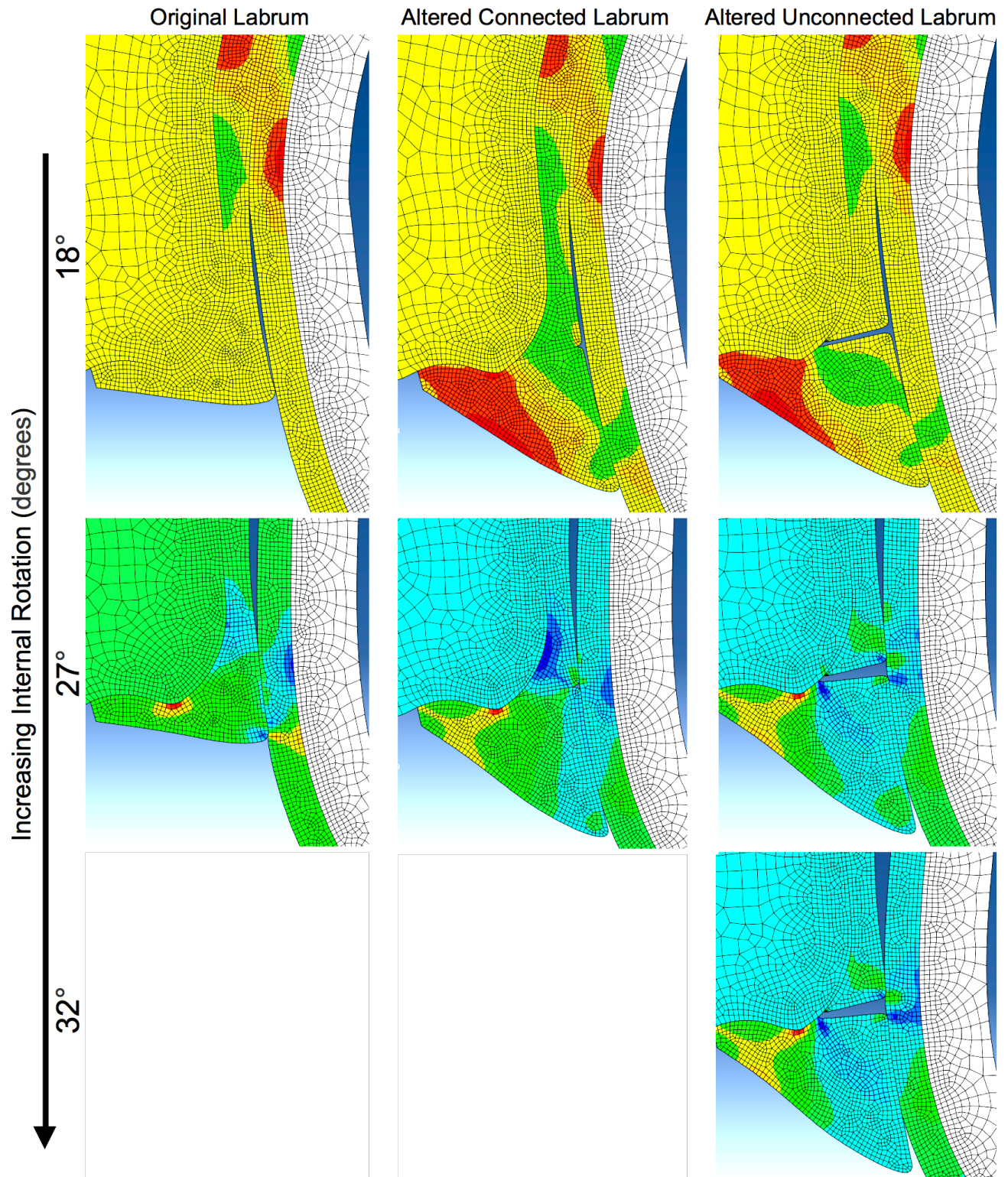


Figure 7.2: The logarithmic shear strain FE results are plotted for the labrum parameter study. All models resulted in great shear strains in the labrum near the labral-bone junction. An elongated labrum resulted in earlier contact with the cam deformity. An elongated unconnected labrum resulted in the widening of the chondrolabral junction upon further entry of the cam deformity and did not result in great shear strains near the cartilage-bone interface of the anterior acetabular articular cartilage as in the other models.

Quantitative Results

Quantitative logarithmic shear strain results for the labrum parameter study for a model with small clearance, large cam protrusion, and large cam width are presented in Figure 7.3. The results for varying labral anatomy for soft tissues including the anterior and posterior acetabular articular cartilage, the labrum, and femoral articular cartilage are shown.

For the original labrum, a steep rise in shear strain occurred simultaneously in the anterior acetabular articular cartilage and the labrum at 18° of internal rotation. For the altered elongated connected labrum, a steeper rise in shear strain occurred, and it occurred earlier in the labrum and anterior acetabular articular cartilage at 13.5° of internal rotation, but not in the positive shear strain component of the anterior acetabular articular cartilage. For the altered elongated unconnected labrum, a steep rise in shear strain occurred earlier in the labrum at 13.5° of internal rotation, while a further rise in shear strain in the anterior acetabular articular cartilage followed at 18° of internal rotation.

The elongated connected labrum resulted in shear strains around up to five times larger than the shear strains produced in the other models in the anterior acetabular articular cartilage and labrum; the positive shear strains (CCW direction) were about two times greater than the negative shear strains in these tissues, meaning they were dominated by shear strains that pushed them into the hip joint. In contrast, the negative component of the shear strains (CW direction) in the elongated unconnected and original labrum models were slightly greater than their positive counterpart in the anterior acetabular articular cartilage, meaning these tissues were dominated by shear strains that pushed them outwards from the hip joint. The altered labrums resulted in greater negative shear strains in the femoral articular cartilage, whereas the original labrum resulted in greater positive shear strains. This result shows that the femoral articular cartilage was dominated by shear strains that pushed it inwards into the hip joint in the altered labrum models and outwards from the hip joint in the original labrum model.

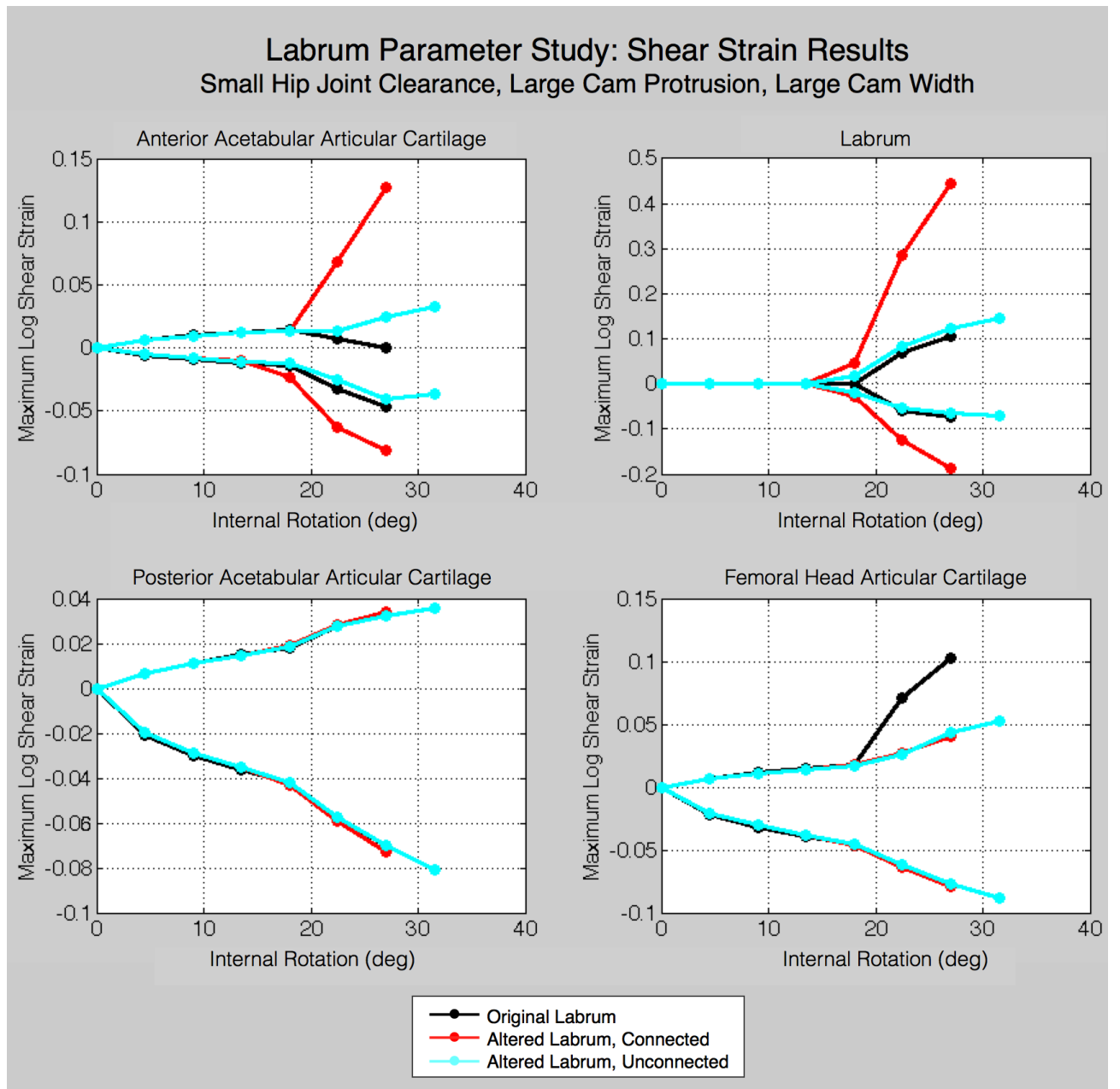


Figure 7.3: The quantitative shear strain FE results for the labrum parameter study for soft tissues are presented. An earlier rise in shear strain in the labrum occurred in the altered labrum models. Shear strains around five times as large were produced in the anterior acetabular articular cartilage and labrum by an elongated connected labrum compared to other models.

7.2.3 Discussion

In all the models, the greatest shear strain in the labrum developed near the labral-bone interface towards the outer rim of the acetabulum. This suggests that this attachment region of the labrum was where damage was most likely to occur regardless of the variations in labral anatomy that were investigated in this chapter. Shear strains in other areas of the labrum, however, were affected by the length and attachment anatomy of the labrum. An elongated labrum resulted in large strains developing in the mid-section and articulating surface of the

labrum, whereas a less extending labrum resulted in large strains developing at the labral tip. Also, an unconnected elongated labrum resulted in large strains at its attachment region directly next to the chondrolabral junction as well as at its tip next to the chondrolabral junction and the articulating surface.

An elongated labrum resulted in early abutment of the labrum which was as expected since it extended closer to the cam deformity. The attachment of the elongated labrum was also altered so that the chondrolabral junction was located more towards the outer rim of the acetabulum. This also meant that the anterior acetabular articular cartilage was also extended towards the outer rim of the acetabulum and closer to the cam deformity. However, a steep and early rise in shear strain did not occur in the anterior acetabular articular cartilage. This was probably because the shape of the acetabular bone at the cartilage-bone interface near the chondrolabral junction curved away from the HJC, making the articular cartilage in this region further from the abutment of the cam deformity. An early and slight increase in magnitude in the negative shear strain component of the anterior acetabular articular cartilage of the elongated connected labrum did occur. The negative shear strain in the anterior acetabular articular cartilage represent the tissue being pushed outwards from the hip joint. So, the connected chondrolabral junction probably allowed the early abutment of the labrum to pull the attached anterior acetabular articular cartilage outwards from the hip joint. In contrast, shear strain did not develop at the cartilage-bone interface of the anterior acetabular articular cartilage in the unconnected labrum model, which suggests that having an unconnected labrum may reduce the likelihood of damage such as cartilage delamination from occurring in this region. However, it is possible that a crack through the cartilage's thickness could occur in this region, and the propagation of this crack could initiate cartilage delamination. This progression of damage was shown in Sample 2 for the *in vitro* damage experiments in Section 5.4.4.

The original labral anatomy with a short extension was the only model where greater positive shear strains occurred in the femoral articular cartilage. This was probably due to its abutment with the labral tip so it was pushed outwards from the hip joint. For the elongated labrum models, the shear strains in the femoral articular cartilage increased smoothly because it contacted with the articulating surface of the elongated labrum rather than its tip. The elongated labrum could have also increased the contact surface area with the femoral articular cartilage and allowed for smoother articulation of the cam deformity into the hip joint. Additionally,

the deepening of the elongated labrum could have further supported the pushing the femoral head articular cartilage into the hip joint rather than outwards.

The greatest shear strains in the anterior acetabular articular cartilage and labrum occurred in the elongated connected labrum model compared to other models. In this model, the articulating surfaces of the labrum and anterior acetabular articular cartilage at the chondrolabral junction did not lie flush with each other as modelled in the original labrum model. Instead, a curved corner was created in the anterior acetabular articular cartilage which was where the greatest shear strain developed. It is possible that if the curved corner was modelled to be flush with the labrum's articulating surface, the resulting shear strains would have been reduced. It is unknown, however, which geometric representation of the chondrolabral junction is more realistic or if both representations model different realistic conditions.

For the elongated connected labrum model, the greatest shear strains developed at the attachment region of the labrum. This possibly resulted because the greatest shear strain in the anterior acetabular articular cartilage occurred at its corner at the chondrolabral junction and was pushed into the hip joint; the anterior acetabular articular cartilage that was connected to the labrum at the chondrolabral junction could possibly have pulled the labrum at its attachment region into the hip joint as well.

To represent the effect that varying labral anatomy had on the cam-type FAI mechanism, the shear strain directions discussed in this section are depicted with black arrows, while red arrows were used to indicate the greatest shear strains that developed in the soft tissues, and a red circle was used to indicate the dominant direction of shear strain in a tissue (Figure 7.4).

For models with the original labrum, the cam abutted against the labrum, pushing the labral tip outwards from the hip joint which pushed the labrum at the labral-bone junction (towards the acetabulum's outer rim) inwards into the hip joint. Also, the anterior acetabular and femoral cartilages were more dominated by shear strains that pushed them outwards from the hip joint in regions near their respective cartilage-bone interface.

For the altered labrum models, the labrum at the labral-bone junction (towards the acetabulum's outer rim) was also pushed inwards into the hip joint, however, their femoral cartilages were more dominated by shear strains that pushed them inwards into the hip joint. Also, shear

Cam-type FAI Mechanism for Labral Anatomy Variations

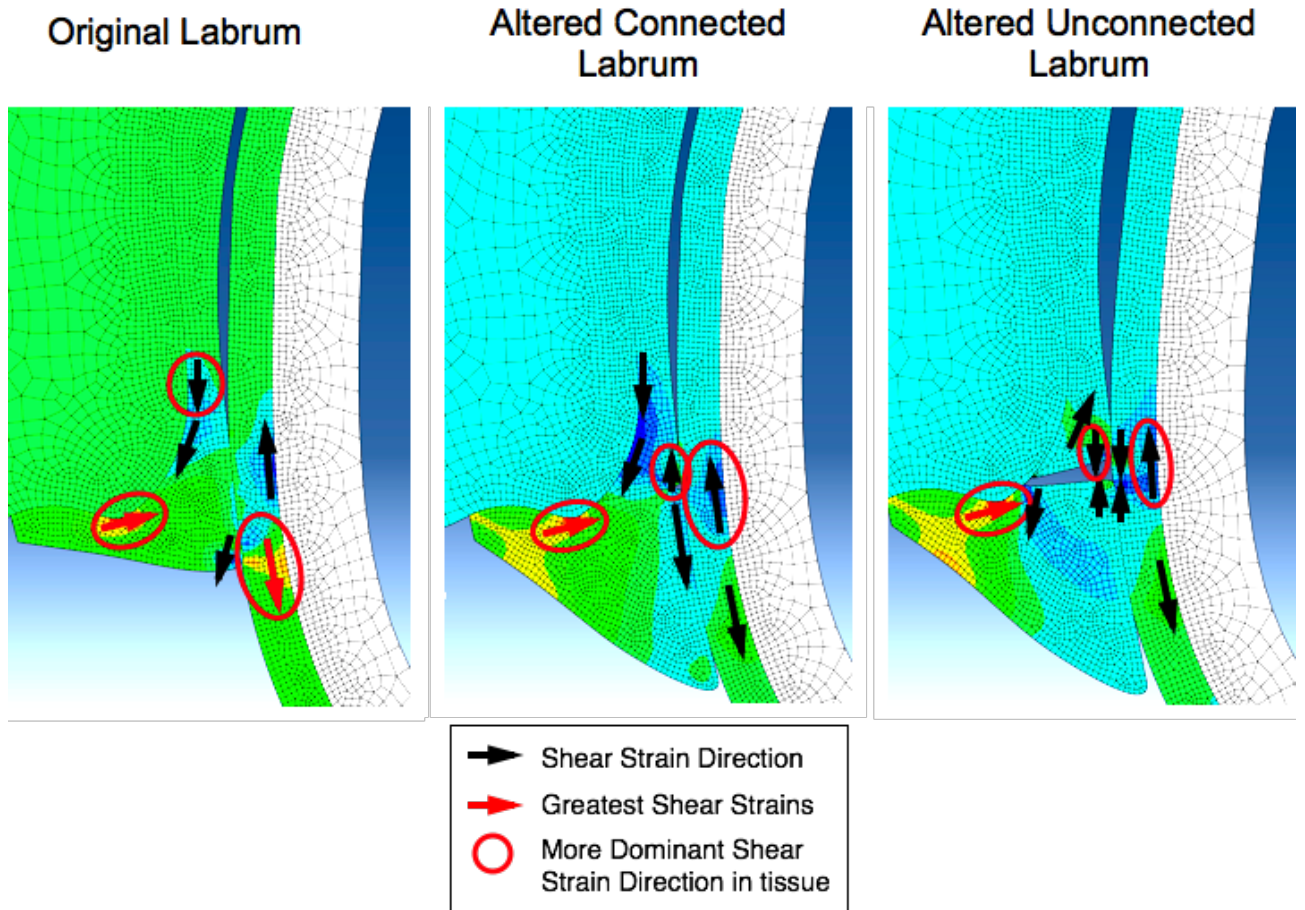


Figure 7.4: Representations of the effect of labral anatomy variation are presented. For the altered labrums, in contrast to the original labrum, cam-type FAI was generally dominated by the pushing of the femoral articular cartilages inwards into the hip joint and did not involve dominating shear strains at the anterior acetabular articular cartilage-bone interface. For the elongated unconnected labrum, all the soft tissues in the region of the chondrolabral junction were pushed inwards into the chondrolabral junction as the junction widened upon further internal rotation.

strains in the anterior acetabular articular cartilage were greatest at its corner (near the chondrolabral junction and the articulating surface) rather than at the cartilage-bone interface. The anterior acetabular articular cartilage corner was pushed inwards for an elongated connected labrum and outwards for an elongated unconnected labrum. The development of large shear strains in this area could possibly give rise to a crack through the acetabular articular cartilage, a type of damage which was not simulated in the parameter study in Chapter 6. The development of large shear strains in this region could have possibly been caused by the curved anatomy of the cartilage in this region or by the fact that the acetabular articular cartilage was modelled to extend further towards the outer acetabular rim.

7.2.4 Conclusion

Investigating the effect of having an elongated labrum and a labrum with a connected or unconnected chondrolabral junction showed that these variations in anatomy affected the cam-type FAI mechanism. Having an elongated labrum and/or wide cam deformity caused earlier contact between the cam deformity and labrum, and therefore earlier development of shear strains in the labrum during internal rotation at flexion. A cam deformity with large protrusion and width consistently caused great shear strains to develop in labrums with different anatomy at their attachment region to the acetabulum, indicating that there was a tendency for damage to occur at the attachment region of the labrum. It was also shown that a crack through the cartilage may possibly initiate at its corner (near the chondrolabral junction and the articulating surface) if it is modelled to be curved (rather than flush with the labrum's articulating surface) and it extends further outwards from the joint towards the acetabular rim.

Additional areas of large shear strain in the labrum, and therefore, other possible areas of damage, occurred depending on the anatomy of the labrum. Shear strains develop at the tip of a labrum if the labrum did not extend very far, whereas for an elongated labrum, shear strains developed along its articulating surface. For an unconnected elongated labrum, shear strains also developed at its attachment region directly next to the chondrolabral junction.

A connected labrum showed that it may be more damaging for the anterior acetabular articular cartilage and labrum compared to an unconnected labrum, resulting in shear strains of up to five times greater. A connected labrum also resulted in greater strains at the anterior acetabular articular cartilage-bone interface, revealing its higher potential to lead to cartilage delamination occurring close to the chondrolabral junction. However, for an unconnected labrum, large shear strains within the cartilage showed that a crack through the anterior acetabular articular cartilage's thickness could have possibly occur near the region of the chondrolabral junction; the propagation of this kind of crack could have potentially caused cartilage delamination to initiate further from the chondrolabral junction.

7.3 Surgical Parameter Study

During reshaping surgery, it is typical that the full width and protrusion of a cam deformity is removed as well as some additional portion of the underlying bone. How much of the cam deformity and bone to remove to relieve a patient from impingement, further damage, and pain is often in question. The current practice of removing an additional portion of the underlying bone upon cam removal compared to removing only the protrusion of the cam (leaving a spherical femoral head) is also in question. Therefore, reshaping surgery was simulated for the full or partial removal of the cam deformity and for the case where the femoral head was shaped to be perfectly spherical. Surgical simulations were conducted on two hip joints, one with a cam that abutted the labrum, and another with a cam that entered the hip joint and affected the acetabular articular cartilage.

7.3.1 Methods

Surgical Simulations

The protrusion of the cam deformity was removed to create a spherically shaped femoral head. Full removal similar to current surgical practice was conducted by removing the entire width and protrusion of the cam deformity and cutting out deeper parts of the femoral bone as well. Partial removal of the cam deformity was created by removing 75% of its cam width and also cutting out deeper parts of the femoral bone (Figure 7.5). In all the surgical simulations, the areas of the femoral bone where the cam deformity was cut was left as bone without a layer of articular cartilage.

Geometry, Mesh Generation, Assigning Boundary Conditions, and Defining Material Properties, Interactions, and Analysis Steps

The geometry of the hip joint models with a small hip joint clearance and 1) a small cam protrusion and middle cam width and 2) a large cam protrusion and width, that were developed in Chapter 6 were used. Similar mesh generation, boundary conditions, material properties, interactions, and analysis steps were applied in the FE models as they were applied in the Internal Rotation at Flexion model in Stage 4 in Section 4.5.1. These analysis formulations

Surgical Parameter Study Geometry

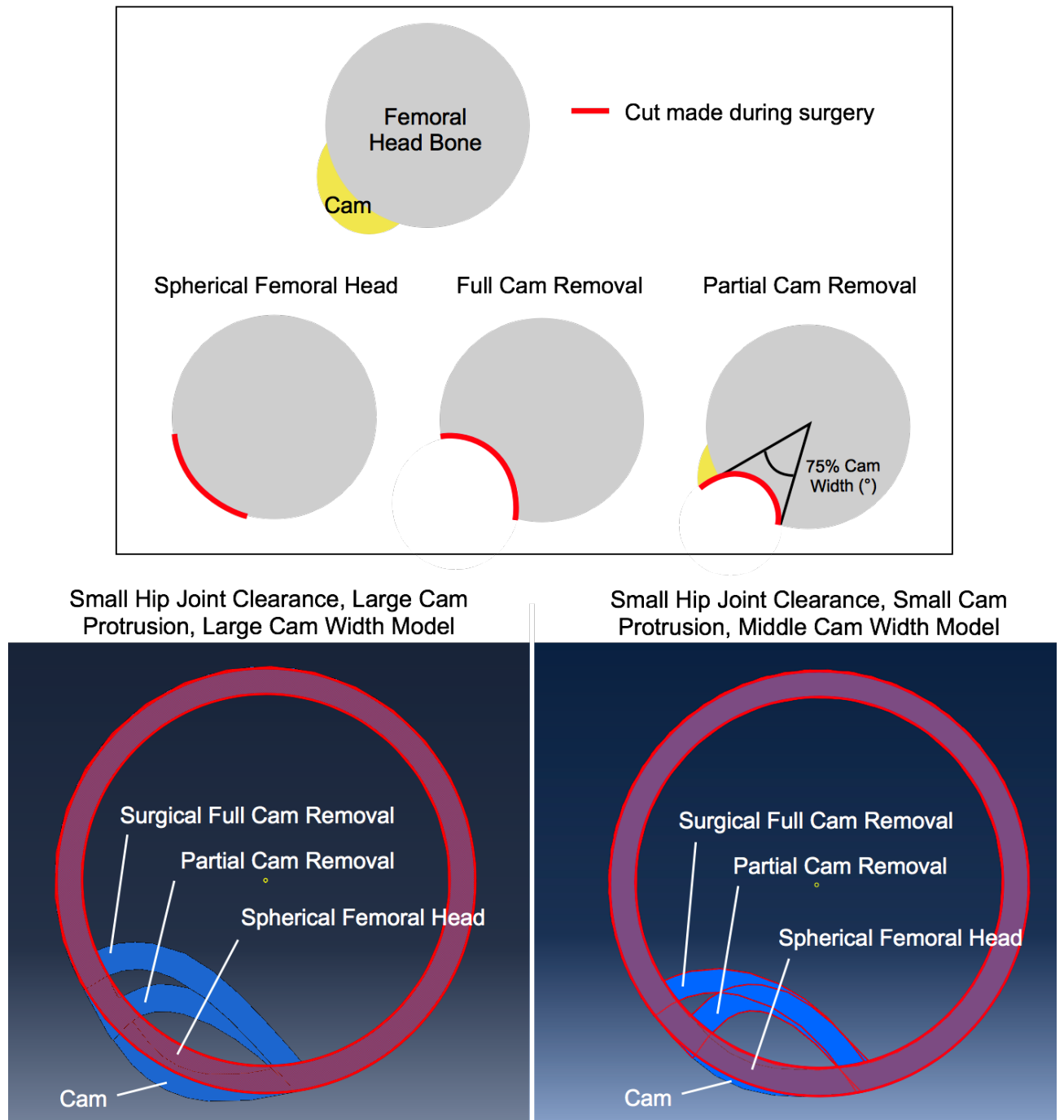


Figure 7.5: The variations of femoral head bony anatomy to be studied in the surgical parameter study for strain behavior with respect to cam-type FAI are displayed. Surgical simulations included full cam removal similar to current surgical practices, partial cam removal, and fully removing the cam protrusion to reshape the femoral head to be spherical (top). The femoral bone of surgical simulation models are presented for hip joints with small clearance and 1) a large cam protrusion and width (bottom left) and 2) a small cam protrusion and middle cam width (bottom right).

created models which simulated the internal rotation of the femoral head during the activity of sitting when the hip was flexed at 40° (with a joint load of 0.1 N medially and 1.92 N anteriorly). In the presence of a cam deformity, these FE models, therefore, simulated cam-type FAI during

the motion of internal rotation at flexion. In the case where surgery was simulated, these models enabled the effect of surgery to be assessed.

7.3.2 Results

Shear Strain

Logarithmic shear strain FE results are presented for the surgical parameter study for a model with small clearance and cam deformities with:

1. A small protrusion and middle width (that enters the hip joint and affects the acetabular articular cartilage) (Figure 7.6)
2. A large protrusion and large width (that abuts the labrum) (Figure 7.7)

They are presented by displaying a matrix of plots where the progression of internal rotation is presented down each column, while across the columns, plots for models with different anatomy (pre- and post- surgery) are given. The greatest negative shear strains (shear strains in the CW direction) are represented in blue, and the greatest positive shear strains (shear strains in the CCW direction) are represented in red.

For the model with a small cam protrusion and middle cam width (Figure 7.6), the cam abutted the labral tip and was still able to enter into the hip joint, causing shear strain to form at the cartilage-bone interface of the acetabular and femoral articular cartilages. For all the variations of surgical simulation, abutment of the labrum was either reduced or eliminated during internal rotation at flexion and resulted in large positive shear strains (CCW direction) near the femoral articular cartilage-bone interface at the anterior edge where the cam had been removed. Positive shear strains in the femoral articular cartilage indicated that the tissue was being pushed outwards from the hip joint in these regions. At 45° of internal rotation, some positive (light yellow) and negative (light blue) shear strains developed near the cartilage-bone interface of the anterior acetabular articular cartilage in the models with a spherical femoral head and full cam removal. The negative shear strains (CW direction) developed along the interface where the acetabular articular cartilage was in contact with the edge of the femoral articular cartilage, which meant the tissue was being pushed outwards from the hip joint in these regions. The positive shear strains (CCW direction) developed further along the interface

compared to the negative shear strains, which meant the tissue was being pushed into the hip joint in these regions.

Very little shear strains developed in the region of the anterior acetabular articular cartilage in the model with partial cam removal. At 45° of internal rotation, the femoral articular cartilage in this model developed positive and negative shear strains at its anterior cut edge near the cartilage-bone interface. This indicated the tissue experienced greater compression in this region due to internal rotation compared to the other surgical models. The tissue could not be wedged any further into the hip joint in this region, so the model's ROM for internal rotation remained at 45° . In contrast, the ROM for internal rotation for the other surgery models was about 20° larger. At ROM's greater than 45° of internal rotation for the spherical femoral head and full cam removal models, the maximum positive shear strain (red) at the edge of the femoral articular cartilage became smaller in area. The surrounding region surrounding it (green) also represented positive shear strains, and the region of positive (green) and negative (dark blue) shear strain in the anterior acetabular articular cartilage were greater compared to smaller ROM's.

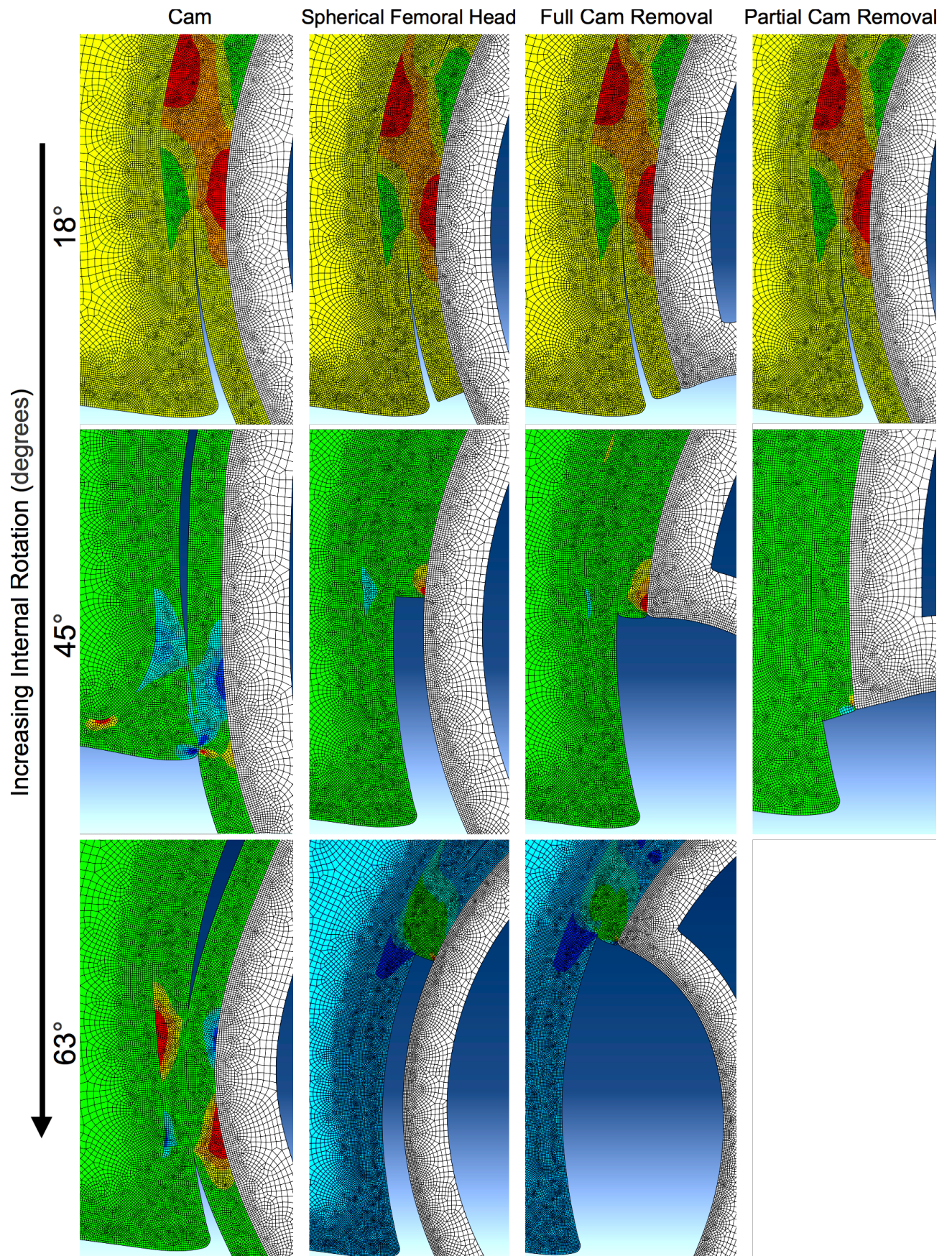


Figure 7.6: The logarithmic shear strain FE results are plotted for the surgical parameter study for a model with small clearance, small cam protrusion, and middle cam width. In all the surgical models, the abutment of the labrum was greatly reduced and large shear strains developed at the femoral articular cartilage's anterior cut edge near the cartilage-bone interface.

For the model with a large cam protrusion and width (Figure 7.7), the cam abutted the labrum and was not allowed to enter further into the hip joint, causing shear strains to form at the labral tip, the labral-bone interface, and in the region of the chondrolabral junction. For all the variations of surgical simulation, internal rotation in the hip joint did not lead to any abutment of the labrum, and the ROM's for internal rotation were much greater, reaching beyond 50°. At 27° of internal rotation, for the spherical femoral head and full cam removal models, large positive shear strains (CCW direction) developed near the femoral articular cartilage-bone interface at the anterior edge where the cam had been removed. For the partial cam removal model, large positive shear strains developed at the femoral articular cartilage-bone interface in the region that would have had a cut edge if the cam was fully removed. Positive shear strains (CCW direction) in these regions indicated that the tissue was being pushed outwards from the hip joint in these areas. For ranges above 40° of internal rotation, the shear strains in the partial cam removal model became more similar to the shear strains in the other surgery models. At these ROM's, the surgery models developed maximum positive (red) and negative (dark blue) shear strains at the cut edge corner of the femoral articular cartilage near its cartilage-bone interface, indicating squeezing of the tissue in this region. Positive shear strains (CCW direction) developed in the anterior acetabular and femoral articular cartilages in the areas where their articulating surfaces were in contact, indicating the femoral articular cartilage was being pushed outwards from the hip joint in these areas, while the acetabular articular cartilage was being pushed into the hip joint in these areas. Negative shear strains (CW direction) developed in the anterior acetabular articular cartilage in the region that was more outwards from the anterior cut edge of the femoral articular cartilage, indicating that the tissue was being pushed outwards from the hip joint in this area.

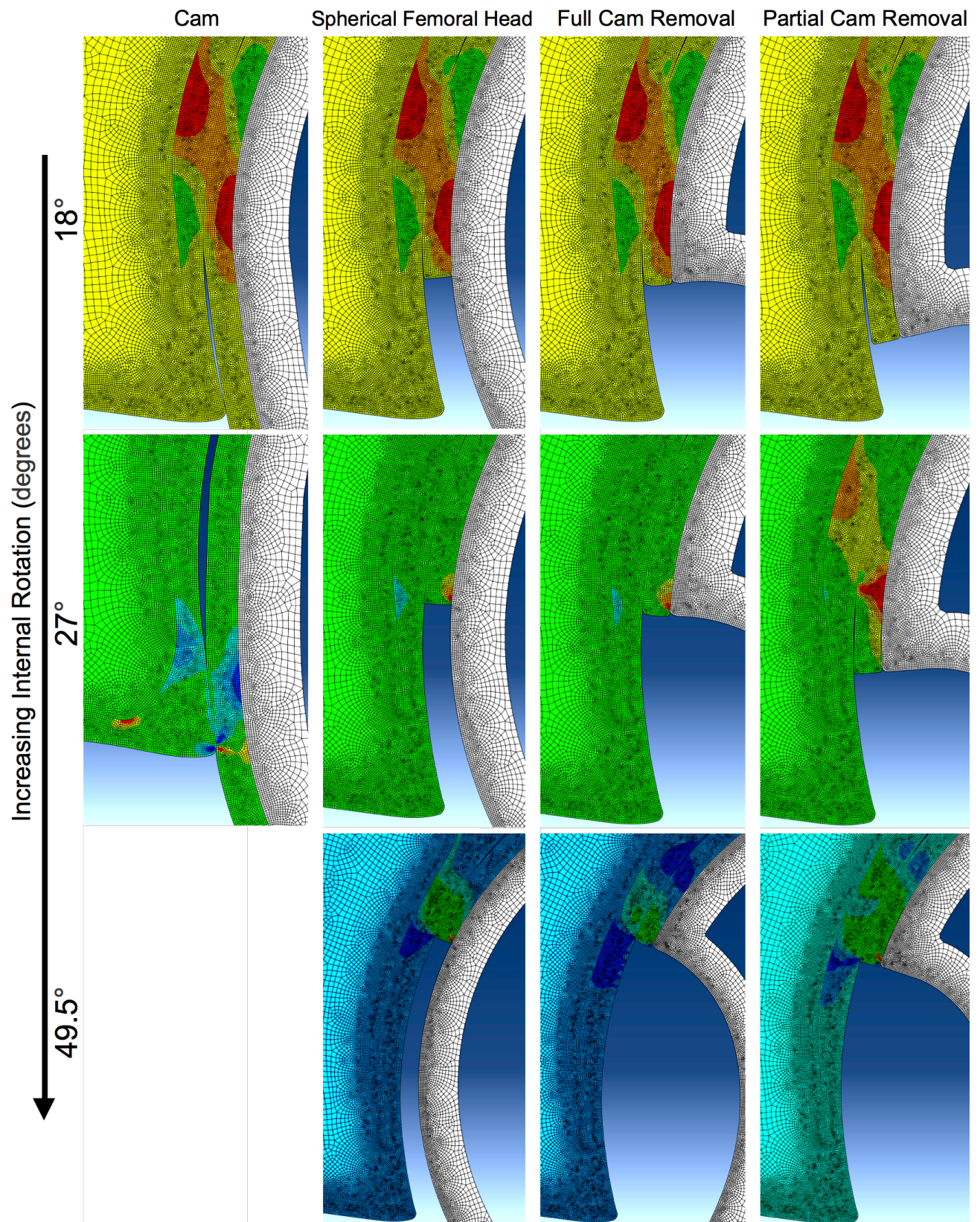


Figure 7.7: The logarithmic shear strain FE results are plotted for the surgical parameter study for a model with small clearance, large cam protrusion, and large cam width. In all the surgical models, no shear strains developed in the labrum and large shear strains developed at the femoral articular cartilage's anterior cut edge near the cartilage-bone interface. Their ROM's were also greatly increased compared to the model with a full cam.

Quantitative Results

Quantitative logarithmic shear strain results are presented for the model with small hip joint clearance and cam deformities with:

1. A small protrusion and middle width (that enters the hip joint and affects the acetabular articular cartilage) (Figure 7.8)
2. A large protrusion and large width (that abuts the labrum) (Figure 7.9)

The results for varying surgical simulations for soft tissues including the anterior and posterior acetabular cartilage, the labrum, and femoral cartilage are shown.

For the model with a small cam protrusion and middle cam width (Figure 7.8), a spherical femoral head and full cam removal led to shear strains being entirely eliminated from the labrum. They also led to reduced shear strains in the acetabular articular cartilages, but increased positive shear strains in the femoral articular cartilage, corresponding to the region of the cut edge of the femoral articular cartilage near its cartilage-bone interface. Partial cam removal led to greatly reduced ROM's, reduced positive shear strains in the labrum, increased positive shear strain in the femoral articular cartilage, and not much change in the acetabular articular cartilages.

For the model with a large cam protrusion and width (Figure 7.9), a full cam deformity was not able to enter the hip joint further than 27° internal rotation, whereas the surgical simulation models resulted in ROM's for internal rotation of approximately 30° larger. The surgical models led to the elimination of the shear strains in the labrum. For internal rotations below 27° , the surgical models also led to reduced shear strains in the femoral and posterior acetabular articular cartilages and reduced negative shear strains in the anterior acetabular articular cartilage compared to the cam model. However, for internal rotations above 27° , the surgical models resulted in gradually increasing and much larger shear strains in the anterior acetabular and femoral articular cartilages compared to the cam model. These shear strains were similar to those that developed in the surgical models for a cam with small protrusion and middle cam width.

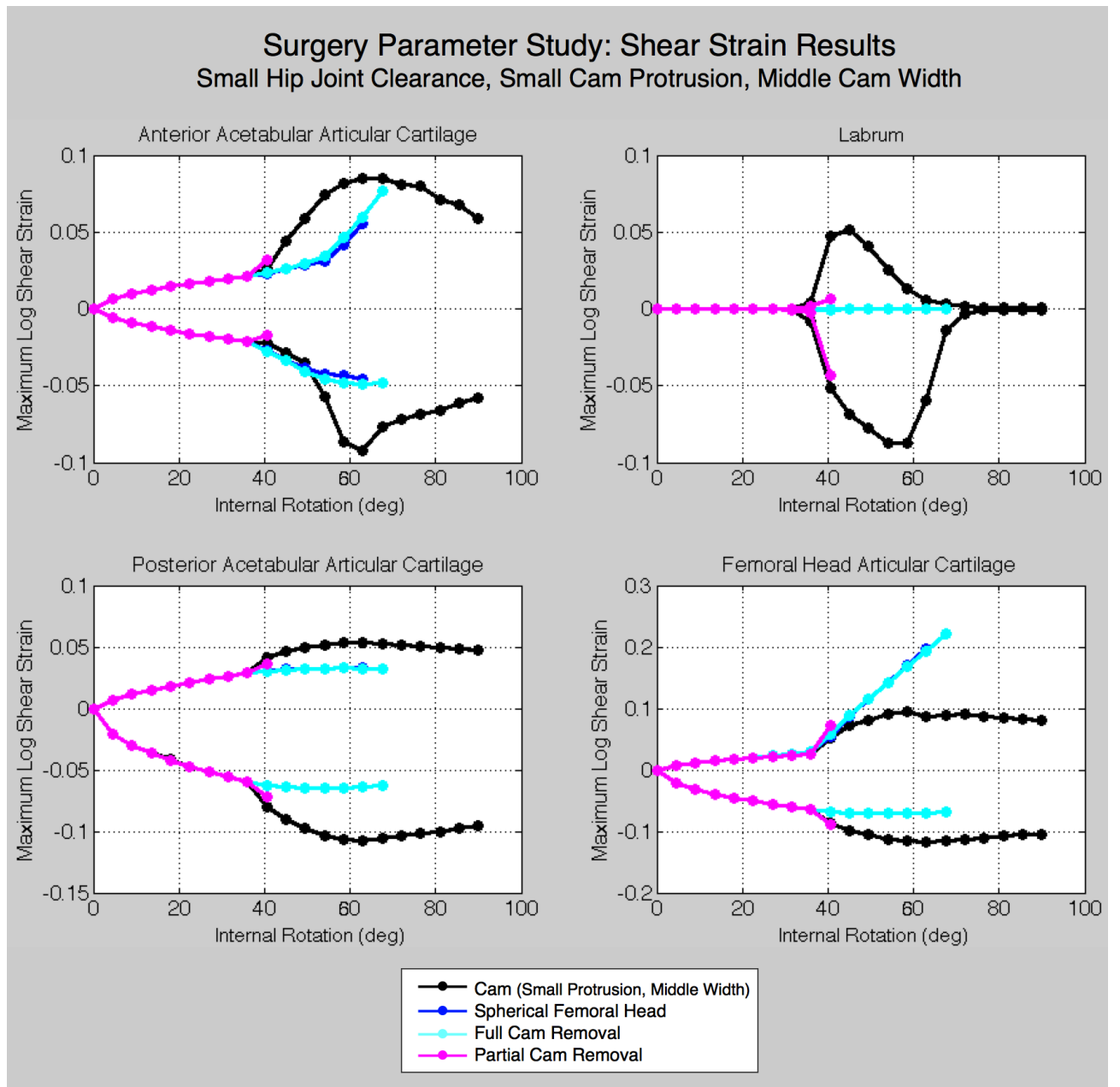


Figure 7.8: The quantitative shear strain FE results for the surgical parameter study for a model with small clearance, small cam protrusion, and middle cam width are presented for soft tissues. A spherical femoral head and full cam removal resulted in shear strains being entirely eliminated from the labrum and reduced shear strains in the acetabular articular cartilages. The positive shear strains in the femoral articular cartilage were increased in all the surgical models (without cams), which corresponded to the region of the cut edge of the femoral articular cartilage near its cartilage-bone interface.

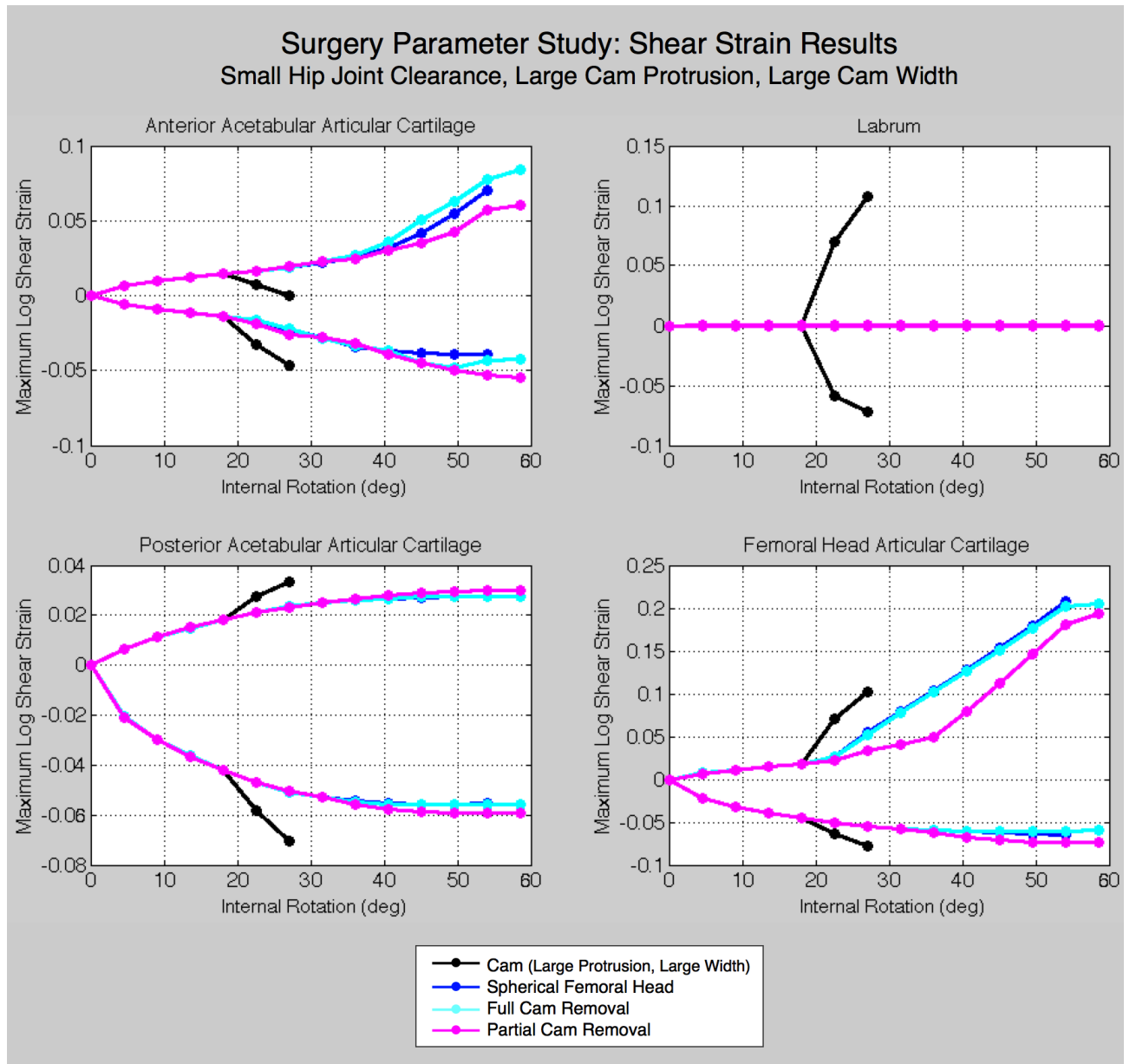


Figure 7.9: The quantitative shear strain FE results for the surgical parameter study for a model with large cam protrusion and width are presented for soft tissues. The surgery models (without cams) led to the elimination of the shear strains in the labrum and much larger ROM's compared to the cam model. However, for internal rotations above 27°, the surgical models resulted in gradually increasing and much larger shear strains in the anterior acetabular and femoral articular cartilages compared to the cam model.

7.3.3 Discussion

For a cam deformity with small protrusion and middle cam width, full removal or removing the cam's protrusion (leaving a spherical femoral head) reduced the shear strains in the acetabular articular cartilage and labrum. Whether the strain reduction would have been enough to prevent any further damage from progressing is not known. It can be hypothesised that at the very least, the reduction in shear strains could possibly slow down the progression of damage.

Large strains in the femoral articular cartilage were also produced in this model, but they are probably not of concern because they are in a region that is not normally loaded during daily functional activities.

For a cam deformity with large protrusion and width, all the surgical simulations eliminated the shear strains in the labrum and greatly improved the ROM of the hip joint. Therefore, the surgical procedures could possibly relieve a patient who has reduced ROM and possibly prevent further labral damage. However, for greater ROM's, the shear strains in the anterior acetabular became greater than the shear strains that were reached at smaller ROM's when a cam was present. These shear strains may possibly be a result of normal loading, but if any damage had previously occurred in the region of the chondrolabral junction, the development of further shear strains in the region in the acetabular articular cartilage could potentially still initiate damage such as cartilage delamination. This meant that if a patient who originally had a labrum-abutting cam underwent surgery, it is possible that post-surgery, they may possibly develop a condition where damage to the acetabular articular cartilage may be of concern.

For both of the cam deformities that were studied, removing the cam's protrusion so that the underlying bone laid flush with the rest of the femoral head (creating a spherical femoral head) resulted in slightly less shear strain in the anterior acetabular articular cartilage compared to when the cam was removed by cutting deeper into the bone of the femoral head. This showed that cutting deeper into the bone may be more damaging compared to reshaping the femoral head to be perfectly spherical. Also, for a cam with a large protrusion and width, partial removal of the cam deformity resulted in even less shear strains in the anterior acetabular and femoral articular cartilages. This was not the case, however, for a cam with a small protrusion and middle cam width where the partial removal caused a much reduced ROM. The cause for the different results arising in the two models for partial cam removal was not clear and would require further parameter studies to understand. However, these results showed that, depending on the shape and size of the cam, partial removal may possibly be the most beneficial surgical treatment, even when compared to when the femoral head is reshaped to be spherical, but it may possibly be more damaging as well.

The common practice of removing deeper portions of the femoral bone upon removal of the cam deformity is thought to possibly be damaging to the acetabular articular cartilage due to

the resulting sharp edge. In this study, the shear strains in the anterior acetabular articular cartilage were shown to be reduced when the cam was fully removed and deeper portions of the bone were cut out. However, this study does not eliminate the possibility that cutting deeper into the bone may be damaging to the acetabular articular cartilage. It is possible that for other loads and motions, it could be more damaging compared to the cutting of the cam deformity so the underlying bone lies flush with the rest of the femoral head.

To represent the effects of surgery, the shear strain directions discussed in this section are depicted with black or white arrows, while red arrows were used to indicate the greatest shear strains that developed in the soft tissues, and a red circle was used to indicate the dominant direction of shear strain in a tissue (Figures 7.10, 7.11). Black or white arrows were chosen to be based on clarity of the arrow against the plots.

For the models with small cam protrusion and middle cam width (Figure 7.10), the spherical femoral head and full cam removal models led to similar mechanisms where the anterior acetabular articular cartilage was more dominated by shear strains that pushed it inwards into the hip joint in the region where its articulating surface was in contact with the femoral articular cartilage. Also, the femoral articular cartilage was dominated by shear strains that pushed it outwards from the hip joint in the region of the cut edge near the cartilage-bone interface. However, the cut edge of the femoral articular cartilage in the full cam removal model experienced additional negative shear strains which pushed it inwards into the hip joint as well, representing the squeezing of the tissue in this region. This squeezing of the femoral articular cartilage also dominated the shear strain behavior in the partial cam removal model.

For the models with large cam protrusion and width (Figure 7.11), all the surgery models led to similar mechanisms to the full cam removal model of a hip joint with small cam protrusion and middle cam width (Figure 7.10). The anterior acetabular articular cartilage was more dominated by shear strains that pushed it inwards into the hip joint in the region where its articulating surface was in contact with the femoral articular cartilage. Also, the femoral articular cartilage was dominated by shear strains that pushed it outwards from the hip joint in the region of the cut edge near the cartilage-bone interface.

Cam-type FAI Mechanism for Surgery Simulations

Small Hip Joint Clearance, Small Cam Protrusion, Middle Cam Width

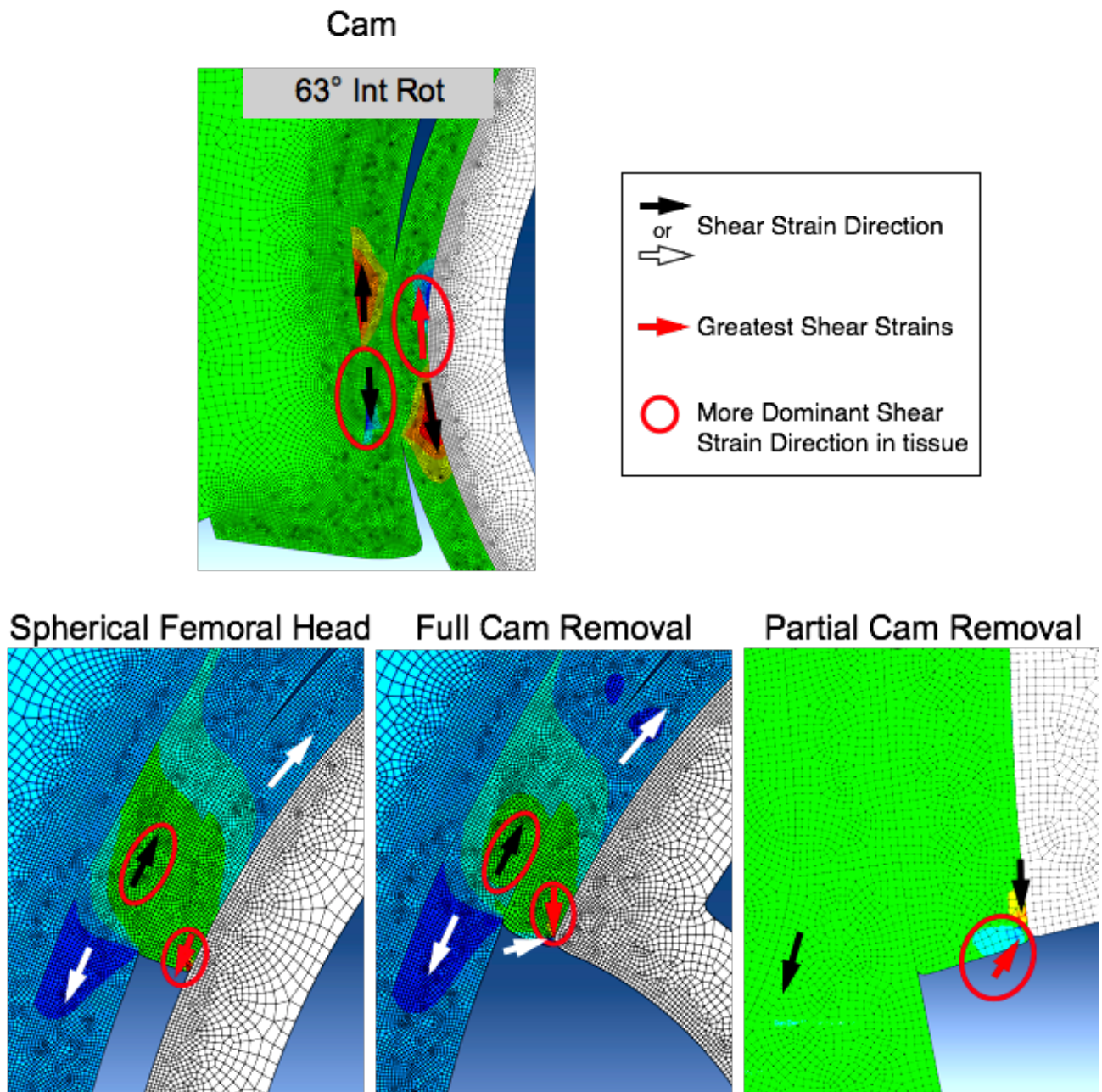


Figure 7.10: Representations of the effect of surgical simulations in hips with small clearance, small cam protrusion, and middle cam width are presented. The spherical femoral head and full cam removal models led to similar mechanisms where the anterior acetabular and femoral articular cartilages were more dominated by shear strains that pushed them inwards into the hip joint and outwards from the hip joint, respectively. The cut edge of the femoral articular cartilage in the full and partial cam removal models experienced positive and negative shear strains, representing the squeezing of the tissue in this region.

Cam-type FAI Mechanism for Surgery Simulations

Small Hip Joint Clearance, Large Cam Protrusion, Large Cam Width

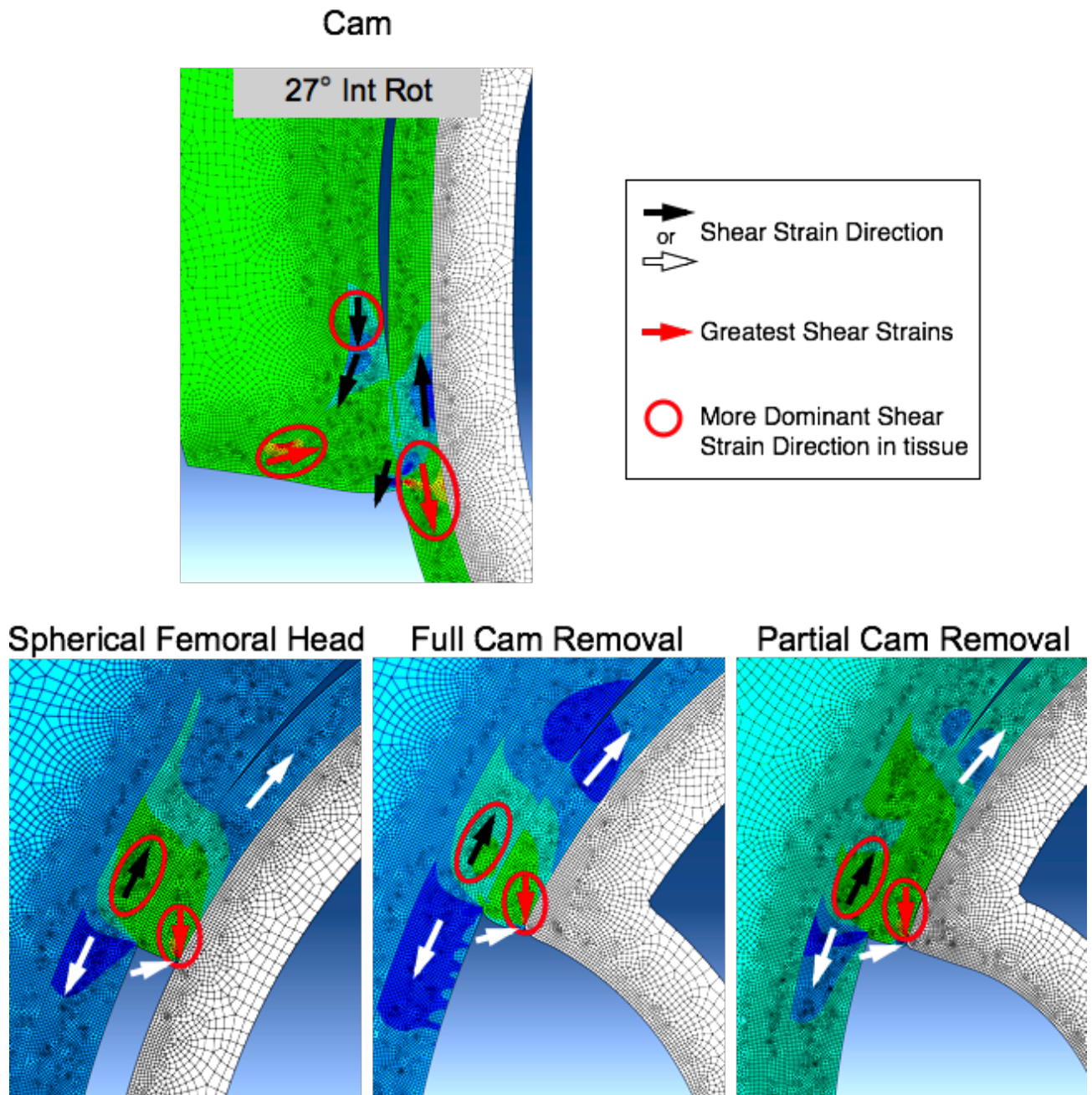


Figure 7.11: Representations of the effect of surgical simulations in hips with small clearance and large cam protrusion and width are presented. All surgery models led to similar mechanisms where the anterior acetabular articular cartilage was more dominated by shear strains that pushed it inwards into the hip joint in the region where its articulating surface was in contact with the femoral articular cartilage. Also, the femoral articular cartilage was dominated by shear strains that pushed it outwards from the hip joint in the region of the cut edge near the cartilage-bone interface.

7.3.4 Conclusion

This study demonstrated that full removal and even only partial removal of a cam deformity may be sufficient in increasing an individual's hip ROM, preventing the abutment of the labrum and therefore eliminating impingement and preventing further labral damage. The removal of a cartilage-affecting cam reduced the shear strains experienced in the anterior acetabular articular cartilage. However, whether the strain reduction was enough to prevent any further damage from progressing in the articular cartilage is not known. Also, the removal of a labrum-abutting cam increased the ROM of a patient's hip, but this also enabled greater shear strains to develop in their anterior acetabular articular cartilage, which may possibly lead to articular cartilage damage.

The removal of the protrusion of a cam deformity so that the underlying bone lies flush with the rest of the femoral head resulted in reduced shear strains compared to when deeper portions of the bone were removed upon cam removal. The partial removal of cams with different morphology led to the greatest reduction in shear strains or decreased ROM. Also, all surgical procedures required the cutting of the femoral articular cartilage which led to large shear strains at the cut edge of the femoral articular cartilage near the cartilage-bone interface; but since this cut edge was in a region not normally loaded during daily functional activities, it could possibly not be of concern.

7.4 Overall Chapter Conclusion

The labrum parameter study in this chapter showed that having a connected chondrolabral junction may be more damaging than having an unconnected chondrolabral junction, resulting in shear strains of up to five times greater. Cartilage delamination may be more likely to occur near the chondrolabral junction in a hip joint with a connected labrum compared to an unconnected labrum. Also, in a hip with a large cam protrusion and width, labrums are very likely to be damaged near their attachment site to the acetabular bone. If they have a short extension, they may also be likely to be damaged at their tip, whereas elongated labrums may also be likely to be damaged along their articulating surface. Furthermore, a crack through the acetabular articular cartilage may result if the labrum extends further outwards from the joint towards the acetabular rim and if its morphology at the chondrolabral junction results in

a curved corner (rather than being flush with the labrum's articulating surface).

The surgical parameter study in this chapter demonstrated that full removal and even only partial removal of a cam deformity may be sufficient in producing transformative results for a patient, relieving them of hip impingement, preventing further abutment and damage of their labrum, increasing their ROM, and hopefully reducing any symptomatic hip pain. The removal of a cartilage-affecting cam deformity can reduce the shear strains experienced in the anterior acetabular articular cartilage. Knowing whether this strain reduction could prevent or slow down the progression of further damage requires additional research. The removal of a labrum-abutting cam deformity may increase the ROM of a patient's hip, allowing for greater shear strains to develop in their acetabular articular cartilage, which may possibly lead to damage such as cartilage delamination. The removal of the protrusion of a cam deformity so that the underlying bone lies flush with the rest of the femoral head may be less damaging compared to when deeper portions of the bone are removed upon cam removal. Partial cam removal may be even less damaging or more damaging, depending on the cam's shape and size. Understanding the mechanisms involved in partial cam removal require further parameter studies to be conducted.

Chapter 8

Discussion

8.1 Overview

Cam-type FAI is a painful mechanical disorder of the hip joint involving the abutment of bony abnormalities in the hip joint during motion, often leading to articular cartilage and labral pathology such as cartilage delamination and labral tears. Cartilage delamination often occurs in the anterior superior region of the acetabulum in cam-type FAI patients and may be a precursor to the progressive degeneration of the acetabular and femoral articular cartilages in the hip joint, ultimately leading to hip OA. Early intervention and treatment of cam-type FAI is, therefore, of great importance to prevent or inhibit the progression of hip OA. The aim of this thesis was to understand the mechanism by which cam-type FAI may lead to soft tissue damage, especially cartilage delamination, to give insight into early intervention and the treatment of cam-type FAI.

Cartilage delamination in relation to cam-type FAI has never been replicated *in vitro* or realistically simulated through FE modelling, and only very conceptual ideas of how cam-type FAI may lead to articular cartilage delamination have been previously suggested. Such conceptual ideas focus on the wedging of a cam deformity into the hip joint, causing compressive and shear forces on the cartilage which may cause it to delaminate (Figure 2.5) [12, 13, 19]. In the present study, a unique hypothesis was made in Section 2.5 with regards to the damage mechanism of cam-type FAI based on the current understanding of articular cartilage behaviour during loading and articular cartilage damage. There is evidence that the earliest change in the collagen network due to mechanical overloading is loosening or breaking of the links between collagen

fibrils (entwinements of the collagen fibrils or nonentwinement-based interactions) due to shear strains along the collagen fibrils [128, 129]. There is also evidence that excessive straining of the collagen fibrils in the collagen network of articular cartilage can lead to breakage of collagen fibres [128, 129]. In addition, articular cartilage has been shown to crack at its cartilage-bone interface or tidemark [130, 131, 132] due to rapid loading at fast strain-rates and large tensile or shear strains [92, 93, 130, 131, 132]. At high strain-rates above 0.1 s^{-1} [93], articular cartilage has been determined to behave similarly to a linear elastic material [90, 91, 92, 93]. Such strain rates are reached during everyday functional activities such as walking, sitting down, or standing up from a chair. Finally, using elastic fracture mechanics, it was determined that cracks are most likely to occur near the cartilage-bone interface oriented at 45° to the interface where shear strains are greatest [135, 136]. Based on this information and research, it was hypothesised that the rapid loading of articular cartilage could cause “micro-damage” to the tissue’s collagen structure, causing the failure of interactions and entwinements between collagen fibres. Further rapid loading of collagen fibres that are less entwined would increase the strain along those fibres, especially those oriented at 45° to the interface where principle shear strains are greatest. Failure of these fibres or a “micro-crack” would make the surrounding tissue more vulnerable to further “micro-damage” and “micro-cracking”. Any tissue degeneration or bone hardening due to large stresses may also make the tissue more vulnerable to further “micro-damage” and “micro-cracking”. Furthermore, a series of “micro-cracks” could form a crack at the cartilage-bone interface or tidemark, initiating cartilage delamination.

Based on this hypothesis, it was determined that the damage mechanism of cam-type FAI should be simulated and assessed using shear strains as a damage criterion where regions of large shear strains would be vulnerable to damage initiation and propagation. Using shear strain as a damage criterion is unique compared to the few FE models of cam-type FAI which have been published [141, 142, 143], which generally compare measured stresses in cam-type FAI patients to those in normal subjects to show that stresses are greater in cam-type FAI patients in the anterolateral area of the hip joint during motions of large flexion or internal rotation. None of the previous studies have examined the shear strains in the hip joint or introduced possible detailed mechanisms for which cartilage delamination or labral tears may occur, making the the findings in this study new and valuable insights.

To provide insight for the development of a model which simulated cam-type FAI, a geometric-

kinematic hip joint model was created with the purpose of understanding which properties of the cam deformity or hip joint anatomy may potentially be the most damaging to soft tissues in the hip joint with respect to cam-type FAI. This model used kinematic data for the anterior impingement test and daily functional activities to drive bony hip joint geometry acquired from CT scans from five normal subjects and five cam-type FAI patients. Proximity algorithms were used to analyse the proximity between the acetabular and femoral bony surfaces to detect hip impingement during motion. The proximity results confirmed that cam-type impingement occurs at internal rotation at large ROM's of flexion as found in other studies [7, 10, 12, 26, 112, 113, 114, 115], but not during functional activities which often cause pain in patients such as walking, sit-to-stand, and stand-to-sit. Earlier impingement was found to occur in cam-type FAI patients compared to normal subjects possibly due to the wider coverage and protrusion of their cam deformities and their smaller hip joint clearance for the femoral head and cam. These three geometric parameters were the most dominant parameters detected, so an FE model was developed to simulate cam-type FAI and further investigate the effect of these parameters in the damage initiation and progression of soft tissues in the hip joint.

A 2D FE model was developed which uniquely utilised realistic acetabular geometry from a cam-type FAI patient, but idealised femoral geometry to allow the parametrisation of the cam protrusion, cam width, and hip joint clearance. The model simulated the activity of internal rotation during the motion of stand-to-sit and represented the soft tissues and bone as isotropic linear elastic materials. The shear strain results of the impingement FE model were validated through *in vitro* experiments which loaded porcine cartilage-bone samples in compression and shear, and used a video-based method for strain measurement of the cartilage tissue samples utilising digital image correlation (DIC). A validation FE model was created to mimic the experimental setup. Importantly, this model used the same contact and material representations and analysis framework as the other models developed in this thesis (impingement FE model). The experimental shear strain results were comparable to the validation FE results, validating the impingement FE model. The experiments also confirmed that under fast physiological loading rates, compression and shear loading can cause shear strains to develop in articular cartilage that are dominantly focused near the cartilage-bone interface. These results are especially important when defining the material definition of articular cartilage because different material definitions may lead to different results. Transversely isotropic models have been shown

to predict peak stresses at the cartilage surface and the cartilage-bone interface, while isotropic models have shown to only predict peak stresses at the cartilage-bone interface [129, 137]. The experiments, therefore, validated the impingement FE model for the calculation of shear strain and the use of isotropic models for strain analysis of cartilage-bone samples undergoing compressive and shear loading at fast physiological strain rates.

To bring greater value to the results produced by the impingement FE model, the use of shear strain as a damage criterion was verified through *in vitro* damage experiments which repetitively applied shear and slowly increasing compression on cartilage-bone samples until damaged occurred. These loading conditions repetitively applied shear strains to the articular cartilage and slowly squeezed the water out from the tissue, reducing its resilience to load. Under such loading conditions, cartilage delamination was able to be replicated repeatedly. Under certain circumstances, a superficial crack which extended to the cartilage-bone interface occurred. In some specimens, subchondral bone damage occurred, possibly due to the excessive lowering of the indenter. In every circumstance where the cartilage or cartilage and bone were cut, creating an edge similar to the edge of the acetabular rim (where cartilage has been found to delaminate), cartilage delaminated at the cut edges. This phenomenon which has been commonly observed in cam-type FAI patients has not been previously reproduced *in vitro*. These experiments have, therefore, confirmed that repetitively applied shear strain can indeed cause cartilage to delaminate from its underlying bone, making large shear strains an appropriate candidate indicator for possible soft tissue damage. More specifically, if large shear strains occur near the cartilage-bone interface, it is an indicator for the possibility of cartilage delamination to occur in this region.

The impingement FE model was used to conduct parameters studies investigating hip joint clearance (ranging between 2.88 to 3.88 mm), cam protrusion (ranging between 0.72 to 2.16 mm), and cam width (ranging between 30 to 90°). It was also used to investigate variations in labral anatomy and to simulate variations in reshaping surgery. It was found that a cam deformity with a small protrusion of 0.72 mm in a hip joint with a hip joint clearance of 2.88 mm was able to allow the cam to enter the hip joint so that large shear strains near the anterior acetabular articular cartilage-bone interface could develop, while a large cam protrusion of 2.16 mm would abut against the labrum and cause large shear strains to develop near the labral-bone interface. Therefore, in general, a small-protruding cam may lead to damage such as cartilage

delamination, whereas a large-protruding cam may lead to damage such as labral tearing at its attachment site.

The width of a cam deformity was shown to possibly affect the location at which damage might occur. In a hip with small hip joint clearance and a small-protruding cam, a larger width caused large shear strains to occur near the cartilage-bone interface further into the hip joint (and for smaller ROM's for internal rotation at flexion). Having a large cam width creates a cam that blends more smoothly with the rest of the femoral head compared to a small cam width which means that smoothly blended cams may be more likely to affect deeper areas of the hip joint, whereas less blended cams may be more likely to affect the hip joint closer to the acetabular rim. In a hip with small hip joint clearance and a large-protruding cam, a larger width caused large shear strains to occur near the labrum's attachment site to the acetabular bone, whereas a narrower width caused large shear strains to occur at the labral tip. It was also demonstrated that having a larger hip joint clearance can lead to reduced shear strains for a relatively small ROM, but can possibly lead to more damaging shear strains for a larger ROM depending on the shape and size of the cam deformity. These discovered trends provide insight into the regions of the cartilage and labrum that might be affected by a cam of a given shape and size. Having such insights can provide valuable understanding of the geometric parameters that might make a specific cam damaging and, therefore, possibly provide guidelines as to who should have surgery and which parts of a cam deformity should be removed during reshaping surgery.

Labrum parameter studies examined the effect of an elongated labrum (with an attachment site focused on the outer part of the acetabular rim) with a connected or unconnected chondrolabral junction. A connected chondrolabral junction was shown to possibly be more damaging than having an unconnected chondrolabral junction, resulting in shear strains of up to five times greater. It was also determined that cartilage delamination may be more likely to occur near the chondrolabral junction in a hip joint with a connected labrum compared to an unconnected labrum. Additionally, it was shown that in a hip with small clearance and a large cam protrusion and width, labrums are very likely to be damaged near their attachment site to the acetabular bone. If they have a short extension, they may also be likely to be damaged at their tip, whereas elongated labrums may also be likely to be damaged along their articulating surfaces.

Surgical parameter studies demonstrated that the removal of a cartilage-affecting cam can reduce the shear strains experienced in the anterior acetabular articular cartilage. Also, full removal and even only partial removal of a cam deformity may be sufficient in producing transformative results for a patient, relieving them of hip impingement, preventing further abutment and damage of their labrum and increasing their ROM. However, the removal of a labrum-abutting cam may increase the ROM of a patient's hip, allowing for greater shear strains to develop in their acetabular articular cartilage, which may possibly lead to damage such as cartilage delamination. This means that reshaping surgery may possibly relieve a patient with a cartilage-affecting cam from developing further cartilage pathology through reshaping surgery. However, reshaping surgery may possibly relieve a labrum-abutting cam from developing further labral pathology, but then allow cartilage pathology to develop as a result. The implications of these findings show that great importance should be placed on creating subject-specific surgical treatment plans because removing a cam deformity may lead to different results in different patients depending on their cam and hip joint anatomy.

Further surgery parameter studies showed that the removal of the protrusion of a cam deformity so that the underlying bone lies flush with the rest of the femoral head may be less damaging (resulting in slightly reduced shear strains) to the anterior acetabular articular cartilage compared to when deeper portions of the bone are removed upon cam removal. These results suggest that current surgical practice where deeper portions of femoral bone are removed upon removal of a cam deformity may possibly not be the optimal way of reshaping the hip joint of a cam-type FAI patient. It was also found that partial cam removal may be even less damaging, but only depending on the cam's shape and size as it can possibly be damaging in certain cases as well. Overall, these results imply that full removal of the cam deformity may not always be required for a successful result, and it is essential to carefully consider a patient's specific cam and hip joint anatomy when planning an optimal treatment.

8.2 Future Work

The 2D impingement FE model developed and used in this thesis can be improved upon by creating a 3D FE model which would allow for the study of more realistic activities and 3D variations in cam and hip joint anatomy. Additional future research which expands on the present work can focus on three areas: 1) using histology and imaging to understand the detailed

damage mechanism of cartilage delamination, 2) the development of improved methods for characterising cam and hip joint anatomy for clinical diagnosis and further parameter studies, and 3) further investigations into surgical simulations using FE for the treatment of Cam-type FAI.

8.2.1 Understanding Cartilage Delamination Using Histology and Imaging

A detailed damage mechanism for how cartilage delamination might occur was hypothesised in this study which involved the propagation of “micro-damage” to the collagen structure and “micro-cracking” or failure of individual collagen fibres, until a series of “micro-cracks” caused cartilage to crack and delaminate (Section 2.5). A possible future study which would investigate the possibility of this damage mechanism could involve the repetitive fast, but still physiological, loading of cartilage-bone samples in shear and compression followed by histology and imaging studies which would allow the collagen structure of the articular cartilage to be examined for “micro-damage”, “micro-cracking”, or any other type of damage. The shear and compression experiments could be conducted over a range of time periods and strain rates and for varying amounts of compression and shear. Also, for some samples, the experiment could be conducted until cartilage delamination occurs, while in other samples, it could be stopped prior to any visible damage. In this way, a spectrum of undamaged to damaged samples would be generated which would allow for investigations of possible “micro-damage” or “micro-cracking” or give rise to other possible damage mechanisms.

A preliminary study was conducted where a Picro-sirius Polarisation method was used to stain and view the collagen of normal and damaged cartilage-bone samples (Appendix C). Samples from the *in vitro* validation and damage studies in Chapter 5 were sectioned, stained with Picro-sirius red stain for collagen staining, and imaged using polarised light so the alignments of collagen fibres and therefore the overall structure of the collagen matrix could be revealed. However, the Picro-sirius Polarisation method was sufficient in showing the structure of the tightly packed collagen fibres in the superficial layer of cartilage, but not the thin and loosely packed collagen fibres in the mid or deep layers. This method was, therefore, not suitable for investigating the damage mechanism of cartilage delamination. Other imaging methods such as scanning electron microscopy (SEM) may possibly be suitable since it has previously been used to image all layers of the collagen structure in the articular cartilage of humans and other

mammalian species [162, 165].

8.2.2 Characterisation of Cam and Hip Joint Anatomy for Clinical Diagnosis and Further Parameter Studies

Characterisation of the cam deformity has not ventured very far from a radiographic measurement of its alpha angle despite the fact that the validity of an alpha angle as an assessment tool for the presence of a cam deformity or cam-type FAI has been much debated [27, 37]. Chapter 3 of this thesis showed that it is essential to measure and characterise an individual's hip joint clearance, cam location, cam protrusion, and cam coverage or width to understand in greater depth when impingement might occur, why it occurs, and help determine if the individual is susceptible to cam-type FAI based on their subject-specific activity level and hip joint geometry. Since the measurement of an alpha angle, does not provide clear information about the listed geometric hip joint parameters, it can be concluded that the clinical standard of taking a single measurement of a person's alpha angle in an AP or cross-table lateral radiograph is an insufficient measurement for determining the presence of a cam deformity that may cause or lead to cam-type impingement. The three geometric parameters studied in Chapter 6 of this thesis including hip joint clearance, cam protrusion, and cam width are examples of only three possible parameters that could provide valuable diagnostic insight into cam-type FAI. More sophisticated ways for characterising the cam and hip joint anatomy would allow for more realistic parameter studies to be conducted and also greatly improve current clinical diagnostic standards.

A unique method for characterising the cam deformity has been developed which involves the segmentation of raw DICOM data to determine the distance from the cam lesion's surface points to the centroid of the femoral head in order to calculate mean bump height, volume, and location [45]. Another method, which may possibly be used for characterisation of not only the cam deformity, but of the femoral and acetabular structures as well, consists of statistical shape modelling (SSM) which enables the comparison of 3D geometries without the need for assuming ideal geometry [47]. It is incredibly important to characterise the entire anatomy of the femoral head and acetabulum as well as the cam since these structures play a significant role in the resulting hip joint contact and shear strains that develop during hip joint loading and motion. Therefore, methods such as SSM are especially promising in the field of characterisation of

the anatomy of the cam and the hip joint anatomies. Finally, FE simulations and parameter studies which focus on the cam-type FAI damage mechanisms and the likelihood of damage due to more sophisticated characterisations such as SSM could bring great insight into the implications of such characterisations parameters. Understanding these implications would be required for such methods to be considered for use in the clinical diagnosis of cam-type FAI.

8.2.3 Further Investigations into Surgical Simulations using FE

The surgical parameter study reported in Chapter 7 presented three variations of how the femoral head could be reshaped to treat cam-type FAI. More sophisticated surgical simulations can be conducted in 2D or 3D using FE models which simulate how subject-specific hip joints have actually been reshaped surgically. Such a study would involve 3D imaging of the hip joint of a cam-type FAI patient prior to surgery and after surgery. In this study, the pre-surgery FE model shear strain results due to various activities could be compared to the subject-specific pathology observed during surgery to gain insight into the damage mechanism of their specific cam deformity and possible relationship with certain activities. To advance this study, subject-specific kinematic data prior to surgery and after surgery could also be acquired and applied to the pre- and post- FE models for a highly accurate representation of their cam-type FAI mechanism prior to surgery and in depth strain analysis of their surgical treatment results.

Another interesting study would be the use of FE models to investigate the optimisation of reshaping surgery for shear strain reduction during extreme or everyday activities. This could potentially serve as a pre-surgical planning tool. It could also possibly transform current surgical treatment by providing patients with a personalised treatment plan where pre-planning optimisation studies of their subject-specific hip joint anatomy can be made to focus on their specific activity level or desired activity level.

8.3 Conclusion

This thesis has provided new insights into the mechanism by which cam-type FAI can lead to cartilage delamination and labral tearing using a validated FE modelling approach. The mechanism by which cartilage can delaminate, involving repetitive shearing and gradual compression, has been verified through *in vitro* damage experiments. Geometric parameters have

been identified which help determine cam deformities which would likely cause OA. Finally, simulations of reshaping surgery have given insight into ways in which a cam can be reshaped which would likely lead to improved outcomes and avoid creating conditions which might allow the delamination of articular cartilage within the acetabulum to initiate or progress.

Thesis Word Count: 39,923

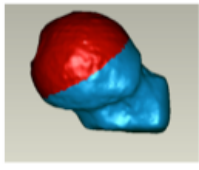
Appendix A

Appendix 1

A.1 Intra- and Inter-Observer Study Participant Instructions

Fitting a Sphere on the Femoral Head Intra- and Inter-observer Study Instructions for Participants

Please following the instructions below for fitting a sphere on the femoral head of a normal subject and cam-type impingement patient:

1. Open Norm4-FH.wrl file
2. Click on "Create Features" 
3. Select Type: Sphere 
4. Select Method: Best Fit 
5. Check "Contact Feature"
6. Select femoral head surfaces by clicking and dragging mouse (selected surfaces will turn red). Keep selecting until all surfaces you think should be a part of the spherical part of the femoral head has been selected (hold control as you click and drag to deselect surfaces). 
7. Press Apply to create Sphere
8. Adjust the sphere's centering around the femoral head by adjusting the selection of the red surfaces or by directly adjusting the x, y, and z components of the centre.
9. Adjust the sphere's diameter for better fit.
10. Record the centre of the sphere and the diameter in the provided file.
11. Click "Apply", then "Next", then "OK" to exit
12. Save file
13. Please repeat for Patient1-FH.wrl

Thank you for your participation!

Figure A.1: The instructions provided to participants in the intra- and inter-observer study for fitting a sphere on the femoral head of a hip joint are shown.

A.2 Proximity Results

Proximity Between the Acetabulum and Femoral Head- Impingement at 0 deg Flexion
Normal Subjects Patients

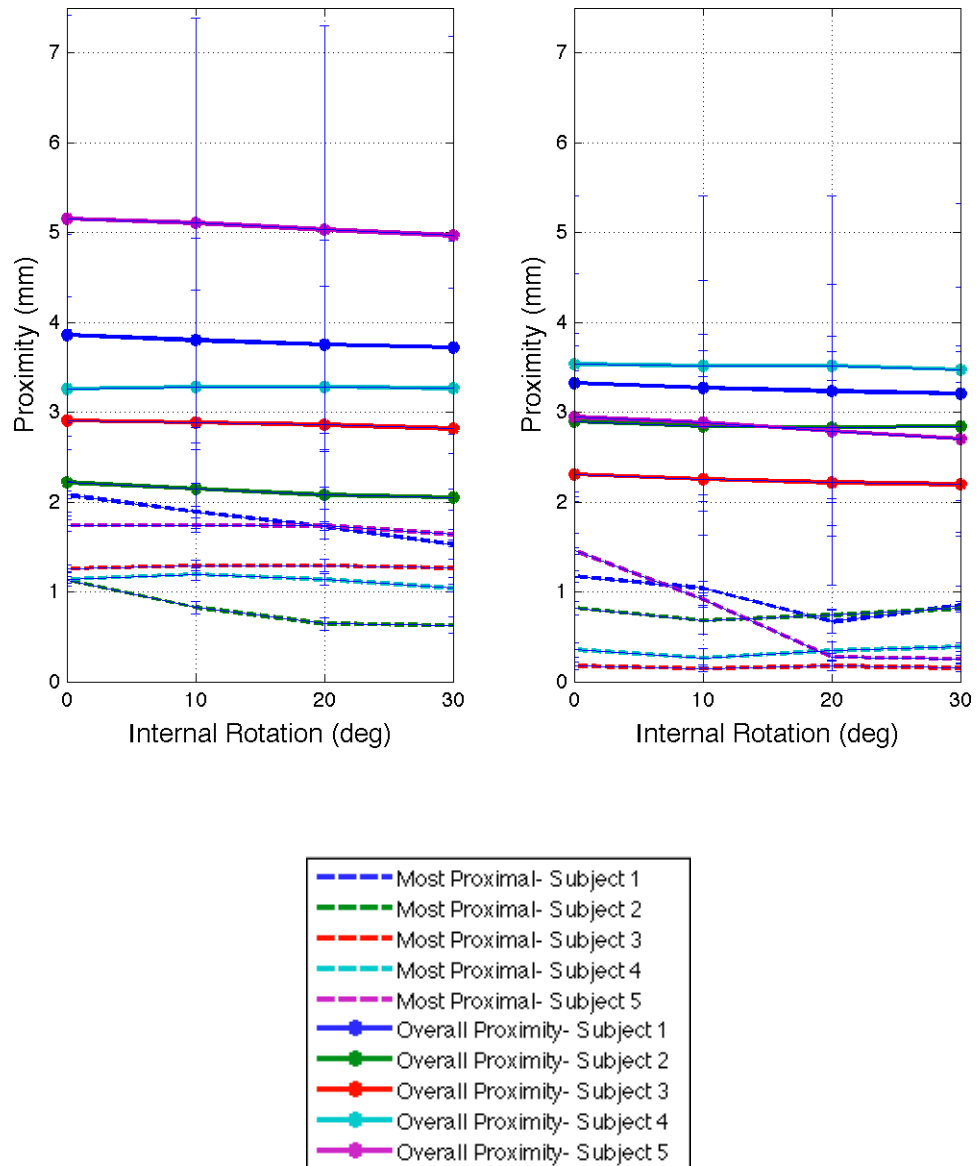


Figure A.2: The individual proximity results for Normal Subjects 1 to 5 and Cam-type FAI Patients 1 to 5 are presented for internal rotation at 0° flexion and 20° adduction; most proximal results are an average of the ten most proximal points; overall proximity is considered between the bony surfaces of the acetabulum and femoral head. Impingement did not occur.

Proximity Between the Acetabulum and Femoral Head- Impingement at 30 deg Flexion Normal Subjects Patients

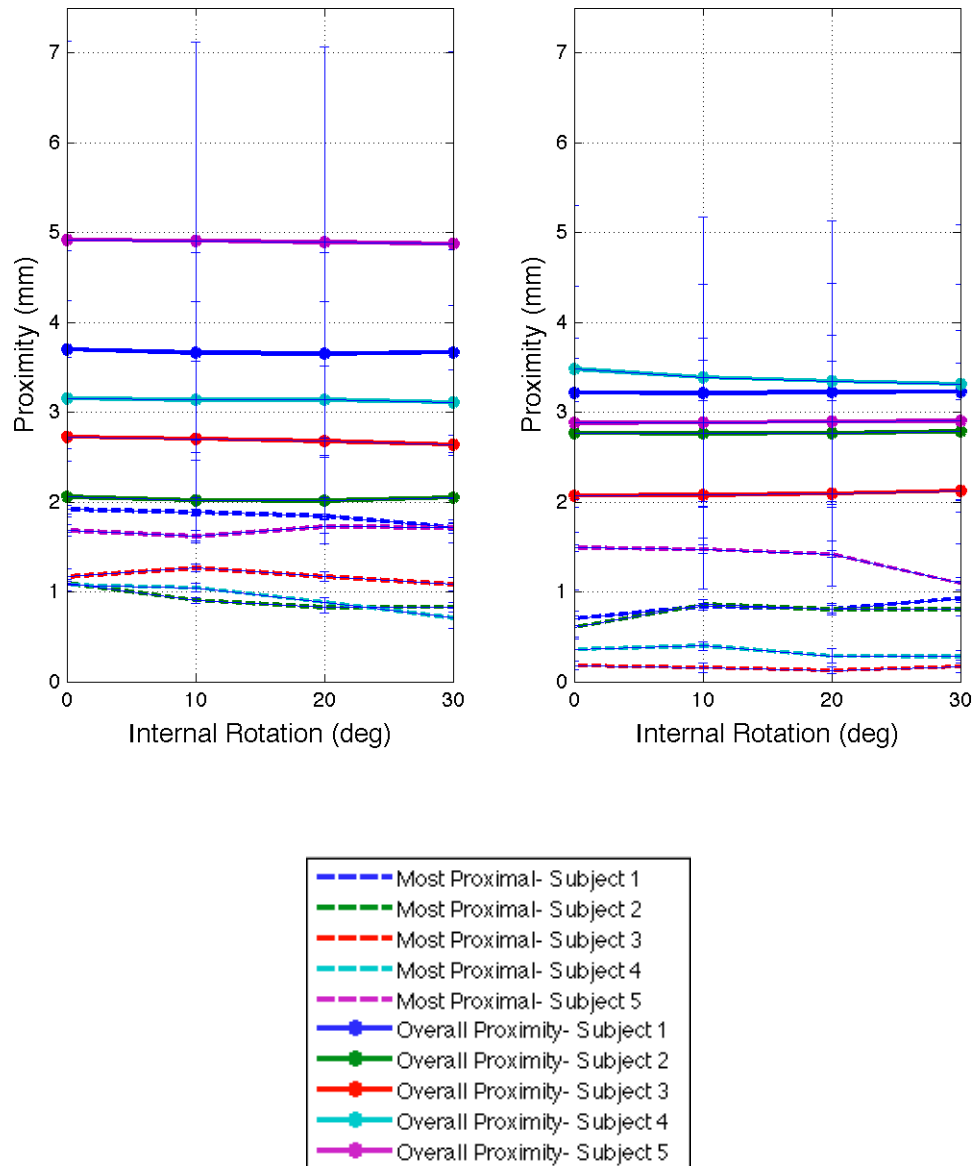


Figure A.3: The individual proximity results for Normal Subjects 1 to 5 and Cam-type FAI Patients 1 to 5 are presented for internal rotation at 30° flexion and 20° adduction; most proximal results are an average of the ten most proximal points; overall proximity is considered between the bony surfaces of the acetabulum and femoral head. Impingement did not occur.

Proximity Between the Acetabulum and Femoral Head- Impingement at 60 deg Flexion Normal Subjects

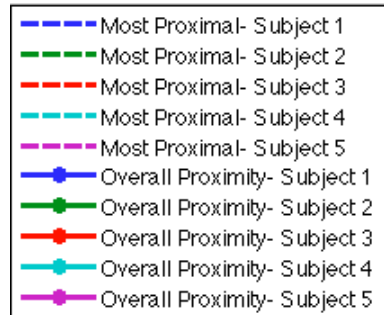
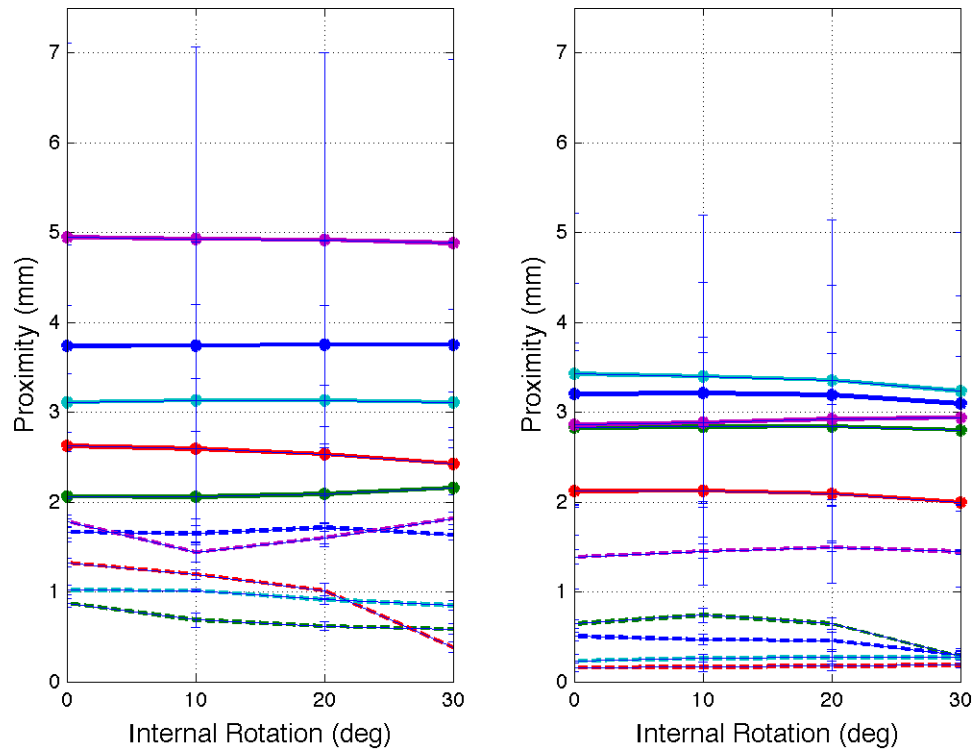


Figure A.4: The individual proximity results for Normal Subjects 1 to 5 and Cam-type FAI Patients 1 to 5 are presented for internal rotation at 60° flexion and 20° adduction; most proximal results are an average of the ten most proximal points; overall proximity is considered between the bony surfaces of the acetabulum and femoral head. Impingement did not occur.

Proximity of Bony Surfaces Between the Acetabulum and Femoral Head- Standing

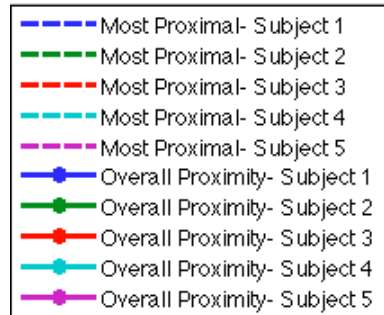
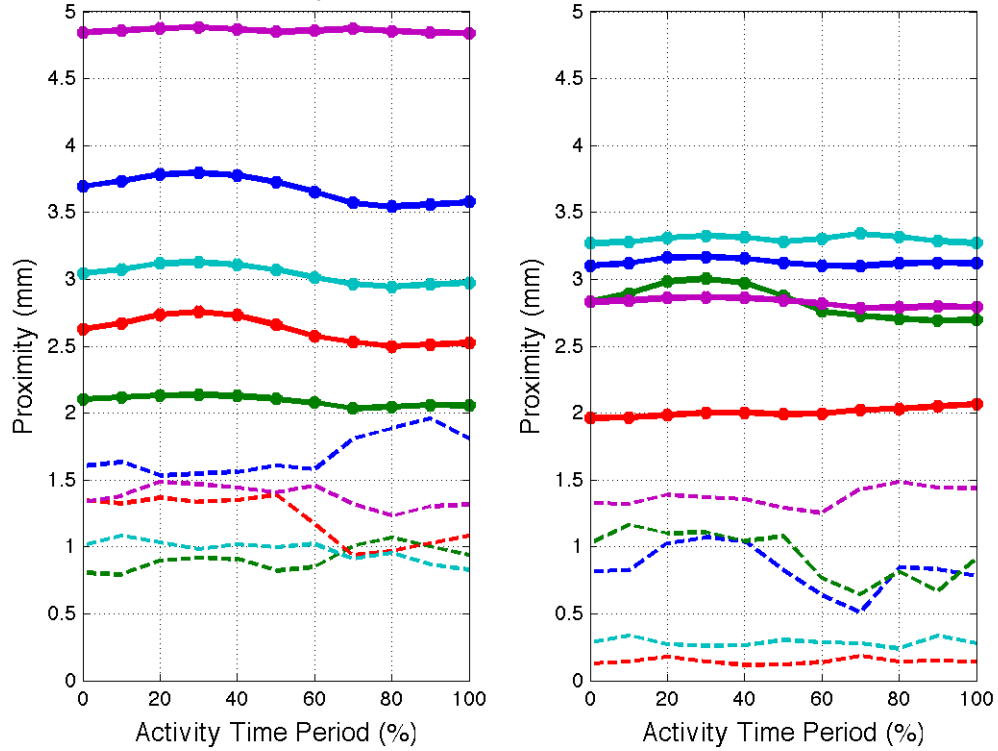


Figure A.6: The individual proximity results for Normal Subjects 1 to 5 and cam-type FAI Patients 1 to 5 are presented for the functional activity of standing; most proximal results are an average of the ten most proximal points; overall proximity is considered between the bony surfaces of the acetabulum and femoral head. Impingement did not occur.

Proximity of Bony Surfaces Between the Acetabulum and Femoral Head- Walking

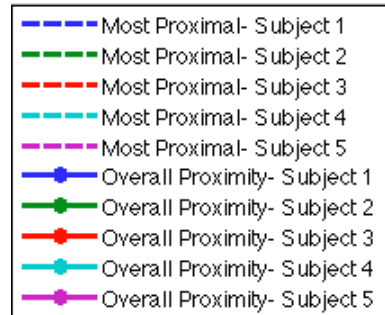
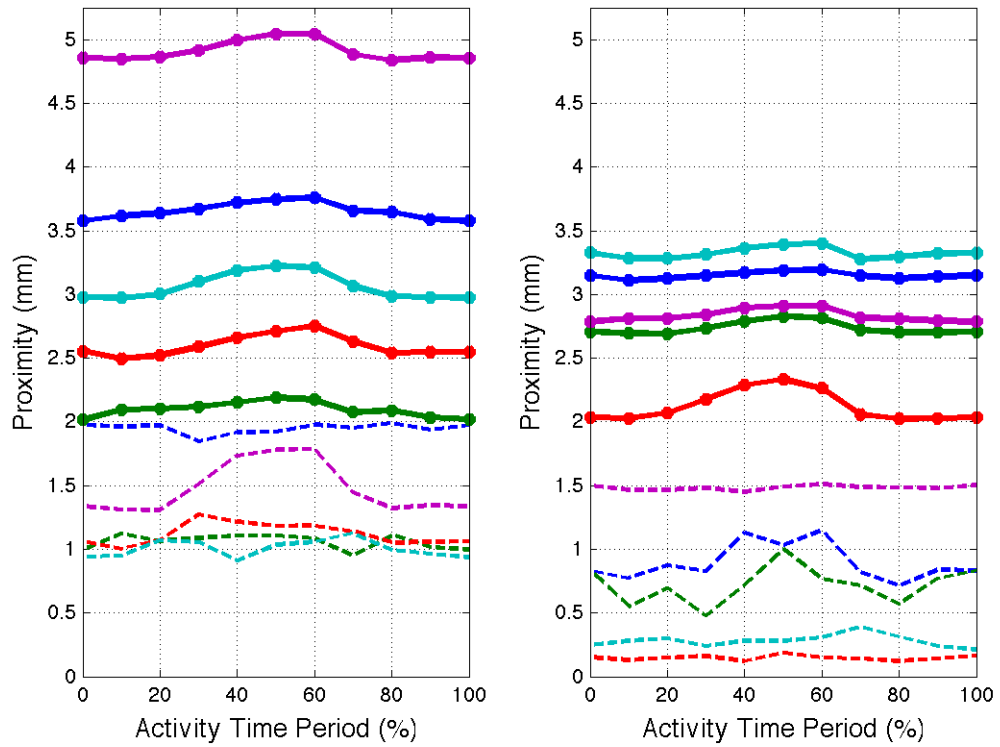


Figure A.7: The individual proximity results for Normal Subjects 1 to 5 and cam-type FAI Patients 1 to 5 are presented for the functional activity of walking; most proximal results are an average of the ten most proximal points; overall proximity is considered between the bony surfaces of the acetabulum and femoral head. Impingement did not occur.

Appendix B

Appendix 2

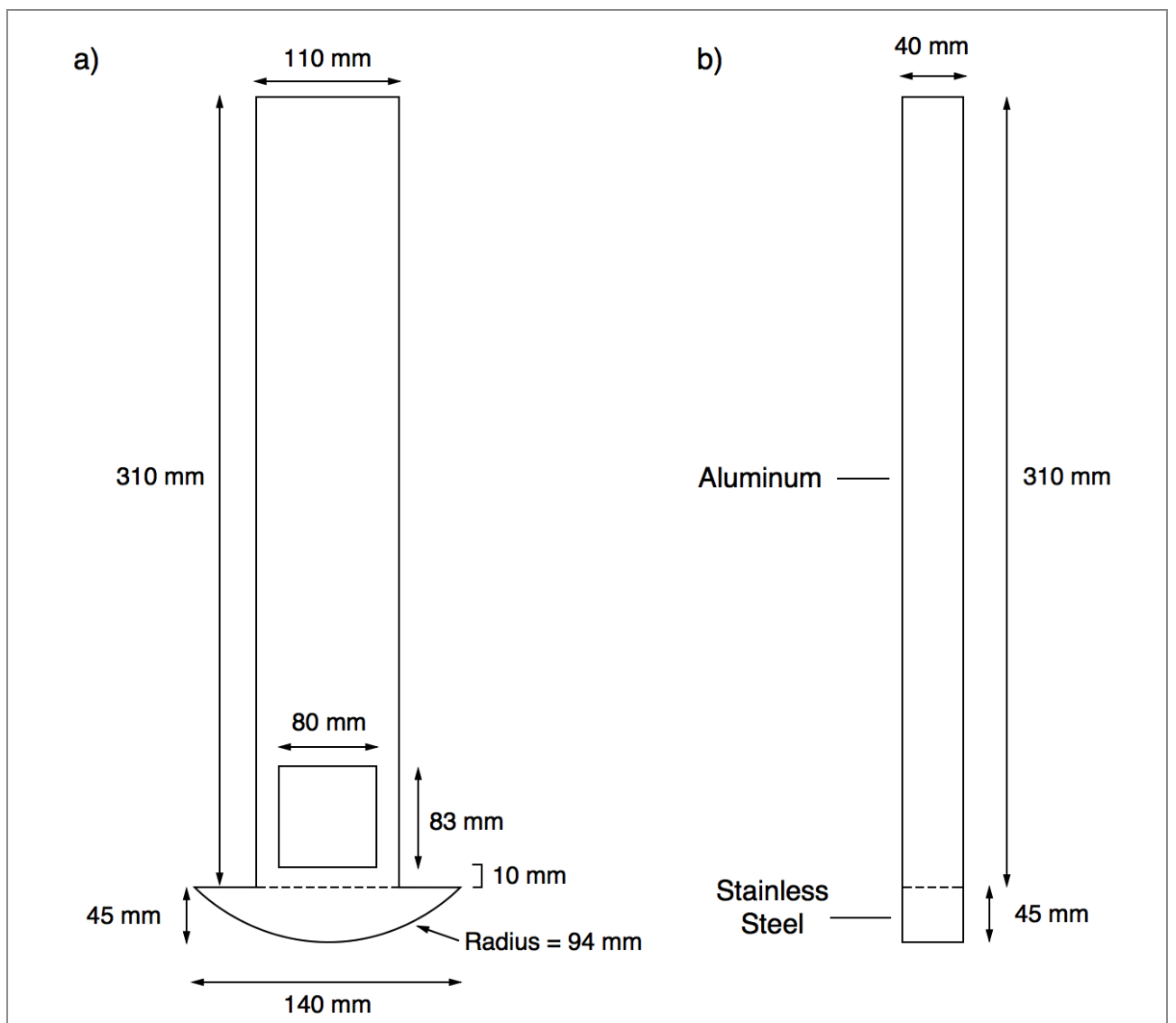


Figure B.1: The CAD drawing of the indenter used in the validation experiments in Chapter 5 is presented from the front view (a) and lateral view (b). The indenter comprised of an aluminum body which attached to the stepper motor of the WASIM apparatus (Figure 5.4) and a curved stainless steel end piece which formed the indenting surface.

Appendix C

Appendix 3

C.1 Preliminary Histology and Imaging Study of Cartilage Damage

C.1.1 Methods

Sample Preparation for Histology

5 mm-diameter cores were cut out of one normal and one delaminated cartilage-bone sample. The delaminated sample was Sample 3 from the set of samples used in the damage experiments. The cores consisted of cartilage, bone, and the cartilage-bone interface. To make sectioning of the samples easier, the samples were left in 8% formic acid for two nights to decalcify the bone. The samples were fit into embedding cassettes, paraffin-embedding was performed, and the samples were cut to 4 to 5 μm slices using a microtome (Leica Microsystems, Wetzlar, Germany).

Staining

Normal and damaged cartilage samples were stained using Mayer's Hematoxylin and eosin (H&E) stain for viewing of the overall tissue structure [166, 167]. Picro-sirius red stain was used for staining of collagen in a normal cartilage sample to investigate if it would be able to detect the collagen structure at the cartilage-bone interface.¹

¹Picro-sirius red staining involves the use of Picric acid, an explosive material, so it was done in collaboration with Ali Shirazi who was experienced in Picro-sirius red staining.

The following protocol was used for Picro-sirius red staining [168, 169, 170]:

1. Deparaffinise/hydrate using series of washes:

- 2x 5 min Xylene
 - 5 min Dirty xylene (1)
 - 5 min Clean xylene (2)
- 2x 2 min 100% Ethanol
 - 2 min Ethanol (1)
 - 2 min Ethanol (2)
- 2x 2 min 90% Ethanol
 - 2 min Ethanol (1)
 - 2 min Ethanol (2)
- 2x 2 min 70% Ethanol
 - 2 min Ethanol (1)
 - 2 min Ethanol (1)
- 2x 5 min distilled water
 - 5 min distilled water (1)
 - 5 min distilled water (2)
- Outline with PAP pen
- 10 min Haematoxylin
- 10 min tap water

2. Stained for 1 hr in Picro-sirius red solution

3. 10 dips of water (0.5% GAA)

4. Quickly dehydrate slides

- 20 dips in 70% Ethanol
- 20 dips in 90% Ethanol
- 20 dips in 100% Ethanol

5. Mount slides onto DPX

6. Proceed to imasine

Imaging

Samples stained with H&E were viewed under a light microscope (BX40F4, Olympus, Tokyo, Japan). The sample stained with Picro-sirius red was viewed under a light microscope with polarised light. Picro-sirius red stain, when viewed under polarised light, can allow for different types of collagen (Type I, II, and III) to be differentiated based on colour. Bundles of thick collagen fibres (2 to 10 μm) [171] are turned red, and as the thickness of collagen fibres decrease, they are turned orange, yellow, and finally green for very thin fibres (0.5 to 1.5 μm) [171] [169, 171, 172, 173].

C.1.2 Histology and Imaging Results

The normal cartilage samples stained with Picro-sirius red and H&E stains and the delaminated cartilage sample stained with H&E are presented in figure C.1. The Picro-sirius red stain coloured the entire thickness of the cartilage sample red. The superficial layer and cartilage-bone interface were coloured a darker red. The H&E stain coloured the entire thickness of the cartilage shades of pink and red, while the nuclei of the cells were coloured black. In the delaminated sample, it was not clear if some portions of the seemingly attached interface were still attached, about to be detached, or were already detached but not separated. Also, some parts of the cartilage slice had folded over on itself, making those areas a much darker red.

The imaging of the sample stained with Picro-sirius red under polarised light is presented in figure C.2. The sample was rotated on a stage so that variations in the birefringence in the tissue could be shown. A similar birefringence occurred when the sample was rotated every 180°. At certain angles, the collagen at the superficial layer of the cartilage was clearly coloured bright yellow. At certain angles, the mid layers of cartilage were coloured light green and shades of less bright yellow. There was no clear differentiation of colour in the collagen in the deeper layers of the cartilage.

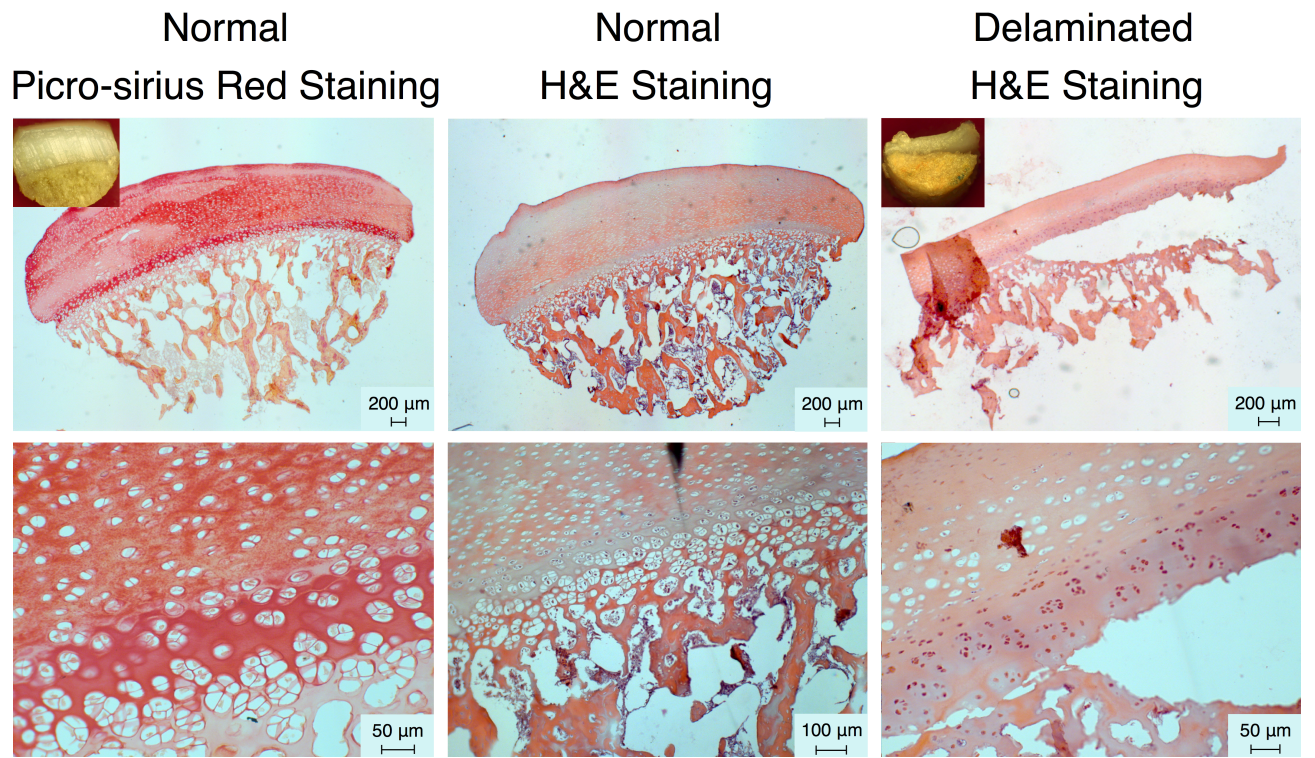


Figure C.1: H&E and Picro-sirius red staining of normal and damaged cartilage-bone samples are presented.

Picro-sirius Red Staining and Polarised Microscopy

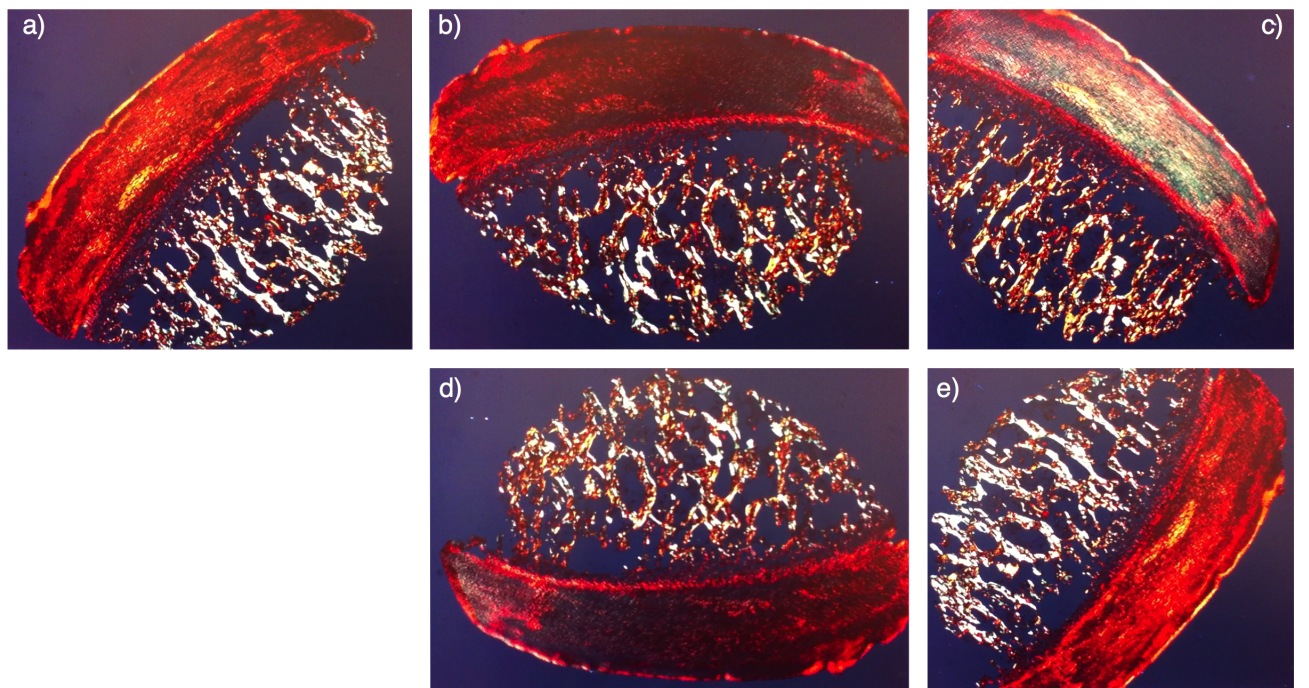


Figure C.2: Images of normal cartilage-bone samples which were stained with Picro-sirius red stain and viewed under polarised light are presented. Images a) to e) represent the sample oriented at different angles to produce variations in birefringence in the cartilage tissue.

C.1.3 Discussion and Conclusion

The Picro-sirius Red staining in combination with applying polarised light was able to show the structure of the collagen fibres in the superficial layer but not in the mid and deep layers of cartilage. This was probably because the strength of the birefringence of collagen increases for thicker fibres and for more tightly packed fibres [174, 175], however, the mid and deep layers of articular cartilage are made up of collagen Type II which is much thinner in diameter compared to collagen Type I and III. Also, the collagen fibres are less packed in the mid and deep layers compared to the superficial layer. Collagen Type II collagen has been demonstrated to be more difficult to image compared to Collagen Type I and III in other studies as well [172, 176].

Picro-sirius Red staining when viewed under polarised light was sufficient in showing the structure of the tightly packed collagen fibres in the superficial layer of cartilage, but not the thin and loosely packed collagen fibres in the mid or deep layers. Since staining using Picro-sirius and polarised microscopy was not successful in detecting the collagen structure in the deep layer of cartilage near the cartilage-bone interface, further histological experiments were not conducted on damaged samples of different damage modes.

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