Effect of Acetabular Labral Tears, Repair and Resection on Hip Cartilage Strains: A 7T MR Study

by

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Abstract

Acetabular labral tears are associated with hip osteoarthritis. A current surgical treatment strategy for a torn labrum, labral resection, has recently shown poor patient outcomes with radiographic signs of osteoarthritis two-years post-operation. Since mechanical factors play a role in the etiology of osteoarthritis, identifying the mechanical role of the labrum may enhance current surgical treatment strategies.

In this pilot study, we assessed the relationship between mean cartilage strain, maximum cartilage strain and the three-dimensional cartilage strain distribution in six human cadaver hips with various pathologic conditions of the labrum. We developed a novel technique of mapping cartilage strain using quantitative magnetic resonance imaging (qMRI). qMRI provides a non-invasive means of quantifying the cartilage strain distribution in the hip in three-dimensions. Each specimen was assessed first with an intact labrum, then after surgically simulating a longitudinal peripheral labral tear, then after arthroscopically repairing the tear, and after labral resection. We validated the precision of the technique through use of an additional specimen which served as a control.

To minimize motion artifact in the high-resolution MR images, we determined that 225 minutes was required for cartilage to reach a steady-state thickness under load. We also determined 16.5 hours was required for cartilage to recover to a steady-state unloaded thickness.

The difference in mean and maximum cartilage strain when the labrum was repaired and resected was assessed using a paired t-test. We found that the resected group had an increased mean and maximum cartilage strain of 4% and 6%, respectively and the 3D cartilage strain distribution was elevated throughout the region of interest. When the condition of the intact labrum was compared to the torn labrum, we found no change in mean and maximum cartilage strain, and little obvious change in the 3D pattern of cartilage strain distribution.

Based on our findings of increased cartilage strain after labral resection when compared to labral repair, we hypothesize that the labrum’s contribution of additional surface area assists in load distribution, which spares cartilage from excessive loads. We therefore recommend that
the longitudinal peripheral torn labrum should not be resected if it is possible to be repaired, because *in vivo*, labral resection may create an environment with increased articular cartilage strain, which is thought to be associated with cartilage degeneration.
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Co-authorship Statement

Sections of this thesis have been submitted as multi-authored papers in peer-reviewed journals. Details of the authors’ contributions are provided.

Submitted Manuscripts


Author’s contribution: Laura Greaves was responsible for conduction of the experiments, analysis and presentation of the findings, and writing and editing of the original paper. David Wilson was the key editor on this paper and provided the original ideas behind the paper. Michael Gilbart conducted the surgical procedures, stimulated discussion and provided editorial assistance. Andrew Yung and Piotr Kozlowski designed the test methods for magnetic resonance imaging and provided editorial assistance.


Author’s contribution: Laura Greaves was responsible for conduction of the experiments, analysis and presentation of the findings, and writing and editing of the original paper. David Wilson was the key editor on this paper and provided the original ideas behind the paper. Michael Gilbart conducted the arthroscopic screening of the specimens, stimulated discussion and provided editorial assistance. Andrew Yung and Piotr Kozlowski designed the test methods for magnetic resonance imaging and provided editorial assistance.

Abstracts


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1.1 Introduction

Osteoarthritis (OA) is a disease characterized by progressive cartilage damage and sclerosis of the underlying subchondral bone within a joint\textsuperscript{100}. OA is one of the most prevalent disorders in the population\textsuperscript{101}. Approximately 10\% of Canadians have osteoarthritis\textsuperscript{1}. Symptoms of osteoarthritis include deep, aching joint pain with mild joint stiffness. The pain increases after use of the affected joint, however long periods of inactivity often result in increased joint stiffness. The patient finds difficulty executing basic activities of daily living, leisure activities or work related activities\textsuperscript{32}. The burden of OA relates not only to its rising prevalence, but also to the cost of the disease to the health care system. In Canada, the total direct and indirect costs of musculoskeletal diseases in 1993 were in excess of $20 billion, exceeding those of cancer ($13 billion) or cardiovascular problems ($19.8 billion)\textsuperscript{78}. Further, the average annual costs incurred by each patient with disabling hip OA due to time lost from employment are $12,200\textsuperscript{41}. With over 12,500 patients undergoing hip replacements per year and the prevalence of hip replacements in Canada continuously rising\textsuperscript{76}, joint preserving surgical strategies are of increasing interest as a means to assure a superior patient quality of life and toward minimizing costs to Canada’s health care system.

It is widely accepted that mechanical factors play a role in the etiology of osteoarthritis. The necessary stimulus to initiate osteoarthritis can be provided by changes in the mechanical environment of the joint\textsuperscript{71, 72, 97,105,15}. The etiology of osteoarthritis of the hip has long been described as either primary (eg, abnormality of cartilage or subchondral bone of undetermined origin) or secondary (eg, associated with injury or with congenital or developmental
However, the term ‘primary hip osteoarthritis’ is diminishing in use because studies have demonstrated that primary osteoarthritis is often secondary to subtle developmental abnormalities such as acetabular dysplasia or femoroacetabular impingement. Patients with acetabular labral pathology often present with symptoms of pain, and mechanical symptoms such as locking in the hip joint at the extreme ranges of motion. Recently, a relationship has been documented between OA and the morphology of the acetabular labrum. In one of the largest retrospective clinical studies, reviewing 436 hip arthroscopies, 261 patients (55%) had labral tears, of which 73% had associated chondral damage. One-half of all labral tear patients had a serious articular cartilage lesion, while only one quarter of patients without a labral tear had a serious lesion. It was concluded that there is a two-fold increase in the relative risk of chondral degeneration in the presence of labral lesions.

Currently surgeons alleviate the pain associated with a torn labrum with labral repair or labral resection procedures. Labral repair involves suturing the torn portion of the labrum back to the acetabular rim, while labral resection involves removing the torn portion of the labrum from the joint. Clinicians report successful case studies where pain was resolved in their patients promptly following labral resection and labral repair, however the one study that evaluated outcomes two-years post operation found patients treated with a labral repair recovered earlier and had superior clinical and radiographic results when compared with patients who had undergone labral resection.
Because of the association of labral tears with OA, recent research studies have focused on determining the mechanical role of the acetabular labrum. The labrum has been shown to increase the surface area in contact with the femoral head by 21%, thereby distributing the loads more evenly in the hip joint, sparing cartilage from excessive loads\textsuperscript{113}. Further, the labrum has been shown to maintain the intra-articular pressure in the hip joint, which plays an important role in minimizing wear to the cartilage and supporting load, sparing the cartilage matrix from excessive loads\textsuperscript{33, 34}.

Quantifying the three-dimensional load distribution in cartilage, rather than quantifying the average load, is important because symptoms and eventual joint degeneration may be related to localized load features. Cartilage degeneration occurs not only over maximum weight bearing regions, but also over areas that are subjected to little or no load\textsuperscript{27, 121}. It is accepted that loading, which ensures that an appropriate balance of nutrients is delivered to the cartilage, is required to maintain cartilage health.

Recent biomechanical studies have been limited because they did not provide a three-dimensional assessment of load distribution in the hip or because of the destructive nature of the chosen biomechanical test instrumentation. Pressure sensitive film\textsuperscript{2, 111, 118} or an array of transducers\textsuperscript{14} are limited because insertion of the film or transducers requires destruction of the joint capsule, which may change joint mechanics. Measurement of hydraulic actuator displacement in a cadaveric hip mounted in a servo-hydraulic materials testing system\textsuperscript{34} provides information of overall joint consolidation rather than a three-dimensional (localized) assessment of joint consolidation. Radiographic techniques\textsuperscript{6, 75} and intra-articular pressure measurement\textsuperscript{34} while the joint is subjected to load also do not provide three-dimensional
assessment of the loading environment. Instrumented prostheses require destruction of the joint capsule.

Using high field MRI to map three-dimensional cartilage strain in response to load is appealing because the method is non-invasive, allowing repeated measurements of the same joint subjected to different loading conditions, and because the method provides information on the distribution of load across the joint. This method has been used before in the hip, however the imaging resolution possible in vivo was not sufficient to detect a consistent change in cartilage thickness with load. This method has been used in the knee joint in an ex vivo study that successfully showed how meniscectomy altered the pattern of strain in tibial articular cartilage.

Further knowledge of the mechanical role of the labrum would be beneficial toward improving current treatment strategies. Specifically, it is not clear how a labral tear affects the three-dimensional cartilage strain distribution in the hip. Further, it is not clear how surgical treatment of a torn labrum such as a labral repair or resection affects the three-dimensional cartilage strain distribution in the hip.

To use quantitative magnetic resonance imaging (qMRI) for strain mapping in repeated measures of the same specimens, the time for cartilage to reach steady-state thickness under load, and the time required to recover thickness once load has been removed, must be known. These time parameters can be used to minimize motion artifact during the lengthy MRI scan while gaining a high resolution strain map of hip cartilage.
Our objectives were:

1. To develop and validate a hip loading device that applies a physiologic compressive load to human cadaveric hip specimens within an MRI scanner.

2. To assess the time required for cartilage to reach a steady-state thickness under continuous physiological loading, and after removing the load.

3. To assess the effect of a torn, repaired and resected labrum on the three-dimensional distribution of hip cartilage strain.
1.2 The Labrum

1.2.1 Hip Joint Anatomy

The hip joint is a ball and socket synovial articulation comprised of the spherical femoral head and cup-like acetabulum of the pelvic bone, both of which are covered with articular hyaline cartilage. The femoral head has an area which is not covered by cartilage called the fovea. The fovea is the site of attachment for the ligamentum teres, which attaches directly from the head of the femur to the acetabulum. The articular cartilage covers the acetabulum in a characteristic horseshoe type pattern, where the central and inferior regions of the acetabulum are left uncovered by cartilage. The rim of the acetabulum is raised slightly by the fibrocartilaginous labrum, which will later be described in section 1.2.2.

The hip joint is encased by a fibrous capsule that attaches to the base of the femoral neck and extends beyond the periphery of the acetabulum. The hip joint is reinforced by three main ligaments: iliofemoral ligament, pubofemoral ligament and the ischialfemoral ligament.

Arterial blood supply to the hip originates from the medial and lateral circumflex arteries. The head of the femur is also supplied by a branch of the obturator artery. The hip joint is innervated by the femoral, obturator, superior gluteal and quadratus femoris nerves.

Many muscles contribute to motion about the hip joint including flexion, extension, lateral rotation, medial rotation, abduction and adduction. The muscles flexing the femur at the hip joint include the psoas major, iliacus, rectus femoris, sartorius and pectineus. The muscles extending the femur include the gluteus maximus, biceps femoris, semitendinosus and
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The muscles involved in lateral rotation include the piriformis, obturator internus, gemellus superior and inferior, quadratus femoris and the biceps femoris.25 Muscles providing medial rotation include the adductor longus, adductor magnus, semitendinosus and semimembranosus.25 The muscles abducting the femur at the hip joint include the piriformis, obturator internus, gemellus superior and inferior, gluteus minimus and medius.25 The muscles providing adduction include the gracilis, pectineus, adductor longus, adductor brevis and adductor magnus.25

1.2.2 Labrum Anatomy

The labrum is a fibrocartilaginous structure that is firmly attached to the acetabular rim. Inferiorly, the labrum bridges across the acetabulum as the transverse acetabular ligament.25 On the non-articulating side of the joint, the labrum attaches directly to the acetabulum (Figure 1-1A). On the articulating side of the joint, the labrum attaches to the acetabular cartilage through a transition zone of calcified cartilage (Figure 1-1A). The labrum is triangular in cross-section, with its base attached to the acetabulum and its apex forming the free edge of the labrum which articulates with the femoral head. Labra with regions of a rectangular cross-section have also been observed.120 The labrum is thickest (Figure 1-1B) in the superior region of the hip joint, and widest in the anterior region.104, 113
The labrum is composed of hydrated tissue, with collagen fiber bundles oriented predominantly in the circumferential direction, aligned with the acetabular rim\textsuperscript{91}. Within the labrum, three distinct layers exist (Figure 1-2). Along the acetabular rim, the collagen fibrillar network within the labrum is random in orientation and composed of chondrocytes, creating a fibrocartilaginous layer\textsuperscript{91}. The adjacent layer is composed of a lamellar layer of collagen fibrils, intersecting at various angles\textsuperscript{91}. The most external surface of the labrum, comprising the majority of the labral substance, is composed of circumferentially oriented collagen fibrils, which blend with the transverse ligament in the acetabulum\textsuperscript{91}. 

Figure 1-2 Scanning electron microscopy images of various layers of the labrum. (Left) Random fibrillar network located adjacent to the acetabulum. (Center) Lamellae of collagen fibrils located adjacent to acetabular layer. (Right) Circumferential orientation of collagen fibrils in the most external layer of the labrum. [With kind permission from Springer Science + Business Media. Archives of Orthopaedic and Trauma Surgery, Vol 123, 6, 2003, page 285, Petersen et al., Fig 2b-d]
1.2.3 The Role of the Labrum

The role of the labrum has not been thoroughly investigated until recently. In addition to providing proprioceptive feedback\textsuperscript{57}, most researchers advocate that a mechanical role of the labrum exists, and that the labrum is integral to the preservation of cartilage health through maintenance of a pressurized intra-articular fluid in the joint\textsuperscript{34} and the contribution of additional surface area\textsuperscript{113}.

1.2.3.1 Labrum and Intra-Articular Pressure

An early study noted a lower pressure reading outside the labrum (inside the joint capsule) compared to measurements inside the labrum (in the acetabular fossa)\textsuperscript{112}. A more recent study supported this finding\textsuperscript{34}. Cadaveric human hip joints were subjected to axial loads and greater intra-articular pressure was recorded in the fat pad of the acetabular fossa when the labrum was present, compared to when it was completely removed (Figure 1-3)\textsuperscript{34}. These findings suggest that the labrum acts like a seal, maintaining intra-articular pressure in the joint.

Figure 1-3 Fluid pressure measurements in the fat pad of the acetabulum after application of a 570 N step load, with and without the labrum\textsuperscript{34}. [Reprinted from the Journal of Biomechanics, Vol 36:2, Ferguson et al., An in vitro investigation of the acetabular labral seal in hip joint mechanics, p.174, 2003, with permission from Elsevier]
The maintenance of intra-articular pressure in the hip joint is thought to play an important role in cartilage preservation through reducing wear. Adhesive, shear-induced wear due to solid-on-solid cartilage contact has been suggested as a friction-induced and heat-induced cause of cartilage degeneration. Using a pendulum apparatus, one study showed that when synovial fluid was removed in cadaveric hip joints, the coefficient of friction was higher compared to when the fluid was present. A maximum change of quadruple the coefficient of friction when the synovial fluid was removed, compared to when it was present, was measured when low (<500 N) compressive loads were applied to the joint. Temperature in the subchondral bone of cadaver human hips was measured and a rise in temperature of 2.5°C was measured when simulating walking. With the measured increase in the coefficient of friction in specimens without synovial fluid, fundamentally there is likely an even higher temperature rise in the joint, which, if greater than 4°C, will decrease protein synthesis in chondrocytes.

The maintenance of fluid pressure in the hip joint also plays an important role in supporting hip loads. Cartilage is composed of 68-85% water by weight and cartilage layers deform predominantly through changes in tissue volume occurring through fluid expression. The presence of a pressurized fluid layer in between the acetabular and femoral cartilage is vital to cartilage health. Fourteen bovine cartilage samples were tested in a confined compression chamber and interstitial fluid pressure was measured after stress was applied to the sample. It was found that the fluid supported approximately 90% of the load for durations as long as 400 seconds after load application. The load supported by the interstitial fluid is thought to spare the cartilage matrix from excessive loading conditions, which may otherwise cause unphysiologically large strains in the cartilage. This “load partitioning” role of fluid pressurization has been highlighted through observations of the natural topography of
cartilage using ultrasound techniques. When the hip joint is loaded, opposing cartilage surfaces naturally maintain surface undulations (approximately 75 microns in depth)\textsuperscript{69}. The presence of such undulations is thought to create a high-resistance pathway of fluid which spares the cartilage matrix from excessive loads\textsuperscript{69}. As illustrated in Figure 1-3, when the labrum was removed and the joint was loaded, the interstitial pressure decreased relative to when the labrum was present and therefore more load was transferred to the cartilage when the labrum was removed.

A finite element analysis has been used to demonstrate the capability of the labrum to seal a layer of pressurized fluid under a 1200 N compressive load\textsuperscript{33}. In the absence of the labrum, the distribution of fluid pressure within the cartilage layers was less uniform and had higher peak pressures, resulting in higher strains within the cartilage matrix (20\% versus 3\%)\textsuperscript{33}. In this finite element model, the load carried by the fluid after 1000 seconds of loading was 94\% with the labrum intact and 92\% without the labrum. After 10,000 seconds of loading, the load carried by the fluid was 80\% with the labrum intact and 73\% without\textsuperscript{33}. Stresses in the cartilage matrix were up to 64\% higher after 1000 seconds of loading and up to 92\% higher after 10,000 seconds of loading in the absence of the labrum\textsuperscript{33}. These findings suggest failure of the labral seal subjects the solid matrix of the cartilage to excessive loads. This may eventually lead to cracks and fissures in the cartilage\textsuperscript{79}.

It has also been suggested that the presence of the labrum attached to the acetabulum protects the subchondral bone beneath the labrum from the synovial fluid acting under cyclic pressure which has been proposed to lead to the separation of articular cartilage from the underlying bone\textsuperscript{74}. 

1.2.3.2 Labrum and Surface Area

The labrum deepens the acetabular socket by 21%, providing an additional 8 cm² of articulating surface area with the femoral head. The labrum’s role in deepening the acetabular socket is perhaps best highlighted by the natural tendency of the labrum to be larger in hips with insufficient acetabular coverage.

The additional surface area provided by the labrum plays a role in constraining the femoral head in the acetabular socket. During extreme ranges of motion, the additional surface area constrains the femoral head as evidenced by an increase in the range of motion of the hip with a torn labrum when in external rotation and abduction relative to the intact condition. Furthermore, the force required to distract the femur has been found to be 60% less after tearing the labrum relative to the intact condition.

Not only does the labrum constrain the femoral head through providing additional surface area, but a healthy labrum also prevents capsular laxity which can cause instability. Supporting the link between labral tears and capsular laxity is a clinical study that reported labral tears in 100% (12 of 12) of patients with capsular laxity. A torn labrum is thought to precede capsular laxity as researchers have theorized that the torn portion of the labrum can push upward, stretching the joint capsule from the inside. A subtle instability may then develop, causing capsular laxity.

The increased surface area provided by the labrum in articulation with the femoral head plays an important role in cartilage preservation. This increase in surface area likely distributes the loads more evenly across the joint, sparing cartilage from excessive loads. The contribution
of the labrum to the surface area in the hip may also provide a greater protective effect of the interstitial fluid, where an increased load is carried by the fluid\(^7\). Further, the constraint of the femoral head provided by the labrum at extreme ranges of motion protects the hip capsule and reduces the potential for abnormal load distribution in the joint due to subtle subluxation
\(^{19,96}\).
1.2.4 Symptoms and Types of Labral Tears

The overall incidence of labral tearing in the general population is currently unknown. Indirect incidence levels are reported through clinician case reports. Clinicians have reported 22% - 87% of patients presenting with groin pain as being diagnosed with a labral tear. Labral tears occur over a wide age range of the population, with patients ranging in age from 14 to 72 years. Cadaveric studies have shown 96% (53 of 55, mean age 78 years) and 61% (33 of 54, mean age 53 years) of a randomly selected population to have labral tears.

Patients with an acetabular labral tear complain of pain, which is almost always (<90%) localized in the anterior hip or groin region. The classical mechanical symptoms associated with a labral tear are clicking, locking or giving way. The clicking sensation in the hip has been reported as the most diagnostic symptom with 100% sensitivity and 85% specificity. However painful catching or clicking does not always indicate the presence of a labral tear, and it is not always necessarily present in those with labral damage. Hence great caution is often advised in the interpretation of mechanical symptoms. Classical symptoms of hip clicking may be misdiagnosed as a labral tear when the cause is from a different source such as a snapping iliobibial tendon or a hypermobile psoas tendon.

A slight decrease in the range of motion of the hip in flexion, rotation, abduction and adduction has also been reported in some patients diagnosed with a torn labrum.

One study collected findings based on 10 subjects arthroscopically diagnosed and treated as acetabular labral tear. Findings on physical examination in these labral tear patients included:

a) Pain when internally rotating the hip while flexed 90 degrees (found in 7 of 10 patients)
b) Pain when axially compressing the hip while flexed 90 degrees and slightly adducted (found in 10 of 10 patients)

c) Tenderness posterior to the greater trochanter (found in 8 of 10 patients)

Based on arthroscopic observations of the morphology of a labral tear in over 350 patients, a classification system has been developed (Figure 1-4)\(^62\). A radial flap tear involves the disruption of the free margin of the labrum with the formation of a discrete flap. This type of tear was present in 57% of patients\(^62\). A radial fibrillated tear has a generally hairy appearance at the free margin of the labrum. These tears were present in 22% of patients, and more common in those with degenerated cartilage\(^62\). Longitudinal peripheral tears, present in 16% of patients, are located along the acetabular insertion of the labrum and the tears are of a variable length\(^62\). Unstable tears (not illustrated) are partially dislocated tears and were present in 5.4% of patients\(^62\).

Findings in 78 hip arthroscopies report 10% of patients having a 10 mm or less tear size, 67% with a 10-20 mm tear size and 23% having a tear size greater than 2 cm\(^90\).
1.2.5 Etiology and Epidemiology of Labral Tears

There are four known causes of acetabular labral tears: trauma, femoroacetabular impingement, dysplasia and degeneration.

1.2.5.1 Labral Tears Caused By Trauma

It is understood that labral tears are caused from incidents of trauma, however trauma is not the only cause. In a study of 55 patients, approximately 50% of patients with labral tears had experienced trauma to the lower extremity. However in a different study of 267 patients, only 19% had experienced trauma. Such incidents of trauma often involve relatively minor events. Published case reports have shown a torn labrum from events such as running, sprinting, rugby, twisting, falling, basketball, a long car trip, or minor automobile accidents. However, case reports from one clinician stated that only 30% of patients remembered the provoking events of the hip symptoms. Because the events tearing the labrum seem sometimes fairly innocuous, the labrum is thought to be a vulnerable structure of the hip.

The specific movements in these relatively innocuous events of trauma have been investigated by a recent study that measured strains in the labrum during controlled testing maneuvers. With specimens in 30° flexion and with application of a 100-lb axial load in combination with a 177 in-lb external rotation torque, strains in the anterior bone-labrum interface of up to 13.6% were found. The authors speculate that during this tested maneuver, hip motion is limited by the tension in the hip capsule, which causes the center of rotation of the joint to shift.
posteriorly, subjecting the anterior labrum to higher strains which through overloading or repetitive loading may lead to separation of the labrum from the chondrolabral junction\(^{19}\).

### 1.2.5.2 Labral Tears Caused By FAI

Femoral acetabular impingement (FAI) is another known cause of labral tears\(^{51, 52, 53, 54, 80, 90, 94}\). It is suggested that due to the reduced anterior femoral head-neck offset present in FAI, a dynamic conflict occurs (impingement) between the femoral head-neck junction and the acetabular rim. This event induces compressive and shear forces within the anterosuperior acetabular rim during flexion and internal rotation. Because the labrum is vulnerable to shear forces\(^{108}\), this load environment leads to labral tearing\(^{65}\). The specific labral tear types associated with FAI have been identified, with peripheral longitudinal tears occurring in patients with cam impingement (abnormality of femoral head), and radial fibrillated tears or ossified labral tears in patients with pincer impingement (abnormality of the acetabulum)\(^{10}\).

### 1.2.5.3 Labral Tears Caused by DDH

Developmental dysplasia of the hip (DDH) is another known cause of labral tears\(^{24, 43, 59}\). The deficient acetabular coverage of the femoral head in DDH leads to instability. It is hypothesized that the resulting anterolateral migration of the femoral head then induces abnormal shear stresses at the labrum\(^{65}\). The characteristic enlarged labrum associated with hip dysplasia\(^{49}\) is thought to initially aid in maintaining the femoral head’s position in the acetabulum, however if the chronic shear stress persists, the labrum will fail by tearing from the acetabular rim, sometimes occurring with a bony fragment\(^{59, 65}\).
1.2.5.4 Labral Tears Caused By Aging

Previous studies report a tear of the labrum as an acquired condition that is highly prevalent in aging adult hips with labral tears present in 61% (33 of 54) to 96% (53 or 55) of cadaver hips\textsuperscript{104, 120}. Because tears are often seen in the absence of all the aforementioned causes of labral tearing (FAI, DDH, trauma), there is general consensus that these labral tears are a part of a degenerative process associated with aging\textsuperscript{3, 16, 45, 62, 63, 66, 93, 114}. Using scanning and transmission electron-microscopes, the density of the collagen fibrils of the labrum was found to decrease with age along with an increased disruption and irregular ridges on the uppermost superficial layer of the labrum\textsuperscript{114}. These morphological findings may be correlated with a decrease in the mechanical properties of the labrum, although labral tensile strength and age were not found to be correlated\textsuperscript{51}. Perhaps the effect of age and mechanical properties with labra tested in compression, such as compressive stiffness, would add to our understanding.

1.2.5.5 Location of Labral Tears

The reported frequency of labral tearing in the anterior, superior or anterosuperior region of the hip ranges from 70% to 100%, and there is a broad consensus in published studies that the anterosuperior region of the labrum is the most vulnerable to tearing\textsuperscript{11, 20, 21, 35, 62, 74, 93, 104, 113}. With the anterosuperior region of the hip known as the weight-bearing region\textsuperscript{2}, labral tears are likely to be more frequent in this region due to the higher mechanical demands\textsuperscript{19}. Posterior and lateral labral fraying have been associated with anterior tearing, and it is thought that these lesions may be secondary lesions related to abnormal joint motion caused by the anterior tear\textsuperscript{92}. It is thought that the locations of labral tears caused by trauma depend on the loading environment of the precipitating event, with anterosuperior labral tears originating
from twisting injuries, and posterior labral tears originating from axial loading of a flexed hip, such as striking the dashboard with the knees during a frontal automotive collision\textsuperscript{18,37}.

1.2.5.6 Types of Labral Tears Commonly Observed

It is not clear from the literature what type of tear is most commonly found, likely due to the limits in study size and the natural variance in tear type with the activity level of the study population. The most common labral lesion has been reported to be the longitudinal peripheral tear involving the separation of the labrum from the articular surface at the labrocartilage junction\textsuperscript{74,104}. While longitudinal peripheral tears were observed, another study reports the most common type of labral tear is a radial flap tear, involving the disruption of the free margin of the labrum with the consequent formation of a discrete flap\textsuperscript{62}. 
1.2.6 Healing Capacity of the Labrum

The arteries of the hip joint capsule are derived from the obturator, the deep branch of the superior gluteal artery, and the inferior gluteal artery\textsuperscript{74}.

The microvasculature of the human hip labrum and its supporting structures has been investigated in the literature\textsuperscript{54, 91, 104}. Generally, the labrum is divided into the capsular side and the articulating side (Figure 1-5), and within these zones, the areas closest to the bony acetabulum, and furthest from the acetabulum (Figure 1-5). The acetabular labrum is nourished by the well-vascularized joint capsule (Figure 1-6) and as such, the vascularity is greatest at the capsular side of the labrum, and throughout one third of the distance toward the articulating side\textsuperscript{91, 104, 54}. These capsular vessels do not penetrate deeply, leaving the articular two-thirds of the labrum avascular, with little potential to heal\textsuperscript{54, 91}. Although vessels have been identified within the adjacent acetabular rim, researchers report infrequent\textsuperscript{54} or no\textsuperscript{74} penetration of these vessels into the substance of the labrum. Within the capsular and articulating sides of the labrum, the distribution of vascularity is similar in all regions (anterior, superior, posterior, inferior), which was a surprising finding to researchers that speculated avascularity was a cause of the frequent tearing in the anterosuperior region of the joint\textsuperscript{54}.

Though clinicians have speculated whether the labrum has potential to repair\textsuperscript{35, 74, 92}. An \textit{in vivo} animal study examining the capability of the labrum to heal after a labral resection showed promising results with replacement of the defect with dense fibrous scar tissue in 89\% of sheep hips (16 of 18)\textsuperscript{77}. This new tissue approximated the original labrum in density, shape, size and location in proximity to the femoral surface\textsuperscript{77}. The subchondral bone remodeled with ingrowth of new blood vessels\textsuperscript{77}. 
Based on these observations, the healing capacity of the labrum is thought to be specific to the type of tear. Longitudinal peripheral tears of the labrum have the potential to heal, with vascular contributions from both the hip capsule and the acetabulum (Figure 1-6). In contrast, intra-substance labral tearing, such as radial flap tears, do not have the potential to heal in the articular two-thirds of the labrum, as this region is avascular. With knowledge of this avascularity in the articulating portion of the labrum, clinicians have sought to provoke bleeding of cancellous bone to promote healing and obtain a stable labral refixation.
Chapter 1: Introduction

1.3 Osteoarthritis

The epidemiology and etiology of osteoarthritis, as well as evidence supporting the association of labral pathology and osteoarthritis are presented.

1.3.1 Epidemiology and Etiology of Osteoarthritis

Osteoarthritis (OA) is a disease characterized by progressive cartilage damage and sclerosis of the underlying subchondral bone within a joint. The disease most commonly affects the hands and weight bearing joints such as the knee and hip. Symptoms of osteoarthritis include deep, aching joint pain with mild joint stiffness. The pain increases after use of the affected joint, however long periods of inactivity often result in increased joint stiffness. The patient finds difficulty executing activities of daily living, either for basic tasks, leisure activities or work related activities.

Approximately 10% of Canadians have osteoarthritis. The burden of OA relates not only to its rising prevalence, but also to the cost of the disease to the health care system. In Canada, the total direct and indirect costs of musculoskeletal diseases in 1993 were in excess of $20 billion, exceeding those of cancer ($13 billion) or cardiovascular problems ($19.8 billion). Further, the annual costs incurred to individuals with disabling hip OA due to time lost from employment and leisure are $12,200.

It has been proposed that mechanical factors play a role in the etiology of osteoarthritis. The necessary stimulus to initiate osteoarthritis can be provided by changes in the mechanical environment of the joint.
Debate is ongoing regarding the exact sequence of events leading to osteoarthritis. One theory suggests that sclerosis of the subchondral bone is the initiating factor, where excessive load causes trabecular microfractures causing a bone healing response in the form of osteophyte creation\textsuperscript{98, 99, 39}. These osteophytes then produce stiffness gradients in the subchondral bone that create stress concentrations and shear the overlying cartilage, leading to general cartilage degeneration\textsuperscript{27, 100}. Another theory suggests that cartilage degeneration is the initiating event. It has been shown that cartilage degeneration occurs not only over heavily loaded regions in the joint, but also in areas that are only subjected to minimal load\textsuperscript{27, 121}. The hypothesis suggests that to maintain cartilage health, nutrients must be delivered to the tissue through regular loading. After loading beyond a threshold, the ability of the cartilage to absorb the vital nutrients is hindered and the cartilage begins to weaken and degenerate. The resulting thinner cartilage has less ability to attenuate forces, which leads to subchondral bone sclerosis\textsuperscript{98}. Regardless of whether bone sclerosis or cartilage degeneration are the initiating factor leading to osteoarthritis, the foundation for both theories is that excessive loading is the root cause of osteoarthritis.

1.3.2 Labral Tears and Osteoarthritis

Labral pathology has been related to changes consistent with osteoarthritis in a large number of clinical studies\textsuperscript{3, 17, 24, 35, 38, 44, 45, 62, 73, 74, 88, 90, 116, cadaver studies\textsuperscript{16, 63, 104}, and animal models\textsuperscript{77}}. In one of the largest retrospective clinical studies, reviewing 436 hip arthroscopies, 261 patients (55\%) had labral tears, of which 73\% had associated chondral damage\textsuperscript{74}. With one-half of all labral tear patients having a serious articular lesion, compared to only one quarter of
patients without an actual tear who had serious lesions, it was concluded that there is a two-fold increase in the relative risk of chondral degeneration in the presence of labral lesions\textsuperscript{74}.

The spatial relation between labral and cartilage lesions was documented in a clinical study examining 18 hemiarthroplasty patients\textsuperscript{63}. Labral lesions and associated cartilage degeneration were consistently found within close proximity of one another, potentially indicative of a direct cause-and-effect relationship. Both labral tears\textsuperscript{11,20, 21, 35, 62, 74, 93, 104, 113} and signs of osteoarthritis\textsuperscript{88} are most frequently found in the anterosuperior region of the hip.

Some studies have specifically attempted to identify the sequence of events relating labral pathology with osteoarthritis\textsuperscript{37,59,74,88}. Based on findings from 120 hip arthroscopies, it was found that most prearthritic hips and almost all early-stage, advanced-stage and end-stage osteoarthritic hips showed detaching tears of the acetabular labrum (Figure 1-7)\textsuperscript{88}. Further, in all five hips showing no apparent cartilage degeneration, there was a detaching tear of the labrum. The authors conclude that labral tears precede osteoarthritis\textsuperscript{88}.

![Figure 1-7 Percentages of detaching labral tears in the anterosuperior (AS), superior (SU) and posterosuperior (PS) quadrants of the hip. Most prearthritic hips and almost all early-stage, advanced-stage and end-stage hips showed detaching tears of the labrum. [Reprinted from Arthroscopy: The Journal of Arthroscopic and Related Surgery, Vol 15:5, Noguchi Y. et al., Cartilage and Labrum Degeneration in the Dysplastic Hip Generally Originates in the Anterosuperior Weight-Bearing Area: An Arthroscopic Observation, p.503,1999, with permission from Elsevier]
One study observed a five-fold increase in findings of labral tears in conjunction with osteoarthritis in elderly patients compared to young patients\textsuperscript{74}. The authors conclude labral tears are a precursor to osteoarthritis. Although it may be argued that this finding is indicative of aging, the authors deem this unlikely given the apparent lag between the incidence of labral tears and chondral degeneration\textsuperscript{74}.

While the aforementioned observations among young and old populations are interesting, future longitudinal studies are necessary to confirm the cause-and-effect relationship between labral tears and osteoarthritis. However, because of the strength of evidence from many studies\textsuperscript{3, 17, 24, 35, 38, 44, 45, 62, 73, 74, 88, 90, 116}, it appears irrefutable that labral tears are related to osteoarthritis.
1.4 Labral Tear Treatment

Initially, labral tears are usually treated conservatively in an attempt to eliminate the need for surgery. However, arthroscopic treatment is reported to be the most successful approach\cite{67}, and as such will be the focus of this section.

1.4.1 Conservative Treatment

The typical preliminary treatment for labral tears include limited weight bearing, nonsteroidal anti-inflammatory drugs, and sometimes physical therapy\cite{5, 67}. However, often the pain recurs when the patient resumes normal activities because the labrum has little potential to heal without surgical intervention\cite{67}. The specific aims of physical therapy are to reduce anteriorly directed forces on the hip by addressing the patterns of recruitment of muscles controlling hip motion\cite{67}. This is done by correcting abnormal hip motion during gait\cite{67}.

1.4.2 Arthroscopic Treatment

Although some surgeons surgically treat a torn labrum through open techniques\cite{3, 40, 58}, potential risks of an open approach include joint dislocation, infection, deep vein thrombosis, avascular necrosis, major nerve or vessel injury and muscle weakness\cite{4, 22, 42}. Due to these risks associated with open operations, arthroscopic techniques are generally preferred\cite{107}. There are two arthroscopic methods to alleviate pain caused by a labral tear, a labral repair and a labral resection. Because labral tears are often found in conjunction with femoroacetabular impingement, hip dysplasia, capsular laxity and osteoarthritis\cite{24, 29, 53, 59, 64, 80, 90, 94, 96, 104, 120}, generally multiple procedures are performed during a single operation. This section will review the operations specific to alleviating pain from the torn labrum. Patients with persistent hip
pain for longer than 4 weeks, clinical signs, and radiographic findings consistent with a labral tear are considered candidates for hip arthroscopy\textsuperscript{56}.

\subsection*{1.4.2.1 Labral Resection}

Labral resection, also known as labral debridement, involves removing the torn portion of the labrum (Figure 1-8). The surgeon aims to remove all torn tissue and leave as much of the healthy labrum as possible\textsuperscript{56}. A flexible ligament chisel is used to cut the torn part of the labrum, leaving only a small portion attached\textsuperscript{56}. Last, a motorized shaver is used to complete the labral resection and to remove the torn portion of the labrum.

\begin{figure}[h]
\centering
\includegraphics[width=\textwidth]{labral_resection.png}
\caption{Arthroscopic labral resection involves removing the torn portion of the labrum. [Copyright Smith & Nephew Endoscopy. Available from: http://endo.smith-nephew.com/]}
\end{figure}

\subsection*{1.4.2.2 Labral Repair}

With increasing knowledge regarding the function and importance of the labrum, new surgical strategies to repair the labrum have been developed. Many clinicians and researchers advocate preserving as much of the healthy labrum as possible\textsuperscript{30,55,56,95}. Briefly, repairing the labrum involves drilling a small hole (< 3 mm in diameter) in the acetabulum, tapping a suture-anchor in this hole, and suturing the torn portion of the labrum back to the acetabular rim.
(Figure 1-9). Fluoroscopy may be used during the procedure to ensure optimal anchor placement\(^{95}\). The suture may pass around, or through, the torn portion of the labrum\(^{95}\). Depending on the size of the labral tear, a second or third anchor may be necessary\(^{95}\).

Figure 1-9 Arthroscopic labral repair involves drilling a hole in the acetabulum, placing a suture anchor in this hole and securing the torn portion of the labrum back to the acetabular rim. [Copyright Smith & Nephew Endoscopy. Available from: http://endo.smith-nephew.com/]

### 1.4.2.3 Short and Long Term Clinical Outcomes

Many clinicians report successful case studies where pain was resolved in their patients promptly following labral resection\(^3,18,46,50,56\) and labral repair\(^{56,95}\).

Two clinical studies have graded patient satisfaction 1 year and 3.5 years after labral resection\(^{31,103}\). The study with a 1-year follow up period reported findings from 28 patients having a labral resection who were questioned about their pain, mechanical symptoms, general activity level, activities of daily living, work ability and ability to return to sporting activities\(^{31}\). Only 46% of patients reported “good to excellent” results within these aforementioned categories, with better outcomes more likely in those without osteoarthritis\(^{31}\). Interestingly, two patients without signs of osteoarthritis 1 year after the labral resection
underwent total hip arthroplasty at an average of 52 months after surgery\textsuperscript{31}. The study with a 3.5 year follow-up period after labral resection reported subject satisfaction based on pain, range of motion and ability to perform certain predefined activities\textsuperscript{103}. The study found 67\% of patients (39 of 58) were “pleased” with the result of the operation while the remaining 33\% of patients were not satisfied\textsuperscript{103}.

A more recent study reviewed the clinical and radiographic results of 52 patients with a torn labrum. Patients treated with a labral repair recovered earlier and had superior clinical and radiographic results when compared with patients who had undergone labral resection\textsuperscript{30}. Radiographic signs of osteoarthritis were significantly more prevalent in those with a labral resection at one and two years post-operation compared to those with a labral repair (Figure 1-10)\textsuperscript{30}. This is in contrast with findings of a different study\textsuperscript{59}. Interestingly, labral resection has not been shown to lead to osteoarthritis in an ovine model\textsuperscript{77}.

Figure 1-10 Clinical measures of osteoarthritis grade in 52 patients having a labral repair compared to a labral resection\textsuperscript{30}.
1.5 Intra-articular Hip Loading

There are many different ways of quantifying force distribution in the hip. Each method has an associated limitation that must be considered in the study design. The following section provides a brief outline of reported test methods and limitations.

1.5.1 Contemporary Methods To Quantify Hip Loading

1.5.1.1 Pressure Sensitive Film

Early studies attempted to quantify the force distribution in the cadaveric hip joint using an array of miniature transducers placed on the femoral head which required destruction of the joint capsule and the creation of cylindrical wells in the superficial cartilage layer for transducer placement. The development of pressure sensitive film greatly facilitated this process as there was no longer a need to create cylindrical wells in the cartilage, and the film provided a higher resolution.

Using pressure sensitive film in human cadaveric hip joints, the pressure distribution in the hip was quantified during simulated activities such as walking and resulting from a lateral load applied to the greater trochanter. One study aimed to quantify changes in mean and maximum pressure in cadaveric hip joints with and without the labrum using pressure sensitive film. Seventeen hips were loaded in a materials testing system simulating single-leg stance. After removing the labrum, a decrease in the maximum pressure in the posterior region of the acetabulum was observed, but no changes were detected with regard to mean or maximum pressure in the anterior or superior regions of the acetabulum.
One limitation of studies using pressure sensitive film is the change in contact mechanics due to the combined effect of adding a relatively thick and stiff material (the film) in the joint. Pressure sensitive film has a thickness of 0.3 mm\(^{122}\), which is large relative to the thickness of articular cartilage in the knee (typically ranging from 1.69 – 2.55 mm\(^{106}\) and hip (typically ranging from 1.35 – 2.0 mm\(^{106}\)). Further, pressure sensitive film has an average effective elastic modulus of approximately 100 MPa in compression, which is larger by a factor of 100 – 300 compared to that of normal articular cartilage\(^{122}\). Through use of finite element modeling, the combined effect of the film’s thickness, stiffness and the measurement precision of the film (approximately 10%)\(^{122}\), has been shown to change maximum true contact pressures by up to 28%\(^{122}\).

However, studies measuring force distribution in cadaver hips using pressure sensitive film have identified the anterosuperior region of the hip as the weight bearing region\(^2,111\), which is supported by studies using more sophisticated techniques such as MRI and implanted instrumented prostheses to measure hip loads\(^{12,81}\), and further complimented by clinical findings of increased prevalence of osteoarthritis in the anterosuperior region\(^{88}\). Another limitation of using pressure sensitive film, inherent to all biomechanical testing with cadavers, is the challenge of simulating the load produced during an activity. Several parameters are involved in the simulation such as force magnitude, force direction, force duration, and the frequency of loading. Proper specimen hydration protocols must be used and if a reproducibility study is necessary, adequate time for cartilage recovery must be allowed prior to commencing the next test.
1.5.1.2 Instrumented Prostheses

Perhaps the most widely referenced technique to quantify the loading environment in the hip is the use of instrumented prostheses *in vivo*. With this *in vivo* method, the challenge of simulating a loading environment is overcome and the assessment of a variety of loading environments is made possible. Further, provided sufficient joint healing time is allowed after the operation to implant the instrumented prosthesis, the measurements can include full strength muscle-skeletal loading, with representative magnitude and direction of muscle forces. However, the use of instrumented prostheses is limited as the numbers of participants in such studies are generally low, the natural incongruity of the femoral head with the acetabulum is lost, and the cartilage-cartilage interface is replaced by a cartilage-metal interface.

In one study, measurements of hip contact forces using instrumented implants were obtained from four patients. Nine different activities were investigated including slow, normal and fast walking, walking up and down stairs, standing up, sitting down, standing on one leg and knee bending. Measurements were normalized to subject body weight and average peak hip contact forces across all four specimens are shown in Table 1-1.

Peak loads were experienced over a short duration (Figure 1-11), however there is a residual compressive force acting across the hip joint at all times, with average amplitude of approximately one bodyweight. This observation provides evidence that the hip is never fully unloaded during daily activities. Figure 1-14 also shows the breakdown by force components. The largest component of the load is consistently in the axial (vertical) direction (Fy).
Table 1-1 Average peak hip loads across four patients, measured with instrumented prostheses.

<table>
<thead>
<tr>
<th>Activity</th>
<th>Average peak hip contact force (% BW)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Slow walking (3.5 km/h)</td>
<td>242</td>
</tr>
<tr>
<td>Normal walking (3.9 km/h)</td>
<td>238</td>
</tr>
<tr>
<td>Fast walking (5.3 km/h)</td>
<td>250</td>
</tr>
<tr>
<td>Up stairs</td>
<td>251</td>
</tr>
<tr>
<td>Down stairs</td>
<td>260</td>
</tr>
<tr>
<td>Standing up</td>
<td>190</td>
</tr>
<tr>
<td>Sitting down</td>
<td>156</td>
</tr>
<tr>
<td>Standing on 2-1-2 legs</td>
<td>231</td>
</tr>
<tr>
<td>Knee bend</td>
<td>143</td>
</tr>
</tbody>
</table>

Figure 1-11 Contact force F (bold, measured in % body weight) of a typical patient during normal walking (Left) and standing on 2-1-2 legs (Right), force expressed as percentage body weight. Peak force marked with dashed line. [Reprinted from the Journal of Biomechanics, Vol 34; 7, Bergmann et al., Hip Contact Forces and Gait Patterns from Routine Activities, 2001, with permission from Elsevier]
1.5.1.3 Roentgenographic Techniques

Cartilage deformation in cadaveric human hips has been measured using roentgenographic techniques, with images taken throughout joint loading of five times body weight in a testing machine\(^5\). This methodology was limited because cartilage deformation was quantified only in two dimensions, and at only 7-12 sites within the imaging plane\(^6\). Nonetheless, the idea of quantifying cartilage deformation in the hip through measurements of bone displacement is plausible because acetabular and femoral hip cartilage are congruent and in contact under load. With modern technology, specifically micro-computed tomographic (CT) imaging, the limitations of the roentgenographic technique could be addressed. In fact, one study has used 3D CT scanning to estimate the projected load-bearing surface in the hip joint before and after periacetabular osteotomy\(^7\). However, because volunteers were scanned while lying supine (i.e. hip joint relatively unloaded), results were based on projected load-bearing area which may introduce error and the consequential change in three-dimensional force distribution induced by periacetabular osteotomy was not quantified. Roentgenographic techniques are also limited due to the concern of patient safety when imaging using x-rays.

1.5.1.4 Hydraulic Actuator Displacement

Cartilage deformation in the intact and labrum-deficient hip has been measured based on the displacement of a hydraulic actuator in a servo-hydraulic materials testing system\(^34\). This study found that cartilage compressed more rapidly without the labrum, relative to the intact case, for the same load. While this study certainly provides useful information for hypothesis generation, measurements from the servo-hydraulic materials testing system are limited because deformation of the entire joint is quantified without isolating the component due to
cartilage compression. Further, no information is gained on the three-dimensional distribution of hip loading, which is a more sensitive means to detect a change in the loading environment.

1.5.1.5 Intra-Articular Pressure Measurement

Intra-articular pressure in the cadaveric hip joint has been recorded throughout loading using a miniature (1 mm in diameter) pressure transducer placed in the fat pad of the acetabular fossa\textsuperscript{34}. The fat pad of the acetabular fossa is a useful region to measure intra-articular pressure in the hip while subjected to load\textsuperscript{34}. This region is very soft and unable to develop significant levels of solid stress within the tissue, which makes pressure measurements possible\textsuperscript{34}.

While results showed significant differences in pressure when the labrum was removed (Figure 1-3), information on the distribution of load across the joint was not available.

1.5.1.6 Finite Element Modeling

Finite element modeling (FEM) is a powerful research tool because the effect of changing any number of parameters (i.e. aggregate modulus of cartilage, permeability of cartilage, bone geometry, body weight, etc.) can be quantified. As opposed to laboratory measurement techniques, the researcher has a great amount of control when determining test parameters. However, FEM is limited by difficulties in validation because testing the same parameters involved in the computer simulated loading environment in the laboratory is technically challenging.
Chapter 1: Introduction

The compressive stress on the articular surface of the human hip joint has been computed in healthy hips relative to osteoarthritic hips using FEM, with no significant difference found\(^\text{13}\). The role of the labrum was also investigated using FEM, and it was found that the distribution of fluid pressure within the cartilage layers was less uniform and had higher peak pressures when the labrum was removed, as described in section 1.2.3.1.

1.5.1.7 Articular Cartilage Explants

Articular cartilage explants are ideal specimens for basic science studies requiring tightly controlled test parameters. Deformation of articular cartilage explants has been evaluated with a cam and follower assembly\(^\text{9}\), a load cell with microprocessor-based feedback\(^\text{8,89}\), and using ultrasound elastography\(^\text{36}\). However, since the cartilage has been extracted from its anatomical environment, results are altered by the boundary conditions of the experiment\(^\text{119}\), the integrity of the matrix at the edges of the tissue sample\(^\text{28}\), the lack of representation of the dynamic changes in contact area\(^\text{68}\) and the lack of intra-articular pressure that regulates the fluid expression from the cartilage\(^\text{79}\).
1.5.2 Using qMRI to Quantify Hip Loads

Cartilage strain mapping using quantitative magnetic resonance imaging (qMRI) has recently been developed\textsuperscript{85, 86, 110}. This method uses ultra high-resolution MR images of cartilage before, during, and after loading to map cartilage strain as a result of joint loading using the following formula:

\[
\text{Cartilage Strain} = \frac{\text{Unloaded cartilage thickness} - \text{loaded cartilage thickness}}{\text{Unloaded cartilage thickness}} \times 100\%
\]

qMRI has a number of advantages over previous techniques. The method reveals the articulating surface of the cartilage and the boundary to subchondral bone over the entire cartilage surface\textsuperscript{110}. The ability to non-invasively quantify cartilage mechanics is a clear advantage of this method, allowing maintenance of the joint capsule and periarticular supporting structures, and therefore allowing repeated measurements of the same joint under various loading conditions.

To date, only one study has used the qMRI technique to measure cartilage deformation in the hip joint. Six volunteers were scanned with a 1.5T magnet after periods of standing or walking for 1 hour, and after lying supine for 70 minutes. Cartilage thickness was measured, and the greatest change in cartilage thickness between unloaded and loaded cartilage was found, expectedly, in the anterosuperior region\textsuperscript{81}. The overall cartilage thickness change due to load was 0.05 mm (Figure 1-12). However, cartilage thickness did not consistently increase in all subjects after resting (which would be expected), which may be indicative of measurement error due to inadequate image resolution, poorly defined periods of activity or inadequate time to rest. This study measured the overall deformation of hip cartilage, grouping acetabular
and femoral cartilage as a single unit. One limitation was in the choice of acquiring the 1.5 T MRI scan with an image voxel size of 0.78 x 0.78 x 1.6 mm. With the combined thickness of femoral and acetabular cartilage ranging from 2.7 – 4.0 mm\textsuperscript{79,106}, the chosen image voxel size could only detect a 20 - 29% change in hip cartilage thickness. The relative size of the hip joint to the imaging coil used in this study was the cause of the poor imaging resolution, and the authors suggested using a magnet of higher field strength (i.e. 3T) in future studies to achieve a higher imaging resolution\textsuperscript{81}. A second limitation was the defined period of activity where subjects were instructed to remain standing or walking for one hour. It could be that some patients walked vigorously for one hour, while others only remained standing. Therefore, some subjects may have experienced larger hip cartilage deformation than others and hence, some subjects may have experienced more hip cartilage recovery than others. A third limitation was in the instruction for the subjects to lie supine for 70 minutes after a period of activity, which may not have been adequate time for the cartilage to recover. A previous in vivo study used MRI to measure cartilage recovery in the knee after exercise found cartilage was still recovering after 90 minutes of rest\textsuperscript{28}.

![Figure 1-12 Population averaged cartilage thickness map of combined femoral and acetabular cartilage a) post weight-bearing and b) post resting. Colour scale in millimeters.](image-url)

This method has been used successfully in the knee joint in an *ex vivo* study\textsuperscript{110}. Using high-field MRI (4.7T), the strain distribution in cadaveric knee specimens was reported with and without the meniscus with an in-plane imaging resolution of 0.16 mm\textsuperscript{110}. Although statistical analysis was not conducted to determine whether load significantly reduced cartilage thickness, strain values up to 72\% (0.68 mm of compression) were reported in the intact knee. When the same knee specimens were scanned again after a meniscectomy, there was a 13\% increase in maximum cartilage deformation and an altered pattern of tibial articular cartilage strain\textsuperscript{110}.

Other studies have examined the unloaded cartilage thickness distribution *in vivo* in patients with normal hips using MRI compared to patients with osteoarthritis or hip dysplasia\textsuperscript{82, 87}. However, there was a lack of standard hip loading protocol prior to imaging the joint, and therefore observed changes in cartilage thickness distribution can be attributed to the variance in hip loading between these patients prior to the MRI scan.

1.5.2.1 Sources of Error Using qMRI

Although qMRI presents several advantages over previous techniques, mainly the ability to non-invasively quantify cartilage mechanics, there are several sources of error that must be carefully considered in the development phase of any study aiming to acquire repeated measurements of cartilage thickness using MRI. Such considerations include, but are not limited to, the orientation of the imaging slice, partial volume effects, imaging resolution and motion artifacts.

If cartilage thickness is measured in two-dimensions, the measurements are highly dependent on the imaging slice orientation. By definition, thickness is the shortest possible measurement...
across the object at the region of interest. It is therefore necessary to select an imaging slice orientation perpendicular to the cartilage surface in the regions of maximum weight bearing, where the largest cartilage deformation is expected\(^1\). Imaging slices oriented perpendicular to the cartilage surface are more sensitive in detecting changes in cartilage thickness with load, because the thickness viewed in the image is the shortest possible measurement. In contrast, if cartilage thickness is measured in three-dimensions, orienting the slice perpendicular to the cartilage surface is less critical. However, if the imaging voxels are anisotropic (i.e. worse imaging resolution through-plane versus in-plane) a perpendicular slice orientation is recommended to reduce error due to the partial-volume effect. In cartilage imaging, partial volume effects arise when a voxel lies over both bone and cartilage, producing a voxel with a lower intensity (appearing gray) compared to a voxel lying completely over cartilage (appearing white). Due to the dependence of cartilage thickness measurements (for 2D measurement) and partial-volume error (for 2D and 3D measurements) on slice orientation, the reproducibility of the qMRI technique is highly dependent on the ability to locate the same imaging slice orientation. Using the qMRI technique in the hip joint is particularly challenging because the curvature of the joint causes most of the imaging slices, except for the central slice, to be subject to partial volume effects\(^2\). With in vitro studies, gadolinium filled fiducial markers are often used for slice location and orientation.

If cartilage deformation is to be measured based on both an unloaded and loaded cartilage image, it is important to ensure the translation and rotation of the joint, induced by the load, only takes place in the image plane; otherwise apparent thickness changes may result from a change in angle between the imaging plane and the cartilage layer\(^4\).
To accurately quantify the thickness of cartilage, which in the hip, ranges in thickness from 2 to 4 mm\textsuperscript{81,106}, a high imaging resolution is of utmost importance. Because each voxel in the MRI image can only be a single intensity, cartilage thickness measurements can be made only within the limits of the imaging resolution. For this reason, high-field MRI is strongly encouraged to obtain a high resolution image, which is essential for depicting cartilage compression\textsuperscript{81}.

High resolution MRI scans require a lengthy acquisition time, where motion artifacts, due to the dynamic nature of cartilage compression, can introduce error. Two options are available to minimize error due to motion artifact. The imaging acquisition time can be reduced at the cost of a lower out-of-plane imaging resolution. Or, a high resolution image can be maintained if the MRI scan is acquired after cartilage has reached a steady-state thickness under load.
1.6 Summary and Direction

➢ It is clear that labral tears are strongly associated with osteoarthritis\textsuperscript{3, 17, 24, 35, 38, 44, 45, 62, 73, 74, 88, 90, 116}, and researchers have proposed that labral tears precede osteoarthritis\textsuperscript{59, 74, 88}. With the labrum contributing a 21% increase in surface area of the hip joint\textsuperscript{113}, thereby distributing loads across the joint, it is likely that labral pathology alters the mechanical environment in the hip, which can predispose the hip to osteoarthritis\textsuperscript{15, 71, 72, 97, 105}. It is not clear how labral tears affect strain distribution in hip cartilage.

➢ Labral repair and labral resection are two commonly used surgical techniques to alleviate pain from a torn labrum. Whether to repair the labrum or resect it depends on the ability of the labrum to heal. Vascularity of the labrum has been shown on the capsular side, and closest to the acetabulum\textsuperscript{54, 77} and therefore peripheral longitudinal tears, present in approximately 16% of cases\textsuperscript{62}, and reported by some as the most frequent type of labral tear\textsuperscript{74, 104} have the potential to heal and therefore can be repaired. Labral tears occur most frequently in the anterosuperior region of the hip joint\textsuperscript{11, 20, 21, 35, 62, 74, 93, 104, 113}. It is not clear how each procedure (repair and resection) affects joint mechanics.

➢ Various methods to measure the loading environment of cartilage have been used previously including pressure sensitive film\textsuperscript{2, 111, 118} or an array of transducers\textsuperscript{14}, hydraulic actuator displacement in a servo-hydraulic materials testing system\textsuperscript{34}, roentgenographic techniques\textsuperscript{6, 75}, intra-articular pressure measurement\textsuperscript{34}, instrumented prostheses\textsuperscript{48, 61, 69, 102}, finite element modeling\textsuperscript{33, 13}, cam and follower assembly\textsuperscript{9}, a load...
cell with microprocessor-based feedback and using ultrasound elastography. Each method has associated limitations. There is currently no method that non-invasively quantifies the three-dimensional force distribution in the joint.

- Quantitative MRI is an attractive tool to measure the strain distribution of natural cartilage because the method is non-invasive, allowing repeated measurements of the same joint subjected to different loading conditions, and the method provides information on the distribution of strain across the joint, from which load distribution can be inferred. This method has been used before in the hip, however the imaging resolution was not sufficient to detect a consistent change in cartilage thickness with load. This method has been used successfully in the knee joint, and tibial cartilage strain was shown to be altered after meniscectomy.

- Labral tears occur over a wide age range of the population, including patients as young as 14 years of age. The current methods available to alleviate pain from a torn labrum may predispose the hip to osteoarthritis within two years after the operation. Total hip arthroplasty is then not a favourable option for these young patients because failure rates in this patient population (<45 years) are high. Joint-preserving surgery with the aim of restoring original joint mechanics, thereby preventing osteoarthritis, is an attractive alternative. Due to the limited methods available to measure contact mechanics at the hip, with the exception of kinematic analyses, no study has biomechanically assessed the change in load environment subjected to the hip joint due to a joint-preserving surgical procedure.
The review of the role and treatment of the acetabular labrum has motivated the generation of the following hypotheses regarding the primary (third) objective:

1. A higher mean and maximum strain will be observed when the labrum is torn compared to intact.

2. A higher mean and maximum strain will be observed when the labrum is resected compared to repaired.
1. Musculoskeletal Disorders such as Arthritis and Osteoporosis. In: Canada NACoAaPHa0, Seniors DoAa, eds; 2005.


Chapter 1: Introduction


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Chapter 2: Deformation and Recovery of Cartilage in the Intact Hip Under Physiological Loads Using 7T MRI

2.1 Introduction

Abnormal joint mechanics are widely believed to be related to pain and cartilage degeneration, and many surgical and conservative treatments for joint conditions seek to restore joint mechanics to normal. Accurate measurement of cartilage deformation in the hip joint could be a valuable tool for applications such as assessing the mechanics associated with cartilage degeneration and evaluating the ability of a particular surgical technique to restore original joint mechanics.

Quantifying force and force distribution at the hip is technically challenging. This is due in part to the joint’s spherical geometry and to the difficulties inherent in inserting transducers without changing joint mechanics. Displacement of a hydraulic actuator in a materials testing system while loading a hip has been measured\(^4\), but this technique is limited because it measures deformation of the entire joint without isolating that of the cartilage, and because it does not assess the distribution of deformation across the joint surface. Roentgenographic techniques have been used to measure cartilage thickness\(^2\), however this measure is based on a two-dimensional projected image. Instrumented prostheses have been used to measure both resultant force and force distribution in the hip\(^6-8\). However, the natural incongruity of the femoral head with the acetabulum is lost and the cartilage-cartilage interface is replaced by a cartilage-metal interface, which limits extrapolation of these results to the natural hip. Pressure

sensitive film has been used *ex vivo*¹ ¹², but it has a limited resolution and interferes substantially with contact mechanics because of its thickness¹⁴.

Cartilage strain mapping using magnetic resonance imaging (MRI) has recently been developed ⁹⁻¹¹ and has a number of advantages over previous techniques. This method uses high-resolution MR images of cartilage to determine its thickness before, during, and after loading and subtracts loaded from unloaded thickness to map cartilage strain. An advantage of this method is that it is non-invasive and allows the joint capsule and periarticular supporting structures to be preserved and repeated measurements of the same joint to be made under various loading conditions. While the method has been used at the knee¹¹, to our knowledge it has not been used to map cartilage strain in the hip.

Experimental design using MR measurements of cartilage strain at the hip requires an understanding of the time required for cartilage to reach equilibrium deformation under constant load and of the time required for cartilage to be restored to full thickness after load is removed. These time parameters must be known to minimize motion artifact in MR images.

Our objective in this study was to answer the following questions:

1. How long does it take for hip cartilage to reach a deformed steady-state cartilage thickness distribution under simulated physiological load, and how much does the cartilage deform?

2. How long does it take for hip cartilage to return to the original cartilage thickness distribution once the load is removed?
2.2 Methods

2.2.1 The Pneumatic Hip Loading Device

We designed and built a device (Figure 2-1) to provide a constant load to human cadaveric hip specimens in a 7.0 T MRI scanner (Bruker Biospin, Ettlingen, Germany). The pneumatic components operating the loading device consisted of a compressor (FP209400RB, Campbell Hausfeld, Harrison OH, USA), an in-line air filter (07F34AC, Parker, Cleveland OH, USA), a 1-way flow valve, a precision regulator (035501040, Parker, Cleveland OH, USA), a digital pressure gauge (2798k22, McMaster Carr, Atlanta GA, USA), a solenoid valve and a custom double-acting pneumatic cylinder (BECO Manufacturing, Laguna Hills, CA). The precision regulator had a repeatability of ±0.14 kPa and was capable of regulating the pressure between 14 and 1034 kPa. The pneumatic piston was made of polypropylene, a material compatible with MRI. All other components were placed outside the magnetic field of the scanner and connected to the loading device by silicone tubing (see Appendix B for design drawings).

The loading device (Figure 2-1) was made to fit in the bore of the MRI scanner. The device was made of an acrylic tube of 5.5” outer diameter with a ¾” wall thickness (Appendix B). The pneumatic cylinder was fixed to the removable end of the acrylic tube. The specimen was mounted to the free end of the pneumatic cylinder and was axially compressed toward the fixed end of the loading device.
The loading device was designed to apply approximately 1980 N of axial compression corresponding to a load of 2.3 times body weight when in single-leg stance, for a body mass of 88 kg.

2.2.2 Cadaveric Human Hip Application

Five human cadaver hip joints from five donors (aged 24-48 years, mean 41.5 years, 1 male, 4 female) were used for the deformation study, and one specimen was also used in the recovery study (38 year old female). Institutional review board approval was obtained for this project (Appendix A). Specimens were dissected of all superficial soft tissues, leaving the capsule intact. Specimens were kept moist at all times using a prepared solution of phosphate buffered saline with protease inhibitors (PBS+) to prevent cartilage degradation or swelling (see Appendix G). Arthroscopic verification confirmed specimens did not have any pre-existing pathology, specifically cartilage degeneration or tears in the acetabular labrum greater than 1 cm in length.
A 3 cm long capsular tear was created in each specimen due to the requirements of a subsequent study.

The pelvis was aligned anatomically with the anterior superior iliac spine and pubic tubercle in the same coronal plane. Using lasers, the femur was potted in a physiological alignment simulating single-leg stance. Next, a laser level was used to guide an axial cut at the level of the anterior inferior iliac spine. The specimen was wrapped in gauze soaked in PBS+ solution to keep the joint moist throughout loading. The specimen was then placed in the loading device and an axial load of 1980 N was applied.

MR imaging was performed with a 3D gradient echo FLASH sequence using a 15 cm inner diameter quadrature volume coil on a 7T MRI system (Bruker Biospin, Ettlingen Germany). Imaging parameters are detailed in Table 2-1, voxel size was 0.11 x 0.11 x 1 mm. Imaging slices were orthogonal to a plane running parallel to the acetabular rim and passed through the longitudinal axis of the anterior inferior iliac spine, AIIS (Figure 2-2).

<table>
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<tr>
<td>Imaging time</td>
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</table>
Chapter 2: Cartilage Deformation and Recovery

Figure 2-2 An imaging slice used in the current study. The imaging slice was orthogonal to a plane that runs parallel to the rim of the acetabulum (white line). This line passed through the longitudinal axis of the AIIS.

Images were obtained prior to loading and at 15 minute intervals after loading. The pixel intensities of each image (after the first) in the center imaging slice were subtracted from the pixel intensities of the preceding image in the center imaging slice at each 15 minute interval, producing a resultant image. Pixels in the resultant image that had a nonzero intensity above the image noise indicated areas of continued deformation. Scanning of the deformation phase was deemed complete when the resultant image showed a random scattering of pixel intensities (i.e. noise) (Figure 2-3). The pneumatic cylinder in the loading device was then retracted slightly for 8 hours, with the resultant image

Figure 2-3 Resultant images where the pixel intensities of an image are subtracted from an image taken 15 minutes prior. (Left) Resultant image shows area of dark pixels indicating continued deformation. (Right) Resultant image shows only noise, indicating deformation has reached steady-state.

A version of this chapter has been submitted for publication in the Journal of Biomechanics
showing no thickness recovery in this time. The piston was then fully retracted (without applying tension to the joint) beginning the cartilage recovery phase and the specimen was scanned again. Thereafter, the specimen was scanned approximately every hour until the recovery phase was deemed complete, again using the pixel intensity subtraction technique as described above. In the recovery phase, the loading device was filled with PBS+ solution in between scans, providing fluid for re-uptake into the cartilage.

2.2.3 Data Processing

At each time interval, cartilage was segmented (Analyze 8.1, Mayo Clinic, USA) as a single unit between the subchondral bone of the femur and the subchondral bone of the femur and the subchondral bone of the acetabulum in the anterosuperior quadrant (weight bearing region) in the center slice. The segmented cartilage edges were exported as two-dimensional data points and a custom-made program (Matlab 7.0, Mathworks) approximated the acetabulum as a circle and measured radial cartilage thickness in more than 200 locations within the slice. The reference point of zero degrees was formed by a line joining the center of the approximated circle to the AllS. The region of interest, chosen to encompass the anterosuperior region of the joint was then set to range from 25° to 45° (Figure 2-4). Mean thickness was computed based on all 200 thickness measurements in one slice in the region of interest.
Approximating the acetabular rim as a circle, mean cartilage thickness was measured radially from the osseous rim of the femoral head, to the osseous rim of the acetabulum in approximately 200 locations within one imaging slice.

For each thickness measurement, the change in cartilage thickness was divided by the unloaded cartilage thickness to calculate cartilage strain.

2.2.4 Precision

The pneumatic piston was tested for precision outside the MRI scanner by pressurizing the inner chamber of the cylinder to 62.3 psi, and recording the resultant force delivered by the pneumatic cylinder using a load cell (25kN capacity, 2.5 N precision) recorded at 10 Hz. Tests were conducted in a materials testing system (Instron 8874) with a position hold signal delivered with WaveMaker software. The piston chamber was pressurized almost instantly, with full load recorded approximately 0.5 seconds after opening the solenoid valve. Three
consecutive 110 minute long tests were performed, including depressurizing and retracting the pneumatic cylinder for approximately 1 minute in between tests.

To determine the repeatability of our method of measuring cartilage deformation, we scanned one specimen while unloaded and after 105, 120, 135 and 150 minutes of loading on four consecutive days. On each day, after loading, the specimen was removed from the magnet bore and submerged in PBS+ solution overnight. Two 1 cm long, 2 mm in diameter gadolinium-filled cylindrical fiducial markers were placed at approximately 90° to one another, in drilled holes in the anterior inferior iliac spine. The fiducial markers served to improve the repeatability of locating an imaging slice with the same orientation and position each day. Cartilage thickness was measured as described above and both the mean coefficient of variation and the average percent difference in mean cartilage thickness over the four days were computed for the unloaded hip, and after 105, 120, 135 and 150 minutes of loading.
2.3 Results

2.3.1 Cartilage Deformation

The pixel intensity subtraction technique indicated steady-state thickness distribution was reached (i.e. resultant image showed only noise) in all specimens after 225 minutes of loading. Unloaded thickness was not obtained in one specimen and therefore overall deformation and strain are reported in four specimens only. Across the four specimens, the average unloaded cartilage thickness was 3.23 mm ± 1.15 mm (mean ± standard deviation). After loading for 225 minutes, the average cartilage thickness was compressed to 2.27 mm ± 0.92 mm. The range of overall cartilage deformation was 0.77 mm to 1.36 mm (Figure 2-6). The average cartilage deformation was 0.96 mm ± 0.27 mm. Average cartilage strain was 30.9% ± 6.0% after 225 minutes of loading. Approximately 20% of the total deformation occurred within the first 15 minutes of loading, with 50% of the deformation reached after 50 minutes of loading (Figure 2-6).

Figure 2-5 A. MRI showing hip cartilage before loading (time 0), white lines mark initial position of cartilage as reference. B. Cartilage thickness after 105 minutes of loading. C. Cartilage thickness after 225 minutes of loading.
Figure 2-6 Mean cartilage thickness change over time in five specimens.

Deformation varied within the region of interest, in terms of both absolute cartilage deformation and strain. Regional variance in cartilage deformation across the imaging slice is illustrated for one specimen in Figure 2-7.
Figure 2-7 Regional variance of cartilage deformation in the center imaging slice across the region of interest in one specimen (specimen 4) over various time points of loading. From dark to light, time 0 (black), 15 min, 60 min, 225 min (white).

2.3.2 Cartilage Recovery

The pixel intensity subtraction technique showed recovery was complete after 16.5 hours. As the specimen shifted when the piston was retracted, the imaging slice also shifted and therefore deformation and recovery phases cannot be directly compared because the imaging slice was different. In the recovery phase, an average thickness of 0.3 mm was recovered (Figure 2-8).
2.3.3 Precision

In the precision tests of the loading device, the mean load over the three tests was 1972 N. The mean and maximum standard deviation as a percentage of the mean load was 0.23% and 1.25%, respectively (Figure 2-9).

Based on the time found to reach steady-state cartilage deformation (225 min) and recovery (16 hr 30 min), the same specimen was imaged unloaded and throughout 225 minutes of loading on each of four days, allowing 16.5 hours of recovery overnight. A higher precision was achieved when comparing cartilage deformation for consecutive days only, rather than comparisons made across more than one day. The mean coefficient of variation in cartilage thickness over consecutive days of data collection was 2.7 ± 1.5 %. The average change in
cartilage thickness in the control specimen over consecutive days of data collection was 0.10 mm.

Figure 2-9 Precision testing measuring the compressive load delivered from the pneumatic hip loading device over time for 3 separate tests.
2.4 Discussion

A device for loading cadaver hip specimens within the bore of a 7T MRI scanner was developed and used to assess cartilage deformation and recovery under simulated physiological load. Hip cartilage reached a steady-state thickness after 225 minutes of loading and deformed an average of 0.96 mm (30.9% mean strain). Hip cartilage thickness recovery reached steady-state 16.5 hours after removing the load.

This study’s results for cartilage strain and time to equilibrium at the hip are generally consistent with results found previously at the knee joint. The current finding that cartilage reached 50% of its final deformation after the first 50 minutes of loading is consistent with previous studies evaluating knee cartilage (Table 2-2). The measurements of mean cartilage strain of 30.9% after 225 minutes of loading are consistent with the finding of a mean deformation of 30% in human femoral cartilage after 214 minutes of loading at 1.5 times body weight and the finding of a mean deformation of 33% in human patellar cartilage after 250 minutes of loading at 2 times body weight. Here we report somewhat less strain in a longer time than the finding of a mean deformation of 36.1% in the medial tibial cartilage of sheep knees after 120 minutes of load at 1.5 times body weight. Differences in strain between this study and others are likely due to differences in image resolution, type of joint and species tested (Table 2-2).

The one specimen that was assessed during the recovery phase remained in the bore of the MR scanner after deformation and throughout the recovery phase. Due to the lack of visual feedback from the loading device, there was an 8 hour period where the load was removed.
from the piston but the piston was not fully retracted. The pixel intensity subtraction technique showed no recovery during the 8 hours the load was removed; similarly, data analysis showed very little cartilage thickness recovery (0.1 mm) in this 8 hour time period. Therefore, recovery time began when the piston was fully retracted.

The present finding of a cartilage thickness recovery time of 16.5 hours is consistent with some studies but not others; however, these differences can be explained with reference to the methods used. The current finding of a cartilage thickness recovery time of 16.5 hours is quite consistent with the finding of a 90% mean thickness recovery after 14 hours of unloading in the tibial cartilage of a sheep specimen\(^1\). However, it is longer than the 98% recovery of the thickness in patellar cartilage reported after only 4 hours of unloading\(^5\). This may be due to the rather poor imaging resolution used in that study, which with an in-plane resolution of 0.31 mm

Table 2-2 A summary of similar studies reporting cartilage strain in vitro.

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<td>250</td>
<td>120</td>
<td>225</td>
</tr>
<tr>
<td>Duration of load (mins) to 50% final deformation</td>
<td>25</td>
<td>50</td>
<td>-</td>
<td>50</td>
</tr>
<tr>
<td>Cartilage Species</td>
<td>Femur</td>
<td>Patellar</td>
<td>Tibial</td>
<td>Acetabular + femoral</td>
</tr>
<tr>
<td>Mean strain (%)</td>
<td>30</td>
<td>33</td>
<td>36.1</td>
<td>30.9</td>
</tr>
</tbody>
</table>

A version of this chapter has been submitted for publication in the Journal of Biomechanics
was almost three times less than our study and others that reported longer recovery times. An image resolution of 0.31 mm would be insufficient to detect the 0.17 mm change in thickness between 4 and 14 hours of recovery measured in the present study (Figure 2-8).

The strengths of the current approach are that image resolution is high, deformation of cartilage is directly observed and validation is possible because the method is non-invasive. One key strength is that we have established the precision of the method, which is necessary to detect small changes in cartilage thickness from one loading condition to the next. Measuring the cartilage deformation at various time points in the same specimen across four consecutive days established the combined effect of several sources of error including the precision of positioning the specimen in the loading device, positioning the loading device in the MRI magnet, the slice prescription, the loads applied by the pneumatic loading device, the segmentation process, and the radial thickness measurement algorithm. In the present study we observed an overall mean cartilage deformation of 0.96 mm which was far greater than the imaging resolution of 0.11 mm and the mean measurement of error of 0.10 mm. A further strength was the high precision of the loading device which applied compressive loads simulating single-leg stance with a mean standard deviation of only 0.23%, which is similar to that of a previously reported MRI loading device designed for cartilage explants.

One limitation of the current approach is that the deformation and recovery data presented here apply to ex vivo conditions and have only indirect application in vivo. The study utilized cadaver specimens with two portal holes and a 3 cm capsular tear created in the joint prior to testing, and the intra-articular pressure in these joints is likely different from the in vivo
condition. A second limitation is the use of a static load as an approximation of physiological loading, which would be better simulated by a varying cyclic load. While this is a simplification, overall joint compression has been shown to be similar for static and cyclic loads of a similar duration. A third limitation is that the present study was limited to a two-dimensional analysis from a single imaging slice due to the need to acquire images quickly with a high in-plane resolution. The steady-state response and the amount of deformation from another region of interest may change, and this limitation must be considered when applying the findings of the present study to other work. Thicker cartilage appears to deform more than thinner cartilage (Figure 2-7). Deformation in other regions of the hip joint also likely depends on the regional cartilage thickness and geometry of the loading environment. A three-dimensional analysis of the steady-state response of cartilage deformation in the hip joint would provide a more complete understanding, although this approach is impractical using MRI alone due to long image acquisition times.

The MRI technique is limited by the lack of immediate quantitative feedback of cartilage thickness. Therefore, the current study used immediate qualitative deformation and recovery feedback from the resultant image. The image corresponding to the final data points in Figure 2-6 and Figure 2-8 produced a resultant image showing noise (Figure 2-3), indicating no further change in cartilage thickness. Although Figure 2-6 and Figure 2-8 do not indicate steady-state thickness has been reached, within the limits of imaging resolution, no further change in cartilage thickness was detectable and therefore would not contribute to motion artifact.
Our results show that it is possible to measure hip cartilage deformation under load using high-field MRI. Quantifying the deformation and recovery time of hip cartilage is essential to minimizing errors introduced by motion artifact in studies of cartilage strain under load using these methods. This new method has potential for assessing changes in hip cartilage strain due to injury or surgical intervention that may be clinically significant.
References:


Chapter 3: Effect of Acetabular Labral Tears, Repair and Resection on Hip Cartilage Strain: A 7T MR Study

3.1 Introduction

Patients with acetabular labral pathology often present with symptoms of pain, and mechanical symptoms such as locking in the hip joint at the extreme ranges of motion. The overall incidence of labral tearing in the general population is currently unknown, however clinicians have reported that 22% - 87% of patients who presented with groin pain had a diagnosis of a labral tear. Labral tears may be caused by trauma, and are strongly associated with femoroacetabular impingement and dysplasia. They are also associated with degenerative changes in the hip.

Labral pathology has been associated with cartilage degeneration in clinical studies, cadaver studies, and studies using animal models. In one of the largest retrospective clinical studies, reviewing 436 hip arthroscopies, 261 patients (55%) had labral tears, and 73% of these had associated chondral damage. While some investigators believe that labral tears precede cartilage degeneration, others believe the reverse, that chondral injuries lead to labral tearing.

It is widely accepted that excessive or altered mechanical loading precipitates cartilage degeneration. Understanding the mechanical role of the acetabular labrum is necessary to design and improve treatment strategies that protect joint cartilage. The labrum maintains intra-articular pressure in the hip joint, which plays an important role in cartilage...
preservation by reducing wear and by supporting part of the load, sparing the cartilage matrix from excessive loads\textsuperscript{40,3, 39, 60}. The labrum also deepens the acetabular socket by 21%, providing an additional 8 cm\textsuperscript{2} of articulating surface area with the femoral head, thereby distributing the loads more evenly across the joint and sparing cartilage from excessive regional loads\textsuperscript{64}. This increased surface area of the labrum is also thought to provide additional constraint of the femoral head in the acetabulum at extreme ranges of motion\textsuperscript{33}.

Labral resection and labral repair are both used arthroscopically to treat symptoms associated with labral tears. Labral resection involves removing the damaged portion of the labrum with minimal disruption to healthy labral tissue\textsuperscript{33}. Labral repair involves suturing the torn portion of the labrum to the acetabular rim using a suture anchor. There are several reports of prompt pain resolution for both labral resection\textsuperscript{2, 11, 27, 29, 33} and labral repair\textsuperscript{33, 56} procedures. However, a recent study found that radiographic signs of cartilage degeneration were significantly more prevalent in those with a labral resection at one and two years post-operation compared to those with a labral repair\textsuperscript{16}.

Quantifying the ability of a surgical technique to restore natural hip joint mechanics is hindered by the limited methods available to measure contact mechanics in the hip. Pressure sensitive film has been used in vitro\textsuperscript{1, 62, 66}, however inserting the film is invasive, the film has a limited resolution, and the film interferes with contact mechanics due to its thickness\textsuperscript{68}. Hydraulic actuator displacement and intra-articular pressure have been measured in hip specimens subjected to compressive load in a servo-hydraulic materials testing system\textsuperscript{18}, but these measurement techniques provide no information about the distribution of load across the joint.
Finite element modeling methods\textsuperscript{17,7} are difficult to validate, and studies of cartilage explanted from the joint are difficult to translate to \textit{in-vivo} applications.

Cartilage strain mapping using quantitative magnetic resonance imaging (qMRI) has recently been developed\textsuperscript{51,52,61} and has a number of advantages over previous techniques. This method uses very high-resolution MR images of cartilage before and after loading to map cartilage strain as a result of joint loading\textsuperscript{61}. The resulting quantification of the distribution of cartilage strain may have more utility to detect an altered loading environment than average strain measurements, since average strain measurements may miss local changes in strain that are clinically important. A second advantage of this approach is that it is non-invasive and allows the joint capsule to be maintained, eliminating the effects of joint disruption to implant transducers and the changes to joint mechanics produced by transducers themselves. While the method has been used at the knee\textsuperscript{61}, to our knowledge it has not been previously used successfully in the hip.

We used high-field MRI strain mapping to answer our research question: Does acetabular labral repair reduce mean and maximum cartilage strain more effectively than labral resection in cadaver hips with simulated labral tears?
3.2 Methods

3.2.1 Specimen Preparation

Seven fresh, previously frozen cadaver human specimens (aged 20-48 years, mean 36.9 years, 4 female, 3 male) were obtained and prepared for testing. Institutional review board approval was obtained for this project (Appendix A). Each specimen, consisting of a hemi-pelvis and femur transected 20 cm distal to the greater trochanter, was dissected of all superficial soft tissues, leaving the capsule intact. An arthroscopic examination was performed prior to testing on each specimen to ensure that no major pre-existing pathology was present (defined as any chondral delamination or chondrosis greater than Outerbridge grade 1 or a labral tear greater than 1 cm in length). Using a laser, with anatomical references, the femur was potted in a physiologic alignment simulating single-leg stance. With the pubic tubercle and anterior superior iliac spine mounted vertically in the same plane, and with medial hemi-pelvis tilt determined from a three-dimensional anatomical model, the pelvis was anatomically aligned when a horizontal (axial) cut was made at the level of the anterior inferior iliac spine. The potting of the femur combined with the axial cut of the pelvis provided a physiologic orientation simulating single leg stance when the specimen was mounted in the pneumatic loading device (Chapter 2). The specimen was kept moist throughout preparation using a solution of phosphate buffered saline with protease inhibitors (PBS+ solution) (see Appendix G).
3.2.2 Loading Device

A MRI compatible pneumatic loading device was developed to apply a constant compressive load simulating single leg stance to a human hip specimen (Figure 3-1). The pneumatic components operating the loading device consisted of a compressor (FP209400RB, Campbell Hausfeld, Harrison OH, USA), an in-line air filter (07F34AC, Parker, Cleveland OH, USA), a 1-way flow valve, a precision regulator (035501040, Parker, Cleveland OH, USA), a digital pressure gauge (2798k22, McMaster Carr, Atlanta GA, USA), a solenoid valve and a custom double-acting pneumatic cylinder (BECO Manufacturing, Laguna Hills, CA). The precision regulator had a repeatability of ± 0.14 kPa and was capable of regulating the pressure between 14 and 1034 kPa. The pneumatic piston was made of polypropylene, a material that does not produce artifacts on MR images. All other components were placed outside the magnetic field and connected to the loading device with silicone tubing traveling through a filter plate into the room housing the MRI scanner. The outer diameter of
the loading device (Figure 3-1) was designed to fit in a 15 cm inner diameter imaging coil. The device was made of an acrylic tube of 5.5" outer diameter with a ¼" wall thickness (Appendix B). The pneumatic cylinder was fixed to the removable end of the acrylic tube. The specimen was mounted to the free end of the pneumatic cylinder and was axially compressed toward the fixed end of the loading device.

The loading device was designed to apply 1980 N of axial compression, which corresponds to 2.3 times body weight for a body mass of 88 kg. This is an average resultant hip force for single-leg stance ⁶. It is less than that for walking quickly or stair climbing (2.5 times body weight)⁶ and greater than that for sitting (1.6 times body weight)⁶. Repeatability of the force applied by the pneumatic piston was assessed with a materials testing machine and a load cell prior to testing each specimen (see section 2.2.4 for details).

### 3.2.3 Imaging Parameters

MR imaging was performed with a fat-suppressed 3D FLASH sequence using a 15 cm inner diameter quadrature volume coil on a 7.0T MRI system (Bruker Biospin, Ettlingen, Germany). Imaging parameters were as follows: TR/TE 25/5.14 ms; flip angle 20°, slice thickness 0.30 mm, in-plane resolution 0.10 mm, imaging matrix 512 x 512 x 170, field of view 5.12 cm, 170 slices per image acquisition, number of averages = 4, imaging time of 2 hr and 25 min. We prescribed slices orthogonal to a plane running parallel to the acetabular rim (Figure 3-2). These slices passed through the longitudinal axis of the anterior inferior iliac spine (AIIS). Two 1 cm long cylindrical gadolinium-filled fiducial markers were placed perpendicular to one another in the
anterior inferior iliac spine, such that the prescribed orientation of the imaging slices was reproducible.

Figure 3-2 The imaging slice was orthogonal to a plane that runs parallel to the rim of the acetabulum (white line). This white line leads to the anterior inferior iliac spine (AIIS).

3.2.4 Protocol

To minimize motion artifact due to the dynamic nature of cartilage deformation under load, we performed a pilot study in which we acquired high-resolution images throughout cartilage deformation and recovery. We determined that steady-state cartilage thickness under constant load of 1980N was reached after 4 hours of loading, and steady-state unloaded cartilage thickness was reached 16.5 hours after this load was removed.23

All specimens were tested intact and after an arthroscopically simulated labral tear, labral repair, and labral resection (Figure 3-3). The testing sequence could not be randomized due to the destructive nature of the tests. Prior to assessing the intact condition, we established an anterolateral portal using a spinal needle, followed by nittinol wire. We then placed a 5.5 mm hip arthroscopy cannula in this visualization portal. We then established an anterior portal using the spinal needle, nittinol wire and a 4.5 mm hip arthroscopy cannula using an outside-in technique under direct visualization. We used a probe for initial visualization and inspection of
the femoral head, acetabulum and labrum. Creation of the simulated labral tear required sharp detachment of the labrum through the hip capsule. Therefore, a capsular tear was created through the anterior portal, prior to testing the intact condition in order to isolate the effect of a labral tear.

On day one, the labrum was intact. The specimen was scanned prior to loading and then again after being loaded for 4 hours. The hip was then unloaded and submerged in the PBS+ solution for 16.5 hours. In the same hip, we then arthroscopically created a 3 cm long longitudinal peripheral labral tear by sharp detachment of the labrum off the osseous acetabular rim in the anterosuperior region. On day two, the hip (torn labrum) was loaded for 4 hours and then scanned. The hip was then unloaded and submerged in PBS+ solution for 16.5 hours. The labral tear in the same hip was then repaired using three 2.3 mm Bioraptor suture anchors (Smith & Nephew Endoscopy, Andover MA, USA), which were inserted using standard arthroscopic techniques with a straight Arthropiercer device (TM, Smith and Nephew) for suture passage through the labrum from a capsular to articular side direction for fixation (one suture pass was used through the labrum per anchor). A Weston sliding locking arthroscopic knot was tied for each anchor. On day three, the hip (repaired labrum) was loaded for 4 hours and then scanned. The hip was then unloaded and submerged in PBS+ solution for 16.5 hours. The 3 cm length of labrum was then resected using sharp knife resection, and on day four, the same hip (resected labrum) was loaded for 4 hours and then scanned.
Figure 3-3 Study protocol for each hip specimen.

3.2.5 Thickness and Strain Measurement

We mapped cartilage thickness and strain in the anterosuperior region of the hip. The acetabular and femoral cartilage were manually segmented as a single unit, from the osseous surface of the femoral head to the osseous surface of the acetabulum (Analyze 8.0, Mayo Clinic, Rochester, MN, USA), in 24 central MRI slices in the anterosuperior region of the hip joint. Cartilage contours were then exported as three-dimensional (3D) data points. When the labrum was intact and the joint was loaded, a reference point was defined as the edge of the osseous
contour of the acetabulum in the middle MRI slice. Using a custom registration algorithm (Matlab 7.0, Mathworks, Natick, MA, USA), the osseous rim of the acetabulum in each condition of the labrum was registered to the osseous rim of the acetabulum when the labrum was intact and loaded. In this way, the defined reference point was aligned in all data sets (Appendix C). Next, a custom program approximated the osseous contour of the acetabulum as a sphere using a least squares sphere fit\textsuperscript{48}. A reference axis was defined as a line joining the reference point in the middle imaging slice to the center of the sphere. Regions of cartilage were defined as rotations from this axis about the centre of the sphere, with the edge of the osseous rim corresponding to 0°. Our region of interest was defined as cartilage between 5° and 40° ranging over 24 slices in the anterosuperior region of the hip joint (Figure 3-4). This chosen anterosuperior region of interest corresponded to what is widely accepted to be the weight bearing region of the hip\textsuperscript{1,62,648}. A custom program measured cartilage thickness radially in 3D in approximately 8,000 locations within the region of interest.

Figure 3-4 Cartilage region of interest chosen in anterosuperior quadrant of the hip.
To compute cartilage strain, at each of the 8,000 locations where cartilage thickness was measured, the loaded cartilage thickness was subtracted from the unloaded cartilage thickness and this value was then divided by the unloaded cartilage thickness.

3.2.6 Control Specimen

To determine the precision of this technique, one specimen underwent the same protocol outlined in Figure 3-3, however the condition of the labrum was left intact (with a portal hole and 3 cm tear in the capsule) at every measurement. Cartilage thickness was measured as described above on each of the four days, and comparisons were made between days 1 and 2 and between days 3 and 4 to quantify uncertainty associated with specimen repositioning, changes in tissues over time, segmentation and sphere fitting, and other factors for comparisons between the intact (day 1), torn (day 2), repaired (day 3) and resected (day 4) labrum. The unloaded thickness was not measured in this specimen, therefore cartilage thickness values were converted to strain using an estimated unloaded thickness based on the average deformation observed in six specimens.

3.2.7 Statistics

We compared mean and maximum cartilage strain when the labrum was torn versus intact, repaired versus torn, and repaired versus resected using paired t-tests.
3.3 Results

3.3.1 Control Specimen

Mean and maximum cartilage strain changed by less than 4% and 3%, respectively, from day-to-day in the control specimen (Table 3-1). Qualitatively, the cartilage thickness distribution did not vary substantially between consecutive days of data collection (Figure 3-5).

Table 3-1 Control specimen mean and maximum strain measurements (in %) across 4 consecutive days.

<table>
<thead>
<tr>
<th>Day</th>
<th>Day 1</th>
<th>Day 2</th>
<th>Day 3</th>
<th>Day 4</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean strain (± SD)</td>
<td>25.8 (±5.8)</td>
<td>29.4 (±6.8)</td>
<td>30.6 (±6.7)</td>
<td>33.1 (±7.9)</td>
</tr>
<tr>
<td>difference from previous day</td>
<td>-3.6%</td>
<td>-1.2%</td>
<td>-2.5%</td>
<td></td>
</tr>
<tr>
<td>Max strain</td>
<td>63.4</td>
<td>65.7</td>
<td>67.5</td>
<td>69.2</td>
</tr>
<tr>
<td>difference from previous day</td>
<td>-2.3%</td>
<td>-1.8%</td>
<td>-1.7%</td>
<td></td>
</tr>
</tbody>
</table>

Figure 3-5 Control specimen cartilage thickness distribution in the anterosuperior region of the hip (in mm) day 1 through day 4.
3.3.2 Intact vs. Torn Labrum

Due to complications in one specimen (specimen 2), comparisons between intact and torn labrum cases were reduced to 5 specimens. The average (±SD) strain in intact specimens was 33.0% (±5.1%) and in the same specimens with a torn labrum the average strain was 32.4% (±5.5%) (Figure 3-6). The mean maximum strain in intact specimens was 45.2% and in specimens with a torn labrum the mean maximum strain was 46.2% (Figure 3-6). The population averaged 3D distribution of cartilage strain (Figure 3-7) shows an inferior shift of the most superior area of high strain when the labrum was torn relative to the intact state, but otherwise the distribution of cartilage strain in the region of interest does not appear to have been substantially changed by the labral tear (see Appendix D for individual specimen thickness and strain maps). For the number of specimens tested, there was no consistent change in mean cartilage strain after a labral tear (Figure 3-8) and the range of observed differences corresponded to the change in strain observed in the control specimen (Table 3-2).

![Figure 3-6 Mean and maximum cartilage strain when the labrum was intact, torn, repaired and resected. Comparisons are valid among adjacent test conditions only. No change in cartilage strain was observed when the labrum was torn vs. intact. Asterisks indicate significant (p<0.05) findings.](image-url)
Table 3-2 Mean and maximum cartilage strain change among various labral conditions relative to strain changes in the control specimen.

<table>
<thead>
<tr>
<th>Strain</th>
<th>Specimen</th>
<th>Intact (%)</th>
<th>Torn (%)</th>
<th>Change (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean</td>
<td>Control</td>
<td>25.8</td>
<td>29.4</td>
<td>-3.6</td>
</tr>
<tr>
<td></td>
<td>1</td>
<td>37.1</td>
<td>33.1</td>
<td>-4.0</td>
</tr>
<tr>
<td></td>
<td>3</td>
<td>30.0</td>
<td>31.9</td>
<td>-1.9</td>
</tr>
<tr>
<td></td>
<td>4</td>
<td>42.0</td>
<td>39.3</td>
<td>-2.7</td>
</tr>
<tr>
<td></td>
<td>5</td>
<td>34.0</td>
<td>35.8</td>
<td>-1.8</td>
</tr>
<tr>
<td></td>
<td>6</td>
<td>21.8</td>
<td>21.8</td>
<td>0</td>
</tr>
<tr>
<td>Max</td>
<td>Control</td>
<td>63.4</td>
<td>65.7</td>
<td>-2.3</td>
</tr>
<tr>
<td></td>
<td>1</td>
<td>53.7</td>
<td>50.3</td>
<td>3.4</td>
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<td></td>
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</tr>
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<td>5</td>
<td>46.6</td>
<td>53.6</td>
<td>-7.0</td>
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<td>6</td>
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<td>35.5</td>
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</tr>
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<td>Control</td>
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<td>-2.5</td>
</tr>
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<td>40.1</td>
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<td>28.0</td>
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</tr>
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<td>42.4</td>
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<td>2.0</td>
</tr>
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<td></td>
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<td>17.0</td>
<td>26.0</td>
<td>-9.0</td>
</tr>
<tr>
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<td>Control</td>
<td>67.5</td>
<td>69.2</td>
<td>-1.7</td>
</tr>
<tr>
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<td>46.9</td>
<td>51.0</td>
<td>-4.1</td>
</tr>
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<td></td>
<td>6</td>
<td>26.6</td>
<td>42.1</td>
<td>-15.5</td>
</tr>
</tbody>
</table>
Figure 3-7 Population averaged cartilage strain distribution in the region of interest when the labrum was intact compared to when the labrum in the same hip was torn. When the labrum was torn compared to repaired, and when the labrum was repaired compared to resected.
Figure 3-8 Individual specimen results when the labrum was intact versus torn, torn versus repaired, and when the labrum was repaired versus resected. Comparisons are valid among adjacent test conditions only. The control specimen is shown as a dotted line.

3.3.3 Torn vs. Repaired Labrum

Mean cartilage strain was significantly decreased when the labrum was repaired compared to torn (p=0.014) (Figure 3-6). There was no significant change in maximum cartilage strain following a labral repair compared to the torn condition. The decrease in mean cartilage strain was evident in all 5 specimens (Figure 3-8). Average (±SD) strain for torn specimens was 32.4% (±5.5%) and for the same specimens with a repaired labrum average strain was 30.4% (±7.5%) (Figure 3-6). Mean maximum strain for specimens with a torn labrum was 46.2% and for specimens with a repaired labrum was 44.0%. Qualitative observation of the population averaged 3D cartilage strain distribution in hips with a torn labrum versus repaired (Figure 3-7)
shows a decrease in cartilage strain following a labral repair in the inferior portion of the region of interest.

3.3.4 Repaired vs. Resected Labrum

Mean and maximum cartilage strain were significantly increased when the labrum was resected compared to repaired ($p=0.020$ and $p=0.023$, respectively) (Figure 3-6). The increase in mean cartilage strain after resection was evident in 5 of 6 specimens (Figure 3-8) and the observed increases in cartilage strain were at least 40% more than the change in strain observed in the control specimen over the same time period (Table 3-2). Average (±SD) strain for repaired specimens was 30.4% (±7.5%) and for the same specimens with a resected labrum average strain was 34.4% (±6.4%) (Figure 3-6). Mean maximum strain for specimens with a repaired labrum was 44.0% and for specimens with a resected labrum the mean maximum strain was 50.0%. Qualitative observation of the population averaged 3D cartilage strain distribution in hips with labral repair versus resection (Figure 3-7) shows an overall increase in strain magnitude throughout the region of interest.
3.4 Discussion

We measured cartilage strain in response to simulated weight bearing load in hip specimens before and after labral repair and resection to assess the effect of these procedures on load distribution in the hip joint. We found significantly more mean and maximum strain in the anterosuperior region of the hip cartilage after labral resection than after labral repair.

The measured differences in mean and maximum cartilage strain between labral repair and resection procedures appear to be due to the different effect each procedure had on joint mechanics. With the exception of one specimen (specimen 5), the mean and maximum differences in strain were all greater than 1.4 times the differences observed over the same testing period for the control specimen. One specimen (specimen 6) had about 9 times the strain difference observed in the control specimen (Table 3-2).

The control specimen was tested to quantify cartilage strain changes due to such factors as day-to-day positioning variation of the specimen in the loading device or in the bore of the MRI magnet, imaging slice prescription techniques, cartilage deformation time (4 hours), cartilage recovery time (16.5 hours), potential swelling induced by the PBS+ solution, cartilage degradation, segmentation techniques, registration techniques, and the cartilage thickness measurement algorithm. The control specimen was particularly important because the order of testing could not be randomized due to the destructive nature of the labral conditions. Based on the increase in strain observed in the control specimen over the four days (Table 3-1), it is evident that cartilage strain when the labrum was intact (Day 1) should not be compared to cartilage strain after the labrum was repaired (Day 3) or resected (Day 4). Similarly,
comparisons should not be made between the torn labrum (Day 2) and resected (Day 4) states. Comparisons are only valid between consecutive days of data collection.

The accuracy of measuring cartilage thickness with MRI using a similar imaging resolution to that used in the present study was previously successfully validated against micro-CT measurements in the knee joint, with findings of an average root mean square difference of 0.08 mm\textsuperscript{61}. Although accuracy findings may differ in the spherically shaped hip joint, and with different techniques of cartilage thickness measurement, the reported relative changes in cartilage strain in the present study remain valid, as verified through use of a control specimen (Figure 3-5).

We were surprised that the labral tear did not produce a large change in cartilage strain, even though we simulated a large tear (3 cm). Many different types of labral tears have been reported, with longitudinal peripheral tears present in approximately 16% of cases\textsuperscript{36}. Although the radial flap tear is the most common (57% of cases)\textsuperscript{36}, we chose to evaluate the effect of a longitudinal peripheral tear because this type of tear has a higher capacity to heal after a labral repair\textsuperscript{32,45}. Labral tears greater than 2 cm in length have been reported in approximately 23% of cases, with the most frequent labral tear size being 1 – 2 cm in length (67% of cases)\textsuperscript{54}. As a conservative approach, a 3 cm long labral tear was chosen in all specimens. One possible explanation for the small change in strain after labral tear is that the contribution of the labrum toward increasing the surface area in contact with the femoral head remained in effect when the labrum had a longitudinal peripheral tear because both ends of the torn labrum remained affixed to the osseous acetabulum and cartilage. Whether the labrum is capable of sparing the
cartilage from excessive loads with a radial flap or radial fibrillated tear remains unknown. Since we did not see a difference in cartilage strain comparing the intact labrum to the torn labrum, it was surprising to see a significant decrease in mean cartilage strain following labral repair compared to the torn labrum condition. Perhaps labral repair results in lower cartilage strain compared to the intact condition, however we were not able to compare these two labral conditions due to the poor precision between day 1 and day 3 in the control specimen. We speculate that labral repair results in a circumferentially stiffer labrum, which may distribute the load more effectively. Our observation of higher cartilage strains when the labrum was resected versus repaired was expected because the torn portion of the labrum was no longer present and therefore unable to contribute to the surface area and distribute the load more evenly across the joint.

The results of the present study are generally consistent with previous studies in the literature. Our finding that labral resection increased cartilage strain compared to labral repair is consistent with the finding that full labral resection increased the final joint consolidation displacement by 21% compared to the intact case. The magnitude of hip cartilage strain reported in the present ex vivo study is consistent with ex vivo findings at the knee using similar methods, but greater than what has been found in vivo. Our finding of hip cartilage strain in the range of 30% was consistent with other ex vivo studies measuring cartilage strain in the knee under load using high field MRI, which found strains of 30% - 43%. In contrast, a previous in vivo study using MRI to quantify cartilage strain in the knee joint reported 2.4 – 8.5% strain in patellar cartilage after 100 knee bends while another study reported patellar cartilage strains up to 7% in high impact loading, and smaller strains for other activities such as squatting.
walking, cycling and running\textsuperscript{13}. An \textit{in vivo} study in the hip found an average strain of 2.3% after 1 hour of walking\textsuperscript{48}. The differences between \textit{in vivo} and \textit{ex vivo} cartilage strain measurement are likely in large part due to the lack of a physiologic intra-articular pressure \textit{ex vivo}. The presence of a pressurized fluid layer in the joint is vital to minimizing cartilage strain since this layer supports approximately 90\% of the load, sparing the cartilage matrix from excessive loading conditions\textsuperscript{3,19,60}. With \textit{in vivo} cartilage strain being so small, measuring a change in cartilage strain due to a surgical procedure is challenging with current technology. Assessment of cartilage strain change \textit{ex vivo} with high load may be the only way to detect the mechanical change imposed on the joint. Although the absolute strain values reported here are substantially different from those \textit{in vivo}, we believe the \textit{ex vivo} finding of increased cartilage strain distribution when the labrum was resected compared to repaired demonstrates the function of the labrum to provide increased surface area with the femoral head, thereby distributing the load more evenly in the joint. Thus, we suspect an \textit{in vivo} study would produce similar results in favor of repairing the labrum.

The observed changes in strain due to labral resection were small, however computing the corresponding increase in load subjected to the hip provides a more intuitive representation of the study results. The corresponding change in load subjected to the hip was determined based on the known change in cartilage thickness and using the cartilage biphasic theory\textsuperscript{46}. The aggregate modulus of cartilage was required to employ the cartilage biphasic theory. We did not measure the aggregate modulus in these specimens, however using a previously reported range of human cartilage aggregate modulus values\textsuperscript{4}, we found the measured overall mean cartilage thickness change of 0.16 mm between repaired and resected labral conditions.
corresponded to an additional 75 to 135 N of load subjected to the cartilage when the labrum was resected versus repaired.

We non-invasively quantified cartilage strain based on 7T MRI scans using a very high image resolution, which was a primary strength of our study. The precision of our measurements was assessed using a control specimen, and we found small changes in cartilage strain (below 4%) over four days of data collection. Our study design was also a strength because each specimen served as its own control, eliminating intra-specimen variability.

Our primary limitations are in the simulation of the in vivo environment with our ex vivo loading system. One limitation in applying the findings of the present cadaveric study to in vivo situations was that we applied a constant load to the cadaver hip in our simulation of in vivo hip loading, as opposed to a cyclic load, which would be a better simulation of most activities. We chose to apply a constant load to simplify the loading device. Although it is unlikely that the hip experiences a constant load of 2.3 times body weight for 4 consecutive hours, a study of six specimens loaded in compression found no change in cartilage deformation between a constant load of 570 N for 1 hr when compared to a cyclic load of 570 N ± 190 N at 1 Hz for 1 hr\(^8\). The 2.3 times body weight constant load that we applied represents an estimate of the average load subjected to the hip on an active day of walking quickly (2.5 times body weight), sitting (1.6 times body weight) and stair climbing (2.5 times body weight)\(^6\).

A second limitation is that we measured strain only within the anterosuperior region of hip cartilage, which we chose to do because this quadrant has been reported to be the weight-bearing region of the hip where cartilage experiences the greatest deformation\(^1,6,2,48\).
Localizing measurements within the anterosuperior quadrant therefore provides a more sensitive means of identifying differences in cartilage strain with labral condition. Further, due to the sphericity of the hip joint, accurate cartilage thickness measurement is limited to this small region of interest because the imaging slice lying perpendicular to the cartilage surface in the anterosuperior region would no longer be perpendicular to the cartilage surface in other quadrants of the hip. When the imaging slice cannot be oriented perpendicular to the cartilage surface, partial volume effects increase (because the through-plane resolution is worse than the in-plane resolution), and hence uncertainty in determining the cartilage-bone interface increases. A third limitation is that our cadaver model likely did not maintain physiological levels of fluid pressure in the joint. Part of the role of the labrum is to maintain high interstitial fluid pressure in the joint, which plays a large role in supporting the load, sparing the cartilage matrix from excessive loads and minimizing heat-induced cartilage wear. Because we could not maintain pressure due to the challenge of sealing the portal hole in the capsule, the cadaver model was only useful for assessing the effects of a change in joint surface area produced by disruptions to the labrum. It is therefore possible that had the sealing role of the labrum been included in this model, larger increases in cartilage strain might have been observed after compromising the labral seal, such as when the labrum was torn or resected.

Last, we report findings from a low sample population of only six specimens.

Our findings suggest that labral resection does not protect hip cartilage from increased strain as effectively as labral repair. Whether the increased magnitude of strain induced in the cartilage due to labral resection was enough to initiate osteoarthritis remains unknown, however the large number of clinical studies, cadaver studies, and animal studies...
models that report a relationship between acetabular labral pathology and cartilage degeneration suggest that these small changes in strain could contribute to the progression to osteoarthritis. The higher cartilage strains with labral resection versus repair found in the present study compliment the findings of a previous clinical study that reported significantly higher grades of osteoarthritis one and two years after patients had a labral resection versus those having a labral repair.

Based on our findings of increased strain after labral resection when compared to labral repair, we recommend that the longitudinal peripheral torn labrum should not be resected if it is possible to be repaired. While the findings of increased cartilage strain with a resected versus repaired labrum support the labral repair technique for a large longitudinal peripheral tear, the type of tear must be considered when deciding whether to repair or resect because not all labral tears have the capacity to heal.
References


Chapter 3: Labral Pathology and Cartilage Strain


A version of this chapter has been submitted for publication in the Journal of Bone and Joint Surgery


Chapter 4: Discussion

4.1 Motivation and Findings

Acetabular labral tears are common and may lead to osteoarthritis. Labral tears are associated with femoroacetabular impingement and dysplasia, and labral tears are commonly diagnosed in patients experiencing groin pain. Labral tear mechanical symptoms include a sensation of locking in the hip joint at the extreme ranges of motion. More concerning are the long term consequences: labral pathology is related to cartilage degeneration as shown through clinical studies, cadaver studies, and animal models.

Labral tears are reported to occur in patients as young as 14 years of age. The threat of osteoarthritis secondary to a labral tear in this young patient population is particularly concerning because a total hip replacement is not an ideal treatment option because failure rates in this patient population (<45 years) are high. Further research is warranted toward effective joint-preserving surgical treatment for acetabular labral tears. Current methods to treat a torn labrum (repair and resection) have not been assessed for their ability to preserve joint mechanics. Quantifying the ability of a surgical technique to preserve joint mechanics is challenging because contemporary test methods do not provide a three-dimensional assessment of load distribution in the hip and the test methods are destructive to the hip joint.

The research presented in this thesis is the first to develop and validate a protocol for a non-invasive method to measure the three-dimensional distribution of cartilage strain in the intact
human hip joint. We used high resolution MR images of loaded hip joints to quantify the three-dimensional cartilage strain distribution in cadaver hips. This allowed a repeated measures design with each hip loaded and scanned after various procedures were performed on the labrum. In chapter 2 it was determined that 225 minutes of constant compression at 2.3 times body weight was required for hip cartilage to reach a steady-state thickness distribution under load and it was also found that 16.5 hours of cartilage recovery was sufficient time for cartilage to reach a steady-state, unloaded thickness. Knowing the time parameter for deformation played an integral role in the study described in chapter 3 by minimizing motion artifacts in the high resolution scan of the loaded hip with various conditions of the labrum. Knowing the recovery time parameter was critical in the study described in chapter 3 because the cartilage strain with each condition of the labrum needed to be evaluated from an equivalent initial cartilage thickness prior to load application.

In the study described in chapter 3, we hypothesized that cartilage strain would be higher when the labrum was torn, however this was not found. With the longitudinal peripheral tear that we simulated in this study, both ends of the torn labrum remained fixed to the osseous acetabulum and cartilage, therefore it is possible that the contribution of the labrum toward increasing the surface area in contact with the femoral head remained in effect with this type of tear. It remains unclear whether the labrum is capable of sparing the cartilage from excessive loads with a radial flap or radial fibrillated tear. We did, however, observe a 4% and 6% higher mean and maximum cartilage strain, respectively, when the labrum was resected versus repaired. Further, we presented three-dimensional cartilage strain distribution maps in the anterosuperior region of the hip joint which showed an overall increase in cartilage strain when
the labrum was resected versus repaired. These increases in cartilage strain were expected because the torn portion of the labrum was no longer present and therefore unable to contribute to the surface area and distribute the load more evenly across the joint. The higher cartilage strains with labral resection versus repair found in the present study compliment the findings of a previous clinical study that reported significantly higher grades of osteoarthritis one and two years after patients had a labral resection versus those having a labral repair\textsuperscript{16}.

4.2 Expected change in cartilage thickness after labral resection versus observed

In chapters 2 and 3, the measured cartilage strain values were compared with previous \textit{in vitro} and \textit{in vivo} studies. However, the expected change in cartilage thickness after labral resection has not yet been compared to what was measured.

In this thesis work, the change in cartilage thickness when the labrum was repaired compared to resected was consistent with a previous study that measured the actuator displacement in human cadaver hip specimens loaded in a materials testing system with the labrum intact and after resecting the entire labrum\textsuperscript{17}. This previous study reported final overall joint consolidation with and without the labrum, normalized by body weight after 1 hour of load. As a rough approximation it was assumed that the test method in this previous study succeeded in isolating cartilage deformation (i.e. there was no osseous deformation). A second assumption was that the change in cartilage deformation was linear with applied load once all the intra-articular fluid had exuded from the cartilage (i.e. once steady-state thickness was achieved). A third assumption was that the contribution of the labrum toward increasing surface area in
contact with the femoral head was the same whether the labrum was intact or repaired. With these assumptions, a change in cartilage thickness of 0.34 mm was expected had the labrum been completely removed compared to when the labrum was repaired. Adjusting this value to reflect the current study’s use of a partial labral resection (removing approximately one-eighth of the labrum) the change in cartilage thickness with 2.3 times body weight when the labrum was resected compared to repaired was expected to be 0.04 mm. Adjusting our results of cartilage thickness change after partial labral resection (0.16 mm) based on the change in thickness observed in the control specimen (0.10 mm), we observed an additional 0.06 mm of cartilage thickness change after partial labral resection, which is similar to the predicted 0.04 mm. Therefore, the expected cartilage thickness change following partial labral resection was similar to our preliminary estimations.

4.3 Relating Strain Measurements to Load

The observed changes in strain due to labral resection were small, however computing the corresponding increase in load subjected to the hip could provide a more intuitive representation of the study results. We computed the corresponding change in resultant hip force based on the measured change in cartilage thickness between labral repair and resection (0.16 mm) and using the cartilage biphasic theory. The aggregate modulus of cartilage was a required parameter in order to employ the cartilage biphasic theory. We did not measure the aggregate modulus in these specimens, however using a previously reported range of human cartilage aggregate modulus values (0.5 – 0.9 MPa)\(^4\), we found the measured overall mean cartilage thickness change of 0.16 mm between repaired and resected labral conditions.
corresponded to an additional 75 to 135 N of force subjected to the cartilage when the labrum was resected versus repaired. Our results suggest that a portion of the 0.16 mm change in thickness between repair and resection can be attributed to day-to-day variation measured in the control specimen. In fact, the change in cartilage thickness induced by labral resection was 0.06 mm above the change seen in the control specimen between day 3 and day 4. A change in thickness of 0.06 mm corresponded to a change in load of 30 to 54 N.

4.4 Study Strengths

As discussed in Chapters 2 and 3, the methodology we developed to quantify the 3D loading environment in the hip is a strength of the research. Previous methods measured overall joint consolidation with cadaveric hips loaded in a materials testing machine, which provided no direct measure of cartilage deformation, nor any detail on the distribution of cartilage deformation in the joint. In the present study, cartilage strain was measured directly at the joint surface. Previously measuring the distribution of load across the cartilage was only possible using an array of miniature transducers, pressure sensitive film, or instrumented prostheses, all of which require destruction of the joint capsule. The joint capsule was maintained throughout data collection, with the exception of the two portal holes necessary for arthroscopic treatment, and the 3 cm capsular tear which was required for our specific study to isolate the effect of a labral tear.

The 7T MRI images acquired in the present study were of an ultra-high resolution (0.1 x 0.1 x 0.3 mm), and therefore error due to partial volume effect was minimized.
The control specimen, which showed very little change in cartilage strain due to various day-to-day factors such as misalignment and cartilage swelling, demonstrated a high level of precision and provided evidence that the changes in overall mean cartilage strain and cartilage strain distribution were indeed caused by altering the condition of the labrum.

Further, because each specimen served as its own control, intra-specimen cartilage strain changes were assessed, eliminating uncertainty in data interpretation due to inter-specimen variability.

4.5 Study Limitations

As described previously in Chapters 2 and 3, there are several limitations in this thesis research.

The primary limitation was our low sample size of only six specimens and the approximation of the loading environment as static. Despite previous research that demonstrated no difference in overall joint consolidation when cadaver hips were subjected to cyclic or static loads, the strain distribution across the surface of the cartilage could differ had the load been applied cyclically.

A second limitation was that we measured strain only within the anterosuperior region of hip cartilage because this quadrant has been reported to be the weight-bearing region of the hip which is where cartilage experiences the greatest deformation. Further, due to partial volume effect error, we chose not to measure cartilage thickness at the edges of the image. Therefore for purposes of minimizing error and providing increased measurement sensitivity to detect a change in cartilage strain, it was deemed appropriate to choose a small region of
interest. However, had it been possible to quantify cartilage strain across the entire hip joint cartilage with various conditions of the labrum, a complete measurement of strain at the hip would have been produced.

Applying the findings of the present cadaveric study to *in vivo* applications is limited because the intra-articular pressure in the hip joint was not monitored, and likely the pressure did not simulate the intra-articular pressure in the hip *in vivo*. The intra-articular pressure in the hip provides a layer of pressurized fluid that supports approximately 90% of the load and therefore spares the articular cartilage from excessive loads\(^3,^{36,51}\). Because of the lack of a physiologic intra-articular pressure in the hip joint, the cartilage strain values reported here of approximately 30% are far greater than would be measured *in vivo*. Cartilage strain has been measured in the knee joint *in vivo*, and it was reported to range from 2.4 – 8.5\(^\%\)\(^{14,12}\). However, the cartilage strain reported in the present *in vitro* study was consistent with other *in vitro* studies measuring cartilage strain in the knee\(^{23,52,54}\). The aim of the present research was to quantify relative changes in cartilage strain due to altering the labrum. Although the relative change in cartilage strain may be less *in vivo*, with a 6% higher maximum strain observed *in vitro* when the labrum was resected versus repaired, it is clear that the labrum is essential to contributing increased surface area. Will this increase in cartilage strain induced with partial labral resection lead to the onset of osteoarthritis *in vivo*? The answer is not known, however the large number of clinical studies\(^2, 9, 11, 18, 19, 21, 22, 32, 37, 38, 44, 45, 55\), cadaver studies\(^6, 33, 50\), and animal models\(^{39}\) reporting a relationship between acetabular labral pathology and osteoarthritis suggests that perhaps it is the very small changes in cartilage strain, less than 6%
measured \textit{in vitro} and perhaps far less than 6\% \textit{in vivo}, that could contribute to the onset of osteoarthritis.

\section*{4.6 Considerations when using MRI to measure tissue thickness \textit{in vitro} and \textit{in vivo}}

An important consideration when using MRI to quantify tissue strain is the signal intensity difference at the edge of the tissue of interest and the surrounding tissues. This interface must be clearly delineated for segmentation accuracy and precision. The intrinsically large difference between the signal intensity of the calcified bone and hydrated articular cartilage was crucial to the success of repeatable cartilage strain measurements (as demonstrated in the control specimen). Quantifying the strain in the labrum at various angles of hip flexion or abduction, for example, would be an interesting research topic; however the MRI sequence would need to be optimized to distinguish the fibrocartilaginous labrum from the acetabular cartilage and the fibrous hip capsule. The gradient echo sequence used in the current study did not provide sufficient delineation between the labrum and surrounding structures.

A second consideration is the time required to prescribe the imaging slice. Locating a repeatable imaging slice orientation and location is critical to achieving a repeatable thickness measurement. Decreasing the size of the gadolinium-filled fiducial markers yielded a more repeatable imaging slice orientation, however this was at the cost of an increased time required to locate the fiducials and hence prescribe the imaging slice. Future work aiming to minimize the time required to locate the fiducials or desired imaging slice orientation would be beneficial.
In the study of labral tears, resection and repair (chapter 4), on days 2 through 4, the imaging slice was defined only after loading the specimen, and consequently, information regarding the dynamic response of cartilage to load through the full load cycle (prior to slice definition) could not be obtained because locating the desired imaging slice required approximately one to two hours. In hindsight, prescribing the imaging slice prior to load on all days of data collection would have provided an advantage because, upon loading the specimen, the specimen underwent almost a pure translation with very little rotation (Appendix E) and therefore the desired slice orientation can be gained post-load rather quickly, requiring only a few adjustments. However, this was not known to be possible until the current studies were completed.

Had it been possible to define the imaging slice faster, information regarding the effect of a torn or resected labrum on the dynamic response of cartilage deformation through the full load cycle could have been obtained, providing a means of assessing the “sealing” role of the labrum. However, a well monitored and maintained intra-articular pressure in the cadaver specimens would have been necessary, which would be a challenge best achieved by conducting the study in vivo.

One difficulty in performing this type of work in vivo is that the use of fiducial markers is extremely limited in vivo because osseous implantation is rarely an option. Rather, anatomical reference points are used, which presents a challenge in the spherical hip joint because there are fewer distinguishing anatomical features in the hip than in the knee. Increasing the field of view to include anatomical references such as the anterior superior iliac spine and pubic
tubercle could be useful, however the imaging resolution would be significantly decreased and hence the repeatability of the chosen landmarks would be questionable.

The orientation of the imaging slice is certainly where detail is warranted when aiming to quantify tissue strain accurately. A previous study quantifying cartilage thickness \textit{in vivo} in the hip joint before and after loading did not provide a detailed description regarding the method of prescribing a repeatable imaging slice, however it was mentioned that the imaging slice was oriented perpendicular to the cartilage. Interestingly, this study found that cartilage thickness did not consistently increase in all subjects post-resting\textsuperscript{41}, which may be due to error associated with measuring thickness in a different imaging slice orientation in the post-resting scan compared to the post-loading scan (see section 1.4.2 for other sources of error).

4.7 Future Directions

4.7.1 Future In Vitro Studies

Applying findings from \textit{in vitro} studies to \textit{in vivo} conditions is inherently challenging due to the difficulty of simulating the \textit{in vivo} joint loading environment. To date, static versus cyclic loading in the hip has only been examined by measuring overall joint consolidation, with no significant difference observed between the two loading regimes\textsuperscript{17}. It has been established that cartilage degeneration occurs not only over maximum weight bearing regions, but also over areas that are subjected to minimal load\textsuperscript{13,57}. Assessing the effect of static versus cyclic joint loading on the distribution of cartilage strain could be a future direction of this research. Information on the dependence of the distribution of cartilage strain on the loading regime, obtainable with the methodology of the present study, could be useful toward enhancing our knowledge of
osteoaartthritis or toward defining appropriate rehabilitation protocols in patients or injured athletes.

The investigation of cartilage strain under a broad range of simulated in vivo environments would also be useful. The present research was limited to small angles of flexion and adduction due to the size of the imaging coil and due to the inherent instability of the cadaver hip when musculature is removed. Larger angles of flexion and adduction would require a sophisticated system to replicate muscle force in order to reduce the flexor and adductor moments at the pelvis while still obtaining physiological loads.

Xtreme-CT is another imaging modality where the distribution of load can be quantified non-invasively. Due to the congruency of loaded hip cartilage, cartilage thickness can be measured based on the distance between the femoral head and the acetabulum. Therefore, cartilage strain in cadaver hip specimens could be quantified using Xtreme-CT. Advantages of using Xtreme-CT include the ability to obtain an ultra high-resolution of 0.041 x 0.041 x 0.041 mm, higher than that used in the present study, and the image can be acquired in 1 to 1.5 hours. Xtreme-CT is more economically feasible compared to MRI and therefore a higher sample size can be achieved. Further, obtaining a repeatable imaging slice is no longer a concern because the image voxels are isotropic and therefore can be reconstructed. Therefore, dynamic measurements of cartilage thickness immediately after loading may be obtained. A disadvantage of Xtreme-CT is the restricted bore size, typically 12 cm in diameter, limiting the species, type of joint tested and design of the loading device.
4.7.2 Future In Vivo Studies

An important but challenging future direction would be to investigate hip cartilage strain with various conditions of the labrum in vivo. With advances in MRI technology such as the open-bore standing MRI, subjects could be scanned (while standing, after a defined period of exercise) before and after a labral repair or resection. Although the imaging resolution would not be as high as that used in the present research, novel image registration techniques could be employed by obtaining an unloaded high-resolution MRI scan of the bone morphology while the patient is lying supine, followed by quick low-resolution scans while the patient is standing in the open-bore MRI.

Combining imaging modalities such as fluoroscopy and MRI could be another method to measure cartilage strain in vivo. A high-resolution unloaded MRI scan of the joint could be acquired with the patient lying supine, followed by quick low-resolution 3D fluoroscopic images of the joint during loading. Registering the osseous morphology from the low-resolution fluoroscopic images with the high resolution MR unloaded reference image could then provide accurate cartilage thickness measurements. Utilizing fluoroscopic images rather than MRI would be a cost-effective approach that may provide an opportunity to increase the sample size.

However, the two methods outlined above are not without limitations. When cartilage strain induced by exercise is only approximately 6%\textsuperscript{14}, it is likely the change in strain due to labral resection is far less, which may be difficult to detect due to the many sources of error associated with image registration and thickness measurement (Appendix F). Also, with regard
to fluoroscopic imaging, the radiation dose subjected to the patient may prompt safety concerns that could limit the study design.

The methodology outlined above for in vivo research studies is certainly not limited to the investigation of cartilage strain at the hip. The methodology could also be used for the investigation of cartilage strain distribution at the knee before and after such operations as a high tibial osteotomy, meniscus repair or a single versus double bundle ACL reconstruction. Further, the methodology could be used in vivo not only with application to osteoarthritis prevention strategies, but toward osteoarthritis treatment strategies such as using the qMRI technique to evaluate or enhance the performance of a new osteochondral plug.

4.8 Contributions and Applications of Findings

This thesis is the first to report differences in hip cartilage strain due to clinically relevant changes in the acetabular labrum. We aimed to achieve the highest imaging resolution possible based on a similar study reporting small changes in tibial cartilage strain following meniscectomy in a sheep knee\(^\text{52}\). The changes in mean and maximum strain (4% and 6% respectively) that we measured when comparing labral repair to resection corresponded to a change in both the mean and maximum cartilage thickness of 0.16 mm. This small change in cartilage thickness provides evidence that a high imaging resolution, such as the voxel size of 0.1 x 0.1 x 0.3 mm used in the present research, is necessary for future studies aiming to quantify a difference in cartilage strain.

Although researchers and clinicians support attempts to preserve as much of the healthy labrum as possible\(^\text{16, 28, 29, 48}\), the findings of increased cartilage strain when the labrum is
resected versus repaired may promote increased awareness of the important role of the labrum. It is generally thought that longitudinal peripheral labral tears have the potential to heal, and therefore should be repaired\textsuperscript{27,39}, however these tears are not always repaired because there are no set standards to do so. The option to resect or repair the labrum also arises in osteotomy procedures for pincer-type femoroacetabular impingement\textsuperscript{46}, where again no standards are in place to repair the labrum (if possible). The findings of the present study may motivate surgeons to enhance their knowledge on labral repair techniques. In addition, the validated technique, including the methodology to determine steady-state cartilage time-parameters could be used to assess various other surgical techniques or external bracing in the hip or in other joints such as the knee or ankle.

4.9 Conclusion

A new technique was developed and validated to measure cartilage strain in the hip joint. We demonstrated the utility of the technique for assessing the changes in hip contact mechanics after a labral tear, repair and resection. By gaining a better understanding of the relative changes in the cartilage loading environment induced by surgical procedures and comparing these changes to clinical findings, in time a better understanding of the particular loading environment that leads to the onset of osteoarthritis may be possible. With increased knowledge of the requirements of a surgical procedure to restore original joint mechanics, better treatment strategies could be employed to eliminate the progression to osteoarthritis.
References


Appendix A: Institutional Review Board Ethics Approval Certificates
Certificate of Expedited Approval
Clinical Research Ethics Board Official Notification

<table>
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<th>NAME</th>
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<td>Wilson, D.R.</td>
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UBC Campus, Vancouver Coastal Health Authority

Co-Investigators:
Gilbert, Michael, Orthopaedics; Greaves, Laura

Sponsoring Agencies:
Smith & Nephew Endoscopy Inc.

Title:
Effect of Acetabular Labral Tears, Repair and Resection on Hip Cartilage Loading: A T1 MR Study

Approval date: 24 July 2006
Term (years): 1
Documents included in this approval:
- Protocol version 1 dated 10 May 2006; Life Legacy
- Consents, Questionnaires and Donor Death Certificate
- Information

Certification:
In respect of clinical trials:
1. The membership of this Research Ethics Board complies with the membership requirements for Research Ethics Boards defined in Division 5 of the Food and Drug Regulations.
2. The Research Ethics Board carries out its functions in a manner consistent with Good Clinical Practices.
3. This Research Ethics Board has reviewed and approved the clinical trial protocol and informed consent form for the trial which is to be conducted by the qualified investigator named above at the specified clinical trial site. This approval and the views of the Research Ethics Board have been documented in writing.

The documentation included for the above-named project has been reviewed by the Chair of the UBC CREB, and the research study, as presented in the documentation, was found to be acceptable on ethical grounds for research involving human subjects and was approved by the UBC CREB.

The CREB approval for this study expires one year from the approval date.

Approval of the Clinical Research Ethics Board by one of:
Dr. Gail Bellward, Chair
Dr. James McCormack, Associate Chair
Dr. John Russell, Associate Chair
Dr. Caron Struhlendorf, Associate Chair
Certificate of Expedited Approval: Renewal

Ethics Committee on Human Research
The University of British Columbia

Ethics Certificate of Expedited Approval: RENEWAL

Certificate of Expedited Approval - Room 214, BC Cancer Centre, Vancouver, BC V 7W 1X8

Appendices
Appendix B: Loading Device Design
Figure B-3 Loading device overall dimensions
Figure B-4 Loading device details of pneumatic entry and exit
Figure B-5 Loading device flange dimensions
Figure B-6 Loading device removable endcap dimensions
Figure B-7 Loading device permanent end cap dimensions
Figure B-8 Manufacturer drawing of custom-made polypropylene pneumatic piston
Appendix C: Data Analysis
As an example, the data analysis process will be outlined for the resected labral condition.

**Step 1:** Register the resected acetabular data points to the intact loaded acetabular data points.

This step takes the resected acetabular points and the intact loaded acetabular points and registers the acetabular points together (Figure C-1).

Figure C-1 Resected acetabular points (blue) are registered to the intact loaded acetabular points (red)

Knowing the transformation matrix applied to the resected acetabular points, the femoral points of the resected contour undergo the same transformation.
Outputs of the code include the shapematched acetabular points from the resected case, the shapematched femoral points from the resected case, the transformation matrix and the shapematch error. The shapematch error is computed using an iterative closest points algorithm between the intact loaded and resected data sets.

**Step 2: Load the aiis reference file**

Next, the aiis reference line from the intact data set is loaded into Matlab. As Analyze exports pixel information only, these data points (and all others) are converted based on the imaging resolution of the MRI scan. The aiis reference line from the intact loaded data set is shown in green in Figure C-2 with the acetabular (red) and femoral (blue) contours from the resected condition.

![Figure C-2 Resected acetabular points (red) and femoral points (blue) shown with the aiis reference line (green) from the intact loaded case.](image)

**Step 3: Calculate thickness**

Code then uses the aiis points from the intact loaded case and prompts the user to enter the file name storing the data points of the shapematched resected acetabular and femoral points.
The code approximates the resected acetabular points as a sphere (Figure C-3, shown in yellow), and the intersection of the center of this sphere to the edge of the osseous acetabulum contour (aiis line), in the center slice is marked as zero degrees. The code then chooses a region of interest where radial cartilage thickness measurements are made between 0 and 45 degrees in slices 65-88. All of approximately 8,000 points are radially measured in thickness from the acetabulum to the femoral head. The error of the measured thickness direction from the perfect radial direction is logged, and the mean error in all 8,000 points, as well as the maximum error is output so that the user is aware of any inaccurate thickness readings.

The code outputs the following: error of acetabular sphere approximation, center coordinates of acetabular sphere, mean thickness, max thickness, standard deviation from mean thickness, minimum thickness, mean error from radial direction, max error from radial direction, minimum angle of the region of interest from the aiis, maximum angle of the region of interest from aiis.

Figure C-3 shows a snapshot of the measurement process. The resected femoral head contour is displayed in red, the resected acetabular contour is shown in green. The intactloaded aiis line is shown in black. The intersection of the black aiis line with the acetabular cartilage is marked as zero degrees from the center of the acetabular sphere. From 0-45 degrees, the code computes radial cartilage thickness measurements in all points. A line is drawn in red from the center of the acetabular sphere to the femoral point of interest, marking the perfect radial direction. The black line extending from this red line shows the measured radial thickness.
direction toward the acetabulum for a given femoral point. The closer this black line extends linearly from the red line, the more representative the thickness is in the radial direction.
Figure C-3 Various views of the thickness measurement process. Femoral points are shown in red and acetabular points are shown in green. The red lines join the point of interest to the center of the acetabular sphere, marking the perfect radial direction. The black line shows the corresponding measured thickness direction from the femoral point to the acetabular point.
Step 4: Generate thickness distribution map

Next, the code outputs a three-dimensional thickness colour map of the region of interest for the resected case, mapped onto the femoral contour.

![Sample cartilage thickness map](image)

Figure C-4 Sample cartilage thickness map (in millimeters) for specimen 5, resected case

This same data analysis procedure outlined above for the resected case is completed for the repaired, torn, intact loaded (without the registration step) and intact unloaded cases, generating a thickness map for all conditions of the labrum.

Step 5: Register to unloaded

To convert the resected labrum cartilage thickness distribution map to a strain distribution map, the following formula was used at each of the 8,000 data points:

\[
\text{Strain} = \frac{\text{unloaded thickness} - \text{loaded thickness}}{\text{Unloaded thickness}} \times 100\% \quad [1]
\]
For reference, the unloaded thickness map for specimen 5 is shown below:

![Unloaded cartilage thickness map](image)

Figure C-5 Unloaded cartilage thickness map (in millimeters) for specimen 5

The first step toward generating a strain distribution map is to register the resected cartilage thickness matrix to the unloaded cartilage thickness matrix. This is accomplished with registering the femoral points of the resected condition to the unloaded condition. Assuming no deformation of the osseous surface of the femoral head, the contour shape of the femoral head should be identical in both the unloaded and loaded condition. Images of the registered unloaded (red) and resected (blue) femoral points are shown in Figure C-6.
Step 6: Subtract resected thickness values from unloaded thickness values.

The code computes the center of the reference (i.e. unloaded) acetabular points approximated as a sphere, and then for each femoral point of the resected points, the code computes the closest unloaded point based on minimizing both the distance between the two points and the angle (in three dimensions) between the two points. The error associated with the distance of the unloaded point is logged. A filter is then used where points that had too high an error to accurately report a strain value are omitted from the strain calculation.

From all 8,000 points, mean strain, maximum strain, minimum strain, standard deviation from mean strain as well as the mean and maximum error associated in choosing the corresponding unloaded point are output.

The resulting strain map for the resected case is shown in Figure C-7, as well as the thickness maps that generated this strain distribution.
Figure C-7 (Above) Cartilage strain distribution map for specimen 5 in the resected case. (Lower left) Unloaded cartilage thickness map for specimen 5, (Lower right) loaded and resected cartilage thickness map for specimen 5. Thickness maps in millimeters.
Appendix D: Individual Cartilage Thickness and Strain Distribution Maps
Figure D-1 Cartilage thickness distribution maps (in mm) for specimens 1 through 3 with various conditions of the labrum. Relatively thin cartilage is displayed in red, and relatively thick cartilage is displayed in blue.
Figure D-2 Cartilage thickness distribution maps (in mm) for specimens 4 through 6 with various conditions of the labrum.
Figure D-3 Cartilage strain distribution maps for specimens 1 through 3 with various conditions of the labrum.
<table>
<thead>
<tr>
<th>INTACT</th>
<th>TORN</th>
<th>REPAIRED</th>
<th>RESECTED</th>
</tr>
</thead>
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<td><img src="image2.png" alt="Image" /></td>
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<td><img src="image11.png" alt="Image" /></td>
<td><img src="image12.png" alt="Image" /></td>
</tr>
</tbody>
</table>

Figure D-4 Cartilage strain distribution maps for specimens 4 through 6 with various conditions of the labrum.
The population averaged strain maps were generated by registering all strain maps with the same labral condition together, and averaging strain measurements based on closest radial points. As a check that the population averaged strain maps were accurate, all data points from individual (specimen 1 through 6) strain maps were averaged, and compared to the average strain from the population averaged strain maps. Results are listed in Table F-1, with all errors of approximately 1%, there was no preference for any labral condition.

![Figure D-5 Population averaged cartilage strain distribution maps for various conditions of the labrum](image)

<table>
<thead>
<tr>
<th></th>
<th>Individual Average strain</th>
<th>Population average strain</th>
<th>Error</th>
</tr>
</thead>
<tbody>
<tr>
<td>Intact</td>
<td>32.96%</td>
<td>32.60%</td>
<td>-1.09%</td>
</tr>
<tr>
<td>Torn</td>
<td>32.38%</td>
<td>32.01%</td>
<td>-1.14%</td>
</tr>
<tr>
<td>Repaired</td>
<td>30.42%</td>
<td>30.24%</td>
<td>-0.59%</td>
</tr>
<tr>
<td>Resected</td>
<td>34.41%</td>
<td>34.11%</td>
<td>-0.87%</td>
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</table>

Table D-1 Average strain computed by averaging all individual data points compared to average strain found by registering strain maps and computing average strain measurements based on radial distances.
Appendix E: Fiducial Movement with Load Application
Figure E-1 Typical specimen rotation induced by compressive load application. (Left) MRI scan showing fiducial placement before load was applied in specimen 6 (upper) and specimen 5 (lower). (Right) MRI scan showing fiducial placement after load was applied in specimen 6 (upper) and specimen 5 (lower).
Appendix F: Quantification of Various Sources of Error
Source of Error 1: Variation in Location of Center of Acetabular Sphere

The variation in the location of the center of the acetabular sphere creates error as the center of the acetabular sphere sets the radial direction in cartilage thickness measurements. The approximation lies in the assumption that the osseous surface of the acetabulum does not change from scan to scan, or due to varying the labral condition. In actuality, error could be introduced through changes to the osseous rim of the acetabulum due to drilling when inserting the suture-anchor, or perhaps due to day-to-day variation in the osseous deformation, which was assumed to be zero.

Figure F-1 shows the center of the approximated acetabular sphere, with data from each
specimen plotted as a different colour. Five data points exist for each specimen (with the exception of specimen 2 where the torn state was not assessed), with the data points consisting of the center of the acetabular sphere when the specimen was unloaded (circled) and when the specimen was loaded while the labrum was intact, torn, repaired and resected. The greatest variation tended to be when the labrum was unloaded versus loaded, likely due to the slight rotation or osseous deformation induced by the load application. Ideally, within each specimen the center of the sphere would not deviate substantially, intra-specimen variation in sphere center location is expected, and does not contribute to inter-specimen measurement error.

The quantified error due to the variance in location of the center of the acetabular sphere is reported in each specimen in Table F-1, with standard deviations all below 1.09 mm. Note this error cannot be directly translated to cartilage thickness measurement error, as this error contributes to a vector orientation error in thickness measurement, and therefore cartilage thickness measurement error would be only a small fraction of the error associated with the variance in the location of the center of the sphere.

An attempt to employ the same spherical center location among labral condition was made, however this method produced unfavorable results as it led to inaccuracies when subtracting images in the creation of strain measurements.

Though the choice of measuring thickness radially introduced error due to this variance in the location of the center of the acetabular sphere, radial thickness measurements are superior than iterative closest point algorithms as the direction of the thickness measurement vector is forced in the same direction each time which plays a key role when comparing cartilage
thickness distribution maps, and results in a much higher accuracy when subtracting thickness measurements point-by-point toward the creation of a cartilage strain distribution map.

Table F-1 Variation in the location of the center of the acetabular sphere across various specimens, and various labral conditions

<table>
<thead>
<tr>
<th>Specimen</th>
<th>Intact</th>
<th>Torn</th>
<th>Repaired</th>
<th>Resected</th>
<th>Mean</th>
<th>Std</th>
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**Source of Error 2: Registration to Load Error in Thickness Measurement**

In order to create a cartilage thickness distribution map for each labral condition within the same specimen, it is important that qualitative comparisons across each cartilage thickness distribution map are possible. In order to do this, the data points from each labral condition are registered to the condition when the labrum was intact and the joint was loaded, this allows sharing of a common reference point, the aiis, as well as a common orientation.

The registration error, computed with an iterative closest point algorithm between the reference image and current data set, was logged in all specimens and is shown in Figure F-2.

![Figure F-2](image)

**Figure F-2** Error associated with registering a given image to the intact loaded image across various specimens and labral conditions.

*Figure F-3* illustrates the average registration error for each labral condition. It is evident that all labral conditions had a similar registration error (within <0.1 mm).
Figure F-3 Average registration error across all specimens when joint was unloaded and when the joint was loaded with a torn, repaired or resected labrum.

**Source of Error 3: Registration to Unload Error in Strain Measurement**

In order to accurately subtract each thickness value from the thickness map of the comparison case from the thickness map of the unloaded case, the geometry of the region of interest must be mapped together through registration. The error associated with registering the reference femoral contour to the unloaded femoral contour was logged and presented in Figure F-4. Figure F-5 illustrates the average measurements, where it is clear that there was no difference in registration to unload error among labral condition.
Figure F-4 Registration to unload error across all specimens and shown for each labral condition.

Figure F-5 Average registration to unloaded data error across all specimens, shown for each labral condition.
Source of Error 4: Error from Global Cartilage Deformation

Another method of measuring error in the data analysis process is to compute the mean cartilage deformation based on the mean thickness values of the unloaded vs. comparison cases using the formula:

\[
\text{Mean Cartilage Deformation} = \text{Mean unloaded thickness} - \text{Mean loaded thickness} \[2\]
\]

The mean cartilage deformation computed from a single calculation using formula [2] (referred to here as the “global mean deformation value”) should, in theory, be equal to the mean cartilage deformation computed from the code, which computed the amount of deformation on a point-by-point level, and then calculated the average deformation. The amount of cartilage deformation computed from the code compared to the amount of cartilage deformation computed from the thickness data is shown in Figure F-6, units of millimeters. Average error from global is shown in Figure F-7, where there is no qualitative change among labral condition.
Figure F-6 Error in mean cartilage deformation computed based on point-by-point measurements, compared to the global mean deformation calculated using mean thickness measurements.

Figure F-7 Average error from global deformation for each labral condition based on data from all specimens.
Source of Error 5: Error Associated with Creating a Cartilage Thickness Map

Cartilage thickness is measured in the radial direction, though it is possible through improper sampling of the osseous contours, imaging resolution, or problems around the edges, that measurements can deviate substantially from the radial direction. The chosen point on the opposing (acetabular) surface in the radial direction is based on optimizing the angle offset of the chosen point from the target point, and the distance the chosen point is away from the target point. The mean and maximum distance (mm) the chosen point was away from the target point were computed in all data sets and are presented in Figure F-8 and Figure F-9, respectively. From specimen to specimen, there was very little change in this error among labral condition.

![Graph showing mean distance chosen point was from the pure radial direction across the entire data set when generating the cartilage thickness distribution map](image-url)
Source of Error 6: Error Associated with Creating a Strain Map

Cartilage strain is measured based on registering the comparison image to the unloaded image, with thickness subtractions based on the closest point in the radial direction. As a means of assuring the correct points were subtracted from one another, the amount of deviation from the radial direction was calculated based on the distance between the chosen point and the target point. This error calculation was based on optimizing the angle offset of the chosen point from the target point, and the distance the chosen point was away from the target point. The mean and maximum distance (mm) the chosen point was away from the target point were computed in all data sets and are presented in Figure F-10 and Figure F-11, respectively. From specimen to specimen, there was very little change in this error among labral condition except perhaps specimen 3 where the resected case had a slightly higher mean radial error, however
putting this error value into context, the distances the chosen points were away from the pure radial direction were all still less than 0.1 mm.

Figure F-10 Mean distance chosen point was from the pure radial direction across the entire data set when generating the cartilage strain distribution map

Figure F-11 Max distance a chosen point was from the pure radial direction within the data set when generating the cartilage strain distribution map
**Source of Error 7: Quantifying the Sphericity of the Acetabulum**

The acetabulum is approximated as a sphere in order to measure thickness radially. Figure F-12 reports the percentage error that the segmented area of the acetabulum deviates from a true sphere.

It is important to note that the error associated with approximating the acetabulum as a sphere was less than 5% throughout all measurements within the study. Further, within each loaded specimen, there was little difference (<1.25%) in the sphericity error among labral condition.

![Figure F-12 Error of segmented region of acetabulum from true sphericity in all specimens](image-url)
Interestingly, the largest change in the acetabular sphericity occurs due to load. This is evident in specimens 1, 3 and 5. This may suggest deformation of the subchondral bone due to application of the 2.3 x BW compressive load, however as this change in sphericity is not evident in all specimens, data collection from more specimens is recommended to thoroughly investigate this phenomenon.
Appendix G: Importance of Using Phosphate Buffered Saline Solution with Protease Inhibitors
Phosphate buffered saline (PBS) (pH 7.4) with protease inhibitors is generally thought to be an ideal bathing solution for articular cartilage because of the ion balance of the phosphate buffered saline and the ability of the protease inhibitors to minimize cartilage degradation.

Proteoglycan aggregates in cartilage contain a large number of negatively charged groups (SO\(^3^-\) and COO\(^-\)) along their glycosaminoglycan chains. When the total ion concentration inside the tissue is greater than the ion concentration in the external bathing solution, the imbalance of ions gives rise to a pressure in the interstitial fluid that is higher than the external bath, causing cartilage swelling.

The cartilage biphasic theory suggests in creep, the analytic solution for the change in cartilage thickness in the axial direction is obtained as defined below:

\[
\left(\frac{\mu(0,t)}{h}\right) = \frac{F_0}{H_\alpha} \left\{1 - 2 \sum_{n=0}^{\infty} \frac{\exp\left(-\left(n + 0.5\right)^2\pi^2H_\alpha t/h^2\right)}{(n + 0.5)^2\pi^2}\right\}
\]

Where \(h\) is the original thickness of the cartilage, \(F_0\) is the applied compressive stress, \(H_\alpha\) is the aggregate modulus and \(k\) is the permeability coefficient of the cartilage. An increase in the dimensions of the cartilage due to swelling, will result in an increase in fluid content and thus increasing the porosity and permeability of the solid matrix, which influence the mechanical behavior of cartilage in compression. Phosphate buffered saline with a pH of 7.4 has the same ion concentration as blood, and therefore it is thought that this fluid does not create a pressure rise in the interstitial fluid, avoiding cartilage swelling.
We used a specific concentration of protease inhibitors. We used 2 mM of phenylmethylsulfonyl fluoride (PMSF), 2mM ethylenediaminetetraacetic acid (EDTA), 10 mM N-Ethylmaleimide (NEM) and 5mM Benzamidine-hydrochloric acid (HCl). These protease inhibitors have been used previously to minimize cartilage degradation in biomechanical studies\(^1,2,6,7\). Cartilage degradation affects the permeability of the cartilage, and the aggregate modulus\(^2,5\), both of which influence the mechanical behavior of cartilage in compression.

Chondroitinase-ABC is an enzyme that targets glycosaminoglycan chains of proteoglycans, and is often used to simulate cartilage degeneration. One study showed the stiffness of articular cartilage bathed in chondroitinase-ABC decreased to approximately 70% of the original stiffness value 20 hours after the enzyme was added\(^2\). However, the stiffness of the cartilage remained at 100% of the original value without chondroitinase-ABC and with protease inhibitors and PBS solution\(^2\).

Cartilage degradation and cartilage swelling due to an improper bathing solution change the behavior of cartilage in compression, and therefore a solution of PBS with protease inhibitors is recommended. The control specimen described in chapter 2 and 3 incorporated the effect of PBS solution on cartilage mechanics, and showed reasonable repeatability (<4%).

References:
