A biomechanical basis for tears of the human acetabular labrum

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ABSTRACT

Objective: Acetabular labral tears predominantly affect young patients and are a source of hip pain in the athlete. Four causes of the initiation of labral tears have been proposed; trauma, hypolaxity of the anterior capsule, dysplasia and bony impingement. A further cause could be reduced biomechanical properties in the area most susceptible to tears. However, no work has defined these properties.

Design: 32 compressive and 32 tensile test samples were harvested from fresh-frozen cadaveric acetabula. The labrum was divided into eight areas to allow comparison around its circumference. Semiconfined compressive testing and tensile testing were performed at a displacement rate of 10 mm/min in a controlled environment of 100% humidity at 37 (SD 1)°C.

Setting: Cadaveric study.

Results: The mean compressive stiffness was 31.75 (SD 16.7) MPa, and the mean tensile elastic modulus was 24.7 (SD 10.8) MPa. The anterosuperior region had a significantly lower compressive elastic modulus than either of the posterior quadrants ($p<0.05$) and a significantly lower tensile modulus to the anterioinferior area ($p<0.05$).

Conclusions: The biomechanical properties in the anterosuperior region may be a contributing factor to the initiation of labral tears.

Labral tears can occur in any age group but mostly affect young adults.[1–4] Usually the symptoms are insidious in onset but may be acute or related to a traumatic event.[3 5] They have been reported to be a source of hip pain in the athlete[4 6–8] and footballers.[9] Pain is usually the main symptom, but patients may present with restricted range of movement, catching or clicking and even instability.[10] The most common area of labral tears appears to be in the anterior–superior region[11–14] and usually involves the articular margin, as opposed to the capsular margin.[15] However, both posterior and posterior–superior tears do occur.[2 16]

Four causes of labral tears have been proposed: trauma, laxity/hypermobility, bony impingement and dysplasia.[17] Trauma has been present in the history of up to 40% of patients with an arthroscopically diagnosed labral tear.[3] Hypolaxity of the anterior capsule, specifically of the iliofemoral ligament, has been suggested as a cause of increased loading of the anterosuperior labrum and found in 23% of 300 cases of labral tears.[18] The hypothesis can be extended to suggest that the increased load leads to labral tearing.

Bony impingement can result from a decreased femoral head–neck junction offset (cam effect), an overhang of the anterior superior acetabular rim (pincer lesion), a retroverted acetabulum or a combination of these bony deformities. The term femoracetabular impingement (FAI) is commonly used to describe the morphological abnormalities of both the acetabulum and proximal femur, which may be a cause of acetabular labral tears.[19 20] Arthroscopic intervention to treat these lesions, especially in athletes, is gaining popularity.[20–22] Even mild dysplastic changes within the hip may be a risk factor for developing labral tears,[23 24] and if the tear is untreated it has been suggested that this could progress to damage of the adjacent chondral surfaces, resulting in eventual osteoarthritis.[12 15 25]

Two distinct types of tears have been described from histological studies: mid-substance tears and detachment of the labrum from the articular hyaline cartilage in the transitional zone.[12]

The labrum itself is a fibrocartilagenous structure with circumferential orientated fibres attached to the acetabular rim and the transverse acetabular ligament.[26] From 9 weeks in utero, the transverse ligament and labrum can be seen continuously around the margin of the acetabulum.[27] It deepens the osseous margin of the acetabulum and extends coverage of the femoral head. The acetabular labrum appears to be well fixed to the underlying bone, with a projection of the bony acetabulum into the undersurface of the labrum and zone of calcified cartilage with a well-defined tidemark.[12]

The labrum increases the surface area and volume of the acetabulum by approximately 60% and 120% respectively.[28] Histological examination has shown the labrum to contain nerve endings, mostly in the superficial articular region.[26 29] Therefore, the structure may play a role in proprioception. The blood supply is from a peripheral anastomosis surrounding the capsular attachment,[26] with blood vessels restricted to the outer third of the labrum.[30] As such, the deep labral structure is relatively avascular.

Scanning electron microscopy (SEM) has revealed three distinct layers: a meshwork of thin fibrils on the articular surface, a deep region that is more layered and the most substantial inner layer of highly circumferentially orientated fibres which, from appearances, carries most of the physiological load. These structures have been postulated to accommodate significant physiological hoop stresses.[30 31]

The actual function of the labrum within the hip joint is debatable and may be multifactorial. Ferguson *et al*[32] focused on the labrum sealing the articular fluid under pressure to produce a fluid film which protects the underlying cartilage and provides a low friction articulation. The labrum also adds resistance in the flow path of fluid being expressed from the cartilage layers of the hip joint under load and therefore enhances the retention of interstitial fluid within the tissue and slows deformation of the cartilage. These effects seem to be most influential on the response of the hip joint to load in the first 500–1000 s after loading.[33] It has also been described as having a valve effect maintaining pressure within the joint.[34] It does not appear to transmit any substantial load across it, as removal of the acetabular labrum does not appreciably change the contact area, load, and mean pressure between the acetabulum and femoral head.[35]

A final cause for the majority of labral tears taking place in the anterosuperior quadrant could be reduced biomechanical properties in this area. One study has reported the mean tensile modulus of just the superior quadrant from human specimens undergoing hip arthroplasty,[36] but no work has been produced on the whole of the labrum and comparisons within it.

Therefore, the aim of this work was to define the compressive and mechanical properties of the acetabular labrum and to evaluate whether there is any change in its mechanical properties around the circumference. Importance was placed on ensuring the samples were fresh-frozen, non-degenerative and tested under physiological conditions.

MATERIALS AND METHODS

This study was approved by a local ethics committee, and eight fresh-frozen cadaveric hip joints were obtained from a tissue bank. The specimens were stored at −20°C when not in use and sprayed with physiological saline during dissection. All the specimens were allowed to thaw thoroughly overnight prior to use. Several groups have found no significant differences in mechanical properties of collagenous tissue after frozen storage.[37–40]

The capsule was excised from its insertion on the femur in order to allow safe disarticulation of the joint without intra-articular iatrogenic damage. The articular surfaces were graded according to the Outerbridge classification,[41] and the gross anatomy and dimensions of the labra were recorded. Any hip that had changes greater than grade I to either articular cartilage surface was discarded from the rest of the study. The acetabulum was then tattooed at each end of the transverse ligament to allow orientation, and the labrum was dissected from any remaining capsule, articular cartilage and underlying bone.

Each labrum was then divided into eight equal sections, in order to allow comparison between them (fig 1). From the previous SEM studies,[30 31] it was known that a substantial area within the labrum is made up of a core region which can be identified by the naked eye. As this appeared to be the most homogeneous layer in which the majority of fibres were co-oriented, it was decided to sample from this region. A custom-made cryo-clamp attached to a microtome was used to prepare the test specimens, ensuring that the cut would run in line with the circumferential collagen fibres. A modified freeze–fracture technique[42] allowed accurate manufacture of the test samples while ensuring that the cut ran in line with the circumferential collagen fibres; this has been shown to not alter the material properties of the tissue.[42]

Figure 2 shows the cross-section of a single portion of the labrum with the core region and the orientation of a tensile test sample within it. All cuts were made along the long axis of the labrum sample and parallel to the long axis of the clamp.

The cutting protocol produced test samples with dimensions of 1×1×8 mm for tensile testing and 3×1×6–9 mm for compressive testing. After each test sample was cut, it was measured while still frozen using electronic callipers (Mitutoyo, Andover, UK); any test samples that deviated by more than 10% from these dimensions were discarded from further testing. By using the central core area, both a compressive and tensile test sample could be harvested from all positions around the labrum.

All testing was carried out on an Instron materials testing machine (Instron 5565, High Wycombe, UK), with accompanying Merlin software (v.4). A 10 N load-cell (accurate to within ±0.25% of load applied) was used; this was calibrated automatically before each testing session. Both compressive and tensile experiments were conducted in an environmental chamber at 37 (SD 1)°C and 100% relative humidity, ensuring that the material properties and water content did not alter during experimentation.[42] The test sample was allowed to adjust to the environmental conditions for 5 min prior to testing. Both the compressive and tensile testing regimes were designed to recreate physiological loading. Therefore, both included a period of precycling to precondition the tissue after its frozen preservation state.[37] All testing was performed at a constant 10 mm/min displacement rate; the sampling frequency was 20 Hz.

The tensile test samples were mounted onto specially designed testing clamps; the clamping area was covered with fine-grain sandpaper to prevent slippage. The tensile clamps were adjusted manually to ensure that the test sample was bearing a tensile tare load of 0.01 N, thus being neither significantly preloaded nor buckled; this position was set to 0 mm. The tare load was used only as an indication that the test sample was not buckled immediately prior to the test commencing.

A precycling regime starting at 0 mm rising to 0.1 N load and returning back to 0 mm displacement was used. All testing was performed at a constant 10 mm/min displacement rate. The precycling part of the test was terminated when a quasistatic state had been reached, that is when the hysteresis curves were overlapping. This was defined as a difference of less than 0.01 mm measured at the ascending 0.05 N load position between two consecutive cycles.

The main tensile testing commenced with loading the test sample up to 1 N also at 10 mm/min and allowing stress relaxation for 5 min; this was achieved by keeping the displacement constant throughout this time. This was followed by loading the test sample up to 1 N also at 10 mm/min. After this period, the test sample was returned to zero displacement and then was loaded to failure or to an upper limit of 7 N; this limit had been chosen to protect the load cell. Once finished, a retrospective measurement of the distance between the two clamps was made to calculate the working length for later analysis.

For the compressive testing, each test sample was mounted into a stainless steel well. The well was machined to a depth of 1 mm and width of 3 mm, allowing an exact fit of the test sample within it. Both ends of the test samples were not enclosed, allowing some free movement of fluid in and out of the sample during testing. The indenter was a 3 mm diameter stainless steel sphere centred on the test sample.

The precycling regime for the compressive test samples started at 0 mm rising to 0.5 N load and returning to 0 mm displacement. The precycling part of the test was terminated when a quasi-static state had been reached; this was defined as a difference of less than 0.01 mm measured at the ascending 0.25 N load position between two hysteresis curves.

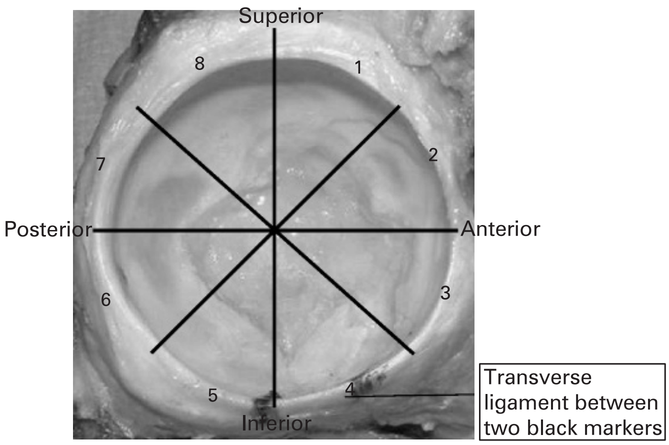

Figure 1 Gross anatomy of the acetabular labrum. The two ends of the transverse ligament are marked with black ink.

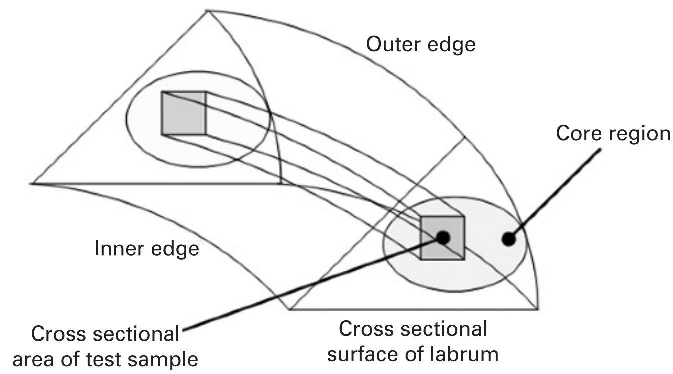

Figure 2 Diagram of a single labral portion with a tensile test sample orientated circumferentially within the core region.

The main testing commenced with loading the test sample up to 1.5 N. The test sample was subsequently allowed to stress-relax for 5 min while the indenter was held stationary at the 1.5 N load position. After this period, the test sample was returned to zero displacement and then was loaded up to 3 N. A second 5 min period of stress relaxation was allowed, and the sample was returned again to zero displacement. Finally, the sample was loaded up to 5 N; a maximum of 5 N was necessary to protect the load cell from damage.

The elastic modulus of the tensile test sample was calculated from the linear portion of the final stress–strain curve; that is after the second period of stress relaxation. The yield stress and the maximum stress were also calculated from this curve as the stress value at the end of the linear portion and the maximum stress value exhibited, respectively.

The compressive stiffness of each compressive test sample was determined from the linear portion of the stress–strain curve at final loading; that is after the second period of stress relaxation.

The strain was recalculated after the second stress relaxation in both tensile and compressive testing due to the change in sample length. The moduli were obtained using linear regression analysis.

The data were compared for significant difference using a one-tailed paired Student t test with an alpha level of 0.05.

RESULTS

Out of the eight hips dissected, four had significant degenerative changes (Outerbridge grade II and above) and were excluded from further testing. This left left right and two left hips for testing with a mean age of 74.8 (range 66–80) years at time of death, none of which were known to suffer from hip pathology. All except one were donated from male patients. Two of the labra had radial fibrillation at the 1 and 11 o'clock positions, respectively, but no other signs of degeneration or tears. The mean measured labral circumference was 168 mm (range 147–181). The width of the labrum varied from a minimum of 2.1 mm to a maximum of 7.6 mm among the specimens.

After cutting, 32 test samples were available for tensile testing and 32 test samples for compressive testing. The mean tensile modulus was 24.7 (SD 10.8) MPa, the anterosuperior area (portions 1 and 2) having a significantly lower tensile modulus to the anterioinferior area (portions 3 and 4) ($p<0.05$), but no significant difference was found between the other quadrants.

The mean compressive stiffness was 31.8 (16.7) MPa. The anterosuperior region (portions 1 and 2) had a significantly lower compressive elastic modulus than either of the posterior quadrants (portions 5 and 6, and portions 7 and 8) ($p<0.05$) (fig 3).

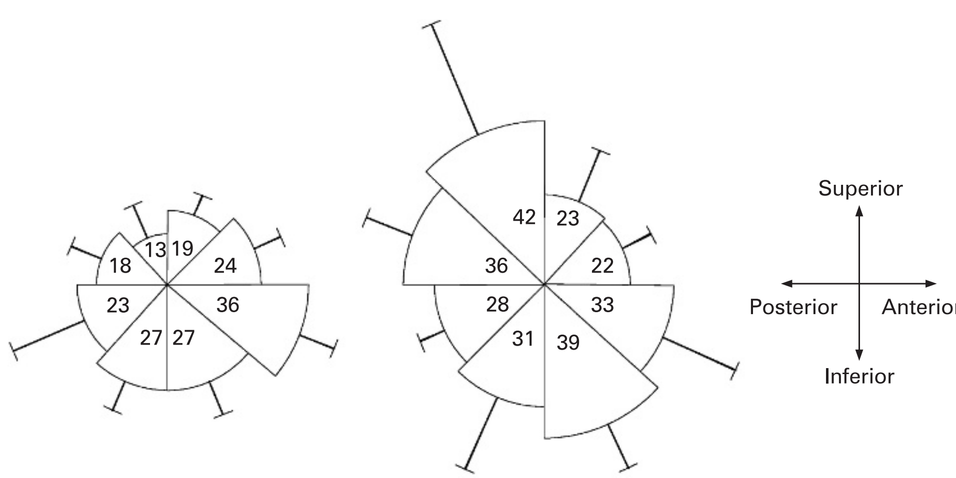

Figure 3 Mean tensile modulus (A) and mean compressive stiffness (B) for each region around the circumference of the labrum. All values are in MPa; error bars show the SD.

DISCUSSION

This paper has presented the tensile and compressive behaviour of the human acetabular labrum around its circumference. The testing was performed on fresh frozen, non-degenerative samples under physiological conditions. This has not been presented before.

There is significant variation around the circumference of the labrum, and the anterosuperior quadrant does have a lower compressive and tensile modulus to some, but not all of the other quadrants. This has significance in the clinical scenario of acetabular tears, which appear more frequently in this region, and it may be concluded that these reduced biomechanical properties are a contributing factor to the propagation of tears.

In reality it is likely that the initiation of tears is multifactorial and that more than one factor needs to be present in order to create a tear. Trauma could affect any part of the labrum, but these results suggest that the anterosuperior quadrant is more susceptible. Hypolaxity of the anterior capsule and the ilio-femoral ligament will increase the tensile loading and possibly shear loading of the anterosuperior labrum. Bony impingement and dysplasia both cause increased compressive loading on this portion. Any of these pathologies combined with the reduced biomechanical properties could lead to failure of the labrum in this region.

The only other published work on the biomechanical properties of the human acetabular labrum produced a tensile modulus of 26.2 (9.1) MPa for the superior quadrant from samples harvested at the time of hip arthroplasty.[36] Therefore, the labrum cannot be defined as "normal." An attempt was made to exclude degenerative specimens in this study through the use of an accepted tool, the Outerbridge classification.[41] Moreover, in our work, the experiments were conducted in a controlled environment, as both hydration status[43 44] and temperature[45] have been proven to alter the biomechanical properties of intra-articular tissue. Therefore, the results of the two papers are not directly comparable and explain the difference in results obtained.

There are limitations to our work. These samples were taken from an older population than the sports players that are affected by acute tears. However, obtaining complete labral specimens from patients in a younger age group is obviously difficult, and it would be hoped that by excluding degenerative samples, these results reflect the mechanical properties of a healthy labrum. Although 64 results were produced from this study, four of the labrums had signs of degeneration and were excluded from testing, and the number of test results was reduced. The authors felt strongly that only healthy specimens be used and accept that the numbers are smaller than was hoped.

The standard deviations for the specimens are large in this study. This is expected in biological tissue and could be reduced by increasing the number of specimens used. However, these limitations do not negate the results presented, which have been statistically robust enough to produce the significant differences presented.

Finally, by electing to test just the core layer of the acetabular labrum, the authors have limited the interpretation of the results. The core layer is by far the largest of the three layers within the substance of the labrum and undoubtedly has the greatest mechanical strength due to its superior collagen structure. This layer was chosen to allow a homogenous sample to be tested and hopefully avoid errors that would have arisen if the whole specimen had been tested due to size differences and shape variation. From the SEM appearances, it is intuitive that this is the only layer with any significant physiological loading[30] and any clinically significant tear would involve this layer.

Even with these limitations, this work is significant, as this is the first time that the compressive and tensile properties for the human acetabular labrum have been presented in the literature. Even with the small numbers, significant differences have been shown which reflect clinical findings and suggest another possible aetiological factor contributing to these lesions.

In conclusion, this study has enabled a comparison of the tensile elastic moduli and compressive stiffness of the acetabular labrum to be made around its circumference under controlled conditions. The results produced suggest that the biomechanical properties in the anterosuperior region may be a contributing factor to the initiation of labral tears.

Competing interests: None.

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