Tensile Strain in the Anterior Part of the Acetabular Labrum During Provocative Maneuvering of the Normal Hip

Christopher J. Dy, Matthew T. Thompson, Matthew J. Crawford, Jerry W. Alexander, Joseph C. McCarthy and Philip C. Noble


---

**Supplementary material**
Commentary and Perspective, data tables, additional images, video clips and/or translated abstracts are available for this article. This information can be accessed at [http://www.ejbjs.org/cgi/content/full/90/7/1464/DC1](http://www.ejbjs.org/cgi/content/full/90/7/1464/DC1)

**Reprints and Permissions**
Click here to order reprints or request permission to use material from this article, or locate the article citation on [jbjs.org](http://www.jbjs.org) and click on the [Reprints and Permissions] link.

**Publisher Information**
The Journal of Bone and Joint Surgery
20 Pickering Street, Needham, MA 02492-3157
[www.jbjs.org](http://www.jbjs.org)
Tensile Strain in the Anterior Part of the Acetabular Labrum During Provocative Maneuvering of the Normal Hip

By Christopher J. Dy, MD, MSPH, Matthew T. Thompson, MS, Matthew J. Crawford, DO, PhD, Jerry W. Alexander, BS, Joseph C. McCarthy, MD, and Philip C. Noble, PhD

Investigation performed at the Institute of Orthopedic Research and Education and The Methodist Hospital, Houston, Texas

Background: Injury of the acetabular labrum is a well recognized cause of hip pain in the young, active patient. The exact mechanism of these injuries remains a subject of speculation, although femoroacetabular impingement and twisting maneuvers have both been proposed as critical factors. We examined the hypothesis that torsional maneuvers of the morphologically normal hip joint generate mechanical strain within the acetabular labrum, particularly in areas that are prone to injury.

Methods: Seven human cadaver specimens were loaded during five separate maneuvers with external rotation or abduction torques applied to the hip in neutral alignment and in moderate flexion or extension. Tensile strain within the acetabular labrum was measured with use of the technique of roentgen stereophotogrammetric analysis.

Results: Substantial tensile strains were generated within the labrum during each of the loading maneuvers, with no significant difference in strain being noted between the maneuvers. Maximum strain in the anterior part of the labrum averaged 13.6% ± 7.8% in the axial direction and 8.4% ± 3.0% in the circumferential direction. The highest mean and maximum strain values were found at the two o’clock position of the labrum, with the highest strain concentration at the bone-labrum interface.

Conclusions: External rotation and abduction maneuvers of the morphologically normal human hip joint in moderate flexion or extension can generate substantial tensile strains in the anterior part of the acetabular labrum. This finding supports the hypothesis that injury to the anterior part of the labrum may occur from recurrent twisting or pivoting maneuvers of the hip joint in moderate flexion or extension without femoroacetabular impingement.

Clinical Relevance: The substantial amounts of tensile strain generated during loading of the morphologically normal hip are indicative of a mechanical process that may accelerate the onset of degenerative disease.

The recent increase in interest in the diagnosis and treatment of lesions of the acetabular labrum reflects the importance of this structure and its role in normal joint function. The labrum is an avascular fibrocartilaginous rim of tissue that encases the femoral head within the acetabulum and restricts motion of the hip, particularly at the extreme ranges of motion. The labrum forms a seal through contact with the articular surface. This sealing function limits the displacement of synovial fluid, which limits loading of the soft tissues of the hip, especially during impact loading. Injury to the labrum is thought to negatively impact the mechanical environment of the hip joint, possibly leading to degenerative changes. Recent investigations have confirmed the role of the intact labrum in enhancing joint stability as disruptions of the labral seal have led to subluxation of the femoral head at the extremes of joint motion.

Observational studies have demonstrated an association between labral injuries and chondral damage, suggesting that a continuum of joint disease exists, beginning with labral lesions and leading to irreversible degenerative joint disease.
Damage to the labrum can occur from a number of sources, including direct trauma to the hip, internal degeneration with aging, and abnormal mechanical wear secondary to hip dysplasia. Labral lesions also have been documented as a common occurrence in anatomic examinations of cadaver specimens, indicating that eventual deterioration of the labrum may be associated with the aging process.\(^5\) Arthroscopic observations also have confirmed the presence of labral injuries in athletic patients with mechanical hip symptoms without radiographic evidence of trauma, dysplasia, or degenerative joint disease.\(^6\) The exact mechanism of labral injury in these clinical scenarios has not yet been clearly delineated. McCarthy et al. proposed that recurrent twisting and pivoting motions may lead to recurrent labral microtrauma, whereas Ganz et al. contended that labral tears are secondary to mechanical impingement of a morphologically abnormal hip joint. Additional investigations of the pathomechanics of labral injury are warranted in light of the potential consequence of irreversible joint degeneration and the opportunities for earlier diagnosis and therapy.

Despite advances in the diagnosis and treatment of labral lesions, there is limited knowledge about the biomechanical properties of the intact labrum. Previous analyses of the labrum include finite element models\(^3\) and isolated tissue studies\(^1\), but the biomechanics of the intact labrum have not yet been described. Ferguson et al. used finite element models\(^2\) to describe the material properties of the bovine labrum and the potential influence of the labrum on joint function, whereas Ishiko et al.\(^7\) described the mechanical properties of the labrum when isolated from the hip joint. An initial clue toward the pathological process causing labral lesions may lie in the previous demonstration of a predominance of lesions in the anterior portion of the labrum and the ensuing suggestion that this area of the labrum may be subject to increased mechanical demands or may be intrinsically weaker. The purpose of the present study was to describe and quantify strain within the intact labrum as it undergoes physiologic mechanical demands, with the goal of determining whether there are indeed areas of the labrum that are predisposed to injury. An experimental study was performed to test the hypothesis that torsional maneuvers of the hip joint generate substantial amounts of mechanical strain within the acetabular labrum, particularly in areas that are prone to injury.

**Materials and Methods**

Seven fresh human hip joint specimens consisting of the hemipelvis, the femur, and overlying soft tissues were obtained from seven cadavers; the donors included six men and one woman who had had a mean age (and standard deviation) of 79 ± 11 years (range, 66 to 92 years) at the time of death. All specimens were screened radiographically to confirm the absence of osseous abnormality. All muscle and soft tissue was removed from each specimen by means of careful dissection, with the hip capsule and labrum being left intact. The frontal plane of the pelvis was defined by a plane target to the pubic symphysis and the anterior superior iliac spine. The epicondylar axis of the femur, based on the most apparent point of each protuberance, was aligned parallel to the frontal plane of the pelvis to define the neutral alignment of the femur. The iliac wing of the hemipelvis was potted in 4-in (10.2-cm)-diameter polyvinylchloride pipe with use of an alignment fixture to set neutral adduction, flexion, and internal rotation. The femur was transected at the junction of the middle and distal thirds, and a threaded rod was inserted into the medullary canal. Each specimen was placed in a loading apparatus that allowed adjustment of flexion, extension, and axial rotation of the femur without constraining translation of the pelvis in the anteroposterior or medial-lateral directions.

During subsequent loading maneuvers, strains that developed in the labrum were measured with use of roentgen stereophotogrammetric analysis. Each specimen was prepared for testing with the creation of a 3-mm incision in the joint capsule. Two rows of chromium-steel spheres (diameter, 0.79 mm) were inserted into the substance of the labrum, parallel to the chondral margin. A third row of markers was fixed to the free edge of the labrum. A clock-face representation of the acetabular rim was used to describe the location of the markers according to the method reported by Beck et al.\(^2\) and Leunig et al.\(^4\). With the middle of the acetabular notch representing the six o’clock position, the markers were inserted at the two, two-thirty, three-thirty, and four o’clock segments of the labrum and were secured with cyanoacrylate adhesive. An additional row of fiduciary markers was also inserted into the osseous pelvis and was secured with cyanoacrylate adhesive.

The loading apparatus holding each specimen was mounted in a radiolucent loading frame and was attached to a vertical pneumatic actuator. A free-weight pulley was incorporated into the design of the loading frame to provide rotational torque during testing. Two radiographic film cassettes were aligned perpendicularly in the testing frame, allowing orthogonal radiographs to be made during the loading maneuvers. After radiographs were made with the hip in the unloaded, neutral position (neutral flexion, rotation, and adduction) (Maneuver A), five additional loading maneuvers were immediately applied to each specimen: (1) Maneuver B (with the hip oriented in 10° of extension and loaded with an external rotation torque of 177 in-lb [20 Nm] and an axial load of 100 lb [445 N]), (2) Maneuver C (with the hip oriented in 30° of flexion and loaded with an external rotation torque of 177 in-lb [20 Nm] and an axial load of 100 lb [445 N]), (3) Maneuver D (with the hip oriented in neutral flexion and neutral rotation and loaded with an abduction torque of 180 in-lb [203 Nm] and an axial load of 20 lb [89 N]), (4) Maneuver E (with the hip oriented in neutral flexion and 20° of external rotation and loaded with an abduction torque of 180 in-lb [20.3 Nm] and an axial load of 20 lb [89 N]), and (5) Maneuver F (with the hip oriented in 10° of extension and 20° of external rotation and loaded with an abduction torque of 180 in-lb [20.3 Nm] and an axial load of 20 lb [89 N]).

During Maneuvers B and C, the external rotation torque was applied with use of the free-weight pulley, whereas during maneuvers D, E, and F, the abduction torque was generated by applying a 20-lb (89-N) force through the pneumatic actuator.
at approximately the middle of the shaft of the femur. Radiographs were made immediately after each loading maneuver, with the load being removed from the specimen after exposure of the radiographs. These particular loading maneuvers were derived from previous investigations by McCarthy et al., in which the authors speculated that torsional maneuvers of the normal hip would lead to labral injury.

Biplanar radiographs of each specimen were analyzed to calculate the spatial location of each radiographic marker with use of roentgen stereophotogrammetric analysis. Each radiograph was digitized with use of a film scanner (Vidar Systems, Herndon, Virginia), and the coordinates of each metallic marker were calculated with respect to the center of the x-ray beam with use of Adobe Photoshop 7.0 (Adobe Systems, San Jose, California) (Fig. 3). Custom computer software corrected the measurement error due to magnification by incorporating the results of a calibration block with markers of known distances that was employed prior to testing. The labrum was assumed to have deformed homogeneously during testing of the hip joint, with the markers returning to their initial positions following release of the mechanical load. The distance between each pair of markers in the strained position (l) was compared with the

Fig. 2
Schematic of the loading apparatus used for testing (Maneuver B). ad = adduction, ab = abduction, and PVC = polyvinylchloride.
corresponding distance in the unloaded, neutral position ($l_0$). Maximum and average linear tensile strains between discrete markers were calculated with use of the formula $(l - l_0)/l_0$. Displacement of markers along the width of the labrum was used to calculate axial strain, and displacement of markers along the circumference of the labrum was used to calculate circumferential strain. A separate study was conducted to evaluate the reproducibility of the strain measurements (see Appendix).

In addition to the linear strain measurements described above, there were maximum strain values at each marker that were not necessarily in the radial or circumferential directions of the labrum. A two-dimensional assumption was made to simplify the displacements into planar directions. Simple plane strain equations were then used to determine the principal directions and the maximum and minimum principal strains (see Appendix). Additionally, for visualization of the area surrounding and between markers, a two-dimensional strain color map was created with use of commercial computer-aided design and finite element software (Unigraphics NX 4.0 and Nastran 4.0; EDS, Plano, Texas). The material properties of the labrum used in the finite element analysis (mean tensile modulus, $66.4 \pm 42.2$ MPa; maximum tensile strain, $45.5\%$) were derived from the previous work by Ishiko et al.\textsuperscript{11}.

The level of significance was set at $p < 0.05$. Outlying data points were identified with use of the extreme studentized deviate single-outlier procedure\textsuperscript{16} and were excluded from additional analysis. Descriptive statistics were calculated for average and maximum labral strain in the axial and circumferential directions. To assess differences in strain within specific regions of the labrum (divided by clock-face segments for axial strain measurements and separated by labral width for circumferential strain measurements), random effects repeated-measures analysis of variance was performed. Differences in axial and circumferential strain between the testing positions were also evaluated with use of random effects repeated-measures analysis of variance. If the results of either analysis of variance test revealed a significant difference in mean strain values, the Fisher least significant difference procedure\textsuperscript{16} was used to compare pairs of groups within the context of the analysis of variance test.

**Femoral Head Motion**

An additional investigation was undertaken to elucidate the motion of the femoral head during the loading maneuvers of the hip because of the potential for femoral head motion to contribute to labral strain\textsuperscript{1,8}. An additional cadaver specimen was prepared for testing, and an array of eight infrared-reflective markers was attached to the pelvis and the femur to enable their three-dimensional positions to be tracked with use of an optical motion-analysis system with a spatial resolution of $\pm 0.3$ mm (PCReflex; Qualysis, Gothenburg, Sweden). A three-dimensional computer model of the specimen was generated by reconstructing computed tomography scan data to allow recreation of the relative positions of the femur and the acetabulum under experimental loading conditions. The loading apparatus holding each specimen was mounted on an x-y displacement table in a biaxial mechanical testing machine (Bionix; MTS, Eden Prairie, Minnesota) and was attached to a vertical hydraulic actuator. The specimen was positioned with use of two of the maneuvers described earlier (Maneuvers B...
and E). Computer analysis of the motion capture data, as described by Alexander et al., was performed following the completion of specimen testing. The maximum external rotation torque (for Maneuver B) or abduction load (for Maneuver E) was identified in the intact state of the specimen, and their corresponding positions were recreated virtually with use of the reconstructed three-dimensional models and motion analysis data. The virtual position of the specimen at these points of maximum torque or load was then superimposed on the position of the specimen prior to the application of the torque or load. This allowed for the calculation of the displacement of the femoral head after the application of rotational torque or abduction forces. Displacement of the femoral head was calculated and was further described with use of the clock-face analogy described earlier.

**Results**

The average axial strain (and standard deviation) was 2.1% ± 1.1% during Maneuver B, 5.0% ± 1.7% during Maneuver C, 5.0% ± 2.5% during Maneuver D, 6.9% ± 1.1% during Maneuver E, and 2.9% ± 2.2% during Maneuver F (Table I). A significant difference between the average strain values during each of these maneuvers was found with use of the analysis of variance test, and the least significant difference procedure revealed that significantly lower strains were present during the external rotation maneuver at 10° of extension (Maneuver B).

![Table I](https://example.com/table1.png)

**Table I: Maximum and Average Axial and Circumferential Strain Values and Maximum Principal Strain in Each Loading Position**

<table>
<thead>
<tr>
<th>Strain</th>
<th>Maneuver B</th>
<th>Maneuver C</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Axial strain</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Average (n = 134)</td>
<td>2.1 ± 1.1</td>
<td>5.0 ± 1.7</td>
</tr>
<tr>
<td>Maximum (n = 20)</td>
<td>10.1 ± 2.6</td>
<td>13.6 ± 7.8</td>
</tr>
<tr>
<td><strong>Circumferential strain</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Average (n = 312)</td>
<td>2.9 ± 0.8</td>
<td>4.7 ± 1.4</td>
</tr>
<tr>
<td>Maximum (n = 45)</td>
<td>8.0 ± 1.2</td>
<td>8.2 ± 2.7</td>
</tr>
<tr>
<td>Maximum principal strain</td>
<td>8.0 ± 5.0</td>
<td>11.1 ± 4.7</td>
</tr>
</tbody>
</table>

*100 lb = 445 N, 177 in-lb = 20 Nm, 180 in-lb = 20.3 Nm, and 20 lb = 89 N.*
and the abduction maneuver at 10° of extension and 20° of external rotation (Maneuver F) when compared with the abduction maneuver at neutral flexion and 20° of external rotation (Maneuver E) (p < 0.05). With the numbers available, there was no significant variation in the average axial strains as a function of circumferential location.

The maximum axial strain was 10.1% ± 2.6% during Maneuver B, 13.6% ± 7.8% during Maneuver C, 13.2% ± 4.6% during Maneuver D, 13.3% ± 4.6% during Maneuver E, and 11.1% ± 4.5% during Maneuver F (Table I and Fig. 4). With the numbers tested, there was no significant difference between the maximum strain values measured during each of these maneuvers. Regional analysis of the data collected during all testing positions revealed that axial strains were largest (maximum value, 17.1% ± 5.1%) in the markers in the superiormost (two o’clock) position as compared with the markers in the two-thirty position (11.1% ± 4.6%), the three-thirty position (9.1% ± 3.1%), and the inferiormost (four o’clock) position (11.9% ± 3.0%). The difference in maximum axial strain values between the regions at the two o’clock and three-thirty positions was significant (p < 0.05).

The average circumferential strain was 2.9% ± 0.8% during Maneuver B, 4.7% ± 1.4% during Maneuver C, 3.1% ± 1.2% during Maneuver D, 3.4% ± 2.0% during Maneuver E, and −0.3% ± 1.6% during Maneuver F (Table I). With the numbers available, there were no significant differences among the mean circumferential strain values during each of these maneuvers.

The maximum circumferential strain was 8.0% ± 1.2% during Maneuver B, 8.2% ± 2.7% during Maneuver C, 8.2% ± 3.0% during Maneuver D, 8.4% ± 3.0% during Maneuver E, and
4.3% ± 3.9% during Maneuver F (Table I and Fig. 5). Analysis of variance testing also revealed a significant difference in maximum strain values between the testing positions, with the least significant difference procedure confirming significantly lower strains during Maneuver F when compared with maximum circumferential strains in the other testing positions (p < 0.05).

Six axial measurements (representing 4% of the 140 axial strain data points) and three circumferential measurements (representing 1% of the 315 circumferential strain data points) were found to be outliers with use of the extreme studentized deviate single-outlier method and were excluded from subsequent analysis. Twenty-six (19.4%) of the remaining 134 axial strains and sixty-nine (22.1%) of the remaining 312 circumferential strains were negative. The negative strain values were included in the statistical analysis.

Analysis of the finite element strain map to visualize areas of increased strain showed the majority of peak strain at the bone-labrum interface (Fig. 6) centered around the two-thirty (anterosuperior) position on the clock face. Lacking reconstructed magnetic resonance images to determine the exact labral edge and bone-labrum interface, the finite element strain map was used only to estimate the approximate location of peak strains, not to give a precise estimate of the maximum and average tensile strain values.

The average maximum principal strain within the labrum during testing was 8.0% ± 5.0% during Maneuver B, 11.1% ± 4.7% during Maneuver C, 8.7% ± 3.0% during Maneuver D, 10.8% ± 4.8% during Maneuver E, and 8.1% ± 4.4% during Maneuver F. With the numbers available, differences between these principal strain values were not significant. Shear strains were calculated and were found to be 0.1% during Maneuver B, 0.6% during Maneuver C, 2.3% during Maneuver D, 2.5% during Maneuver E, and 2.4% during Maneuver F.

Evaluation of the motion of one femoral head during the provocative maneuvers revealed that the application of an external rotation torque caused the femoral head to displace 1.3 mm anteriorly. The application of an abduction torque with the hip oriented in 10° of extension (Maneuver B) caused the femoral head to displace 1.6 mm in an anterior direction.
mm in an anteroinferior direction (three to five o’clock segments). The application of an abduction load caused the femoral head to displace 1.6 mm in an anterior direction (three o’clock segment) (Fig. 7).

Discussion

Prior to conducting the current study, we hypothesized that, during torsional loading of the hip, substantial mechanical strains are generated within the labrum, particularly in areas that are prone to injury. Our findings confirmed this hypothesis as substantial tensile strains were observed throughout the anterior portion of the acetabular labrum during provocative loading maneuvers. Statistical analysis of the strain data revealed several observations that may elucidate the mechanism of labral disruption. Under the controlled loading conditions, maximum strains in the anterior part of the labrum averaged 13.6% ± 7.8% in the axial direction and 8.4% ± 3.0% in the circumferential direction. Substantial strains were developed in all of the maneuvers tested, although each maneuver involved loading of the joint in orientations within 30° of neutral, without femoroacetabular impingement. Moreover, the degree of flexion or extension in each of the testing maneuvers did not significantly affect the amount of labral strain generated; in addition, substantial labral strains were generated during maneuvers in which the hip was loaded in external rotation or abduction.

These findings call into question the theory of Ganz and colleagues that all labral lesions arise during repetitive activities performed with the hip flexed to 90° to 100°. Moreover, our results support the conclusion that labral injury may occur in the absence of femoroacetabular impingement in positions other than flexion and internal rotation. One such mechanism, initially proposed by McCarthy et al., is that during athletic maneuvers involving the hip, especially those involving twisting and pivoting, excessive strain develops within the labrum, leading to mechanical disruption and, more specifically, creates a “watershed lesion” at the junction between the anterior part of the labrum and the adjacent articular cartilage.

To investigate the mechanism for the development of labral strains with external rotation and abduction moments, we tracked the displacement of one femoral head during loading. The application of an external rotation torque with the hip in 10° of extension caused displacement of the femoral head toward the anteroinferior quadrant of the acetabulum. Similarly, anterior subluxation was observed when an abduction torque was applied to the femur in an externally rotated position. It is entirely plausible that these displacements would generate stretching of the anterior part of the labrum, ultimately leading to labral lesions, particularly during activities that load the hip at higher levels than the controlled conditions of our experimental maneuvers. Furthermore, analysis of the finite element strain map revealed that the highest strain values were concentrated at the bone-labrum interface. This finding, in view of abduction or external rotation maneuvers in which the femoral head may pull the labrum away from its osseous and cartilaginous underpinnings, supports the theory that watershed lesions may occur in response to the repetitive loads applied during these maneuvers.

Arthroscopic and cadaver studies have confirmed that the majority of labral lesions occur in the anterior portion of the labrum. An effort was made in the current study to pinpoint areas of high labral strain to better understand the preponderance of both labral and chondral injury within this area. The highest values of mean and maximum strain were found at the two o’clock position and were significantly greater than those at the three-thirty position. This finding supports the conclusion that these lesions arise in response to a strain concentration that develops during physiologic loading and not through overload of intrinsically weak tissue within a focal area of the anterior part of the labrum itself. These findings also support McCarthy’s theory that athletic maneuvers performed in positions of extension or minimal hip flexion may be responsible for anterior labral injury and the creation of watershed lesions of the labrum and cartilage. It is important to note that our findings do not exclude the possibility that, in patients with abnormal morphologies of the acetabular rim or femoral head-neck junction, labral tears may still occur secondary to femoroacetabular impingement. However, the current study does suggest that anterior labral lesions may result from two parallel pathomechanical pathways: femoroacetabular impingement and recurrent athletic maneuvers.

The calculation of tensile strain in the current study was based on the displacement of labral markers before and after the application of dynamic loading to the hip joint. One of the major limitations of the present study is the planar assumption of labral strain. Like other studies of tensile strain in soft-tissue structures, our method of strain calculation relies on the assumption that the markers lie in a single plane and that the labrum deforms homogeneously without discontinuities. One additional assumption of our strain measurements is that each labral marker returned to its initial unloaded position following each testing position. This assumption was tested by performing a repeatability study, which showed that the labrum returned to its initial unloaded position following the release of the mechanical load (see Appendix). In addition, this repeatability study also confirmed the precision and reproducibility of the labral strain measurements themselves.

In the current study, we attempted to describe the regional distribution of labral strain. In doing so, we made the assumption that the markers were distributed in a similar manner in each of the specimens. Through a standardized and limited dissection, the markers were inserted in the same general area of the labrum in each specimen and, although the exact location of the markers in each specimen cannot be assured, it is the relative location of the markers (further superior or inferior) that is most important for understanding the distribution of labral strains in the current study. An additional limitation associated with our strain measurement involves the use of cyanoacrylate adhesive to fix the metallic markers within the labrum. This technique may alter the material properties of the area immediately surrounding each marker. The effects
of the adhesive are likely minimal, however, as the area to which the adhesive was applied was relatively small as compared with the remainder of the labrum.

Although the strains generated by the maneuvers studied in our experiments were considerable, they did not generate fraying or tearing of the labrum. Previous reports of the material properties of the labrum have been based on explanted specimens dissected free of the acetabular margin and loaded uniaxially. Two such studies demonstrated that the average maximum strain at failure was 26.5% in bovine tissue\(^1\) and 48.5% in humans\(^2\). Although the highest maximum strain value in the present study was lower than the ultimate strains reported in previous studies, this finding indicates that the maneuvers that were simulated in our testing protocol do not strain the labrum near failure. Also, strains that develop in the labrum in situ are likely attenuated by the dynamic stabilization provided by the intact hip capsule.

The limitations of the current study also include those inherent in the mechanical study of cadaver specimens. The age of the donors of the specimens that were used in the present study was generally greater than the age at which labral lesions initially become symptomatic. As mentioned earlier, the specimens were prepared with dissection of the soft tissues external to the hip capsule. The integrity of the remaining joint capsule may affect the amount of stabilization provided. In addition, the possibility exists that the removal of other soft-tissue structures surrounding the capsule, which may have a role in stabilizing the hip joint, altered our ability to accurately simulate the in vivo kinematics of the articulation.

The current study presents an evaluation of tensile strain within the acetabular labrum during dynamic maneuvering of the intact human hip joint. The positions chosen for testing generate considerable strains within the anterior part of the labrum and are thought to represent positions encountered during daily life and athletic activities, thus allowing for a better understanding of the normal physiology of the labrum. The presence of substantial anterior strains during these loading maneuvers indicates that activities that involve abduction and external rotation in slight flexion or hyperextension may predispose the joint to the development of labral injury. These findings provide a foundation for additional investigation of the pathomechanical process responsible for the development of labral injury and subsequent degenerative disease of the hip.

### Appendix

The description of Strain Measurement Reproducibility and the Principal Strain Equation for the Case of Plane Strain are available with the electronic versions of this article, on our web site at jbs.org (go to the article citation and click on “Supplementary Material”) and on our quarterly CD-ROM (call our subscription department, at 781-449-9780, to order the CD-ROM).