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Early Deformation of Hip Articular Cartilage Under A High Load Before and After Labral Excision

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Abstract

Background: It has been reported that the function of the labrum is to ensure that an increased surface area encountered the femoral head and spares the cartilage from excessive strain. The purpose of this study was to determine whether early deformation of the hip articular cartilage occurs under high-load conditions in the presence/absence of the labrum.

Methods: The hip joints of 4 beagle dogs were retrieved. A continuous static load of 80 Kg was applied for 2.5 hours. Magnetic resonance imaging (MRI) was performed immediately after the loading and at 30, 60, 90, 120, and 150 minutes after the loading. The load was then removed, and the specimens were stored for 14 h to allow the cartilage to recover. The labrum was then carefully removed, and the experiment was performed again.

Results: The maximum percentage change in cartilage thickness was 35.3 ± 17.4% when the labrum was intact and 55.7 ± 7.5% after the labrum had been excised (p=0.060). Labral excision resulted in a reduction in cartilage thickness. Cartilage thickness was significantly decreased in the normal and labral excision models immediately after loading (p=0.003 and p=0.022, respectively).

Conclusion: The labrum plays a role in dispersing loads equally across the joint cartilage and reduces the load placed on the maximum weight-bearing region of cartilage. Early articular cartilage deformation occurred under high-load conditions both before and after the excision of the labrum.

Keywords: Hip Joint; Labrum; Cartilage; MRI

Introduction

Deformation of the joint cartilage by loading is a concern for the maintenance of the function of joints. Reduction of strain in joint cartilage is important to keep normal function of joints.

The labrum increases the effective depth of the hip joint's socket and the coverage of the femoral head. The induction of negative pressure in the hip joint and mechanical interaction between the labrum and the base of the femoral head have been found to decrease cartilage deformation during loading [1,2]. Previous experimental studies have measured the tibial articular cartilage deformation induced by static loading under steady state conditions before and after medial meniscectomy and demonstrated that medial meniscectomy increased and decreased central and peripheral medial tibial cartilage deformation, respectively [3]. It has also been reported that labral repair resulted in a significant reduction in mean cartilage strain compared with that seen after labral tears and that labral resection resulted in higher cartilage strain than labral repair [4].

Previous theoretical analyses have shown that during joint loading there is little time for fluid to escape due to the low permeability of cartilage and so the volume of cartilage does not change much [5,6]. Herberhold et al. reported that under in situ conditions the cartilage deformation produced during physiologically relevant periods of loading for several seconds or minutes, only represents a small fraction of that seen in the final steady state [7].

We developed a device that can accurately and precisely measure cartilage deformation in the beagle hip joint under axial compressive loading. The purpose of this study was to compare the femoral and acetabular articular cartilage deformation induced under high physiological loading conditions simulating stumbling during walking before and after labral resection. In hip joints that have been subjected to labral resection, static loading can reduce the contact area and increase the maximum extent of cartilage deformation, which results in changes in the pattern of articular cartilage deformation. We hypothesized that (1) in hip joints that had been subjected to labral resection static loading would increase the maximum extent of cartilage deformation, (2) and early deformation of the hip articular cartilage would occur under high-load conditions simulating stumbling [8] both before and after labral excision.

Materials and Methods

Loading apparatus

We have developed a loading system that can be used to place loads on the hip joints of beagles. The device is designed to be used with magnetic resonance imaging (MRI). The loading system consists of a combination of the loading equipment, which is used in the MRI room, and an air flow control device, which is placed outside of the MRI room. The loading equipment consists of box (width: 138 mm, depth: 90 mm, height: 91 mm), in which the beagle hip joint is placed, and a piston (diameter: 63 mm). Compressed air is delivered from outside the room in order to drive the piston (stroke length: 46 mm) and place a load on the hip joint. The hip joint is fixed in place in the box using bone cement, and the inside of the box is filled with joint fluid. The pelvic side of the hip joint is fixed to the tip of the piston,

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and the femoral side is attached to the bottom of the box in a neutral position. The loading equipment was composed of polycarbonate (polyvinyl chloride/acrylics), and silicon grease was used together with nitrile rubber pneumatic seals to produce the piston ring. As for the air flow control device, a compressor, an air filter, and a regulator (Hitachi Construction Machinery, Tokyo, Japan) were connected in series, and the abovementioned loading equipment was connected to the regulator via a nylon tube (internal diameter: 6 mm). A one-touch joint was used to connect the equipment to enable its smooth assembly/disassembly. The maximum generating pressure of the compressor was 4.2 MPa. The connection tube and connector had allowable pressure values of ≥ 1 MPa.

Preparation of the animals

All experiments were conducted in accordance with the relevant animal welfare regulations and guidelines. Hip joints were retrieved from four beagles (approximate body weight: 20 kg), and the soft tissue outside the capsule was removed. The specimens were then frozen and stored until the experiment.

Evaluation of cartilage thickness using MRI

A 1.5 T MRI scanner (EXCELART Vantage XGV 1.5 T; Toshiba Medical, Tokyo, Japan) and a dual coil (φ 70 mm flexible coil, φ 150/70 mm flexible coil) were used to evaluate cartilage thickness. Fat-suppressed proton-weighted images (repetition time (TR)/echo time (TE): 27/15, number of excitations (NEX): 3.0, phase-encoding (PE) matrix: 200, rotation (RO) matrix: 160, PE field of view (FOV): 6.3 mm, RO FOV: 5.0 mm, resolution: 0.39 mm × 0.18 mm, slice thickness: 1.0 mm × 25) were acquired. Subtraction images were created using fat-suppressed T2*-weighted images (TR/TE: 27/15, NEX: 3.0, PE matrix: 200, RO matrix: 160, PE FOV: 6.3 mm, RO FOV: 5.0 mm, resolution: 0.31 mm × 0.31 mm, slice thickness: 1.0 mm × 25).

Each hip joint was placed in a neutral position, and coronal images were used for the evaluations. Specifically, the cross-sectional view that passed through the center of the femoral head was extracted. The thickness of the articular cartilage in the maximum weight-bearing region, which included both the femoral head and acetabulum, was measured on the digital images.

We mapped cartilage thickness and strain in the superior weight-bearing region of the hip. The acetabular and femoral cartilage was always in contact, and so it was difficult to distinguish between them. Therefore, the acetabular and femoral cartilage was manually segmented as a single unit. The contours of the cartilage tissue were then exported as 3D data points. Every coronal image of the hip joint image was sampled (total number of images: 45 to 50), and it took 8 to 10 minutes to acquire each series of images. Reconstructed images that passed through the center of the femoral head were used to obtain the cartilage thickness measurements. The edge of the osseous contour of the acetabulum in the middle MRI slice was used as a reference point. On each image, a grid of 15-20 radial test lines in the superior weight-bearing region was selected. Then, the orthogonal distance through the cartilage was manually measured. The measurements were summed, and the mean distance/thickness of the acetabular and femoral cartilage was calculated. The measurements were obtained three times, and the mean cartilage thickness was determined.

Calibration of pneumatic compressive load

The pressure applied to the sample was carefully adjusted via a calibration process to achieve a load of 80 kg (785 N), which was considered to simulate the weight placed on the hip joint by stumbling during walking. Air pressure of 0.29 MPa was required to accomplish this.

To evaluate the accuracy and repeatability of the loads produced by the device, pressurization up to 1 MPa and depressurization to 0 MPa was performed 5 times, and the applied load was measured every 0.1 MPa. This experiment was performed three times.

Application of a compressive load to the normal and labral excision models and MRI

Four unilateral beagle hip joints were cryopreserved for 12 h. Each specimen was fixed in place in the loading device and filled with bovine serum (Figure 1). The first MRI scan was performed before the loading. Then, a continuous static load of 80 kg simulating stumbling during walking was applied along the load axis for 2.5 h [8]. Further MRI scans were acquired immediately after the loading and at 30, 60, 90, 120, and 150 minutes after the loading. The compressive load was then removed, and the specimen was taken out of the loading device and stored in saline solution in a refrigerator for 14 h to allow the cartilage to recover. Next, the joint sample was warmed to room temperature and scanned once again to determine the extent to which the thickness of the cartilage had recovered. The labrum was then carefully removed with a scalpel, and the hip joint without the labrum was placed back into the loading chamber. The abovementioned loading process was then repeated. MRI scans were performed before the loading; immediately after the loading; and at 30, 60, 90, 120, and 150 minutes after the loading.

An orthopedic instructor who specialized in hip surgery and imaging analyses of the hip joint obtained all of the radiographic measurements.

Statistical evaluation

Intra observer reliability studies of the cartilage thickness measurements were performed using the intraclass correlation

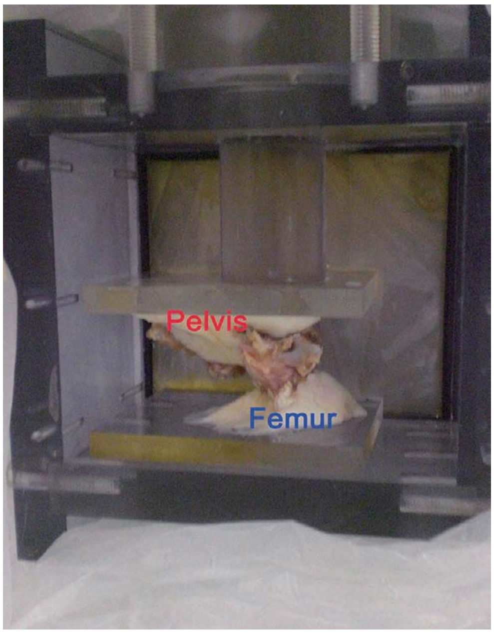

Figure 1: The loading equipment in which the beagle hip joints were fixed in place using bone cement.

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coefficient. The changes in cartilage thickness that occurred in each specimen were evaluated using the paired Student's t test. Statistical analyses were performed using SPSS software, version 19.0 (SPSS Inc., Chicago, Illinois, USA).

Results

Accuracy and repeatability of the loads produced by the device

The intraclass correlation coefficients obtained in the three pressurization and depressurization experiments were 0.998, 0.999, and 1.000, respectively. Thus, it was confirmed that it possible to apply a predefined load in an accurate and repeatable manner using the device (Figure 2).

Cartilage thickness

Three sets of cartilage thickness measurements were obtained in order to determine the intra observer reliability of the cartilage thickness data. The resultant mean intra class correlation coefficient was 0.935 (range, 0.899-0.971).

The mean initial thickness of the acetabular/femoral cartilage was 1.03 mm (range, 0.90–1.20 mm). The extent to which cartilage thickness had recovered by 14 h after the loading procedure was also measured. Accordingly, it was demonstrated that at 14 h after the loading procedure the mean percentage thickness of the articular cartilage in the superior weight-bearing region of each sample was 92%, 99%, 100%, and 100% of its original unloaded value, respectively (mean: 99.7 ± 3.9% recovery).

We evaluated the time-dependent changes in cartilage thickness in the superior weight-bearing region using coronal images passing through the center of the femoral head. The time-dependent changes observed in the normal and labral excision models are shown in Figure 3. The maximum percentage change in cartilage thickness was 35.3 ± 17.4% when the labrum was intact and 55.7 ± 7.5% after labral excision. As a result, it can be stated that labral excision caused a 20% increase in the maximum percentage change in cartilage thickness; however, this difference was not significant (p=0.060).

Subtraction coronal MRI passing through the center of the femoral head of specimen 4 are shown in Figure 4. The white portion indicates the change in the thickness of the articular cartilage observed after loading. A thicker white portion indicates a greater change in the thickness of the articular cartilage. Cartilage thickness was decreased by the excision of the labrum. The extent of the reduction was significant in specimens 1, 3, and 4 (p<0.001, p=0.002, p=0.001, respectively). Cartilage thickness was also reduced in specimen 2, but the change was not significant (p=0.200).

While the reduction in cartilage thickness seen immediately after loading in the normal model was 18.8 ± 8.5%, it was 37.5 ± 8.7% in the labral excision model (p=0.772). Cartilage thickness decreased significantly immediately after loading in both the normal and labral excision models (p=0.003, p=0.022). The reduction in cartilage thickness observed immediately after loading was 57.5 ± 25.3% and 68.8 ± 17.8% of the final recorded reduction value in the normal and labral excision model, respectively (p=0.493).

Discussion

Deformation of the joint cartilage by loading has been a concern for the maintenance of the function of joints. Previous studies have examined cartilage strain by measuring cartilage thickness in the unloaded state and after loading on MRI. For example, *ex vivo* studies have investigated cartilage strain in patellofemoral joints from 6 cadavers [7], sheep knees before and after meniscectomy [3,9] and hip joints from 6 cadavers that had been subjected to simulated labral tearing and repair [2]. These studies used 1.5 T, 4.7 T, and 7 T MRI scanners, respectively. Cartilage strain has also been studied *in vivo*. Interestingly, it was reported that among studies in which a load was applied to the knee or hip for 2 to 4 h the mean magnitude of the recorded cartilage strain was smaller in the *in vivo* studies [10-12] than in the *ex vivo* studies [2,3,7]. For example, in a study of 6 normal specimens, Herberhold et al. reported that the change in the maximal thickness of the knee joint cartilage was 2.8%, which was much smaller than that reported in *ex vivo* studies, probably due to the much shorter period of loading employed in the latter study [7]. Our results support these findings.

We speculate that the labrum plays a role in dispersing loads equally across joint cartilage and reducing the load placed on the cartilage in the maximum weight-bearing region. In experiments using beagle hip joints and MRI, we found that the change in cartilage thickness induced during loading was increased by labral resection. In a study involving 6 cadavers, Ferguson et al. reported that full labral resection increased the extent of the final displacement of the cartilage by 21% compared with that seen in intact joints [1]. On the other hand, Greaves et al. found that labral repair caused a 2% decrease in mean cartilage strain compared with that seen in a torn labrum (p=0.014), and that labral resection produced 4% and 6% increases in mean and maximum cartilage strain, respectively, compared with labral repair (p=0.02) [4]. The significant

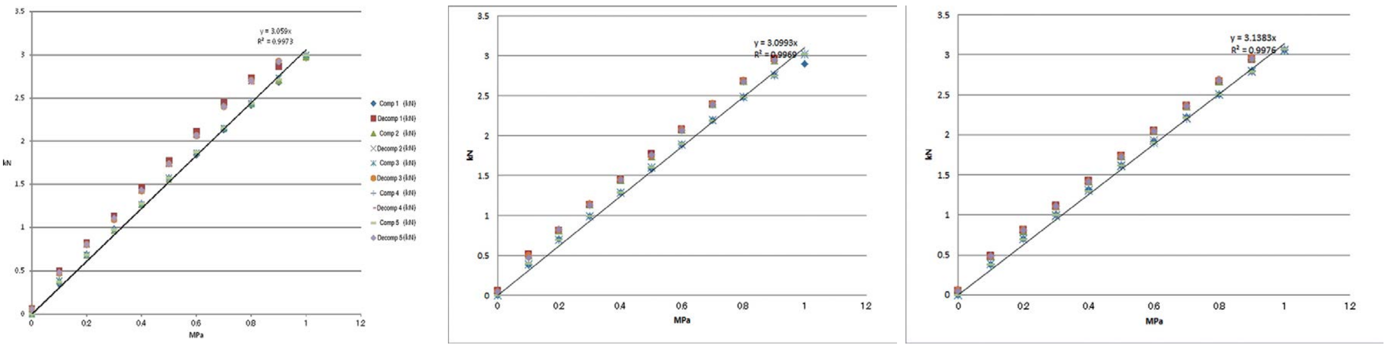

Figure 2: A hip joint was subjected to repeated compression and decompression and the correlation between the pressure induced and the applied load was evaluated 3 times (2a-2c).

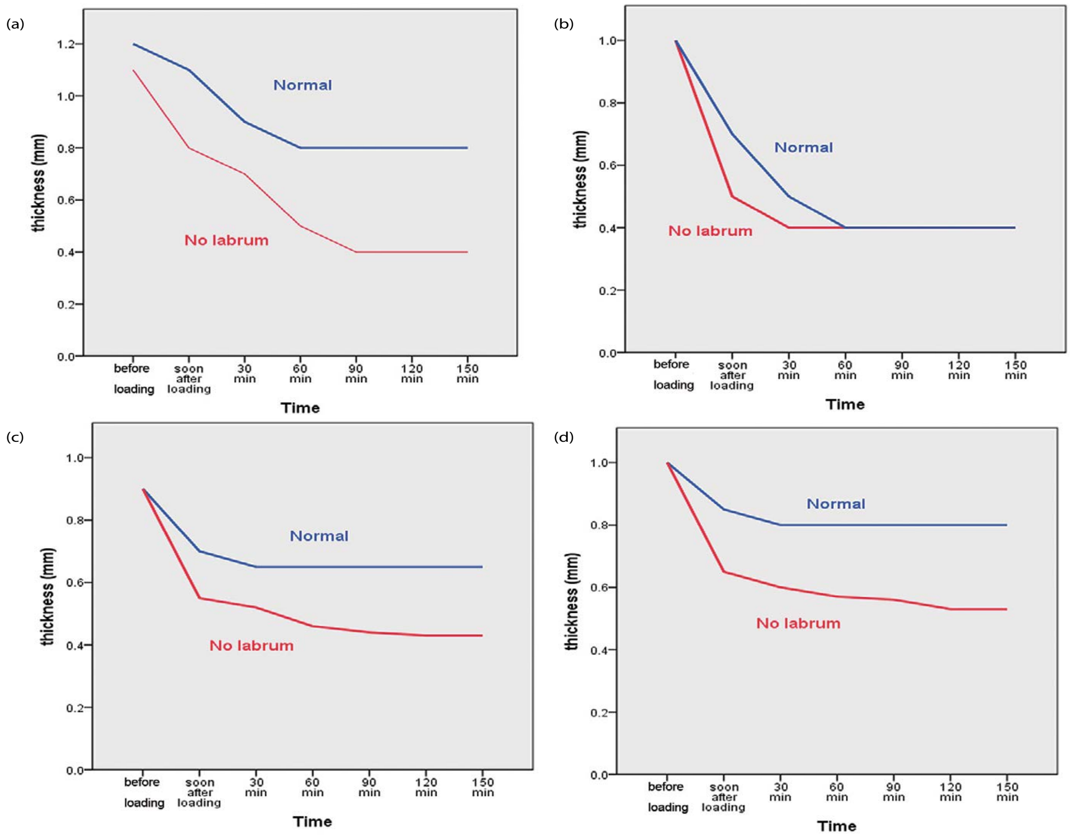

Figure 3: Time-dependent changes in cartilage thickness in all 4 beagles. (a) specimen 1 Cartilage thickness changes of 0.8 mm and 0.4 mm were seen in the labral excision and normal models, respectively. (b) specimen 2 (c) specimen 3 (d) specimen 4.

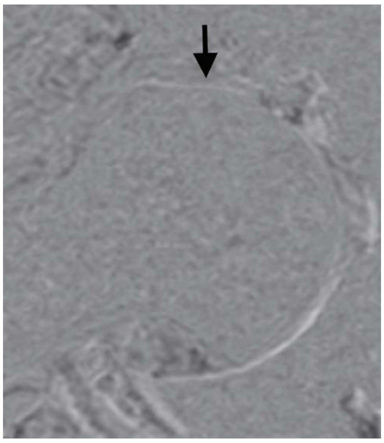

Figure 4A: Subtraction image of the normal model of specimen 4.The change in articular cartilage thickness seen at 150 minutes after loading is shown in white. A thicker white portion indicates a greater change in the thickness of the articular cartilage. The thickness of the cartilage in the superior weight-bearing region changed by 0.20 mm.

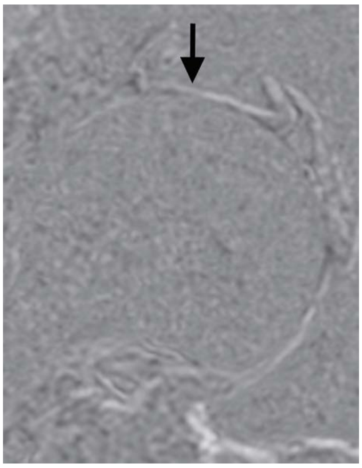

Figure 4B: Subtraction image of the labral excision model of specimen 4. The change in articular cartilage thickness seen at 150 minutes after loading is shown in white. The thickness of the cartilage in the superior weight-bearing region changed.

reduction in mean cartilage strain seen after labral repair compared with that observed after labral tearing might have resulted from the production of a circumferentially stiffer labrum, which might distribute loads more effectively. Higher cartilage strain values are observed when the labrum is resected because the torn portion of the labrum is no longer present and so is unable to distribute loads evenly across the joint. These findings demonstrate that the labrum increases the surface area that is in contact with the femoral head, sparing cartilage from excessive strain and are supported by our results.

Resolution of MR images might be important for an analysis of the thickness of joint cartilage. Very high resolution images seem to be preferable for accurately measuring changes in cartilage thickness. Although the resolution of the 1.5 T MRI scanner used in our study might not have been high enough, the changes in articular cartilage thickness caused by labral excision were clearly depicted on subtraction images obtained using a suitable dual coil (Figure 4).

Attainment of an equilibrium state of joint cartilage in physiological loading has been studied to analyze function of the cartilage. It was reported that articular cartilage requires hours of loading to reach an equilibrium state [4]. In an *ex vivo* imaging study of the patellofemoral joint, Herberhold et al. studied the deformational behavior of the articular cartilage during the static loading of 150% of the subject's body weight [7]. The changes in the thickness of the patellar/femoral cartilage seen after 8 minutes of compression were only 25%-30% of the equivalent values recorded after 214 minutes of compression. In addition, mean and maximal percentage in situ cartilage thickness changes of 44% and 57% were recorded in the patellar cartilage after 3.5 h of static loading with a force equivalent to 150% body weight (mean contact pressure=3.6 MPa). Of the total cartilage thickness change recorded at the end of the study, only ~7% (3% absolute deformation) occurred during the first minute of loading, and only 25% (11% absolute deformation) occurred during the first 8 minutes of loading. These findings show that under in situ conditions only a small fraction of the final change in cartilage thickness occurs during physiologically relevant loading periods; i.e., several seconds or minutes. These results are consistent with previous theoretical analyses which suggested that there is little time for fluid to escape during loading due to the low permeability of cartilage, and therefore, cartilage volume does not change much [5,6]. Hydrostatic pressurization of the interstitial fluid, which is considered to protect the solid matrix from elastic deformation for several hundred seconds after the onset of loading is assumed to be a crucial factor in protecting the matrix from undue stress and tissue damage during normal loading [5,13]. In our experiments, however, the percentage changes in cartilage thickness seen immediately after loading were 57.5 ± 25.3% and 68.8 ± 17.8% of the final values recorded in the normal and labral excision models, which were much greater than those recorded in previous studies. The applied load of 80 kg was based on the highest hip joint force generated during walking. In the present study, we found that greater cartilage deformation occurred immediately after loading that was observed in previous studies, suggesting that cartilage deforms much faster when it is subjected to continuous high loading. Articular cartilage that is subjected to loading in the physiological range does not deform much initially as the fluid in the cartilage bears the compressive load and is not pushed out of the articular cartilage matrix due to the low permeability of the articular cartilage matrix. However, when the cartilage is loaded beyond its failure load, it might result in the bursting of the cartilage matrix, allowing rapid fluid extravasation. Thus, continuous loads that simulate stumbling during walking might be high and result in articular cartilage matrix damage.

The other limitations of the present study include the limited number of animals used and the differences between the hip joint characteristics of beagles and humans. The resolution of the MRI (1.5 Tesla) used to measure cartilage thickness was relatively low. These limitations should be addressed in future studies.

Conclusion

The labrum plays a role in equally dispersing loads across joint cartilage and reducing the load applied to the cartilage in the maximum weight-bearing region. Early deformation of the articular cartilage occurs under high-load conditions simulating stumbling during walking in both before and after labral excision.

Conflict of Interest Statement

The authors declare that they have no conflict of interest.

Authors' Contributions

HI organized the entire study, carried out the experiments and drafted the manuscript. TN participated in the design of the study and carried out the experiments. TS carried out the experiments, analyzed the data and performed the statistical analysis. YN carried out the experiments and analyzed the data. MH designed and made the loading apparatus. HT participated in the design of the study and carried out the experiments and analyzed the data. All authors read and approved the final manuscript.

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