Articular cartilage MR imaging and thickness mapping of a loaded knee joint before and after meniscectomy

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Summary

Objective: We describe a technique to axially compress a sheep knee joint in an MRI scanner and measure articular cartilage deformation. As an initial application, tibial articular cartilage deformation patterns after 2 h of static loading before and after medial meniscectomy are compared.

Methods: Precision was established for repeated scans and repeated segmentations. Accuracy was established by comparing to micro-CT measurements. Four sheep knees were then imaged unloaded, and while statically loaded for 2 h at 1.5 times body weight before and after medial meniscectomy. Images were obtained using a 3D gradient echo sequence in a 4.7 T MRI. Corresponding 3D cartilage thickness models were created. Nominal strain patterns for the intact and meniscectomized conditions were compared.

Results: Coefficients of variation were all 2% or less. Root mean squared errors of MR cartilage thickness measurements averaged less than 0.09 mm. Meniscectomy resulted in a 60% decrease in the contact area (P = 0.001) and a 13% increase in maximum cartilage deformation (P = 0.01). Following meniscectomy, there were greater areas of articular cartilage experiencing abnormally high and low nominal strains. Areas of moderate nominal strain were reduced.

Conclusions: Medial meniscectomy resulted in increased medial tibial cartilage nominal strains centrally and decreased strains peripherally. Areas of abnormally high nominal strain following meniscectomy correlated with areas that are known to develop fibillation and softening 16 weeks after medial meniscectomy. Areas of abnormally low nominal strain correlated with areas of osteophyte formation. Studies of articular cartilage deformation may prove useful in elucidating the mechanical etiology of osteoarthritis.

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Key words: Articular cartilage, Meniscectomy, Cartilage thickness, Contact area, Osteoarthritis, MR imaging, Joint loading.

Introduction

Osteoarthritis (OA) is among the most common diseases in orthopedics and a leading cause of physical disability and health care expense throughout the world. Approximately 11% of people aged over 65 and 6% of entire U.S. adult population suffer from symptomatic OA1. Various factors, including mechanical insults such as meniscectomy, intra-tibial fracture, mechanical malalignment, and ligamentous instability, alter the mechanical environment on the articular surface and can initiate OA2.

Meniscectomy is a common and successful surgical procedure for treating pain and mechanical locking of the knee after a meniscus tear. However, several studies have shown that a large number of meniscectomized patients subsequently develop OA3,4. Post-meniscectomy OA of the knee is generally believed to be a consequence of biomechanical alterations at the knee due to the absence of a meniscus to distribute the joint reaction force, absorb impact force, and maintain joint stability.

Experimental and computational models have demonstrated increases in articular cartilage contact stress following meniscectomy6,6, and in animal models articular cartilage degradation and subchondral bone changes have been reported7–11. More recently, Appleyard et al. and Oakley et al. mapped region-specific cartilage morphology and material properties following meniscectomy in a sheep model of OA12,13. Sixteen weeks after medial meniscectomy, the medial tibial articular cartilage developed fibillation and softened centrally, while osteophytes developed peripherally13.

Various mechanical factors are believed to initiate and promote the progression of OA14. Among these factors, joint unloading has been postulated to result in decreased articular cartilage fluid pressure and a progression of the subchondral growth front15,16. Osteophytes are commonly seen in such areas18. Conversely, articular cartilage has been noted to be fibrillated in areas of pathologic overloading16.

Though contact stress has been previously studied in the laboratory, the region-specific deformation in the articular cartilage tissue itself has not been previously mapped in a knee before and after meniscectomy. To understand how known alterations of contact stress translate to known patterns of articular cartilage degeneration, it is necessary to characterize the deformation of the articular cartilage tissue. It is likely that the local deformation of the tissue is...
involved in transmitting the mechanical signal from the surface to the chondrocyte, the active element embedded in the tissue that is responsible for matrix synthesis and degradation. Further, it is likely that areas of increased cartilage deformation are more susceptible to mechanical damage and fibrillation.

Various methods have been described to determine articular cartilage thickness or deformation in an intact joint. Among those methods, computed tomography (CT) and scanning electron microscopy (SEM) have been shown to be capable of providing high resolution cross sectional images of cartilage\(^1\),\(^2\). CT scans, however, do not delineate the interface between soft tissues, such as cartilage—cartilage or cartilage—meniscus boundaries, well when entire intact joint is scanned and these surfaces are pressed together. SEM is incapable of revealing the thickness of the articular cartilage over its entire surface and through its depth, and necessitates destroying the specimen, making repeated measurements of the same specimen impossible.

MRI is a promising technique for articular cartilage imaging because it reveals the superficial surface and the rim to subchondral bone over the entire cartilage surface. It also has the advantage of being non-invasive, allowing maintenance of periarticular supporting structures, and allowing repeated measurements of the same joint under various loading conditions. Several studies have evaluated the deformation of articular cartilage in tissue plugs and exposed articular surfaces with mechanical loading\(^1\),\(^2\), and another has evaluated the in vivo and in vitro deformation of articular cartilage from the patella-femoral joint during mechanical loading\(^1\),\(^2\). At the tibio-femoral joint, MRI has been used in vivo to evaluate the deformation of articular cartilage after various activities such as running\(^1\),\(^2\). There have been, however, no MRI studies done on the patella-femoral joint to evaluate articular cartilage deformation while load is applied, and no MRI studies have evaluated articular cartilage deformation following OA-inducing changes to the joint such as meniscectomy. Information about articular cartilage deformation in an intact knee joint with and without a meniscus could shed light on the pathogenesis of OA after meniscectomy. The purpose of this study was to develop a device that can accurately and precisely measure cartilage deformation in a sheep knee joint under axial compressive loading at a physiologic magnitude. As an initial application of this device, we compared tibial articular cartilage deformation before and after medial meniscectomy under static loading and steady state conditions. We postulated that the pattern of tibial plateau articular cartilage deformation in a meniscectomized knee is different than in an intact knee. Specifically, we hypothesized that in a statically loaded knee joint following meniscectomy, the contact area will decrease, the maximum cartilage deformation will increase, and the pattern of articular cartilage deformation will be significantly changed. We further hypothesized that changes in articular cartilage deformation with meniscectomy will relate to articular cartilage degeneration patterns that are seen in vivo.

Methods

LOADING SYSTEM

The entire loading system consists of a cylindrically shaped MR compatible pneumatic loading apparatus and an electric air flow control device. The loading apparatus (Fig. 1) holds a joint specimen and applies uniaxial compressive load across the tibio-femoral joint inside a 4.7 T MRI scanner. A flow control device (Fig. 2) controls the air flow to generate both static and cyclic compressions. For imaging of cyclically loaded specimens, it also provides a gate signal to the MRI scanner synchronized with the loading cycle.

The loading apparatus is sized to test a sheep knee. It consists of an outer cylinder, a supporting shell, a loading shell, and an air bladder. The outer cylinder is a container holding the specimen and fluid, and is a frame for the loading mechanism. The outside diameter of the cylinder fits into the 6 cm radio-frequency coil of the 4.7 T MRI scanner. The cylinder is made of high strength cast acrylic to maintain its original shape under high pressure, to resist fatigue failure after numerous loading cycles, and to allow X-ray transmission should micro-CT applications be needed. The supporting shell is also made of high strength cast acrylic and is fixed on the bottom part of the outer cylinder. An acrylic pin on the supporting shell anchors the outer part of the tibia when the specimen is in place, preventing anterior/posterior rotation and medial/lateral movement of the tibial portion of the specimen. PVC tubing is used for the loading shell because of its flexibility. The loading shell is driven by an inflatable air bladder in the gap between the loading shell and the outer cylinder, and is guided by two guide pins on the top of the outer cylinder. The bladder has two air inlets to

![Diagram of loading apparatus](image)
achieve even inflation and deflation. The bladder is made of urethane coated nylon using radio-frequency welding process which can hold up to 1.4 MPa (200 psi).

In this study, only static compressive loading was used. A joint sample was placed into the loading chamber between the loading and supporting shells. The loading device was filled with a saline solution of 5 mmol/L gadolinium (Magnevist, Schering Ag, Berlin, Germany) and proteinase inhibitors to keep the sample hydrated and to limit enzymatic degradation of the tissue. When the specimen was to be loaded, the solenoid valve was opened by the electronics and compressed air inflated the bladder pushing the loading shell toward supporting shell and applying a uniaxial compressive force across the joint. The tibial part of the specimen was completely fixed. The femoral part of the joint was constrained in internal/external rotation and anterior/posterior translation, but was unconstrained in the other directions.

SAMPLE PREPARATION

Four fresh sheep hind limbs were collected from a local butcher shop. The sheep were 1 year old and weighed 50 kg. The leg was positioned at 45 degrees of knee flexion angle simulating the normal standing position of sheep and frozen until the experiment. While the tissue was frozen, the joint was cored from the medial to the lateral side with a 4.5 cm hole saw. The cored specimen was large enough to include the entire intact tibio-femoral joint, both collateral and cruciate ligaments, and both menisci. The joint sample was then frozen until testing.

CALIBRATION OF PNEUMATIC COMPRESSIVE LOAD

To calibrate the load applied by the pneumatic loading chamber, a phantom joint specimen was made of acrylic, and a commercial load cell was used to measure the force between the phantom joint surfaces. A 6 cm diameter acrylic rod was cut in half and was machined to accommodate a load cell between the top and bottom halves. The load cell was then placed between top and bottom halves of the phantom and the entire phantom was positioned in the pneumatic loading chamber. The loading device was then turned on, and the force across the phantom joint was recorded from the load cell. The applied pressure inside the air bladder was carefully adjusted by a calibration.
process to achieve a 1.5 times body weight force (650 N) across the tibio-femoral joint. Air pressure of 0.4 MPa (60 psi) was required to accomplish this.

IMAGING AND CARTILAGE CONTACT AREA AND THICKNESS MEASUREMENT

Imaging was performed on a Unity Inova console (Varian, Inc., Palo Alto, CA) controlling a 4.7 T, 15 cm horizontal bore magnet (Oxford Instruments, Ltd., Oxford, UK) with GE Techron Gradients (12 G/cm) and a volume coil with an inner diameter of 6 cm (Varian, Inc., Palo Alto, CA). Scout images oriented coronal to the joint were acquired to visualize the extent of interest. An entire 3D image of whole knee joint was acquired with a T1 weighted 3D gradient echo sequence (TR/TE: 22/6 ms; Matrix: 512 × 512 × 128, zero-filled to 1024 × 1024 × 128; FOV: 60 × 60 mm²; Slice thickness: 60 mm; NEX: 2) providing 58 × 58 μm² in-plane resolution and 470 μm slice thickness. A total of 128 sagittal images of the knee joint were acquired and the total acquisition time to image one sample was 100 min.

Articular cartilage boundaries and subchondral bone interfaces in the 128 sagittal slices obtained for each joint sample were carefully segmented using custom segmentation software. All segmented lines were combined and a solid 3D surface was generated using 3D modeling software, GeoMagic Studio (Raindrop Geomagic, Inc., Research Triangle Park, NC). Four 3D cartilage models (unloaded intact joint, statically loaded intact joint, unloaded recovered joint, statically loaded meniscectomized joint) for each joint specimen were created.

To calculate contact area, locations of tibial articular cartilage contact to either femoral articular cartilage or meniscus were manually marked on the segmentation data. The areas of contact were calculated using GeoMagic Studio. Cartilage thickness was calculated by determining the perpendicular distance from the subchondral bone surface to the articular cartilage surface using MATLAB (The MathWorks, Inc., Natick, MA) and custom code designed in our laboratory. Articular cartilage thickness data were collected for approximately 10,000 sample points on the joint surface for each loading condition [Fig. 4(a)].

To compare cartilage deformation before and after load application, each 3D loaded cartilage model was aligned with the unloaded intact 3D cartilage model. Subchondral bone surfaces were used as a reference to register the position of each model because the subchondral bone remained nearly undeformed throughout the tests. The alignment process was performed using the “Best Fit Alignment” function of GeoMagic Studio. The aligned pairs of loaded and unloaded cartilage models were prepared to calculate the amount of cartilage deformation. Surface normal lines at each sample point on the unloaded subchondral bone surface were created. The amount of cartilage deformation was calculated by subtracting the thickness data at the points of the deformed cartilage surface (thickness 1, Fig. 4(b)) from the thickness at the points of the undeformed cartilage surface (thickness 2, Fig. 4(b)). This process was also performed using custom MATLAB code. To visualize these data, nominal strain plots were made by calculating the percent thickness change at each point on the articular surface before and after meniscectomy.

To investigate the change in articular cartilage strain over the medial femoral condyle surface with meniscectomy, the proportions of the articular cartilage surface that had high deformation (>60% nominal strain), moderate deformation (20–60% nominal strain), and low deformation (<20% nominal strain) were calculated for each of the four specimens in the loaded intact and loaded meniscectomy cases.

PRECISION AND ACCURACY OF QUANTITATIVE CARTILAGE THICKNESS MEASUREMENT

To determine the precision of our overall cartilage thickness measurement procedure (inter-scan precision), we scanned the same unloaded intact knee joint on three consecutive days. After finishing each day’s scan, the sample was stored in saline-proteinase inhibitors solution in a refrigerator until the next scan. Three 3D cartilage models were created and aligned with each other. This procedure tested the potential variation in thickness measurement across different scans due to the scanning procedure itself, including the effects of positioning and orientation of the specimen, and the effects of segmentation and registration. To test the precision of the manual segmentation and 3D registration step itself (inter-segmentation precision), one of the three MRI data sets was selected, and this data set was segmented on three different days. 3D models from each segmentation result were aligned and the thickness differences were compared.

To test the accuracy of measurements, MRI thickness measurements were compared with micro-CT measurements of the same specimen. Accuracy validation of MRI cartilage thickness measurements using micro-CT as the standard has been done in numerous previous studies. Micro-CT scanning was performed on vivaCT 40 scanner (SCANCO Medical AG, Bassersdorf, Switzerland). On the micro-CTs, a total of 1568 images were acquired (Matrix: 1024 × 1024; FOV: 38.9 × 38.9 mm²; Slice thickness: 38 μm) providing 38 × 38 × 38 μm³ isotropic pixel size. Total acquisition time was 80 min.
For accuracy testing, we chose to scan only the proximal tibia because we found that the CT scanner could not delineate the cartilage–cartilage contact interface well, even if a contrast agent was used. A proximal tibia was isolated and stored in a solution of saline and proteinase inhibitors. The sample was then scanned in the MRI scanner. The same sample was then scanned in a micro-CT scanner. A total of three different proximal tibias were scanned and six corresponding 3D models, three for MRI scans and three for micro-CT scans, were created. For the three pairs of images, segmentation and 3D model construction were performed.

MR IMAGING OF STATISTICALLY LOADED INTACT AND MENISCECTOMIZED KNEE SPECIMENS

Four unmatched sheep hind limbs were obtained fresh from a local butcher. Joint samples were prepared as described above. Each of the four joint samples was scanned four times: (1) unloaded intact, (2) statically loaded intact, (3) unloaded recovery, and (4) statically loaded meniscectomized. To thaw the specimen, a cored frozen knee joint was bathed at room temperature in a saline solution of gadolinium and proteinase inhibitors for 12 h. The completely thawed intact joint sample was then placed inside the loading chamber with the tibio-femoral joint surface perpendicular to the loading axis [Fig. 1(a)]. The loading apparatus was then filled with a solution of saline, gadolinium and proteinase inhibitors to keep the sample hydrated, and the intact unloaded joint was scanned.

Following the unloaded scan, 650 N of static compressive load was applied across the joint, for 2 h. Based on previous work by other investigators, we felt that this was the sufficient time to achieve steady state cartilage deformation.\(^\text{25}\) The magnitude of the load, approximately 1.5 times body weight is close to the knee joint reaction force for normal walking in sheep.\(^\text{34}\) After 2 h of loading, the compressed joint sample was scanned again. The specimen was never moved, so that each slice obtained during the loaded scan exactly matched a slice in the unloaded scan.

The compressive load was then removed and the specimen was taken out of the loading device and stored in the solution of saline, gadolinium and proteinase inhibitors in a refrigerator for 14 h to allow cartilage thickness recovery. After 14 unloaded hours, the joint sample was warmed to room temperature and scanned once again to determine the amount of cartilage thickness recovery. The medial meniscus was then carefully removed with a scalpel keeping the collateral and cruciate ligaments intact. The meniscectomized joint was then placed back into the loading chamber with the saline–gadolinium–proteinase inhibitor solution. Accurate realignment of the joint after meniscectomy was achieved due to the constraints of the cylindrical loading chamber and the remaining intact ligaments.

All images were segmented and cartilage thickness measurements were made according to the procedure described in a previous section (Imaging and cartilage contact area and thickness measurement). Three deformed and one undeformed 3D cartilage thickness models were created and the deformed models were aligned with undeformed cartilage model. The aligned pairs of (1) statically loaded vs unloaded, (2) recovery vs unloaded, and (3) meniscectomized loaded vs unloaded, were prepared to calculate cartilage deformation. The thickness comparisons for all of the three pairs were done using the technique described in a previous section (Imaging and cartilage contact area and thickness measurement).

STATISTICAL EVALUATION

To determine the inter-scan and the inter-segmentation precision of MR thickness measurements, we calculated coefficient of variations (CV (%), (standard deviation/average) × 100) of mean and maximum thickness of the 3D models (three different scanning data and three different segmentation data). The accuracy of MR measurement was validated by the root mean square (RMS) and percentile ([CT – MRI]/CT) of differences between MRI and CT measurements over entire 3D cartilage surface.

Contact area, mean and maximum deformation, and proportions of articular cartilage high, moderate, and low deformation areas before and after meniscectomy were compared between the intact and meniscectomy cases using paired Student’s t tests.

Results

PRECISION AND ACCURACY OF ARTICULAR CARTILAGE THICKNESS MEASUREMENT

Inter-scan CVs were 1.88% and 1.96% and inter-segmentation CVs were 1.10% and 2.00% for mean and maximum thicknesses, respectively. The RMS (percentile) differences between MRI and micro-CT thickness measurements over the entire cartilage surface were 0.099 mm (8.4 ± 7.1%), 0.074 mm (6.6 ± 7.3%), and 0.081 mm (4.4 ± 9.2%) for the three different knee joints.

ARTICULAR CARTILAGE CONTACT AREA AND DEFORMATION WITH AND WITHOUT THE MEDIAL MENISCUS

Sample unloaded and loaded sagittal MR images through the center of the medial condyle of a knee are shown in Fig. 5. The contact area change on the tibial articular cartilage due to the medial meniscectomy is shown in Fig. 6. The medial contact area was reduced by approximately 60% after meniscectomy \((P = 0.001)\). In contrast, the lateral side where the meniscus was intact had no significant changes in contact area.

The 3D cartilage models and thickness maps of one of the four tested specimens are shown in Fig. 6. The mean (maximum) medial and lateral cartilage thicknesses of intact unloaded joint were 0.95 ± 0.05 mm \((1.92 ± 0.11 \text{ mm})\) and 0.86 ± 0.09 mm \((1.27 ± 0.52 \text{ mm})\), respectively. When the joint was loaded, medial meniscectomy resulted in a deeper and wider dimple in the medial tibial articular cartilage [Fig. 6(b and c)].

Percent cartilage thickness change (nominal strain) was then calculated. The nominal strains of one of the four tested knees with and without medial meniscus are displayed in Fig. 7. Across the four specimens tested, mean and maximum nominal strains in the medial tibial articular cartilage were \(36.1 ± 4.5\% \text{ and } 72.1 ± 7.1\%\), respectively, when the meniscus was intact, and \(31.0 ± 2.1\% \text{ and } 85.3 ± 4.0\%\) after medial meniscectomy. As a result, meniscectomy caused a statistically significant 13% increase in maximum deformation \((P = 0.01)\), but no significant change in mean nominal strain \((P = 0.08)\).

In this static, steady state loading condition, we defined nominal compressive strain less than 20% as being low deformation, 20–60% as being moderate deformation, and greater than 60% as being high deformation. Using
this definition, across all specimens, the proportions of tibial articular cartilage surfaces experiencing low, moderate, and high compressive nominal strains in the intact and meniscectomy cases were calculated (Fig. 8). After 2 h of static loading at 1.5 times body weight and the medial meniscus intact, a broad area (64%) of the articular cartilage surface was moderately deformed on the medial tibial plateau under static load. This area roughly corresponded to the area of meniscus contact. Thirty percent of the surface had low deformation and 6% of the surface had high deformation.

In contrast, when the medial meniscus was removed, the area of moderate deformation on the medial tibial plateau was reduced to 35% (P = 0.0007) of the articular cartilage surface, while the area of low deformation was increased to 49% (P = 0.009) of the surface, and the area of high deformation was increased to 16% of the surface (P = 0.028). Medial meniscectomy did not alter these proportions significantly on the lateral side of the joint.

Recovery of cartilage thickness after 14 h of unloading was recorded for all four specimens. We found that the mean medial tibial plateau articular cartilage thickness after 14 unloaded hours in each of the four samples was 85%, 98%, 87%, and 85% of the original unloaded thickness (average 88.8 ± 6.2% recovery).
Discussion

In this study we described a new method of axially loading a whole tibio-femoral joint in an MRI scanner and measuring the resulting 3D deformation of articular cartilage. This device allowed us to define the magnitudes and patterns of articular cartilage nominal strain under various loading conditions. We investigated and defined the precision and accuracy of this new technique, and as an initial application, we investigated the deformation of tibial plateau articular cartilage after 2 h of static compressive loading before and after medial meniscectomy.

The new loading device and measurement technique were found to have a high degree of accuracy and precision. With MRI scanning, the precision of the measurement depends on factors associated with image acquisition and image post-processing. Precision of the measurement was tested with (1) repeated scanning of one sample, and (2) repeated segmentation of a single scan. In both cases, the CVs of mean and maximum thicknesses were calculated and all CVs were 2% or less. Thus, the sum of each individual error (scanning, segmentation, and registration error) embedded in our measurements was less than 2%. The low inter-scan CV values are consistent with a previous report that susceptibility induced geometric distortion does not pose a significant problem in quantitative cartilage imaging, as long as high resolution T1 weighted gradient echo sequences are selected. Inter-segmentation CV showed that a person can consistently reproduce 3D cartilage models with our high resolution MR images.

The accuracy of our measurement was tested using micro-CT scans. Micro-CT was selected as a "gold standard" because (1) it is non-invasive, (2) the border between the calcified and uncalcified cartilage, and the articular cartilage surface is clearly delineated, (3) no significant geometric distortion is created with scanning, (4) the same post-processing method as in the MRI measurement can be used, and (5) the micro-CT in-plane image resolution is higher than other imaging modalities. The accuracy test revealed that percentile differences between MRI and CT measurements were 4–8% (RMS difference: 70–100 μm). This compares favorably with the 10–15% that has been reported in a similar study on human cadaver tibial articular cartilage. Moreover the RMS errors between MR and micro-CT data (70–100 μm) were very small, amounting to less than two in-plane pixels (1 pixel = 58 μm) in the MRI scans.

To determine whether the initial condition for the intact and meniscectomy situations was different enough to influence the results and conclusions of the experiment,
the recovery of cartilage thickness before re-compressing with the meniscectomy condition was measured. We expected full thickness recovery after 14 h, and found that the mean thickness recovery was about 90%. Because of the high compressive strains (over 80%) seen centrally following meniscectomy, and because of the additional time that passed after meniscectomy during static loading, we feel that this recovery was complete enough to not significantly influence the findings and conclusions of this study.

It is important to note that the magnitude of nominal compressive strain we report in this study is greater than the magnitude of nominal compressive strain one would expect to find in vivo. In this study, we statically loaded the knee joint to 1.5 times body weight for 2 h. This is nearly the peak joint reaction force of a sheep34 and applying this force statically to a knee joint for 2 h is not physiologic. The definitions of low, moderate, and high nominal strains that we used in this study are relative to the current study only, and cannot be used to interpret strain magnitude in in vivo situations. We expect, however, that the patterns of articular cartilage contact area and deformation seen in this study will reflect the patterns seen in knees that are loaded physiologically.

The magnitude of nominal strain we report in this study, however, is comparable to the nominal strain seen in other statically loaded in vitro studies18,25. We measured a mean nominal strain in the sheep medial tibial articular cartilage to be 36% with the meniscus intact, and 31% following meniscectomy. In a study using MRI to measure the time-dependent articular cartilage strain in a human patella-femoral joint loaded to 1.5 times body weight for 3.5 h the mean nominal strain in the articular cartilage was reported to be 30%26. To compare to the in vivo situation, Mosher et al. found that mean tibial articular cartilage thickness loss after 20 min of running was 14%26.

Medial meniscectomy alters the pattern of medial tibial plateau articular cartilage deformation under static loading. 3D cartilage models from the high resolution MR images showed that medial meniscectomy resulted in deeper and wider dimples in the articular cartilage surface compared to the meniscus intact case (Fig. 6). This finding was consistent with a previous study that investigated the effects of meniscectomy on the deformation of tibial plateau articular cartilage in rabbits using freeze fracture techniques and SEM18.

Measurements showed that medial meniscectomy reduces the medial contact area on the tibial plateau articular cartilage by approximately 60%. This is consistent with previous studies showing a 50–70% contact area reduction5. As a result of this reduction in contact area, a smaller percentage of the medial tibial articular cartilage surface is involved in transmission of load across the joint. With the meniscus intact, 64% of the articular cartilage surface experienced deformation in the range of 20–60% nominal strain, what we defined as “moderate” for this statically loaded study. Following meniscectomy, only 35% of the medial tibial articular cartilage surface was deformed in this range, and the proportion of low and high articular cartilage deformation areas was increased (Fig. 8). Interestingly, though meniscectomy is generally thought of as increasing contact stress on the condyle, it does so for a small area of the condyle, and a corresponding small percentage of the condyle experiences abnormally high deformation. A larger percentage of the tibial articular cartilage is unloaded due to the absence of a meniscus, and this articular cartilage experiences abnormally low nominal strain. Meniscectomy thus results in a significant increase in maximum nominal strain but insignificant changes in mean nominal strain.

The change in articular cartilage deformation after meniscectomy suggests a mechanism for OA development following meniscectomy. The central areas that are directly loaded by the medial femoral condyle following meniscectomy experience an increase in nominal strain (Fig. 7). This area develops deep fibrillation and softening in articular cartilage 16 weeks after medial meniscectomy13. Areas around the periphery of the condyle experience a decrease in nominal strain following meniscectomy (Fig. 7). This is where osteophytes develop following medial meniscectomy13. The general pattern of cartilage degeneration in areas of pathologic overloading, and osteophyte formation in areas of unloading has been previously described14. Excessive deformation of the articular cartilage may result in increased matrix damage and degradation by the chondrocytes, whereas decreased loading may lead to cartilage hypertrophy, endochondral ossification, and osteophyte formation.

Although sheep knee joints have anatomic and geometric differences from human knees, the pattern of central articular cartilage fibrillation and peripheral osteophyte formation on the sheep tibia following meniscectomy generally matches the pattern of degeneration commonly observed in the human tibia. Though the exact quantitative measure of articular cartilage deformation and nominal strain we report in the present study may be different than in humans, we do expect that the general observations made in this study — that meniscectomy results in a pattern of articular cartilage with greater deformation centrally and a large area of relatively undeformed articular cartilage peripherally, would be applicable to humans, and may provide insight into the mechanism of human cartilage degeneration following meniscectomy.

In summary, we successfully developed a loading system which can be used in a high field MRI scanner to image a whole sheep tibio-femoral joint while applying a magnitude of compressive load that is comparable to the peak joint reaction force during normal sheep gait. We validated the accuracy and precision of resulting articular cartilage thickness measurements and used this device to apply a static compressive load of 1.5 times body weight for 2 h, obtaining high resolution cross-sectional images of the tibial articular cartilage in the unloaded state, loaded intact, and loaded following medial meniscectomy. We found that meniscectomy results in a change in the pattern of medial articular cartilage nominal strain, and that this change relates to previously reported patterns of articular cartilage fissuring and softening and patterns of osteophyte formation in the joint. With this device, we are currently studying articular cartilage deformation under dynamic loading conditions, and hope that these ongoing studies will provide valuable insights into the mechanical influences on OA development.

Acknowledgments

We thank JJ&F market for providing the tissue specimens used in this study, Ocean Vendors for fabricating RF welded air bladders, and Dr. Garry E. Gold for his advice regarding the MR imaging. This work was supported by a Whi-}

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