In vivo cartilage contact deformation in the healthy human tibiofemoral joint

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Objectives. *In vivo* cartilage contact deformation is instrumental for understanding human joint function and degeneration. This study measured the total deformation of contacting articular cartilage in the human tibiofemoral joint during *in vivo* weight-bearing flexion.

Methods. Eleven healthy knees were magnetic resonance (MR) scanned and imaged with a dual fluoroscopic system while the subject performed a weight-bearing single-leg lunge. The tibia, femur and associated articulating cartilage were constructed from the MR images and combined with the dual fluoroscopic images to determine *in vivo* cartilage contact deformation from full extension to 120° of flexion.

Results. In both compartments, minimum peak compartmental contact deformation occurred at 30° of flexion ($24 \pm 6\%$ medial, $17 \pm 7\%$ lateral) and maximum peak compartmental deformation occurred at 120° of flexion ($30 \pm 13\%$ medial, $30 \pm 10\%$ lateral) during the weight-bearing flexion from full extension to 120°. Average medial contact areas and peak contact deformations were significantly greater than lateral compartment values ($P < 0.05$). In addition, cartilage thickness in regions of contact was on average 1.4- and 1.1-times thicker than the average thickness of the tibial and femoral cartilage surfaces, respectively ($P < 0.05$).

Conclusions. These data may provide base-line knowledge for investigating the effects of various knee injuries on joint contact biomechanics and the aetiology of cartilage degeneration.

KEY WORDS: Compartmental cartilage deformation, Knee joint, Biomechanics, Magnetic resonance, Cartilage thickness.

Introduction

A tremendous amount of research has been performed on the articular cartilage function in the human knee joint, seeking to understand better the biomechanical mechanisms of cartilage degeneration. Numerous studies have reported on the cartilage morphology and contact behaviour in human knees. Cartilage morphology of the knee has been extensively investigated using both cadaveric and living knee specimens with various techniques, such as needle probes, ultrasound, stereophotogrammetry and MRI [1–7]. These studies have provided valuable data about articulating surface area and thickness distributions in the knee. Many *in vitro* studies have also investigated tibiofemoral contact area and contact pressure under simulated loading conditions using thin film pressure sensors, casting materials, imaging techniques and finite element simulation [8–15].

Recently, several studies have presented data on tibiofemoral contact kinematics in living subjects. Both open and closed MRI modalities have been used to determine tibiofemoral contact areas and locations for a variety of activities [16–20]. Past studies have examined joint contact by analysing the interarticular distance between the bony surfaces of the tibiofemoral joint [21–23]. A combination of MR imaging and fluoroscopy has also been used to analyse cartilage contact locations [24, 25]. In addition to contact, deformation of articular cartilage after functional loading has been investigated using MR images [26].

While these previous studies have provided valuable data on cartilage function, the *in vivo* magnitude of tibiofemoral articular cartilage deformation during functional joint loading conditions remains unclear. The information on deformation among non-diseased knees could be used to provide norms with abnormal diseased cartilage deforming more than healthy cartilage. Also, the location of deformation during loading could identify where

loading and later injury might occur. In this study, our objective was to measure total compressive contact deformation in the tibiofemoral joint compartment during *in vivo* weight-bearing knee flexion of human subjects. To support these observations, the morphology of each living knee was analysed to determine the articulating cartilage surface area and thickness distribution. Using kinematic data, contact location and cartilage thickness, total contact deformation was determined throughout the full range of knee flexion.

Methods

MRI and dual fluoroscopic imaging

Eleven subjects (average age 31 ± 8 yrs, age range 23–40 yrs, five males and six females, five left and six right) without previous injury to the lower limbs and having healthy knees (normal quadriceps strength and limb alignment, absence of proprioception defects and symptoms of joint pathology) documented by clinical examination and MRI were recruited with informed consent. The study was approved by our institutional review board. To reduce the effects of residual cartilage deformation confounding the datum cartilage thickness, subjects were asked to refrain from all strenuous activity such as lifting, running, stair climbing for at least 4 h prior to their visit and to remain non-weight bearing for ~1 h prior to the MRI of the knee. Each subject was asked to lay supine with their knee in a relaxed, extended position while sagittal plane images were acquired with a 3.0 Tesla MR scanner (Siemens, Malvern, PA, USA). The MR scanner was equipped with a surface coil and employed a three-dimensional (3D) double echo water excitation sequence [field of view = $16 \times 16 \times 12$ cm, voxel resolution = $0.31 \times 0.31 \times 1.00$ mm, time of repetition (TR) = 24 ms, time of echo (TE) = 6.5 ms and flip angle = 25°]. Each scan lasted for ~12 min. The images were then imported into solid modelling software (Rhinoceros; Robert McNeel and Associates, Seattle, WA, USA) to construct 3D surface mesh models of the tibia, fibula, femur and articulating cartilage [22, 24]. The meshes were assembled using a point density of 80 vertices/cm² and triangular facets with an average aspect ratio of 2.

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The intensifiers of two fluoroscopes (OEC 9800; GE, Salt Lake City, UT, USA) were positioned in orthogonal planes and the subject's knee was located within the field of view while performing a single lunge activity so that fluoroscopic images could be taken simultaneously in the anteromedial and anterolateral directions [27, 28]. The subject was first asked to stand upright with the knee fully extended and the feet in a stance position. Then, using a hand-held goniometer to measure flexion angle, the subject was asked to flex their knee to 30°, 60°, 75°, 90°, 105° and 120° while their upper body remained upright (Fig. 1A). At each flexion angle, the subject was asked to pause for 5 s while simultaneous fluoroscopic images were taken. Throughout the experiment, the lower limb being tested supported the subject's body weight, while the other limb provided stability. The time elapsed between the MR scan and the lunge activity was ~15 min.

The fluoroscopic images and 3D bony surface models of the knee were then imported into the solid modelling software to create a virtual dual fluoroscopic imaging system using a previously presented method (Fig. 1B) [28]. The virtual system was created by generating two virtual cameras to recreate the position of the X-ray sources and placing the acquired fluoroscopic images representing the image intensifiers. The kinematics was then determined by manually manipulating the tibial and femoral models of the knee separately in six degrees of freedom (6DOF) so that their projections as viewed from the virtual cameras matched the bony contours of the fluoroscopic images. MR and dual fluoroscopic imaging techniques have been described and validated in detail in previous publications [27, 29].

Cartilage contact deformation

The *in vivo* knee kinematics represented by the matched knee models at each flexion angle was combined with the anatomical surface models of cartilage to determine cartilage contact during knee flexion. The tibiofemoral knee kinematics and cartilage morphology, including average cartilage thickness and contact area, of the 11 healthy human subjects are described in supplementary data Appendix A (supplementary data are available at *Rheumatology* Online). The validation of determining articular cartilage contact in the knee joint is discussed further in supplementary data Appendix B (supplementary data are available at *Rheumatology* Online).

FIG. 1. (A) Weight-bearing single-leg lunge; (B) recreating *in vivo* kinematics using a virtual orthogonal fluoroscopic environment.

In this study, a strain measure termed *compartmental contact deformation* was defined for each mesh vertex as the amount of penetration divided by the sum of the tibial and femoral cartilage surface thicknesses. Penetration was computed by determining the minimum Euclidian distance connecting a vertex of the reference cartilage mesh to the intersecting cartilage surface (Figs 2A and 3). Cartilage thickness was calculated by finding the smallest Euclidian distance connecting a vertex of the articular surface to the cartilage–bone interface [24]. Recently, a validation study on contact deformation measurement showed an accuracy of $\pm 4\%$ when this technique was used to detect cartilage deformation in human ankle joints [30].

Based on previous literature, two radial coordinate systems referenced to the femoral anatomy were used to uniquely determine the location of peak compartmental contact deformation on the femoral condyles (Figs 2B and C) [25]. The sagittal plane was defined perpendicular to the transepicondylar axis and passing through the distal–proximal axis. The 'contact angle' (α) was constructed by fitting a circle to the posterior curvature of each condyle in the sagittal plane (Fig. 2B). Next, a line was drawn from the centre of the circle to the point of interest on the cartilage surface defined as the 'contact line'. Contact angles were measured with respect to the distal–proximal axis. Reference geometry for the 'deviation angle' (β) was constructed by fitting a circle to each condyle in a plane passing through the contact line and perpendicular to the sagittal plane (Fig. 2C). Then a line was drawn from the centre of this circle to the point of interest. The angle between this line and the sagittal plane was defined as the deviation angle. Positive sense was given to deviation angles constructed from points of interest nearest the femoral notch.

The location of the peak compartmental deformation on the tibial plateaus was then recorded for each articulating surface. Locations of peak compartmental contact deformation were referenced to Cartesian coordinate systems on the tibial plateaus

FIG. 2. (A) Sagittal section of a right knee showing the definition of contact area and cartilage penetration. (B) Contact angle for femoral condyle in the sagittal plane for a right knee. (C) Deviation angle for medial and lateral femoral condyles in the coronal plane for a right knee. Dashed line shows delineation of medial and lateral femoral condyle cartilage boundaries. (D) Cartesian coordinate system for medial and lateral tibial plateaus of a right knee. The definition of X- and Y-axes correspond to the contact location data of Table 1.

(Fig. 2D) [24, 25, 31]. Circles were fit to the posterior edges of the medial and lateral tibial plateaus. The centre of each circle was the origin of the coordinate system. Next, the medial–lateral axis (X -direction) for the medial and lateral tibial plateaus was created by drawing a line through the origin of each coordinate system, parallel to the posterior edges of the tibial plateaus. Finally, the anterior–posterior axis (Y -direction) for each tibial plateau was created by drawing a line through the origin of the coordinate system perpendicular to the respective medial–lateral axis. The two quadrants nearest the tibial spine were defined as the ‘inner half’, and the remaining quadrants the ‘outer half’.

Statistics

The data was analysed statistically to examine cartilage deformation as a function of flexion angles during weight-bearing knee flexion and its relation to cartilage contact area and thickness. A two-way repeated measures analysis of variance (ANOVA) and the Student–Newman–Keuls (SNK) *post hoc* test were used to determine statistical differences in thickness, contact area and peak deformation between medial and lateral compartments and as a function of flexion angle during the weight-bearing flexion activity. Differences were determined to be statistically significant when $P < 0.05$.

Results

Magnitude of compartmental contact deformation

The peak contact deformation varied with flexion in both medial and lateral compartments (Fig. 4). Peak contact deformation in the medial compartment increased from $25 \pm 9\%$ at full extension to $30 \pm 13\%$ at 120° of flexion, but not significantly different along the flexion path ($P = 0.4618$). In the lateral compartment, peak contact deformation followed a similar trend with deformation of $22 \pm 10\%$ at full extension and $30 \pm 10\%$ at 120° of flexion. Peak deformation at 0 – 75° of flexion in the lateral compartment was found to be statistically lower than the higher flexion angles of 90 – 120° ($P = 0.0116$, 0.0016 and 0.0001 , respectively) (Fig. 4).

In general, peak contact deformation in the medial compartment was higher than that found in the lateral compartment at all flexion angles except 120° . Statistical differences in peak contact deformation between the compartments were noted at 30° ($P = 0.0013$) and 60° of flexion ($P = 0.0012$). Maximum peak deformation occurred in both compartments at 120° of flexion ($30 \pm 13\%$ medial, $30 \pm 10\%$ lateral) while the minimum peak deformation occurred at 30° of flexion ($24 \pm 6\%$ medial, $17 \pm 7\%$ lateral) (Fig. 4).

Location of contact deformation

Peak deformation, in general, was in the inner halves of both the tibial and femoral condyle cartilage (Fig. 5 and Table 1). The cartilage thickness located in areas of contact was significantly greater (1.3 times) than the total average thickness for all flexion angles in both medial and lateral compartments of the tibia (Fig. 5A). Lateral tibial cartilage in contact was, on average, 1.3-times thicker than cartilage contact in the medial plateau ($P = 0.0013$).

Regions of contact for the medial femoral condyle were, on average, 1.1 times thicker than the average medial condyle cartilage thickness (Fig. 5B). This difference in cartilage was statistically significant for 0 – 75° and 120° of flexion ($P = 0.0153$). In the lateral femoral condyle, cartilage thickness at contact was significantly greater (1.2-times) for flexion angles of 30 – 75° ($P = 0.0001$, 0.0024 and 0.0019). At later flexion angles of 90 – 120° , cartilage thickness at contact was 1.1-times thicker than the average cartilage thickness.

Discussion

Knowledge of *in vivo* cartilage contact deformation is critical for understanding cartilage function and biomechanical mechanisms related to cartilage degeneration. This information is also important for tissue engineering that aims to develop replacement materials for diseased cartilage. This study examined the compartmental cartilage deformation of the tibiofemoral joint during *in vivo* weight-bearing flexion of the normal knee.

Fig. 3. Illustration of the calculation of compartmental contact deformation. The MR images of the knee joint after ~ 1 h non-weight bearing (A) are used to determine the respective cartilage thicknesses of the femur (red outlines) and tibia (blue outlines) at rest (B). After matching the MR models to the fluoroscopic images captured during weight-bearing lunge (C), compartmental cartilage deformation is calculated by dividing the amount of penetration (1) by the sum of the femoral (2) and tibial (3) cartilage surface thicknesses, as illustrated in (D). (See colour figure online).

FIG. 4. Top: illustrative colour maps of the cartilage deformation displayed on the tibial plateaus at 0°, 75° and 120° of flexion, with intense red indicating higher deformation values. Bottom: peak tibiofemoral contact deformation for all flexion angles in the medial (blue bars) and lateral (yellow bars) compartment (* denotes statistical significance for $P < 0.05$). (See colour figure online).

FIG. 5. Thickness of cartilage in regions of contact for the (A) tibia and (B) femur for all flexion angles (* denotes statistical significance for $P < 0.05$). The black lines on the cartilage layers indicate the contact location throughout the range of flexion.

TABLE 1. Average value $\pm$ SD of peak tibiofemoral deformation location in the medial and lateral compartments for each flexion angle of the single-leg lunge

Flexion angle (deg)	Tibia				Femur			
	Medial		Lateral		Medial		Lateral	
	X (mm)	Y (mm)	X (mm)	Y (mm)	α (deg)	β (deg)	α (deg)	β (deg)
0	3.9 $\pm$ 6.1	6.1 $\pm$ 4.8	-7.3 $\pm$ 4.3	0.9 $\pm$ 2.3	-42 $\pm$ 10	30 $\pm$ 17	-12 $\pm$ 8	12 $\pm$ 7
30	3.5 $\pm$ 3.7	0.2 $\pm$ 3.3	-2.1 $\pm$ 4.4	-5.3 $\pm$ 2.8	5 $\pm$ 8	25 $\pm$ 12	21 $\pm$ 6	9 $\pm$ 10
60	1.4 $\pm$ 3.4	-3.6 $\pm$ 3.0	-2.6 $\pm$ 5.4	-7.7 $\pm$ 3.4	49 $\pm$ 8	19 $\pm$ 12	49 $\pm$ 6	15 $\pm$ 12
75	0.8 $\pm$ 2.9	-3.8 $\pm$ 2.9	-4.0 $\pm$ 5.0	-8.4 $\pm$ 2.8	66 $\pm$ 6	16 $\pm$ 9	61 $\pm$ 6	16 $\pm$ 13
90	-0.2 $\pm$ 4.4	-3.2 $\pm$ 3.0	-4.5 $\pm$ 4.6	-8.7 $\pm$ 2.4	79 $\pm$ 7	12 $\pm$ 9	73 $\pm$ 7	16 $\pm$ 13
105	1.0 $\pm$ 5.1	-2.9 $\pm$ 3.5	-3.4 $\pm$ 3.8	-9.6 $\pm$ 2.2	93 $\pm$ 7	14 $\pm$ 12	89 $\pm$ 6	13 $\pm$ 13
120	1.4 $\pm$ 4.2	-3.9 $\pm$ 3.8	-3.1 $\pm$ 4.0	-11.4 $\pm$ 2.1	109 $\pm$ 6	17 $\pm$ 9	105 $\pm$ 5	9 $\pm$ 11

X: location on tibia in medial-lateral direction, positive if lateral to anterior-posterior axis; Y: location on tibia in anterior-posterior direction, positive if anterior to medial-lateral axis; α : contact angle; β : deviation angle; deg: degrees.

A large amount of *in vivo* and *in vitro* data has been reported on cartilage morphology in the literature [1, 3, 4, 6, 7, 32–35]. Despite anatomical diversity, most literature agrees upon the same qualitative trends in cartilage morphology. The average thickness values from this study compare favourably with *in vivo* and *in vitro* data from the literature with values lying within the range of 1.6–2.4 mm (2.1 mm in this study) for the medial tibia, 1.8–3.5 mm (2.3 mm) for the lateral tibia, 1.6–2.3 mm (2.2 mm) for the medial femur and 1.6–1.8 mm (2.2 mm) for the lateral femur [5, 6, 34, 36]. These data also confirm the difference in cartilage thickness between the medial and lateral tibial plateaus.

The location of cartilage contact has previously been analysed by plotting the motion of the contact area centroid [16–20, 24]. In this study, contact location was defined as the peak compartmental contact deformation found in the tibial plateaus and femoral condyles. Qualitatively, this method had similar contact patterns to results presented using the contact area centroid [25]. Contact locations defined by the peak deformation were located more posteriorly in the lateral compartment when compared to the medial compartment. This result was consistent with the observed internal tibial rotation during weight-bearing flexion of the knee in previous studies [22, 25].

Data in this study showed that total cartilage contact deformation was greater in the medial compartment compared with the lateral compartment (Fig. 4). In addition, contact area was also greater in the medial compartment (Fig. A1), while cartilage was found to be thicker in the lateral compartment. Caution is advised when interpreting these findings. The increased contact area in the medial compartment could be explained by the complex cartilage geometry of the knee joint, with the medial compartment demonstrating a greater contact congruity (concave tibial surface) than the lateral compartment (convex tibial surface) [14, 37]. On the other hand, based on these data one might infer that the articular contact forces were greater in the medial compartment, as there are reports in the literature that more than half of the joint reaction forces of the knee pass through the medial compartment [38]. Contact deformation alone as obtained in this study is not sufficient to quantitatively describe the load distribution between the medial and lateral compartments. In the future, the load distribution could be calculated using 3D finite element calculation if accurate constitutive behaviour of the cartilage is known.

Loading effects on the average cartilage thickness and volume have been studied in previous research; however, there is no consistent conclusion regarding the effect of loading on cartilage morphology [3, 39–45]. Reported data have shown that joint immobilization is accompanied with a decrease in cartilage thickness [42, 44, 46]. Conversely, in certain cases cartilage thickness has reportedly increased with moderate exercise [3, 45, 46]. Other studies have noted that there is no apparent difference in cartilage thickness between athletes and inactive individuals [3, 40, 43].

The disparity in published results shows the difficulty in finding a correlation between average morphological parameters and the effects of *in vivo* cartilage loading. A localized examination of the cartilage morphology may provide a better insight into the relation between loading and morphology [25]. Data from this study indicated that the cartilage–cartilage contact area was only about 30% of the overall tibial cartilage area, but local thickness in regions of contact was significantly higher than the average thickness of the cartilage (Fig. 5). This finding was corroborated by previous publications, and lends credence to the hypothesis that cartilage–cartilage contact loading might be important to maintain local cartilage thickness [1, 24].

The magnitude of the peak compartmental cartilage deformation was measured to reach 30% during a weight-bearing single-leg lunge. Only a few studies have reported *in vivo* cartilage deformation in the literature. One recent study used MRI techniques to examine cartilage deformation after a jumping

activity and reported a maximum volumetric deformation of 7.2% in the lateral tibial plateau [36]. Values from these studies are a precursor to examining *in vivo* stress and strain of the articular cartilage. Our study further presented data on local values of deformation at discrete locations of the tibiofemoral joint. While volumetric and peak compartmental deformation may not have equivalent magnitudes, they both show similar trends in relative intensity. Also, it is imperative to take into consideration the loading conditions involved and the presentation of the data when making comparisons within the literature.

It should be noted that there were certain limitations presented in this study. The primary caveat to this study was that only cartilage–cartilage contact was investigated. This was due to the constraints of the imaging technique: motion and deformation of the meniscus is not detectable in the fluoroscopic images. In the literature, MR images have been used to determine meniscal motion at different flexion positions of living subjects [47, 48]. However, detecting real-time meniscus deformation remains a challenge and only very few quantitative data exist on *in vivo* tibiomeniscal kinematics. It was shown with axial load-bearing MRI and open MRI system that the menisci translate posteriorly with flexion [33, 49–51]. The lateral meniscus moves more than the medial, and the anterior horns more so than the posterior. These findings could be related to the contact location changes observed in the present study, as the location of peak deformation was located more posteriorly in the lateral compartment when compared with the medial compartment with increased flexion.

In this study, an unloaded cartilage datum was defined based on an *in vivo* knee, which might include any latent deformation and might contribute to inaccurate morphology. It should be mentioned that the cartilage-to-cartilage contact was determined as the overlapping of the cartilage surfaces, which may underestimate cartilage contact area and does not allow for determining the amount of deformation unique to either the tibial or femoral cartilage separately. In the future, 3D finite element calculation could be performed to better estimate actual contact area. Loading was also limited to a weight-bearing single-leg lunge. A complete understanding of articular contact mechanics requires an examination of a full spectrum of functional activities, such as gait, stair climbing, etc. Furthermore, it should be noted that no ground reaction forces were measured in this study. In the future, a force plate will be incorporated into the system.

In conclusion, this study investigated tibiofemoral cartilage contact deformation during a single-leg lunge of living subjects. It was found that the peak compartmental cartilage contact deformation ranged from 22% to 30% and was found to be significantly higher in the medial compartment compared with the lateral compartment. In addition, contact data presented in this study confirmed that cartilage–cartilage contact is primarily in the thickest regions of the tibiofemoral cartilage. The data presented in this study provides an additional baseline understanding of the normal cartilage function as well as a quantitative reference for future investigation of cartilage degeneration. In addition, the presented information on normal cartilage deformation during *in vivo* weight-bearing flexion of the knee may lend a deeper insight into the coupling of cartilage morphology and joint loading, which might be useful for the surgical improvement of the various tibiofemoral joint pathologies.

Rheumatology key messages

- Cartilage deformation ranges from 22% to 30% and is higher in the medial knee compartment.
- The data provides an insight in the normal cartilage function and a quantitative reference for future investigation of cartilage degeneration.

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Supplementary Data

Supplementary data are available at *Rheumatology* Online.

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