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In-vivo Tibiofemoral Cartilage Deformation during the Stance Phase of Gait

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Abstract

The knowledge of articular cartilage contact biomechanics in the knee joint is important for understanding the joint function and cartilage pathology. However, the in-vivo tibiofemoral articular cartilage contact biomechanics during gait remains unknown. The objective of this study was to determine the in vivo tibiofemoral cartilage contact biomechanics during the stance phase of treadmill gait. Eight healthy knees were magnetic resonance (MR) scanned and imaged with a dual fluoroscopic system during gait on a treadmill. The tibia, femur and associated cartilage were constructed from the MR images and combined with the dual fluoroscopic images to determine in vivo cartilage contact deformation during the stance phase of gait. Throughout the stance phase of gait, the magnitude of peak compartmental contact deformation ranged between 7% and 23% of the resting cartilage thickness and occurred at regions with thicker cartilage. Its excursions in the anteroposterior direction were greater in the medial tibiofemoral compartment as compared to those in the lateral compartment. The contact areas throughout the stance phase were greater in the medial compartment than in the lateral compartment. The information on in vivo tibiofemoral cartilage contact biomechanics during gait could be used to provide physiological boundaries for in vitro testing of cartilage. Also, the data on location and magnitude of deformation among non-diseased knees during gait could identify where loading and later injury might occur in diseased knees.

INTRODUCTION

The investigation of in vivo articular cartilage contact deformation in the knee joint is important for understanding the biomechanical aspects intrinsic to normal joint function and its associated pathologies. Despite the anatomical diversity, most literature agrees upon the same qualitative trends in cartilage morphology, with cartilage in the lateral tibial plateau being thicker than that in the medial plateau [1–5]. Using magnetic resonance (MR) imaging techniques, the location of tibiofemoral cartilage contact was found to pivot through the medial compartment,

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Author Disclosure of Potential Conflict of Interest

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with posterior femoral translation pronounced in the lateral compartment during knee flexion [6–8] and standing [9,10]. At the location of cartilage contact, cartilage thickness was significantly thicker than in non-contact regions [11], suggesting a link between loading and cartilage morphology. In contrast to the analysis of cartilage thickness and contact location, the deformation response of cartilage under in vivo physiological loading – a precursor to examining in vivo stress and strain of articular cartilage – has received limited attention. One recent study used MR techniques to examine cartilage deformation after a jumping activity and reported a maximum volumetric deformation of 7.2% [12]. Values up to 30% were described during a quasi-static lunge, when calculating cartilage contact deformation at discrete locations of the tibiofemoral joint [13].

During gait though, cartilage is subjected to cyclic loading and the loads are applied at fast rates, and act for short durations [14], much different from the loading seen during a lunge activity. Furthermore, certain kinematic phenomena that were measured during weightbearing flexion, such as the above described pivot through the medial compartment, have been shown to be opposite during level walking [15,16]. Therefore, the objective of this study was to determine the in vivo tibiofemoral cartilage contact biomechanics, including contact location, size of contact area, thickness at contact area, and resultant magnitude of cartilage contact deformation during the stance phase of gait – the predominant human locomotive activity. We used a combined dual fluoroscopic imaging system (DFIS) and MR image technique, which was previously employed to study the cartilage biomechanics during the single-leg lunge [13], and which was recently validated for the investigation of dynamic knee joint motion during gait [17].

MATERIAL AND METHODS

Subjects

Eight healthy subjects (aged 32–49 years, six males, two females, average body mass index of 23.5 kg/m^2) without history of knee joint pathology (i.e. injury, surgery, or systemic disease) were recruited for gait analysis using the DFIS technique [15]. Knee pathology was excluded upon physical and radiographic (MRI and X-ray) examination. The study was approved by the Institutional Review Board, and written consent was obtained from all subjects prior to participation in the study.

Imaging of the knee during treadmill gait and reproduction of kinematics

Each participant had one knee scanned (five left and three right) using a 3-Tesla MR scanner (MAGNETOM Trio[®], Siemens, Erlangen, Germany) and a double-echo water excitation sequence. The images were used to create 3D models of the bony surfaces of the tibia and femur in solid modeling software (Rhinoceros[®] 4.0, Robert McNeal and Assoc., Seattle, WA). The DFIS, previously validated for treadmill gait analysis [17], was employed to capture the knee motion during the stance phase of gait. In terms of measurement accuracy, using this technique it was possible to determine knee position with an accuracy of $0.16 \pm 0.61^\circ$ in rotation and $0.24 \pm 0.16 \text{ mm}$ in translation when compared to radiostereometric analysis (RSA). The cartilage contact area could be determined with $14 \pm 11\%$ error of measurement and finally, the cartilage thickness based on MR meshed models with $0.04 \pm 0.01 \text{ mm}$ error of measurement (corresponding to a $1.8 \pm 1.6\%$ difference).

Each subject practiced the gait on the treadmill for one minute at a treadmill speed of 1.5 miles per hour i.e. 0.67 m/s . Two thin-pressure sensors (Force Sensor Resistor, Interlink Electronics Specifications, Camarillo, CA) were attached to the shoes to record the heel-strike and toe-off of both feet. Two laser-pointers were incorporated into the DFIS to facilitate the knee positioning within the field of view of the fluoroscopes during stance phase. During the

adjustment the subject's natural gait and stride length were not altered. The knee was imaged during three consecutive strides with 30 snapshots per second.

The fluoroscopic images were imported into the modeling software and placed in calibrated planes to reproduce the position and orientation of the fluoroscopes. The 3D MR-based knee model was imported into the software and manipulated in six degrees of freedom (6DOF) until the projections of the model matched the bony outlines on the fluoroscopic images. This process was repeated at each 10% of stance phase starting from heel-strike until toe-off. Thereby, the series of image-matched knee models reproduced the motion of the subject's knee during the entire stance phase.[15]

Tibiofemoral cartilage contact

The tibial and femoral cartilage surfaces were constructed from the MR images of each subject. 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 gait. Cartilage contact was determined as the overlap area of the tibial and femoral cartilage surfaces (Fig. 1). The method for determination of cartilage contact has been described as the area of overlap of the cartilage models of tibia and femur ; [18]. Contact area (mm^2) was determined at each 10% of the stance phase. Cartilage deformation (%) was calculated by the penetration of cartilage mesh models (mm) divided by the thickness of cartilage at the same location (mm), multiplied by 100. Cartilage thickness (mm) was calculated in the regions of peak contact deformation.

To describe the changes in the tibiofemoral contact during the stance phase, we utilized a previously established technique whereby the locations of the peak contact deformation are recorded in tibial and femoral coordinate systems. Two radial coordinate systems referenced to the femoral anatomy were created. 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 femoral condyle in the sagittal plane (Fig. 2A). Next, a line was drawn from the center of the circle to the point of interest on the cartilage surface defined as the "sagittal contact line". Contact angles were measured with respect to the distal-proximal axis. Reference geometry for the "deviation angle" (β) was created by fitting a circle to each condyle in a plane passing through the coronal contact line and perpendicular to the sagittal plane (Fig. 3A). A line was drawn from the center 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 contact centroids on the tibial plateaus was recorded for each articulating surface referenced to Cartesian coordinate systems on the tibial plateaus. The center of each circle was the origin of the coordinate system. The medial-lateral axis (X axis) 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 (Fig. 4A). The anterior-posterior axis (Y axis) 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".

Statistical analysis

A two-way repeated measures analysis of variance (ANOVA) and a post-hoc Student-Newman-Keuls test were used to determine statistical differences in contact area, contact kinematics, peak contact deformation and cartilage thickness at peak deformation between medial and lateral compartments as a function of time. The independent variables were: percent stance phase and laterality. The dependent variables were: cartilage deformation, peak contact

deformation, contact kinematics and cartilage thickness. Differences were considered statistically significant when $p < 0.05$.

RESULTS

Medial tibiofemoral compartment

On the medial femoral condyle, peak deformation moved posteriorly and towards the intercondylar notch of the femur from 0 to 20% of stance phase as represented by a decrease in contact angle α and deviation contact angle β (-21.2° to 13.0° and -35.1° to 46.6° , respectively) (Fig. 2B, **3B**). Thereafter, the peak deformation moved anteriorly and away from the intercondylar notch as represented by α and β angles of -24.0° and 28.2° , respectively, at 60% stance. From there the direction of motion of peak contact deformation reversed and at toe-off the α and β angles were 24.8° and 47.8° , respectively.

On the tibial side, from heel-strike to 20% stance, the location of peak contact deformation shifted 2.3 mm anteriorly (Fig. 4). This was followed by a 2.0 mm posterior shift from 20 to 60% of stance phase. From 60% stance phase to toe-off, peak contact location moved 4.7 mm anteriorly. With regards to motion in the mediolateral direction, the peak contact first moved 0.4 mm laterally i.e. towards the intercondylar eminence of the tibia (heel-strike to 20% phase) then 0.5 mm medially (20 to 50% stance phase) and again laterally (1 mm) until toe-off.

The average overall cartilage thickness was 2.0 ± 0.2 mm and 2.3 ± 0.5 mm on the tibial and femoral condyles, respectively (Fig. 5). The average peak contact deformation at heel-strike was $8 \pm 5\%$ (Fig. 6A) and the tibial and femoral cartilage thicknesses at that location were 2.5 ± 1.2 mm and 2.4 ± 0.8 mm, respectively (Fig. 5). Peak deformation increased from heel-strike to 30% of stance phase ($p < 0.05$) where it reached the first amplitude ($23 \pm 6\%$) with tibial and femoral thicknesses of 2.6 ± 0.7 mm and 2.1 ± 0.6 mm, respectively. Thereafter, peak contact deformation decreased between 30% and 60% of the stance phase ($p < 0.05$) when it was measured $16 \pm 7\%$ and the tibial and femoral cartilage thicknesses were 2.8 ± 0.8 mm and 2.2 ± 0.8 mm, respectively. Peak contact deformation reached the second amplitude at 80% of stance phase ($22 \pm 5\%$ with tibial and femoral cartilage thicknesses 2.7 ± 0.8 mm and 2.1 ± 1.0 mm, respectively). At toe-off peak deformation was $17 \pm 4\%$ with corresponding tibial and femoral cartilage thicknesses of 2.8 ± 0.5 mm and 2.3 ± 1.0 mm, respectively.

The changes in contact area followed the changes in magnitude of contact deformation (Fig. 6B). From heel-strike to 30% stance phase the contact area increased from 235 ± 111 to 467 ± 61 mm² ($p < 0.05$). Thereafter, it decreased to 354 ± 97 mm² at 50% stance phase and increased to a second peak at 80% stance phase when it was measured 428 ± 87 mm². From 80% stance phase to toe-off the size of contact area decreased to 260 ± 140 mm² ($p < 0.05$).

Lateral tibiofemoral compartment

The lateral contact kinematics followed the patterns that were observed in the medial compartment, but with lower magnitudes (Fig. 2B, **3B**). After an initial posteriorly and laterally (towards the intercondylar notch of the femur) directed motion from 0 to 20% of stance phase (an increase in α and β angle -3.0° to 7.0° and 30.1° to 41.3° , respectively) the location of peak contact deformation moved anteriorly and away from the intercondylar notch with α and β angles of -5.2° and 29° , respectively, at 60% stance. From that point, the direction of motion of peak contact reversed and peaked at toe-off with α and β angles reaching 17.4° and 39.5° , respectively.

On the tibial side, the motion of peak contact deformation was of small magnitudes. From heel-strike to 80% stance, the peak cartilage contact deformation moved < 1.0 mm anteriorly and 0.7 mm posteriorly, thereafter. In the mediolateral direction, the peak contact deformation first

moved 0.6 mm laterally (heel-strike to 20% stance phase), then 0.9 mm medially (20 to 50% stance phase) and then 0.9 mm laterally again (Fig. 4B).

The average overall cartilage thickness was 2.3 ± 0.3 mm and 2.2 ± 0.4 mm on the tibial and femoral condyles, respectively (Fig. 5). The average peak contact deformation at heel-strike reached $7\pm3\%$ (Fig. 6A) and the tibial and femoral cartilage thicknesses at this location were 2.7 ± 0.7 mm and 2.2 ± 0.7 mm, respectively (Fig. 6). Peak deformation then increased and at 30% of stance phase it reached the first amplitude ($16\pm4\%$, $p<0.05$) and the tibial and femoral thicknesses were 2.7 ± 0.6 mm and 2.1 ± 0.7 mm, respectively. Thereafter, the peak contact deformation decreased until 50% of the stance phase ($p<0.05$) to $9.0\pm4.0\%$ with tibial and femoral cartilage thicknesses of 2.6 ± 0.7 mm and 2.0 ± 0.8 mm, respectively. Peak contact deformation reached its second amplitude at 80% of stance phase ($15\pm3\%$ with tibial and femoral cartilage thicknesses of 2.5 ± 1.0 mm and 2.3 ± 0.6 mm, respectively). At toe-off, the peak deformation was $9\pm4\%$ with corresponding tibial and femoral cartilage thicknesses of 2.9 ± 0.5 mm and 2.6 ± 1.0 mm, respectively.

The changes in contact area in the lateral compartment followed those observed on the medial side (Fig. 6B). From heel-strike to 30% stance the contact area increased from 200 ± 84 to 411 ± 159 mm². Thereafter, it decreased to 329 ± 97 mm² at 50% stance ($p<0.05$) and increased again to reach a second peak at 80% stance (451 ± 109 mm²). From 80% stance to toe-off, the contact area decreased to 331 ± 167 mm².

Comparison of medial and lateral compartment contact

Throughout the stance phase, the anteroposterior contact excursions in the lateral tibiofemoral compartment (1.6 ± 0.4 mm) were significantly smaller than in the medial compartment (3.6 ± 0.3 mm) (Fig. 4) ($p<0.05$). Concurrently, peak cartilage deformation was higher in the medial compartment than in the lateral compartment. The cartilage of the medial tibial condyle was thinner than the lateral condyle. From heel-strike to 50% stance, the average cartilage contact area of the medial compartment was higher than that of the lateral compartment. Conversely, later in the stance phase (50% stance phase to toe-off), the average cartilage contact area of the lateral compartment was higher than the medial compartment.

DISCUSSION

Understanding of in-vivo articular cartilage contact biomechanics in human knee joint during gait is instrumental not only for the comprehension of cartilage development, normal function, or disease pathogenesis but also for the development of therapies aimed at restoration of cartilage physiology. Physiological boundaries could be provided for the loading conditions of cadaveric cartilage specimens when designing in vitro experiments. In this study we investigated the tibiofemoral cartilage contact characteristics during the stance phase of gait using the combined DFIS and MR imaging technique. During the stance phase, excursions of peak contact deformation location and the contact area were greater in the medial than in the lateral tibiofemoral compartment. On average, the peak deformation was larger in the medial compartment than in the lateral compartment. Furthermore, peak cartilage contact deformation was located in regions with thicker tibial articular cartilage. These data indicated that the medial and lateral compartments of the knee experienced different contact biomechanics.

The data of contact location changes during the stance phase of gait were consistent with the 6DOF tibiofemoral motion patterns [15] [16,19]. Minimal translation of the lateral femoral condyle on the tibial plateau was measured during the stance phase [15]. Consistently, the data of this study indicated that the contact locations on the lateral tibial plateau surface moved within a range of less than 1.0 mm. The 6DOF kinematics data demonstrated that at heel-strike, the medial femoral condyle moved anteriorly on the tibial plateau. During midstance phase, as

the knee extends back, the medial femoral condyle shifted posteriorly. Throughout terminal extension and pre-swing the medial femoral condyle moved anteriorly on the tibia. The cartilage contact locations moved in the same pattern on the tibial cartilage surface during the stance phase. The shift of contact was significantly greater in the medial than in the lateral compartment, supporting the previous observation of a lateral pivot feature during stance phase [15,16,19]. This is also supported by the measurements of contact angle α which showed significantly greater excursions in the lateral compartment. Furthermore, the changes in contact angle α reached considerable magnitudes (on average up to 25°) suggesting that rolling of the femur on the tibia, rather than sliding, has a substantial contribution to the tibiofemoral motion during the stance phase. The positive values of contact angle β demonstrated that peak is occurring on the inner portions of the femoral condyles and closer to the tibial spine. This is the location where cartilage degeneration is frequently reported[20].

Much effort has been devoted to the study of in-vivo cartilage contact mechanics [7,17]. Li et al. [21] measured the positions of contact centroids during lunge and noted that the location of the contact centroid in the medial tibiofemoral compartment did not change significantly with flexion. On the other hand, the contact location in the lateral compartment changed significantly with flexion in both mediolateral and anteroposterior directions. Similar observations of greater contact excursions in the lateral compartment were obtained by Bingham et al.. These contact kinematics were consistent with the 6DOF knee kinematics during weightbearing knee flexion [7], which showed that the lateral femoral condyle had larger posterior translation than the medial condyle. Therefore, both the articular contact kinematics and the 6DOF knee kinematics during weightbearing flexion were not consistent with those observed during gait as measured in the present study. As indicated by Kozanek et al. [15], knee kinematics are activity-dependent. The cartilage contact kinematics during the stance phase of gait reported in this work may not be directly applied to explain cartilage functions during other different activities.

Another notable finding of this study was that the increases in cartilage deformation and contact area occurred after heel-strike. This corresponds to the first peak ground reaction force during the stance phase. The second peak contact deformation and contact area occurred at $\sim 80\%$ of stance phase, which corresponds to the second peak ground reaction forces during stance phase. Further, we observed that the peak deformation and area were greater in the medial tibiofemoral compartment, implying that the contact force might be higher in the medial compartment. It has been shown that forces acting across the medial compartment during stance phase are significantly higher than those of the lateral compartment [16,22,23], which along with the knee adduction torque [24] might account for the greater contact area and deformation.

Finally, we found that the maximal deformation was less than 25% throughout the stance phase. This compares well with other reports in the literature. Bingham et al. [13] reported magnitudes of contact deformation up to 30% of tibiofemoral cartilage thickness during lunge. Van de Velde et al. [18] measured contact deformation using the same methodology in posterior cruciate ligament-deficient knees with contact deformation values in the range of 15 to 25%. These data may be useful in designing ex-vivo experiments to investigate cartilage cellular activities where a physiological deformation is critical for examination of relevant cellular responses. The data are also important for designing constitutive behavior of cartilage repairing materials that should bear deformation levels experienced by cartilage during gait. In addition, we observed that articular contact during gait was located on the inner portion of the medial femoral condyle which was also reported to be the most frequent site of degenerative lesions in primary knee osteoarthritis [20].

Certain limitations of this study should be noted. Cartilage contact was determined as the overlap of rigid cartilage surface models obtained at the time of MR scanning, which may underestimate the contact area in-vivo [13]. In the future, 3D finite element calculation that

uses the overlap as displacement boundary conditions needs to be performed to calculate the actual in-vivo cartilage contact areas. We did not include the menisci into the model since they cannot be visualized on the fluoroscopic images. We studied the stance phase of gait because the swing phase could not be captured in its entirety within the imaging zone of the DFIS. Therefore, a description of the recovery of cartilage tissue after loading was not possible. Another limitation was that ground reaction forces were not measured due to the difficulty of integrating a force plate to the treadmill. Nonetheless, we believe the current findings provided a comprehensible insight in the changes in in-vivo tibiofemoral cartilage contact deformation, and identified important directions for future research.

In conclusion, this study presents the first in-vivo investigation of tibiofemoral articular cartilage contact deformation and kinematics during gait. The results showed that the peak contact deformation ranged from 7% to 23% of resting cartilage thickness and occurred in regions of thicker cartilage. We also found that the medial and lateral compartments experienced different patterns of articular contact biomechanics. The excursions of the peak contact deformation in the anteroposterior direction were greater in the medial than in the lateral tibiofemoral compartment, confirming that during the stance phase of gait the motion of the medial condyle in the transverse plane is greater and concurrently, the center of rotation is on the lateral side of the joint. Furthermore, the contact area was greater in the medial compartment. Finally, the contact kinematics during gait had different characteristics from those measured during single-leg lunge illustrating that extrapolating non-ambulatory measurements to ambulatory activities is unreliable.

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Fig. 1.
Determination of articular cartilage contact area and deformation.

Fig. 2. (A) Definition of sagittal plane contact angle α (°); (B) Sagittal plane contact angles of the describing the location of the peak contact deformation as a function of time during the stance phase of gait. Error bars indicate standard deviations.

Fig. 3. (A) Definition of coronal plane deviation contact angle β (°); (B) Deviation contact angles β as a function of time during the stance phase of gait. Error bars indicate standard deviations.

Fig. 4. (A) Excursions of the peak cartilage contact on the tibial plateau; (B) Anteroposterior and mediolateral kinematics of peak contact deformation on the medial and lateral tibial plateau during the stance phase of gait. Error bars indicate standard deviations.

Fig. 5. Cartilage thickness at the point of peak contact deformation on **(A)** femoral and **(B)** tibial articular surfaces. The red and green lines represent the average overall thickness in the lateral and medial compartments, respectively. Error bars indicate standard deviations.

Fig. 6. Cartilage contact characteristics during the stance phase of gait: **(A)** peak cartilage contact deformation and **(B)** cartilage contact area on the medial and lateral tibiofemoral compartments. Error bars indicate standard deviations. Asterisk denotes significant difference at $P < 0.05$. Because of the numerous comparisons we only present the relevant comparisons i.e. between the maximal and minimal values.