Effects of in vivo exercise on ankle cartilage deformation and recovery in healthy volunteers: an experimental study

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S U M M A R Y

Objective: To monitor ankle cartilage 3D volume changes after in vivo exercise and during recovery.
Method: Based on 3D MRI, 3D volumes of talar and tibial cartilage were calculated before and after 30 bilateral knee bends in 12 healthy volunteers. 3D volumes were calculated at five time points (one pre- and four post-scans) determining deformation and recovery for both cartilage plates of interest. Post-scans ran immediately after the exercise and were repeated according to a 15 min interval. 3D volumes were subjected to repeated measures GLM. Additionally, relative area use during deformation was compared between plates using a Wilcoxon Signed Ranks test and its correlation with deformation was investigated using Spearman’s rho.
Results: Mean 3D volume change percentages for talar cartilage after the exercise were: -10.41%, -8.18%, -5.61% and -3.90%. For tibial cartilage mean changes were: -5.97%, -5.75%, +0.89% and +1.51%. No significant differences in relative volume changes between both cartilage plates existed. Although no significant differences in relative surface area use between plates were revealed, a moderate to strong correlation with deformation existed.
Conclusion: Ankle cartilage endures substantial deformation after in vivo loading that was restored within 30 min for the talus. Overall cartilage contact area involvement might be associated with cartilage quality maintenance in the upper ankle. Talar cartilage is suggested to play a critical role in intra-articular shock attenuation when compared to tibial cartilage.

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Introduction

In Western society, osteoarthritis (OA) is one of the most frequent causes of pain, malfunction, and disability in adults. Due to the aging population, the prevalence of OA is expected to increase up to 40% the next 10 years labelling OA the fourth leading cause of disability. Therefore, knowledge on predictors for disease onset and progression are paramount in furthering OA management and prevention.

Mechanical factors have long been implicated in this disease’s aetiology. On the one hand, lack of cartilage conditioning to frequent loading has been hypothesized to predispose for degeneration when encountered with impulsive loads. On the other hand, joint tissue vulnerability to biomechanical insults has been suggested to depend in part upon the tissue’s resilience. Given the intriguingly low ankle idiopathic OA prevalence as compared to the knee, the study of ankle cartilage resilience to loading might enhance insights into its contribution in cartilage quality maintenance. Insight into joint degeneration aetiology requires knowledge of cartilage deformation in this joint. In this regard, the extensive work of Kuettner & Cole and co-workers using human cadaver specimen showed ankle (i.e., talar) cartilage to contain a higher proportion of proteoglycans to water when compared to knee (i.e., femoral) cartilage of the same cadaver. Combined with a lower hydraulic permeability due to a smaller effective pore size, these structural features have been suggested to result into higher dynamic stiffness to loading when compared to the knee. Hence, less deformation during in vitro loading has been proposed to explain in part the remarkably low prevalence
of idiopathic ankle OA. Surprisingly, the first in vivo studies reporting on ankle cartilage deformation under body weight suggested that ankle cartilage may be undergoing large deformations during daily activities. Using a dual-orthogonal fluoroscopic and magnetic resonance imaging (MRI) technique, considerable cartilage deformational strains were registered in the overlapping tibio-talar layers. Additionally, these authors' previous study reported considerable talar cartilage volume changes after four different exercises when compared to the available reports on knee joints. Hence, this apparent contradiction between in vitro and in vivo loading responses urges further investigation.

The study of cartilage deformational behaviour after in vivo loading conditions provides a means to encompass in vivo resiliency. Cartilage deformational behaviour entails that changes in 3D morphology (e.g., volume) are monitored before and after weight-bearing exercise. Registration of recovery after loading additionally allows for a comprehensive evaluation of cartilage resilience capacity. In vivo studies addressing recovery, however, are few in number, especially regarding the ankle joint. This is to our knowledge -- the first study monitoring cartilage recovery processes in the upper ankle following in vivo loading bouts.

The objectives of this study were twofold. The first objective was to investigate the changes in 3D talar and tibial cartilage volumes after an in vivo weight-bearing exercise and during recovery and, hence, determine the time required to restore initial volumes (i.e., recovery time). The second objective was to determine whether the two cartilage plates of interest displayed similar tendencies in deformation and recovery times by means of volume change comparisons at the different time points under study. It was hypothesized that ankle cartilage endures considerable deformation after in vivo loading. Since talar cartilage is known to show decreased stiffness and increased permeability in creep indentation experiments when compared to distal tibial cartilage, higher talar than tibial cartilage volume changes were expected to be observed. Considering ankle cartilage decreased hydraulic permeability, recovery time was hypothesized to proceed relatively slow for both cartilage plates displaying dominant talar cartilage involvement in intra-articular stress attenuation. Finally, in view of ankle cartilage stiffness, it was hypothesized that surface areas involved would contribute to ankle cartilage considerable volume changes.

Materials and methods

Subjects

Twelve healthy able-bodied subjects (six men, six women) participated voluntarily. All subjects were recruited from the local community or university campus.

Inclusion criteria were: age 20–40 years, Body Mass Index (BMI) 20–30 kg/m², injury free at the time of study, sports participation maximum three times/week, no changes in regular life style the week prior to the actual study appointment.

Exclusion criteria were: history of surgical or arthroscopic procedures, traumatic ligament injuries or chronic ankle instability, cartilage injury or degenerative pathology to the ankle joint, a history of fractures at the lower leg or foot as well as contraindications to MRI. On recruitment, eligibility was verified using a standard questionnaire.

All subjects were instructed not to practice sports the day before testing or on the testing day and to avoid running, lifting heavy weights and taking stairs 4 h preceding the actual experimental procedures. The right lower limb was the dominant limb in all participants and was defined as the limb the subject would choose to kick a ball. Informed consents ratified by the local ethics committee were obtained from all subjects. Subject demographics are depicted in Table I.

Experimental procedures

One weight-bearing exercise was examined. For every subject the testing procedures occurred at the same time of day.

In vivo exercise

The exercise consisted of 30 knee bends until maximal ankle dorsal flexion in 1 min. Maximal dorsal flexion was restrained to anterior rocking of the lower leg over the foot without heel lifts while lowering the upper leg horizontally. To control for a correct and standardised performance, the exercise was carried out under a researcher’s supervision and performed barefoot next to the scanner magnet.

MRI of cartilage morphology

Before (one pre-scan; tpre) and after the exercise (four post-scans; tpost0–15–30–45), high-resolution images of the right talocrural joint were obtained with a dedicated phased array high-resolution 8-channel Foot–Ankle coil (Invivo, Gainesville, FL, USA) on a 3 T Trio Tim magnet (Siemens medical solutions, Erlangen, Germany). Hence, a sagittal 3D double echo steady state sequence was applied with fat suppression by means of water excitation (sag3D DESS WE). The following parameters were implemented: partition thickness 0.4 mm, 104 partitions, echo time 5.5 ms, repetition time 15.6 ms, flip angle 28°, field of view 105 mm and matrix 384 pixels (in-plane resolution 0.3 × 0.3, interpolated to 0.125 × 0.125, acquisition time 07'19")

After 1 h of standardized physical rest, the pre-scans were performed followed by the exercise under study. Within 90 s after exercise cessation, the first post-scan is started (i.e., tpost0) and repeated with 15 min-intervals up to 45 min after the exercise (i.e., tpost15–30–45). The sequence of events is displayed in Fig. 1.

Data analysis

Three-dimensional reconstruction, volume calculation, model registration and surface area calculation were performed using a commercial solid modelling software package (Mimics version 13.1, Materialise, Leuven, Belgium). No custom codes were used.

Talar and tibial cartilage 3D reconstruction and volume calculation

MR image stacks were subsequently segmented to generate a 3D reconstruction of talar and tibial cartilage. A semi-automatic segmentation procedure was implemented based on grey value-oriented threshold (lower and upper threshold set at 105–533 respectively) and a slice-by-slice manual correction to digitize talar and tibial cartilage by masking (Fig. 2). Manual correction was preceded by a region growing algorithm to dispose of abundant voxels. Subsequently, applying contour interpolation, 3D cartilage plates were reconstructed and 3D volumes were calculated for pre- and post-scans. 3D volumes were calculated summing the pertinent voxels within the obtained binary volumes.

<table>
<thead>
<tr>
<th>Table I</th>
<th>Means (S.D.) for subject demographics</th>
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<tbody>
<tr>
<td>Parameter</td>
<td>Means (S.D.)</td>
</tr>
<tr>
<td>Age</td>
<td>29.08 (5.02)</td>
</tr>
<tr>
<td>BMI</td>
<td>22.08 (1.75)</td>
</tr>
<tr>
<td>Physical activity score</td>
<td>8.49 (0.98)</td>
</tr>
</tbody>
</table>

- Physical activity scores were determined using the reliable and valid Baecke questionnaire quantifying physical activity level during work, sports and leisure time activities.
Determination of relative surface area use for talar and tibial cartilage

For every pre-3D volume, total coverage area was determined by means of surface triangulation. To calculate the area of predominantly loaded surfaces during the exercise, the first post-3D volume was imported as an STereoLithography (STL)-format and superimposed on the pre-3D volume. Registration of models was guided by means of navigating their respecting contours matching the cartilage–bone interface using the index scan images as a reference. Predominantly loaded surfaces were identified as those regions where post-3D volumes remained covered by the pre-3D volumes\(^{14}\). Using orthogonal cut, regions of interest were distracted from the pre-3D model and their distinct surface areas were determined as calculated by the software package (Fig. 3).

Based on four test–retest measurements conducted in all participants prior to the actual experiment\(^{15,26,28}\), 3D volumetric measurements’ intra-tester reliability and inter-scan short-term precision error for talar cartilage attained an Intra-Class Correlation Coefficient (ICC) = 0.99 and Root Mean Square Coefficient of Variation (RMS CV) = 3.3% respectively and for tibial cartilage an ICC = 0.98 and RMS CV = 4.8% respectively. All segmentations were performed by a single researcher with 2 years of experience in cartilage segmentation at the time of analysis and who was blinded to the time sequence of scanning.

Statistical analysis

Absolute 3D volumes for talar and tibial cartilage were calculated at all time points. Relative 3D volume change percentages were calculated for each post-scan relative to the pre-scan using the following equation: \[\left(\frac{3D \text{ volume pre-scan}}{3D \text{ volume post-scan}}\right) \times 100\]\(^{14,24,27}\). To compare changes in volume after the exercise and between cartilage plates, a General Linear Model (GLM) with repeated measures was applied. Within-subject factors were ‘time of scanning’ (tpre and tpostt0–15–30–45) and ‘cartilage plate’ (talar and tibial). ‘Gender’ was allocated as between-subject factor. To adjust for multiple comparisons of main effects, a Bonferroni post-hoc test was implemented.

Relative surface area use was calculated using the following equation: \[\left(\frac{\text{predominantly loaded surface}}{\text{total cartilage coverage area}}\right) \times 100\]. To investigate the correlation between relative surface area use and 3D volume change, a Spearman’s rho correlation coefficient was calculated. To test the hypothesis that significant changes exist in relative surface area use between plates, a Wilcoxon Signed Ranks test was applied.

Level of significance for all tests was set at \(\alpha < 0.05\). PASW statistical package (version 18.0, Chicago, IL, USA) was used for all analyses.

Results

Absolute 3D volumes of talar and tibial cartilage at the five time points (i.e., tpre, tpost0–15–30–45)

Absolute mean (S.D.) talar cartilage 3D volumes at the five consecutive time points were: 2,207.20 (460.86) mm\(^3\), 1,966.79 (382.95) mm\(^3\), 2,017.73 (394.88) mm\(^3\), 2,083.98 (464.19) mm\(^3\), 2,119.11 (446.27) mm\(^3\). Accordingly, absolute mean (S.D.) tibial cartilage 3D volumes were: 1,427.49 (305.31) mm\(^3\), 1,342.27 (228.45) mm\(^3\)
mm$^3$, 1,335.21 (268.46) mm$^3$, 1,434.63 (335.41) mm$^3$, 1,454.62 (349.84) mm$^3$. In Fig. 4, the absolute mean volumes course is displayed for talar and tibial cartilage.

Relative 3D volume changes: deformation and recovery time of talar and tibial cartilage (i.e., $t_{postt0}$ to $t_{45}$)

Relative mean (S.D.) talar 3D volume change percentages at the four post-exercise time points respectively were: $-10.41\%$ (6.89; $P = 0.006$), $-8.18\%$ (6.07; $P = 0.009$), $-5.61\%$ (6.39; $P = 0.216$), and $-3.90\%$ (5.52; $P = 0.300$). All volume changes compared to baseline were significant except for the volume changes calculated at the two last time points (i.e., $t_{postt30}$ to $t_{45}$).

Accordingly, relative mean (S.D.) tibial 3D volume change percentages were: $-5.97\%$ (9.61; $P = 0.072$), $-5.75\%$ (7.94; $P = 0.364$), $+0.89\%$ (8.71; $P = 0.611$), and $+1.51\%$ (7.43; $P = 0.428$). No volume changes were significant when compared to baseline.

Given the number of subjects included, no significant between-subject effects for 'gender' were observed for both talar and tibial volume changes ($P = 0.776$ and $P = 0.965$ respectively). Mean (S.D.), median (95% Confidence Interval (CI)) as well as each subject’s individual volume change percentages for talar and tibial cartilage and stratified according to gender are displayed in Tables II and III.

Comparison of deformation and recovery time between talar and tibial cartilage (i.e., $t_{postt0}$ to $t_{15–30–45}$)

No significant differences between plates were revealed at all time points ($P = 0.103$, $P = 0.346$, $P = 0.163$, $P = 0.194$ for $t_{postt0}$ to $t_{15–30–45}$ respectively).

Relative surface areas used and the relationship with deformation in both talar and tibial cartilage (i.e., $t_{postt0}$)

For talar surfaces a mean (S.D.) 35.47 (10.25)% of the available surface was loaded, for tibial cartilage a mean (S.D.) 46.25 (22.02)%.

The Wilcoxon Signed Ranks test revealed no significant difference in relative use of surfaces between plates ($P = 0.176$). Spearman’s rho correlation analysis revealed a moderate to strong correlation between relative surface area use and 3D volume decrease observed at $t_{postt0}$ (i.e., talar cartilage: $r_s = 0.60$, $P = 0.044$; tibial cartilage: $r_s = 0.71$, $P = 0.036$).

Discussion

The main findings of this study suggest considerable deformation for talar as well as tibial cartilage after 30 knee bends. Talar cartilage recovered within 30 min after exercise cessation. For tibial cartilage, no volume changes were significant.
The first objective was to monitor deformation and recovery time after 30 knee bends for talar and tibial cartilage respectively. Present talar and tibial cartilage deformation fits with a previous in vivo study reporting mean talar volume decreases of 8.3% after a similar exercise\(^1\). Compared to significant volume decreases of 1.2–7.2% captured in tibio-femoral compartments after a variety of exercises (e.g., Refs.17,24,27), considerable deformation is observed in both cartilage plates investigated in this study. Absence of significance within the tibial cartilage plate, however, might be due to the combination of smaller volume changes to be measured with relatively larger precision errors, thus, requiring a relatively larger sample to reach 95% confidence. Additionally, as described below, scan duration and/or intervals might have hampered capturing tibial cartilage involvement in intra-articular stress attenuation.

Although applying a different technique to monitor in vivo deformational responses, Wan et al.\(^12\) and Li et al.\(^13\) similarly concluded with ankle cartilage being subjected to substantial deformation in daily life. Using a dual fluoroscopic and MRI technique, these authors determined real-time peak compressive strains (i.e., defined as the cartilage penetration divided by the thickness of the two overlapping cartilage layers at each vertex) of 34.5% and 38% in the overlapping tibio-talar cartilage layers under body weight. As the same technique documented on peak deformational strains of 22–30% and 10.5–12.6% in the tibio-femoral compartment\(^20,29\), the notion of ankle cartilage deforming substantially might be supported.

Surprisingly, in vivo observation does not appear to agree with previous in vitro biomechanical studies. In vitro studies (e.g., unconfined compression measurements, indentation probe testing) showed ankle (talar) cartilage to present with increased stiffness to loading (i.e., higher dynamic stiffness and lower permeability) when compared to knee (femoral) cartilage\(^9,19,30,31\). However, in vitro and in vivo measurements do not necessarily conflict. Outcome of in vivo deformation is suggested to depend on several factors other than local material properties alone\(^6,14\). In this respect, in vivo conditions display some characteristics that cannot be met during in vitro tests and, hence, might influence deformation outcome. Factors that come into play are the unknown physiological loads exerted onto the joint, variability and complexity of load distribution during dynamic activities including surface areas involved, joint lubrication, (in-)congruence,
variability in cartilage thickness distribution, mechanics of cartilage—cartilage contact, etc.\textsuperscript{14}. Additionally, actual in vivo outcome depends upon the manner in which deformation is expressed (i.e., overall volume change, local thickness change, deformational strains in overlapping layers). Cartilage material properties such as dynamic stiffness and permeability are thus suggested to contribute to in vivo deformational outcomes in this study, however, these properties are not the sole determining factors.

Actual changes in overall volume depend upon [local thickness changes × surface areas loaded]. In view of ankle cartilage increased stiffness, these authors previously proposed that the considerable overall volume changes are more likely to be caused by relatively extensive use of surface areas during joint articulation, rather than by considerable local thickness changes\textsuperscript{15}. In this regard, the ankle joint is known to adapt a state of increased congruence when loaded in order to assume a position of inherent stability\textsuperscript{12,31}. As congruence is, next to bony constitution (i.e., tight ankle mortise) attended to by the overlapping cartilage layers\textsuperscript{34}, small articular surfaces and limited range of motion might give way to substantial relative cartilage surface area involved\textsuperscript{4,35}. The fact that these authors revealed a moderate to strong correlation between ‘degree of deformation’ and ‘relative surface area used’ for both talar and tibial cartilage in the present study supports this hypothesis. Hosseini et al.\textsuperscript{29} similarly suggested that cartilage—cartilage contact brings about the difference in in vivo deformational responses determined between ankle and knee joints.

Alternatively, notwithstanding its stiffness, ankle cartilage might just experience larger deformation because of the higher intra-articular stresses to be attenuated by the articular surfaces when compared to other joints.

Nonetheless, for a complete comprehension of the present findings, functional intra-individual comparisons between joints

<table>
<thead>
<tr>
<th>Table II</th>
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<tbody>
<tr>
<td>Mean (S.D.) and median (95% CI) talar volume change percentages compared to baseline for the entire study sample as well as stratified according to gender, and for individual subjects for the four post-exercise time points</td>
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</table>

<table>
<thead>
<tr>
<th>Subjects</th>
<th>tpost0</th>
<th>tpost15</th>
<th>tpost30</th>
<th>tpost45</th>
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<tbody>
<tr>
<td>Total (N = 12)</td>
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<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Mean (S.D.)</td>
<td>-10.41 (6.89)</td>
<td>-8.18 (6.07)</td>
<td>-5.61 (6.39)</td>
<td>-3.90 (5.52)</td>
</tr>
<tr>
<td>Median (95% CI)</td>
<td>-9.09 (-18.72, -5.04)</td>
<td>-6.81 (-11.45, -3.13)</td>
<td>-4.26 (-6.90, -1.36)</td>
<td>-2.30 (-5.56, -0.58)</td>
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<tr>
<td>Female (N = 6)</td>
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<td></td>
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<tr>
<td>Mean (S.D.)</td>
<td>-8.66 (7.96)</td>
<td>-6.90 (7.70)</td>
<td>-6.30 (7.44)</td>
<td>-4.33 (7.81)</td>
</tr>
<tr>
<td>Median (95% CI)</td>
<td>-5.69 (-23.67, -2.82)</td>
<td>-3.25 (-21.84, -1.74)</td>
<td>-4.16 (-21.16, -1.36)</td>
<td>-1.54 (-20.10, 0.00)</td>
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<tr>
<td>Male (N = 6)</td>
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<tr>
<td>Mean (S.D.)</td>
<td>-12.17 (5.81)</td>
<td>-9.47 (4.21)</td>
<td>-4.91 (5.78)</td>
<td>-3.47 (2.35)</td>
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<tr>
<td>Median (95% CI)</td>
<td>-11.18 (-19.32, -5.03)</td>
<td>-10.63 (-15.12, -3.97)</td>
<td>-4.90 (-14.27, 2.65)</td>
<td>-3.07 (-6.72, -0.79)</td>
</tr>
</tbody>
</table>

Note that mean outcomes are prone to drift in data to a limited extent. Medians, however, remain to display considerable volume changes according to similar courses.

<table>
<thead>
<tr>
<th>Table III</th>
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<tr>
<td>Mean (S.D.) and median (95% CI) tibial volume change percentages compared to baseline for the entire study sample as well as stratified according to gender, and for individual subjects for the four post-exercise time points</td>
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<th>Subjects</th>
<th>tpost0</th>
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<th>tpost30</th>
<th>tpost45</th>
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<tr>
<td>Total (N = 12)</td>
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<tr>
<td>Mean (S.D.)</td>
<td>-5.97 (9.61)</td>
<td>-5.75 (7.94)</td>
<td>+0.89 (8.71)</td>
<td>+1.51 (7.43)</td>
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<tr>
<td>Median (95% CI)</td>
<td>-6.65 (-14.55, 2.56)</td>
<td>-4.46 (-14.24, -0.38)</td>
<td>+1.81 (-8.37, 9.25)</td>
<td>+1.50 (-6.15, 7.90)</td>
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<tr>
<td>Female (N = 6)</td>
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<tr>
<td>Mean (S.D.)</td>
<td>-4.23 (10.17)</td>
<td>-4.25 (7.33)</td>
<td>-4.37 (12.53)</td>
<td>-0.94 (6.99)</td>
</tr>
<tr>
<td>Median (95% CI)</td>
<td>-4.86 (-18.10, 6.48)</td>
<td>-4.98 (-14.24, 7.047)</td>
<td>3.72 (-9.25, 26.39)</td>
<td>-0.93 (-9.54, 10.80)</td>
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<tr>
<td>Male (N = 6)</td>
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<td></td>
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</tr>
<tr>
<td>Mean (S.D.)</td>
<td>-6.99 (10.95)</td>
<td>-8.26 (9.94)</td>
<td>+0.66 (10.20)</td>
<td>+3.96 (7.62)</td>
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<tr>
<td>Median (95% CI)</td>
<td>-7.30 (-22.52, 7.96)</td>
<td>-4.44 (-22.53, 1.02)</td>
<td>0.45 (-13.97, 11.62)</td>
<td>5.45 (-9.70, 12.64)</td>
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</tbody>
</table>

Note that mean outcomes are prone to drift in data to a limited extent. Medians, however, remain to display considerable volume changes according to similar courses.
remain appropriate as well as the incorporation of in vivo local thickness change computations.

Next to deformation, this study also includes recovery time. As pressure is relieved, fluid influx effects cartilage recovery according to a biphasic exponential function. In this study, talar cartilage volume changes were significant until 30 min after 30 knee bends. Patellar cartilage volumes have been shown to require more than 90 min for volume restoration after 100 knee bends. Sixty minutes after a 20 km run no significant volume changes could be detected anymore in tibial knee cartilages of experienced runners. Considering the difference in loading regimen, current talar cartilage recovery might be suggested comparable or even slightly slower than knee cartilage recovery. Nonetheless, recovery proceeded gradually, approximately linear in time.

Relatively slow recovery times reflect decreased hydraulic permeability characterizing ankle cartilage superficial layers. As low ankle cartilage permeability has been associated with increased in vitro stiffness to loading, this observation might underline the role of surface area involvement in the degree of deformation discovered.

Although no changes within plates were significant, tibial cartilage remarkably displayed positive volume changes following 15 min after exercise cessation. Assuming this effect only comes from loading in the resting state prior to the exercise seems reasonable to be argued upon. However, as a 1 h rest period is commonly applied in the literature (e.g., Refs. 14, 17, 24, 27), this study’s participants were instructed to refrain from excessive loading before the experimental procedures. Additionally, during scanning procedures, axial loading was prevented by instructing the participants to keep their feet in a relaxed position within the Foot–Ankle coil. Furthermore, if tibial cartilage demands extended recovery before exercise, volume changes would not resolve this fast within post-exercise recovery time spans. For these reasons, the authors propose that mechanisms other than unresolved pre-exercise recovery were likely to have contributed to this volume increment. In this respect, negative pressurization resulting from different cartilage plates’ recovery times within the congruent talocrural joint, is proposed to induce an additional ‘swelling’ of tibial cartilage. Alternatively, volume increases might be effected by fluid displacement from tibial to talocrural cartilage. Because of differences in Poisson’s ratio between fibular, tibial and talus surfaces, propensity for more fluid transport through the solid matrix of the tibial and talar cartilage as opposed to fibular cartilage is indicated. Additional increases in volume are physiologically feasible as cartilage has already been described to be subjected to an equilibrium compression state during daily life. Finally, considerable variation in cases of little deformation and/or positive volume changes measured with precision errors up to 4.0% has been previously reported. In this study, the second objective was to compare the course of deformation and recovery between both cartilage plates of interest. In this study, no significant differences in volume change percentages between plates were resolved after the exercise and at all the following time points under study. Differences in significant time-effects within plates, however, suggest potential differences between plates that possibly could not be statistically proven due to the relatively limited sample size. In this respect, a tendency is noted towards smaller tibial cartilage deformation when compared to talar cartilage. As these authors could not distinguish in relative surface area use between plates, smaller tibial cartilage deformation is most likely due to its decreased permeability and increased stiffness when compared to talar cartilage. Hence, larger deformation of talar cartilage resulting into an apparently slower recovery process might endorse the notion of talar cartilage more critically involved in intra-articular stress attenuation during joint articulation than tibial cartilage. As decreased tibial cartilage deformation possibly accounts for decreased involvement in cushioning intra-articular stresses, higher shear and compression stresses for the talar surfaces might be encountered. Athanasiou et al. proposed that disparities in the mechanical properties between two articulation surfaces produce dissimilar strain fields. The fact that several pathologic processes are mostly manifested in the talus (i.e., transchondral fractures of the talus, OCD lesions) should support this hypothesis. Furthermore, it is these authors’ contention that the relatively limited involvement of tibial cartilage in cushioning impact loads by means of less deformation, adds up to the difficulties in detecting significant volume changes during deformation and recovery in this study as mentioned before.

Unlike dual-orthogonal fluoroscopic and MRI techniques providing with real-time monitoring, deformation in this study was evaluated through post-exercise effects to avoid motion artefacts MR imaging is susceptible to. Consequently, scan duration might underestimate true deformation and/or miss early recovery. As the majority of volume gain should be registered during early recovery, talar cartilage volume decreases were significant until 30 min after exercise cessation. As in between the first and second post-scan (~ twice the scan duration), no significant difference in volume could be achieved (mean difference: 50.95 mm³; P = 0.135), these authors suggest that recovery occurred slow enough to be captured by the scan sequence. On the other hand, tibial cartilage was suggested to show limited contribution to shock attenuation by means of deformational responses. Although the course in absolute volumes suggests restoration of the pre-exercise state before 30 min after exercise cessation (Table III), scan sequences and/or intervals were possibly too long since no significant difference in volume could be established between all distinct time points.

Additionally, as high-resolution imaging is paramount to produce accurate measurements, thin cartilage layers are known to produce relatively larger precision errors. In this regard, the combination of the precision error and the registered (or expected) volume change is crucial to allow for statistical significance to be established. As a guideline, the minimal interval of change that can be detected with 95% confidence in a single individual is 2.8 times the precision error. Hence, it is reasonable that talar cartilage changes reached significance where tibial cartilage did not. Although this study was the first including single tibial cartilage plates, attaining significant effects for “time” in case of larger errors requires larger samples to increase the likelihood for significance. Notwithstanding the ability to detect talar cartilage deformation (i.e., tps0t0), relevant precision errors were too extensive to resolve late recovery (from tps3t30) changes with 95% confidence. Nonetheless, early recovery of talar cartilage is not missed which encompasses the most important and critical changes after pressure release. Although the importance of precision should be acknowledged, one needs to mention that the present precision errors concur with or were smaller than ranges reported previously when applying high-resolution ankle cartilage morphology imaging.

In view of overall volume changes, this study did not include local thickness change calculations next to loaded surface areas. Assessing local thickness changes would enable (1) confirming the hypothesis that extent of deformation is primarily driven by surface area use or not, (2) exploring whether load transfer is characterized as being (in-)homogenous. Nonetheless, the latter has been addressed before and was considered beyond the scope of this study. Additionally, given the circumstances of thin cartilage within a congruent joint, these authors suggest that overall 3D volume changes might be preferred over local thickness in successfully detecting morphometric differences within single plates. In this respect, measurement of local thickness changes would rely even more on high precision data processing which is recognized as an inherent technical challenge when dealing with thin cartilage layers.
Conclusion

The present results reveal considerable in vivo deformation for both talar and tibial cartilage that might be primarily driven by relatively extensive surface areas involved in joint articulation. The suggested substantial volume changes recovered relatively slow, more specifically within 30 min for talar cartilage. Tendencies towards smaller deformation for tibial when compared to talar cartilage and slower recovery occurring in the latter were disclosed, suggesting critical involvement of talar cartilage in shock attenuation in the upper ankle. Limitations in sample size, precision error, scan duration and/or intervals possibly hampered adequate monitoring of tibial cartilage deformational behaviour. In a key next step, the study of cartilage deformational behaviour in the OA patient should be addressed. Additionally, other recreational or sports-specific activities such as running or jumping, warrant attention.

Author contributions

Van Ginckel An: conception and design, analysis and interpretation of the data, drafting of the article, critical revision of the article for important intellectual content, final approval of the article. Van Ginckel Ans: conception and design, analysis and interpretation of the article, drafting of the article, critical revision of the article for important intellectual content, final approval of the article. Almqvist Karl Fredrik: conception and design, critical revision of the article for important intellectual content, technical support, final approval of the article. Verstraete Koenraad: conception and design, critical revision of the article for important intellectual content, final approval of the article. Witvrouw Erik: conception and design, critical revision of the article for important intellectual content, final approval of the article.

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Conflict of interest

No conflicts of interest were declared.

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