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Knee cartilage MRI with in situ mechanical loading using prospective motion correction

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Abstract

Purpose—To assess the feasibility of high resolution knee cartilage MRI with in situ mechanical loading using optical tracking to compensate for motion.

Methods—In vivo cartilage MRI with in situ mechanical loading is demonstrated on a clinical 3T system for the patellofemoral as well as for the tibiofemoral knee joint using a T1-weighted spoiled 3D gradient-echo sequence. Prospective motion correction is performed with a moiré phase tracking system consisting of an in-bore camera and a single tracking marker attached to the skin.

Results—Rigid-body approximation required for prospective correction with optical motion tracking is fulfilled well enough for the patellofemoral as well as for the tibiofemoral joint when the tracking marker is attached to the knee cap and the shin, respectively. Presaturation proves to be efficient in suppressing pulsation artifacts from the popliteal artery and residual motion artifacts primarily arising from non-rigid motion of the posterior knee compartment.

Conclusion—The proposed technique enables knee cartilage imaging under in situ mechanical loading with sub-millimeter spatial resolution devoid of significant motion artifacts and thus appropriate for cartilage volumetry. It has the potential to provide new insight into the biomechanics of the knee and might complement the panoply of diagnostic MR methods for osteoarthritis.

Keywords

osteoarthritis; knee; cartilage; MRI; prospective motion correction; in situ mechanical loading

INTRODUCTION

Osteoarthritis of the knee joint represents a major cause of morbidity in the adult population. While X-ray imaging can only diagnose the disease at an already advanced and irreversible state, magnetic resonance imaging (MRI) can be used to probe early structural and biochemical changes in articular cartilage associated with osteoarthritis (1,2). Cartilage thickness as well as T_2 and $T_{1\rho}$ relaxation times have been shown to serve as diagnostic markers for chondromalacia of the knee joint at an early stage, providing insight into the degree of cartilage damage and dysfunction (3,4) and enabling treatment evaluation after cartilage repair surgery (5).

Since chondromalacia is associated with altered mechanical properties of the cartilage, the response of these tissue parameters to mechanical loading is of particular interest. A plethora of MRI studies have investigated changes in cartilage thickness, as well as T_2 , in response to different types of static or dynamic loading conditions such as knee bending or running (6–11). Cartilage deformation has also been studied with in-situ mechanical loading in MRI experiments using cadaver joints (12–15). However, simulating physiological loading in situ is challenging due to the complexity of the knee joint and the load distribution over several knee compartments. Furthermore, the mechanical properties of the cartilage may change during and after extraction. The long-term objective is a diagnostic method complementing the morphological and relaxometric information provided by standard clinical knee cartilage imaging with additional biomechanical information. To date, only few in vivo studies have performed knee cartilage MRI with in situ loading. The challenge of building a MR-compatible loading device that enables physiologically meaningful loading paradigms has been addressed in several studies, though most of them aimed to investigate joint kinematics (16). However, the biggest challenge arises from the subject motion that is inevitably associated with knee loading. Two works have investigated T_2 relaxation changes induced by in situ loading, applying load to the stretched leg and focusing on the tibiofemoral compartment (17,18).

In contrast, loading of the patellofemoral compartment requires knee flexion, which impairs the stability of the experimental setup and therefore gives rise to considerable motion artifacts in the acquired MR images, especially since this flexed position has to be maintained for several minutes during the scan. Connolly et al. investigated patellofemoral contact mechanics in healthy subjects as well as in subjects with patellofemoral pain syndrome by applying isometric in situ loading at various knee flexion angles (19). However, they only applied very weak loading (10 % of the maximum voluntary force, 10–13 N) to ensure adequate image quality without significant motion artifacts. In vivo cartilage MRI studies of the patellofemoral joint with strong isometric loading have not been performed to date.

Magnetic resonance imaging of the brain with prospective motion correction using an external camera-based system was demonstrated by Zaitsev et al. (20) and has been validated for various sequences and setups at several different sites (21–27). Recently, the method has been refined by using a moiré phase tracking (MPT) system, based on an in-bore camera and a passive marker that generates moiré patterns to quantify out-of-plane rotations.

This enables motion detection in all six degrees of freedom from a single image with high precision and with minimal subject discomfort (28). Since only rigid-body motion can be corrected with this method, the focus of motion correction has so far been on brain MRI.

In this proof-of-concept work we show that small involuntary knee motion during a MR examination can be reliably tracked and corrected via a marker attached to the patella or the shin, depending on whether the patellofemoral or the tibiofemoral cartilage is to be imaged. For the first time, knee MRI with prospective motion correction is demonstrated, thus enabling the quantification of cartilage deformation even in the patellofemoral joint under in situ mechanical loading.

MATERIALS AND METHODS

The experiments were performed on a Magnetom 3T Trio a TIM system (Siemens Healthcare, Erlangen, Germany). Prospective motion correction was performed with a tracking system (Metria Innovation, Milwaukee, USA) consisting of a single camera mounted inside the scanner bore (Fig. 1a) and a tracking marker with a multilayer structure, which generates moiré patterns for accurate orientation measurement (28). Communication with the tracking system was implemented directly on the real-time control unit of the scanner as previously proposed by Zaitsev et al. (20). The tracking system reported marker motion in all three translational and three rotational degrees of freedom with a position update rate of 80 frames/s and a tracking precision in the order of 0.02 mm in translation and 0.01° in rotation (root mean square) (28). Under the rigid-body approximation, motion of any point $\vec{x}$ in the imaged object undergoes a transformation, which can be described by $\vec{x}' = \mathbf{R} \cdot \vec{x} + \vec{t}$, where the matrix $\mathbf{R}$ describes a rotation and the vector $\vec{t}$ a translation. Using the tracked motion data, RF pulse and receiver frequencies of the MRI sequence were updated in real time to account for the translation $\vec{t}$, while the 3D gradient waveform $\vec{g}$ was updated to account for the rotation according to $\vec{g}' = \mathbf{R} \cdot \vec{g}$. For more details the reader is referred to pertinent literature (20,29). All slice objects defined in the sequence were updated so that the imaging volume as well as the presaturation slabs followed the knee motion. The coordinate update was performed with an averaged latency of about 35 ms. It was implemented to take place once per TR right before the excitation pulse. Motion correction was performed without a scan time penalty. Inter-scan 'position locking' corrected for subject motion between the scans and ensured that all motion-corrected imaging scans of a session were performed with the initially planned geometric setup.

All subject studies were conducted with the approval of our institutional review board. Knee loading was realized with a home-built MR-compatible loading jig (Fig. 1b). The jig was designed so that the subject was able to press against a vertical plate while a shoulder fixation kept him in a stable position and prevented him from sliding backwards. The MRI scans were started at least 50 s after the onset of loading to ensure approximately steady-state cartilage deformation (30). All measurements were performed with a T1-weighted spoiled 3D gradient-echo sequence using slab-selective water excitation. In some experiments, the signal from the posterior part of the knee was erased with a coronal presaturation slab to mitigate artifacts arising from non-rigid-body motion and pulsatile arterial flow in the posterior part of the knee.

For imaging of the patellofemoral joint, the tracking marker was fixed to the knee cap of the subject using a double-sided adhesive tape and an 11 cm loop coil was used for signal reception (setup 1, Fig. 1c,d). To ensure strong loading of the patellofemoral joint, imaging was performed with a knee flexion angle of approximately 50°. For validation purposes, axial 3D images were obtained from one healthy subject with the following parameters: TE = 9.8 ms, TR = 26 ms, selective excitation of axial slab with flip angle = 15°, readout bandwidth = 130 Hz/pixel, readout along the right-left (RL) direction, FOV = (160 × 160 × 44) mm³, in-plane resolution of the axial images: 0.6 mm, through-plane resolution: 2 mm, scan duration = 2 min 30 s. The imaging setup is shown in Fig. 2. The imaging protocol was run with and without loading as well as with and without motion correction. To assess the influence of non-rigid-body motion and pulsatile arterial flow, all experiments were performed twice with and without a presaturation slab applied to the posterior part of the knee. An alternative remedy for the residual motion and pulsation artifacts in the region of interest is switching the readout direction to AP, since these artifacts appear only along the phase encoding directions, allowing imaging protocols without presaturation and thus with a shorter TR. Therefore the measurement protocol was slightly modified and a high-resolution scan was performed in six healthy subjects with the readout switched to the AP direction, using prospective motion correction with TE = 6.9 ms and TR = 16 ms. Furthermore, the resolution was increased to 0.4 mm in plane and 1 mm through plane, amounting to a scan duration of 3 min. Scans without and with loading were performed with this protocol.

For imaging of the tibiofemoral compartment, the tracking marker was taped to the shin close to the tuberositas tibiae and a birdcage circularly polarized extremity coil was used for signal reception (setup 2, Fig. 1e,f). When setting up the experiment, the camera and leg have to be positioned such that the tracking marker is visible for the camera and is not hidden by the rungs of the coil. Imaging was performed in one healthy subject with knee flexion of approximately 10°. Initially, 3D images with the readout along the FH direction were acquired for demonstration purposes. Then high-resolution 3D images were acquired with the following parameters: TE = 6.9 ms, TR = 16 ms, selective excitation of axial slab with flip angle = 15°, readout bandwidth = 130 Hz/pixel, FOV = (160 × 160 × 44) mm³, resolution: 0.4 mm (AP), 0.8 mm (RL), 0.5 mm (FH), scan duration = 3 min. The protocol was run four times with and without knee loading as well as with and without motion correction.

From the high-resolution data obtained with setup 1, patellar cartilage compression was determined to investigate the influence of mechanical loading on the patellofemoral joint. The load-induced decrease of the patellar cartilage thickness at the lateral facet was quantified in six subjects using the viewing application of the Syngo software (Siemens Healthcare, Erlangen, Germany). From the images obtained with setup 2 in one subject, cartilage compression in the tibiofemoral joint was quantified accordingly. Standard deviations were determined through measurements by several experts.

RESULTS

Figure 3 shows the results of the MRI experiments performed with setup 1. Even scans without knee loading benefitted from motion correction, as can be seen from the reduction

of artifacts in Fig. 3f compared to Fig. 3b. The quality of the uncorrected images acquired with knee loading was strongly affected by motion (Fig. 3c,d), but with prospective motion correction an image quality similar to the measurements without loading was obtained (Fig. 3g,h). In the measurements without a coronal presaturation slab, pulsation artifacts from the popliteal artery impair the image quality. These artifacts are usually removed with a proximal transverse presaturation slab, but a coronal slab has the advantage of crushing all signal from the posterior part of the knee, which does not fulfill the rigid-body approximation with respect to the knee cap. Since the foot plate of the loading jig slightly yielded under the strong force applied by the subject, the knee position changed from the unloaded to the loaded setup (Fig. 3c,d). Position locking ensured that the imaging volume did not change in between motion-corrected scans (Fig. 3g,h). However, there is still a slight slice mismatch between the unloaded and loaded setup, which is due to the decreased flexion angle of the knee in the loaded setup, which in turn is associated with slight skin motion with respect to the patella. This mismatch can be reduced with a more rigid loading jig.

Motion-corrected high-resolution images (resolution: 0.4 mm in-plane, 1 mm through-plane) acquired with setup 1 and the readout along the AP direction are shown in Fig. 4. Pulsation artifacts from the popliteal artery appear along the left-right phase encoding direction, but no motion artifacts are visible in the patellar cartilage region. The two images (left: without loading, right: with loading) show the same slice of the knee cap and the patellar cartilage, while the femoral bone and the soft-tissue structures at the posterior part of the knee are slightly displaced due to load-induced muscle activation. The boundary between the patellar and the femoral cartilage is visible more clearly in the image acquired with loading than in the image acquired without loading. This observation was confirmed in scans on other subjects. Since loading of the flexed knee causes a lateralization of the patella, patellar cartilage compression is strongest in the lateral region. Quantification was performed on the transverse images and yielded a measureable load-induced decrease of patellar cartilage thickness at the lateral facet in five out of six subjects (Table 1).

Figure 5 shows a sagittal image acquired with setup 2 using motion correction and the readout along the FH direction. The image quality is strongly impaired by artifacts arising from plantaris motion, which does not fulfill the rigid-body requirement and thus cannot be corrected. Therefore further experiments were performed with the readout in the AP direction. The results of the high-resolution experiments performed with setup 2 are presented in Fig. 6. There is no obvious improvement through motion correction for the measurements without loading. However, for the loaded setup significant ghosting and blurring of the cartilage contours can be recognized in the uncorrected images (Fig. 6c), which can be prevented with motion correction (Fig. 6d). Comparing the motion-corrected images acquired with the unloaded and loaded setup demonstrates load-induced cartilage compression at the lateral tibiofemoral compartment (compare Figs. 6b and 6d). This compression is slightly disguised by motion-induced blurring in Fig. 6c. Quantification yielded a decrease of the overall (tibial + femoral) cartilage thickness from (5.5 ± 0.2) mm to (5.0 ± 0.2) mm in the region of strongest compression (see arrows in Fig. 6).

DISCUSSION

Strong knee loading always gives rise to involuntary motion, which has hampered MRI studies with in situ mechanical loading to date. The results presented in this work demonstrate that knee cartilage imaging with in situ loading can be improved by prospective motion correction. For imaging of the patellofemoral knee compartment, the rigid-body approximation is fulfilled well when the tracking marker is attached to the knee cap. In contrast to a birdcage coil, a loop coil does not restrict the knee flexion angle, which is a prerequisite for proper loading of the patellofemoral cartilage. However, it should be noted that the bore size of the scanner and the fixed focal length of the camera restricted knee flexion to approximately 60°. Using a loop coil of appropriate size yields sufficient SNR for quantitative cartilage studies and ensures marker visibility at all times, while artifacts arising from non-rigid-body motion in the posterior knee compartment are mitigated through the localized coil sensitivity. For imaging of the tibiofemoral compartment the marker should be fixed at the shin close to the tuberositas tibiae. It has to be noted that marker stability can be impaired in obese subjects with thicker subcutaneous fat tissue. Apart from that, marker visibility is an issue when a birdcage coil is used for signal reception. In this work, a circularly polarized extremity coil with thin rungs was used so that with proper alignment of the camera and the knee inside the coil even stronger motion during the scan did not affect marker visibility. However, the best image quality is usually achieved using phased-array coils with a closed shape. This problem could be circumvented by attaching the marker to the distal part of the shin outside the coil. However, a larger distance between the marker and the knee will affect the accuracy of correction, as small errors in measurement of marker angle due to skin motion will result in large translation errors in the region of interest. Even though the tibiofemoral joint can be loaded without knee flexion, measuring the load-induced tibiofemoral cartilage deformation in the bent knee might provide additional insight into the knee biomechanics. For such measurements a standard birdcage coil, which restricts knee flexion, would be inappropriate and a stretchable coil array should be used instead (31).

For the investigation of cartilage deformation, the same protocols have to be run with the unloaded and loaded setup for comparison. Due to the limited rigidity of the available loading jig, knee positions can be significantly different between the unloaded and the loaded setup, so inter-scan position locking and image alignment is an important feature for such experiments. The acquisition of scout images and scan planning can then be performed with the unloaded setup without having to readjust the volume of interest and the presaturation slab between further scans. In particular for imaging of the tibiofemoral joint, proper positioning of a coronal saturation slab is a delicate task since on the one hand as many non-rigid structures from the posterior knee compartment as possible should be saturated, but on the other hand the investigated cartilage should not be affected. Motion artifacts from the posterior knee compartment could also be mitigated by using a 2D sequence with coronal slices instead of a 3D sequence. However, for quantitative volumetric cartilage measurements 3D sequences are preferred since they enable high spatial resolution in all three dimensions with sufficient SNR in an acceptable scan time. In our experiments, no additional knee stabilization with foam pads was used. Especially for setup 2, this could

reduce subject motion. However, foam padding would influence the load distribution over the knee so that the setup would no longer reflect physiological loading conditions.

Since pulsation and non-rigid-body motion give rise to artifacts along the phase encoding directions, presaturation of the popliteal artery and the plantaris is strongly recommended for 3D imaging protocols if phase encoding is performed along the AP direction. As demonstrated in this work, switching the readout to the AP direction is another viable remedy, albeit at the expense of some minor drawbacks. Since geometric distortions due to susceptibility inhomogeneities only occur along the readout direction, they should have a stronger influence on cartilage volumetry when the readout is perpendicular to the cartilage surface. Fat-water chemical shift artifacts along the readout direction can also impair the accuracy of volumetric measurements, but they can be neglected when selective water excitation is used as in this work. Non-rigid-body motion of the soft tissue on the medial side of the knee cap (e.g., through muscle twitching of the vastus medialis during loading) might give rise to artifacts in the cartilage region when phase encoding is applied along the RL direction. Such artifacts were not observed in our experiments though. Measurements without presaturation are not only desirable because of the shorter scan time, but also because of the better contrast between cartilage and synovial fluid for shorter TR. When long high-resolution scans are performed with presaturation in vivo, the minimal TR is not only increased through the time needed for the additional pulses and gradients, but also through SAR limitations, which require either a TR increase or a flip angle reduction for excitation. The protocols used in this work (in-plane resolution 0.5 mm, scan duration = 3 min) did not show major artifacts and are therefore a good compromise for cartilage volumetry studies with mechanical loading on subjects and patients who might not be able to maintain the loaded position for a longer period of time. As demonstrated in Fig. 4, loading improves the visibility of the boundary between patellar and femoral cartilage. This phenomenon is probably due to load-induced flow between the cartilage surface and the synovial fluid during the experiment (2). Furthermore, it has been shown that loading leads to increased anisotropy in the surficial collagen fibre network, which in turn gives rise to a shortening of cartilage T2 (12).

The setup restrictions associated with optical tracking as described in this work could be avoided by navigator-based motion correction techniques, which have already been proposed for head imaging (32–36). However, such methods, which require the interleaved acquisition of navigator echoes, increase the scan time significantly. Alternatively, a self-navigated image-based technique such as Prospective Acquisition CorrEction (PACE) could be applied without a scan time penalty, but has a far lower update rate than the method described here and would probably be hampered by the more complicated geometry of the knee compared to the head (37).

With the proposed method, mechanical properties of the knee cartilage can be studied in vivo with meaningful in situ loading paradigms. A 3D gradient-echo sequence, as applied in this work, can be used for assessing cartilage deformation under loading. The results obtained in this proof-of-concept work suggest a decrease of cartilage thickness in the loaded compared to the unloaded setup. However, systematic assessment of cartilage deformation would require a more sophisticated loading jig, to enable control over the

loading force during the experiment (16–18), as well as automatic or semi-automatic 3D cartilage segmentation for volumetric measurements (38,39).

It should be noted that the motion correction methodology can also be readily implemented with any other sequence. Another promising application would be a turbo spin echo measurement for the investigation of T_2 relaxation changes due to altered water concentrations in the loaded cartilage, thus probing the integrity of the chondral proteoglycan-collagen matrix (1,3). A turbo spin-echo sequence has recently been validated with prospective motion correction for brain imaging (40). Combining static imaging of the loaded knee with kinematic measurements acquiring single images over a range of motion (16,30,41–43) may provide new insights into knee biomechanics, paving the way to better diagnosis and treatment of osteoarthritis.

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FIG. 1.

Experimental setup: MPT camera mounted in the scanner bore (a); MR-compatible loading jig with adjustable shoulder plates (b); setup for imaging of the loaded patellofemoral compartment (setup 1) with the tracking marker taped to the knee cap and a loop coil used for signal reception (c, d); setup for imaging of the loaded tibiofemoral compartment (setup 2) with the tracking marker taped to the shin and a birdcage coil used for signal reception (e, f).

FIG. 2.

Imaging setup for cartilage imaging of the patellofemoral joint (setup 1), using a 3D gradient echo sequence with the yellow box showing the field of view. In some experiments, the signal from the posterior part of the knee was suppressed with a coronal presaturation slab to avoid pulsation artifacts from the popliteal artery.

FIG. 3.

Transverse images acquired with setup 1. Measurements were performed without (a–d) and with (e–h) motion correction, without (a, b, e, f) and with (c, d, g, h) knee loading as well as without (a, c, e, g) and with (b, d, f, h) a presaturation slab.

FIG. 4.

High-resolution transverse images of the patellofemoral joint (resolution: 0.4 mm in-plane, 1 mm through-plane) acquired with setup 1 and with motion correction, once without and once with loading. The two images show the same slice of the knee cap and the patellar cartilage, while the femoral bone and the soft-tissue structures at the posterior part of the knee are slightly displaced due to load-induced muscle activation. A decrease of the patellar cartilage thickness can be observed in the lateral region (arrows).

FIG. 5. Sagittal image acquired with setup 2 using motion correction. The image quality is impaired by artifacts (dashed arrows) from non-rigid plantaris (solid arrows) motion.

FIG. 6. High-resolution sagittal images of the tibiofemoral joint acquired with setup 2. Measurements were performed without (a, b) and with (c, d) knee loading as well as without (a, c) and with (b, d) motion correction. The white arrows indicate the region of strongest cartilage compression where the distance between tibial and femoral bone was measured.

Table 1

Patellar cartilage thickness in mm at the lateral facet as measured from the high-resolution transverse images acquired without and with mechanical loading in six healthy subjects.

Subject	1	2	3	4	5	6
w/o loading	3.2 ± 0.1	4.9 ± 0.1	3.7 ± 0.1	4.1 ± 0.1	3.2 ± 0.1	3.7 ± 0.1
with loading	2.8 ± 0.1	4.2 ± 0.1	3.4 ± 0.1	3.8 ± 0.1	3.2 ± 0.1	3.3 ± 0.1