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A B_1 -Insensitive High Resolution 2D T_1 Mapping Pulse Sequence for dGEMRIC of the HIP at 3 Tesla

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Abstract

Early detection of cartilage degeneration in the hip may help prevent onset and progression of osteoarthritis in young patients with femoroacetabular impingement. Delayed gadolinium-enhanced MRI of cartilage is sensitive to cartilage glycosaminoglycan loss and could serve as a diagnostic tool for early cartilage degeneration. We propose a new high resolution 2D T_1 mapping saturation–recovery pulse sequence with fast spin echo readout for delayed gadolinium-enhanced magnetic resonance imaging of cartilage of the hip at 3 T. The proposed sequence was validated in a phantom and in 10 hips, using radial imaging planes, against a rigorous multipoint saturation–recovery pulse sequence with fast spin echo readout. T_1 measurements by the two pulse sequences were strongly correlated ($R^2 > 0.95$) and in excellent agreement (mean difference = -8.7 ms; upper and lower 95% limits of agreement = 64.5 and -81.9 ms, respectively). T_1 measurements were insensitive to B_{1+} variation as large as 20%, making the proposed T_1 mapping technique suitable for 3 T.

Keywords

femoroacetabular impingement; MRI; hip; cartilage; dGEMRIC; radial MRI; T_1 mapping; B_{1+} inhomogeneity

In recent years, it has been increasingly recognized that femoroacetabular impingement (FAI), a condition in which structural abnormalities of the femoral head–neck junction and/or acetabulum cause mechanical blockage in the terminal range of hip motion, can lead to osteoarthritis of the hip (1,2). In FAI, the abnormal contact between the acetabular rim and femoral neck causes chondral and labral damage, which can progress over time and result in osteoarthritis of the hip joint if the underlying cause of impingement is not addressed surgically (3).

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MR Imaging has emerged as the preferred diagnostic modality for suspected FAI due to its multiplanar image acquisition capability and its high soft tissue contrast. The acetabular cartilage's and labrum's position and orientation within the pelvis make MR imaging of these structures in three orthogonal planes susceptible to partial volume effects. One approach to minimize partial volume averaging is to image the acetabular rim and cartilage in a set of rotating radial planes (4). Imaging in rotating radial planes exploits the geometry of the hip joint and allows orthogonal display of the whole acetabular rim around its circumference. This imaging technique has been shown to be useful in identifying obliquely oriented tears in the anterosuperior and posterosuperior sections of the labrum (4).

Corrective surgical procedures aimed at removing the bony abnormalities of FAI and treating the associated labral and cartilage abnormalities are less likely to be successful in patients presenting with extensive articular cartilage injuries (5), for whom the only viable treatment is arthroplasty. Therefore, it is imperative to detect cartilage damage in its early stages. Cartilage that appears morphologically normal in routine MRI may already be irreversibly compromised in early osteoarthritis. MR-based biochemical imaging techniques, such as delayed gadolinium-enhanced MRI of cartilage (dGEMRIC) (6,7), have been proposed as an early diagnostic tool for the evaluation of chondral lesions. In dGEMRIC, negatively charged contrast agent (Gd-DTPA^{2-}) is administered prior to an exercise protocol, to exploit the different Gd-DTPA kinetics between the healthy and compromised cartilage, and imaging is performed to measure delayed contrast enhancement of compromised cartilage, which reflects the local concentration of glycosaminoglycans in an inverse relationship (6,7). The areas with depleted glycosaminoglycans will have higher concentrations of Gd-DTPA^{2-} , which will be reflected in the measured T_1 . Therefore, dGEMRIC provides an indirect visualization of glycosaminoglycans loss, which is an early sign of cartilage degeneration (8).

In recent years, a fast 2-angle T_1 mapping (F2T1) pulse sequence based on three-dimensional (3D) gradient echo readout has been introduced (9) and validated (10) for dGEMRIC in the hip. The F2T1 pulse sequence is more time-efficient than two-dimensional (2D) multipoint inversion recovery and saturation–recovery (SR) pulse sequences, which are problematic for clinical use due to their long acquisition times (8,11). The F2T1 sequence has been proposed to acquire dGEMRIC datasets covering the entire hip joint with isotropic spatial resolution, which can then be reformatted during postprocessing in rotating radial planes of the hip joint (12,13). These studies showed that dGEMRIC images reformatted during postprocessing in rotating radial planes depict cartilage damage in the anterior–superior region of the acetabulum (12), where cartilage injury typically occurs in FAI patients. These previously reported 3D dGEMRIC results were obtained at 1.5 T with approximately $0.80 \text{ mm} \times 0.80 \text{ mm} \times 0.80 \text{ mm}$ isotropic spatial resolution and acquisition times in the order of 9–10 min (12,13) or more, depending on the number of partitions needed to sample the whole 3D volume without aliasing artifacts. Given the small dimensions of hip acetabular cartilage (14), it may be necessary to further increase the spatial resolution and reduce the scan time to minimize the loss in spatial resolution due to patient motion. One approach to increase the spatial resolution and/or reduce the scan time is to perform 3D dGEMRIC at 3 T and trade increased signal-to-noise ratio (SNR) for higher resolution and/or faster imaging (e.g., higher acceleration), respectively, at the expense of

reduced accuracy due to increased B_{1+} variation within the hip at 3 T. The loss in accuracy can be partially compensated with a corresponding B_{1+} mapping method (15,16), where the resulting flip angle maps can be used to correct the T_1 map.

We propose to exploit increased SNR afforded by 3 T imaging by performing high spatial resolution 2D T_1 mapping in radial imaging planes to take advantage of the geometry of the hip joint (4,12). Therefore, the aim of this study was to develop a new B_1 -insensitive 2D T_1 mapping pulse sequence with high in-plane resolution for dGEMRIC in radial planes of the hip. Our approach was to image the hip using a fast spin echo (FSE) pulse sequence at 3 T to achieve high spatial resolution with adequate SNR and employ a B_1 -insensitive saturation pulse to perform uniform T_1 weighting. The scan time of the proposed pulse sequence is 1 min and 20 s per 2D slice. Compared with the previously reported 3D dGEM-RIC pulse sequence, the proposed pulse sequence is relatively less sensitive to patient motion. We validated the proposed pulse sequence against a rigorous multipoint SR pulse sequence at 3 T, by comparing measured T_1 in a phantom and in the hip cartilage of FAI patients. We also compared the accuracy and SNR efficiency of the proposed pulse sequence against the 3D F2T1 pulse sequence in phantom experiments.

MATERIALS AND METHODS

Pulse Sequence

We modified a FSE pulse sequence to perform two image acquisitions with two different T_1 weightings. The first FSE image acquisition was acquired after applying a saturation pulse with a SR time delay (TD) on the order of T_1 , to achieve a good balance between T_1 sensitivity and SNR. Based on previous dGEMRIC studies at 1.5 T (13) and 3 T (17), we anticipated T_1 of normal cartilage at 3 T to be on the order of 700–800 ms. As such, we used TD = 700 ms to achieve a good balance between T_1 sensitivity and SNR for the SR acquisition (18). In the proposed SR acquisition with TD = 700 ms, tissues with relatively short T_1 values (<350 ms), for which the recovery of the longitudinal magnetization is nearly complete, and tissues with long T_1 values (>2100 ms), for which the recovery of the magnetization is minimal, are especially susceptible to random error. The second FSE image [i.e., proton density (PD)] acquisition was performed with TR on the order of 5 T_1 s and without the saturation pulse. T_1 was calculated pixel-wise by dividing the SR image, I_{SR} , by the PD image, I_{PD} , to correct for the unknown equilibrium magnetization (M_0), and then solving the ideal SR experiment described by the Bloch equation governing T_1 relaxation:

$$\frac{I_{SR}=M_0(1 - e^{-TD/T_1})}{I_{PD}=M_0}=(1 - e^{-TD/T_1}), \quad [1]$$

$$T_1=\frac{-TD}{\log(1 - \frac{I_{SR}}{I_{PD}})} \quad [2]$$

The modified FSE pulse sequence was implemented on a whole-body 3 T MRI scanner (Verio, Siemens Healthcare, Erlangen, Germany) equipped with a gradient system capable of achieving a maximum gradient strength of 45 mT/m and a slew rate of 200 T/m/s. The radio-frequency (RF) excitation was performed using the transmit body coil, and a 32-element “cardiac” coil array (Invivo, Orlando, FL) was employed for signal reception. The relevant imaging parameters are: field of view = 190 mm × 190 mm, acquisition matrix = 320 × 320, in-plane resolution = 0.6 mm × 0.6 mm, slice thickness = 5 mm, turbo factor = 13, FSE readout duration was 143 ms, TE = 10 ms, refocusing flip angle was 180°, generalized autocalibrating partially parallel acquisitions (19) with an acceleration factor = 1.8 and receiver bandwidth = 161 Hz/pixel. A fat suppression pulse was used to avoid chemical shift artifacts at the bone–cartilage interface. TR (including the saturation pulse, recovery time, and FSE readout duration) was 850 ms and 4000 ms for SR and PD acquisitions, respectively. Total scan time for both SR and PD acquisitions was 1 min 20 s per slice.

We used a B_1 -insensitive saturation pulse (20) to achieve uniform T_1 weighting within the hip at 3 T. The hybrid adiabatic-rectangular pulse train is comprised of three nonselective RF pulses: nonselective rectangular 140° pulse, nonselective rectangular 90° pulse, and nonselective adiabatic half-passage pulse. The crusher gradients inserted between RF pulses are cycled to eliminate stimulated echoes. Spoiler gradients are applied before the first RF pulse and after the third RF pulse to dephase the transverse magnetization.

To validate the T_1 measurements calculated with Eq. 2, we acquired four additional SR images, with TD = 350, 1050, 1750, and 2450 ms (Fig. 1). Total scan time for the four additional SR images was 1 min 40 s per slice. These additional SR images were combined with the aforementioned SR image with TD = 700 ms and the PD image, to perform a two-parameter (M_0 , T_1) nonlinear fit of Eq. 1. All six images were acquired in series to minimize image registration errors. Total scan time to acquire all six images was 3 min per slice.

We compared the proposed 2D FSE pulse sequence against the 3D F2T1 pulse sequence in two phantom experiments. For the first phantom experiment designed to compare their sensitivity to B_{1+} variations, 2D T_1 mapping pulse sequence was performed with the aforementioned protocol, and 3D F2T1 imaging was performed with the following parameters: spatial resolution = 0.8 mm × 0.8 mm × 0.8 mm, flip angles = 5 and 30°, TE/TR = 3.5/20 ms, receiver bandwidth = 130 Hz/pixel, 144 partitions, 22% partition over sampling, 41% partition over sampling, generalized autocalibrating partially parallel acquisitions acceleration factor = 1.8, partial Fourier factor 6/8 in the phase-encoding direction, and scan time = 13 min 16 s. Prior to the 3D F2T1 sequence, a B_{1+} mapping prescan, based on a stimulated echo pulse sequence (15,16), was performed to correct the T_1 maps calculated from the 3D F2T1 images. The T_1 maps with B_{1+} correction were computed using the Siemens inline reconstruction algorithm on our 3 T scanner equipped with VB17 software platform. For the second phantom experiment, designed to compare their SNR efficiencies, both our proposed 2D T_1 mapping and 3D F2T1 mapping methods were performed with full k -space encoding (i.e., no generalized autocalibrating partially parallel acquisitions acceleration and no partial Fourier imaging), where the scan time was 2 min and 15 s and 31 min and 48 s, respectively, to calculate the SNR as the ratio of the mean signal and standard deviation of background noise (21).

Phantom Imaging

We imaged a spherical mineral oil phantom with a known T_1 (~550 ms) in the coronal plane to determine the sensitivity of the saturation pulse to clinically relevant B_{1+} variations within the hip at 3 T. To avoid signal saturation of the oil phantom, the phantom experiment was performed without the fat suppression pulse. Image acquisition was repeated with B_{1+} scale of the saturation pulse manually adjusted from 0.8 to 1.2 (0.1 steps) of its nominally calibrated B_{1+} value. Nominal B_{1+} was determined using the automated RF transmit calibration procedure. The upper limit of 20% B_{1+} variation was based on our preliminary experience with hip imaging at 3 T.

In the second experiment, the phantom was comprised of approximately 9% glycerol in distilled water to emulate relaxation times of hip cartilage (measured $T_1 = 730$ ms; measured $T_2 = 37$ ms). For the 3D data, SNR was measured only in a 2D plane that corresponds to the 2D FSE plane. To account for the difference in voxel sizes, we normalized the SNR by the voxel size. The normalized SNR efficiency was then calculated as the normalized SNR divided by the square root of the scan time.

Hip Imaging

Patients with hip pain and positive physical examination for FAI were imaged after a double dose (0.2 mmol/kg) intravenous injection of Gd-DTPA²⁻ (Magnevist®, Bayer Healthcare) and 15 min walking on a treadmill at controlled speed (9). The dGEMRIC pulse sequence was applied after the clinical protocol, ~45 min after administration of Gd-DTPA. Ten hips (6 left, 4 right) were scanned in nine consecutive patients (mean age = 36 ± 10 years). All images were acquired in a radial plane that included the anterior–superior region of the acetabulum. Human imaging was performed in accordance with protocols approved by the Human Investigation Committee at our institution; all subjects provided written informed consent.

Image Analysis

Image processing was performed using an in-house developed software. For each hip, the six images acquired at different time points (see Fig. 1) were spatially registered to the PD image to compensate for motion. In particular, we used affine transformation to register only a user-defined region of interest (ROI) comprising the entire hip joint.

After deidentification and randomization of the patient data, two observers (RL and DK) manually segmented a ROI over the weight-bearing portion of the hip articular cartilage (10), extending from the lateral bony edge, not including the labrum, to the edge of the acetabular fossa. For each ROI, the software calculated an analytically solved T_1 map based on the formula in Eq. 2 (using $TD = 700$ ms and PD). As a reference measurement, the software also calculated a two-parameter six-point fitted T_1 map based on Eq. 1, using all six images and a global optimization algorithm (22). Observer 1 (RL) repeated the image analysis after 14 days from the first analysis to assess intraobserver variability. Interobserver variability was assessed between observer 1 and observer 2 (DK), comparing the average T_1 value in the cartilage ROI for each hip. The two independent observers were blinded to patient identity and each other.

Statistical Analysis

For each ROI, the difference between the analytic T_1 and the six-point fit T_1 was calculated pixel-wise to display the spatial distribution of error for each analysis session. We performed the Pearson correlation and Bland-Altman (23) analyses using the mean T_1 value in each ROI.

Noise Analysis

To estimate the T_1 error, we performed a theoretical analysis using Eq. 1 for reference T_1 mapping (i.e., 6-point SR experiment) and Eq. 2 for analytic T_1 mapping, as a function of true T_1 ranging from 600 to 1200 ms (5 ms steps). The lower (normal—200 ms) and upper (normal + 400 ms) limits of the T_1 range were based on assuming normal cartilage T_1 equal to 800 ms (see results). To estimate clinically relevant white Gaussian noise, in a 27-year-old male volunteer, we acquired two PD image acquisitions in radial planes of the hip with full k -space encoding and $TR = 10$ s ($>5 T_1$). In addition, we acquired a noise map using the same pulse sequence without RF excitation. The hip articular cartilage was segmented manually, and the SNR was calculated as the ratio of the mean cartilage signal and standard deviation of noise derived from the noise map. The average of two PD SNR measurements was 127.5. Given that our proposed PD acquisition performs generalized autocalibrating partially parallel acquisitions acceleration 1.8, we anticipate a PD SNR of 95. Assuming $M_0 = PD$, clinically relevant white Gaussian noise was estimated as $0.0105 M_0 (= M_0/95)$. The theoretical noise analysis was repeatedly performed 100 times using a numerical phantom with 100 pixels to mimic the typical number of pixels in the segmented hip cartilage, where identical amount of white noise was added to the numerical PD and SR images. The influence of white noise on T_1 accuracy was estimated by performing linear regression analysis on the calculated and true T_1 values and calculating root-mean-square-error. Reported linear regression statistics and root-mean-square-error values represent the mean $\pm$ standard deviation over 100 measurements.

RESULTS

Figure 2 shows analytic and six-point T_1 maps of the phantom, as well as the percentage difference map. T_1 in the phantom was 562 ± 21 ms with the analytic method and 561 ± 15 ms with the six-point fit method, and RMS of percent difference was 2.8%, suggesting that they are quantitatively equivalent. T_1 measurements with the analytic method were 567, 565, 561, 561, and 563 ms for B_{1+} scales 0.8, 0.9, 1.0, 1.1, and 1.2, respectively. Consistent with the previous work in the heart at 3 T (20), the phantom T_1 values were similar throughout (less than 1% difference with respect to the average value), suggesting that the saturation pulse is insensitive to B_{1+} variation as large as 20%. In contrast, T_1 measurements using the 3D F2T1 pulse sequence with B_{1+} correction were 559, 574, 585, 612, and 630 ms for B_{1+} scales 0.8, 0.9, 1.0, 1.1, and 1.2, respectively, suggesting that even with B_{1+} correction the 3D F2T1 pulse sequence was sensitive to clinically relevant B_{1+} variation (Fig. 3).

For the glycerol phantom experiment, the normalized SNR efficiency was 10.3 and 4.3 for the 2D FSE and 3D F2T1, respectively. The higher SNR efficiency of 2D FSE over 3D F2T1

is largely due to the difference in flip angles (90–180° vs. 5–30°; 2D FSE vs. 3D F2T1, respectively).

Figure 4 shows, for one representative case, the six radial images acquired with different TDs. This image series exhibits consistently good image quality. For pixels within the ROIs, global optimization (22) using the six available values allowed an accurate fitting of the SR curve shown in Fig. 1 to calculate T_1 . Analytic and six-point fit T_1 maps are shown for one hip in Fig. 5, together with a map and a histogram of the percent difference between the two. The range of the color bars were chosen to span the distribution of values in the ROIs. In this particular hip, the pixel-wise percent difference between analytic and six-point fit T_1 ranged from –6.4 to 6.8%, and the RMS of percent difference was 3.2%.

The mean T_1 over 10 hips was 823 ± 189 ms, 808 ± 183 ms, and 797 ± 132 ms, for the two sessions of observer 1 and the single session of observer 2, respectively. The fact that mean T_1 of cartilage was on the order of 800 ms confirms our choice in TD of 700 ms. The top row of Figure 6 shows the correlation between analytic and six-point fit T_1 for the 10 hips, whereas the bottom row shows Bland-Altman plots that illustrate the agreement between the two T_1 measurements. The Person correlation coefficient of determination R^2 was larger than 0.95 in all cases ($P < 0.001$), suggesting that the two measurements were strongly correlated. According to the Bland-Altman analysis, analytic and six-point fit T_1 values were in good agreement (mean difference = –8.7 ms, i.e., ~1%; upper and lower 95% limits of agreement = 64.5 and –81.9 ms, respectively). Pearson and Bland-Altman statistics for observer 1, analysis 2, and observer 2 are shown in Table 1. As summarized in Table 1, the intra/interobserver variability in T_1 calculated from the same SR data with the analytic method was –10.4/11.9 ms, and the upper (mean plus 1.96 standard deviation) and lower (mean minus 1.96 standard deviation) 95% limits of agreement were 34.1/118.3 ms and –54.9/94.5 ms, respectively. Using the six-point fit, the intra/interobserver variability in T_1 was –14.8/11 ms, whereas the upper and lower 95% limits of agreement were 38.0/144.7 ms and –67.6/122.7 ms, respectively.

Figure 7 shows representative dGEMRIC T_1 maps of a 53-year-old male patient in six rotating radial planes of the hip joint. The total scan time to acquire all six T_1 maps was 8 min. Both raw SR and PD images exhibited good image quality, and these T_1 maps depict the hip cartilage with adequate spatial resolution.

For the theoretical noise analysis, root-mean-square-error values were 27.3 ± 1.6 and 20.3 ± 1.6 ms for the analytic and 6-point fit T_1 , respectively, compared with true T_1 ranging from 600 to 1200 ms. Linear regression statistics were comparable between the analytic and 6-point T_1 mapping methods (see Table 2).

DISCUSSION

This study describes a new 2D T_1 mapping pulse sequence for dGEMRIC in the hip joint with a clinically acceptable scan time of 1 min 20 s per slice. Compared with a rigorous six-point SR acquisition (3 min per slice), the proposed T_1 mapping acquisition produced

accurate results in vitro and in vivo, suggesting that the two acquisitions are quantitatively equivalent. The intra and interobserver agreements for T_1 calculations were good.

Conventional 2D T_1 mapping pulse sequences based on multipoint inversion recovery or SR with FSE readout (24–26) are clinically not feasible due to their long acquisition times. T_1 mapping pulse sequences based on gradient echo readout (8,27) are more efficient than FSE based pulse sequences, but they are inherently low in SNR and sensitive to B_{1+} inhomogeneities at 3 T. Our proposed 2D pulse sequence resulted in good image quality, because we used FSE readout at 3 T. Furthermore, it achieved uniform T_1 weighting through the use of a robust saturation pulse (20). This robust saturation pulse has been shown to effectively saturate the magnetization within the whole heart at 3 T (20). We anticipate B_{1+} variation is lower within the hip than within the heart. Our phantom experiments showed that, compared with 3D F2T1 pulse sequence, our proposed 2D T_1 mapping pulse sequence yields higher SNR efficiency and lower sensitivity to B_{1+} variations. We performed our phantom experiment assuming B_{1+} variation as large as 20%, based on our preliminary experience with hip imaging at 3 T. We demonstrated that our proposed T_1 mapping pulse sequence is insensitive to up to 20% B_{1+} variation.

We validated our pulse sequence against a rigorous T_1 mapping method based on a six-point SR acquisition. A potential issue with this acquisition approach in vivo is patient motion. While we used affine transformation to perform image registration of the entire hip joint, there was small residual motion between images which could have affected T_1 calculation for some of the pixels. We anticipate the motion to be less of an issue for the two-point SR acquisition of 1 min and 20 s than the full six-point SR acquisition of 3 min. One approach to further minimize the registration error is to perform interleaved acquisition between SR and PD. Another potential approach is to develop more accurate retrospective image registration methods. A limitation with our pulse sequence is that it may be sensitive to partial volume averaging along the through-plane direction, even with the preferred imaging in radial planes of the hip joint. Another limitation is that it may suffer minor loss in in-plane spatial resolution due to apodization in k -space with echo-train-length of 13.

The mean T_1 of cartilage was on the order of 800 ms. As such, our choice TD = 700 ms for the SR image acquisition represents a good balance between T_1 sensitivity and SNR, and TR = 4000 ms for the PD image acquisition was sufficient. For imaging tissues with different T_1 , both TD for SR and TR for PD acquisitions need to be adjusted.

In conclusion, we developed a new high-resolution 2D T_1 mapping sequence, suitable for dGEMRIC in radial planes of the hip at 3 T. We showed that the T_1 measurements are accurate, repeatable, and reproducible. The technique could be applied to measure cartilage T_1 in other joints, and it is particularly suitable for applications at 3 T, because it is insensitive to B_{1+} inhomogeneities.

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FIG. 1.

Plot of SR acquisition used in this study. The five SR acquisitions were acquired with TDs 350, 700, 1050, 1750, and 2450 ms. The PD acquisition was acquired with TR = 4000 ms and without the saturation pulse. The analytical T1 measurement was made using the SR image with TD = 700 ms and PD image (i.e., Eq. 1). The two-parameter fit of the ideal SR equation was made using all six images. Note that all six images were acquired in series, to minimize image registration errors.

FIG. 2.

T_1 maps were calculated using the analytic and the six-point fit method for a spherical mineral oil phantom with a known T_1 (~550 ms). The phantom was imaged on a coronal plane without the fat suppression pulse. The difference between the two T_1 maps was calculated pixel-wise for the entire phantom.

FIG. 3.

T_1 measurements as a function of B_{1+} scale ranging from 0.8 to 1.2 (0.1 steps). The 3D F2T1 pulse sequence was sensitive to B_{1+} scale ranging from 0.8 to 1.2, whereas our proposed 2D T_1 mapping pulse sequence was insensitive to the same B_{1+} scale range.

FIG. 4.

Six radial images acquired with different TD for one of the hip used in the study. T_1 was calculated rigorously by fitting the SR curve with the signals of the six images. T_1 was also calculated with the analytic formula in Eq. 1 using the second and last image. TD values were chosen assuming T_1 in the order of 700–800 ms in healthy hip cartilage at 3 T, so that the image at TD $\frac{1}{4}$ 4 s corresponds to proton density.

FIG. 5.

For all cases, the weight-bearing portion of hip cartilage was segmented from the lateral bony edge to the edge of the acetabular fossa. T_1 maps were calculated using the analytic and the 6-point fit method for each ROI (top row) and the percent difference between the two was computed pixel-wise (bottom row). The RMS of percent difference was 3.2% for the hip in this figure.

FIG. 6.

Average T_1 of the hip cartilage calculated with the analytic T_1 mapping sequence vs. rigorous 6-point fitting at 3 T. The top row shows correlation plots and the bottom row shows Bland-Altman plots (95% confidence intervals are displayed) of the T_1 values calculated with the two methods. The three columns correspond to the two test sessions of observer 1 and the single session of observer 2.

FIG. 7.
Representative dGEMRIC T_1 maps a 53-year-old male patient in six rotating radial planes of the hip joint at 3 T. These T_1 maps depict the hip cartilage with adequate spatial resolution.

Table 1

Summary of Bland–Altman and Pearson Analysis

Agreement type	Difference (ms)	Upper 95% limit (ms)	Lower 95% limit (ms)	R^2 (Pearson)
Analytic—6-point fit (observer 1, session 1)	−8.7	64.5	−81.9	0.98
Analytic—6-point fit (observer 1, session 2)	−4.3	71.5	−80.1	0.97
Analytic—6-point fit (observer 2)	18.6	77.2	−40.0	0.95
Intra-observer (session 2—session 1: analytic in gray, 6-point fit in white)	−10.4	34.1	−54.9	0.98
	−14.8	38.0	−67.6	0.98
Inter-observer (observer 2—observer 1: analytic in gray, 6-point fit in white)	11.9	118.3	−94.5	0.88
	11	144.7	−122.7	0.92

Table 2

Linear Regression Statistics for the Theoretical Noise Analysis, Using Eq. 1 for Reference T_1 Mapping (i.e., 6-Point SR Experiment) and Eq. 2 for Analytic T_1 Mapping, as a Function of True T_1 Ranging from 600 to 1200 ms (5 ms Steps)

Measurement pair	Slope	Bias (ms)	R^2	RMSE (ms)
Analytic T_1 vs. true T_1	1.01 ± 0.01	9.60 ± 9.08	0.99 ± 0.00	27.3 ± 1.6
6-Point fit T_1 vs. true T_1	1.00 ± 0.01	0.35 ± 9.26	0.99 ± 0.00	20.3 ± 1.6

Assuming M_0 = PD, Clinically relevant white Gaussian noise was estimated as $0.0105 M_0$ ($= M_0/95$). The theoretical noise analysis was repeatedly performed 100 times using a numerical phantom with 100 pixels to mimic the typical number of pixels in the segmented hip cartilage, where identical amount of white noise was added to the numerical PD and SR images. RMSE values are 27.3 ± 1.6 and 20.3 ± 1.6 ms for the analytic and 6-point fit T_1 , respectively, compared with True T_1 ranging from 600 to 1200 ms.