Published in final edited form as:

NMR Biomed. 2016 April ; 29(4): 420–430. doi:10.1002/nbm.3474.

Capturing Fast Relaxing Spins with SWIFT Adiabatic Rotating Frame Spin-lattice Relaxation ($T_{1\rho}$) Mapping

Jinjin Zhang¹, Mikko Nissi^{1,2,3,4}, Djaudat Idiyatullin¹, Shalom Michaeli¹, Michael Garwood¹, and Jutta Ellermann¹

¹Center for Magnetic Resonance Research and Department of Radiology, University of Minnesota, Minneapolis, MN, United States ²Department of Applied Physics, University of Eastern Finland, Kuopio, Finland ³Research Unit of Medical Imaging, Physics and Technology, University of Oulu, Oulu, Finland ⁴Diagnostic Imaging Center, Kuopio University Hospital, Kuopio, Finland

Abstract

Rotating frame spin-lattice relaxation, with characteristic time constant $T_{1\rho}$, provides a means to access motion-restricted (slow) spin dynamics in MRI. Due to their restricted motion, these spins are sometimes characterized by a short transverse relaxation time constant T_2 and thus can be difficult to detect directly with conventional image acquisition techniques. Here we introduce an approach for three-dimensional adiabatic $T_{1\rho}$ mapping based on a magnetization prepared Sweep Imaging with Fourier Transformation (MP-SWIFT) sequence which captures signal from almost all water spin populations including the extremely fast relaxing pool. A semi-analytical procedure for $T_{1\rho}$ mapping is described. Experiments on phantoms and musculoskeletal tissue specimens (tendon, articular and epiphyseal cartilages) were performed at 9.4 Tesla for both the MP-SWIFT and fast spin echo (FSE) read outs. In the phantom with liquids having fast molecular tumbling and single-valued $T_{1\rho}$ time constant, the measured $T_{1\rho}$ values obtained with MP-SWIFT and FSE are similar. Conversely, in normal musculoskeletal tissues, $T_{1\rho}$ values measured with MP-SWIFT are much shorter relative to values obtained with FSE. Studies of biological tissue specimens demonstrated that $T_{1\rho}$ -weighted SWIFT provides higher contrast between normal and diseased tissues as compared to conventional acquisitions. Adiabatic $T_{1\rho}$ mapping with SWIFT readout captures contributions from the otherwise undetected fast relaxing spins allowing more informative $T_{1\rho}$ measurements of normal and diseased states.

Keywords

rotating frame relaxation; bound water; ultrashort T2; SWIFT; relaxometry; Musculoskeletal

Introduction

The time constant $T_{1\rho}$ characterizes spin-lattice relaxation in the rotating frame (1). $T_{1\rho}$ is sensitive to the interactions between motion-restricted water molecules and the corresponding local macromolecular environment (2,3). With $T_{1\rho}$, the slow motional regime can be accessed at high magnetic field (2,3). Previous studies have shown that $T_{1\rho}$ is sensitive to structural alterations in musculoskeletal tissues, often providing more information than might be obtained from analysis of conventional longitudinal or transverse relaxations (2,4-7). Specifically, $T_{1\rho}$ relaxation has been shown to be sensitive to the proteoglycan (PG) loss in the early stage of articular cartilage degeneration (8-14) and in cases of vertebral disc degeneration (15), and has been found utility in the study of epiphyseal cartilage of the developing skeleton (16).

$T_{1\rho}$ is traditionally measured by applying a continuous wave (CW) pulse which “locks” the magnetization to the effective field direction (17). The CW spin-lock (CW-SL) pulse creating an effective field (in the kHz range) in the ‘y’ direction (in the first rotating frame) is usually embedded between 90°_x and 90°_{-x} rotations that are accomplished with short hard pulses or adiabatic half passage pulses (18). Alternatively, relaxation in the rotating frame can be studied using a time-dependent spin-locking field, for example, created by a train of frequency-modulated adiabatic full-passage (AFP) pulses (19-22). This technique, entitled “adiabatic $T_{1\rho}$ ”, was recently shown to provide sensitive detection of structural changes in articular cartilage (13,14,23). Adiabatic $T_{1\rho}$ characterizes relaxation during RF irradiation when the magnetization vector is aligned with the direction of the time-dependent effective field ($B_{\text{eff}}(t)$) (Fig.1). As compared with the conventional CW-SL method, the adiabatic $T_{1\rho}$ method is less affected by inhomogeneity of B_1 and static field B_0 due to the offset independent adiabaticity property of the AFP pulses (24). The adiabatic relaxation methods are inherently sensitive to a broader range of molecular motions as compared to the CW method (19). Further, it has been recently demonstrated that adiabatic $T_{1\rho}$ reduces magic angle dependence in relaxation mapping (25). All these features are critical for tissue characterization and especially during *in vivo* study when homogeneity of the B_1 and B_0 fields could be altered by patient motion (23).

In real tissues, multiple spin populations which are characterized by short and long relaxation time constants have been shown to exist. Different T_2 or $T_{1\rho}$ relaxation times were found in tendon or cartilage tissues by using rapid imaging sequence or spectroscopy methods (26-28). Water bound tight or closely associated to macromolecules tends to have short T_2 and $T_{1\rho}$ values, thus has conventionally been undetected due to very short transverse relaxation time (26,27). Characterization of these pools and the correlation of MR properties with tissue microstructure are critically important (29). On the other hand, conventionally, $T_{1\rho}$ relaxation measurements are performed by interleaving the readout of an imaging sequence with magnetization preparation (MP) blocks used for generating $T_{1\rho}$ weighting. In tissues having fast relaxing components in sub-millisecond range, e.g. in highly ordered tissues or highly mineralized tissues, the measurement of $T_{1\rho}$ using conventional sequences as read out is problematic due to considerable signal decay during the echo time (TE) before the detection (30).

In this paper, we introduce a three-dimensional (3D) adiabatic $T_{1\rho}$ relaxation time mapping utilizing an ultrashort T_2 -sensitive sequence for readout, in addition to the existing 3D $T_{1\rho}$ relaxation time mapping techniques (31-34). Multiple MRI pulse sequences capable of preserving signal from fast relaxing spins have been developed in recent decades (35-40). In general all these developed techniques are applicable to use as readout for the proposed $T_{1\rho}$ relaxation mapping with preserved short T_2 components. In this paper, the Sweep Imaging with Fourier Transformation (SWIFT) sequence was used as the readout sequence (39,41). SWIFT is an emerging technique that allows imaging of almost all biological tissues including tissues with ultra-short T_2 relaxation times in the microsecond time scale (39,41,42). With appropriate parameters, SWIFT provides nearly pure proton density weighted imaging, and by utilizing magnetization preparation embedded within this sequence, different types of contrasts (T_1 , T_2 , $T_{1\rho}$, $T_{2\rho}$ etc.) can be generated (42-45). Here, we describe the experimental setup and semi-analytical procedure to generate and quantify the $T_{1\rho}$ relaxation with SWIFT readout, and demonstrate application of $T_{1\rho}$ for quantification of osteochondral and tendon specimens.

Methods

Pulse sequence

The pulse sequence diagram is shown in Fig.1b. MP blocks are embedded after each m number of SWIFT readout periods (m TR periods). Each MP block is composed of four AFP pulses of the hyperbolic secant HS n family (here n is the stretching factor) (46), each having a duration of t_{180} with phases prescribed according to the MLEV4 phase cycling scheme (47). The minimal duration for one MP block is defined as $T_{mp} = 4t_{180}$. The variable duration of the MP train is equal then to $T_{mp} = T_{mp} \cdot N_{mp}$, with N_{mp} as the number of MP blocks.

Steady state signal equation

In regular SWIFT without any magnetization preparation, the repeated excitation drives the spin system to the steady state with equilibrium magnetization:

$$A_{ss} = M_0(1 - E_1)/(1 - E_1 \cos \theta), \quad [1]$$

where $E_1 = e^{-\frac{TR}{T_1}}$ and θ is the excitation flip angle (39,48). With the preparation block, the steady-state magnetization generated during MP-SWIFT is different and depends on T_{mp} . Simulations based on Bloch equations show the evolutions of M_z to different steady states for different T_{mp} settings (Fig. 2). Of relevance, as shown in Fig. 2, two competitive processes, i.e., signal decay with time constant $T_{1\rho}$ and signal recovery with T_1 , cause the steady-state signal to oscillate about an average value. This oscillating steady state is used for the reconstruction of 3D images. The average steady-state signal intensity in this case is given by (see Appendix A):

$$S(T_{\text{mp}}) = \frac{A_{\text{ss}}(1-B)}{2} \frac{1+E_{\rho}}{1-BE_{\rho}}, \quad [2]$$

where $E_{\rho} = e^{-\frac{T_{\text{mp}}}{T_{1\rho}}}$ and $B = E_1^m \cos^m \theta$. After normalization by $\frac{(1-B)}{2} S(T_{\text{mp}}=0)$, the Eq. 2 becomes

$$S(T_{\text{mp}}) = \frac{1+E_{\rho}}{1-BE_{\rho}}, \quad [3]$$

By separating out the exponential decay E_{ρ} and taking the logarithm:

$$\ln S_e(T_{\text{mp}}) = \ln E_{\rho} = \ln \frac{S(T_{\text{mp}}) - 1}{S(T_{\text{mp}})B + 1} = -\frac{T_{\text{mp}}}{T_{1\rho}}. \quad [4]$$

S_e is the modified signal, which is a mono-exponential function in this case. On the

logarithmic scale, S_e is represented by a straight line with slope equal to $-\frac{1}{T_{1\rho}}$. In practice, $T_{1\rho}$ can be estimated from the set of data measured with different T_{mp} . Obviously, the signal also depends on T_1 relaxation, and thus an independent acquisition for measuring T_1 is required. Here, the variable flip angle method with SWIFT readout was used to acquire T_1 maps (49).

Two water pools: the fitting strategy

In the previous section the description of the single pool spin system was presented. In the case of real tissues, multiple pools of water populations should be considered (27,50). The most practical model is a two pool system, characterized by two different proton populations: free water and bound water (27), having “slow relaxation” with long $T_{1\rho}(T_{1\rho}^s)$ and “fast relaxation” with short $T_{1\rho}(T_{1\rho}^f)$, respectively. The signal having contributions from both slow and fast relaxing components can be described as:

$$S(T_{\text{mp}}) = S_s(T_{\text{mp}}) + S_f(T_{\text{mp}}) = C_s \frac{1+E_{\rho s}}{1-BE_{\rho s}} + C_f \frac{1+E_{\rho f}}{1-BE_{\rho f}}, \quad [5]$$

where $E_{\rho s} = e^{-\frac{T_{\text{mp}}}{T_{1\rho}^s}}$ and $E_{\rho f} = e^{-\frac{T_{\text{mp}}}{T_{1\rho}^f}}$, and C_s and C_f are normalized populations of the two components separately ($C_s + C_f = 1$). The average relaxation time constant ($T_{1\rho}^{\text{ave}}$) is the slope of $\ln(S_e)$ at $T_{\text{mp}} \rightarrow 0$:

$$\frac{1}{T_{1\rho}^{\text{ave}}} = \frac{C_s}{T_{1\rho}^s} + \frac{C_f}{T_{1\rho}^f}, \quad [6]$$

In principle, the relaxation parameters can be estimated by fitting the experimental data using Eq. 5. The quality of fitting, however, critically depends on the relative difference between intrinsic relaxation parameters of the two pools and also on the sufficiency of sampling points of the relaxation decay curves. Note that the relaxation time constants of the two components should be sufficiently different in order to be separable with a given noise level (51). Even if these two components are very different, an adequate sampling of the short component could be problematic because of the limited duration of the AFP pulses which require sufficient peak power to satisfy adiabaticity. Thus, the reliability of experimental mapping of the exact $T_{1\rho}^s$, $T_{1\rho}^f$ or even $T_{1\rho}^{\text{ave}}$ values is questionable. Therefore, we propose to use two values, $T_{1\rho}^a$ and $T_{1\rho}^s$, representing the asymptotic average relaxation time and the slow relaxation time, respectively. The $T_{1\rho}^a$ value can be determined from the slope of the line connecting the first two data points on the logarithmic scale (red solid line in Fig. 3a). The value of $T_{1\rho}^a$ approaches $T_{1\rho}^{\text{ave}}$ (empty circles in Fig. 3b) when $T_{\text{mp}} \rightarrow 0$, as shown in Fig. 3b. With increasing T_{mp} , the contribution of the fast relaxing component vanishes, which allows the long $T_{1\rho}$ component to be estimated from the tail of the relaxation decay curve (see Appendix B for details). A representative example of the fitting is shown in Fig. 3a.

Experimental details

Phantoms—Test experiments were performed on three separate phantoms containing 0.1, 0.1, and 0.4 mM MnCl_2 in DI water, agar, and DI water, respectively. These solutions were poured in glass tubes immersed in a bigger tube containing saline solution. The T_2 values of these three phantoms were all above 20 ms.

Tissue specimens—Two specimens were studied: a piece of mature bovine Achilles tendon (size = $1 \times 1 \times 1 \text{ cm}^3$), and a juvenile femoral condyle from a 12 week-old pig (size = $3 \times 3 \times 3 \text{ cm}^3$) including both articular and epiphyseal cartilage layers. The animal specimens were placed in tubes and immersed in Fluorinert (3M, Minnesota, USA) to match the magnetic susceptibility.

All images were acquired on a 9.4 T-31 cm bore horizontal animal MRI scanner (Agilent Technologies, Santa Clara, CA USA), using a volume transmit/receive coil (Agilent Technologies) or a homemade coil with three parallel loops of a 22 mm diameter. All simulations and data processing were performed using MATLAB (R2012b, The Mathworks). For $T_{1\rho}$ measurement, variable numbers of MP blocks were added after every 16 SWIFT readout periods ($m = 16$ in Eq. 2) to obtain images with different T_{mp} . The experimental parameters are summarized in Table 1. Different durations of the AFP pulses (t_{180}) were utilized in order to test the effect of different weighting of fast and slow relaxing components. The peak RF power was matched to ensure adiabatic condition of the AFP pulses, and the time-bandwidth product of the pulse (R) was adjusted to keep excitation bandwidth of the pulse (BW_{mp}) the same for different pulse durations (24). Thus, the $B_{\text{eff}}(t)$ functions were the same for the different t_{180} values (20). The specific values of the maximal effective frequency and the averaged effective frequency are shown in Table 1. The time to acquire 128,000 radial spokes with each setting of T_{mp} varied from 5 to 11

minutes, depending on T_{mp} . The spokes in k-space were isotropically distributed on a sphere using the pseudo-random Halton approach to minimize signal oscillation effects on the steady state (52). All MP-SWIFT $T_{1\rho}^a$ and $T_{1\rho}^s$ maps were generated using linear least squares fitting on a pixel-by-pixel basis in the logarithmic coordinate of the transformed signal (S_e) according to the developed algorithm. For T_1 measurement, images with flip angles of 1, 3, 5, 7, 10 degrees were acquired, then fitted to the Ernst equation using a non-linear least square algorithm to generate the corresponding T_1 map (49). BW = 125 kHz and TR = 2.6 ms were used for all SWIFT acquisitions.

For comparison, the $T_{1\rho}$ relaxation time constants were also measured with a single-slice 2D fast spin echo (FSE) sequence used as readout following a number of MP blocks. The preparation pulse parameters were matched to the MP-SWIFT (i.e., the same AFP pulses, duration and RF power). With FSE, TR = 5000 ms, effective TE = 5.5 ms, ETL = 4, thickness = 1 mm, matrix size = 256×256 and bandwidth = 130 kHz were used. The choice of 2D FSE which was used here for comparison with MP-SWIFT is justified because it is a commonly used as readout for $T_{1\rho}$ measurements and has sufficient SNR to guarantee good accuracy. The long TR in FSE destroys the steady-state condition and therefore signal decays exponentially during the length of T_{mp} , and thus, $T_{1\rho}$ maps were generated using linear least squares fitting on a pixel-by-pixel basis to the exponential function:

$$S = ke^{-T_{mp}/T_{1\rho}} \quad [7]$$

For comparison with FSE, SWIFT images were averaged to the same thickness (= 1 mm) as the FSE images.

To assess the reproducibility of the proposed technique, a distal humeral trochlea of a 10-week old pig (size = $3 \times 3 \times 3 \text{ cm}^3$) was scanned 3 times during the same day utilizing the same parameter set as was used for the porcine femoral condyle (Table 1).

Results

The signal-to-noise ratio (SNR) of the SWIFT image in all experiments with $N_{mp} = 0$ was higher than 60. The SNR decreases as T_{mp} increases, but was greater than 20 for all T_{mp} . SNR was calculated by taking the ratio of the mean of the signal intensities in a given ROI and the standard deviation of the signal of a ROI from the background.

In Fig.4 the $T_{1\rho}$ relaxation time constants of the MnCl_2 phantoms (with $T_2 > 20\text{ms}$) measured by FSE and SWIFT sequences are shown. The $T_{1\rho}$ of the 0.1 mM of MnCl_2 solution was approximately 320 ms. The addition of agar to the 0.1 mM MnCl_2 solution changed the $T_{1\rho}$ time constant to approximately 200 ms as measured by either sequence. The $T_{1\rho}$ values of the phantom with higher concentration of MnCl_2 at 0.4 mM were around 85 ms in both FSE and SWIFT measurements (Fig. 4). The $T_{1\rho}$ of saline was around 2.6 s with both SWIFT and FSE measurements, as expected for the fast tumbling spins. This also further verified that the signal equation for the single component is accurately described by Eq. 2.

The $T_{1\rho}$ decay curves of a bovine tendon specimen obtained with the MP-SWIFT sequence at different t_{180} values are shown in Fig. 5. Because tendon is a highly ordered tissue and characterized by short T_2 , it is practically undetectable with conventional MRI sequences. Therefore, only measurements with SWIFT are presented. From the data points shown in Fig. 5, it is apparent that the S_e decay has multiple components with different decay rates. Since the effective fields produced with $t_{180} = 0.3$ and 1.2 ms were kept the same, some data points are overlapped as expected. In the case of $t_{180} = 2.0$ ms, the fitted $T_{1\rho}^a$ is equal to 16.8 ms, whereas for t_{180} equal to 0.3 ms, $T_{1\rho}^a$ decreases to 6.0 ms. These results suggest that reducing t_{180} increases weights of fast relaxing spins. Fittings of $T_{1\rho}^s$ with $T_{mp} > 5$ ms demonstrate the method of extraction of information about slowly relaxing spins. In both $t_{180} = 0.3$ and 1.2 ms cases, the fitted $T_{1\rho}^s$ value and the population are similar.

$T_{1\rho}$ maps and decay plots of a juvenile porcine femoral condyle are shown in Fig. 6. The juvenile femoral condyle reveals the homogenous appearing epiphyseal cartilage (z2 in Fig. 6a), in addition to the articular cartilage in the most superficial layer of the tissue (z1 in Fig. 6a). Longer $T_{1\rho}$ in the articular cartilage compared to the epiphyseal cartilage is observed. The zonal organization is similar in both SWIFT $T_{1\rho}^a$ (Fig. 6a) and FSE $T_{1\rho}$ maps (Fig. 6b), however with different absolute values. A focal lesion due to vascular failure within the epiphyseal cartilage is easily identifiable (black arrows, Fig. 6a). The focal lesion near the interface of cartilage and bone reveals higher $T_{1\rho}$ values in all $T_{1\rho}$ maps. The corresponding hematoxylin and eosin (H&E) stain histology shows the necrotic cartilage matching the focal lesion (Fig. 6c). The $T_{1\rho}$ decays on the logarithmic scale of the 2D FSE and 3D SWIFT $T_{1\rho}$ measurements are shown in Fig. 6d and Fig. 6e, respectively. For normal tissues, the signal obtained with FSE readout (blue line in Fig. 6d) is apparently single component decay, while the signal from SWIFT readout (blue line in Fig. 6e) reveals multiple decay rates. Interestingly, the lesion shows multiple components even in the FSE measurement (red data points in Fig. 6d). The fitted $T_{1\rho}^a$ and $T_{1\rho}^s$ values from the ROI in the lesion are very similar in both SWIFT (Fig. 6e) and FSE (Fig. 6d) measurements (red points and lines). This implies the absence of fast relaxing components within the lesion. SWIFT $T_{1\rho}^s$ maps obtained using the strategy described above are shown in Fig. 6f. Notably, the SWIFT $T_{1\rho}^s$ map is very similar to the FSE $T_{1\rho}$ map in Fig. 6b. The apparent population map of the fast relaxing spins, which was generated from C_s in the SWIFT measurement, is shown in Fig. 6g. The overall fraction of the fast relaxing component is around 50% in epiphyseal cartilage and 30% in articular cartilage. The fraction of the fast relaxing component in the area of the lesion is close to 2%. The T_1 value measured in the lesion area was ~ 1600 ms, which is longer than the T_1 of normal surrounding tissues (1100 ms). The contrast between normal tissue and the lesion is shown in Fig. 6h. The maximal relative relaxation time's difference (RRTD) between normal and lesion tissues $RRTD = [T_{1\rho}(\text{lesion}) - T_{1\rho}(\text{normal})] / T_{1\rho}(\text{normal})$ from SWIFT $T_{1\rho}^a$ map is about 6 times higher than that obtained with FSE readout.

Since an independent acquisition aiming at measuring T_1 is needed for determination of $T_{1\rho}$ values, the dependence of the deviation in $T_{1\rho}$ estimation (defined a

$D_{T_{1\rho}} = \frac{(T_{1\rho}^{\text{fitted}} - T_{1\rho}^{\text{real}})}{T_{1\rho}^{\text{real}}}$) on the deviation of T_1 ($D_{T_1} = \frac{(T_1^{\text{fitted}} - T_1^{\text{real}})}{T_1^{\text{real}}}$) is analyzed and shown in Fig. 7a. As can be seen, $D_{T_{1\rho}}$ does not have an apparent dependency on the $T_{1\rho}$ values, and is roughly linear with the D_{T_1} . From the theory section and Appendix B, the obtained $T_{1\rho}^s$ and C_f will deviate from the true values if the approximation $T_{1\rho}^s \gg T_{1\rho}^f$ is not fully satisfied. Figure 7b-d shows examples of the estimated deviation of $T_{1\rho}^s$, C_s and C_f

originating from the fitting procedure. Within the range of $\frac{T_{1\rho}^s}{T_{1\rho}^f}$ ratio from 10 to 30, the error from the fitting model is less than 15% in most cases. The fitting error source from SNR were also analyzed by Monte-Carlo simulation and showed in Fig. 7e & f.

For the reproducibility experiment, the measured relaxation time constants and population from the three separate scans with SWIFT readout on the same specimen in the epiphyseal cartilage layer from the same ROI were [average $\pm$ standard deviation (coefficient of variation) over the three scans]: $T_{1\rho}^a = 65.1 \pm 2.7$ ms (4.14%), $T_{1\rho}^l = 271 \pm 8.0$ ms (2.95%), $C_f = 0.59 \pm 0.03$ (5.08%).

Discussion

In this work measurement of adiabatic $T_{1\rho}$ in 3D using the SWIFT sequence was proposed and analyzed. Sensitivity of the sequence to different spin populations, including both fast and slow relaxing spins, was shown and the application to the musculoskeletal system was exemplified.

In biological tissues, several types of motional processes contribute to the relaxation mechanisms. Depending on the tissue, more than one relaxation mechanism may be operative simultaneously, but with different relative contributions. Generally, the more rapidly relaxing component is assigned to water that is more tightly bound to macromolecules (such as proteoglycans and collagens in cartilage), and the slow relaxing component is assigned to bulk water.

In the phantom study of single component relaxation with comparatively long T_2 , FSE- $T_{1\rho}$ matches well with the SWIFT- $T_{1\rho}$ in phantoms with different $T_{1\rho}$ values (Fig.4). For the juvenile porcine femoral condyle (Fig. 6), the measured FSE- $T_{1\rho}$ in the articular and epiphyseal cartilage is around 300 and 200 ms, respectively. This is similar to the values previously reported by Toth etc. on juvenile goat specimens using FSE readout (16), but higher than the adiabatic $T_{1\rho}$ values measured in mature specimens with FSE readout, which fall in 100-200 ms range (13,14,23). Note besides specimens, the specific parameters used in these studies were also not the same. While only one relaxing component was found in these previous studies with FSE readout, here by using SWIFT, apparent multi-component decay rates are observed in biological specimens (Fig.5 & 6). The measured $T_{1\rho}^a$ values using SWIFT were all much shorter than the $T_{1\rho}$ measured with FSE (Fig.6). In Fig.6, the $T_{1\rho}$ values of a focal lesion in a juvenile porcine femoral condyle specimen are much longer than those in the normal surrounding tissues in both FSE and SWIFT measurements. $T_{1\rho}$ within the lesion is much closer in SWIFT and FSE mapping comparing to normal tissues, which

implies the lack of fast relaxing spins within the area of concern. It is consistent with the histological finding of chondronecrosis, a disease process characterized by the degradation of the normal, tightly bound collagen and proteoglycan structures. This is also in agreement with the population map of the short component shown in Fig.6g, where in the lesion region, the lowest population of fast relaxing component is close to 2%.

From the apparent population map of the fast relaxing spins shown in Fig.6g, the overall fraction of the fast relaxing component was around 30% in the articular cartilage. The fast relaxing component originated from water with highly restricted motion, which is more tightly bound to proteoglycan and collagen in cartilage. Prior reports estimated the population of water pool bound to proteoglycan in cartilage at about 15-30% (26,27,50,53), and at 3-6% for the pool of water bound to collagen (50). Thus the total population of fast relaxing spins is measured in same range as our result. In contrast, the populations of the fast relaxing component in epiphyseal cartilage in the developing skeleton from our experiments are around 50% which is a considerably higher value relative to that obtained in articular cartilage. This is expected due to difference in matrix composition between epiphyseal and articular cartilage, for example, the greater concentration of glycosaminoglycan in epiphyseal cartilage (54). A full explanation will require a separate analysis which is beyond the scope of the present work.

The fact that the abnormal tissue within the lesion is dominated by slowly relaxing spins, while the surrounding tissue remains normal with multiple pools of water, reveals the different range of sensitivity of $T_{1\rho}$ values obtained with SWIFT and FSE readouts. The fitting for $T_{1\rho}^s$ of the slow relaxing components showed in Appendix B was mainly for the purpose of providing proof of principle. In practice, the map of $T_{1\rho}^a$ would be more informative and useful. As illustrated in the theory section, $T_{1\rho}^a$ is a parameter approaching to $T_{1\rho}^{ave}$ which is the weighted harmonic mean of the long and short $T_{1\rho}$ values. The slope for first two data points was calculated as the asymptotic value. $T_{1\rho}^a$ contains weighting from $T_{1\rho}^s$, $T_{1\rho}^l$, C_s , C_l and. From Fig.6, the SWIFT $T_{1\rho}^a$ map not only showed the similar multilayer features as the FSE $T_{1\rho}$ map, but also provided much higher relative contrast between normal and lesion tissues as compared to the $T_{1\rho}$ maps obtained with FSE readout. Therefore, with the capability of measuring the full dynamic range of the entire spin population in biological tissues, SWIFT $T_{1\rho}^a$ map can maximize contrast for detecting pathology.

The fitting of $T_{1\rho}^a$ only uses the first two data points. The first points are crucial since they include contribution from fast relaxing spins. In the adiabatic $T_{1\rho}$, the first point is obtained without any preparation pulse which can preserve maximal fast relaxing signal. In the case of the CW-SL $T_{1\rho}$ mapping, the first point is measured in presence of two 90° pulses with zero spin-lock time to keep the condition the same for measurements of all points in the relaxation decay. During these pulses the signal from spins having relaxing times comparable to the pulse durations could decay significantly. The latter is why the adiabatic version of $T_{1\rho}$ was preferred for the present study.

There are several limitations of the developed method and analytical formalism, which need to be elaborated further. Firstly, since the approximation $T_{1\rho}^s \gg T_{1\rho}^f$ was used during the fitting, the fitted $T_{1\rho}^s$, C_s and C_f have intrinsic deviation from the fitting model in addition to experimental errors. Therefore, if separation of the multiple components is desirable, a more extended fitting strategy will be needed to better separate different relaxing components. Secondly, four AFP pulses were used in one MP block which may lead to SAR limitation for in vivo applications, particularly at high field. To reduce SAR and increase efficiency, the AFP preparation was applied every m TR ($m = 16$ in this paper) instead of every single TR. The SAR of adiabatic pulses could be compared with that of CW- $T_{1\rho}$ by calculating the root-mean-square RF power; however, due to the differences in pulse modulations, the relaxation phenomena probed by the two methods would be different (20). The benefit of adiabatic pulses is that they offer additional flexibility for alleviating SAR issues (24). Finally, an independent T_1 measurement is required for the quantification of $T_{1\rho}$, which leads to a longer total acquisition time. In future work, a view-sharing method might be utilized to reduce the total measurement time (55). Furthermore, the assumption of a single T_1 component, which was applied in both the T_1 and $T_{1\rho}$ measurements, was based on our earlier work in which multiple T_1 components were not observed in the cartilage system (49). To be more precise, for the systems with multiple T_1 components, further extension of the analytical procedure will be needed. Despite these limitations, many MRI applications may benefit from the improved contrast provided by approaches similar to those introduced here.

Conclusion

In conclusion, a method called MP-SWIFT has been described for measuring $T_{1\rho}$ in tissues containing fast relaxing spins. The signal equation of the proposed method and a semi-analytical procedure for the $T_{1\rho}$ mapping were derived. SWIFT- $T_{1\rho}$ and FSE- $T_{1\rho}$ values matched well with each other in the case of spins having long T_2 values. However, in normal biological tissues, multiple $T_{1\rho}$ relaxing components were observed in SWIFT measurements. Besides information about slowly relaxing spins, both simulations and experimental results suggest that SWIFT $T_{1\rho}^a$, a parameter approaching the weighted harmonic mean of the $T_{1\rho}$ of all decaying components, contains extra contributions from fast relaxing spins which are undetected with FSE readout. The relative contrast ratio between normal and lesion tissues in SWIFT $T_{1\rho}^a$ map was also much higher than in the FSE $T_{1\rho}$ map. The potential application for detecting degeneration in the musculoskeletal system has been demonstrated, but the extent to which this method can be used remains to be shown. Furthermore, $T_{1\rho}$ is not only applicable in musculoskeletal tissues, but has also been associated with neurological disorders such as Parkinson's and Alzheimer's diseases (56,57), and has been proposed recently for applications in cardiac (58) and abdominal (59) MRI. The developed 3D SWIFT $T_{1\rho}$ relaxation mapping technique in this paper also has high potential for studying these diseases.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

Acknowledgments

We thank Dr. Ferenc Toth for providing the animal specimens and histology.

Grant sponsor: NIH P41 EB015894, WM KECK foundation, Academy of Finland #260321 and #285909.

References

1. Redfield AG. Nuclear spin thermodynamics in the rotating frame. *Science*. 1969; 164(3883):1015–1023. [PubMed: 17796604]
2. Sepponen RE, Pohjonen JA, Sipponen JT, Tanttu JI. A method for T1 rho imaging. *J Comput Assist Tomo*. 1985; 9(6):1007–1011.
3. Rommel E, Kimmich R. Volume-selective determination of the spin-lattice relaxation time in the rotating frame T1 rho, and T1 rho imaging. *Magn Reson Med*. 1989; 12(2):209–218. [PubMed: 2615628]
4. Duvvuri U, Charagundla SR, Kudchodkar SB, Kaufman JH, Kneeland JB, Rizi R, Leigh JS, Reddy R. Human knee: In vivo T1(p)-weighted MR imaging at 1.5 T - Preliminary experience. *Radiology*. 2001; 220(3):822–826. [PubMed: 11526288]
5. Duvvuri U, Goldberg AD, Kranz JK, Hoang L, Reddy R, Wehrli FW, Wand AJ, Englander SW, Leigh JS. Water magnetic relaxation dispersion in biological systems: the contribution of proton exchange and implications for the noninvasive detection of cartilage degradation. *Proc Natl Acad Sci U S A*. 2001; 98(22):12479–12484. [PubMed: 11606754]
6. Borthakur A, Mellon E, Niyogi S, Witschey W, Kneeland JB, Reddy R. Sodium and T1rho MRI for molecular and diagnostic imaging of articular cartilage. *NMR in biomed*. 2006; 19(7):781–821.
7. Keenan KE, Besier TF, Pauly JM, Han E, Rosenberg J, Smith RL, Delp SL, Beaupre GS, Gold GE. Prediction of glycosaminoglycan content in human cartilage by age, T1rho and T2 MRI. *Osteoarthritis Cartil*. 2011; 19(2):171–179. [PubMed: 21112409]
8. Duvvuri U, Reddy R, Patel SD, Kaufman JH, Kneeland JB, Leigh JS. T1rho-relaxation in articular cartilage: effects of enzymatic degradation. *Magn Reson Med*. 1997; 38(6):863–867. [PubMed: 9402184]
9. Duvvuri U, Kudchodkar S, Reddy R, Leigh JS. T-1 rho relaxation can assess longitudinal proteoglycan loss from articular cartilage in vitro. *Osteoarthritis Cartil*. 2002; 10(11):838–844. [PubMed: 12435327]
10. Regatte RR, Akella SV, Borthakur A, Kneeland JB, Reddy R. Proteoglycan depletion-induced changes in transverse relaxation maps of cartilage: comparison of T2 and T1rho. *Acad Radiol*. 2002; 9(12):1388–1394. [PubMed: 12553350]
11. Jordan CD, McWalter EJ, Monu UD, Watkins RD, Chen W, Bangerter NK, Hargreaves BA, Gold GE. Variability of CubeQuant T1rho, quantitative DESS T2, and cones sodium MRI in knee cartilage. *Osteoarthritis Cartil*. 2014; 22(10):1559–1567. [PubMed: 25278065]
12. Kumar D, Souza RB, Singh J, Calixto NE, Nardo L, Link TM, Li X, Majumdar S. Physical activity and spatial differences in medial knee t1rho and t2 relaxation times in knee osteoarthritis. *J Orthop Sports Phys Ther*. 2014; 44(12):964–972. [PubMed: 25353261]
13. Rautiainen J, Nissi MJ, Liimatainen T, Herzog W, Korhonen RK, Nieminen MT. Adiabatic rotating frame relaxation of MRI reveals early cartilage degeneration in a rabbit model of anterior cruciate ligament transection. *Cartil Osteoarthritis*. 2014; 22(10):1444–1452.
14. Rautiainen J, Nissi MJ, Salo EN, Tiitu V, Finnila MA, Aho OM, Saarakkala S, Lehenkari P, Ellermann J, Nieminen MT. Multiparametric MRI assessment of human articular cartilage degeneration: Correlation with quantitative histology and mechanical properties. *Magn Reson Med*. 2015; 74:249–259.
15. Wang YX, Zhao F, Griffith JF, Mok GS, Leung JC, Ahuja AT, Yuan J. T1rho and T2 relaxation times for lumbar disc degeneration: an in vivo comparative study at 3.0-Tesla MRI. *Eur Radiol*. 2013; 23(1):228–234. [PubMed: 22865227]

16. Toth F, Nissi MJ, Wang L, Ellermann JM, Carlson CS. Surgical induction, histological evaluation, and MRI identification of cartilage necrosis in the distal femur in goats to model early lesions of osteochondrosis. *Osteoarthritis Cartil.* 2014
17. Santyr GE, Fairbanks EJ, Kelcz F, Sorenson JA. Off-resonance spin locking for MR imaging. *Magn Reson Med.* 1994; 32(1):43–51. [PubMed: 8084236]
18. Grohn HI, Michaeli S, Garwood M, Kauppinen RA, Grohn OHJ. Quantitative T-1 ρ and adiabatic Carr-Purcell T-2 magnetic resonance imaging of human occipital lobe at 4 T. *Magn Reson Med.* 2005; 54(1):14–19. [PubMed: 15968651]
19. Michaeli S, Sorce DJ, Springer CS Jr, Ugurbil K, Garwood M. T1 ρ MRI contrast in the human brain: modulation of the longitudinal rotating frame relaxation shutter-speed during an adiabatic RF pulse. *J Magn Reson.* 2006; 181(1):135–147. [PubMed: 16675277]
20. Mangia S, Liimatainen T, Garwood M, Michaeli S. Rotating frame relaxation during adiabatic pulses vs. conventional spin lock: simulations and experimental results at 4 T. *Magn Reson Imaging.* 2009; 27(8):1074–1087. [PubMed: 19559559]
21. Taheri S, Sood R. Spin-lock MRI with amplitude- and phase-modulated adiabatic waveforms: an MR simulation study. *Magn Reson Imaging.* 2006; 24(1):51–59. [PubMed: 16410178]
22. Blicharski J. Nuclear magnetic relaxation in rotating frame. *Acta Phys Pol A.* 1972; 41:223.
23. Ellermann J, Ling W, Nissi MJ, Arendt E, Carlson CS, Garwood M, Michaeli S, Mangia S. MRI rotating frame relaxation measurements for articular cartilage assessment. *Magn Reson Imaging.* 2013; 31(9):1537–1543. [PubMed: 23993794]
24. Tannus A, Garwood M. Adiabatic Pulses. *NMR in biomed.* 1997:10.
25. Nissi MJ. Orientation anisotropy of rotating frame and T2 relaxation parameters in articular cartilage. *Proc Natl Acad Sci U S A.* 2013; 3552
26. Wang N, Xia Y. Dependencies of multi-component T2 and T1 ρ relaxation on the anisotropy of collagen fibrils in bovine nasal cartilage. *J Magn Reson.* 2011; 212(1):124–132. [PubMed: 21788148]
27. Du J, Diaz E, Carl M, Bae W, Chung CB, Bydder GM. Ultrashort echo time imaging with bicomponent analysis. *Magn Reson Med.* 2012; 67(3):645–649. [PubMed: 22034242]
28. Liu F, Choi KW, Samsonov A, Spencer RG, Wilson JJ, Block WF, Kijowski R. Articular Cartilage of the Human Knee Joint: In Vivo Multicomponent T2 Analysis at 3.0 T. *Radiology.* 2015; 277(2): 11.
29. Pauli C, Bae WC, Lee M, Lotz M, Bydder GM, D'Lima DL, Chung CB, Du J. Ultrashort-echo time MR imaging of the patella with bicomponent analysis: correlation with histopathologic and polarized light microscopic findings. *Radiology.* 2012; 264(2):484–493. [PubMed: 22653187]
30. Du J, Carl M, Diaz E, Takahashi A, Han E, Szevenyi NM, Chung CB, Bydder GM. Ultrashort TE T1 ρ (UTE T1 ρ) imaging of the Achilles tendon and meniscus. *Magn Reson Med.* 2010; 64(3):834–842. [PubMed: 20535810]
31. Aronen HJ, Ramadan UA, Peltonen TK, Markkola AT, Tantt J, Jaaskelainen J, Hakkinen AM, Sepponen R. 3D spin-lock imaging of human gliomas. *Magn Reson Imaging.* 1999; 17(7):1001–1010. [PubMed: 10463651]
32. Borthakur A, Wheaton A, Charagundla SR, Shapiro EM, Regatte RR, Akella SVS, Kneeland JB, Reddy R. Three-dimensional T1 ρ -weighted MRI at 1.5 Tesla. *J Magn Reson Imaging.* 2003; 17(6):730–736. [PubMed: 12766904]
33. Regatte RR, Akella SV, Wheaton AJ, Lech G, Borthakur A, Kneeland JB, Reddy R. 3D-T1 ρ -relaxation mapping of articular cartilage: in vivo assessment of early degenerative changes in symptomatic osteoarthritic subjects. *Acad Radiol.* 2004; 11(7):741–749. [PubMed: 15217591]
34. Li X, Han ET, Busse RF, Majumdar S. In vivo T(1 ρ) mapping in cartilage using 3D magnetization-prepared angle-modulated partitioned k-space spoiled gradient echo snapshots (3D MAPSS). *Magn Reson Med.* 2008; 59(2):298–307. [PubMed: 18228578]
35. Hafner S. Fast imaging in liquids and solids with the Back-projection Low Angle Shot (BLAST) technique. *Magn Reson Imaging.* 1994; 12(7):1047–1051. [PubMed: 7997092]
36. Madio DP, Lowe IJ. Ultra-Fast Imaging Using Low Flip Angles and Fids. *Magn Reson Med.* 1995; 34(4):525–529. [PubMed: 8524019]

37. Robson MD, Gatehouse PD, Bydder M, Bydder GM. Magnetic resonance: an introduction to ultrashort TE (UTE) imaging. *J Comput Assist Tomo.* 2003; 27(6):825–846.
38. Weiger M, Brunner DO, Dietrich BE, Muller CF, Pruessmann KP. ZTE imaging in humans. *Magn Reson Med.* 2013; 70(2):328–332. [PubMed: 23776142]
39. Idiyatullin D, Corum C, Park JY, Garwood M. Fast and quiet MRI using a swept radiofrequency. *J Magn Reson.* 2006; 181(2):342–349. [PubMed: 16782371]
40. Wu Y, Ackerman JL, Chesler DA, Graham L, Wang Y, Glimcher MJ. Density of organic matrix of native mineralized bone measured by water- and fat-suppressed proton projection MRI. *Magn Reson Med.* 2003; 50(1):59–68. [PubMed: 12815679]
41. Zhang J, Idiyatullin D, Corum CA, Kobayashi N, Garwood M. Gradient-modulated SWIFT. *Magn Reson Med.* 201510.1002/mrm.25595
42. Garwood, M.; Idiyatullin, D.; Corum, C.; Chamberlain, R.; Moeller, S.; Kobayashi, N.; Lehto, L.; Zhang, J.; O'Connell, R.; Tesch, M.; Nissi, M.; Ellermann, J.; Nixdorf, D. *SWIFT Encyclopedia of Magnetic Resonance.* John Wiley & Sons; 2012. Capturing Signals from Fast-relaxing Spins with Frequency-Swept MRI: SWIFT.
43. Kendi AT, Khariwala SS, Zhang J, Idiyatullin DS, Corum CA, Michaeli S, Pambuccian SE, Garwood M, Yueh B. Transformation in mandibular imaging with sweep imaging with fourier transform magnetic resonance imaging. *Arch Otolaryngol Head Neck Surg.* 2011; 137(9):916–919. [PubMed: 21930980]
44. Zhang J, Chamberlain R, Etheridge M, Idiyatullin D, Corum C, Bischof J, Garwood M. Quantifying iron-oxide nanoparticles at high concentration based on longitudinal relaxation using a three-dimensional SWIFT look-locker sequence. *Magn Reson Med.* 2014; 71(6):1982–1988. [PubMed: 24664527]
45. Ellermann J, Idiyatullin D, Corum C, Moeller S, Ling W, Garwood M. The ultrastructure of bone using selective saturation in SWIFT at 9.4 T. *Proc Natl Acad Sci U S A.* 2009; 849
46. Garwood M, DelaBarre L. The return of the frequency sweep: designing adiabatic pulses for contemporary NMR. *J Magn Reson.* 2001; 153(2):155–177. [PubMed: 11740891]
47. Levitt MH, Freeman R, Frenkiel T. Supercycles for Broad-Band Heteronuclear Decoupling. *J Magn Reson.* 1982; 50(1):157–160.
48. Ernst RR, Anderson WA. Application of Fourier transform to magnetic resonance spectroscopy. *Rev Sci Instrum.* 1966; 37:93.
49. Nissi MJ, Lehto LJ, corum CA, Idiyatullin D, Ellermann JM, Grohn OHJ, Nieminen MT. Measurement of T1 Relaxation Time of Osteochondral Specimens Using VFA-SWIFT. *Magn Reson Med.* 201510.1002/mrm.25398
50. Reiter DA, Lin PC, Fishbein KW, Spencer RG. Multicomponent T2 relaxation analysis in cartilage. *Magn Reson Med.* 2009; 61(4):803–809. [PubMed: 19189393]
51. Degroen P, Demoor B. The Fit of a Sum of Exponentials to Noisy Data. *J Comput Appl Math.* 1987; 20:175–187.
52. Wong TT, Luk WS, Heng PA. Sampling with Hammersley and Halton Points. *J Graph Tools.* 1997; 2(2):9–24.
53. Reiter DA, Roque RA, Lin PC, Irrechukwu O, Doty S, Longo DL, Pleshko N, Spencer RG. Mapping proteoglycan-bound water in cartilage: Improved specificity of matrix assessment using multiexponential transverse relaxation analysis. *Magn Reson Med.* 2011; 65(2):377–384. [PubMed: 21264931]
54. Cobb JG, Kan JH, Gore JC. T1rho mapping of pediatric epiphyseal and articular cartilage in the knee. *Journal of magnetic resonance imaging : JMRI.* 2013; 38(2):299–305. [PubMed: 23650103]
55. Altbach MI, Outwater EK, Trouard TP, Krupinski EA, Theilmann RJ, Stopeck AT, Kono M, Gmitro AF. Radial fast spin-echo method for T2-weighted imaging and T2 mapping of the liver. *J Magn Reson Imaging.* 2002; 16(2):179–189. [PubMed: 12203766]
56. Michaeli S, Oz G, Sorce DJ, Garwood M, Ugurbil K, Majestic S, Tuite P. Assessment of brain iron and neuronal integrity in patients with Parkinson's disease using novel MRI contrasts. *Mov Disord.* 2007; 22(3):334–340. [PubMed: 17149719]
57. Borthakur A, Sochor M, Davatzikos C, Trojanowski JQ, Clark CM. T1rho MRI of Alzheimer's disease. *NeuroImage.* 2008; 41(4):1199–1205. [PubMed: 18479942]

58. Muthupillai R, Flamm SD, Wilson JM, Pettigrew RI, Dixon WT. Acute myocardial infarction: tissue characterization with T1rho-weighted MR imaging--initial experience. *Radiology*. 2004; 232(2):606–610. [PubMed: 15215547]
59. Wang YX, Yuan J, Chu ES, Go MY, Huang H, Ahuja AT, Sung JJ, Yu J. T1rho MR imaging is sensitive to evaluate liver fibrosis: an experimental study in a rat biliary duct ligation model. *Radiology*. 2011; 259(3):712–719. [PubMed: 21436087]

Abbreviations

$T_{1\rho}$	spin-lattice relaxation time in rotating frame
SWIFT	sweep imaging with Fourier transformation
MP-SWIFT	magnetization prepared-SWIFT
FSE	fast spin echo

Fig.1.

(a) Illustration of adiabatic $T_{1\rho}$ relaxation in the Larmor rotating frame. During the adiabatic rotation the magnetization follows the effective field B_{eff} . (b) Schematic of the magnetization prepared SWIFT sequence. MP blocks are embedded after each m number of SWIFT readout periods (mTR periods). Each MP block is composed of four AFP pulses, with consecutive phases prescribed according to the MLEV4 scheme.

Fig.2.

Bloch simulations showing evolution of the signal to the steady-state during MP-SWIFT pulse sequence with different T_{mp} values. The observed signal oscillation is the result of two processes: T_{1p} decay during each T_{mp} (black portion) and T_1 recovery during m SWIFT readouts (color-coded points). The parameters used for simulations were: $T_1 = 500$ ms, $T_{1p} = 40$ ms, $TR = 2.6$ ms, $m = 16$, $t_{180} = 1.5$ ms and SWIFT flip angle = 2° .

Fig.3.

(a) Fitting example of the simulated decay $S(T_{mp})$ (red circles). The parameters used for simulations are: $T_{1\rho}^f = 10$ ms (50% population) and $T_{1\rho}^s = 150$ ms (50% population). T_1 was assumed to be the same for both fast and slow relaxing components, $T_1 = 1000$ ms. The red solid line shows the fitting using just the first two data points ($T_{1\rho}^a$) to provide an estimation approaching $T_{1\rho}^ave$. The grey dash-dot line shows the resulting $T_{1\rho}^s$ and C_s obtained from the intermediate fitting (black dash line) (see Appendix B for details). (b) Examples of the asymptotic $T_{1\rho}^a$ approaching $T_{1\rho}^ave$ at different conditions. The empty circles are the values of $T_{1\rho}^ave$.

Fig.4.

(a) Representative $T_{1\rho}$ time constant maps of MnCl_2 doped agar (smaller tube) merged in the bigger tube containing saline. (b) Example plots of the signal evolution for both saline and MnCl_2 doped agar. Both FSE and SWIFT plots are indicating single component decay. (c) Comparison of the adiabatic $T_{1\rho}$ values of three different phantoms measured with FSE and SWIFT sequences. 400 data points were plotted for each phantom. The solid line is the line of identity, representing perfect correspondence. The dash line is the linear fit of the averaged $T_{1\rho}$ values from the three data sets, with $r = 0.999$ and $p < 0.01$. Each data point corresponds to one voxel in the $T_{1\rho}$ maps.

Fig.5.

MP-SWIFT images of the bovine tendon specimen acquired with different T_{mp} (a) and graphs (b,c) representing S_e signal measured in ROI (shown in the image by yellow square) with two different t_{180} durations. The data points did not fall on the same line indicate multiple decay rates. Since the effective field for pulses with durations of $t_{180} = 0.3$ and 1.2 ms were kept the same, some data points were overlapped as expected. (b,c) The solid lines are drawn by connecting first two data points, the slopes of which determines $T_{1\rho}^a$ in experiments with different t_{180} . The asymptotic $T_{1\rho}^a$ decreases with decreasing t_{180} demonstrating increased weighting from the fast relaxing spins. The dash lines are the fittings of data points with $T_{mp} > 5$ ms from which the contribution from slowly relaxing spins can be retrieved.

Fig.6.

$T_{1\rho}$ maps and example of the fitting strategies of the juvenile porcine femoral condyle specimen. The maps in (a) are the same map presented with different color scale. A gradient of the $T_{1\rho}$ values along the direction of the cartilage depth is observed in all maps. Below the articular cartilage (labeled as $z1$ in (a)) is the epiphyseal cartilage (labeled as $z2$ in (a)). SWIFT $T_{1\rho}^a$ values of the normal epiphyseal growth and articular cartilage are much shorter than $T_{1\rho}$ measured by FSE (b), while the SWIFT $T_{1\rho}^s$ (f) values are close to FSE $T_{1\rho}$ (b). (c) H&E stain of the same specimen revealing the focal area of chondronecrosis matching the lesion detected in MR images. The signal from a ROI in normal tissues (blue line in d) from FSE experiment is apparently single component decay, while the signal from SWIFT experiment (blue line in e) exhibits multi-component decay. The signal from the ROI in the lesion region shows multi-exponential decay even in the FSE measurement (red data points in d). Of note, the $T_{1\rho}$ values within the lesion region is comparatively similar for SWIFT and FSE measurements, which implies the absence of fast relaxing components within the lesion. This conclusion is also supported by population map of the short component shown in (g).

(h) The average values of $T_{1\rho}$ from ROI over the normal epiphyseal cartilage and the focal lesion areas. The relative contrast between normal and lesion tissues in the SWIFT $T_{1\rho}^a$ map is much higher than in the FSE $T_{1\rho}$ map.

Fig.7.

(a) The dependence of the deviation of $T_{1\rho}$, $D_{T_{1\rho}} (= \frac{(T_{1\rho}^{\text{fitted}} - T_{1\rho}^{\text{real}})}{T_{1\rho}^{\text{real}}})$, on the deviation of

T_1 , $D_{T_1} (= \frac{(T_1^{\text{fitted}} - T_1^{\text{real}})}{T_1^{\text{real}}})$. (b), (c) & (d) show the relative deviations

$D_v = \frac{|V_{\text{fitted}} - V_{\text{real}}|}{V_{\text{real}}}$, with v representing, $T_{1\rho}^s$, C_s or C_f respectively. The deviations

originated from the approximations used in the fitting procedure. $T_{1\rho}^f$ was fixed to 10 ms.

For the $\frac{T_{1\rho}^s}{T_{1\rho}^f}$ ratio in the range from 10 to 30, the intrinsic error from the fitting model was less than 15%. (e) & (f) Dependence of the fitting error on SNR by using Monte-Carlo simulation. $T_{1\rho}^f=10$ ms and $T_{1\rho}^s=250$ ms was used with equal population. For each data

point in (e) & (f), the fitting process was repeated 1000 times with noise chosen from the corresponding normal distribution randomly.

Table 1

Parameters Used In FSE and SWIFT $T_{1\rho}$ Measurements

Sample	t_{180} (ms)	N_{mp}	BW_{mp} (kHz)	ω_{1eff}^{max} (kHz)	ω_{1eff}^{avg} (kHz)	$\theta(^{\circ})$	Resolution (μm^2)
Phantom 1	1.5	SWIFT: 0, 2, 4 FSE: 0, 4, 8	10	5	5	2	130×130
Phantom 2	3	SWIFT: 0, 1, 2 FSE: 0, 2, 4	10	5	5	2	130×130
Phantom 3	3	SWIFT: 0, 1, 2 FSE: 0, 2, 4	4.8	5	2.7	2	130×130
Tendon	0.3	SWIFT: 0, 1, 2, 3, 4, 6, 8, 12, 16	18	10	8.5	4	78×78
	1.2	SWIFT: 0, 1, 2, 3, 4, 5	18	10	8.5	4	78×78
Femoral condyle	2	SWIFT: 0, 1, 2, 3, 4 FSE: 0, 1, 2, 3, 4, 6	10	5	5	3	130×130