Published in final edited form as:

NMR Biomed. 2020 November ; 33(11): e4391. doi:10.1002/nbm.4391.

Rapid Single Scan Ramped Hybrid Encoding for Bicomponent T2* Mapping in a Human Knee Joint: A Feasibility Study

Hyungseok Jang^{1,*}, Alan B McMillan², Yajun Ma¹, Saeed Jerban¹, Eric Y Chang^{3,1}, Jiang Du¹, Richard Kijowski²

¹Department of Radiology, University of California San Diego, San Diego, CA 92103, USA

²Department of Radiology, University of Wisconsin Madison, Madison, WI 53705, USA

³Radiology Service, VA San Diego Healthcare System, San Diego, CA 92037, USA

Abstract

The purpose of this study is to determine the feasibility of using a single scan ramped hybrid encoding (RHE) method for rapid bicomponent T2* analysis of the human knee joint.

The proposed method utilizes RHE to acquire UTE and subsequent gradient echo images at 16 different echo times ranging between 40 μ s and 30ms in a single scan. In the proposed RHE technique, UTE imaging was followed by 14 gradient recalled echo imaging, where an additional UTE image was obtained within the first readout by oversampling single point imaging (SPI) encoding. The single scan RHE method with 9-minute scan time was performed on human cadaveric knee joints from six donors and in vivo knee joints from four healthy volunteers at 3T. A bicomponent signal model was used to characterize the short T2* and long T2* water components. Mean bicomponent T2* parameters for patellar tendon, anterior cruciate ligament (ACL), posterior cruciate ligament (PCL), and meniscus were calculated.

In the experimental results, the RHE technique provided bicomponent T2* parameter estimations of tendon, ACL, PCL, and meniscus which were similar to previously reported values in the literature.

In conclusion, the proposed single scan RHE technique provides rapid bicomponent T2* analysis of the human knee joint with a total scan time less than 9 min.

Keywords

Hybrid Encoding; T2*; Bicomponent; UTE; Knee; SPI; PETRA; RHE

INTRODUCTION

Ultrashort echo time (UTE) magnetic resonance (MR) imaging techniques can capture rapidly decaying signals within collagen-rich musculoskeletal tissues such as tendon, ligament, and meniscus (1). UTE techniques have been used to measure short T2* relaxation

*Corresponding Author: Hyungseok Jang, Ph.D., University of California, San Diego, Department of Radiology, 200 West Arbor Drive, San Diego, CA 92103-8226, Phone (858) 246-2225, h4jang@ucsd.edu.

time by acquiring multiple echoes and fitting the image data using a single-component reconstruction model. While single-component short T2* analysis can detect disease-related and trauma-induced changes in tendons (2–4), menisci (5–7), and ligaments (8–10), T2* relaxation time is a nonspecific parameter, which depends on field strength and is highly sensitive to the magic angle effect. Therefore, it may be difficult to be translated as a quantitative biomarker for clinical assessment due to the multiple competing biological and technical factors that can influence the measurement (e.g., magic angle effect, field strength, and inhomogeneity).

Bicomponent models have recently been used to improve the specificity of T2* analysis of musculoskeletal tissues by characterizing both the fast relaxing water component bound to the macromolecular matrix and the slow relaxing bulk water component (11–16). Even in the most currently available UTE imaging techniques for bicomponent T2* mapping, it is common to acquire only a limited number of echoes within each single scan due to the gradient error induced by hardware imperfections, including eddy currents (17–20) and imperfection induced by the power amplifier, as well as mechanical/thermal vibrations (21,22), which is more problematic in the non-Cartesian radial sampling strategy. Thus, these sequences need to be repeated multiple times to acquire the 12 to 16 echoes typically required for reliable bicomponent T2* analysis (11–15), resulting in long scan times for whole knee joint coverage.

A rapid imaging technique utilizing ramped hybrid encoding (RHE) has been recently proposed to provide more efficient UTE imaging (23), which is a hybrid imaging scheme of pointwise encoding time reduction with radial acquisition (PETRA) (24) and conventional radial UTE imaging (25). RHE benefits from single point imaging (SPI) (26), which is known to be less susceptible to eddy currents owing to the nature of pure phase encoding with near-zero readout duration (27–29). In the literature, it has also been shown that RHE is able to acquire multiple UTE images in a single scan by oversampling SPI (27).

In this study, bicomponent T2* analysis of the patellar tendon, ligaments, and meniscus of the human knee joint was demonstrated at 3T using an extended single scan RHE approach with oversampled SPI. The proposed method acquires 16 echoes ranging between 40 μ s and 30ms in a single acquisition and provides complete anatomic coverage of the knee joint in a clinically feasible scan time (<9min). This study was performed to evaluate the feasibility of the single scan RHE method in ex vivo experiments with six human cadaveric knee joints and in vivo experiments with four knee joints from healthy volunteers.

MATERIALS AND METHODS

Single Scan Multi-Echo RHE

RHE has been proposed to optimize encoding time (or readout duration) for radial UTE imaging, which may reduce blurring when imaging tissues with rapid T2* decay (23). The RHE technique is further modified to acquire multiple echoes, as shown in Figure 1-a. As in the previously described RHE method (23), gradients are switched on before the RF coil dead time and ramped up to the maximum gradient. The missing data due to the RF coil dead time is filled using SPI, also known as constant time encoding or pure phase encoding.

Figure 1-b illustrates a 2D example of how k-space is sampled using the single scan RHE technique. The number of phase encoding steps in each axis (NSPI in figure 1-b) is determined by considering the desired FOV in each axis. One modification from the original RHE method made for this study is the application of a Shinnar-Le Roux (SLR) RF pulse with minimum phase (30) to allow slab selection to be performed with minimized dephasing. The effective gradient during the RF pulse, G_{RF} , is set to zero, as in zero- G_{RF} RHE sequences (27,31,32). Instead, a slab selection gradient is applied during the RF excitation.

In the proposed RHE technique, two UTE images (i.e., UTE_1 and UTE_2 in Figure 1-a) are obtained in the first readout gradient using SPI data acquired at two different echo times (TEs) as in (27). After the UTE acquisition, 14 consecutive non-UTE images are acquired using fly-back radial gradient echo train imaging. The non-UTE images are also hybrid-encoded to keep the sampling pattern consistent between TEs, though there is no RF coil dead time and the consequent missing data in central k-space. This approach is beneficial since SPI, which is robust to eddy currents and magnetic susceptibility (27,33,34), is used to sample the central k-space. Gradient spoiling was used along with RF spoiling.

Fat-Saturation

The single scan multi-echo RHE sequence was implemented on a 3T clinical MR system (MR750, GE Healthcare, Milwaukee, WI). A GE-provided chemical shift-based fat-saturation (fat-sat) pulse (width=8ms, offset frequency=-440Hz, and BW=500), was utilized to suppress fat signal that may have contaminated T2* curve fitting. To reduce the overhead in scan time due to the additional fat-sat module, the fat-sat preparation was performed in front of four consecutive acquisition of spokes, as shown in Figure 1-c, which has been used in UTE imaging (12,13,35,36). The efficiency of fat-sat majorly depends on T1 recovery of the saturated fat signal at the imaging. Assuming fat-sat module length of 12 ms and spoke-to-spoke timing (τ) of 36ms, T1 of fat signal of 360ms (typical T1 of fat at 3T), and number of spokes per fat-sat (NSP) of 4, the fat signal is expected to be suppressed by 100%, 90.5%, 81.9%, and 74% for the four consecutive acquisitions at 0ms, 36ms, 72ms, and 108ms. On average, we expect approximately 86.6% suppression of fat signal with NSP of 4. This is a rough approximation to show multi-spoke based fat-sat UTE imaging is feasible, not considering the efficiency of the fat-sat pulse itself.

Experimental Setup

An ex vivo study was performed on human cadaveric knee joints from six male donors (age: 64.0 ± 5.4 years). An in vivo study was performed on the knee joints of four healthy male volunteers (age: 33.0 ± 1.2 years) in compliance with Health Insurance Portability and Accountability Act (HIPAA) regulations, with approval from the University Institutional Review Board, and with all subjects providing written informed consent. The single scan RHE sequence was performed on the six human cadaveric knee joints and the knee joints of the four healthy volunteers using a 3T MR scanner (MR750) and an 8-channel Transmit/Receive extremity coil (InVivo, Orlando, FL). Foam padding was used to firmly secure the knee joints within the coil to minimize motion during MR imaging.

The single scan RHE sequence was performed using the following imaging parameters: flip angle (FA) = 15°, G_{RF} = 0 mT/m, G_{max} = 47.5 mT/m, spoiling gradient applied in x-y-z direction, an RF-spoiling seed = 117 degree, slew rate = 180 mT/m/ms, NSPI = 19, readout BW = 500kHz, number of radial spokes = 11185, number of SPI encoding = 2629, readout duration per echo = 0.64ms, matrix size = 256x256x30, FOV = 16x16x9cm³, number of spokes per fat-saturation pulse = 4, RF-to-RF timing (τ) = 35.5ms, length of fat-sat module = 12ms, TEs = [0.04, 0.11 1.40, 2.70, 4.30, 6.00, 8.00, 10.00, 14.00, 16.00, 18.00, 20.00, 24.00, 26.00, 28.0,0 and 30.00ms], and total scan time = 8min 53sec.

k-Space Trajectory Measurement

In this study, a dynamic SPI-based gradient measurement technique was utilized to measure k-space trajectory during the long readout (37), where 1D SPI was performed with a spherical phantom by using the phase encoding gradient that was prescribed by linearly scaling the targeted RHE readout gradient in each axis (1201 equally spaced phase encoding steps). The 1D SPI encoding was repeated three times in the physical x, y, and z axis with the gradient in the corresponding axis only turned on, which yielded 1D-projected images. The width of 1D profile was used to estimate the dynamically changing FOV over the phase encoding time delays, which was directly converted to k-space trajectory at the given phase encoding time delays based on the following FOV equation in phase encoding:

$$FOV(t) = \frac{\pi N_p}{\gamma \int_0^t G(\tau) d\tau} \quad (1)$$

, where γ is a gyromagnetic ratio of proton, N_p is the number of phase encoding steps, and $G(\tau)$ is the targeted gradient waveform to measure.

The readout gradient waveform was measured once prior to the ex vivo and in vivo studies, using a body coil and a spherical water phantom (GE 46-265826G6 TLT Head Sphere MRI Phantom, GE Healthcare, Waukesha, WI) with total scan time = 4min 48sec.

Image Reconstruction and Data Analysis

The images were reconstructed using convolution gridding with an oversampling rate of 1.5 and a kernel width of 5-pixel and with the k-space trajectory acquired by the dynamic SPI-based gradient measurement technique. Iterative convolution-based density compensation was applied to the convolution gridding (38). Motion registration was not applied since no rigid motion was observed within the data. A bicomponent exponential signal model was fitted to the data using a non-linear least square method under the assumption of a Rician-distributed noise (12,39) to characterize the short T2* water component bound to the macromolecular matrix and the long T2* bulk water components in each pixel based on M3 method in (12). The initial condition for short T2*, long T2*, and fraction of short T2* tissues was set as 2ms, 20ms, and 80%, respectively. The lower bound was set to 0 for the three parameters. The upper bound was set to 20ms, 200ms, and 100%, respectively. The averaged values for pixels of patellar tendon, anterior cruciate ligament (ACL), posterior cruciate ligament (PCL), and meniscus were calculated. Pixels with T2* values greater than

50ms were excluded from analysis due to unreliable fitting of long $T2^*$ components for the range of echoes used for the single scan RHE technique.

RESULTS

Qualitative Image Analysis

Figure 2 shows single scan RHE images reconstructed using the nominal and measured k-space trajectory. Images reconstructed using the nominal trajectory show strong ringing artifacts near tissue-to-tissue and tissue-to-air interfaces (indicated by the yellow arrows), while ringing artifacts are suppressed on the images reconstructed using the measured trajectory.

Figures 3-a and 3-b show representative images at 16 echoes acquired using the single scan RHE technique from a human cadaveric knee joint (61-year-old male donor) and from the knee joint of a healthy volunteer (34-year-old male), respectively. There is a monotonic decay of signal for all musculoskeletal tissues, with much stronger decay occurring prior to 2ms for the patellar tendon. Note that streaking artifacts are visible for the single scan RHE images due to the undersampled radial sampling (40–44). In later echoes, the images are slightly degraded by susceptibility artifacts due to B0 inhomogeneity causing signal drop off and morphological distortion (indicated by the green arrow). However, the artifacts do not significantly affect image quality in structures of interest such as the patellar tendon, ACL, PCL, and meniscus.

Bicomponent $T2^*$ Parameter Estimation

Figure 4 shows bi-exponential $T2^*$ mapping results from a representative human cadaveric knee joint (61-year-old male). In this knee joint, the mean and standard deviation of fitted parameters for the patellar tendon are: f_s (fraction of short $T2^*$ tissues) = $79.5 \pm 8.1\%$, $T2^*_S$ (short $T2^*$) = 1.5 ± 0.8 ms, and $T2^*_L$ (long $T2^*$) = 25.8 ± 8.9 ms. The signal versus time plot (pixel-wise signal decay from the pixel in the region indicated by a yellow arrow) shows a typical bi-exponential decay curve. In the ACL, mean and standard deviation of the fitted parameters are: f_s = $46.2 \pm 19.7\%$, $T2^*_S$ = 3.0 ± 2.6 ms, and $T2^*_L$ = 18.4 ± 7.4 ms. In the PCL, mean and standard deviation of the fitted parameters are: f_s = $60.4 \pm 21.7\%$, $T2^*_S$ = 13.6 ± 4.1 ms, and $T2^*_L$ = 16.6 ± 4.9 ms. Signal decay for both the ACL and PCL shows oscillation, which is likely due to noise and interference from fat signal caused by suboptimal fat-saturation and partial volume effect. In the meniscus, mean and standard deviation of the fitted parameters are: f_s = $66.4 \pm 19.8\%$, $T2^*_S$ = 6.8 ± 3.4 ms, and $T2^*_L$ = 18.0 ± 7.0 ms. Signal decay for representative pixel (selected from the region indicated by the yellow arrow) shows a typical bi-exponential decay curve.

Figure 5 shows bi-exponential $T2^*$ results from a representative knee joint of a healthy volunteer (34-year-old male). In this knee joint, the mean and standard deviation of the fitted parameters for the patellar tendon are: f_s = $80.5 \pm 9.9\%$, $T2^*_S$ = 1.3 ± 4.2 ms, and $T2^*_L$ = 22.1 ± 10.1 ms. Signal decay for a representative pixel (at the yellow arrow) shows a typical bi-exponential decay curve with rapidly decaying short component. In the ACL, the mean and standard deviation of the fitted parameters are: f_s = $48.7 \pm 25.5\%$, $T2^*_S$ = 2.7 ± 2.3 ms,

and $T2^*_L = 20.5 \pm 9.6$ ms. In the PCL, the mean and standard deviation of the fitted parameters are: $f_s = 49.6 \pm 28.6$ %, $T2^*_S = 3.4 \pm 3.6$ ms, and $T2^*_L = 17.4 \pm 6.8$ ms. The ACL and PCL (selected from the region indicated by the yellow arrows) exhibit weaker bi-exponential signal decay than tendon and meniscus. In the meniscus, the mean and standard deviation of the fitted parameters are: $f_s = 59.9 \pm 16.8$ %, $T2^*_S = 2.8 \pm 0.7$ ms, and $T2^*_L = 17.9 \pm 6.6$ ms. A representative pixel in the middle of the posterior meniscus (at the yellow arrow) shows a typical bi-exponential decay curve.

Table 1 shows the mean and standard deviation of bi-exponential $T2^*$ parameter estimation performed on all six *ex vivo* knee joints and all four *in vivo* knee joints using the proposed single scan RHE method. The f_s parameters previously reported in the literature is ~80-90 % for tendon (12,13,45), ~79 % for ACL (45), ~75 % for PCL (45), and ~50-70 % for meniscus (7,46). The single scan RHE technique shows similar parameter estimation for patellar tendon and meniscus as reported values. In ligaments, the f_s showed underestimation compared with the previously reported value (45).

Supporting Figure S1 and Table S1 show the fitted parameters with or without inclusion of the data point at UTE2. Overall, higher f_s is observed with the additional data point at UTE2.

DISCUSSION

Our study has demonstrated the feasibility of using a single scan RHE technique for rapid bicomponent $T2^*$ analysis of tendon, ACL, PCL, and meniscus of the human knee joint at 3T with a total scan time less than 9min. The proposed method benefits from slight oversampling of the SPI region to obtain two UTE echoes at 0.04ms and 0.11ms. Although the echo spacing is small, signal decay is clearly observed between the first and second UTE echoes, as shown in the signal versus time plots in Figures 4 and 5. The contribution of the data point at UTE2 in the parameter fitting is shown in the supporting document. A higher degree of oversampling in SPI could achieve larger echo spacing between the UTE echoes at the expense of increased scan time. The hybrid encoding strategy covering the central region of k-space with SPI may provide more reliable $T2^*$ parameter estimation than conventional UTE sequences, as SPI is more robust to off-resonance effect that may shift and blur k-space center in the regular multiple acquired echoes. The efficacy of RHE for the morphological imaging has been shown in the literature (23,27,32). However, this potential benefit of the RHE technique in the quantitative MR imaging needs to be further investigated in future studies.

In this study, we used fat-saturation preparation to reduce fat signal contamination to the targeted signal. However, the results in Figures 4 and 5 imply that bicomponent $T2^*$ mapping for ACL and PCL may be challenging due to the partial volume effect with the surrounding adipose tissues occurring through the slices, while tendon and meniscus demonstrate less oscillatory signal decay, which is presumably why we obtained different parameters from the previously reported values in the literature (45). Moreover, it has been recently reported that fat-saturation pulses can influence the UTE signal from short $T2^*$ tissues (36). Alternatively, bicomponent $T2^*$ analysis using the single scan RHE method would greatly benefit from improvements in fat-suppression techniques (47–49). Another

possible approach is to utilize tricomponent $T2^*$ fitting including fat signal fraction. We have recently shown the feasibility of tricomponent analysis in bone imaging (50), which reduces the fat signal contamination in the parameter mapping. Combined with the proposed single scan multi-echo RHE method, more accurate $T2^*$ parameters could be achieved with a reduced scan time.

The total scan time for the single scan RHE technique is 8min 53sec, which is much faster than most conventional multi-acquisition-based methods currently used for bicomponent $T2^*$ mapping (13,45) where multiple scans are performed. If pure radial UTE sequence without SPI is used to acquire 16 echoes in four scans with the matching imaging parameters, the total scan time will be ~29min. The single scan multi-echo RHE sequence achieved more than 68% reduction in the scan time. Furthermore, using the single scan RHE technique, all 16 echoes needed for bicomponent signal fitting are acquired during the same acquisition. Thus, our approach is inherently less sensitive to subject motion when compared to conventional multi-acquisition based methods in which co-registration of images acquired in different scans is needed to compensate for motion (11–15). Note that it is challenging to perfectly correct for subject motion using co-registration, especially when the motion is non-rigid, as it may cause unpredictable errors in parameter estimation, especially for small joint structures such as the meniscus.

There are other approaches which can be used to reduce scan time for the bi-exponential $T2^*$ analysis. The methods primarily used are based on undersampling k-space to accelerate acquisition. This can be done in k-space (i.e., number of spokes) (51) or p-space (i.e., number of echoes) (52). The undersampled data in the k-space or p-space can be recovered by novel reconstruction algorithms based on compressed sensing (51) or deep learning (53). These methods of acceleration can also be applied to the proposed single scan RHE acquisition, which will be investigated in future studies.

Our study has several limitations. First, ex vivo knee joints were obtained from older donors (age: 64.0 ± 5.4), while in vivo subjects were recruited from a younger population of healthy volunteers (age: 33.0 ± 1.2). In particular, the estimated $T2^*_S$ for the ACL and PCL in the ex vivo experiments were higher than the estimated $T2^*_S$ in the in vivo experiments. This discrepancy was presumably due to age-related degeneration of the ligaments in the older ex vivo cadaveric knee joints. Second, the sample size in this feasibility study was small, with only six ex vivo knee joints and four in vivo knee joints imaged using the single scan RHE method. Third, the last echo time in the RHE imaging protocol was 30ms, which may not be sufficient to fit the long $T2^*$ signal precisely, leading to errors in fitting $T2^*_L$ and its relative fraction. Fourth, the current single scan RHE technique could not be used for bicomponent $T2^*$ analysis of cartilage since cartilage has water components with longer $T2^*$ relaxation times than tendon, ligament, and meniscus. (11–15,54). The RHE method may need to be modified to include additional later echoes to precisely measure bicomponent $T2^*$ parameters for cartilage; however, this may exacerbate the eddy current effect. More investigation will be required to optimize the imaging parameters. Finally, our feasibility study did not investigate the ability of the single scan RHE method to perform bicomponent $T2^*$ analysis of degenerative tendon, ligament, and meniscus in symptomatic knee joints.

CONCLUSIONS

Our study has demonstrated the feasibility of using a single scan RHE technique for reliable bicomponent T2* analysis of tendon, ligament, and meniscus within the human knee joint at 3T with whole knee coverage in a clinically feasible scan time.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

ACKNOWLEDGEMENTS

The authors acknowledge grant support from NIH (R01 AR068373, R01 AR062581, R01 AR068987, and R01 EB026708), Veterans Affairs (Merit Awards 1I01CX001388 and 1I01RX002604), and GE Healthcare. We thank Dr. Fang Liu at Massachusetts General Hospital for support in data processing.

Abbreviations used

UTE	ultrashort echo time
RHE	ramped hybrid encoding
SPI	single point imaging
PETRA	pointwise encoding time reduction with radial acquisition
ZTE	zero time echo imaging
FOV	field-of-view
TR	repetition time
FA	flip angle
ACL	anterior cruciate ligament
PCL	posterior cruciate ligament

REFERENCES

1. Chang EY, Du J, Chung CB. UTE imaging in the musculoskeletal system. *J. Magn. Reson. Imaging* 2015;41:870–883 doi: 10.1002/jmri.24713. [PubMed: 25045018]
2. Koff MF, Pownder SL, Shah PH, Yang LW, Potter HG. Ultrashort echo imaging of cyclically loaded rabbit patellar tendon. *J. Biomech.* 2014;47:3428–3432 doi: 10.1016/j.jbiomech.2014.08.018. [PubMed: 25234349]
3. Grosse U, Springer F, Hein T, et al. Influence of physical activity on T1 and T2* relaxation times of healthy Achilles tendons at 3T. *J. Magn. Reson. Imaging* 2015;41:193–201 doi: 10.1002/jmri.24525. [PubMed: 24347267]
4. Grosse U, Syha R, Hein T, et al. Diagnostic value of T1 and T2* relaxation times and off-resonance saturation effects in the evaluation of achilles tendinopathy by MRI at 3T. *J. Magn. Reson. Imaging* 2015;41:964–973 doi: 10.1002/jmri.24657. [PubMed: 24817378]
5. Chu CR, Williams AA, West RV, et al. Quantitative Magnetic Resonance Imaging UTE-T2 * Mapping of Cartilage and Meniscus Healing After Anatomic Anterior Cruciate Ligament

- Reconstruction. *Am. J. Sports Med* 2014;42:1847–1856 doi: 10.1177/0363546514532227. [PubMed: 24812196]
6. Williams A, Qian Y, Golla S, Chu CR. UTE-T2* mapping detects sub-clinical meniscus injury after anterior cruciate ligament tear. *Osteoarthr. Cartil* 2012;20:486–494 doi: 10.1016/j.joca.2012.01.009.
 7. Juras V, Apprigh S, Zbý Š, et al. Quantitative MRI analysis of menisci using biexponential T2* fitting with a variable echo time sequence. *Magn. Reson. Med* 2014;71:1015–1023 doi: 10.1002/mrm.24760. [PubMed: 23606167]
 8. Biercevicz AM, Walsh EG, Murray MM, Akelman MR, Fleming BC. Improving the clinical efficiency of T2* mapping of ligament integrity. *J. Biomech* 2014;47:2522–2525 doi: 10.1016/j.jbiomech.2014.03.037. [PubMed: 24792580]
 9. Biercevicz AM, Murray MM, Walsh EG, Miranda DL, Machan JT, Fleming BC. T2* MR relaxometry and ligament volume are associated with the structural properties of the healing ACL. *J. Orthop. Res* 2014;32:492–499 doi: 10.1002/jor.22563. [PubMed: 24338640]
 10. Biercevicz AM, Miranda DL, Machan JT, Murray MM, Fleming BC. In Situ, noninvasive, T2*-weighted MRI-derived parameters predict ex vivo structural properties of an anterior cruciate ligament reconstruction or bioenhanced primary repair in a porcine model. *Am. J. Sports Med* 2013;41:560–6 doi: 10.1177/0363546512472978. [PubMed: 23348076]
 11. Juras V, Apprigh S, Szomolanyi P, Bieri O, Deligianni X, Trattnig S. Bi-exponential T2* analysis of healthy and diseased Achilles tendons: an in vivo preliminary magnetic resonance study and correlation with clinical score. *Eur. Radiol.* 2013;23:2814–2822 doi: 10.1007/s00330-013-2897-8. [PubMed: 23760303]
 12. Liu F Assessment of different fitting methods for in-vivo bi-component T2* analysis of human patellar tendon in magnetic resonance imaging. *Muscle, Ligaments Tendons J.* 2017;7:163 doi: 10.11138/mltj/2017.7.1.163.
 13. Kijowski R, Wilson JJ, Liu F. Bicomponent ultrashort echo time T2* analysis for assessment of patients with patellar tendinopathy. *J. Magn. Reson. Imaging* 2017;46:1441–1447 doi: 10.1002/jmri.25689. [PubMed: 28263448]
 14. Chang EY, Du J, Iwasaki K, et al. Single- and Bi-component T2* analysis of tendon before and during tensile loading, using UTE sequences. *J. Magn. Reson. Imaging* 2015;42:114–120 doi: 10.1002/jmri.24758. [PubMed: 25223714]
 15. Chang EY, Du J, Statum S, Pauli C, Chung CB. Quantitative bi-component T2* analysis of histologically normal Achilles tendons. *Muscles, Ligaments, Tendons J.* 2015;5:58–62. [PubMed: 26261782]
 16. Bouhrara M, Reiter DA, Celik H, et al. Incorporation of rician noise in the analysis of biexponential transverse relaxation in cartilage using a multiple gradient echo sequence at 3 and 7 tesla. *Magn. Reson. Med* 2015;73:352–366 doi: 10.1002/mrm.25111. [PubMed: 24677270]
 17. Boesch CH. in *Superconducting Magnets : Optimization of Corrections and Quantitative Characterization of Magnet / Gradient Systems.* 1991;284:268–284.
 18. Jehenson P, Westphal M, Schuff N. Analytical method for the compensation of eddy-current effects induced by pulsed magnetic field gradients in NMR systems. *J. Magn. Reson* 1990;90:264–278 doi: 10.1016/0022-2364(90)90133-T.
 19. Ahn CB, Cho ZH. Analysis of the eddy-current induced artifacts and the temporal compensation in nuclear magnetic resonance imaging. *IEEE Trans. Med. Imaging* 1991;10:47–52 doi: 10.1109/42.75610. [PubMed: 18222799]
 20. Wysong RE, Madio DP, Lowe IJ. A novel eddy current compensation scheme for pulsed gradient systems. *Magn. Reson. Med* 1994;31:572–575. [PubMed: 8015414]
 21. Wu Y, Chronik B a, Bowen C, Mechefske CK, Rutt BK. Gradient-induced acoustic and magnetic field fluctuations in a 4T whole- body MR imager. *Magn Reson Med* 2000;44:532–6. [PubMed: 11025508]
 22. Foerster BU, Tomasi D, Caparelli EC. Magnetic field shift due to mechanical vibration in functional magnetic resonance imaging. *Magn. Reson. Med* 2005;54:1261–1267 doi: 10.1002/mrm.20695. [PubMed: 16215962]
 23. Jang H, Wiens CN, McMillan AB. Ramped hybrid encoding for improved ultrashort echo time imaging. *Magn. Reson. Med* 2016;76:814–825 doi: 10.1002/mrm.25977. [PubMed: 26381890]

24. Grodzki DM, Jakob PM, Heismann B. Ultrashort echo time imaging using pointwise encoding time reduction with radial acquisition (PETRA). *Magn. Reson. Med* 2012;67:510–518 doi: 10.1002/mrm.23017. [PubMed: 21721039]
25. Robson MD, Gatehouse PD, Bydder M, Bydder GM. Magnetic Resonance: An Introduction to Ultrashort TE (UTE) Imaging. *J. Comput. Assist. Tomogr* 2003;27:825–846 doi: 10.1097/00004728-200311000-00001. [PubMed: 14600447]
26. Emid S, Creyghton JHN. High resolution NMR imaging in solids. *Phys. B+C* 1985;128:81–83 doi: 10.1016/0378-4363(85)90087-7.
27. Jang H, Liu F, Bradshaw T, McMillan AB. Rapid dual-echo ramped hybrid encoding MR-based attenuation correction (dRHE-MRAC) for PET/MR. *Magn. Reson. Med* 2018;79:2912–2922 doi: 10.1002/mrm.26953. [PubMed: 28971513]
28. Han H, MacGregor RP, Balcom BJ. Pure phase encode magnetic field gradient monitor. *J. Magn. Reson* 2009;201:212–217 doi: 10.1016/j.jmr.2009.09.011. [PubMed: 19815435]
29. Jang H, Lu X, Carl M, et al. True phase quantitative susceptibility mapping using continuous single-point imaging: a feasibility study. *Magn. Reson. Med* 2019;81:1907–1914 doi: 10.1002/mrm.27515. [PubMed: 30325058]
30. Horch RA, Wilkens K, Gochberg DF, Does MD. RF coil considerations for short-T2 MRI. *Magn. Reson. Med* 2010;64:1652–1657 doi: 10.1002/mrm.22558. [PubMed: 20665825]
31. Blunk Y, Moffat BA, Kolbe SC, Ordidge RJ, Cleary JO, Johnston LA. Zero-gradient-excitation ramped hybrid encoding (zG RF -RHE) sodium MRI. *Magn. Reson. Med* 2019;81:1172–1180 doi: 10.1002/mrm.27484. [PubMed: 30252156]
32. Jang H, Ma Y, Searleman AC, et al. Inversion recovery UTE based volumetric myelin imaging in human brain using interleaved hybrid encoding. *Magn. Reson. Med* 2020;83:950–961 doi: 10.1002/mrm.27986. [PubMed: 31532032]
33. Wiens CN, Artz NS, Jang H, McMillan AB, Reeder SB. Externally calibrated parallel imaging for 3D multispectral imaging near metallic implants using broadband ultrashort echo time imaging. *Magn. Reson. Med* 2017;77:2303–2309 doi: 10.1002/mrm.26327. [PubMed: 27403613]
34. Wiens CN, Artz NS, Jang H, McMillan AB, Koch KM, Reeder SB. Fully phase-encoded MRI near metallic implants using ultrashort echo times and broadband excitation. *Magn. Reson. Med* 2018;79:2156–2163 doi: 10.1002/mrm.26859. [PubMed: 28833407]
35. Ma Y, Jerban S, Carl M, et al. Imaging of the region of the osteochondral junction (OCJ) using a 3D adiabatic inversion recovery prepared ultrashort echo time cones (3D IR-UTE-cones) sequence at 3 T. *NMR Biomed.* 2019;32:e4080 doi: 10.1002/nbm.4080. [PubMed: 30794338]
36. Jang H, Carl M, Ma Y, et al. Fat suppression for ultrashort echo time imaging using a single-point Dixon method. *NMR Biomed.* 2019:e4069 doi: 10.1002/nbm.4069. [PubMed: 30768813]
37. Jang H, McMillan AB. A rapid and robust gradient measurement technique using dynamic single-point imaging. *Magn. Reson. Med* 2017;78:950–962 doi: 10.1002/mrm.26481. [PubMed: 27699867]
38. Johnson KO, Pipe JG. Convolution kernel design and efficient algorithm for sampling density correction. *Magn. Reson. Med* 2009;61:439–47 doi: 10.1002/mrm.21840. [PubMed: 19165893]
39. Gudbjartsson H, Patz S. The rician distribution of noisy mri data. *Magn. Reson. Med* 1995;34:910–914 doi: 10.1002/mrm.1910340618. [PubMed: 8598820]
40. Block KT, Uecker M, Frahm J. Undersampled radial MRI with multiple coils. Iterative image reconstruction using a total variation constraint. *Magn. Reson. Med* 2007;57:1086–98 doi: 10.1002/mrm.21236. [PubMed: 17534903]
41. Block KT, Uecker M, Frahm J. Model-based iterative reconstruction for radial fast spin-echo MRI. *IEEE Trans. Med. Imaging* 2009;28:1759–69 doi: 10.1109/TMI.2009.2023119. [PubMed: 19502124]
42. Song J, Qing HL. Improving non-cartesian MRI reconstruction through discontinuity subtraction. *Int. J. Biomed. Imaging* 2006;2006:1–9 doi: 10.1155/IJBI/2006/87092.
43. Chan RW, Ramsay EA, Cheung EY, Plewes DB. The influence of radial undersampling schemes on compressed sensing reconstruction in breast MRI. *Magn. Reson. Med* 2012;67:363–377 doi: 10.1002/mrm.23008. [PubMed: 21656558]

44. Feng L, Grimm R, Block KT, et al. Golden-angle radial sparse parallel MRI: Combination of compressed sensing, parallel imaging, and golden-angle radial sampling for fast and flexible dynamic volumetric MRI. *Magn. Reson. Med* 2014;72:707–717 doi: 10.1002/mrm.24980. [PubMed: 24142845]
45. Liu J, Nazaran A, Ma Y, et al. Single- and Bicomponent Analyses of T2* Relaxation in Knee Tendon and Ligament by Using 3D Ultrashort Echo Time Cones (UTE Cones) Magnetic Resonance Imaging. *Biomed Res. Int* 2019;2019:1–9 doi: 10.1155/2019/8597423.
46. Diaz E, Chung CB, Bae WC, et al. Ultrashort echo time spectroscopic imaging (UTESI): an efficient method for quantifying bound and free water. *NMR Biomed.* 2012;25:161–168 doi: 10.1002/nbm.1728. [PubMed: 21766381]
47. Berglund J, Ahlström H, Johansson L, Kullberg J. Two-point dixon method with flexible echo times. *Magn. Reson. Med* 2011;65:994–1004 doi: 10.1002/mrm.22679. [PubMed: 21413063]
48. Wang Y, Li D, Haacke EM, Brown JJ. A three-point dixon method for water and fat separation using 2D and 3D gradient-echo techniques. *J. Magn. Reson. Imaging* 1998;8:703–710 doi: 10.1002/jmri.1880080329. [PubMed: 9626890]
49. Reeder SB, Pineda AR, Wen Z, et al. Iterative decomposition of water and fat with echo asymmetry and least-squares estimation (IDEAL): Application with fast spin-echo imaging. *Magn. Reson. Med* 2005;54:636–644 doi: 10.1002/mrm.20624. [PubMed: 16092103]
50. Lu X, Jerban S, Wan L, et al. Three-dimensional ultrashort echo time imaging with tricomponent analysis for human cortical bone. *Magn. Reson. Med* 2019;82:348–355 doi: 10.1002/mrm.27718. [PubMed: 30847989]
51. Lustig M, Donoho D, Pauly JM. Sparse MRI: The application of compressed sensing for rapid MR imaging. *Magn. Reson. Med* 2007;58:1182–95 doi: 10.1002/mrm.21391. [PubMed: 17969013]
52. Huang C, Bilgin A, Barr T, Altbach MI. T2 relaxometry with indirect echo compensation from highly undersampled data. *Magn. Reson. Med* 2013;70:1026–1037 doi: 10.1002/mrm.24540. [PubMed: 23165796]
53. Zhu B, Liu JZ, Cauley SF, Rosen BR, Rosen MS. Image reconstruction by domain-transform manifold learning. *Nature* 2018;555:487–492 doi: 10.1038/nature25988. [PubMed: 29565357]
54. Shao H, Chang EY, Pauli C, et al. UTE bi-component analysis of T2* relaxation in articular cartilage. *Osteoarthr. Cartil* 2016;24:364–373 doi: 10.1016/j.joca.2015.08.017.

Figure 1. Ramped hybrid encoding (RHE). (a) Pulse sequence diagram for single scan RHE, (b) hybrid encoding scheme, and (c) fat-saturation preparation and the following multi-spoke imaging. In the single scan modification of RHE shown in (a), readout gradient with zero amplitude is used to allow slab selection. Note that in (a), two UTE images are acquired using SPI data obtained at two different TEs. Note that the fat-saturation module is played for each 4 excitations (τ : RF-to-RF timing).

Figure 2.

Single scan RHE images reconstructed with the nominal k-space trajectory and the measured k-space trajectory (a healthy volunteer, 34-year-old male). Note that ringing artifacts (indicated by yellow arrows) are suppressed in images reconstructed with measured k-space trajectory.

Figure 3. Reconstructed single scan RHE images. (a) An ex vivo cadaveric knee joint from 61-year-old male donor and (b) an in vivo knee joint from a healthy volunteer (34-year-old male). The images are affected by streaking artifact due to undersampling (indicated by red arrow) and susceptibility artifact due to B0 inhomogeneity (indicated by green arrow), where the susceptibility artifacts are exacerbated with longer TE. However, these artifacts do not significantly degrade image quality in structures of interest within the knee joint. Note that the ex vivo cadaveric knee shows stronger susceptibility artifacts in the muscle, presumably due to the change in local magnetic susceptibility induced by degenerated tissues with hemorrhage.

Figure 4.

Bicomponent $T2^*$ parameters estimated in a representative ex vivo knee joint (61-year-old male donor) for the patellar tendon, ACL, PCL, and meniscus. Patellar tendon and meniscus show a typical bicomponent UTE $T2^*$ signal decay (in the pixel indicated by a yellow arrow), while ACL and PCL show less obvious bicomponent behavior, presumably due to the degenerated tissue, partial volume effect, or inadequate fat suppression.

Figure 5.

Bicomponent $T2^*$ parameters estimated in a representative in vivo knee joint (34-year-old male healthy volunteer) for the patellar tendon, ACL, PCL, and meniscus. Examples of signal decays in the ACL and PCL (in the pixel indicated by a yellow arrow) exhibit weaker bi-exponential signal decay than tendon and meniscus.

Table 1.

Mean values of estimated bicomponent T2* parameters for tendon, ligament, and meniscus in the ex vivo and the in vivo human knee joints.

Subject	Region	f_s (%)	$T2^*_S$ (ms)	$T2^*_L$ (ms)
<i>ex vivo</i> (n=6)	Patellar Tendon	75.3 ± 4.2	1.7 ± 0.9	26.4 ± 4.2
	ACL	43.3 ± 5.1	6.1 ± 3.7	21.6 ± 4.9
	PCL	50.5 ± 10.8	8.8 ± 3.4	20.6 ± 3.5
	Meniscus	64.1 ± 6.4	7.6 ± 2.6	17.9 ± 1.9
<i>in vivo</i> (n=4)	Patellar Tendon	80.0 ± 2.6	1.2 ± 0.1	21.8 ± 0.8
	ACL	35.8 ± 7.6	4.5 ± 2.3	21.9 ± 2.0
	PCL	39.7 ± 5.8	3.5 ± 0.4	15.5 ± 1.4
	Meniscus	65.5 ± 3.3	3.6 ± 0.5	21.4 ± 2.1