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RESEARCH ARTICLE

Fat suppression for ultrashort echo time imaging using a single-point Dixon method

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Purpose: In ultrashort echo time (UTE) imaging, fat suppression can improve short T_2^* contrast but can also reduce short T_2^* signals. The conventional two-point Dixon (2p-Dixon) method does not perform well due to short T_2^* decay. In this study, we propose a new method to suppress fat for high contrast UTE imaging of short T_2 tissues, utilizing a single-point Dixon (1p-Dixon) method.

Methods: The proposed method utilizes dual-echo UTE imaging, where UTE is followed by the second TE, chosen flexibly. Fat is estimated by applying a 1p-Dixon method to the non-UTE image after correction of phase errors, which is used to suppress fat in the UTE image. In vivo ankle and knee imaging were performed at 3 T to evaluate the proposed method.

Result: It was observed that fat and water signals in tendons were misestimated by the 2p-Dixon method due to signal decay, while the 1p-Dixon method showed reliable fat and water separation not affected by the short T_2^* signal decay. Compared with the conventional chemical shift based fat saturation technique, the 1p-Dixon based approach showed much stronger signal intensities in the Achilles, quadriceps, and patellar tendons, with significantly improved contrast to noise ratios (CNRs) of 11.8 ± 2.2 , 16.0 ± 1.6 , and 26.8 ± 1.3 with the 1p-Dixon method and 0.6 ± 0.2 , 4.6 ± 1.0 , and 17.5 ± 1.4 with regular fat saturation, respectively.

Conclusion: The proposed 1p-Dixon based fat suppression allows more flexible selection of imaging parameters and more accurate fat and water separation over the conventional 2p-Dixon in UTE imaging. Moreover, the proposed method provides much improved CNR for short T_2 tissues over the conventional fat saturation method.

KEYWORDS

Dixon, fat saturation, fat suppression, MSK, single point, tendon, UTE

1 | INTRODUCTION

Recent development in ultrashort echo time (UTE) imaging techniques¹⁻⁵ has enabled MRI to detect signal from tissues with short T_2^* , such as cortical bone, tendons, and ligaments. Many preclinical UTE studies have been conducted in the field of musculoskeletal (MSK) imaging and have

Abbreviations used: 1p-Dixon, single-point Dixon; 2p-Dixon, two-point Dixon; 3p-Dixon, three-point Dixon; BW, bandwidth; CNR, contrast to noise ratio; MSK, musculoskeletal; NuFFT, non-uniform FFT; SNR, signal to noise ratio; UTE, ultrashort echo time.

shown that the resolved UTE signal can help to directly image the morphology of certain short T_2^* tissues and to detect lesions.⁶⁻¹¹ In addition, it has been reported that the UTE signal allows for more reliable multi-component T_2^* fitting to better characterize lesions quantitatively in the short T_2^* tissues.¹¹⁻¹⁴

Fat suppression techniques are widely used in many MRI applications (e.g. MSK and abdominal imaging) to obtain better image contrast with suppressed signal from adipose tissues. This is commonly achieved by using fat saturation or fat separation techniques. In fat saturation, the fat signal is selectively excited and spoiled before imaging. This is the most commonly used method in clinical MRI protocols due to its simplicity of implementation and its compatibility with a variety of MRI sequences. Fat separation methods usually require images at various TEs to utilize phase information over the different TEs to estimate fat and water components.

UTE imaging can also benefit from fat suppression for morphological and quantitative imaging, but unfortunately conventional fat suppression methods are problematic with UTE imaging. Specifically, while UTE imaging is ideal for imaging of short T_2^* tissues, these tissues demonstrate a broad spectrum; thus, chemical shift based fat saturation pulses may undesirably suppress their signals in addition to those of fat. Fat separation based methods, such as the two-point Dixon (2p-Dixon),¹⁵⁻¹⁷ the three-point Dixon (3p-Dixon),¹⁸ and the iterative decomposition of water and fat with echo asymmetry and least-squares estimation (IDEAL)¹⁹ methods require longer acquisition time to obtain multiple images at different TEs. Moreover, the regular 2p- or 3p-Dixon methods do not consider water signal decay, and therefore the estimated fat and water signals can be deviated in UTE imaging of short T_2 tissues.

In this study, we propose a new framework to suppress fat signal with UTE imaging by utilizing a single-point Dixon (1p-Dixon) approach, which allows better UTE contrast for tissues with short T_2^* decay. In the proposed method, a dual-echo UTE imaging scheme is utilized to acquire a UTE image and a gradient recalled echo image in a rapid single scan. The second echo image chosen at a TE with the phase difference between water and fat close to $\pi/2$ or $3\pi/2$ is used for the 1p-Dixon based fat and water separation after phase errors are corrected. We demonstrate the feasibility and efficacy of the proposed fat suppression in in vivo UTE ankle and knee imaging.

2 | METHODS

2.1 | 1p-Dixon

The 1p-Dixon approach has been proposed for rapid fat and water separation in the literature.^{20,21} This method directly decomposes fat and water components from the complex MR signal after correcting for phase error, as shown in the following equation:

$$F = \text{Im}(S e^{-i\phi}) / \sin(\theta) \quad W = \text{Re}(S e^{-i\phi}) - F \cos(\theta) \quad (1)$$

where F and W denote the estimated fat and water signal intensities, S is the complex MR signal as an input, ϕ is a phase error due to phase offset in an RF coil and B_0 field inhomogeneity, and θ is the phase difference between fat and water signals due to chemical shift. As this equation implies, fat and water signals become maximum when $\theta = \pi/2$ or $3\pi/2$, where maximum signal to noise ratio (SNR) can be achieved. On the other hand, this equation becomes ill posed with $\theta = \pi$ or 2π , which cannot be solved. Therefore, when selecting the TE for the 1p-Dixon technique, the TE at which fat and water signals are in phase or out of phase should be avoided, and it is desirable to select the TE near $\theta = \pi/2$ or $3\pi/2$ to achieve better SNR as imaging parameters allow.

To allow accurate estimation of phase difference between water and fat at a prescribed TE, a multi-peak fat model was employed, as introduced in Reference 22, where five main peaks, at 434, 332, 486, -76, and 52 Hz with relative amplitudes of 0.69, 0.13, 0.09, 0.05, and 0.04 in 3 T, were considered in the calculation of θ at the selected T_E as follows:

$$\theta = \angle \left(\sum_{p=1}^5 \alpha_p e^{i2\pi \Delta f_p T_E} \right) \quad (2)$$

where Δf_p denotes the chemical shift frequencies at the main peaks and α_p denotes the corresponding relative amplitudes. An example of the 1p-Dixon fat and water separation can be found in Supporting Figure S1.

2.2 | Dual-echo UTE imaging

Figure 1 shows the pulse sequence diagram used for the proposed fat suppression method based on the 1p-Dixon approach, which utilizes dual-echo UTE imaging. For more efficient coverage of k -space, a 3D-Cones trajectory was used as shown in Figure 1A (although any trajectory can be used in general, such as a radial trajectory).^{23,24} UTE acquisition was performed immediately after the RF coil deadtime (32 μ s in the MR system used for this study), which was followed by acquisition of the second echo, where the TE can be chosen flexibly as shown in Figure 1B. The fat signal in the UTE images was then estimated and suppressed by applying the 1p-Dixon method to the non-UTE image at TE2.

FIGURE 1 Dual-echo UTE 3D-cones imaging. A, pulse sequence diagram; B, k-space trajectory. UTE acquisition (TE1) is followed by acquisition of the second echo (TE2), which is used for a 1p-Dixon based fat/water separation. Since the fat and water separation is performed independently from the UTE image, signal decay between TE1 and TE2 does not affect the result

2.3 | Fat suppression in UTE image

Fat suppression can simply be performed by subtracting the fat signal, as estimated at TE2, from the UTE image. However, due to the short T_2^* decay between the two echoes, there may be some residual signals after subtraction in the adipose tissues. Therefore, better fat suppression can be achieved by globally scaling the fat image. The scaling factor can be set by the user to achieve a desired degree of fat suppression in the UTE image. However, in this study the scaling factor was automatically calculated by solving the following equation, where signal intensities in all voxels from the UTE, estimated water, and fat images were used:

$$[U_1 U_2 \dots U_n] = [S_W S_F] \begin{bmatrix} W_1 & W_2 & \dots & W_n \\ F_1 & F_2 & \dots & F_n \end{bmatrix} \quad (3)$$

where U_i , W_i , and F_i denote signal intensities of UTE, water, and fat images at a pixel index i ; S_W and S_F are two unknowns, which are the scaling factors for water and fat; and n is the number of total pixels in the targeted 3D image. Since this equation is over-determined, linear least square fitting was used to obtain S_F .

2.4 | Experimental setup

To evaluate the proposed method, in vivo experiments were performed on a 3 T scanner (MR750, GE Healthcare, Waukesha, WI, USA) with healthy volunteers: two subjects (36- and 26-year-old males) for the ankle imaging and two subjects (35- and 36-year-old males) for the knee imaging, in accordance with IRB approved protocols.

Imaging parameters for the ankle imaging were as follows: GE receive-only 8-CH ankle coil, slab-selective Shinnar-Le Roux pulse with minimum phase,^{25,26} flip angle of 10° , TE1/TE2/TR = 32 μ s/2.9 ms/24 ms, number of spokes = 10 607, readout bandwidth (BW) = ± 125 kHz, FOV = $120 \times 120 \times 128$ mm³, matrix size = $256 \times 256 \times 32$, scan time = 4 min 16 s. External field map acquisition was performed with matched imaging parameters, except for the following parameters: TE1/TE2 = 32 μ s/2.2 ms, matrix size = $128 \times 128 \times 32$, number of spokes = 6515, and scan time = 2 min 10 s.

Imaging parameters for the knee were as follows: GE transmit/receive 8-CH knee coil, 100 μ s hard pulse with flip angle of 20°, TE1/TE2/TR = 32 μ s/2.7 ms/20 ms, number of spokes = 11 625, readout BW = $\pm$ 125 kHz, FOV = 160 $\times$ 160 $\times$ 96 mm³, matrix size = 256 $\times$ 256 $\times$ 32, scan time = 3 min 53 s. External field map acquisition was performed with matched imaging parameters, except for the following parameters: TE1/TE2 = 32 μ s/2.2 ms, matrix size = 128 $\times$ 128 $\times$ 32, number of spokes = 7859, and scan time = 2 min 38 s.

Additional single-echo UTE imaging was performed for comparison using the conventional fat saturation method using a GE standard clinical fat saturation pulse, with matched imaging parameters except for the following: T_{E1} = 32 μ s, fat saturation pulse width = 16 ms, offset frequency = -440 Hz, BW = 500 Hz, five spokes obtained per single fat saturation pulse, and scan time = 4 min 53 s for ankle imaging and 4 min 35 s for knee imaging. Additional data were acquired for 2p-Dixon, where two images were obtained with TE1 = 32 μ s and TE2 = 1.1 ms or 3.3 ms, with all other parameters matched to the 1p-Dixon acquisition in each experiment. A dual-echo acquisition was performed to obtain two images at TE = 32 μ s and 3.3 ms within a single scan, where the TE of 3.3 ms was the shortest attainable out-of-phase echo with the given spatial resolution. An additional single-echo acquisition was performed with the same imaging parameters to obtain the out-of-phase echo at TE = 1.1 ms.

2.5 | Data processing

Images were reconstructed using offline reconstruction codes based on non-uniform FFT (NuFFT).²⁷ To correct for errors in k -space trajectories, a gradient impulse response function was applied to the nominal k -space trajectories,²⁸ which was input to the NuFFT reconstruction pipeline. After the k -space data in each channel were reconstructed, the reconstructed images in individual channels were combined into a complex image.²⁹

For the field map estimation, the individual phase images at TE = 32 μ s and 2.2 ms were spatially unwrapped using the 3D unwrapping algorithm provided in the FMRIB Software Library (v5.0).³⁰ Based on the two unwrapped phase images, a field map image was estimated, where the generated field map was smoothed with a 3D Gaussian filter with standard deviation of 1 mm. Phase offset, which can be caused by phase evolution during the RF pulse or data acquisition,³¹ was obtained by smoothing a phase map at TE = 32 μ s (the first echo from the dual-echo UTE acquisition for 1p-Dixon), using a 3D averaging filter with kernel width of 2 mm. The scaling factor, S_F , found for fat suppression in the 1p-Dixon process was 2.1 or 2.1 for the ankle imaging, and 1.6 or 1.7 for the knee imaging, respectively. All data processing was performed using MATLAB 2017b (MathWorks, Natick, MA, USA) with the Ubuntu 16.04 (Canonical, London, UK) operation system.

To compare the proposed method and the conventional fat saturation method in terms of image contrast, contrast to noise ratio (CNR) was calculated based on the signal intensity in the tissues of interest (i.e. tendon), with respect to the neighboring fat tissues. The measurement was repeated three times in the adjacent three 2D slices, and the mean and standard error were calculated.

3 | RESULTS

3.1 | In vivo ankle imaging

Figure 2 shows in-phase and out-of-phase images, which are inputs for the conventional 2p-Dixon, and the resultant water and fat images from a 36-year-old healthy volunteer. As can be seen, the signal decay between UTE and TE2 is interpreted as coexistence of fat and water, partially nulling each other's signal. This causes a misestimated, residual water signal in bone marrow (green arrows) and fat signal in the Achilles tendon (red arrows). As a result, water estimated in the Achilles tendon shows decreased signal intensity (yellow arrows), which does not accurately reflect the actual water proton density in the UTE image. Figure 2B shows the results with smaller echo spacing (TE1 = 32 μ s and TE2 = 1.1 ms), which demonstrates less residual water and fat signal (green and red arrows) due to smaller signal decay between the two echoes. However, the estimated water signal in the Achilles tendon still shows diminished intensity compared with the UTE image. Moreover, note that two separate scans were required to obtain images with the small echo spacing (1.1 ms), leading to doubled scan time.

Figure 3A shows the UTE image, the second TE image used for 1p-Dixon, the estimated water and fat images, the resultant fat-suppressed UTE image, and the UTE image with conventional fat saturation as a comparison. The proposed 1p-Dixon based method provides strong signal for the Achilles tendon, while the UTE image with conventional fat saturation exhibits strongly decreased signal intensity in the Achilles tendon (indicated by red arrows in the magnified images in Figure 3B). For the two subjects, the proposed 1p-Dixon was able to suppress fat and visualize the Achilles tendon well in the UTE image. The measured CNR for the Achilles tendon with respect to bone marrow was 11.8 ± 2.2 or 6.5 ± 0.6 in the proposed 1p-Dixon method and 0.6 ± 0.2 or 0.8 ± 0.5 in the conventional fat saturation for the two subjects, respectively.

3.2 | In vivo knee imaging

Figure 4 shows comparisons between fat and water separations in the knee performed using 2p-Dixon (A) and 1p-Dixon (B) methods, respectively, from a 35-year-old healthy volunteer. As in the ankle imaging experiment, the short T_2^* decay is misinterpreted as the coexistence of fat and water

FIGURE 2 The 2p-Dixon technique for UTE ankle imaging. The 2p-Dixon technique applied for UTE imaging with dual-echo acquisition (TE = 32 μ s/3.3 ms) (A) and echo shifted acquisition (TE = 32 μ s/1.1 ms) (B). In the dual-echo acquisition, the shortest attainable echo spacing is limited; therefore, an out-of-phase image was obtained at TE = 3.3 ms. In the echo shifted acquisition, UTE images and out-of-phase images at TE = 1.1 ms were obtained in two separate scans. Note that water and fat are misestimated and residual signals are shown due to the signal decay between UTE and TE2, as indicated by green and red arrows. 2p-Dixon with larger echo spacing in (A) shows stronger residual signals due to the larger signal decay. Note that the water signal in the Achilles tendon is underestimated due to the signal decay (yellow arrow)

signals. Therefore, the fat signal is misestimated in tissues containing short T_2^* components, as clearly indicated in the magnified images in Figure 4 C; meanwhile, the 1p-Dixon based method does not show such misestimated fat signal.

Figure 5A shows the UTE image, the second TE image used for 1p-Dixon, the estimated water and fat images, the resultant fat-suppressed UTE image, and the UTE image with conventional fat saturation as a comparison. As shown in the magnified images in Figure 5B, the proposed 1p-Dixon based approach shows strong signal intensity in the quadriceps and patellar tendons, with less spatial variation in the fat suppressed UTE image, while the UTE image with fat saturation exhibits gradually varying signals, as indicated by red arrows. The pattern of signal variation in the UTE image with fat saturation shows high similarity to the estimated water images at TE (2.7 ms), where most short T_2^* signals have already decayed, as indicated by green arrows. This observation implies that broad spectrum short T_2^* components were partially affected by the fat saturation pulse. For the two subjects, the proposed 1p-Dixon was able to suppress fat and clearly visualize the tendons in the UTE image. The measured CNR for the quadriceps tendon with respect to the adjacent adipose tissue was 16.0 ± 1.6 or 10.9 ± 2.1 in the proposed 1p-Dixon

FIGURE 3 The 1p-Dixon based fat suppression for in vivo UTE ankle imaging. A, the UTE image, the second echo, the estimated water image and fat image, the UTE image with the proposed 1-pt Dixon fat suppression, and the UTE image with the conventional fat saturation technique; B, the corresponding magnified images in the ROI A to F. As indicated by red arrows in B, the fat saturation pulse affects signals from the short T_2^* components (i.e. The Achilles tendon), where signals appear much lower than UTE signals without fat saturation. Meanwhile, the fat-suppressed UTE image with the proposed 1p-Dixon approach shows similar signal intensities to the original UTE image

FIGURE 4 Fat and water separation in in vivo knee imaging. A, the conventional 2p-Dixon based fat and water separation performed with echo shifted acquisition ($TE = 32 \mu s / 1.1 \text{ ms}$); B, the proposed 1p-Dixon based fat and water separation with a non-UTE image at $TE = 2.7 \text{ ms}$; C, magnified views of the patellar tendon (ROI A and C) and bone (ROI B and D). As shown in C, the 1p-Dixon approach was not affected by UTE signal decay since it was performed with a single non-UTE image at $T_E = 2.7 \text{ ms}$, allowing more robust fat and water separation for short T_2^* tissues

method and 4.6 ± 1.0 or 2.6 ± 1.0 in the conventional fat saturation for the two subjects, respectively. The measured CNR for the patellar tendon with respect to the infrapatellar fat pad was 26.8 ± 1.3 or 24.1 ± 4.1 in the proposed 1p-Dixon method and 17.5 ± 1.0 or 9.9 ± 1.9 in the conventional fat saturation for the two subjects, respectively.

4 | DISCUSSION

The proposed 1p-Dixon based fat suppression method has high flexibility with regards to the TE selected for the second echo, which is advantageous over the conventional 2p-Dixon method, where echo spacing is limited by imaging parameters such as spatial resolution, FOV, and tolerance to the error in echo spacing in the fat and water separation algorithm. Moreover, we have shown that the 1p-Dixon based method is not affected by short T_2^* decay since only one non-UTE image is used for fat and water estimation. By comparison, the 2p-Dixon method exhibited misestimated, residual fat and water signals affected by the fast signal decay between the two echoes.

Using in vivo experiments, we have shown that the proposed method preserves water signals from the short T_2^* components in UTE images well, while a fat saturation pulse suppresses short T_2^* signals. This observation can be explained by the fact that short T_2^* tissues have a broad spectrum that can be partially affected by a chemical shift based fat saturation pulse. This hypothesis was also indirectly verified by comparing

FIGURE 5 The 1p-Dixon based fat suppression for in vivo UTE knee imaging. A, the UTE image, the second echo, the estimated water image and fat image, the UTE image with the proposed 1p-Dixon fat suppression, and the UTE image with the conventional fat saturation technique; B, the corresponding magnified images in each ROI (A-L). Compared with the conventional fat saturation technique, the 1p-Dixon based approach shows stronger signal intensities in the patellar and quadriceps tendons with less spatial variation than UTE images with regular fat saturation (red arrows). The pattern of signal variation in UTE images with regular fat saturation shows high similarity to that of the water images at TE2 (at 2.7 ms), where most short T_2^* signals have already decayed to near zero; therefore, the broad spectrum short T_2^* components are partially affected by the fat saturation pulse (green arrows)

the water image at TE2 with the fat saturated UTE image in Figure 5, where the pattern of signal variation showed high similarity to the pattern in the water images at a later echo, where most of short T_2^* signals had already decayed to near zero.

Another possible approach to achieving better contrast for UTE imaging is echo subtraction, where an in-phase, later echo image is subtracted from a UTE image. This approach allows good contrast for the tissues with short T_2^* , but all tissues with long T_2^* are suppressed along with fat, which is demonstrated in Supporting Figure S2. Moreover, the short T_2^* signal detected in the subtraction image is not a pure UTE signal, but a difference between two echoes, which cannot be used for quantitative UTE imaging. On the other hand, the proposed 1p-Dixon based approach selectively suppresses fat signal in the UTE image, which does not affect UTE signal; thus, the detected signal can be used for quantitative imaging.

In the knee experiment better contrast can be obtained by using the Ernst angle for tissues of interest. For example, a flip angle of 24° for bone ($T_1 \sim 220$ ms), 13° for quadriceps tendon ($T_1 \sim 800$ ms), and 11° for cartilage ($T_1 \sim 1100$ ms) will increase the signal intensity in the tissues.^{32,33} However, in this study, we focused on showing the comparison between conventional fat saturation methods and our proposed method with all other parameters matched within the experiment, where we were able to see diminished tendon signal in UTE imaging with fat saturation. The patellar tendon in Figure 5 shows higher signal than that in Figure 3 due to the magic angle effect, which increases the T_2^* (narrower spectra); thus, the fat saturation pulse has less overlap with the spectrum of the patellar tendon, and therefore less signal attenuation (the longer the T_2^* , the lower the signal attenuation by the fat saturation pulse). The Achilles tendon is oriented parallel to the B_0 field, thus demonstrating a significantly shorter T_2^* than the patellar tendon, with more resultant signal attenuation due to the fat saturation pulse.

The proposed method is very simple and flexible, and therefore has the potential to be incorporated into various types of UTE imaging scheme, such as UTE- T_2^* mapping,^{12,34} UTE- T_1 mapping,^{35,36} and UTE-magnetization transfer imaging.^{37,38} All of these quantitative UTE techniques may benefit from the proposed 1p-Dixon method, which is expected to improve quantification by removing interference from the fat signal. Another advantage of this method is that the estimated fat and water reflect actual transverse magnetization at TE2, which can be utilized for quantitative imaging, such as T_2^* mapping, by using the fat-suppressed UTE signal and the water signal at TE2.

One limitation of the proposed method is that the accuracy of fat and water separation strongly depends on the accuracy of estimation of a field map. In this study, we performed additional low resolution dual-echo imaging to achieve an accurate field map estimation. Although this imposes additional scan time, it is relatively short (~ 2 min) and this scan can be further optimized in the future. In addition, more advanced methods for field map estimation without the external acquisition are also available,^{15,16,20} and can further reduce the total scan time and remove

possible issues caused by inter-scan motion. Another limitation of this method is the employment of a fat scaling factor to compensate for the multi-peak T_2^* decay of fat and other systematic errors, such as discrepancy between the prescribed TE and the effective TE, which is associated with non-zero iso-delay determined by the RF pulse (shape, phase, and duration) and non-zero readout duration depending on the encoding scheme and k -space trajectory. We have shown that this approach works for the applications where accurate estimation of fat fraction is not necessary, such as fat suppression in UTE images. However, these issues must be addressed to achieve truly quantitative fat estimation with the proposed method, and this will be further investigated in our future studies.

5 | CONCLUSIONS

In this study, we demonstrated the feasibility and efficacy of a new framework for fat suppression with UTE imaging utilizing a 1p-Dixon approach. This approach has the advantage of flexible echo spacing and demonstrates improved fat estimation over the conventional 2p-Dixon method. Moreover, the proposed method preserves short T_2^* signals, which may be greatly suppressed with conventional fat saturation methods. The new 1p-Dixon technique provides much improved CNR for short T_2^* tissues, which is beneficial for morphological and quantitative UTE imaging.

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SUPPORTING INFORMATION

Additional supporting information may be found online in the Supporting Information section at the end of the article.

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