UTE Imaging With Simultaneous Water and Fat Signal Suppression Using An Efficient Multi-Shot Inversion Recovery Preparation

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Introduction: Direct imaging of the very short T2 tissues frequently encountered in the musculoskeletal system using MRI requires specialized pulse sequences with very short echo times (TEs). To address this challenge, ultrashort TE (UTE) sequences typically begin data acquisition as soon as possible after the RF excitation with k-space acquired in a center out fashion [1]. Magnetization inversion is an important tool used in conjunction with UTE MRI to generate contrast and selectively eliminate signals in the image, such as those from fat or water in muscle. Traditionally, one k-space spoke is acquired after each application of a single IR or dual IR pulse [2]. In this work we explore the use of several UTE k-space spokes after the application of a single adiabatic IR pulse. Theoretical calculations, simulations, and in-vivo experiments were used to optimize the sequence parameters such as TI and flip angle to obtain a time efficient acquisition.

Theory: The pulse sequence is shown in Fig.1. The adiabatic IR pulse is repeated every TR period, during which N separate k-space spokes are obtained during short time intervals τ (see Fig.1B for an example with N=5). The TI time is defined as the time from the center of the IR pulse to the center spoke and is set to approximately the null-point of fat signals. Tracking the recursive steady state z-magnetization, an equation can be derived for the MR signal (S_i) excited by the i^{th} excitation pulse with excitation flip angle θ. Not all tissue signals are inverted by the IR pulse. This can be accounted for by allowing a variable inversion efficiency Q, with a range of Q=-1 signifying full inversion to Q=1 signifying no disturbance to the z-magnetization. Very short T2 tissues typically get saturated by the IR pulse (Q=0) [3]. To simplify the signal equations, T2* decay during the excitation RF pules was not taken into account in the following equations:

$$S_i = \rho_i \left(\frac{\left[(1-E_w)E_pQ + (1-E_p) + (1-E_\tau)E_pE_wQ\sum_{j=1}^{N-1}(\cos(\theta))^j {E_\tau}^{(j-1)}\right]}{1-(\cos(\theta))^N {E_\tau}^{(N-1)}E_pE_wQ}(E_\tau\cos(\theta))^{(i-1)} + (1-E_\tau)\sum_{k=1}^{i-1}(E_\tau\cos(\theta))^{(k-1)}\right)\sin(\theta) \quad [1]$$

Equation 1 can be simplified for the case of short T2 tissues (Q=0), so that:

$$S_i = \rho_i\left((1-E_p)(E_\tau\cos(\theta))^{(i-1)} + (1-E_\tau)\sum_{k=1}^{i-1}(E_\tau\cos(\theta))^{(k-1)}\right)\sin(\theta) \quad [2]$$

Using the following definitions: $E_w \equiv e^{-\frac{(TR-TI-\tau(N-1)/2)}{T1}}$, $E_p \equiv e^{-\frac{(TI-\tau(N-1)/2)}{T1}}$, $E_\tau \equiv e^{-\frac{\tau}{T1}}$

Materials and Methods: Volunteer imaging was performed using an 8-channel knee coil using a 3T GE HDxt clinical MR scanner on a healthy male volunteer (age 70). The 3D Cones sequence employed a unique data sampling trajectory scheme that samples MRI data along twisting paths that follow evenly spaced cone surfaces in 3D [4]. It samples data starting from the center of k-space and twisting outwards from there with data acquisition starting as soon as possible after the RF excitation. To minimize scan time (to less than 5 minutes), anisotropic FOV encoding together with slab-selection was used to excite and encode only the region of interest. IR preparation was performed using 8.6ms Silver-Hoult (SH) adiabatic IR pulses [5,6] in order to minimize inversion sensitivity to B1 and B0 inhomogeneity. Relevant sequence parameters were TR=50ms, N=5 and TI=22ms, with θ=10º-80º.

Results: Fig.2 shows the theoretical signal curves using Eq.[1]. These generally agree well with the experimentally measured ROI curves shown in Fig.3 (overall signal amplitudes are arbitrary and depend on the unknown underlying spin density ρ). The deviation of the bone signal can be attributed to the rapid signal decay of the transverse magnetization of bone, which has a short T2* (300-400μs) and is comparable to the constant excitation RF duration (300μs) used in our experiments. As was shown in [7-9] this leads to higher optimum flip angles. Finally, Fig.4 shows the result of the volunteer scan at θ=50º, which shows excellent marrow and muscle signal suppression with high cortical bone SNR.

Conclusion: IR prepared 3D UTE imaging can be used effectively to suppress long T2 signals in UTE imaging. The theoretical equations can be used to determine the sequence parameters such as TI, and excitation flip angle to optimize long T2 suppression and short T2 SNR. In this optimization, a compromise has to be made between high SNR for cortical bone (which is optimized at higher flip angles) and sufficient long T2 suppression (which was best suppressed at intermediate flip angles). The theoretical equations presented here can also be extended to analyze the signals behavior of other types of preparation pulses, such as fat saturation, followed by multi-spoke UTE imaging.

References: [1] Rahmer et al. MRM 55:1075–1082 (2006), [2] Du et al, MRI 29 (2011) 470–482, [3] Li et al, MRM;68(3):680 (2012), [4] Gurney et al. MRM 55:575-582 (2006), [5] Conolly et al, JMR 78,440-458 (1988), [6] Silver et al, JMR 59, 347-351 (1984), [7] Carl et al MRM, 2010. 64: 481-490, [8] Springer et al MRM 206:88–96 (2010), [9] Carl et al MRI 32-8: 1006–1011 (2014)

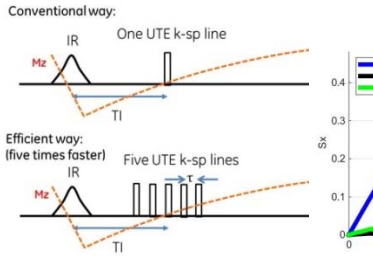

Fig.1: Schematic diagram of IR preparation used with the 3D UTE pulse sequence.

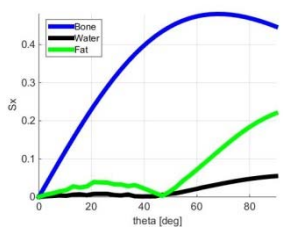

Fig.2: Theoretical signals [a.u.] for cortical bone, marrow fat and water in muscle.

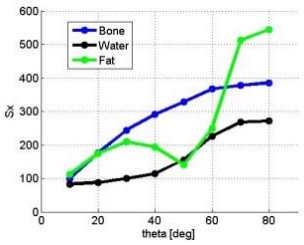

Fig.3: Experimental ROI signals [a.u.] for cortical bone, marrow fat and water in muscle.

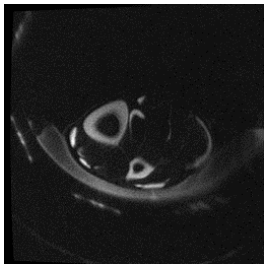

Fig.4: Representative axial IR prepped UTE image of cortical bone of the tibia and fibula.