Published in final edited form as:

Magn Reson Med. 2015 June ; 73(6): 2204–2211. doi:10.1002/mrm.25361.

Simple and Robust Saturation-Based Slice Selection for Ultra-Short Echo Time MRI

Kevin D. Harkins¹, R. Adam Horch^{1,2}, and Mark D. Does^{1,2,3,4}

¹Institute of Imaging Science, Vanderbilt University ²Biomedical Engineering, Vanderbilt University ³Radiology and Radiological Sciences, Vanderbilt University ⁴Electrical Engineering, Vanderbilt University

Abstract

PURPOSE—To present a new method for localizing signal within a 2D slice suitable for ultrashort echo time (UTE) imaging, called saturation-based UTE (sat-UTE). The new method digitally subtracts two acquisitions that are non-selectively excited with and without selective saturation of the slice of interest.

METHODS—Sat-UTE was compared with half-pulse and double-half pulse excited UTE within phantoms, as well as 3D-UTE within ex-vivo femur and in-vivo tibia. Numerical simulations were also used to quantify the effects of slice profile broadening and signal component amplitudes for quantitative UTE.

RESULTS—Sat-UTE is robust to suppress out-of-slice signal, and produces short T_2 signal decay curves comparable to 3D-UTE, but has a lower SNR efficiency compared to other slice-selective methods.

CONCLUSIONS—The proposed method is useful for fast, quantitative evaluation of short T_2 signals, and is insensitive to gradient performance.

Keywords

ultrashort echo time; UTE; slice selective; saturation; 2D

INTRODUCTION

Ultra-short echo time (UTE) magnetic resonance imaging (MRI) methods are designed to acquire signals with transverse relaxation (T_2) time-constants that are too short to be observed with conventional pulse sequences. In order to observe such short T_2 signals, radio frequency (RF) excitation is typically achieved with a hard (i.e., non-selective) waveform, which provides the minimum RF pulse duration and does not require gradient refocusing prior to signal acquisition (1–3). This approach works well for minimizing the T_2 -weighting

of the acquired signal, but restricts the acquisition to be 3D since no spatial selectivity is imparted in the excitation.

For 2D-UTE, the current standard for slice-selective excitation is through variable rate selective excitation (VERSE) RF half-pulses (HP), where full slice selectivity is provided by adding the complex signals acquired in response to two half-pulses. Each pulse alone excites signal within and beyond the desired slice profile, but when combined, out-of-slice signal is theoretically canceled (4–6). This approach does not require gradient refocusing prior to acquisition, but the HP excitation waveforms are generally longer in duration than a hard pulse, resulting in some loss of sensitivity to short- T_2 signals.

While several groups have had success with HP-excited 2D-UTE, the method is known to be sensitive to imperfections in the slice-selective gradient field, and it can be challenging to achieve accurate and echo-time independent 2D slice selection on some clinical systems. Small inconsistencies between the prescribed and actual gradient fields, potentially caused by eddy currents, propagation delay, or the transient response of the gradient amplifier can result in relatively large out of slice signal that is not canceled when the two HP acquisitions are combined (6,7). This is a particular problem when utilizing UTE-MRI to quantify short T_2 signal fraction or relaxation time-constant, such as those being developed to characterize the health of cortical bone (8,9). The HP performance can be improved with comprehensive eddy current compensation (10,11) or out-of-slice signal suppression (12), but on some systems the effective use of HP excitation for 2D-UTE requires time-consuming gradient and/or RF waveform characterization and will always result in some additional T_2 weighting compared to non-selective excitation for 3D-UTE.

One simple strategy to reduce the sensitivity of HP excitation to gradient waveform imperfections is through the extension to the double-half pulse (DHP) method (7). With this approach, two half-pulses are run consecutively within a single TR period. Together, the two half pulses act like one fully refocused slice selection for long T_2 signals, while still looking like a regular half pulse for short T_2 signals. Thus, the slice selection of long T_2 signals does not require phase cancellation of signals from two acquisitions, although good gradient waveform characterization is still needed to appropriately align the RF pulses with the k-space trajectory and intermediate to short T_2 signals behave as they do in the HP method.

This paper presents an alternate method of slice-selection that is suitable for UTEMRI and is comparatively insensitive to the slice-select gradient field waveform. This method uses a radiofrequency (RF) saturation pulse to define the slice of interest, as is referred to herein as sat-UTE. An simplified pulse sequence diagram for sat-UTE is shown at the top of Fig 1. Like the HP and DHP methods, this approach is comprised of two acquisitions that are digitally combined to cancel out-of-slice signal. Both acquisitions result from a non-selective hard RF pulse, which excites all signals within the volume of the coil. The two acquisitions are distinguished by the absence ($\phi = 0$) or presence ($\phi \neq 0$) of a slice selective saturation pulse. Any signal excited by the slice-selective pulse is dephased by the G_{slice} waveform prior to hard pulse excitation. In-slice signal is then calculated as the difference between the two acquired signals, and gradient-field imperfections are canceled since identical gradient waveforms are used for both acquisitions.

METHODS

Sat-UTE

Neglecting the time required for the RF pulses and assuming transverse magnetization is completely spoiled between RF pulses, the signal equation from the slice of interest in sat-UTE is given by

$$S_{\text{sat-UTE}} = \sin(\alpha) e^{-TE/T_2^*} \left[\frac{1 - e^{-(T_D+T_S)/T_1}}{1 - \cos(\alpha) e^{-(T_D+T_S)/T_1}} - \frac{1 - e^{-T_S/T_1} + (1 - e^{-T_D/T_1}) e^{-T_S/T_1} \cos(\phi)}{1 - \cos(\phi) \cos(\alpha) e^{-(T_D+T_S)/T_1}} \right] \quad [1]$$

where α is the flip angle of the excitation pulse, ϕ is the flip angle of the saturation pulse, T_S is the time between the saturation and the excitation pulses, and T_D is the time between one excitation pulse and the next saturation pulse, such that the repetition time (TR) = $T_S + T_D$. The chosen value of ϕ will ultimately determine the contrast and SNR of the sat-UTE image, with a 90° pulse resulting in complete saturation and the peak theoretical SNR. However, in practice, ϕ may be limited by B_1 and gradient limits of the MRI system.

Note that this signal equation does not account for two additional tissue-dependent factors that will broaden the slice profile and therefore alter the signal amplitude. First, the response of magnetization to the slice-selective saturation pulse will be T_2 -dependent, resulting in broadened slice profile for shorter T_2 s (6). A similar effect occurs with HP- and DHP-UTE. Second, this same effect results in saturation of macromolecular proton magnetization ($T_2 \sim 10 \mu\text{s}$) over a broadened slice profile. Through dipole-dipole coupling or chemical exchange, this saturation effect can be transferred to the protons with greater T_2 (long enough to be observed in the UTE acquisition), resulting in a further broadened slice profile for the sat-UTE image. This magnetization transfer effect will depend on tissue characteristics as well as RF pulse amplitude and repetition rate (i.e., TR). Thus, depending on acquisition parameters, the signal amplitude in sat-UTE may be greater than predicted by Eq [1] in tissues with particularly short T_2 and large macromolecular content. The T_2 -induced slice profile broadening also occurs in HP and DHP, but the magnetization transfer induced broadening does not.

Half-pulse and double-half-pulse RF excitation

HP (4) and DHP (7) RF excitation schemes for 2D-UTE were tested for comparison. To minimize the nominal echo time, VERSE excitation schemes were used to continue excitation during the gradient ramp down (13). Slice selective gradient waveforms for HP and DHP excitation were generated to minimize the duration of a single half-pulse given constraints of gradient amplitude (400 mT/m), maximum slew rate (1600 mT/m/ms), and gradient acceleration (30,000 mT/m/ms²). Bipolar gradient lobes were used to minimize long-term eddy currents within the sequence. The DHP excitation scheme was generated through sequential excitation by two half-pulses of opposite gradient polarity, and two passes were performed for HP- and DHP-excitation with opposite gradient polarity to cancel out of slice signal. HP- and DHP-RF waveforms were generated with gradient-induced B_0 correction based upon gradient waveforms measured using a variant of the method by Duyn et al. (14).

Phantom and cadaver bone imaging

Two-dimensional UTE studies were performed using sat-, HP-, and DHP-UTE methods on a 9.4 Tesla, 31-cm horizontal bore magnet with Varian/Agilent DirectDrive console (Santa Clara, CA) using a 38-mm Litz quadrature coil (Doty Scientific, Columbia, SC). A phantom of four tubes was constructed to test for out-of-slice signal contamination imparted by various 2D-UTE excitation schemes. Two tubes were filled with MnCl_2 solutions with $T_2 \sim 100$ ms, while another two tubes were filled with 4% agarose gel ($T_2 \sim 30$ ms). All four tubes were oriented orthogonal to the prescribed slice plane—one set of tubes (one MnCl_2 and one agarose) traversed the slice, while the other two tubes were placed about 4 mm outside the slice. Imaging was also performed a human cadaver femur midshaft obtained from the Vanderbilt Anatomical Donor Program Phantom.

Pulse sequence details for all methods were as follows. A 184- μs gauss-shaped excitation VERSE (13) half-pulse ($\alpha = 20^\circ$) was used for HP, two such pulses were used for DHP ($\alpha = 10^\circ$, each), and sat-UTE used an 80- μs hard excitation ($\alpha = 20^\circ$) and a 250- μs gauss-shaped saturation pulse ($\phi = 90^\circ$). For all excitation schemes, k-space data were collected along 128 radial spokes with a receiver bandwidth of 250 kHz, and an acquisition window of 0.32 ms. Images were collected at 20 echo times (TE, defined as the time between the end of the excitation pulse and the start of signal acquisition) logarithmically spaced between 0.1 and 2 ms, and TR = 20 ms. For all three sequences, signals were acquired in two conditions (with/without saturation for sat-UTE and with gradient polarity reversal for HP/DHP). All radial spokes were acquired with 4 averaged excitations and were run in a block with 128 dummy scans run before the start of each condition block, resulting in a scan time ≈ 25 s per image. For comparison, 3D-UTE images were acquired using 4096 radial spokes (15), 20° hard pulse excitation, TR = 20 ms, and the same readout parameters as for 2D acquisitions. To measure the slice profile of sat-UTE, saturation RF and gradient pulses were prepended to the 3D-UTE excitation, and 3D images were collected with $\phi = 0^\circ$ and again with $\phi = 90^\circ$. These two images were subtracted to create 3D resolved images of the sat-UTE saturation profile.

To reconstruct images, raw signals were density compensated (16–18) and re-gridded using the measured k-space readout trajectory (14) to a Cartesian grid with a Kaiser-Bessel kernel (19). Re-gridded data were reconstructed with an inverse Fourier transform onto a 40 mm length field of view with 64 encoding in each imaging direction.

In vivo human imaging

Example 3D-UTE and 2D sat-UTE images were collected from the lower leg of a healthy volunteer on a 3T Philips Achieva system with a Philips 8 channel knee coil. Human studies were approved by the Vanderbilt Institutional Review Board. Sat-UTE images were collected from a prescribed 6-mm thick slice with a $20 \text{ cm} \times 20 \text{ cm}$ field of view and 1.39 mm nominal in-plane resolution using the following parameters: TR = 24 ms, TE = 0.08 ms, $\alpha = 15^\circ$ with a 100- μs hard pulse, and $\phi = 30^\circ$ with a 0.750 ms sinc-gauss pulse, and was peak B_1 limited. K-space was sampled with 132 points acquired with a 256 kHz bandwidth along each of 264 radial views. Each condition was acquired with 528 dummy scans and 6 averaged excitations, resulting in a scan time ≈ 101 s. In order to demonstrate magnetization

transfer effects, a second sat-UTE image was collected with $TR = 6$ ms and $\alpha = 7^\circ$, resulting in ≈ 25 s scan time. Images were reconstructed as described above except an assumed ideal read-out k-space trajectory was assumed. The 3D-UTE data were collected with the same geometry and read-out protocols, with 34848 radial views, and a scan time $\approx 3 \frac{1}{2}$ min. The 3D images were reconstructed using built-in Philips code.

Bloch simulations

Bloch equation simulations were used to evaluate slice profiles and signal to noise ratio (SNR) as a function of T_2 for HP- and sat-UTE excitation schemes. Sat-UTE simulations were run using the same sequence parameters as used for the in-vivo studies, above. For HP-UTE, excitation was simulated using 300- μ s duration $\alpha = 7^\circ$ VERSE half-pulses, which were approximately the shortest duration attainable for typical gradient and RF limitations of a 3 T clinical scanner ($G_{\max} = 30$ mT/m, $T_{\text{rise}} = 0.2$ ms, $B_{1\max} = 15$ μ T). Signal amplitudes were evaluated at the end of RF excitation (a nominal TE = 0 ms), and the effect of T_2 induced slice profile blurring was assessed by measuring the profile full width half max (FWHM).

RESULTS

Figure 2 shows HP-, DHP-, sat-, and 3D-UTE images acquired at TE = 0.1 ms and 1.0 ms. Corresponding plots of signal vs. TE within the four tubes are shown in Fig 3. While 3D-UTE images show no signal from the tubes that are placed outside of the desired slice, HP-UTE images contain significant out-of-slice signal. Degradation of the slice profile is caused by gradient field imperfections and misalignment of the two half pulses, leading to discrepancies in the phase of out-of-slice signal between the two passes. Transient gradient field imperfections cause out-of-slice signal present in the two passes to come in and out of phase, and large variations in observed image intensity with echo time (Fig 3). The out-of-slice signal and signal variations with echo time are markedly lower in DHP demonstrating reduced effect of gradient waveform imperfection on selection of long T_2 signals. However some amount out-of-slice signal contamination remains due to imperfect alignment of the two half-pulses in excitation k-space. Note that these results represent the best efforts of the authors to optimize the gradient performance of the HP and DHP scans on one system. The exact nature of the HP and DHP signals vs. echo time will depend on hardware-specific characteristics, including the gradient amplifier and eddy currents, which will vary from system to system.

Since sat-UTE does not depend upon phase reversal to cancel out-of-slice signals, it is not sensitive to gradient waveform performance, and, as seen in Fig 2, out-of-slice signal is suppressed to the noise floor. Likewise, Fig 3 demonstrates smooth signal decay from the phantoms within the image slice, indicating an echo-time independent slice profile. Also note that for this phantom and these sequence parameters, there is no apparent difference in the efficacy of the sat-UTE method for the MnCl_2 and agarose phantoms, indicating no significant magnetization transfer effects.

Figure 4 shows sat-UTE and 3D-UTE images of the ex-vivo femur at TE = 0.1 ms along with magnitude differences images (subtraction of the images acquired at TE = 0.56 ms

from the images acquired at $TE = 0.1$ ms). Also shown in Fig 4 are plots of signal vs. TE for ROIs drawn in the cortical bone and marrow, red and green respectively, where the signals are normalized to the marrow signal at minimum TE . The oscillatory behavior of the signal in marrow and bone is consistent with the presence fat with chemical shift ≈ 1400 Hz (3.5 ppm at 9.4T). The sat- and 3D-UTE signal decays closely correspond in the marrow, while cortical bone signal was greater relative to marrow in sat-UTE compared to 3D-UTE. The signal amplitude discrepancy can be explained by T_2 and MT induced slice profile effects presented in Figs 5-7, below.

The sat-UTE slice profiles measured with 3D-UTE are shown in Fig 5, where the profiles were normalized to the maximum intensity. The slice profile in the cortical bone is broadened relative to that in the marrow due to some combination of MT effects and short T_2 blurring during excitation and/or readout. To illustrate the effect of T_2 , the results of the Bloch simulations of slice profiles are shown in Fig 6. The FWHM of simulated slice profiles, normalized to that with a long T_2 , are shown in left panel of Fig 6. At gradient and RF limits of a typical clinical scanner, HP- and sat-UTE excitations exhibit a similar slice profile FWHM for T_2 s greater than about 500 μ s. Below 500 μ s, the HP-excited slice profile is significantly broader due the additional time required for RF excitation on the gradient ramp down; however, measurement of such short T_2 s with any UTE sequence is difficult due to significant signal decay during readout.

The right panel of Fig 6 shows the integrated signal through the slice (solid) and peak profile amplitude (dashed) across a range of T_2 s for each excitation scheme. Since the in-slice signal from sat-UTE is derived from only one of the two excitations, its long T_2 SNR is theoretically half of that of the HP-UTE signal, which utilizes the addition of signals from two excitations. The asymptote of the peak profile amplitude for sat-UTE on the far right of this panel is slightly less than 0.5 due to incomplete saturation by the $\phi = 30^\circ$ saturation pulses used in the simulation (see Eq [1]). Moving to the left, lower values of T_2 result in a sat-UTE signal increase due to short T_2 related broadening of the slice profile. This signal increase persists until $T_2 \approx 100$ -200 μ s, below which signal loss due to T_2 relaxation on the transverse plane during the excitation pulse dominates any slice profile broadening caused by such a low T_2 . This biphasic response of total signal vs. T_2 is much less pronounced for HP-UTE because there is an additional effective T_2 - weighting effect during the 300 μ s duration HP excitation compared with the 100 μ s hard pulse excitation used in the sat-UTE simulations.

Example sat-UTE (with $TR = 6$ and 24 ms) images and a closely corresponding slice from a 3D-UTE ($TR = 6$ ms) acquisition from a human tibia in vivo are shown in Fig 7. SNRs from muscle for these three images were 38, 48 and 134, respectively, which are close in relative values to theoretical predictions considering differences in voxel size, number of acquired signals, TR , and α . Comparing the sat-UTE image at $TR = 24$ ms with the image from the 3D-UTE acquisition demonstrate similar image quality, with similar contrast and resolution of muscle, fat, and bone. The sat-UTE image acquired with shorter TR demonstrates a reduction in contrast between the cortical bone & muscle compared to the surrounding fat & bone marrow. This is presumably due to a greater slice profile in cortical bone and muscle due to magnetization transfer effects, which were made greater by more rapid RF saturation

pulses. In general, this effect will depend on complex magnetization transfer characteristics of the tissue as well as sequence timings and RF pulse power, but the results for sat-UTE image acquired at $TR = 24$ ms demonstrates that it can be largely avoided at the expense of increase scan time. Also note that plastic and foam support used to stabilize the leg are brighter in the sat-UTE images compared to the 3D, another demonstration of the T_2 -induced slice broadening effect in sat-UTE.

DISCUSSION

This paper presents a new method for performing 2D-UTE imaging through subtraction of two non-selective excitations, with and without a slice-selective saturation preparation, and is somewhat similar to previously published methods that combine signal after a sequence of spectroscopic inversion/saturation pulses to achieve the desired contrast and encoding (20,21). The primary benefit of the proposed method is greatly reduced sensitivity to gradient waveform errors, which manifests in HP-UTE as a time-varying out-of-slice signal contamination. The primary cost is a factor of $2\times$ in maximum SNR, because sat-UTE destructively combines in-slice signal, while HP-UTE destructively cancels out-of-slice signal, but constructively combines in-slice signal.

Phantom images and analysis presented in Fig 2-3 demonstrate slice-selection performance of the sat-UTE method in comparison to HP- and DHP-UTE. Overall, it is apparent that sat-UTE offers superior slice-selectivity and echo-time invariance amongst the three 2D methods evaluated. When evaluated in comparison to 3D UTE on cadaveric human femur, sat-UTE shows similar but not identical image quality and signal amplitudes, as presented in Fig 4. The discrepancy in cortical bone signal amplitudes between sat- and 3D-UTE (Fig 4, bottom), points to the primary limitation of the sat-UTE method in comparison with 3D UTE— T_2 -dependent slice profile broadening of cortical bone signal compared to that of bone marrow. The effect is demonstrated Fig 5 with plots of the experimentally measured slice profiles for cortical bone and bone marrow. Although this slice broadening effect will alter image contrast in sat- compared to 3DUTE, the simulation results in Fig 6 show that the effect is not entirely avoided by using HP-UTE.

The increase in the slice profile FWHM and decrease in net and peak signal amplitude with decreasing T_2 of HP-UTE shown in Fig 6 are in qualitative agreement with previous studies on T_2 relaxation and HP excitation (22). For signals with a T_2 greater than about 500 μ s, (and considering the B_1 and gradient limitations of a typical clinical MRI system) the left panel of Fig 6 shows that the slice broadening in sat-UTE excitation is comparable to HP-excited signals, meaning that a similar amount of out-of-slice signal is introduced with both HP and sat-UTE methods. The right panel of Fig 6 depicts a fundamentally different behavior of the total signals as a function of TE for sat and HP-UTE, but the nearly monotonic behavior of the total HP-UTE signal does not reflect a methodological advantage, but rather the unavoidably greater T_2 weighting present in HP-UTE compared to sat-UTE. Implementing sat-UTE with longer duration excitation pulse would result in similar monotonic response of total signal vs T_2 .

Unlike HP-UTE, sat-UTE slice profiles can also be broadened due to MT effects. Relative to fat signal, the shorter-TR sat-UTE image in Fig 7 has higher signal in muscle (known to exhibit an MT effect (23)) than the longer-TR image, and this change in contrast does not originate from changes in T_1 -weighting at shorter TR. The contribution of MT to the slice profile will be a function of the bandwidth and flip angle of the saturation pulse and TR, and may not be simply calculated without a good understanding of the MT characteristics of the tissue. However, this additional MT-based contrast may be reduced by increasing TR (and, therefore, increasing scan time).

In addition to slice profile characteristics, there are other differences between sat- and HP-UTE. In HP-UTE, the longitudinal magnetization remains in the same steady state regardless of which gradient polarity is used in HP excitation, which means that magnetization within a slice can remain in steady state while the two gradient conditions are alternated for acquisition of each radial view in k-space. In contrast, in sat-UTE, the steady state condition changes between the sat/no-sat preparations, requiring dummy scans in the transition between conditions with and without the saturation pulse. In order to keep scan times short, this means that all views from one condition should be required before transitioning to the other condition. Such a scheme introduces the potential for motion- or B_0 -drift-induced errors in out of slice signal subtraction. Techniques exist to monitor and correct for B_0 shifts (24) or motion in radial scans (25), although no such problems are apparent in the in vivo scans presented in Fig 7.

Implementation of the sat-UTE excitation is simple compared to HP-UTE, as standard slice selective pulses can be used for saturation. When considering multi-slice UTE acquisitions, sat-UTE slice offsets can be realized with the addition of linear phase to the saturation pulse RF pulses, whereas HP-UTE (and DHP) requires the phase of each RF pulse to be pre-computed for each offset slice (6), and generally becomes more difficult on slices further from isocenter. However, the aforementioned steady-state requirement also means multiple sat-UTE slices cannot be interleaved within a single TR period. To collect multiple slices, at least one acquisition is needed without the saturation pulse, and then in-slice saturation on multiple slices can be acquired serially. With an increased number of slices, the maximum time interval between the acquisition of the sat and no-sat signals also increases, increasing the sensitivity to motion- or B_0 -drift induced error in out-of-slice signal cancellation.

CONCLUSION

A method of 2D-UTE imaging is demonstrated that calculates the difference between two nonselective excitations with and without an in-slice saturation preparation period. The method is insensitive to imperfections in gradient-field performance, which make accurate HP-UTE challenging on some MRI systems, at the cost of a relative factor of 2 in SNR efficiency.

ACKNOWLEDGEMENTS

This work was supported by NIH grant EB001744 and EB014308. The authors would like to thank Mary Katherine Manhard for help in data collection.

APPENDIX

Within a single TR period, the sat-UTE sequence consists of two RF pulses, the excitation pulse with a flip angle α and the saturation pulse with flip angle ϕ , and two delays, T_D and T_S , as depicted in Fig. 1. Given a longitudinal magnetization $M(t=0^-)$ immediately prior to the excitation pulse and neglecting relaxation during the pulse, the magnetization at the end of the excitation pulse is

$$M(t=0^+) = M(t=0^-) \cos(\alpha). \quad [\text{A1}]$$

Longitudinal relaxation prior to the next saturation pulse results is

$$M(t=T_D^-) = M_0 - (M_0 - M(t=0^+)) \exp(-T_D/T_1), \quad [\text{A2}]$$

where M_0 is the equilibrium magnetization. After the saturation pulse, again neglecting relaxation during the RF pulse, the magnetization is

$$M(t=T_D^+) = M(t=T_D^-) \cos(\phi), \quad [\text{A3}]$$

and after the T_S delay,

$$M(t=T_R^-) = M_0 - (M_0 - M(t=T_D^+)) \exp(-T_S/T_1). \quad [\text{A4}]$$

Given the steady state condition,

$$M(t=T_R^-) = M(t=0^-) \quad [\text{A5}]$$

and solving for $M(t=0^-)$, one obtains

$$M(t=0^-) = M_0 \left(\frac{1 - e^{-T_S/T_1} + (1 - e^{-T_D/T_1}) e^{-T_S/T_1} \cos(\phi)}{1 - \cos(\phi) \cos(\alpha) e^{-(T_D+T_S)/T_1}} \right), \quad [\text{A6}]$$

and the transverse magnetization signal is thus

$$S(\phi) = M(t=0^-) \sin(\alpha) e^{-T_E/T_2^*}. \quad [\text{A7}]$$

Note that in the absence of saturation, $\phi = 0$, Eq [A6] reduces to the standard signal equation for spoiled gradient echo imaging, given that $T_R = T_D + T_S$. The final observed sat-UTE signal results from subtracting the signal acquired with saturation, $S(\phi)$, from that acquired without, $S(0)$, which gives Eq [1].

References

1. Tyler DJ, Robson MD, Henkelman RM, Young IR, Bydder GM. Magnetic resonance imaging with ultrashort TE (UTE) PULSE sequences: technical considerations. J. Magn. Reson. imaging. 2007; 25:279–89. [PubMed: 17260388]

2. Rahmer J, Börnert P, Groen J, Bos C. Three-dimensional radial ultrashort echo-time imaging with T2 adapted sampling. *Magn. Reson. Med.* 2006; 55:1075–82. [PubMed: 16538604]
3. Glover GH, Pauly JM, Bradshaw KM. Boron-11 imaging with a three-dimensional reconstruction method. *J. Magn. Reson. imaging.* 1992; 2:47–52. [PubMed: 1623280]
4. Pauly J, Conolly S, Nishimura D, Macovski A. A Slice Selective Excitation for Very Short T2 Species. *Proceedings of the Society of Magnetic Resonance in Medicine.* 1989:28.
5. Bergin C, Pauly J, Macovski A. Lung Parenchyma: Projection Reconstruction MR Imaging. *Radiology.* 1991:777–781. [PubMed: 2027991]
6. Pauly J. Selective Excitation for Ultrashort Echo Time Imaging. *eMagRes.* 2012; 1:1–7. doi. 10.1002/9780470034590.emrstm1271.
7. Josan S, Pauly JM, Daniel BL, Pauly KB. Double half RF pulses for reduced sensitivity to eddy currents in UTE imaging. *Magn. Reson. Med.* 2009; 61:1083–9. [PubMed: 19235919]
8. Du J, Takahashi AM, Chung CB. Ultrashort TE spectroscopic imaging (UTESI): application to the imaging of short T2 relaxation tissues in the musculoskeletal system. *J. Magn. Reson. imaging.* 2009; 29:412–21. [PubMed: 19161197]
9. Manhard MK, Horch RA, Harkins KD, Gochberg DF, Nyman JS, Does MD. Validation of quantitative bound- and pore-water imaging in cortical bone. *Magn Reson Med.* 2013 doi: 10.1002/mrm.24870.
10. Wansapura JP, Daniel BL, Pauly J, Butts K. Temperature mapping of frozen tissue using eddy current compensated half excitation RF pulses. *Magn. Reson. Med.* 2001; 46:985–92. [PubMed: 11675651]
11. Lu A, Daniel BL, Pauly JM, Pauly KB. Improved slice selection for R2* mapping during cryoablation with eddy current compensation. *J. Magn. Reson. imaging.* 2008; 28:190–8. [PubMed: 18581340]
12. Josan S, Kaye E, Pauly JM, Daniel BL, Pauly KB. Improved half RF slice selectivity in the presence of eddy currents with out-of-slice saturation. *Magn. Reson. Med.* 2009; 61:1090–5. [PubMed: 19319972]
13. Hargreaves BA, Cunningham CH, Nishimura DG, Conolly SM. Variable-rate selective excitation for rapid MRI sequences. *Magn. Reson. Med.* 2004; 52:590–7. [PubMed: 15334579]
14. Duyn JH, Yang Y, Frank J a, van der Veen JW. Simple correction method for k-space trajectory deviations in MRI. *J. Magn. Reson.* 1998; 132:150–3. [PubMed: 9615415]
15. Wong ST, Roos MS. A strategy for sampling on a sphere applied to 3D selective RF pulse design. *Magn. Reson. Med.* 1994; 32:778–84. [PubMed: 7869901]
16. Pipe JG, Menon P. Sampling density compensation in MRI: rationale and an iterative numerical solution. *Magn. Reson. Med.* 1999; 41:179–86. [PubMed: 10025627]
17. Johnson KO, Pipe JG. Convolution kernel design and efficient algorithm for sampling density correction. *Magn. Reson. Med.* 2009; 61:439–47. [PubMed: 19165893]
18. Zwart NR, Johnson KO, Pipe JG. Efficient sample density estimation by combining gridding and an optimized kernel. *Magn. Reson. Med.* 2012; 67:701–710. [PubMed: 21688320]
19. Fessler, J a.; Sutton, BP. Nonuniform fast fourier transforms using min-max interpolation. *IEEE Trans. Signal Process.* 2003; 51:560–574.
20. Ordidge R, Connelly A, Lohman J. Image-Selected in Vivo Spectroscopy (ISIS). A New Technique for Spatially Selective NMR Spectroscopy. *J. Magn. Reson.* 1986; 294:283–294.
21. Foo TK, Sawyer AM, Faulkner WH, Mills DG. Inversion in the steady state: contrast optimization and reduced imaging time with fast three-dimensional inversion-recovery-prepared GRE pulse sequences. *Radiology.* 1994; 191:85–90. [PubMed: 8134602]
22. Robson MD, Gatehouse PD. Consequences of T2 relaxation during half-pulse slice selection for ultrashort TE imaging. *Magn. Reson. Med.* 2010; 64:610–5. [PubMed: 20665804]
23. Stanisiz GJ, Odrobina EE, Pun J, Escaravage M, Graham SJ, Bronskill MJ, Henkelman RM. T1, T2 relaxation and magnetization transfer in tissue at 3T. *Magn. Reson. Med.* 2005; 54:507–12. [PubMed: 16086319]

24. Durand E, van de Moortele PF, Pachot-Clouard M, Le Bihan D. Artifact due to B(0) fluctuations in fMRI: correction using the k-space central line. *Magn. Reson. Med.* 2001; 46:198–201. [PubMed: 11443728]
25. Welch EB, Rossman PJ, Felmlee JP, Manduca A. Self-navigated motion correction using moments of spatial projections in radial MRI. *Magn. Reson. Med.* 2004; 52:337–45. [PubMed: 15282816]

Figure 1. Pulse sequences illustrating the excitation period for sat-UTE. Slice selection by digitally combining signal from two passes, with ($\emptyset \neq 0$) and without ($\emptyset = 0$) slice selective saturation.

Figure 2.

Cropped 2D-UTE images from a phantom consisting of 4 tubes. Two tubes were filled with MnCl_2 (A and C) while another two tubes were filled with agarose (B and D). Tubes C and D were placed within the slice of interest, while tubes A and B were placed ≈ 4 mm outside the slice. Sat-UTE shows improved out-of-slice signal suppression compared to HP- and DHP-UTE.

Figure 3.

Signal vs. TE within four ROIs corresponding to the four tubes in Fig. 2. Sat-UTE signal is insensitive to gradient waveform imperfections and out-of-slice signal contamination, compared to HP- and DHP-UTE. Note that the exact nature of the HP and DHP plots will depend on hardware-specific characteristics, including the gradient amplifier imperfections and eddy currents, that will vary from system to system.

Figure 4.

Top: sat-UTE and 3D-UTE images of a human femur midshaft scanned ex vivo, with ROIs drawn in the cortical bone and bone marrow. Middle: sat-UTE and 3DUTE differences images (TE = 0.1 ms – TE = 0.56 ms). Bottom: Signal vs TE comparisons of sat-UTE and 3D-UTE for ROIs within the marrow and cortical bone.

Figure 5. Sat-UTE slice profiles measured within the marrow and cortical bone. The slice profile within the bone is broadened due to T_2 relaxation during the saturation pulse.

Figure 6.

(left) The normalized FWHM and SNR calculated from simulated slice profiles for HP-UTE and sat-UTE excitation as a function of signal T_2 . (right) The solid line indicates total integrated signal due to slice selection, while the dashed line indicates the peak of the slice profile.

Figure 7.

Sat-UTE and 3D-UTE images of a human tibia from a healthy volunteer. The difference in muscle contrast in the two sat-UTE images are due to MT-based slice profile broadening in short TR images.