Title
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Permalink
https://escholarship.org/uc/item/5pm8c54k

Journal
Magnetic resonance imaging, 43

ISSN
0730-725X

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Publication Date
2017-11-01

DOI
10.1016/j.mri.2017.06.007

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Peer reviewed
Technical note

Effects of fat saturation on short T2 quantification

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1. Introduction

The T2 relaxation properties of MRI signals play an important role in the assessment of pathology [1,2] (R3.4). Many musculoskeletal (MSK) tissues (e.g. cortical bone, tendon, and ligaments, etc.) have very short transverse relaxation times. This short relaxation time requires specialized pulse sequences such as ultrashort echo time (UTE) for optimal signal acquisition and quantification [3,4]. Multiple-echo UTE acquisitions further allow the assessment of the T2s of these tissues, either using a single echo train or by combining two or more interleaved multi-echo scans [5,6]. Generally, these multiple-echo UTE scans can be performed either with or without Fat Saturation (FS).

Short T2 signals decay rapidly as a function of TE, and, via their Fourier transform relationship, have a corresponding broad Lorenzian line-shape (R2.2). As shown in Fig. 1A (blue), these FS prep-pulses typically do not significantly overlap with the main water signal peak of longer T2 soft tissues (such as muscle, etc.). Nevertheless, the broad spectral distributions associated with very short T2 tissues can cause significant overlap with the FS pulse (Fig. 1A, red). This overlap may result in measureable reduction in the short T2 signals [7]. Here we analyzed the effects that FS pulses of various bandwidths (BW) have on the overall decay curve, since all subsequent echoes are effected equally. Often however, the MSK tissues of interest may contain structures with multiple short T2 components [8,9], which simultaneously contribute to signals in the region of interests (ROIs). Such components with different T2 values can get attenuated by varying amounts (defined here as a reduction factor Q), e.g. shorter T2 tissues that have broader linewidths get attenuated more. Therefore, the overall signal decay is altered and a simple single component T2 fitting model may yield corresponding altered results. Fig. 1B shows theoretical signals (at TE = 0) of two different T2s (T2A = 10 ms and T2B = 0.3 ms) after application of FS pulses with various BWs using simulations (non-FS corresponding to FS pulse BW = 0). The simulations are based using the Bloch equations to track the magnetization vectors using small discrete time steps (1 μs) during the application of RF pulses (R2.3). For Fat Saturation a clinical non-adiabatic pulse (default duration 8 ms) was used (R1.1/R2.4). As expected, the short T2 signals get attenuated more than the longer T2 signals.

The combined signal in an ROI containing both tissues is the simple sum:

\[ S_{\text{Tot}}(TE) = Q(T_{2A})\rho_A e^{-\frac{TE}{T_{2A}}} + Q(T_{2B})\rho_B e^{-\frac{TE}{T_{2B}}} \] (1)

As a simplifying example, let’s consider an ROI with equal spin population (ρA = ρB):

\[ S_{\text{Tot}}(TE) = Q(T_{2A})e^{-\frac{TE}{T_{2A}}} + Q(T_{2B})e^{-\frac{TE}{T_{2B}}} \] (2)

Looking at Fig. 1B (see dotted circles) (R1.2) for T2A = 10 ms and T2B = 0.3 ms, using a FS pulse with BW = 400 Hz, this results in a final total...
signal:
\[ S_{T2}^{FS}(TE) = 0.93e^{-\frac{TE}{T_2A}} + 0.7e^{-\frac{TE}{T_2B}} \]  

while the case without the application of FS (NFS) reduces to:
\[ S_{T2}^{NFS}(TE) = e^{-\frac{TE}{T_2A}} + e^{-\frac{TE}{T_2B}} \]

Eqs. (3a) and (3b) demonstrate that FS weights the signal more towards the longer T2 tissue and therefore results in a higher fitted value for T2. The systematic effects on single component T2 fitting is shown in Fig. 1C which results in overall T2 values lying between the two components. For higher FS pulse BWs, the shorter T2 components are attenuated more and the overall measured T2 therefore increases.

3. Methods

Imaging was performed on a clinical 3T GE HDxt MR scanner using a 8 channel T/R knee coil (R1.4/R1.5/R2.4/R3.3). The experimental phantom setup, as shown in Fig. 2A, was used to study the effects of Fat Saturation on T2 quantification. Test tubes containing water were doped with different amounts of MnCl2 (to achieve T2 values of ~10 ms and ~0.3 ms). The test tubes then were imaged using a coronal multi-echo UTE sequence with TE ranging from 0.032 ms to 8.8 ms in increments of 0.8 ms. Images were obtained without fat saturation and with fat saturation using BWs between 200 and 650 Hz. To increase scan efficiency, up to three UTE k-space spokes were acquired per FS pulse. Since these UTE acquisitions are very short (<5 ms) the fat suppression was not significantly affected (R3.1). ROI measurements were obtained in the individual tubes and combined (summed) to emulate multi-component signals. Single component exponential fitting was performed to obtain T2 values.

Furthermore, in-vivo scans were also performed on a healthy male volunteer (age 72 y.o.) in accordance to the institutional IRB approval (R1.4). Sagittal multi-echo images (see Fig. 3) were obtained without FS, as well as with two separate FS pulse BWs. T2 values were obtained using a single component fit in the patella tendon, meniscus, posterior cruciate ligament (PCL), and femoral-tibial cartilage.

4. Results

Fig. 2B shows the signal reduction as a function of FS BW for the two individual phantom tubes. These curves agree well with the general trend shown in the theoretical simulated curves in Fig. 1B. The measured values of T2 of the combined signals are shown in Fig. 2C as a function of the FS pulse BW. These measured values also show good agreement with the theoretical simulations.

Several sagittal UTE images of the in-vivo knee (with FS) are shown in Fig. 3 at different TE values. A pronounced signal drop can be observed in short T2 tissues such as the meniscus (arrow), while the signal intensities for longer T2 tissues such as muscle do not change much over the range of TE values. Fig. 4A shows the single-component fitted T2 values of the in-vivo slices. For all anatomies a similar systematic increase of the measured T2 can be observed for larger FS pulse BWs, as expected. The larger dispersion displayed by the cartilage T2 indicates a higher sensitivity of single component fitting on fat saturation. By comparison, the measured T2 of muscle only increased by ~5%. (R3.9) Fig. 4B shows the measured fat signals for the three different FS BWs. As expected, the highest fat signal is observed for the FS BW = 0 (non-FS), while increasing the FS BW increases the fat suppression efficiency (R2.1).
5. Discussion

We have shown that Fat Saturation can have notable effect on the measured single component T2 values of multi-component tissues. Since the dispersion is less pronounced for smaller FS bandwidths, longer FS pulses can help to minimize the adverse effects of fat saturation. However, this needs longer FS pulse will also result in less complete fat saturation, this may require a compromise (R3.7). Our tests have shown that Fat Saturation can have notable effect on the measured single component T2 values of multi-component tissues.

Fig. 2. The simple phantom setup is shown in panel A. B–C) Experimental verification of the corresponding theoretical simulations in Fig. 1B–C, which show a similar trend.

Fig. 3. Sagittal in-vivo knee slices at various TE's. Although the signal intensities for longer T2 tissues such as muscle do not change much over the range of TE values, a more pronounced signal drop can be observed in short T2 tissues such as the meniscus (arrow).
primarily focused to 3 T where the main fat peak is shifted by ~440 Hz from the main water peak. At 1.5T our results need to be scaled accordingly to account for the closer proximity of the fat and water peaks (~220 Hz) (R1.5).

Alternatively, multi-component fitting can be used as another possible solution. For example, the effect of fat saturation on bi-component signal decay fitting is currently being studied by our group. Although a bi-component fit will allow more accurate calculation of both T2 values, the fit values of the relative spin density fit will still be altered by the FS pulse. This is because the fat saturation pulses alter the available longitudinal magnetization (particular for short T2 tissues) prior to UTE excitation. During the subsequent data acquisition these pools therefore contribute less signal and are underestimated in the parameter fitting (R1.6, R3.5) Currently, we are developing a correction algorithm based on our theoretical work in this paper to allow correct extraction of the relative spin densities even for scans with fat saturation.

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