Improved transverse relaxometry: a new fitting model with stimulated echo compensation

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Introduction

Transverse relaxometry – mapping transverse relaxation times (T_2) or rates ($R_2=1/T_2$) – is an important investigational tool for quantitation. Potential clinical applications include detection of hippocampal sclerosis in epilepsy **[1]**. Research applications include myelin imaging **[2]** and protocol design. Data is traditionally acquired with a multi-echo spin-echo sequence. Complete images are formed at each echo time and relaxation rates are extracted via exponential fitting. This approach requires 180° refocusing angles (RAs) to generate signal modulated exclusively by T_2 decay. In practice, this stipulation is chronically violated **[3]**: finite refocusing widths (RWs), non-rectangular slice profiles, transmit calibration errors, and field-focusing – a particularly problematic effect at high field – all collude to alter the RAs. Deviations in RAs introduce stimulated-echo signal pathways, which include T_1 weighting and thus confound the T_2 measurement.

Here, we present a robust fitting model for transverse relaxometry with multi-echo spin-echo data. This model compensates for stimulated echo contributions arising from non-ideal RF pulse shape, width, and amplitude. Our model improves relaxometry reliability at all field strengths and permits efficient data acquisition with thin refocusing widths.

Methods

We integrate echo amplitudes, computed with the extended phase graph algorithm **[4]**, over the slice profile to obtain an aggregate decay curve. This curve serves as the objective function for non-linear fitting. Slice profile shapes and widths, prescribed RAs, and sequence timings are all known; T_1, T_2, transmit error/field-focusing, and a scaling factor are unknown. Unfortunately, T_1 and T_2 effects are indistinguishable; however, assuming $T_1 \gg T_2$, which is valid for most tissues, we can extract all remaining unknowns. This assumption stipulates that magnetization exposed to T_1 processes via stimulated pathways undergoes negligible relaxation, but is preserved for subsequent echo formation. While not perfect, this only applies to signal evolving in stimulated pathways and is superior to simply neglecting this eventuality.

Our proposed model was verified experimentally in Mn^{2+} doped water phantoms and in the human brain at 4.7 T. Gold standard T_2 values were obtained with wide refocusing pulses (RW=5x excitation width), 180° RAs, and exponential fitting. We assessed the accuracy of our model at various RWs and RAs (latter not shown in abstract). *In-vivo* data were collected at 1.2 x 1.2 x 5 mm^3 resolution with a train of 20 Gaussian RF pulses and 15 ms echo spacing. Images were collected with various RWs and RAs; our proposed model and the exponential fit were compared to the gold standard.

Results and Discussion

Signal decay from a 0.3 mM Mn^{2+} phantom acquired with non-ideal refocusing is shown in **Fig. 1**. The true T_2 is 27 ms; exponential fitting erroneously returns T_2=42 ms while our proposed model reports T_2=25 ms. Model accuracy, determined experimentally, is shown in **Fig. 2** at three RWs. A relaxation rate map processed with our model is shown in **Fig. 3**. Comparison with true relaxation rates, **Fig. 4**, validates our model for *in-vivo* applications.

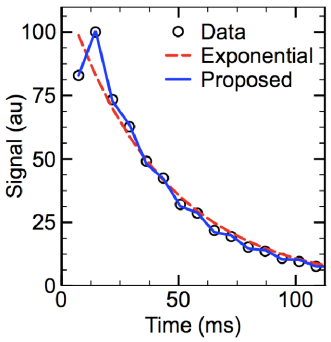

Fig 1: Measured signal decay with non-ideal conditions of RW=excite-width and α_R=120°. Our proposed model provides an improved fit over the exponential.

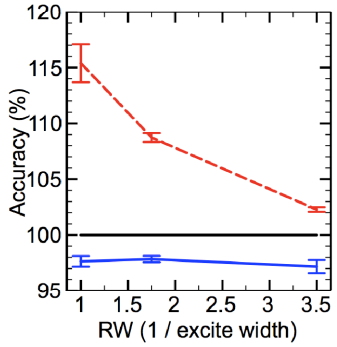

Fig 2: Experimentally determined T_2 accuracy of the proposed (solid blue) and exponential (dashed red) fits as functions of refocusing width.

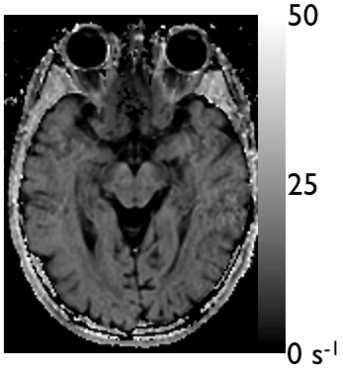

Fig 3: R_2 map of a healthy 27 year old volunteer processed with our proposed model. Intensity scale from 0 to 50 s^{-1}.

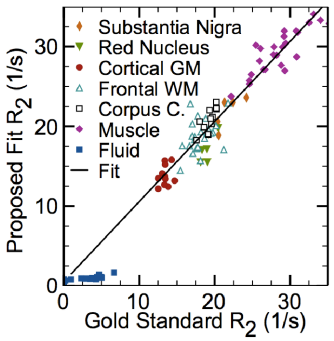

Fig 4: Scatter plot from regions in Fig. 3 (and other slices) of R_2 from our method versus the true R_2. Weighted linear regression yields a slope of 1.00 ± 0.03.

Conclusions

Stimulated echoes due to refocusing errors severely degrade traditional relaxometry data. Our model compensates for slice profile, finite refocusing width, and amplitude deviations to provide accurate relaxometry measurements. This technique permits relaxometry at very high field strength, where field focusing biases measurements, and improves multi-slice acquisition efficiency at all field strengths by permitting thin refocusing pulses.

References **[1]** Bernasconi A. et al. (2000). Neuroimage, 12(6), 739-46. **[2]** MacKay A. et al. (1994). Magn Reson Med, 31(6), 673-7. **[3]** Crawley A. P. et al. (1987). Magn Reson Med, 4(1), 34-47. **[4]** Hennig J. (1988). Journal of Magnetic Resonance, 78(3), 397-407.