Modeling of T2* decay in water/fat imaging: comparison of one-decay and two-decay models

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INTRODUCTION

Chemical shift-encoded techniques for quantitatively measuring the presence of fat have a number of important applications in MRI, including bone marrow, muscle, brain, liver and heart studies. However, inaccurate modeling of the acquired signal can result in significant bias in the estimates of fat amplitude or fat fraction. Specifically, modeling the T2* decay (or the decay rate R2*=1/T2*) of the signal has been shown to be necessary in order to avoid such bias [1]. In recent works, the possibility of modeling separate decay rates for water and fat has been proposed [2,3]. Even though the two-decay model is more accurate (potentially lower bias), it suffers from increased noise sensitivity (higher standard deviation) with respect to the one-decay model due to the need to estimate an additional nonlinear parameter. In this work, we analyze quantitatively the tradeoff between bias and standard deviation using simulation, phantom and in vivo data.

METHODS

Phantom construction: A water fat phantom was constructed by mixing water and oil in separate vials with fat fractions (%): 0, 10, 20,30,40,50,60,70,100, as described in [4,5].

Data acquisition: Phantom and in-vivo data were acquired using a spoiled GRE sequence with monopolar readout, 8 TEs with initial TE=1.43ms and TE spacing=2.23ms. The phantom acquisition was performed both with flip angle=8°, TR=500ms (giving SNR≈30), and with flip angle==25°, TR=2000ms (giving SNR≈90). Each type of acquisition was repeated 128 times to obtain "Monte-Carlo" measurements.

Processing: The complex-valued data were processed by fitting (nonlinear least-squares) using single-peak/multi-peak fat models; each with one-R2* and two-R2* decay models. Multi-peak (6-peak) calibration and gold standard for fat amplitude were obtained from a 32-point acquisition in the phantom.

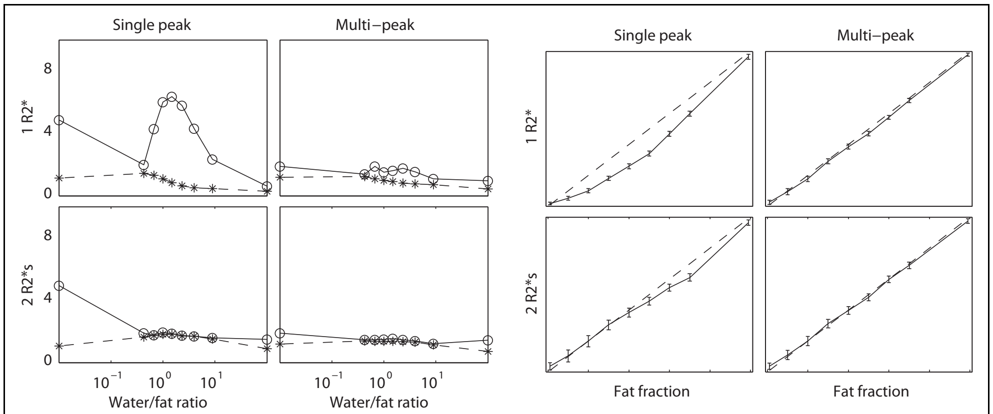

Figure 1. Phantom results for performance comparison of different fitting models in the case of moderate SNR (~30): single-peak and multi-peak fat, with a single R2* or two R2* decays. (Left) Standard deviation (stars) and root mean squared error (RMSE, circles) for fat amplitude estimation, as a function of true water/fat ratio. (Right) Mean with standard deviation bars for fat fraction (FF) estimation, for true FF ranging from 0% to 100%.

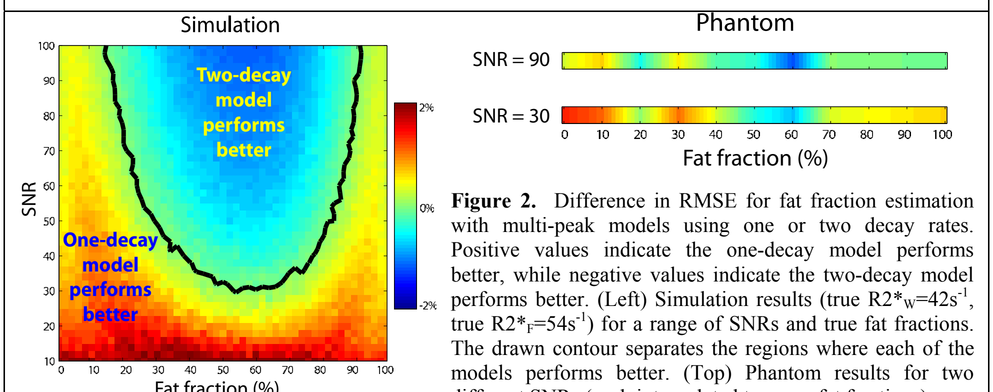

Figure 2. Difference in RMSE for fat fraction estimation with multi-peak models using one or two decay rates. Positive values indicate the one-decay model performs better, while negative values indicate the two-decay model performs better. (Left) Simulation results (true $R2^*_W$=42s^{-1}, true $R2^*_F$=54s^{-1}) for a range of SNRs and true fat fractions. The drawn contour separates the regions where each of the models performs better. (Top) Phantom results for two different SNRs (each interpolated to cover fat fractions).

RESULTS AND DISCUSSION

Figure 1 shows the phantom results for fat amplitude and fat fraction estimation. As expected, the two-R2* models produce lower bias and higher standard deviation relative to the corresponding one-R2* models.

The difference between mean R2* estimates for water and fat ($R2^*_F - R2^*_W \approx 12$ s^{-1}) from the two-R2* multi-peak model was in good agreement (assuming $R2^*_{\{W,F\}} = R2_{\{W,F\}} + R2'$, with R2' common for water and fat) with R2 estimates obtained using a spin-echo experiment, where $R2_F - R2_W \approx 11$ s^{-1}.

Figure 2 shows a direct comparison of multi-peak models with one-R2* and two-R2*. Note that, at sufficiently high SNR and fat fractions close to 50%, the two-R2* model is preferable. However, for low SNR or fat fractions close to 0% or 100%, the one-R2* model is preferable because the increased noise sensitivity in the two-R2* estimates outweighs the reduced bias attainable with the more sophisticated model. Figure 3 shows in vivo results (liver), in good agreement with the simulations and phantom data.

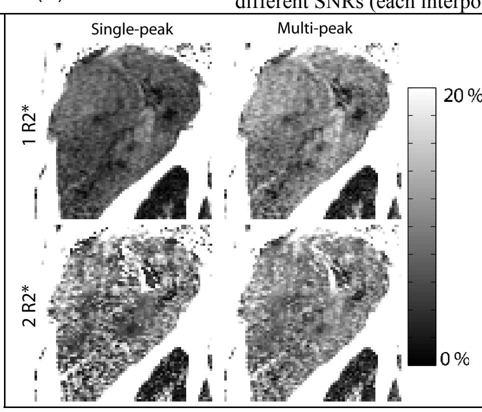

Figure 3. In vivo liver imaging results. The relative results of the different models are in good agreement with the simulation and phantom studies: two R2*s result in significantly noisier estimates compared to one R2*. In single-peak modeling, however, 1 R2*s provides a lower mean FF estimate (presumably due to bias, as in the phantom results). In the multi-peak case, one-R2* has approximately the same mean as two-R2*.

CONCLUSION: Among the single-peak fat models, a two-R2* model is preferable to a one-R2* model in order to account for the broader effective fat peak composed of several actual peaks. Among the multi-peak models, a one-R2* model is preferable for a clinically relevant range of fat fractions and SNRs.

REFERENCES: [1] Yu H et al., Magn Reson Med 60:1122-1134, 2008. [2] Chebrolu VV et al., ISMRM 2009, p. 2847. [3] Bydder M et al., Magn Reson Imaging, 26:347-359, 2008. [4] Bernard CP et al., J Magn Reson Imaging, 27:192-197, 2008. [5] Hines CD et al., Magn Reson Med, 30:1215-1222, 2009.