Solving the free water elimination estimation problem by incorporating T_2 relaxation properties

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Introduction

The free water elimination (FWE) diffusion model accounts for partial volume effects that occur when voxels in diffusion tensor imaging (DTI) volumes contain both a tissue and a free water compartment[1-2]. A downside of FWE is that the model fitting problem is ill-conditioned[3-4]. Advanced parameter estimation techniques that incorporate regularization usually succeed in stabilizing the model fit but, as a trade-off, impose model assumptions that are likely to bias the results[1,5-6]. In this work, we exploit that the T_2 relaxation times of white matter and cerebrospinal fluid are very different. By accounting for the associated echo time (T_E) dependency of the signal decay, the model parameters can be estimated more precisely, accurately, and robustly.

FWE-T_2 model

T_2-weighting — Diffusion weighting

$$S_i = S_0\left(\underbrace{(1-f)e^{-T_E/T_2^{tissue}}e^{-b_i \boldsymbol{g}_i^T \boldsymbol{D} \boldsymbol{g}_i}}_{\text{Tissue}} + \underbrace{f e^{-T_E/T_2^{fw}}e^{-b_i d}}_{\text{free water}}\right)$$

S_0 = signal without diffusion weighting on $T_E = 0$; b_i, $\boldsymbol{g}_i$ = diffusion weighting strength and direction; $\boldsymbol{D}$ = diffusion tensor; d = diffusion of free water at body temperature ($= 3\ \mu\text{m}^2/\text{ms}$)

Results

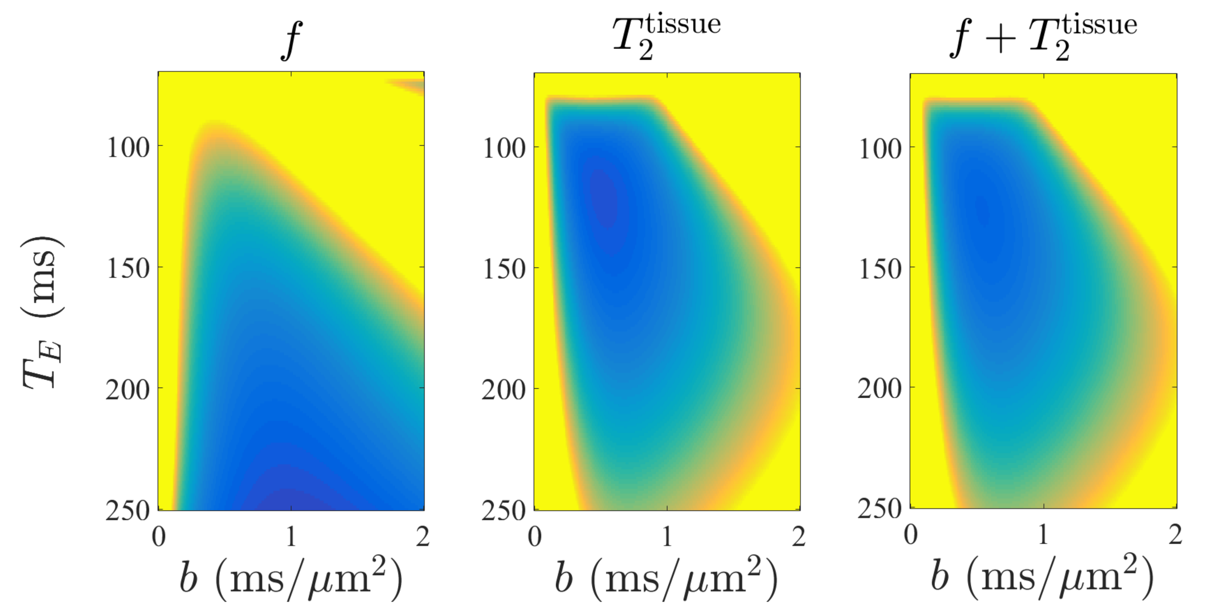

Figure 1: The coefficient of variation for the FWE-T_2 model with T_2^{fw} estimation, derived from the Cramér-Rao lower bound (assuming Gaussian noise), for f, T_2^{tissue}, and a combination of both. A typical WM voxel with free water was assumed. Protocol: $5 \times b = 0$, $30 \times b = 1$ with $T_E = 70$ms and 1 extra shell with variable b and T_E(30 dirs).

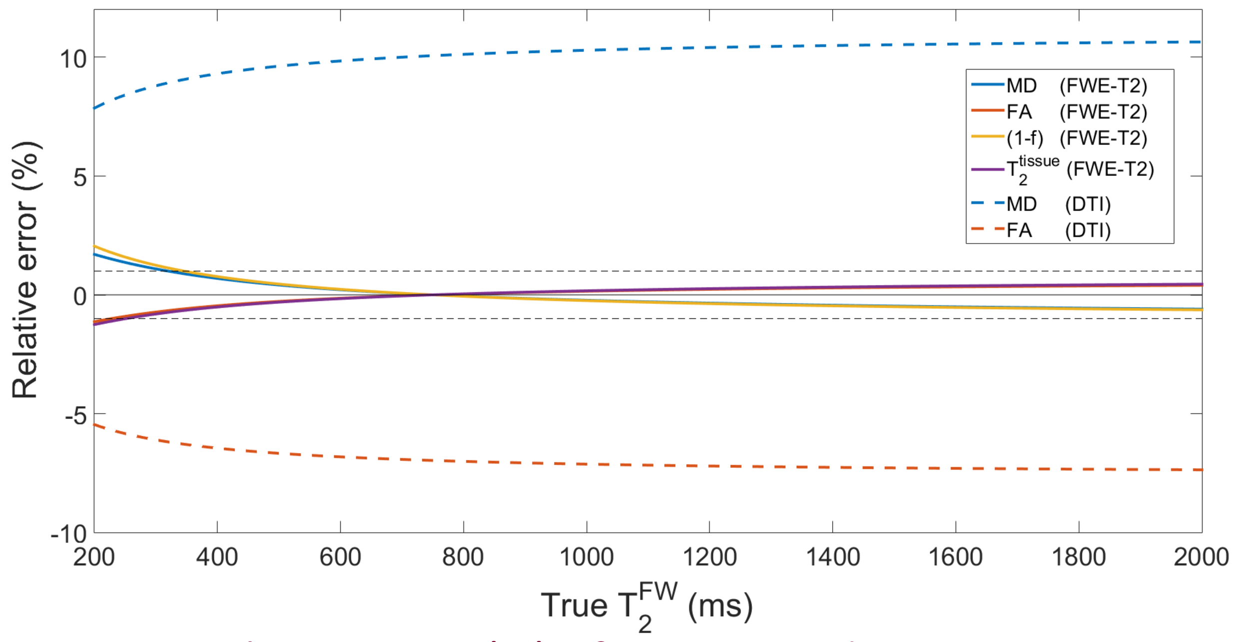

Figure 3: Relative error (%) of FWE-T_2 and DTI parameters in function of the true value of T_2^{fw} (FWE-T_2 model assumes a fixed $T_2^{fw} = 750$ms).

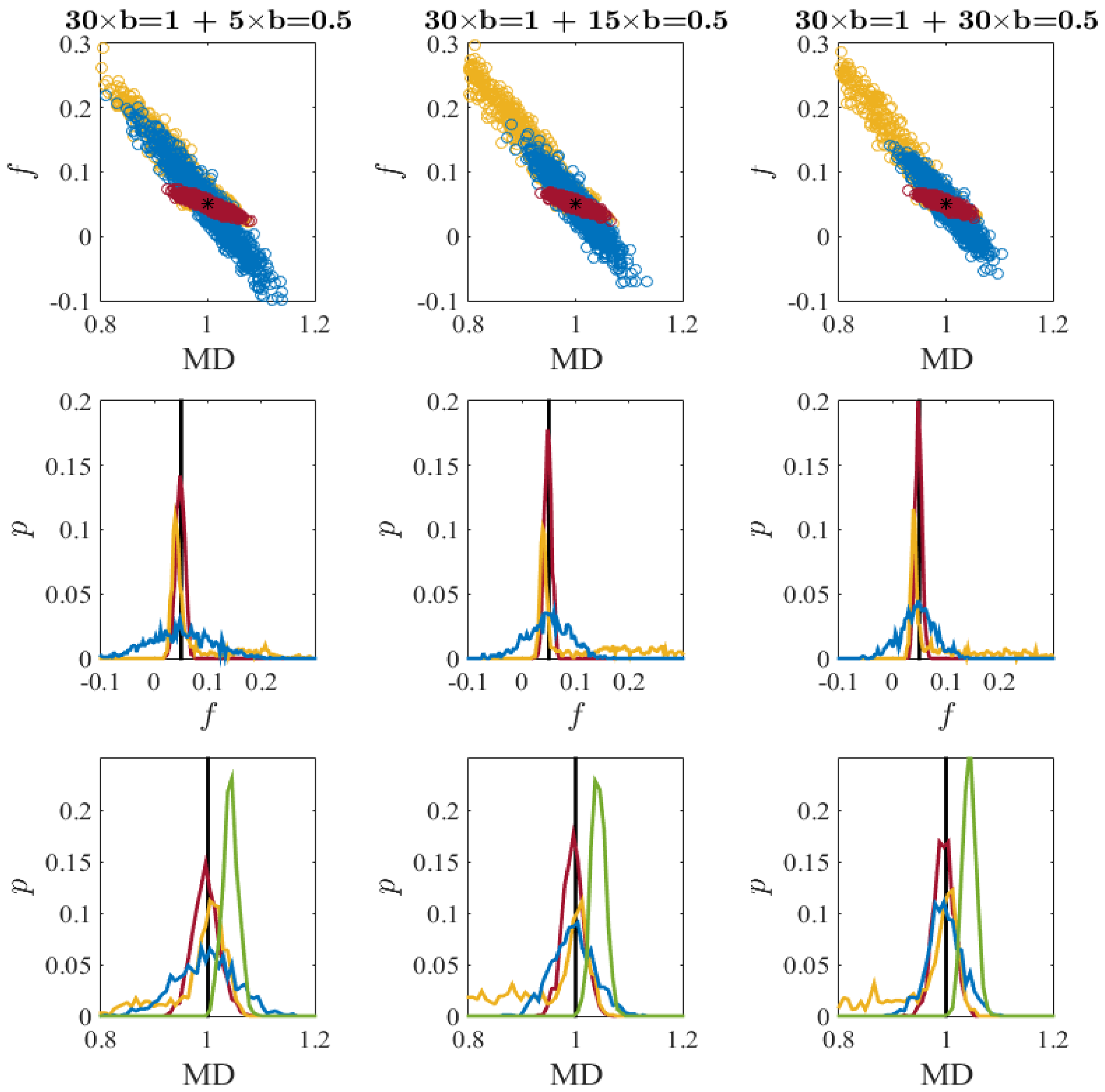

Figure 2: Single voxel Monte Carlo simulations assuming a typical WM voxel with a free water compartment and SNR=50 on $b = 0$ and $T_E = 100$ms signal. Comparison between: FWE-T_2 w/o T_2^{fw} est., FWE-T_2 with T_2^{fw} est., FWE, DTI and true value. Protocol: $5 \times b = 0$ ($T_E = 70$ms), $30 \times b = 1$ ($T_E = 70$ms) and $5/15/30 \times b = 0.5$ ($T_E = 120$ms or 70ms for DTI and FWE).

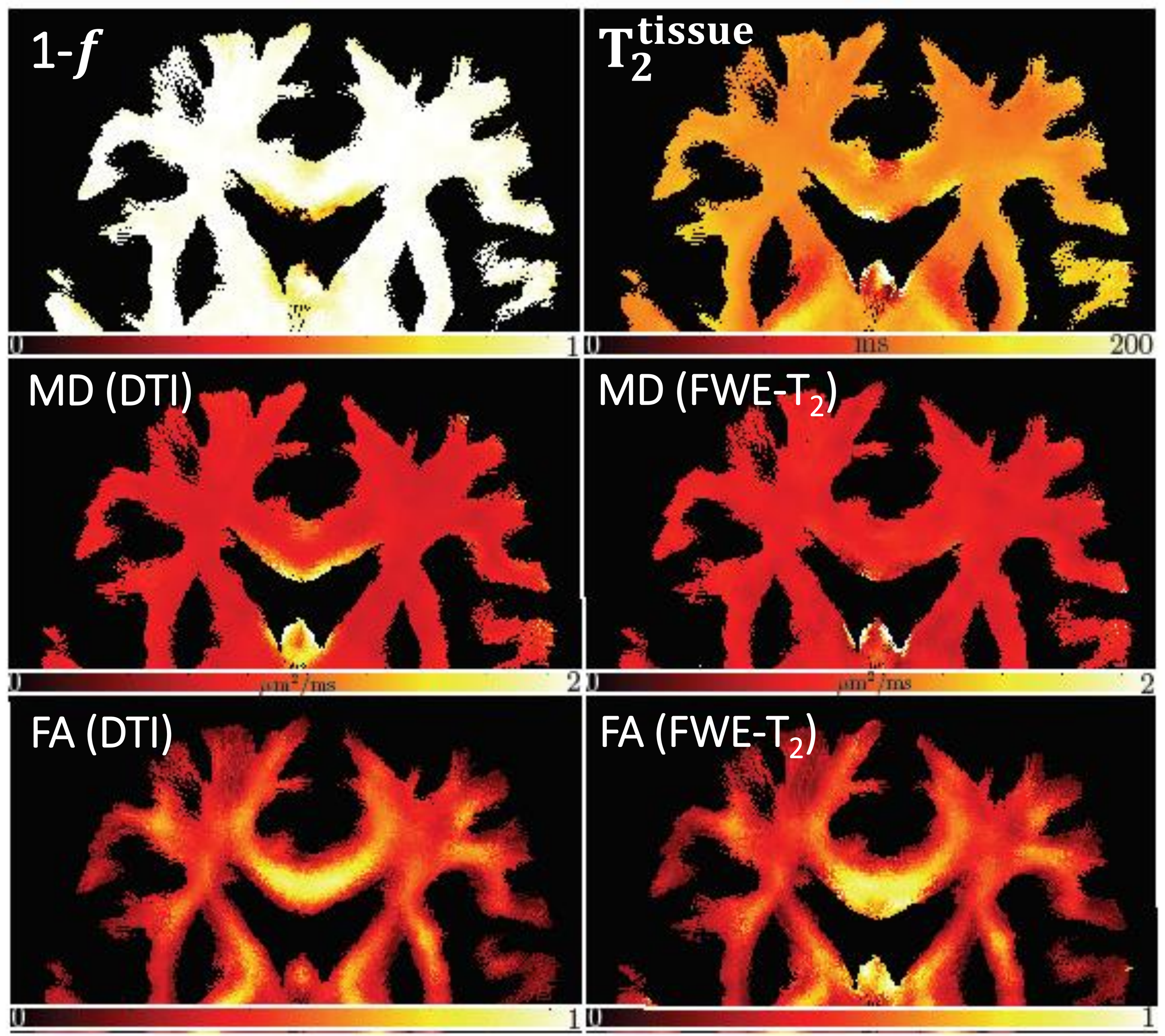

Figure 4: Real data results of FWE-T_2 (w/o estimation of free water T_2 value) and DTI. A healthy volunteer was scanned on a 3T scanner with $1 \times b = 0$, $30 \times b = 0.5$ ($T_E = 120$ms) and $30 \times b = 1$ ($T_E = 70$ms). The data was denoised[7] and subsequently corrected for Gibbs ringing[8], eddy current distortions and motion[9].

Discussion and conclusion

In this work, we propose an extension to the FWE model that incorporates the T_E dependencies of both the tissue and free water compartments. We showed that in addition to a standard DTI acquisition protocol, ideally, only a sparsely sampled additional shell should be acquired with $b \approx 0.5\,\text{ms}/\mu\text{m}^2$ and $\text{TE} \approx 120\text{ms}$ (fig. 1). Monte Carlo simulations indicate that the FWE-T_2 model with a prior estimated T_2^{fw}, is substantially better compared to FWE or FWE-T_2 with estimation of T_2^{fw}, in terms of both the precision and accuracy of f and MD estimation. Fixing the T_2 of CSF is justified because the increase in precision (fig. 2: red versus yellow) outweighs the potential drop in accuracy for the FWE-T_2 model parameters, where the errors will typically be smaller than 1% (fig. 3). Finally, clinical data shows that neglecting to account for free water partial volume effects, biases the estimation of diffusion properties of brain tissue close to CSF regions.

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