Computer Simulations of 3D MPRAGE in Human Brain with Inclusion of Inadvertent Magnetization Transfer Effects

G. Matson[1], and H. Liu[1]
[1]CIND, Veterans Affairs Medical Center, San Francisco, California, United States

Introduction: In the human brain, both white matter and gray matter contain significant amounts of bound water. Although the bound water is not directly seen in conventional MRI experiments, due to its very short T2, the effects of RF pulses on the bound water can influence the free water signal through proton exchange. This effect is known as magnetization transfer (MT) [1], and when MT effects are unintentionally produced, the result has become known as inadvertent MT. Our simulations show that, for conventional 3D MPRAGE experiments at 4.0 T, the inadvertent MT effects are significant, but not substantial. At 4.0 T and higher strength fields, the B_1 inhomogeneity over the human head becomes quite significant [2], producing deleterious effects, including regionally varying contrast and loss of signal to noise (S/N). Although B_1 shimming has become a popular approach for reducing the B_1 inhomogeneity, B_1 shimming requires both the MRI instrument and the excitation coil be equipped with multiple RF channels. Another approach, suitable for 3D MPRAGE experiments, is the use of RF pulses which produce uniform tips even in the presence of non-uniform B_1 fields (B_1 insensitive pulses) [3,4]. We have recently designed our own B_1 insensitive pulses [ISMRM 2010, submitted]. However, as the B_1 insensitive pulses are invariably much longer than the rectangular pulses they replace in the conventional 3D MPRAGE experiment, the inadvertent MT effects can be quite substantial. Thus, our aim was to develop a simulation program for 3D MPRAGE experiments that incorporated MT effects into the simulation.

Methods: Our simulation program was written in Matlab, and provides results both with and without consideration of MT effects. Without consideration of MT effects, the program follows roughly along the lines outlined by Deichmann et al. [5]. For consideration of MT effects, we used the two compartment exchange model provided by Helms and Hagberg [6]. Briefly, the model of reference 6 provides for the propagation of the saturation of both the free and bound pools due to RF pulses, and due to relaxation and exchange. The saturation is given by $\eta = 1 - M/M^0$, where M represents the z-component magnetization. The relative saturation induced by an RF pulse is designated by δ, where $\delta = 1 - M^+/M^-$ where M^+ represents the z-component magnetization prior to the pulse, and M^- the magnetization following the pulse. With the use of f for free and b for bound fractions, reference 6 shows that, using $\boldsymbol{\eta} = (\eta_f\ \eta_b)^T$, the saturation evolves due to an RF pulse according to: $\boldsymbol{\eta}^+ = \begin{pmatrix} 1-\delta_f & 0 \\ 0 & 1-\delta_b \end{pmatrix} \boldsymbol{\eta}^- + \begin{pmatrix} \delta_f \\ \delta_b \end{pmatrix}$, and evolves with time according to $\boldsymbol{\eta}(t) = \mathbf{T}(t)\ \boldsymbol{\eta}(0)$, where the elements of the matrix **T** are complicated functions of relaxation times and exchange rates [6]. The effect of the RF pulse on the bound water was calculated following the prescription provided by Li et al. [7], using a super-Lorentzian lineshape. We have named our simulation package MatMRI, and will make it available on the web.

Results: Here we show our 3D MPRAGE simulations at 4.0 T using our B_1 insensitive pulse with a maximum B_1 strength of 7 μT and a length of approximately 2 ms. Figure 1 shows the menu for the simulations, where the number of RF pulses has been reduced to 36 to avoid exceeding SAR limitations, and the time between pulses lengthened to 20 ms to enable the collection of the equivalent of multiple lines of k-space (The additional data collection is required if the time for the MRI examination is to remain unchanged). Figure 2 shows the results for Mz magnetization without consideration of MT effects, using a tip angle of 10 degrees. The fact that both the white matter and gray matter magnetizations are rising during the data collection suggests that a larger tip could be used. However, Fig. 3, which includes MT effects, presents a dramatically different picture. The larger MT effects on the white matter reflect the fact that the bound pool is larger in the white matter than in the gray matter. Other simulations show that use of a larger tip angle exacerbates the reduction of the white matter as the data is collected (Results not shown).

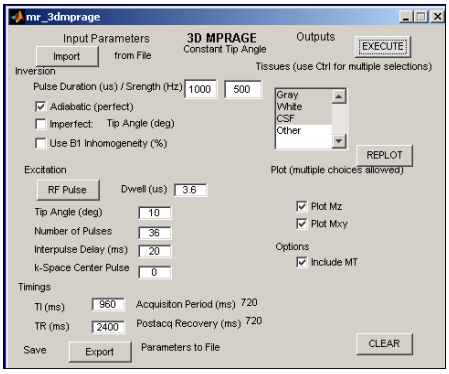

Fig. 1. Menu for the MatMRI simulations

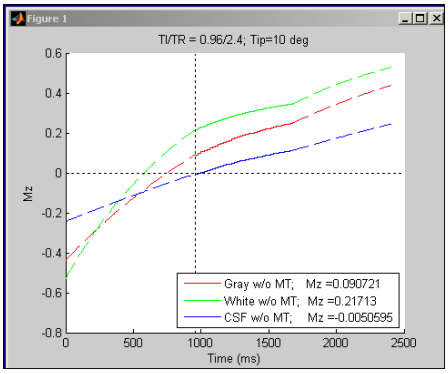

Fig. 2. Mz simulations without MT effects.

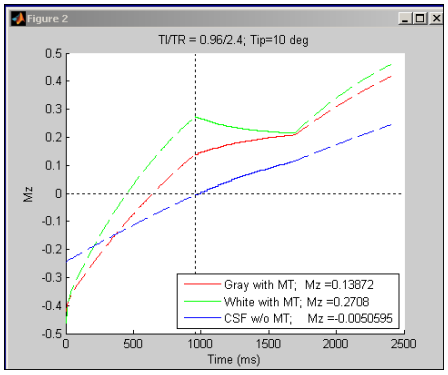

Fig. 3. Simulations with MT effects included.

The simulations also enable the calculation of variable tip angles to maintain the magnetization of one particular tissue type constant, with a prescribed reduction at the edge of k-space. However, as the different tissue types require different recipes of tips, the use of variable tips provided only a relatively modest improvement in the efficiency of the MPRAGE experiment.

Conclusions: Particularly with the use of longer RF excitation pulses, the inadvertent MT effects must be taken into account to properly design MRI experiments to make effective use the longer pulses, and obtain optimal contrast between gray and white matter. The MatMRI program provides a flexible tool for altering the parameters of the 3D MPRAGE experiment to optimize the experiment.

References: [1]. SD Wolf and RS Balaban, Magn Reson Med 10, 135, 1989. [2]. JT Vaughan et al. Magn Reson Med 46, 24, 2001. [3]. N Boulant et al. Magn Reson Med 61, 1165, 2009. [4]. JE Moore et al. Intl Soc Magn Reson Med 17, 2579, 2009. [5]. R Deichmann et al. NeuroImage 12, 112, 2000. [6]. G Helms and GE Hagberg, Concepts Magn Reson 21A, 37, 2004. [7]. JG Li, SJ Graham, and RM Henkelman, Magn Reson Med 37, 866, 1997.