Quantification of magnetization transfer and relaxation rates by MT-prepared multi-echo EPI

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Introduction:

In clinical applications, magnetization transfer (MT) is commonly quantified using rapid gradient echo sequences by analogy to continuous wave experiments (1-3). We derived an analytic solution for both the dynamic and steady state of a progressive saturation experiment (4). Using repetitive MT-pulses and multi-echo EPI for single-shot read-out the relaxation and transfer rates of both pool can be quantified when sampling the transition to steady state at different pulse repetition periods (PR).

Theory:

The general solution for a binary spin-bath of free water (f) and macromolecules (m) with linear exchange kinetics assumes free evolution during the whole PR (4). Model 1: In tissue, the macromolecular content ($F = M_{0m}/(M_{0m}+M_{0f})$ is small, and the difference in relaxation rates (R_{1m}-R_{1f}) can be neglected compared to the sum of exchange rates ($k_{mf}+k_{fm}$). The free water signal can then be described by analogy to progressive partial saturation (5), by means of an apparent saturation (d_{app}):

$$d_{app} = d_f + F(d_m - d_f)\,(1 - exp(-(R_T - R_R)\,PR))\,/\,(1 - (1-d_m)\,exp(-R_T\,PR)). \quad [1]$$

R_T is the fast apparent transfer rate, and R_R is the slow apparent relaxation rate observed during free evolution. The direct saturation caused by one MT-pulse (d_f) increases during PR by an MT-contribution proportional to the differential saturation of the pools, $d_m - d_f$. The dependence on the number of MT-pulses (N) is thus given by

$$M_{zf}/M_{0f}(n) = \{1 - exp(-R_R\,PR) + d_{app}\,exp(-R_R\,PR)\,[(1-d_{app})\,exp(-R_R\,PR)]^N\}/\{1-(1-d_{app})\,exp(-R_R\,PR)\}. \quad [2]$$

Model 2: When discarding the assumptions named above, different longitudinal relaxation rates in the pools have to be considered. Their weighted average yields the observed $R_R = (1-F)R_{1f} + F\,R_{1m}$. The steady state and d_{app} have to be corrected by small terms proportional to $F(R_{1m}-R_{1f})$ (4).

Methods and Materials:

Measurements were performed at 1.5 Tesla on 6 healthy adults using a Siemens Vision and the standard head coil. Gaussian MT-pulses of 6.4 ms durations and 720° nominal flip angle were repetitively applied at 1 kHz offset. An axial slice through the centrum semiovale was measured by a 16-echo EPI sequence (TE=50, 100, …800 ms, 5mm thickness, 20 cm FOV, 64x64 matrix). The dynamic behaviour and the steady state were sampled at various PR (8 - 200 ms) (Fig. 1). Fat suppression was accommodated in the last PR interval.

ROIs in cortical grey matter (GM) and central white matter (WM) were evaluated by a global fit to the complete multi-echo data. The Levenberg-Marquardt procedure of IDL 4.0.1 (Research Systems Inc., Boulder, CO) was extended to three independent variables (N, PR, TE). In GM, CSF signal was modelled by a second T2-component; in WM, we modelled the T_2-decay into the Rician noise (6). The saturations were constrained between 0 and 1; progressive saturation of CSF (7) was accounted for by a T_1 of 3.5 s and a saturation of 0.1%. To simplify the partial derivatives of the signal dependence, a convenient parameter set was chosen for MT. Macromolecular ratio, kinetic and relaxation parameters were then calculated from the fitted parameters.

Results:

Evaluation of WM (T_2 = 80.8±8.1 ms) was more reliable than GM (T_2 = 76.7±6.4 ms), which was affected by motion. The fitted apparent relaxation rates (WM: 1.21± 0.08 1/s; GM: 0.69±0.15 1/s) were in line with known T_1 values at 1.5 Tesla. In WM, Model 2 (unequal relaxation in the pools) significantly reduced the residues ($p < 0.05$, F-test). This also affected the fitted RR and RT ($p < 0.001$, paired T-test), and thus the MT-parameters. No such differences were seen in GM, where Model 1 gave more consistent results.

Macromolecular ratios were higher in WM (f = 24.5±11.6%) than in GM (7.4±3.0%) (p = 0.003). For WM, the pool-specific relaxation times were determined from Model 2 as 1.59±0.28 s for bulk water, and 0.28±0.10 s for macromolecules. The apparent transfer rates were 14±4 1/s in WM, and 32±15 1/s in GM. The macromolecular saturation was d_m=47±13% in WM and d_m=68±24% GM; the direct effect on free water was less than 1%.

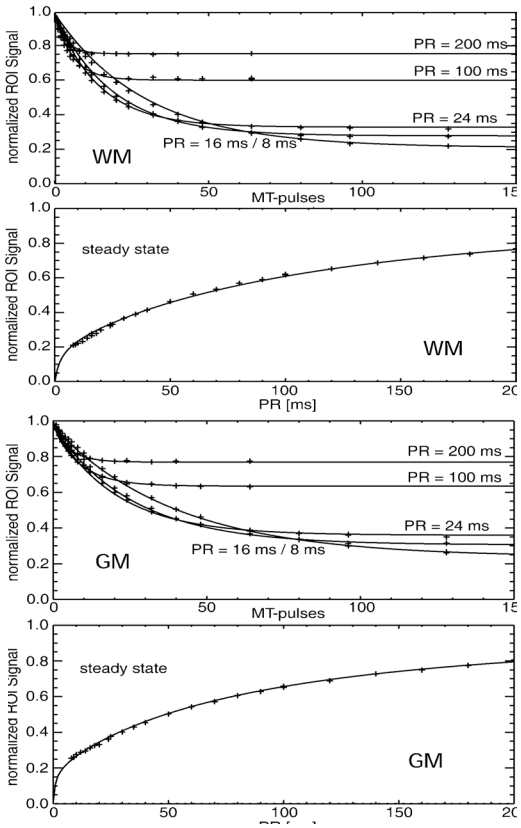

Figure 1: Sampling scheme and fitted model. Here, the signal appears corrected for CSF and normalized. For clarity, only the first echo is shown.

Discussion:

This study suggests that the T_1 of macromolecules is considerably shorter than commonly assumed (1 s). Macromolecular ratios were higher and the backward rates were slower than determined at short TE by gradient-echo MT (1-3). This may be explained by myelin water contributing to the observed MT effect (8). When discarding the multi-echo capability, quantitative MT-mapping of the whole brain may become feasible by a suitable permutation of the slice order.

References:

1: Sled JG, Pike GB. *Magn Reson Med* 2001; 46:923-931
2: Yarnykh VL. *Magn Reson Med* 2002; 47:929-939
3: Ramani A, et al. *Magn Reson Imaging* 2002; 20:721-731
4: Helms G, Hagberg GE. *Conc Magn Reson* 2004; 21A: 37-62
5: Helms G, Piringer A. *NMR Biomed* 2004; in press.
6: Jones DK, Basser PJ. *Magn Reson Med* 2004; 52:979-93.
7: Helms G, Piringer A. *Magn Reson Imag* 2001; 19:803-811
8: Stanisz GJ, et al. *Magn Reson Med* 1999; 42:1128-1136