Effect of Magnetization Transfer on Quantification of CBF

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Introduction

Continuous arterial spin tagging methods have been extensively used for quantification of tissue perfusion (1). Assuming that labeled water protons in blood diffuses rapidly into tissue and exchange with macromolecular protons, cerebral blood flow (CBF, in a unit of ml/g/min) can be calculated as

$$CBF = \frac{\lambda}{E}\left(\frac{1}{T_{1t}}+\delta\right)\frac{M_t^{cont}-M_t^{label}}{M_t^{label}+(2\alpha-1)M_t^{cont}},$$

where λ is water-tissue partition coefficient, *E* = water extraction fraction, δ is a component that accounts for exchange between tissue water and macromolecules (e.g., 0 in one coil ASL), T_{1t} is T_1 of tissue, M_{cont} and M_{label} are magnetization in control and labeled states, and α is spin tagging efficiency (2). To evaluate whether this equation is valid for CBF quantification, we measured CBF at different magnetization transfer (MT) effects with and without the suppression of blood spins during data collection.

Methods

Ten male Sprague-Dawley rats weighting 300-350 grams were initially anesthetized with 5% isoflarine, orally intubated, and then maintained with 1.5% isoflarine in 50:50 O_2:N_2 mixture. Rectal temperature, blood pressure and blood gases were maintained within physiological ranges.

All MRI measurements were performed on a 9.4T Varian system. Two actively decoupled surface coils were used; a neck coil for spin labeling, and a surface head coil for data detection. To measure CBF values with different MT effects within the same animal, we developed a pulse sequence to induce various MT effects using the two-coil system without changing spin tagging (Fig. 1). This pulse sequence repeatedly applies a pair of 100-ms RF pulses during 8-sec spin labeling period; one is applies to the neck coil for spin labeling (DEC), and the other is applied to the head coil to generate various MT effects (TX). The frequencies of control and transmitter pulses were +8500 Hz with respect to the resonance frequency. Spin tagging frequencies were -8500 Hz. In order to generate different MT effects, five RF power level of transmitter were used in a randomized order, achieving steady-state magnetization to a level of ~0.27M_0 (Fig 2). Since active decoupled switching coils were used, there are no asymmetric MT effects between the two images, which were confirmed experimentally. After spin preparation, single 2mm-thick images were acquired using an adiabatic-version double-spin echo EPI technique with TE of 25 ms, TR = 10 s, matrix size of 64 x 32, and FOV of 3.0 x 1.5 cm^2. To investigate the effect of blood pool contribution to CBF quantification, studies in the same animal were performed before and after injection of 1 mg/kg iron oxides (MION). For simplicity, E is assumed to be 1.0 and λ is 0.9 ml/g. In vitro T_1 and T_2 of arterial blood withdrawn before and after the injection of MION were determined using standard spectroscopic techniques. MION decreased blood T_1 from 2.30 ± 0.19 to 1.14 ± 0.06 sec (n=10). Before the injection of MION, we assumed α=0.4 (because of 50% duty cycle), while after injection, α decreased to 0.4*exp(-transit time*ΔR_1 of blood) = 0.32. Relative spin tagging efficiencies were confirmed experimentally. A ROI of the cerebral cortex was chosen, based on anatomic images. The same ROI was used to compare CBF values measured with different MT effects.

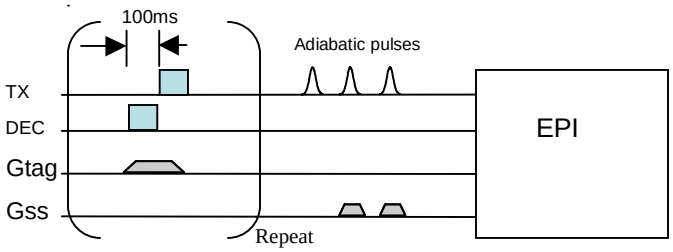

Fig 1. The sequence diagram of this experiment. TX; transmitter channel for generating MT effects and obtaining images (head coil). Gtag; z-axis gradient of 1G/cm. Gss; slice-select gradient. DEC; arterial spin labeling channel (neck coil)

Results and Discussion

By increasing a RF power level, the saturation of macromolecule and resulting reduction of water proton signal were observed, as expected (3). Magnetization transfer ratio (MTR) was defined as $(M_0-M_{sat})/M_0$ where M_{sat} and M_0 is magnetization with and without MT effects, respectively (4). After injection of MION, in vitro T_2 of arterial blood changed from 40.0 ± 1.8 ms to 14.1 ± 3.6 ms (n = 10). Effective T_2 of blood in vivo is shorter than in vitro due to flow contribution. Therefore, the blood signal in our spin-echo measurement with TE of 25 ms can be significantly reduced after the MION injection. Without any MT effect, the measured CBF in the rat cortex was 1.89 ± 0.29 and 1.09 ± 0.19 ml/g/min (n = 10) without and with MION, respectively. This difference can be potentially due to the many factors including blood contribution. However, CBF values with various MT effects should be similar if the model is correct. The calculated CBF remained constant when blood signal was suppressed by MION (squares), while the calculated CBF was closely dependent on MT effect before the injection of MION (circles). When the blood component is minimized, the small difference between CBF values measured with different MT effects is likely due to using an assumption of δ = 0 for all calculation. Dependence of the calculated CBF values on MTR was calculated; 0.1 and 1.05 ml/g/min with and without the suppression of blood pool signals, respectively. Regardless of the suppression of the blood pool during the data acquisition (not during the spin tagging time), the extraction fraction should be constant. Thus, the large difference of CBF values with MT effects without MION can be explained by different vascular contributions. Magnetization transfer effects saturate water signals in brain tissue with a high density of macromolecules, while, it has a little effect in tagged blood spins due to constant flowing of inverted spins from outside FOV of the surface coil into the imaging slice and a low density of macromolecules (3). Thus, relative contribution of blood signals increase when MT effect is larger, resulting in a larger overestimation of CBF. Especially in MTR level of 0.73, CBF measurement without the suppression of vascular pool can be overestimated by about 40%.

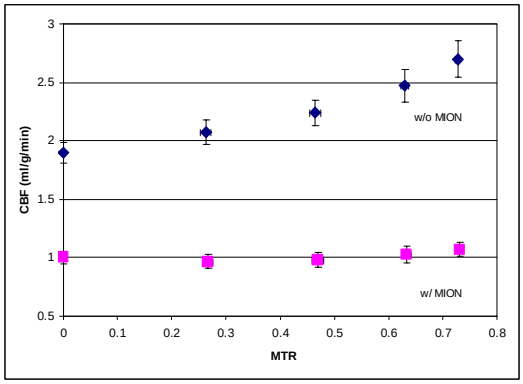

Fig 2. Dependence of CBF changes on different MTR with and without MION (error bars; SEM)

References

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