Cerebral arterial blood volume and blood flow measured by arterial spin labeling with diffusion-weighted gradient and MOTIVE methods

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Introduction

Diffusion-weighted gradients can reduce arterial blood vessel contribution to arterial spin labeled (ASL) images (1). According to diffusion-weighted spectroscopic studies of blood substitutes (perfluorocarbons), *b*-value of >50 s/mm^2 can eliminate arterial blood contribution (2). By using flow-crushing diffusion-weighted gradients, this phenomenon can be utilized to determine the amount of suppressed signals in ASL, which is equivalent to cerebral arterial blood volume (CaBV). A theoretical framework to determine arterial blood volume with diffusion-weighted gradients was described in this study. Similarly, CaBV can be measured by the *mo*dulations of *ti*ssue and *ve*ssel signals (MOTIVE), which relies on ASL with various magnetization transfer (MT) levels (3). To determine whether the CaBV value measured by both diffusion-weighted gradient and the MOTIVE methods is consistent, ASL experiments were performed in the same animals with a *b*-value of 0 and 100 s/mm^{2}with four different MT levels each.

Theory

In ASL measurements, control and arterial spin labeled images (S_{lab} and S_{cont}, respectively) are acquired. Difference between the two images (referred to as "ASL" signal) with diffusion-weighted gradient of *b* value at echo time TE, $\Delta S(b)=(1-v_a)\Delta S_{tissue}(TE) + v_a\Delta S_a(TE)$, where v_a is the arterial blood volume fraction, and ΔS_{tissue} and ΔS_a are signal changes in tissue and arterial blood induced by ASL. When the arterial blood contribution is removed by diffusion-weighted gradients, only the tissue compartment remains. By comparing normalized ASL signals with ($\Delta S(b)/S_{cont}(b)$) and without ($\Delta S(b=0)/S_{cont}(b=0)$) diffusion-weighted gradients, v_a can be calculated as $v_a = [\Delta S(b=0)/S_{cont}(b=0) - \Delta S(b)/S_{cont}(b)]/[2\alpha - \Delta S(b)/S_{cont}(b)]$ where α is the spin labeling efficiency. CBF was determined by CBF = $\lambda/T_1\cdot(\Delta S/(S_{lab}+(2\alpha-1)\cdot S_{cont})$ where λ is the blood-to-tissue partition coefficient (0.9 ml/g).

Methods

Eight male Sprague-Dawley rats (300-350 g) under 1.5% isoflurane were used. To modulate CBF and CaBV, the ventilation rate was adjusted to change the pCO_2 level. MRI measurements were performed on a 9.4T/31 cm Varian system with two actively switching detunable surface coils (neck and head coils). To modulate a level of MT effect without changing arterial spin labeling efficiency, a pair of RF pulses, a 100-ms spin labeling pulse in the neck coil followed by a 100-ms MT-inducing pulse in the head coil, was repeated during 8 sec (see Fig.1 of ref. 4). ASL images were acquired using a diffusion-weighted, adiabatic single-shot double spin echo planar imaging sequence with TE of 36 ms, TR = 10 sec, slice thickness = 2 mm, matrix size = 64 × 32 and FOV = 3.0 × 1.5 cm^2. Diffusion-sensitizing unipolar gradients were applied around the first 180° RF pulse. *b* values of 0 and 100 sec/mm^2 and four MT levels (M_{sat}/M_0) of 0, 0.27, 0.45, and 0.65 were applied in a randomized order. CaBV was determined as shown at the theory section. CBF value was calculated from data at each condition (two b values, four MT levels). By comparison, CaBV and CBF were also calculated by the MOTIVE method (3).

Results and Discussion

Eleven data sets were obtained from 8 animals under various pCO_2 levels, varying from 25~50 mmHg. Fig. 1 shows calculated CBF values in the cortical ROI, obtained from data with a *b* value of 0 (Fig. 1a) and 100 sec/mm^2 (Fig. 1b). Each symbol indicates each data set with four MT levels. When no crushing gradient was used, calculated CBF value (or $\Delta S/S$) increases as MT level increases (Fig. 1a) because of relatively higher arterial blood volume contribution in a higher MT level. When a *b* value of 100 sec/mm^2 was used, the calculated CBF values (or $\Delta S/S$) were independent of MT levels (Fig.1b). It can be explained that the contribution of arterial volume was removed by the appropriate strength of diffusion-weighted gradient. The calculated CaBV is independent of MT levels (confirmed experimentally), and its average across 11 data sets is 0.9 ± 0.3 % (mean ± SD). Arterial blood volume and CBF measurements between MOTIVE and diffusion-weighted gradient methods were compared under various pCO_2 conditions (Fig. 2). These two CaBV values were highly correlated with each other (R^2 = 0.82) (Fig. 2a). However, CaBV values obtained from the diffusion-weighted gradient method were significantly lower than those with MOTIVE method ($p < 0.01$). Similarly, calculated CBF values with b = 100 sec/mm^2 were highly correlated with (R^2 = 0.98) but slightly higher (7%) than the CBF values from the MOTIVE method (Fig. 2b). This difference suggests that the both methods have slightly different vascular contributions to the "apparent" tissue pool. The MT-based MOTIVE approach may separate signals from vasculature and tissue better than the diffusion-weighted approach. The CaBV in the MOTIVE measurement may contain a portion of capillary blood volume, and/or a diffusion-weighted gradient of 100 sec/mm^2 may not remove the entire arterial blood volume. Nonetheless, the ASL method with and without crushing arterial blood signals can detect CaBV and CBF, and can be easily applicable to humans.

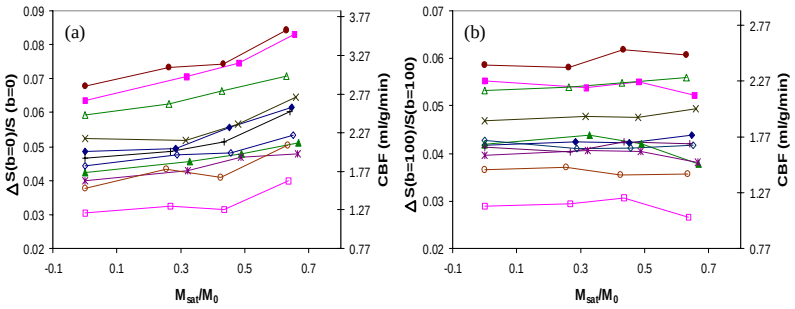

Fig.1. ASL data at the graded MT levels with a b-value of 0 (a) and 100 sec/mm^2 (b).

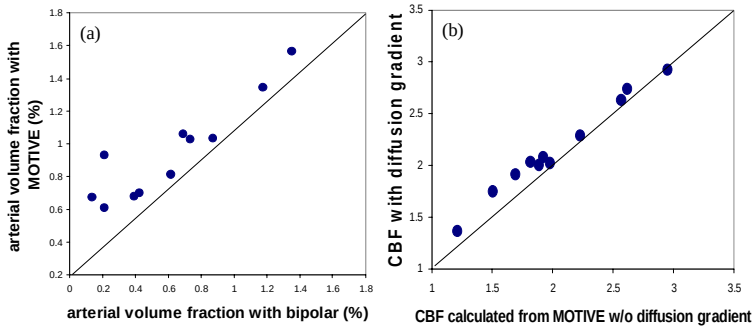

Fig. 2. The comparison of arterial blood volume fraction (a) and CBF (b). CBF: ml blood/g tissue/min.

References

1. Ye et al., MRM 37:226-235 (1997) 2. Duong et al. MRM 43:393-402 (2000) 3. Kim & Kim, ISMRM abstract submitted (2005) 4. Kim & Kim, ISMRM abstract 1372 (2004)

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