Source of IVIM in Arterial Spin Labeled Signals

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Introduction

When pulsed magnetic field gradients are applied, MR signal intensity decreases due to diffusion and intravoxel incoherent motion (IVIM) of water (1-2). Le Bihan et al. found that two distinct apparent diffusion coefficient (ADC) components exist in vivo (2): one is pure diffusion of water molecules, and the other is presumably related to incoherent motions due to capillary blood flow ("pseudo-diffusion"). According to this theory, pseudo ADC value is correlated to perfusion rate and its volume fraction is capillary volume fraction. By applying IVIM into arterial spin labeling (ASL) measurements, Silva et al. separated un-extracted labeled spins in capillaries from tissue spins, and measured an extraction fraction (E) of labeled water into tissue (3). Since the fraction of the pseudo ADC component is much larger than capillary blood volume fraction (1-2%), the exact source of fast pseudo ADC is controversial (4). We propose that the fast pseudo-diffusion component occurs from arterial blood vessels, especially in ASL measurements. To test our hypothesis, IVIM effects in ASL measurements were measured in conjunction with the modulation of contribution of arterial blood vessels using variable post spin-labeling delay time (5).

Theory

In the conventional IVIM model, normalized ASL signal can be described as $\Delta S(b)/\Delta S(0) = (1-f)\cdot e^{-bD} + f\cdot e^{-bD^*}$ where f is the fraction of the fast pseudo ADC component. In Silva et al's study, f is considered as the fraction of un-extracted labeled spins, i.e., 1 – E (3). In our model, f is assumed to arterial volume fraction v_a. To evaluate whether f is related to E or v_a, IVIM experiments can be combined with different post-spin labeling delay times w under MT saturation. Contribution of arterial blood to ΔS(b) can be reduced by introducing a post-labeling delay time w. The relative signal contribution of the arterial blood pool will be dependent on w, transit time from the carotid artery to arterial vessels in a pixel τ_a, and transit time from the carotid artery to capillary τ_c. Contribution of arterial blood volume will be reduced (Fig. 1) in the case of $\tau_a < w < \tau_c$, and eliminated when $w > \tau_c$. Based on our model, normalized ΔS(b) will be bi-exponential decay as a function of b value until $w < \tau_c$, and turn to a single-exponential decay afterwards. If f is related to E, f should be constant, irrespective of w.

Methods

Ten male Sprague-Dawley rats (300-350g) under 1.5% isoflurane were used. All MRI measurements were performed on a 9.4T/31 cm Varian system. Two actively detunable surface coils were used; one was for generating arterial spin labeling in the carotid arteries in the neck, and the other for generating MT effects and obtaining brain images. To generate a ~60% MT saturation level without changing the ASL efficiency, a pair of RF pulses, a 100ms-long spin tagging pulse in the neck coil followed by a 100ms MT-inducing pulse in the head coil, was repeated during 8-sec (see Fig.1 of ref. 6). The MT effect enhance the relative contribution of arterial blood to ASL signals because MT selectively suppress tissue signals. All 2-mm thick coronal images were acquired using a diffusion-weighted, adiabatic single-shot double spin-echo planar imaging sequence with TE of 36 ms, TR = 10 sec, matrix size = 64×32 and FOV = 3.0×1.5 cm^2. Diffusion-sensitizing gradients were applied around the first 180° RF pulse. To examine characteristics of IVIM in ASL data, both w and diffusion-weighted gradient strength were modulated at a randomized order. In five animals, w was 0.1, 0.2, 0.3 0.4 and 0.5 sec and b value was 0, 3, 10, 20, 30, 50, 100 and 150 sec/mm^2, and in another five animals, w was 0.1, 0.3, 0.5 and 0.7 sec and b value was 0, 3, 10, 20, 30, 50, 70, 100 and 200 sec/mm^2.

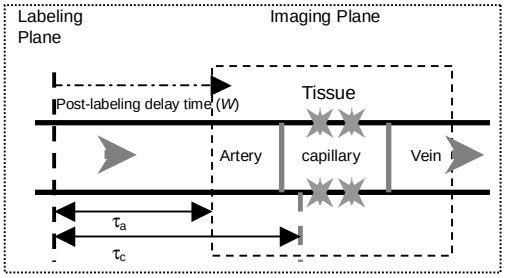

Fig. 1. Delivery scheme of labeled arterial spins. Labeled spins move into the artery pool, then capillaries and exchange with tissue water. τa: transit time to artery, τc: transit time to capillary.

Results and Discussion

To determine the relationship between the D* component in the IVIM model and post-spin labeling delay time, normalized ASL signal (log scale) was plotted as a function of b value in Figure 2. All data across animals were averaged (n = 10 for post-labeling delay time of 0.3 and 0.5 s, and n = 5 for 0.7 s delay). ASL data in a logarithmic scale was linearly fitted to determine whether a single diffusion compartment is observed. If an intercept of the fitted line is close to zero, a single diffusion coefficient can explain the IVIM data, which is the case of w of 0.7 s. However, data with a shorter delay time has a non-zero intercept, suggesting that the IVIM signals consist of two compartments. When a post-labeling delay is sufficiently long for unlabeled fresh spins to fill the arterial blood volume, the contribution of the arterial compartment will be eliminated, while the tissue compartment is dominant. To analyze IVIM data quantitatively, ΔS(b)/ΔS(0) was fitted by a bi-exponential function in each animal; apparent diffusion coefficients of fast and slow diffusion components and their fractions were determined (Table). The ADC value of the slow diffusion compartment is invariant to w, suggesting this pool is indeed tissue, which is consistent with previous observations (3). If the fast D* component is due to extraction fraction of water spins, its fraction should be independent of w. However, D* and its volume fraction (v_a) decreased as w increased, demonstrating that the fast diffusion compartment is closely correlated to w. When the post-labeling delay time increases, the contribution of arterial volume to ASL signal reduces, resulting in decreases in effective D* and f. Our data suggest that the fast component of IVIM in ASL is not likely from un-extracted spins in capillaries and veins, and water extraction fraction may not be determined. Since the fast D* component originates from arterial blood, arterial blood volume and CBF values can be determined by modulating flow-crushing gradient strength.

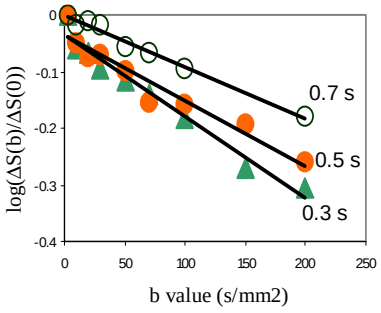

Fig 2. ASL signal attenuation curves in logarithm scale versus the b values with the modulation of post-labeling delay time. Triangles: 0.3 s delay, filled circles: 0.5 s, open circles: 0.7 s.

Delay time (sec)	0.1	0.2	0.3	0.4	0.5	0.7
D (× 10^{-3} mm^2/sec)	1.2 ± 0.3	1.2 ± 0.3	1.3 ± 0.4	1.2 ± 0.1	1.2 ± 0.3	1.0 ± 0.3
Fraction (v_a)	0.17 ± 0.03	0.16 ± 0.02	0.11 ± 0.02	0.05 ± 0.03	0.07 ± 0.05	0.02 ± 0.08
D* (× 10^{-3} mm^2/sec)	400.8 ± 275	366.5 ± 342	311.0 ± 250	122.4 ± 98	198.7 ± 96	37.1 ± 28

References

1. Stejskal and Tanner,*J Chem Physics* 42:288-292 (1965) 2. Le Bihan et al., *Radiology* 168:497-505 (1988). 3. Silva et al., *MRM* 38:232-237 (1997). 4. King et al., MRM 24:288-301 (1992) 5. Alsop and Detre, *JCBFM* 16:1236-1249 (1996) 6. Kim and Kim, *ISMRM abstract* 1372 (2004)

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