Magnetization dispersion effetcs on quantitative perfusion imaging for pulsed and continuous arterial spin labeling

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Introduction:

For quantitative perfusion imaging with arterial spin labeling (ASL), kinetic models were developed taking into account the dynamics of the inflow, magnetization decay and outflow of the magnetically labeled spins carried by blood flow [1]. The arterial transit time (δt) is a key local variable which has to be included in the kinetic models to estimate the CBF (*f*). Most experiments assume an instant and uniform arrival of the tagged blood without any dispersion of the magnetization. However, due to cardiac pulsation, non-uniform cross-sectional flow profile and complex vessel networks, the distribution of δt has a statistical nature instead of being uniform [2]. Based on perfusion territory maps, it has been reported that the dispersion of the tag is higher in regions which have a longer δt because of the longer time for dispersive effects to build up [3]. In this study, we have investigated the regional effects of magnetization dispersion on quantitative perfusion imaging for varying distances between tagging and imaging regions by implementing pulsed (PASL), pseudo-continuous (PCASL) and dual-coil continuous (DC-CASL) arterial spin labeling encoding schemes. Longer distances between tagging and imaging region as typically used for continuous tagging schemes were found to cause larger magnetization dispersion effects and to enhance the regional bias on the quantitative CBF estimation up to 37% in comparison to assuming plug flow.

Methods:

Five male subjects were scanned on a 3T Siemens MAGNETOM Trio TIM scanner using a 12-channel head coil. Four 8 mm axial slices were positioned at the level of the corpus callosum and were acquired in ascending order. For PASL, the PICORE-QUIPSS II encoding scheme was used with a spatial tag width of 10 cm and 700 ms of tag duration. A C-shaped FOCI pulse was used for inversion of the tagging slab positioned 3cm proximal to the image slices. For PCASL experiments, a single-channel birdcage coil was used to apply a train of 24° Gaussian pulses with aduration of 600 μs to adiabatically invert the flowing blood separated by a 600 μs gradient rephasing delay. The tagging plane was positioned 8cm proximal to the magnetic isocenter and the tag duration was set to 500ms. For DC-CASL, RF was transmitted using two rectangular surface coils with a size of 5 x7 cm positioned at the neck of the subject right above the carotid arteries, 17cm proximal to the isocenter, and the tag duration was set to 500ms. A total RF power of 8W corresponding to a flip angle of 55°/ms at the position of the carotid arteries was used with an optimum gradient strength of 1.8 mT/m. To span the kinetic curves for each of the ASL schemes, we used nine inversion times (TI)/post labeling delay (PLD) times ranging in [100, 250, 500, 750, 1000, 1250, 1500, 1750, 2000] ms. For all 3 sequences, 30 alternating control and tag images were acquired with the following sequence parameters: [TR/TE= 3000/19 ms, voxel size= 3.5 x 3.5 x 8 mm, 64x64 matrix, partial Fourier factor=6/8, EPI read-out]. Each data set was fit to the standard [1] and the Hrabe-Lewis [2] model using a least-square fit. Five ROIs were defined on the gray matter masked CBF images for each subject based on the territorial characteristics of the major brain feeding arteries, namely anterior cerebral artery (ACA), posterior cerebral artery (PCA) and middle cerebral artery (MCA).

Results:

Fig. 1 shows the five defined ROIs overlaid on the perfusion image for a representative subject. Fig. 2 shows the δt maps calculated by fitting the Hrabe-Lewis model to the measured kinetic curves for PASL (left), PCASL (middle) and DC-CASL (right) sequences. The δt values exhibit regional variations across the brain and the increase in distance between the tagging and the imaging region for different ASL encoding schemes (PASL: 3cm, PCASL: 8cm, DC-CASL: 15-17cm) leads to higher δt values. A quantitative comparison of the effects of longer transit times on the magnetization dispersion is shown in Fig. 3 (top), where the fitted δt values from the Hrabe-Lewis model were plotted against the fitted σ values. There is a strong correlation between δt and σ with r=0.8 for PASL, r=0.94 for PCASL and r=0.93 for DC-CASL (r: Pearson correlation coefficient). Fig. 3 (bottom) compares the effect of increased σ on the ratio of CBF estimates from the Hrabe-Lewis and standard models for different ASL sequences. It is obvious that with increasing σ, the relative difference in CBF estimation between the Hrabe-Lewis and the standard model becomes higher. Fig. 4 (top) shows the regional variations of the fitted δt values obtained with different sequences. Fig. 4 (bottom) shows the ROI-based comparison of the estimated CBF values obtained with different sequences. ROI5, close to the PCA, has the highest CBF values, closely followed by ROI4 and ROI3, which are located near the MCA. ROI2, positioned close to the ACA, has the lowest CBF estimation. While the values found with DC-CASL seem to be slightly higher than those determined with PCASL and PASL for all ROIs, there is no significant difference between the CBF estimates of different sequences.

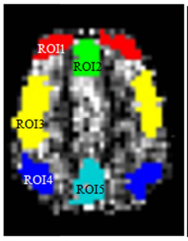

Figure1

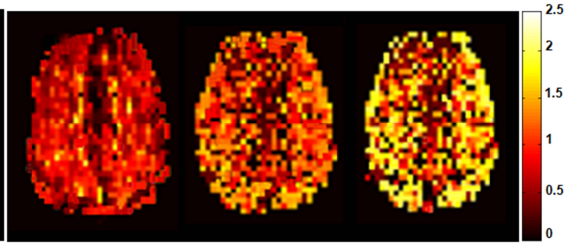

Figure 2

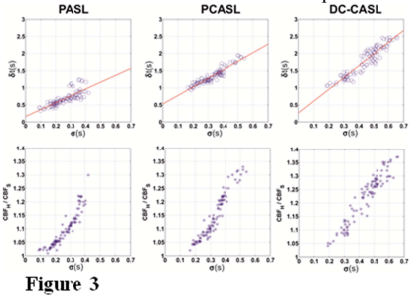

Figure 3

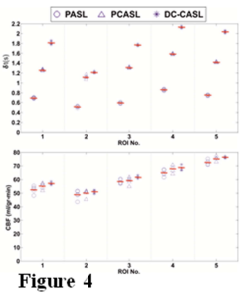

Figure 4

Discussion:

An accurate perfusion measurement over a wide range of brain regions requires a careful assessment of δt which varies across the brain due to the distribution of velocities and path lengths between the tagging and imaging regions. For continuous tagging schemes, the longer distance between the tagging and imaging planes leads to larger dispersion effects potentially amplifying the regional bias on the CBF estimation. The strong correlation between δt and σ shown in Fig.3 implies that the effect of magnetization dispersion is more pronounced for longer transit times because there is more time for dispersion to accrue. As a consequence of increasing σ, relative differences in CBF estimation between Hrabe-Lewis and standard kinetic model become higher. Fig. 4 (top) confirms what we expect from the arterial anatomy as shorter arrival times were found for the major branches of ACA [3]. The order of the δt distribution among the ROIs is also almost equal for all sequences where ROI2 has the shortest δt and ROI4 has the longest, which is clear evidence that the structural pattern of transit time distribution is basically independent of signal acquisition parameters (i.e. tag duration, distance between the tagging and imaging region, encoding scheme). There is a significant ($p < 0.01$, two tailed t test) discrepancy between the perfusion of ROI5 close to the PCA and ROI2 close to ACA and the variation of CBF across the ROIs is consistent for different sequences. Similar to δt, the regional variation of CBF is also not affected by the differences in measurement methods.

References:

[1] Buxton, R.B. (1998), MRM, vol.40, pp. 383-396. [2] Hrabe, J. (2004), JMR, vol. 167, pp. 49-55. [3] Gallichan, D. (2009), MRM, vol. 61, pp. 686-695.