Quantification of Relative Perfusion Changes in Functional Brain Imaging Using Diffusion Weighted FAIR: Comparison of T_1 Model and Kinetic Model

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Introduction: Pulsed arterial spin labeling (ASL) techniques are useful in the measurement of relative cerebral blood flow changes (ΔrelCBF) because of their high temporal resolution. However, in order to quantify perfusion changes, at least three possible sources of systematic errors should be addressed. First, transit delay (δt), the time for the tagged blood to reach the imaging slice from tagging region, shortens during functional activation. Although FAIR (1) provides the thinnest spatial gap between the tagging region and the imaging slice, it is not clear whether a decrease in transit delay will cause significant overestimation of ΔrelCBF. Second, venous tagged blood in small veins might increase flow or volume during task state and artificially cause higher signal. Third, the optimal inversion time (TI) needs to be chosen. In the T_1 model (2), it was shown that maximum FAIR signal is achieved at TI $\approx T_1$ of the brain tissue. However, in the kinetic model (3), the signal peaks at the time when the tail of the tag reaches the imaging slice, which in turn is determined by the size of the tag. In FAIR, the proximal tag size is set by the sensitivity of the RF coil at the proximal end, which also sets the limit for the non-slice selective tag width. The time width (τ) (tag size / average blood velocity along the tag axis through tagging region) of a tag also decreases during functional activation. The appropriate TI would need to be even shorter for the tagged blood to stay in the arrival phase so that the signal would be proportional to CBF. This contradicts the TI selection predicted by the T_1 model. Moreover, the perfusion signal will be overestimated because of the presence of intravascular signal at short TI. In this study, we use diffusion weighting (DW) to minimize the signal contribution from the venous flow and the intravascular signal, and apply both the T_1 and kinetic models for the quantification of ΔrelCBF to address these issues.

Methods: For both T_1 and kinetic model simulations, T_1 of gray matter = 1.33 sec, T_1 of blood = 1.5 sec, λ = 0.9 ml/g, TR = 2 sec, and CBF = 60 and 114 ml/100ml/min (ΔrelCBF = 90%) during task and rest states were used. For the kinetic model, δt was assumed to be 0.15 and 0.3 sec, and τ to be 1 and 1.2 sec for task and rest states respectively. All images were acquired using a 3T Bruker Biospec 30/60 scanner. A local head gradient coil and an endcapped quardrature birdcage coil were used. Single-shot blipped gradient echo EPI acquisition was used with FOV 24 cm and 8 mm slice thickness. The inversion slice was 20 mm thick. TI was varied from 0.5 to 1.5 sec. TE = 43.2 ms and TR = 2 sec. For DW studies, a bipolar gradient pulse was applied in the slice-selective direction with a total duration of 16 ms and b factor of 1 s/mm^2 (4) corresponding to a velocity encoding of 1.37 cm/sec. Typically 150 repetitions were acquired with 60 sec on/off finger tapping episodes. FAIR perfusion images were created by performing pairwise subtraction of consecutive images. Functional perfusion images were created by subtraction of the averaged resting and active state perfusion maps. Relative perfusion changes were calculated using both the T_1 and kinetic models.

Results: Fig. 1a and 1b show the simulated FAIR signal as a function of TI using the T_1 and kinetic models respectively. The calculated ΔrelCBF using only one TI point for both models are shown in Fig. 1c and 1d. The curves differ sharply. To correctly estimate ΔrelCBF, it is necessary, when using the T_1 model, to assume no δt, and, when using the kinetic model, to assume no *changes* in δt and the TI is shorter than the time when the end of the tag reaches the slice during activation. Using the simulated parameters, ΔrelCBF was significantly overestimated due to the decrease in δt. Fig. 2 shows FAIR images during rest at TI = 0.9 sec with and without DW and the corresponding functional maps. The signals in veins were destroyed with DW gradients. Fig. 3 shows the measured FAIR signal *vs.* TI during task and rest states with and without DW. The estimated δt using the kinetic model without DW is 0.006 and 0.15 sec during task and rest respectively, and 0.14 and 0.28 sec with DW respectively. The signal drops with DW, which indicates that some of the tagged blood is intravascular arterial or venous blood. The application of DW gives a better representation of the time when the end of the tag reaches the capillary bed. Fig. 3b shows the calculated ΔrelCBF using the T_1 model with each TI point and the kinetic model with at least two TI points to estimate δt first. The calculated ΔrelCBF using the kinetic model is 38.2% without DW and 88.7% with DW. This supports the idea that ΔrelCBF happens at the capillary level and half of the signal is intravascular. With only one TI data point, ΔrelCBF is underestimated without DW and mostly overestimated with DW, which essentially correlates with the simulation results.

Conclusions: The kinetic model better describes the perfusion signal obtained with pulsed ASL. To quantify ΔrelCBF with FAIR, diffusion weighting and two TI points before the tail of the tag reaches the imaging slice are necessary.

References:

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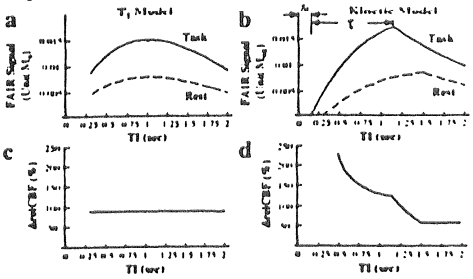

Figure 1. Simulated FAIR signal *vs.* TI using (a) T_1 model and (b) kinetic model. Calculated ΔrelCBF using a single TI point assuming FAIR signal follows (c) T_1 model and (b) kinetic model.

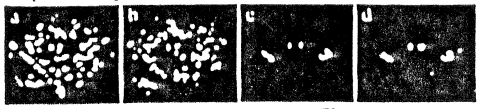

Figure 2. FAIR images during rest state at TI = 0.9 sec (a) without DW and (b) with DW, and (c, d) corresponding functional maps.

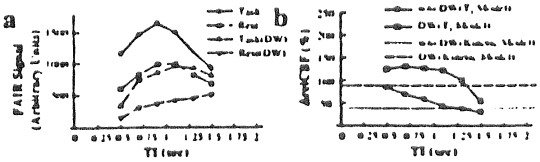

Figure 3. (a) Measured FAIR signal and (b) calculated ΔrelCBF

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