Effects of pulsatile flow on arterial input function and CBF quantification in continuous ASL

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[Introduction] Regional CBF is generally quantified by comparing arterial input function (AIF) and the regional distribution of the tracer. Although AIF is measured in PET and dynamic susceptibility contrast MRI by arterial sampling or time intensity curve, AIF on arterial spin-labeling (ASL) CBF signals is calculated assuming an arbitrarily shaped AIF without considering pulsatile flow profiles. Full width of AIF is on the order of a few minutes in PET and ~10 s in PWI, whereas it is only a few seconds in ASL. Thus, the flow dynamic profile of the short AIF may affect ASL CBF quantification. In this study, we modeled the effect of different flow profiles on the ASL signals. This model was applied to the arterial ASL signals obtained at different post-labeling delays with and without diffusion gradients in anesthetized monkey to model the AIF in cASL measurements. Finally, the potentially errors in CBF quantification was evaluated over wide ranges of hemodynamic and MRI measurement parameters.

[Models] Three common flow profiles in artery are illustrated in **Fig 1A** where τ is a labeling duration, T is a spreading time due to laminar, and ς is a cut-off ratio to approximate the plug flow profile. In square profile, $T \approx 0$ and in laminar, $\varsigma=1$. Profiles of labeled spins in a big artery can be expressed as $\Delta S/S = \kappa Ca(t)$; where κ is the labeling profile at the labeling plane. Three patterns of profiles can be expressed as; when $t < \omega + \tau - T/2$, $Ca(t) = \min[(\max[(1-(t-\omega-T/2)^2/T^2), 0]), 1]$, when $\omega + \tau - T/2 < t$, $Ca(t) = \max[(1-(t-\omega-\tau-T/2)^2/T^2), 0]$. **Fig 1B** shows the calculated flow profiles of labeled spins are shown with dashed lines and ones accounting for the T_1 decay of the blood (T_1b) with solid lines. Assuming 10-50cm/s of velocities in arteries, labeled spins momentarily pass through an arterial voxel at tracer arrival time to artery (δa). The ASL signals in a big artery can be, therefore, simplified as; when $\omega < \delta a - T/2 + \varsigma T$, $\Delta S/S = \kappa \exp(-\omega/T_1b)$, when $\delta a - T/2 + \varsigma T < \omega$, $\Delta S/S = \max[\kappa\{1-(\delta a-\omega-T/2)^2/T^2\}\exp(-\omega/T_1b), 0]$. $\Delta S/S$ taking into consideration the plug flow profile was calculated to estimate potential errors in CBF quantification due to flow dynamics.

[Materials and Methods] Rhesus monkeys (n=5, 5.8-7.5 kg) were anesthetized under 0.9-1.1% isoflurane. MRI was performed on a 3T Siemens Trio. CBF was measured using a separate neck coil for cASL (1) with and without a diffusion gradient (b=30 s/mm^2). The MRI parameters were: single shot GE-EPI, TR=4900ms, TE=32ms, FOV=96×96 mm, matrix=64×64, thickness 1.5mm, labeling duration 2.0s, labeling gradient of 0.3G/cm, and seven post-labeling delays (ω) for each imaging slice accounting for the slice acquisition time.

[Results & Discussion] **Fig 1C** shows the simulations of $\Delta S/S$ in the arteries vs post-labeling delays (ω). **Fig 2** shows $\Delta S/S$ maps at different delays (ω) without (**A**) and with (**B**) diffusion gradient. Assuming that the diffusion gradient could selectively eliminate the arterial component, arterial ASL maps (**Fig 2C**) were obtained by subtraction of the two images. The arterial $\Delta S/S$ as a function of ω are plotted for the middle cerebral artery (MCA) and the perforating arteries in the caudate in **Fig 3**. Among the three common flow profiles, plug flow pattern fit best for both arteries (**Fig 3**, dashed lines). The resultant parameters from fitting were: spreading times, T=1s in MCA and 0.6s in caudate, plug flow ratios, ς = ~0.5 in both lesions. The estimated potential errors of CBF without account for flow dynamics were simulated by comparing square and plug flow profiles (**Fig 4**). These results showed that the CBF without accounting for flow dynamics could contribute up to 5% error under normal CBF conditions (**Fig 4A**). Moreover, such errors became more significant at long transit time (δ), such as in diseased state. The errors could also become more pronounced at lower magnetic field where T_1 is shorter (**Fig 4B**).

[Conclusions] We detected and modeled the plug flow profile in the arteries in the monkey brain in our ASL measurements. Hemodynamic parameters were estimated to model the AIF on cASL. Errors in CBF due to pulsatile flow profile were up to 5% under normal blood flow conditions. However, these errors could become more significant at lower field where T_1 is shorter and with long transit time. The latter may have implications in cerebrovascular diseases where transit time is delayed.

References] 1) Zhang X et al. Neuroimage 34:1074-83, 2007. 2) Ye FQ et.al, Neuroimage 6: 104-112, 1997. 3) Alsop D and Detre J JCBFM 16: 1236-1249, 1996. 4) Gach HM et.al MRM 47: 709-719, 2002

Fig 1: Flow profiles and simulated ASL signals

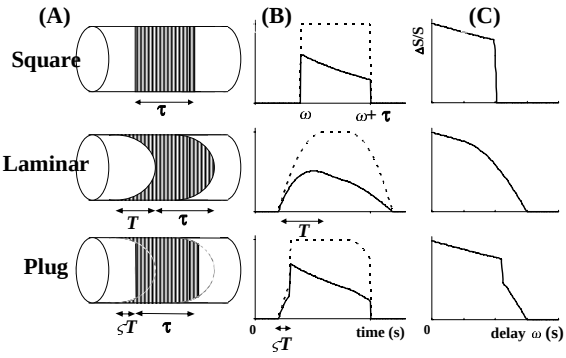

Fig 2; ΔS/S maps at short delay with b = 0, 30 mm^2 and difference map (arterial ΔS/S map)

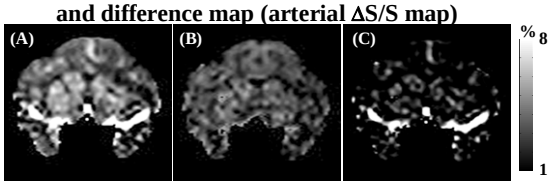

Fig 3: Fitting the model to ΔS/S in two arteries

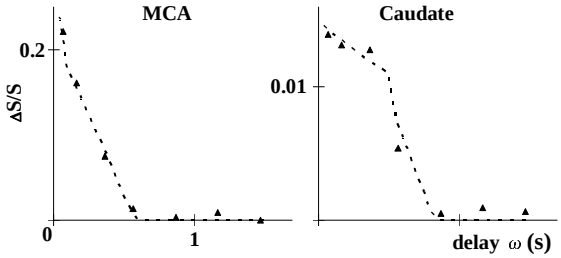

Fig 4; Errors in tissue CBF without account for flow dynamics