QUANTIFICATION OF FLOW RATES IN SHORT VESSEL SEGMENTS FROM ARTERIAL SPIN LABELING DYNAMIC ANGIOGRAPHY

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Target audience: Scientists and clinicians with an interest in angiography and blood flow quantification.

Introduction: Good collateral blood flow networks in the brain can ameliorate the damaging effects of cerebrovascular diseases, such as stroke [1]. Typically only qualitative information about the degree of vessel disease, and the existence and source of collateral blood flow can be obtained by conventional angiographic techniques [2]. The purpose of this work was to quantify blood flow rates in small segments of brain-feeding arteries from angiographic images. This was achieved through the fitting of a new mathematical model of blood flow within a pre-defined vessel-segment. The work focused on non-invasive, vessel-selective, arterial spin labeled (ASL) MR angiography (VEPCASL) [3], although the method is also potentially applicable to data from other angiographic techniques such as X-ray digital subtraction angiography (DSA).

Theory: Pseudo-continuous (PC) ASL produces a tagged blood bolus with an 'ideal' box-car shaped signal profile. Assuming an axial blood velocity profile of $U(r) = \bar{U}((\gamma+2)/\gamma)(1-(r/R)^{\gamma})$, where $\bar{U}$ is the average axial velocity, R is the vessel radius, and γ is a measure of bolus dispersion: $2 < \gamma < \infty$. The transit of the ASL blood bolus through a vessel is modeled using the mass transport equation to obtain an expression for the concentration of the ASL tag at any point in the vessel. Integrating this expression over the volume of a region-of-interest (ROI) gives the shape of the tracer concentration-time curve in the ROI. This shape is scaled by S_0, a calibration factor giving the ASL signal per unit volume of tagged blood, and modified with an attenuation term $R(t)$ to account for longitudinal relaxation of arterial blood, and signal attenuation caused by RF excitation pulses. Before imaging commences at $t = t_0$: $R(t < t_0) = exp(-t/T_{1b})$ and after: $R(t > t_0) = \exp(-t_0(1/T_{1b} - 1/\tau))exp(-t/\tau)$ where $1/\tau = 1/T_{1b} - \ln(\cos\alpha)/T_R$. T_{1b} is the longitudinal relaxation time of arterial blood, and τ is the effective longitudinal relaxation time when RF excitation pulses are applied with a flip angle α and repetition time T_R. Putting these terms together gives the final model (eq. 1), which is shown in Fig. 1.

$$S_{ROI} = S_0 R(t) BF \cdot \left\{-[t_1 - t][1 - t_1/t]^{2/\gamma} \cdot H(t - t_1) + [t_2 - t][1 - t_2/t]^{2/\gamma} \cdot H(t - t_2) + [t_3 - t][1 - t_3/t]^{2/\gamma} \cdot H(t - t_3) - [t_4 - t][1 - t_4/t]^{2/\gamma} \cdot H(t - t_4)\right\} \quad (1)$$

where $t_2 = \gamma x_2/\left((\gamma+2)\bar{U}\right)$, $t_3 = \gamma(x_1 + \Delta)/\left((\gamma+2)\bar{U}\right)$, $t_4 = \gamma(x_2 + \Delta)/\left((\gamma+2)\bar{U}\right)$ and $t_1 = \gamma x_1/\left((\gamma+2)\bar{U}\right) = t_2 + t_3 - t_4$. x_1 and x_2 are the distances from the tagging plane to the start and end of the ROI respectively, and Δ is the spatial width of the ASL tagged blood bolus.

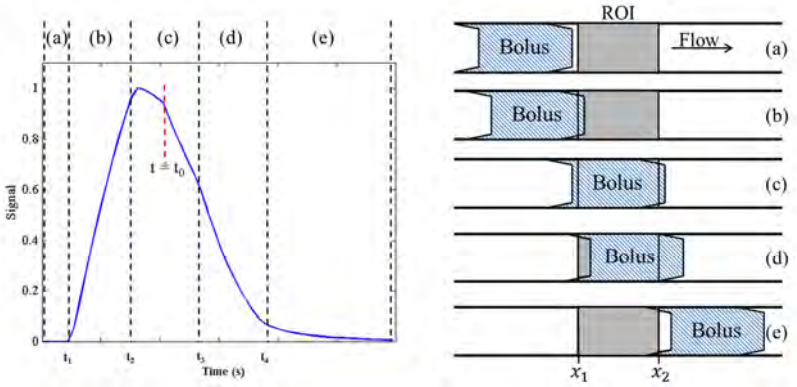

Fig. 1 – The modeled ASL MR signal in an ROI (left) and the stages of the bolus' passage, through the ROI, that create it (right)

Materials and methods: VEPCASL angiography images of both a flow phantom and six healthy volunteers, scanned under protocols approved by local ethics and institutional committees (data acquisition performed, and sequence parameters described, by [4]), were used to test the flow quantification method. The flow model (Fig. 1) was fitted to the measured VEPCASL signals summed over an 'input' region (in) and over a larger downstream region (ROI) (Fig. 2). γ was set to 12 based on initial experiments on the flow phantom data. For each image time series the calibration factor S_0 was calculated by [5]. Assuming no dispersion of the ASL bolus occurs between the 'input' region and the downstream ROI, i.e. the bolus width remains constant between the two regions: $t_{2in} = t_{4in} + t_{2ROI} - t_{4ROI}$. Curve fitting was achieved by minimizing the root-mean-squared difference between the measured and modeled region signals. A total of six parameters were estimated from curve fitting: BF, t_{4ROI}, t_{3ROI}, t_{2ROI}, t_{4in}, t_{3in}.

Results: The model was able to successfully describe the shapes of the signals measured in the flow phantom at a range of flow rates. Accurate quantification of the water flow rates in the phantom could be achieved using vessel segments as short as *20 mm* (2 *mm* 'input' region, 2 *mm* gap, and 16 *mm* downstream ROI). Using such a region, water flow rates in the phantom were calculated to within ±15% of the known values. The flow model fitted the volunteer data well and the flow rates estimated were all physiologically plausible. Fig. 2 shows example data from one healthy volunteer. Assessment of blood flow in the left posterior cerebral artery (LPCA) showed a large proportion of the flow was tagged in the left internal carotid (LICA). The left posterior communicating artery (LPCoA), a vessel that often appears absent on MR angiograms, provided a collateral route in this case.

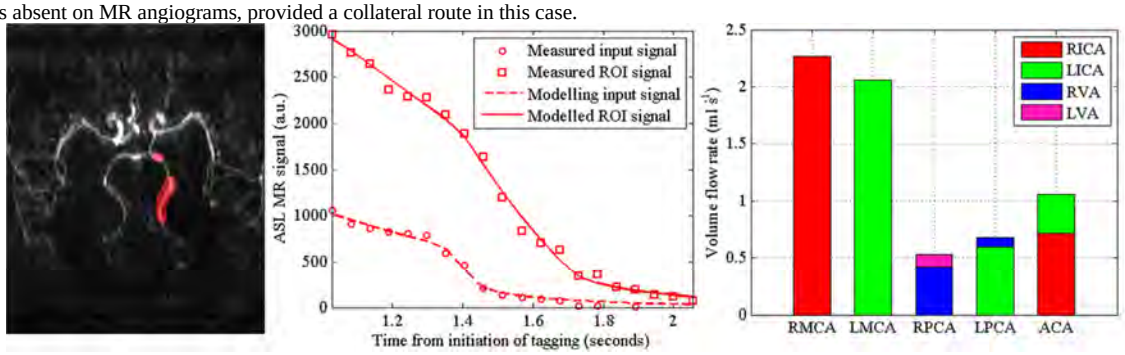

Fig. 2 – Transverse view of the circle of Willis of a healthy volunteer (left), red highlighting indicates the 'input' region and downstream ROI considered. Signals from blood tagged in the LICA, measured in the two regions, are shown with the associated optimal flow models (center). A bar chart shows blood flow rates estimated from the angiogram (right). The bar colors indicate the afferent vessel origin of the detected blood.

Discussion: The flow model described here allows the quantification of blood flow rates in much smaller arterial segments than those used by previously described methods [5,6]. Hence, this method could be used to quantify blood flow rates along the whole of the arterial tree supplying the brain. This would potentially allow vessel segments not directly visible on the images to be inferred based on differences in flow in connected vessels.

Conclusion: Estimation of blood flow rates in the vessels of the circle of Willis was achieved by fitting the flow model (Fig. 1) to measured VEPCASL signals. Collateral flow through the LPCoA was observed and quantified based on its effect on the estimated flow rate in the left posterior cerebral artery (LPCA).

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References:
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