Quantification of Vessel-Encoded Arterial Spin Labeling Dynamic Angiography with Auto-Calibration

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Introduction: It was recently shown [1] that vessel-selective dynamic angiograms of the cerebral vasculature can be generated non-invasively by combining a vessel-encoded pseudo-continuous arterial spin labeling (VEPCASL) preparation [2] with a time-resolved FLASH readout. This technique can provide useful information about the morphological and functional status of brain feeding arteries in patients with cerebrovascular disease. However, the resulting images have so far been qualitative, making it difficult to make firm conclusions from the data (e.g. assessment of whether a low signal level is truly due to low blood volume or due to T_1 decay of the VEPCASL label) or to compare across subjects. Here a framework is described in which such data can be quantified to give volume flow rates (ml/s) in each major artery along with arrival time and bolus dispersion information.

Theory: Unlike conventional ASL, where the labeled blood water accumulates in the tissue, the signal measured in angiographic acquisitions arises only from arterial blood passing through the voxel. Thus it is only possible to measure arterial blood volume and not flow rate from the time series in any voxel. However, it is possible to estimate volume flow rate if an ROI can be defined for any given vessel in which the entire labeled bolus of blood water is present. By estimating the volume of blood within this ROI, V, and knowledge of the VEPCASL pulse train duration, τ, the volume flow rate can be calculated: $F = V/\tau$. In order to define an ROI containing the entire bolus, τ must, in general, be small, leading to poor SNR. However, it is possible to use a parameterized model to fit to the signal time series measured with a longer τ. A signal that would have been obtained with a small τ can then be simulated. The signal was modeled as: $S(t) = A\,[c_{ideal}(t) \otimes D(t')]\,R(t)\,T(\delta_t)$ (Eq. 1), where $A = S_0\,v$ is a scaling parameter (S_0 is the calibration factor describing the signal per unit blood volume and v is the blood volume within the voxel); $c_{ideal}(t)$ is the ideal arterial input function: rect, with onset at the blood arrival time δ_t and duration τ; $D(t')$ is a gamma variate dispersion kernel (parameterized by a sharpness, s, and time to peak, p [3]); $R(t) = \cos^{N(t)}\alpha$ is the FLASH RF modulation function (where $N(t)$ is the number of RF excitation pulses of flip angle α experienced by the magnetization at time t); and $T(\delta_t) = \exp(-\delta_t/T_{1,blood})$ accounts for the T_1 decay of the VEPCASL label.

Calibration: Calibration factors for these data are derived by plotting a profile of A perpendicular to a large vessel which runs parallel to the imaging plane. An assumption is made that arteries are (on average) circular in cross-section, allowing a curve to be fitted to this profile with two parameters: the vessel diameter and the calibration factor, S_0.

Methods: This quantification approach was applied to four data sets, acquired as described in [1]: transverse and coronal views from a healthy volunteer and a patient with vertebral artery stenosis, acquired under protocols agreed with local ethics and institutional committees. The coil sensitivities were removed from the raw data prior to processing using a maximum *a posteriori* solution to the Bayesian framework of [4] with two vessels per class, which can adapt to motion between scans and boost SNR. The derived model (Eq. 1) was fitted to each vascular component in each voxel within a mask derived by thresholding the mean signal intensity over time to exclude voxels that were noise dominated. The five parameter fit (A, δ_t, s, p, σ, where σ is the noise standard deviation) was performed using a constrained Bayesian maximum *a posteriori* procedure to regularize the fitting procedure to sensible parameter values and provide an estimate of the error on each parameter which could be propagated through to the quantification stage. Once parameter maps were derived, the calibration factor, S_0, was calculated from an average of ten profiles through different vessel segments. A calibrated simulated signal, $S'(t)$, was then calculated assuming a very short labeling duration, τ' = 1 ms, corrected for T_1 and RF effects, such that the signal is equal to the estimated volume of labeled blood within the voxel at the specified time. This process was repeated in the absence of dispersion ($s \to \infty$, $p = 0, \tau' = 50$ ms) for cases in which the ROI is small, where dispersion effects are likely to invalidate the assumption that the entire labeled bolus lies within the ROI at some time. The total labeled blood volume within manually drawn ROIs was calculated at each simulated time point and divided by τ' to give an estimated volume flow rate, F. If the quantification is perfect, we would expect F to increase as the labeled bolus washes in, plateau while the bolus remains within the ROI, then fall off as the bolus leaves the ROI or exchanges into tissue. The mean and standard deviation within the plateau region were calculated, yielding an estimate of F and its error.

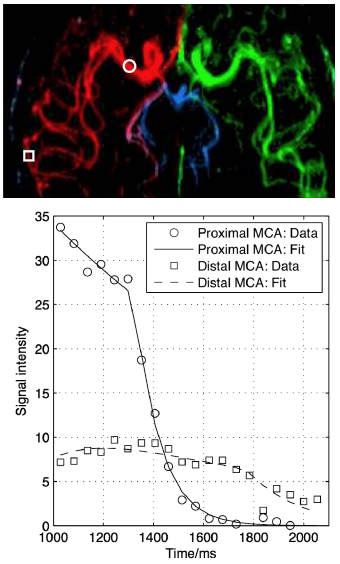

Figure 1: Theoretical model fit to the data in the proximal (circles) and distal (squares) right MCA, overlaid on a map of parameter A (top), color coded to show all components simultaneously (RICA = red, LICA = green, RVA = blue, LVA = purple). Time is relative to the start of labeling.

Results: The theoretical model (Eq. 1) gave a good fit to the data (Fig. 1). The calculated calibration factors, S_0, were consistent both between transverse and coronal data sets and between subjects. The expected plateau region of estimated volume flow rate vs time was seen in many, but not all cases (Fig. 2). Quantification with dispersion was generally more robust except where the ROI was small and could not encompass the dispersed bolus. Final F values (Fig. 3) are in good agreement with previous studies [5] and collateral flow into the PCA territories is clear in this patient. There is some contamination of VA signal into the MCA and ACA territories in the coronal data. To within error estimates, measurements of F agree between transverse and coronal views and F for each feeding (input) artery is equal to the total contribution to each of the outgoing vessels.

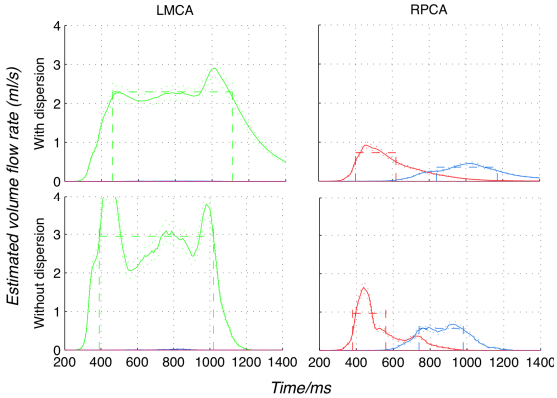

Figure 2: Example plots of estimated volume flow rate vs time in a patient using the transverse view for both MCAs and PCAs with (upper) and without (lower) dispersion. Colors represent the origin of the blood, as per Fig. 1. Dashed lines show the estimated plateau region and mean volume flow rate within it.

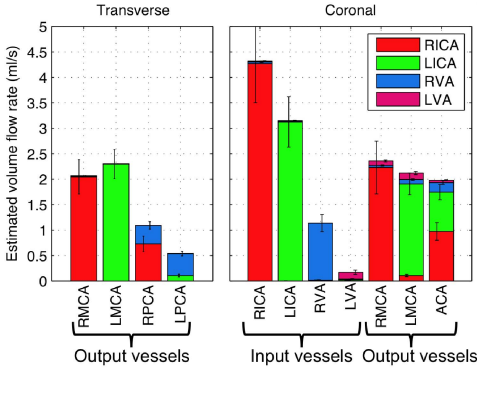

Figure 3: Final estimates of the volume flow rate in each vessel segment from a patient in transverse (left) and coronal (right) views. All results from simulated signals with dispersion except for the input vessels where the ROIs do not encompass the dispersed simulated bolus.

Discussion: The proposed technique allows the quantification of vessel-encoded dynamic angiography without the need for additional calibration scans, and yields volume flow rate measurements in good agreement with previous studies. Errors can occur where the ROI is smaller than the simulated bolus or where blood leaves the ROI or vessel mask before the full bolus washes in. We hope to improve upon these limitations in future work and validate the method in phantoms.

References:

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