Limit of Detection of Localized Absolute Changes in CBF Using Arterial Spin Labeling (ASL) MRI

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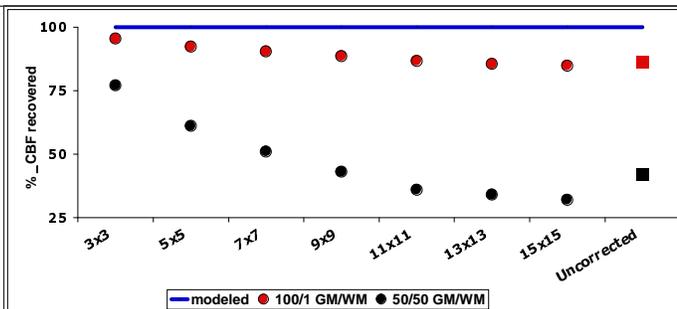[1]PICS, Radiology, Columbia University, New York, NY, United States, [2]Columbia University

Introduction: One of the main advantages of arterial spin labeling (ASL) MRI is its ability to provide absolute quantification of cerebral blood flow (CBF)[1]. This feature is highly relevant in assessing absolute changes in CBF due to activation or disease. As is the case with most imaging methods, the accuracy of the ASL measurement is impeded by the partial volume effects (PVE), which are a consequence of the limited spatial resolution in human brain imaging. Low spatial resolution results in increased voxel heterogeneity. Therefore, the acquired signal, which reflects a volume-averaged contribution from all the tissues with a voxel, cannot distinguish gray matter from the white matter or CSF. This problem is exacerbated in ASL because the measured signal is a ratio of two images, each of which is affected by PVE. Recently, a method that corrects for PVE in ASL imaging has been developed[2]. The method is based on linear regression algorithm, which estimates the pure tissue signals by modeling the intensity from a given voxel as a weighted sum of the contributions from individual tissues with the weighting coefficients being the tissues' fractional volumes[2]. The main advantage of this method is that it yields pure, tissue-specific CBF maps that are independent of voxels' tissue heterogeneity[2]. On of its drawbacks is the inherent spatial smoothing; the larger the regression kernel, the larger the smoothing effect[2]. Hence, while a larger regression kernel might be more suitable for baseline measurements, it can underestimate changes in CBF due to activation or disease. The goal of this ongoing study is to investigate the limit of detection of absolute CBF changes using ASL. Here we show results from simulated data analyzed using both the PVE-correction method (with varying regression kernel sizes) and the conventional one that does not correct for PVE. Validity of the results is shown on an experimental stroke model from a patient with left thalamic infarct.

Methods - Simulations: A theoretical ASL CBF image was constructed based on simulated EPI control and label images. Simulation of the EPIs was based on SPM5 templates for gray matter (GM), white matter (WM) and CSF tissue probability maps; tissue magnetization ratios were assumed $m_{CSF}:m_{GM}:m_{WM}=1.68:1.23:1.00$[2]. GM and WM CBF were assumed 107 and 30 (mL/100g·min), respectively[2]. Matlab code was written that allowed for a 2D step function in CBF to be applied with varying width (in the xy plane) and location across the brain. Results shown here correspond to a step function of width = 5x5x1 voxels (corresponding to an activated region = 15x15x9 mm^3.) **The patient** was a 54 yo, right-handed male, diagnosed with sub-acute left-thalamic infarct. Written consent was obtained as approved by the institutional IRB. **Imaging** was done on a 1.5T scanner (Philips) using a standard transmit-receive coil as previously described[3]. Briefly, single shot spin-echo EPI CASL images were acquired with: TR/TE/=4s/36ms, $\theta=90^o$; FOV=220×198 mm^2; acq.matrix=64 × 58; 15 slices, thickness/gap = 8mm/1mm; post-label delay=1000ms; labeling time=2.0s. A high resolution, 3D T_1 (SPGR) image was acquired: TE/TR=3 ms/34 ms, $\theta=45^\circ$; 100 slices; slice thick.=1.5mm/1mm gap; FOV=240×240mm^2; acq. matrix=256 × 256. **PVE-correction**: Pure tissue-specific CBF maps were obtained for the simulated and patient data as previously described[2]. To assess the effect of regression kernel size, analysis was run with varying kernel sizes in the xy-plane; the z-dimension=1 (corresponding 9mm). ASL CBF images were also obtained from the conventional, PVE-uncorrected data.
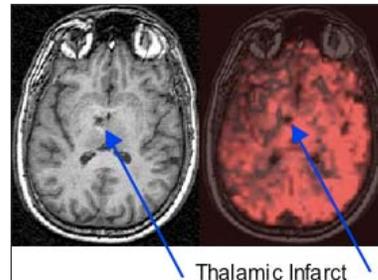
Results: Fig.1 shows results from simulated data. A stepwise increased activated region, ΔCBF=50%, was centered on a 100% GM voxel. The larger the kernel size, the lower the amount of activation recovered (see also Fig.2). Furthermore, for larger kernel sizes, the observed activated regions were spread onto the unactivated voxels due to the smoothing effect. Results were compared with those from the conventional, PVE-uncorrected method (squares). The PVE-uncorrected method yielded results similar to the corrected method for larger regression kernels.
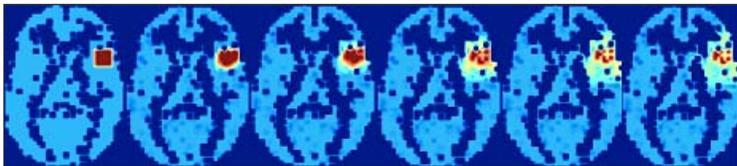
Fig.1 Simulated pure GM CBF images. The one on the far left shows the simulated increased activation of ΔCBF =50% of baseline centered on a 100% GM voxel. The following images (left to right) show results from the PVE-corrected method with kernel sizes of 3x3, 5x5, 9x9, 11x11, 13x13, respectively. Note the dilution effect of the larger kernels.

Fig.3: Images of: SPGR (left) and PVE-corrected CBF (right). Note good detection of the ischemic region. The observed left-right asymmetry is consistent with PET data (not shown).

Fig.2: % ΔCBF activation recovered via convetional, PVE-uncorrected method (squares) and PVE-corrected with varying regression kernel sizes (circles). A stepwise ΔCBF=50%of baseline was centered at a 100% GM voxel (red) and a voxel comprised of 50% GM and 50% WM

Experimental results from data acquired on a stroke patient are shown in Fig.3. In this case, the control EPI image was PVE-corrected using an 11x11x1 regression kernel while the ASL (control-label) difference image was PVE-corrected using a 3x3x1 regression kernel. The CBF of the lesioned area was 18% lower when measured using the PVE-corrected method as compared to the conventional, uncorrected, method, consistent with the expectation from the simulation data (not shown.)

Discussion: We have shown the feasibility of the PVE-corrected ASL MRI to detect local changes in CBF. The method is optimal when the 'activated' region is approximately the same size as the regression kernel used for PVE-correction. Although a direct comparison with conventional method is not straightforward, in regions with high voxel heterogeneity, the PVE-corrected method could recover ~50% more activation than the conventional, uncorrected method.

References: [1]Buxton et al., MRM 40(3) 1998; [2]Asllani et al., MRM in press; [3]Asllani et al., JCBFM, 28(4) 2008.