A SIMPLE MODIFICATION FOR REDUCING SCANNING TIME AND MOTION ARTEFACTS IN CLINICAL IMPLEMENTATIONS OF 3D-PCASL

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Target audience: This work will be of interest to clinicians and basic scientists measuring blood flow in the brain using ASL perfusion MRI.
Purpose: Arterial spin-labelled (ASL) perfusion MRI is used to measure blood flow quantitatively and non-invasively. As outlined in a recent review[1] it is employed in an increasing number of investigations including cerebrovascular disease, tumour vascularization and pharmacological imaging. Pseudo-continuously labelled ASL with a 3D fast-spin echo stack-of-spirals readout[2] and background suppression[3,4] (3D-PCASL) is one of the options currently available commercially. 3D readout approaches have recently been suggested as the preferred option for clinical applications of ASL[5]. We show that it is possible to acquire 3D-PCASL data with the same effective in-plane resolution and very similar signal-to-noise ratio (SNR) in a considerably shorter time by judicious choice of the number of spiral interleaves and the number of data points per interleaf. This modification may offer substantially improved temporal resolution in perfusion studies; and facilitate the use of ASL in some patient populations who can only tolerate very short scans.
Methods: Two sets of ASL perfusion MRI data were acquired on a 3T GE Discovery MR750 system (General Electric, Waukesha, WI, USA) from 3 healthy volunteers (2 male, 1 female) using 3D-PCASL. For both data sets the field-of-view was 24 cm, the slice-thickness (Δz) was 4mm, the number of slices (N_s) was 36, the number of averaged labelled-control pairs (NEX) was 3, the bandwidth (BW) was ±62.5kHz, the label duration was 1.45 s and the post-label delay was 2.025 s. Acquisition one used the conventional acquisition parameters on GE systems (as reported in the initial publication[4] and in a number of subsequent studies[6-8]); 8 spiral interleaves (N_{int}) with 512 data points per interleaf (N_p). These parameters results in an effective in-plane voxel size of 3.64 mm in a total acquisition time of 281 s. The parameters for acquisition two were chosen by exhaustively searching over the allowed ranges $3 < N_{int} < 16$ and $512 < N_p < 1024$ to determine if the same effective in-plane voxel size could be achieved in a quicker scan time. The following combinations and were found: $N_{int} = 7$ and $N_p = 580$ (248 s), $N_{int} = 6$ and $N_p = 672$ (216 s), $N_{int} = 5$ and $N_p = 800$ (183s) and $N_{int} = 4$ and $N_p = 992$ (151s). We therefore used $N_{int} = 4$ and $N_p = 992$ for acquisition two as it had the greatest time saving. The echo time (and the echo spacing (Δt)) was 10.5 ms for acquisition one and 14.4 ms for acquisition two. The repetition time (TR) was 4.84 s for acquisition one and 5.02 s for acquisition two. The readout window for acquisition one was 4096 μs and 7936 μs for acquisition two. The impact on the SNR was quantified as the ratio of the SNR in acquisition two to the SNR in acquisition one. The SNR for each acquisition is proportional to $\Delta x \Delta y \Delta z \sqrt{N_s N_{int} N_p NEX}/\sqrt{BW}$. Because the voxel size, number of slices, number of averages and bandwidth were unchanged the predicted relative SNR is given by $SNR_2/SNR_1 = \sqrt{N_{int,2} N_{p,2}}/\sqrt{N_{int,1} N_{p,1}}$. The effect of the increased echo-spacing (Δt) on the through-plane blurring (caused by T2 decay during signal readout) was quantified analytically using the full-width half-maximum (FWHM) of the point spread function (PSF) i.e. $FWHM = 2.\Delta t. N. \Delta z/\pi T_2$ for T_2=200 ms (this formula can be derived from the analytic Fourier transform of the signal in a centrically ordered k_z acquisition).
Results: The resulting calculated CBF maps for both acquisitions for all three subjects are shown in Fig 1. The FWHM of the through-plane PSF was calculated to be 4.83 mm for acquisition one and 6.59 mm for acquisition two (36% increase). The predicted SNR ratio SNR_2/SNR_1 was 0.984 (this was confirmed using measurements in a phantom).

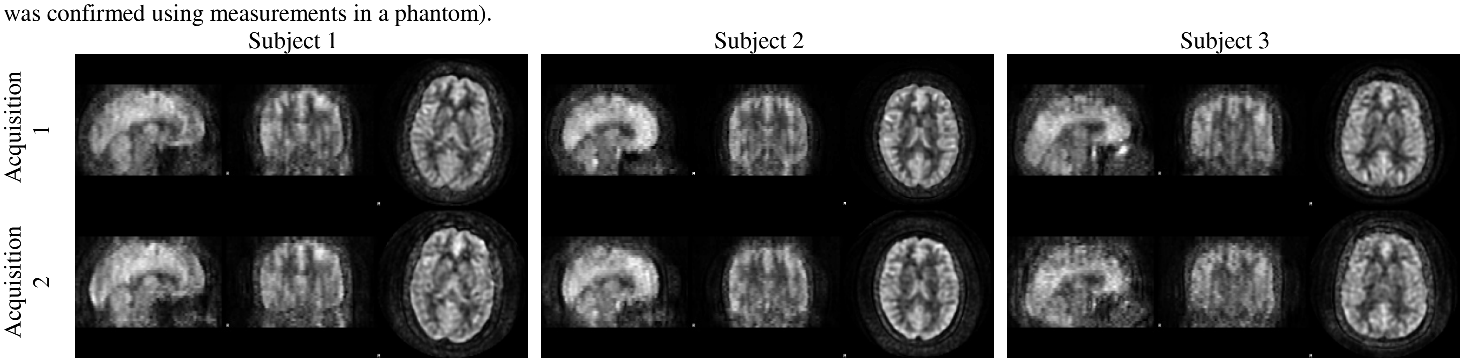

Fig. 1 Sagittal, coronal and axial projections of the calculated CBF Maps (0-100 ml/100g/min) from three subjects for both data acquisition schemes

Discussion and Conclusions: We have shown that CBF maps derived with 3D-PCASL can be produced with the same effective in-plane resolution and SNR as previously in 54% of the acquisition time (2 minutes 31 s vs. 4 minutes 41 s) with no appreciable loss of image quality. This was achieved by simply reducing the number of spiral interleaves and increasing the number of data points per interleaf. This change comes at the cost of a theoretical 36% increase in the z-blurring PSF. However, this z-blurring was not visually obvious in the sagittal and coronal projections of the calculated CBF maps. An additional drawback could be increased in-plane blurring in acquisition two due to the greater impact of resonance offsets over the longer readout window. Again this was not visually obvious in the calculated CBF maps. If only smaller time savings are needed (i.e. using $N_{int} = 7$ and $N_p = 580$) then smaller increases in z-blurring and in-plane blurring would result. As well as increasing patient throughput and providing higher temporal resolution in pharmacological studies that utilise perfusion MRI, we believe this modified acquisition will improve data quality as the shorter scan may reduce the impact of subject motion. We are currently evaluating the impact of our suggested protocol in neuro-pediatric patients at 1.5T.

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