Parallel Transmit Vessel Selective Arterial Spin Labelling: Phantom and In-Vivo Results

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Introduction

Vessel selective arterial spin labelling (ASL) provides visualisation and quantification of the perfusion territory from a feeding set of arteries. Methods to selectively label blood include multidimensional RF pulses [1], oblique labelling slabs [2], transverse gradients and phase cycling [3], precessionary gradients [4] and utilising a spatially localised labelling coil [5]. Separate labelling coils yield high SNR and no magnetisation transfer effects in the imaging volume, and are typically used to label either the right or left carotid artery. However, the spatial profile of the surface coil extends across the neck, causing partial labelling of the contralateral arteries. It has been shown in computer simulations that parallel transmission [6] can be used to improve vessel specificity in vessel selective ASL [7, 8]. In this work, parallel transmit vessel selective CASL is performed in-vivo with a two coil ASL labelling array with the aim of improving vessel specificity compared to using individual coils.

Method

Imaging was performed on a 3T Siemens Tim Trio (Erlangen, Germany) whole-body scanner, interfaced to a two-channel in-house built transmitter system based around a re-cycled spectrometer (MR5000, Surrey Medical Imaging Systems, Guildford, UK), equipped with 1W nominal RF power amplifiers (ZHL-3A, Mini-Circuits Inc. Brooklyn, NY). Each channel drove an element of an ASL labelling transmit array, comprising of two PIN diode detonable, 45mm ID circular surface coils tuned and matched to 123.2MHz. A whole-body birdcage coil was used to transmit the imaging RF pulses, and a 32 channel receive-only head coil for reception. B_1 maps were acquired using an AFI sequence [9]. To improve the SNR of the B_1 maps in regions of low excitation, the coils were driven in linear combinations, then inversely transformed to produce individual coil maps [10]. A 3D time of flight (TOF) angiogram was then acquired to locate the arteries within the neck. In-house software with a graphical user interface was used to select label and no-label regions and then calculate appropriate amplitudes and phases for the ASL labelling coil array by optimising the total B_1 field such that the field at the no-label region was minimised, whilst retaining the B_1 magnitude within the label region [7]. Due to coupling between coil elements, amplitudes and phases were fine-tuned manually to ensure the B_1 null coincided exactly with the 'no-label' artery. A continuous ASL (CASL) sequence with interleaved background suppression [11] and single shot 3D GRASE readout [12] was used to acquire perfusion weighted MR volumes.

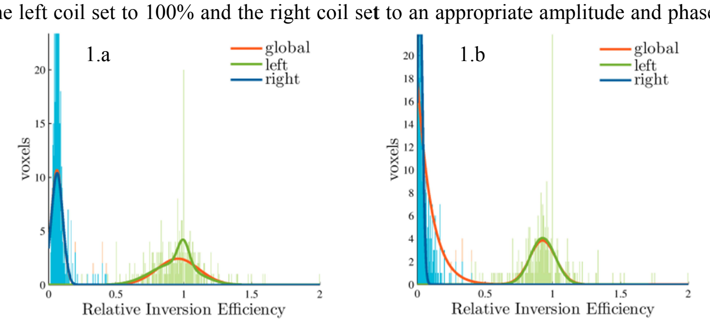
Masks were created from the non-vessel selective ΔM images, and segmented manually into left and right ROIs. Relative inversion efficiency maps were calculated by dividing vessel selective ΔM images by non-vessel selective ΔM images, from which histograms were produced of all voxels within each mask. Dual Gaussian functions were fit to each histogram. Images have been enlarged by a factor of 3 using a lanczos filter. Image processing was performed in Matlab (The Mathworks Inc., Natick, MA).

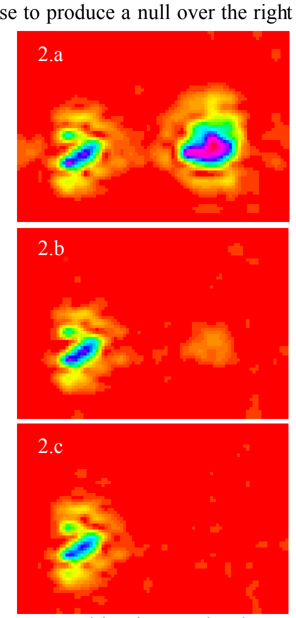
Phantom

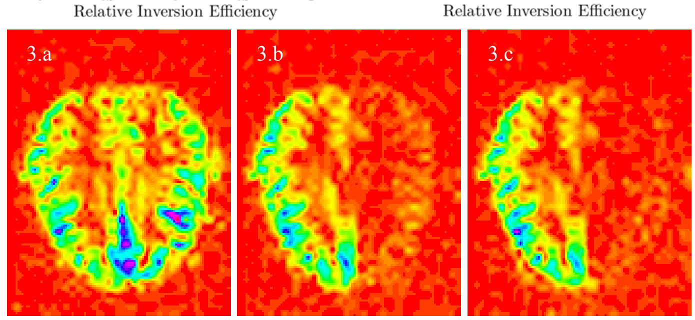
A flow phantom with separate left and right perfusion territories, each fed by inflow pipes separated by 6cm was used as a controlled environment for testing this method. Tap water doped with a small amount of $CuSO_4$ was circulated around the phantom with a peristaltic pump. ASL labelling coils were positioned either side of the phantom, in line with the inflow pipes. CASL labelling parameters were 3.5s labelling duration, 0.5s post labelling delay, 5mT/m labelling gradient. 3D GRASE acquisition parameters were FOV=135×240×60mm, TE=14.9ms, TR=4350ms, 64×36×12 image matrix, 30 averages, 6/8 partial fourier and 2×GRAPPA accelerated. ASL volumes were acquired for: (a) labelling with both coils at 100% amplitude and zero phase difference, (b) selective labelling with the right or left coil at 100%, (c) selective labelling with the labelling coil at 100% and the opposite coil set to an appropriate amplitude and phase to produce a null over the contralateral inflow pipe.

In-Vivo

In-vivo ASL volumes were acquired of a healthy 30 year old female volunteer with CASL parameters 2s labelling duration, 1s post labelling delay and 3.5mT/m labelling gradient. 3D GRASE parameters were similar to the phantom, except with FOV=162×288×60mm. Labelling coils were placed either side of the neck. Data was acquired for (a) labelling with both coils at 100% amplitude and zero phase difference, (b) selective labelling with the left side coil at 100%, (c) selective labelling of the left side carotid and vertebral arteries with the left coil set to 100% and the right coil set to an appropriate amplitude and phase to produce a null over the right carotid artery.







Results

Relative inversion efficiency histograms of the phantom measurements are shown for the left coil transmitting, and parallel transmit configuration in 1.a and 1.b, respectively. The parallel transmit configuration shows a decrease in the centre of mass of the right ROI distribution in comparison to using the left coil independently. Figure 2 shows difference images for the central slice of the phantom measurements for 2.a both coils transmitting , 2.b left coil only transmitting, 2.c parallel transmit configuration. Figure 3 shows difference images for the central slice of the in-vivo measurements for 2.a both coils transmitting, 2.b left coil only transmitting, 2.c parallel transmit configuration. Both the phantom and in-vivo results show a reduction in contaminated labelling on the right side/hemisphere, yet still retain labelling efficiency on the left side/hemisphere.

Discussion and Conclusion

In this work parallel transmission methods have been applied to vessel selective arterial spin labelling, with the aim of improving vessel specificity and reducing contralateral labelling in comparison to using a single surface coil placed on the side of the neck. Both phantom and in-vivo results show a noticeable reduction in contralateral labelling with minimal perturbation to the labelling efficiency of the left perfusion territory. Typical amplitudes for the contralateral coil were 10-20%; power deposition for this technique is not much more than using a single labelling coil on one side of the neck. Using additional coils would increase flexibility in the size and depth of the B_1 null, further attenuating contralateral labelling or allowing more vessels to be targeted. Further work will aim to reduce the coupling between the two coils so that no manual correction is required of the amplitudes and phases.

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