BOLD signal phase and magnitude dependence on vessel geometry

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Introduction

Functional MRI signal from gradient-echo EPI is dominated by intravascular BOLD changes in large cortical and pial veins.[1] This increases the effective point spread function of the fMRI technique, since these larger veins can extend from several to tens of millimeters from the site of neuronal activity. While many clever techniques, such as spin-echo EPI, spiral, diffusion-weighted EPI, and perfusion-weighted methodologies, have been proposed to suppress the macrovascular BOLD effect, the use of gradient echo EPI has been virtually ubiquitous in BOLD applications over the past decade because of its speed, multislice capability, ease of acquisition, and robust signal. It has been proposed that the phase response of large vessels can be used to distinguished macrovascular and microvascular effects since larger vessels can be considered essentially linear and oriented while capillaries are generally randomly oriented.[2] Because the large vessels are generally orientated compared to the random orientation of microvessels, they can contribute significantly to both the phase and magnitude compared to microvessels that only effect the magnitude.

Methods

To validate that phase changes are associated with only the macrovasculature, a three-dimensional deterministic simulation of MRI signal has been used. The deterministic diffusion model, first proposed by Bandettini and Wong,[3] was extended to three dimensional analysis. The MRI simulation uses the iterative multiplication and convolution of three-dimensional arrays according to $\mathbf{M}_k = (\mathbf{M}_{k-1} \times \mathbf{R}) * \mathbf{G}$ where $\mathbf{M}_k$ is the magnetization array at the kth time point, $\mathbf{R}$ is a relaxation and precession array accounting for T_2 decay and differences in precession frequency, and $\mathbf{G}$ is a Gaussian diffusion kernel derived from the Einstein equation and the Gaussian distribution in three dimensions assuming isotropic Gaussian diffusion. The net voxel signal is the sum of all magnetizations. $T_{2,tissue}$ was 66 ms and $T_{2,blood}$ was computed from literature values based on oxygenation.[4] All simulations were conducted with static magnetic field of 4T, gradient echo time of 15 ms, uniform blood oxygenation ranging from 60 to 76%, hematocrit of 0.4, 1.5 μm^2/ms diffusion rate, and 256 point isotropic three-dimensional computation grid. The vessel radius r was modulated with oxygenation Y using the nominal radius r_n according to $r = r_n (Y - 0.6)^{1.00465}$.

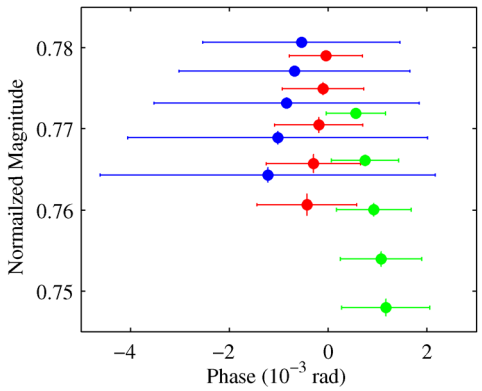

Figure 1 Phase versus magnitude plot for capillary networks in voxels with 350 (blue), 750 (red), and 1500 (green) µm resolution as Y increased from 0.6 to 0.76.

Three isotropic voxels of size 375 µm, 750 µm, and 1500 µm were used in all simulations. Five random distributions of 3 µm radius vessels were distributed uniformly in position and orientation to achieve approximately 2% blood volume. The vessels covered a volume twice the voxel size to ensure uniformity and the computational grids for the voxels were 500 µm, 1000 µm, and 2000 µm to minimize edge effects during convolution. Fixed orientation vessels were placed in the voxel at angles of 0 to 90 degrees. Four combinations of size and spacing were use for simulations: (1) 15 µm radius vessels separated by 250 µm, (2) 50 µm vessels separated by 1000 µm, (3) 125 µm radius vessels separated by 1500 µm, and (4) 500 µm radius vessels separated by 2000 µm. Simulations were done for each fixed vessel orientation and for the random distributions separately and combined.

Results

Phase variations were $1.1 \pm 0.1 \times 10^{-3}$ radians, $0.4 \pm 0.2 \times 10^{-3}$ radians, and $0.6 \pm 0.3 \times 10^{-3}$ radians for 375 µm, 750 µm, and 1500 µm resolution respectively (Fig 1). Since an image with signal to noise ratio of 100 has a phase error of 1×10^{-2} radians, this is below the detectable limit and will not effect fMRI time course phase. For combined vessel orientations, the phase changes depended on the vessel orientation and vessel size (Fig 2). The larger radius fixed vessels produce greater correlated phase and magnitude changes that dominate the changes within the smaller voxels because of the larger volume fraction of the large vessels.

Discussion

Phase changes in fMRI occur in large vessels and are usually associated with a large magnitude change. Although phase changes only occur with large vessels, large vessel BOLD changes do not always produce phase changes. This is particular true of vessels near the magic angle since the frequency offset effects that create the phase dependence are zero at that angle. The phase and magnitude changes are near linear, particularly for very large vessels, which would result in a linear relationship between magnitude and phase. Therefore some large vessel effects can be isolated and removed using previous described techniques.[2] These techniques would be most applicable to high-resolution fMRI studies in which the phase changes are most significant.

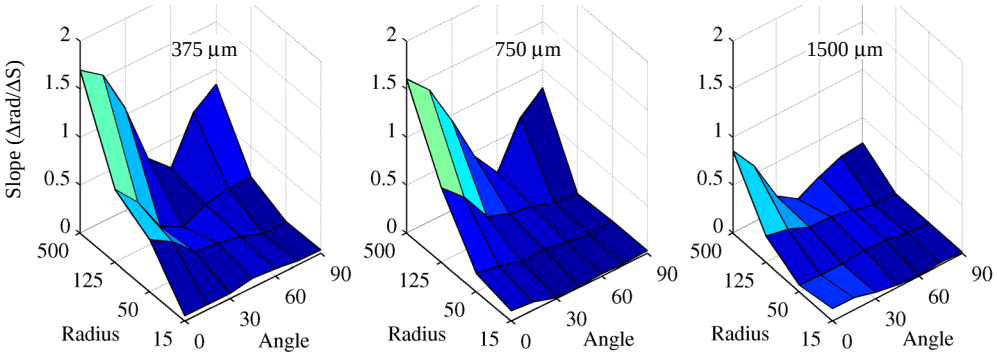

Figure 2 Slope of phase versus magnitude with changing Y for voxels of 375, 750, and 1500 µm size with fixed vessel of 15, 50, 125, and 500 µm radius at angles from 0 to 90 relative to the main magnetic field.

References

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3. Bandettini PA, Wong EC. Int J Imag Syst Tech 1995;6:133-152.
4. Silvennoinen MJ et al. Magn Reson Med 2003;49:47-60.