Modeling of Phase Changes in BOLD fMR

Z. Feng[1,2], A. Caprihan[1], K. Blagoev[3], F. Zhao[4], and V. D. Calhoun[1,2]

[1]The MIND Institute, Albuquerque, NM, United States, [2]Dept. of ECE, University of New Mexico, Albuquerque, NM, United States, [3]Los Alamos National Laboratory, Los Alamos, NM, [4]Emory/Georgia Tech University, Atlanta, Georgia

INTRODUCTION: The complex-valued tissue signal for BOLD fMRI signal change described in (1) indicates that both the magnitude and phase of fMRI signal contain physiologic information. However, so far, virtually all fMRI studies have analyzed only the magnitude images. Phase changes have been repeatedly observed and reported (2-4). It has been suggested that these phase changes are from the frequency shift of the IV signal (4) or the tissue water's frequency shift caused by local brain temperature change (5). There are different approaches to attempt to utilize the phase data (6-9). All of these approaches improved detectability at a fixed false-alarm rate than the magnitude-only approach for the signals at low signal-to-noise ratios (SNRs). However, those approaches do not address the underlying causes of the phase change and the mechanisms; the 'Lorentz sphere' concept was extended to calculate the magnetic field distribution in the heterogeneous tissue (10). However, no simulation based on the theory of the Lorentz sphere has yet been performed using human BOLD fMRI data. In this paper, we first present the theory of the Lorentz sphere and express the local magnetic field in the material. Next, we report on the simulations for the 3-D Gaussian magnitude change, which provides some insight into the phase model presented in the theory section. Then we discuss the application of the phase model to BOLD fMRI data.

THEORY: In each voxel, according to a two-component model (11,12), the volume-averaged magnetic susceptibility χ can be calculated from the volume-weighted average of magnetic susceptibilities of the EV tissue χ_t and the IV blood χ_b: $\chi = f\chi_b + (1-f)\chi_t$ where f is the relative blood volume fraction. Subsequently, the volume averaged magnetization for a voxel at position **r** can be written as: $M(r) = \chi(r) \cdot B_0 / \mu_0$. Let us determine a volume of activation (VOA) where the blood susceptibility and/or the blood volume would change during the brain activation. Then, a Lorentz sphere can be drawn around a position of interest **r** within the VOA, as shown in Fig. 1 (10). Note that the size of this Lorentz sphere is different from that introduced above in the classical solid state physics textbook. It ranges from sub-millimeter to millimeter scale, comparable with imaging-voxel size. Assuming that $\vec{M}$ is continuous in the macroscopic scale, then the magnetic field change can be subsequently written as

$$\Delta\vec{B}(r) = \Delta\vec{B}_d(r) + \frac{\mu_0}{3}\Delta\vec{M}(r) + \sum \Delta\vec{B}_b(r),$$

where the demagnetization field is given by (13,14)

$$\vec{B}_d(r) = \frac{1}{4\pi}\iint_S \frac{(\chi(r') - \chi_{out}(r'))(\vec{B}_0 \cdot \hat{n})(\vec{r} - \vec{r}')}{|\vec{r} - \vec{r}'|^3} ds' - \frac{\mu_0}{4\pi}\iiint_V \frac{\nabla \cdot \vec{M}(r)(\vec{r} - \vec{r}')}{|\vec{r} - \vec{r}'|^3} dv'$$

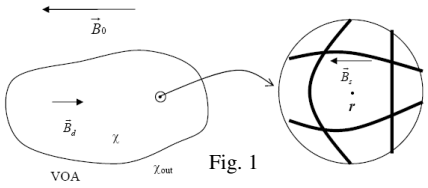

Fig. 1

If we assume that the detected fMRI magnitude change is linearly proportional to ΔM(r) (15) provided that TE is greater than the characteristic time t_c and diffusion effect is ignored, based on the magnitude results the spatial distributions of ΔM(r) and the VOA can be obtained. $\Delta\vec{B}_b(r)$ is the local magnetic field change resulting from the blood vessels. The contribution of $\Delta\vec{B}_b(r)$ to the fMRI phase change is zero in the case where the blood vessels are randomly distributed (12,15).

We performed simulations in order to understand the theory described above better. For example, if the activated region of the magnitude change of signal ΔM(r) is 3-D Gaussian with the standard deviations σ_x, σ_y and σ_z respectively. We can see that when the long axis of magnitude change ΔM(r) is parallel to the magnetic field, the resulting phase change is positive. Therefore, the volume-averaged magnetization effect dominates. When the long axis is perpendicular to the magnetic field, the resulting phase change is negative; thus, the demagnetization effect dominates. The long axis of ΔM(r) is some degree off the magnetic field. As expected, the results are combinations of positive and negative phase changes as shown in Figure 2, due to the volume-averaged magnetization and demagnetization effects.

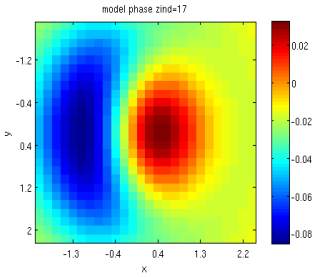

Fig. 2

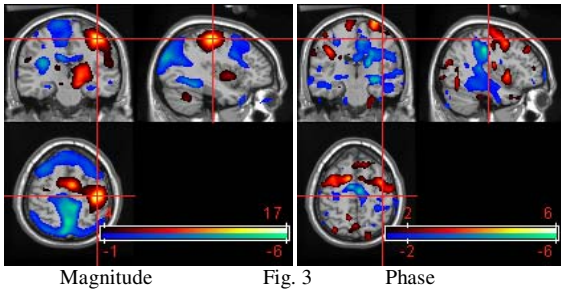
Magnitude

Fig. 3 Phase

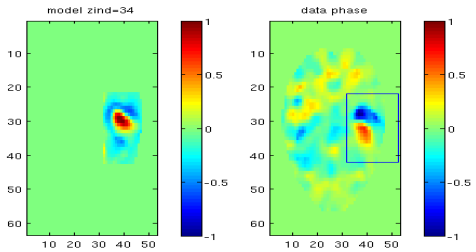

Fig. 4

RESULT: Observations of the magnitude and phase change data indicate the peak of the magnitude change is near the sign change of the phase change. This statement is also true for 16 different subjects. Fig. 3 shows the magnitude and phase change for 16 averaged subjects. The results of phase model and data at z = +48 mm for subject A are shown in Fig. 4. The modeled phase has a similar structure as the data in the box (task-related signal change area). They have similar patterns of positive and negative peaks, as well as zero crossing. The similarity exists for different slices and different subjects. In conclusion, based on the previously developed Lorentz sphere model, this paper simulates the phase change of fMRI BOLD data from humans for the first time. Our results are encouraging and suggest the phase data is providing useful information. Further work is needed in order to add this technique of phase change modeling to improve the analysis of BOLD fMRI in the future.

REFERENCE: (1). Hoogenraad, F.G., et al., MRM, 2001. 45: p. 233-246. (2) Lee, A.T., et al., MRM, 1995. 33: p. 745-754. (3) Hoogenraad, F.G., et al., MRM, 1998. 39: p. 97-107. (4) Menon, R.S., MRM, 2002. 47: p. 1-9. (5) Yablonskiy, D.A., et al., Proc Natl Acad Sci USA 2000. 97: p. 7603-7608. (6) Lai, S., ISMRM, 1997. p.1671. (7) Nan, F.Y., et al., IEEE Trans Med Imaging 1999.18: p. 320-329. (8) Calhoun, V.D., et al., MRM, 2002. 48: p. 180-192. (9) Rowe, D.B., et al., Neuroimage, 2004. 23: p. 1078-1092. (10) Zhao, F., et al., MRM, 2007. 57: p. 520-527. (11) Ogawa, S., et al., Proc Natl Acad Sci USA 1990. 87: p. 9868-9872. (12) Ogawa, S., et al., Biophys J, 1993. 64: p. 803-812. (13) Durrant, C., et al., Concepts Magn Reson 2003. 18A: p. 72-95. (14) Zhao, F., et al., ISMRM, 2006. 14: p. 2790. (15) Yablonskiy, D.A., et al., 1994. 32: p. 749-763.