Phase map based investigation of intravoxel signal dephasing in gradient echo MRI

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Introduction:

Macroscopic static magnetic field inhomogeneities accelerate signal loss in gradient echo images due to intravoxel signal dephasing [1]. This imposes limitations upon any application of low resolution gradient echo imaging, such as fMRI. Due to the complex geometry of the field inhomogeneities, signal and BOLD sensitivity depend critically on voxel geometry, echo time, slice orientation, or correction schemes. We present a method based on single-echo high resolution phase imaging that allows to simulate and investigate intravoxel signal dephasing and BOLD sensitivity in a systematic manner.

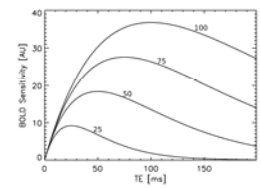

Fig 1.: Theoretical BOLD sensitivity curves for four different T2*-values (25, 50, 75, 100 ms).

Methods:

Background: Field inhomogeneities lead to a variation of the precession frequency (Eq. 1) and therefore to an accelerated T2' decay. Spins that do not precess on resonance with the rotating frame of reference accumulate a phase shift during the time between excitation and sampling (Eqs. 1 and 2). This phase shift manifests itself in the phase of gradient echo images, with a 2π ambiguitiy due to phase wraps. Based on an unwrapped high resolution phase image, the T2' decay can be predicted for a variety of data acquisition schemes. In some applications of gradient echo imaging, such as fMRI or the assessment of signal changes induced by hyperoxia or hypercapnia, the BOLD sensitivity σ_{BOLD} (Eq. 3, Fig 1.) [2] rather than the signal itself is of interest. *Data Acquisition:* Flow compensated 3D gradient echo images [3] were acquired on a 1.5 T system (Siemens Magnetom Vision): TR=42ms, TE=25ms, Flip angle=25°, Matrix=512×256×72, FOV=25.6×19.2×10.8 cm³. *Reconstruction*: Zero filling to 512 × 384 × 206, inverse FFT, and phase unwrapping using a quality map driven region growing approach [4]. *Simulation:* The temporal behavior of the complex signal $S(j_0, k_0, l_0)$ in a low resolution voxel at position (j_0, k_0, l_0) due to intravoxel dephasing was simulated based on the high resolution phase map ϕ (Eq. 4). The sum runs over all high resolution voxel positions (j, k, l) that contribute to the low resolution voxel. The magnitude in each high resolution voxel was set to unity ($\rho(j, k, l) = 1$). Different imaging parameters were simulated, such as: the in-plane resolution and slice thickness by grouping the high resolution voxels in different ways; the angle a between the slice and AC-PC-direction by rotating the phase data; z-shimming [5] by adding phase gradients G_z^{ϕ}; echo time TE by scaling the unwrapped phase. Signal characteristics and σ_{BOLD} were evaluated for all combinations of these parameters. Among the many parameters tested in the following only the interplay between z-shimming and α is discussed.

$$\omega(\vec{r}) = \omega_0 + \Delta\omega(\vec{r}) = -\gamma \cdot (B_0 + \Delta B(\vec{r})) \quad [1]$$

$$\phi(\vec{r}) = -\gamma \cdot \Delta B(\vec{r}) \cdot t \quad [2]$$

$$\sigma_{\mathrm{BOLD}} \approx T_E \cdot S(T_E) \quad [3]$$

$$S(j_0, k_0, l_0; t) = \sum_{j,k,l} \rho\,(j,k,l)\; e^{i\phi(j,k,l;t)} \quad [4]$$

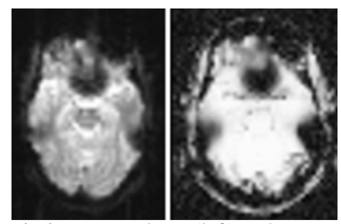

Fig 2: Measured EPI (left) and simulated image with the same TE and geometric parameters.

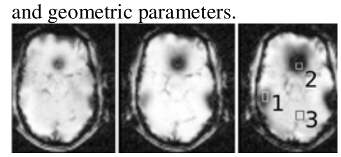

Fig 2: A z-shim of $G_z^{\phi} = 35\pi$ Rad cm⁻¹ (left) reduces dephasing in the region near the sinus compared to no z.-shim (middle) or even a negative z-shim of $G_z^{\phi} = -35\pi$ Rad cm⁻¹ (right).

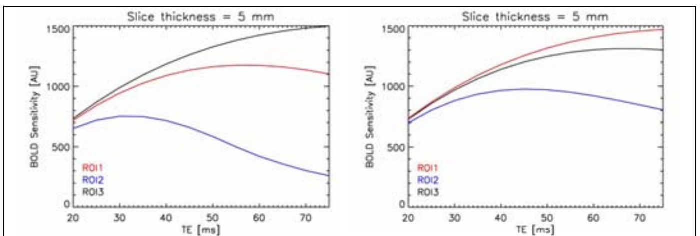

Fig 4.: σ_{BOLD} without (left) and with an optimized combination of z-shim and AC-PC angle. Left α=0, G_z^{ϕ} =0; Right: α=30, G_z^{ϕ}=35π Rad cm⁻¹.

Results: The same slice for different z-shimming gradients is shown in Fig. 3. Three regions were investigated. ROI 1 in a region with background inhomogeneities near the auditor canal, ROI 2 in the notoriously inhomogeneous region produced by the sinuses, and ROI 3 in a homogeneous region. A comparison of simulated data and an original EPI image with the same spatial resolution and echo time is displayed in Fig. 2. The simulated image and the EPI image image display very similar patterns of signal loss. The effects of z-shimming in combination with optimized a are shown in Figs. 3,4 and 5. In the critical ROI2 z-shimming improves σ_{BOLD} by 35% and shifts its maximum from TE = 32 ms to TE = 45 ms, which is a TE typically employed in fMRI. ROI 1 is also improved by about 30 %, whereas in the previously homogeneous ROI 3 σ_{BOLD} is reduced by about 20 %. These findings are also reflected in the σ_{BOLD} maps (Fig 5.), where the signal voids in ROI1 and ROI2 rare educed by proper scan parameters.

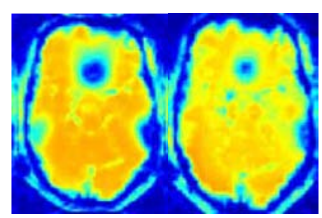

Fig 5: Simulated σ_{BOLD} maps before (left) and after optimizing α and G_z^{ϕ}.Left: α=0, G_z^{ϕ}=0, Right: α=30°, G_z^{ϕ}=35° Rad cm⁻¹ (blue = low σ_{BOLD}; orange = high σ_{BOLD}).

Discussion: The advantage of this approach over the acquisition of a series of different EPI data acquisition schemes is that it uses a scan that is not affected by image distortion or signal cancellation. One high resolution single echo scan is sufficient for the simulation and offline testing and optimization of acquisition parameters, including TE. The phase is acquired with short TE compared to EPI and with high resolution: 512 simulation voxels fit into a simulated EPI voxel of 4 × 4 × 4 mm³. Therefore, no signal loss occurs in regions where EPI already suffers from large signal voids. Acquisition schemes used in fMRI studies of difficult brain regions, such as amygdalae or the orbitofrontal cortex, may be tailored on an indiviudal basis prior to the fMRI session, or validated retrospectively. The method is scalable to other field strengths. *Limitations:* A global T2 of 100 ms of grey matter (where the BOLD effect occurs) was assumed. Partial volume effects between grey matter and white matter or grey matter and CSF due to a different magnitude of the signal are therefore not represented with high accuracy. However, in the inhomogeneous ROIs the rapid T2' decay dominates the signal behavior. Different T1-times, on the other hand, are not relevant for the simulation of EPI, where TR is in the range of 2 to 5 seconds.

References: **[1]** Reichenbach, J. R., et al. J Magn Reson Imaging, Mar-Apr 1997. 7(2):266-279. **[2]** Deichmann, R., et al. NeuroImage, 2002. 15(1):120-35 **[3]** Reichenbach, J. R. et al. NMR Biomed , Nov-Dec 2001. 14(7-8):453-46 **[4]** Witoszynskyj, S., et al. In Proc Intl Soc Magn Reson Med 13 . 2005 page 2249.7.**[5]** Merboldt, K. D., et al. NeuroImage, 2001. 14(2):253-7.