Dynamic Compensation of B0 Field Inhomogeneities Restores Complex fMRI Time Series Activation Power

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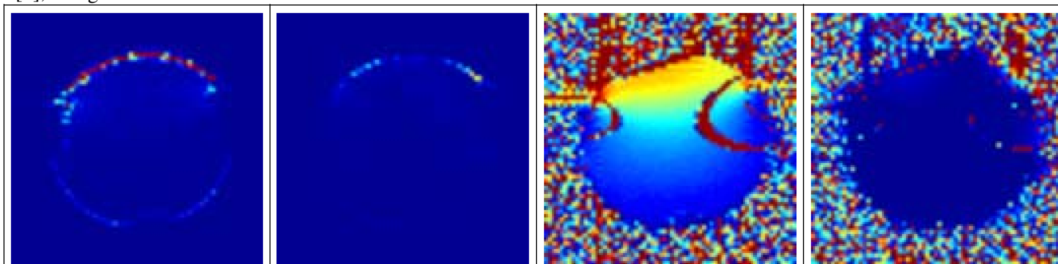
Introduction: In MRI, Echo Planar Imaging (EPI) techniques are vulnerable to dynamic fluctuations in the homogeneities of the main magnetic field (B0) that can result from subject respiration [1] or motion inside or outside the field of view. When EPI is used in fMRI, "noise" is induced as a direct effect of these dynamic phenomena and manifests itself in both the magnitude and phase time series. Magnitude effects are seen because the signal from a certain voxel location in the object being scanned is improperly encoded and thus will be shifted, or warped, to a different voxel location depending on the value of B_0 at that location during acquisition, while phase effects occur because the direct relationship between phase and B_0 offset from ideal perfect resonance [2]. As that offset changes, so does the measured phase, and this has a large impact on activations computed from complex-valued data (as opposed to magnitude-only data) which can provide increased statistical power [3] as well as eliminate BOLD activations in areas of draining veins [4]. In order to compensate for these errors caused by time varying B_0 field homogeneity, it is possible to utilize the knowledge that for each repetition in a time series, the spins are given an initial phase Φ_0 at time t=0 that is independent of the B_0 field homogeneities and is determined solely by the characteristics of the RF pulse. The phase of the i^{th} image, Φ_i, is the phase of the spins at the echo time (TE) and the B_0 field offset, $\Delta\omega_i$ (in rad/sec), is given by $\Delta\omega_i = (\Phi_i - \Phi_0)/TE$. The value of Φ_0 is difficult to measure with much certainty, but assuming Φ_0 does not change to a large degree between acquisitions, it is possible to solve for the *difference* between the B_0 field offsets present between the i^{th} and the j^{th} images, denoted by $\delta\omega_{ij}$, as $\delta\omega_{ij} = \Delta\omega_i - \Delta\omega_j = (\Phi_i - \Phi_j)/TE$. By setting j=1, the difference in the B0 field between the 1st image and every other image in the series can be calculated. After correcting each image for this difference, effects of changes in B_0 homogeneities through time are reduced resulting in a more robust time series.

Methods: A spherical phantom filled with SiO_2 oil was imaged on a 3T MRI scanner (General Electric, Milwaukee, WI) while the main magnetic field was manually perturbed by moving a water filled object along the inferior-superior axis at .2 Hz outside the imaging field of view during the acquisition. A series of 276 images were acquired using gradient echo EPI (matrix = 64x64, FOV = 24cm, slice thickness = 3.8 mm, TE = 25 ms, TR = 1 s, flip angle = 45 degrees). A human subject was also scanned using the same sequence and scan parameters as described for the phantom experiment. During the series acquisition, the subject was cued to breath heavily at a .167 Hz rate as well as to perform a block design bilateral finger tapping task (20s rest followed by 16 epochs of 8s active, 8s rest). Activation statistics (Z statistics) were calculated using a general linear model with compensation for a constant and linear trend. The first 10 data points were discarded and the block design reference function was delayed 4 seconds to account for the hemodynamic delay. Raw maps of $\delta\omega_i$ were processed in a manner similar to [2]. A 2D polynomial (4th order) is fit to the raw map and then masked and allowed to fall smoothly to zero. The processed field maps can be used to correct the images using any preferred method, with the "conjugate phase" method described in [5], being chosen in this case.

Results: Inspection of the power spectrum of both the magnitude and phase voxel time series from the phantom experiment reveals a large peak at .2 Hz (the rate of field perturbation). The images in Figure 1 display the value of that peak in the power spectrum corresponding to each voxel. As expected, the effect in the magnitude is large in areas of high spatial contrast, most noticeably along edges, while the effect in the phase is much more global and smoothly varying. After calculating the maps of $\delta\omega_i$ as discussed above and performing the correction, the effect is greatly diminished in both the magnitude and phase time series. For the human data, Figures 2(a) and (c) show magnitude-only activation maps before and after correction, respectively. They appear nearly identical, which is likely due to the fact that at 3T with the level of B_0 modulation induced in this case by the heavy breathing, there is little effect on the magnitude, especially in areas that do not have high spatial contrast. The benefit of the correction in this case is shown in Figures 2(b) and (d), which displays thresholded [6] activation statistics, calculated using the complex constant phase method [3] before and after correction, respectively. The "noise" in the phase due to the B_0 modulation greatly increases the residual variance estimates (with little change to the coefficient estimates) thereby reducing the statistical power of the complex constant phase model. Correction reduces the phase noise thereby restoring the increased detection power of the complex time series activation analysis.



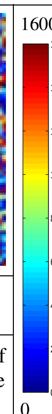


Fig. 1a: Uncorrected Magnitude Power Spectrum, .2 Hz	**Fig. 1b:** Corrected Magnitude Power Spectrum, .2 Hz	**Fig. 1c:** Uncorrected Phase Power Spectrum, .2 Hz	**Fig. 1d:** Corrected Phase Power Spectrum, .2 Hz

Figure 1: Images shown are of the value of the power spectrum at the .2 Hz frequency (corresponds to the frequency of B_0 field perturbations) for each voxel time series. As expected the contribution of the .2 Hz frequency to the voxel time series is clearly reduced after applying the correction.

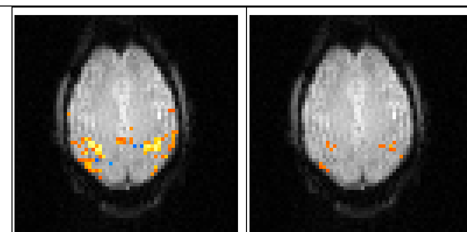


Fig 2a: CP uncorrected. **Fig 2b**: CP uncorrected.

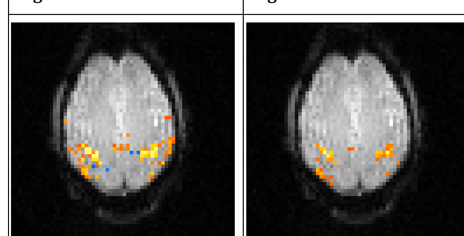


Fig 2c: MO corrected. **Fig 2d**: CP corrected.

Figure 2: Z statistic maps at a .05 Bonferroni threshold.

Discussion: A major advantage of the method described here is that it can be carried out using the phase information inherent to standard gradient echo EPI images, and can thus be applied without any special pulse sequence. The method is very robust and will be reliable given that the assumption of a constant initial phase (Φ_0) holds true. In general it should, but hardware instability and motion within the transmit coil large enough to affect the phase homogeneity of the RF pulse can cause that assumption to fail. It is worth emphasizing that the correction method described here does not provide an absolute correction for the B_0 field inhomogeneities, but rather a correction for changes in B_0 through time. This method in conjunction with a static field map collected before the time series acquisition will provide all the necessary information to apply an absolute field correction to every image in the entire time series.

References: **1.** PF Van de Moortele et al.: Magn Reson Med 47:888-895, 2002. **2.** P Jezzard, RS Balaban: Magn Reson Med 34:65-73, 1995. **3.** DB Rowe, BR Logan: NeuroImage 23:1078:1092, 2004. **4.** AS Nencka, DB Rowe: NeuroImage 37:177-188, 2007 **5.** DC Noll et al.: IEEE Trans Med Imaging 10:629-637, 1991 **6.** BR Logan, DB Rowe: NeuroImage 22:95-108, 2004.

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