Noise reduction in fMRI GRE EPI acquisitions using real-time navigators

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Introduction: Gradient-Echo Echo-Planar Imaging (GRE EPI) is the method of choice for functional magnetic resonance imaging (fMRI) and many approaches have been proposed for temporally stabilising acquired signals and reduce variations caused by physiological noise and/or hardware imperfections ([1-5]). Since field inhomogeneities of higher orders, which can result from such variations, are difficult to compensate for retrospectively, a prospective method might be of advantage. We will here show the first results of a prospective correction method using an approach presented by Splitthoff et al ([5]), which acquires self-referencing projection scans, so-called navigators, before every EPI slice readout for determining in real-time (<8ms) in-plane inhomogeneities of up to first order. Unlike other methods, no registration or comparison to a reference scan is required. The measured inhomogeneities are then corrected for during the echo-planar read-out.

Methods: In phantom measurements the mentioned method ([5]) was shown to detect purposely set inhomogeneities with an extremely high accuracy; in in-vivo experiments a high correlation with the separately measured breathing signal was demonstrated. This high accuracy could only be achieved by using an array receiver coil, with the best results achieved with a twelve-element coil. Although methods for reconstructing complex image data from coil arrays have been suggested ([6]), unfortunately no gold standard exists up to now and even the reconstruction environment shipped with scanners is not producing reliable results. Thus, it was not possible for us to demonstrate the effects of the method on the phase of the data. We therefore have chosen to investigate a much more difficult but important aspect: the reduction of noise in magnitude fMRI images. One healthy volunteer was scanned on a 3T Magnetom Trio (Siemens Medical Solutions, Erlangen, Germany) with the following parameters: 128x128 matrix, 20 slices, $2^3 mm^3$ resolution, TR=2s, TE=36ms with a 6/8 partial Fourier reconstruction. The fMRI paradigm was chosen to be a visual stimulation with a block size of 40s, consisting of 20s rest and 20s stimulation; 15 blocks were presented, leading to a measurement time of 600s. Two runs were performed, the first one with navigators and correction enabled, the second one with navigators but without correction. In order to include possible hardware imperfections into the analysis, care was taken, that the scanner had not been used for one hour before the first measurement; between the two runs again a pause of one hour was inserted. SPM5 was used for realigning the data and computing a design matrix; the volumes were restricted to an area common to both data sets. The design matrix was then interpolated and fitted, using Matlab (The Matworks Inc.), to the voxel time courses with the correct temporal offset.

Results & discussion: Following the standard fMRI analysis the square roots of the power spectra of the residuals were calculated and averaged amongst the 500 highest values of the regressor, to give the results shown in Figure 1. Comparing the two averaged spectra, it can be seen that especially the very low frequencies are much more prominent in the uncorrected data set; this indicates slow modulations in the signal. One explanation for this can be found when looking at the image consistency expressed in difference images in Figure 2. To account for initial fluctuations, we chose to subtract the 295^{th} data set from the 15^{th} (both have the same activation status so that activation differences should be minimum); shown is the result for a central slice. As can be seen in Figure 2.a, the image remains stable in the corrected data set. On the other hand, in the uncorrected one a small hardware imperfection seems to have led to an inhomogeneity in phase encoding direction and thus a small compression of the FOV. This is very likely to be one of the causes for the slow changes that can be seen in the spectrum. Another possibility is the change in susceptibility artifacts caused by subject motion, as indicated by the images in Figure 3. Here difference images of axial EPI scans of a phantom are presented. The phantom was rotated along the sagital axis by 5°; bulk motion was corrected using the PACE ([7]) method. As can be seen, the motion related susceptibility artifacts are drastically reduced when prospective field inhomogeneity correction is active. Motion-to-distortion interaction might therefore indeed be the cause of the increased lower frequencies in the uncorrected data set in Figure 1. As for the higher frequencies: the breathing frequency for the mentioned volunteer measurements lay for both runs at about 0.1Hz; unfortunately, no significant changes can be observed in that range, which might lead to the question, whether the echo time penalty of about 8ms for the method is justified. The inability of the presented technique to compensate for the breathing-related signal variations is puzzling, bearing in mind that the navigators seem to be able to detect these variations reliably. Since the data presented here are of very preliminary character, we hope to obtain further results by measuring more volunteers with different fMRI paradigma as well as at higher field strengths. Nonetheless, Figures 2 and 3 indicate the type and scale of the magnetic field homogeneity changes that do occur during normal fMRI measurements, rendering the prospective correction necessary.

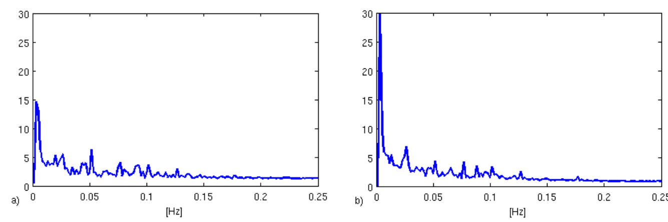

Figure 1: The averaged spectra for a) the corrected data set and b) the uncorrected data set

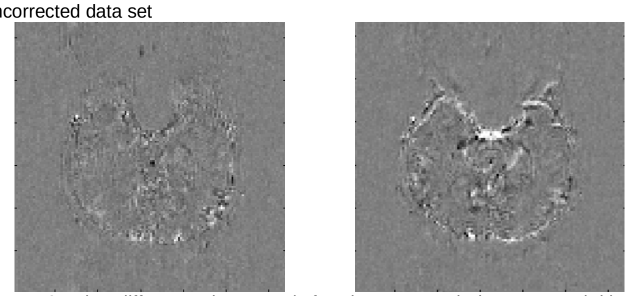

Figure 2: The difference images a) for the corrected data set and b) the uncorrected data set

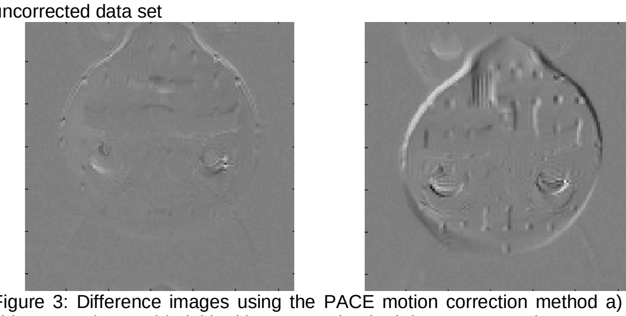

Figure 3: Difference images using the PACE motion correction method a) with shim correction enabled, b) without; note that both images are at the same scale.

References: [1] Pfeuffer et al., MRM 47, 344-353, 2002;[2] Ward et al., MRM 48, 771-780, 2002;[3] van der Kouwe et al., MRM 56, 1019-1032, 2006;[4] van Gelderen et al., MRM 57, 3363-368, 2007;[5] Splitthoff et al., ISMRM, Berlin, 2007; [6] Walsh et al., MRM 43, 682-690, 2000; [7] Thesen et al., MRM 44, 457-465, 2000.

Acknowledgements: This work was supported in equal parts by the German Federal Ministry of Education and Research, grant #01EQ0605, and the German Research Foundation, grant #422/2-1.