Combined between slice motion and susceptibility distortion correction for fMRI with extreme motion

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Target audience: Interested in EPI distortion correction, particularly applied to fMRI data with large subject motion.

PURPOSE Echo planar imaging (EPI), a technique widely used in fMRI due to short acquisition times, suffers from spatial distortion caused by magnetic field inhomogeneity due to magnetic susceptibility differences. Several methods have been developed to correct for this type of geometrical distortion [1,2]. Recently there has been a great interest in fMRI studies of fetuses and young children where large head movement may occur during acquisition. However existing distortion correction methods have not been tested in the presence of such large and rapid head motion. In this study we applied the correction approach in [2] requiring two echoes only for the initial step (and hence more time efficient than [1]). Due to subject movement between two consecutive slices or frames, the original method was adapted to work in a slice by slice manner and the temporal phase unwrapping step was skipped. Moreover, volume to volume and slice to slice movement corrections were performed for both the susceptibility distortion corrected and uncorrected datasets. **METHODS** Resting state fMRI data were acquired on one healthy volunteer on a 3T Philips Achieva scanner using a dual echo EPI sequence with TE1=15.6ms, TE2=45ms, TR=3000ms, FA=79˚, spatial resolution of 3.0x3.0x3.0mm^3 and FOV=80x80x40. During three acquisitions, consisting of a total of 100 time frames each, the subject stayed still (A1), performed small (A2) and large (A3) nodding head movements in the middle third of the acquisition. Phase unwrapping was performed according to [1], with inter-slice and inter-frame offset correction based on the mean values of unwrapped phase across the slice and frame. The ϕ_0 map representing the phase component independent on echo time and spatial magnetic field inhomogeneity was calculated according to the classical phase correction approach [1] using the phase data acquired for the first time frame with two different TEs. In the following time points, the magnetic field map was estimated using the information from only one echo, ϕ_{TE_n}, according to the equation: $\Delta B_0(x,y,z,t_n) = (\phi_{TE_n}(x,y,z,t_n) - \phi_0(x,y,z))/\gamma TE_n$. Appropriate voxel shifts to correct for the distortion were calculated based on the final field map. Due to signal instability occurring in the first frames of an fMRI acquisition, field calculation and correction were only performed from 6th time frame onwards.

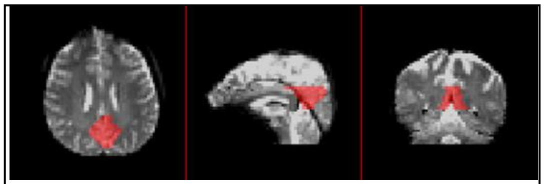

Fig 1. Default mode network ROI used for in time consistency calculation, overlaid on the original image.

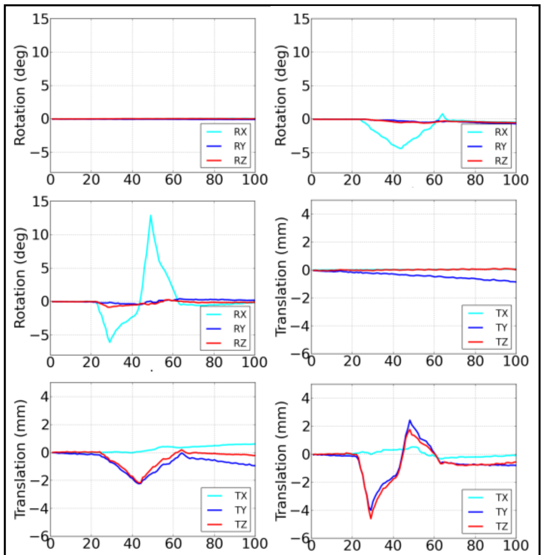

Fig 2. Subject motion parameters measured post-hoc: rotation (left) and translation (right), for acquisitions A1 (stationary, top), A2 (small motion, middle) and A3 (large motion, bottom).

Subsequently volume registration and cascaded slice to volume registration [3] were applied to both distortion corrected and uncorrected data. Consistency of the correction for a typical fMRI study was examined using a the region of interest (ROI) corresponding to the expected location of default mode network (DEF) in original images (Fig 1) and evaluating the root mean square difference (RMSD) in intensity values in this region. We then assume that we need to remove any large scale intensity differences arising from motion to permit accurate detection of subtle bold effects in this region. To ensure the consistency of location of this region in the differently deformed and corrected images, we locally re-aligned this ROI to each time series being compared. **RESULTS** Subject head motion measured post-hoc from the slice motion estimates ranged from -5˚ to 1˚ rotation around the *x*-axis, -2-1mm translation along *y*- and *z*-direction for A2 and -5˚-15˚ rotation, -5 to 3mm translation for A3 (Fig 1) [3]. In the stationary acquisition (A1) translation ~1mm in *y*-direction was observed. RMSD values obtained for three acquisitions, plotted for both corrected and uncorrected data, with and w/o additional volume and slice correction are presented in Fig 3. **DISCUSSION** RMSD values corresponding to in time consistency calculated for DEF ROI were improved by the susceptibility distortion correction in all acquisitions, for original, volume and slice corrected data (solid vs dotted lines, Fig 3). Volume and slice correction further lowered RMSD values, which is especially visible in case of the acquisitions containing motion, A2 and A3, between frames 20th and 65th (Fig 3, compare with Fig 2). The results including only volume correction and both volume and slice correction were comparable for this ROI. The trend for RMSD to increase in time may be arising from by a phase drift present in the EPI data series. **CONCLUSION** We showed that an existing susceptibility distortion correction method modified to work on a slice by slice manner can improve fMRI images suffering from small and large motion. Volume and slice correction additionally improved the temporal consistency of the data, especially in case of significant subject motion. These results are particularly promising for the experiments with moving subjects, and may have important applications to the study of small children or foetuses. Further research will also incorporate intensity correction and investigate spin history effects present in the images.

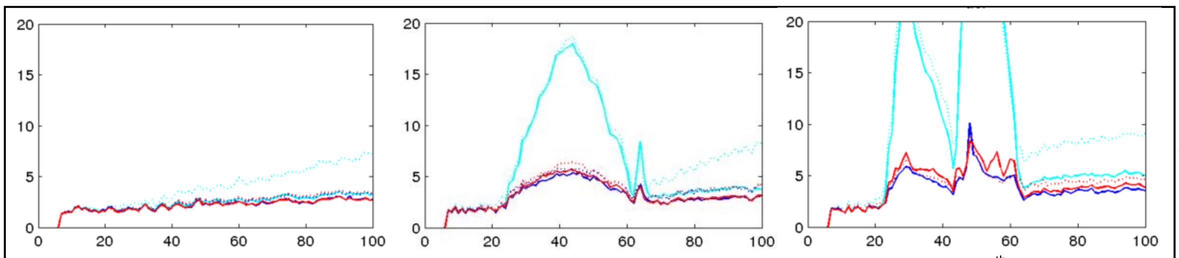

Fig 3. Root-mean-square difference (RMSD) calculated within default mode ROI for all time frames vs 6th frame plotted for A1 (stationary, left), A2 (small motion, middle) and A3 (large motion, right); original, volume corrected and additionally slice corrected images; with (solid line) and w/o (dotted line) susceptibility distortion correction.

References [1] Hutton at el., *Neuroimage.* 2002, 16:217-40; [2] Lamberton at el., JMRI. 2007, 26(3):747-55 [3] Seshamani et al., ISBI, 2013.