

Analysis of ghosting artifacts for real-time motion correction applications using EPI

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Introduction: Patient motion is a well-known issue in MRI. Echo-Planar Imaging (EPI) is a fast acquisition technique that can reduce the effect of patient motion. However, motion that occurs between EPI volumes requires either retrospective volume-to-volume realignment or prospective techniques to correct for this motion. Recent work has shown the advantages of the real-time prospective approach, whereby a camera is used to track motion and the logical gradients are updated in real-time to correct for head rotation [1, 2], as shown schematically in Fig. 1. The problem with prospective motion correction in the presence of head rotation is that in EPI, gradient hardware delays between odd-even echoes can result in different ghosting parameters for each rotation of the logical gradients, which would require an estimation of the ghost parameters for each different head rotation. This is problematic as it is scan time intensive. Here we test whether the ghost correction parameters estimated from the first EPI scan can be applied to oblique angles that are limited to the expected extent of patient head motion, in order to test whether this extra reference scan is necessary for real-time prospective motion correction applications.

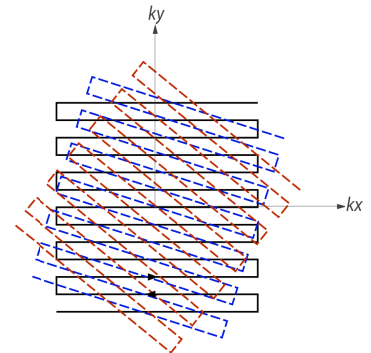


Fig 1 -To simulate the effect of rotating EPI gradients on the image quality for real-time motion correction applications the logical gradients are rotated to the expected extent of

Methods: Data were acquired on a homogeneous spherical agar phantom and on a healthy human volunteer using a

3T whole-body MRI unit (GE Discovery MR750), an 8-channel head coil, and a high-performance gradient system (50 mT/m, SLR=200 mT/m/s). To simulate scan plane updates for prospective motion correction, EPI data were acquired at 9 different angles (by rotation of the logical gradients) ranging from 0° - 20°, with 2.5° increments in the axial, coronal, and sagittal planes. The following imaging parameters were used: single-shot EPI, acquisition matrix = 64 x 64, FOV = 26 cm, slthck/gap = 5 mm/2mm, TR/TE = 4 s/17 ms, partial Fourier with 24 overscans. Additionally, the scanner gradient delays were estimated using balanced reference estimator delays for each physical gradient (X, Y, and Z) [3] and were found to be 1.076μs, 0μs, and 0.964μs, respectively. Three different post-processing methods performed to investigate the effects of ghosting. 1) “First angle” ghost-correction calibration estimation was performed on the first EPI angle (0°) using an image entropy-based approach [4]. The same constant and linear phase ghost calibration parameters from the first angle were then applied to each of the successive EPI angles. 2) “Per angle” ghost correction was performed by separately estimating, and applying, the entropy-based ghost calibration parameters for each angle. 3) The ideal “Phantom” ghost calibration parameters from the 0° EPI phantom data were applied to the *in vivo* data. A quantitative metric for the amount of ghosting is calculated as follows: the 0° angle was used as a reference image for both the phantom and *in vivo* data. The absolute difference between the reference image and each subsequent image was obtained by first rotating the ghosted image back to the orientation of the reference image using bi-cubic interpolation. The metric was then calculated as the total energy (square of the Frobenius norm) of each difference image.

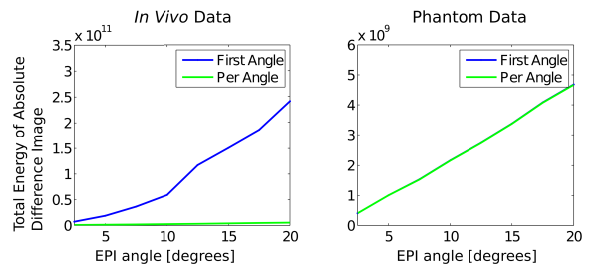


Fig 2 - Numerical comparison of ghosting correction of (left) phantom data and (right) *in vivo* data. Each trace represents one of the correction parameters used.

Results: Fig. 2 plots the ghosting metric (total energy) vs. EPI-blade angle for the *in vivo* (left, taken from the sagittal scan where there was no ghosting) and phantom data (right). Fig. 3 shows the ghosting artifacts in the *in vivo* images after post-processing using the three methods. In Fig. 4, the ghosting artifacts are seen to increase at

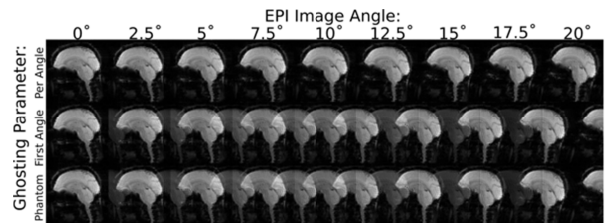


Fig 3 - Ghosting artifacts observed in *in vivo* data after correction using various parameters

increasing EPI angles.

Discussion: For “first angle” and “per angle” methods in both *in vivo* and phantom data, total energy increases as the angle of rotation increases, with the rate of increase being most severe for the *in vivo* “first angle” method. This suggests that ghost correction on a first angle basis is insufficient as rotation angles increase. In the phantom data, “per angle” and “first angle” methods overlap (Fig. 2, right), suggesting that the data is more robust to ghost correction. There is no advantage to using method 3 on the *in vivo* data for correction (Fig. 3) as there is no observable reduction in ghosting. In addition, the physical delay of the sagittal (Fig. 4, row 3) scan (logical Y gradient) is 0μs, resulting in the least amount of ghosting artifact of all cases, and explains why “compensation blips” (which causes gaps in k-space) were not necessary in this case. For axial and coronal cases (Fig. 4, rows 1, 2), however, the importance of “compensation blips” is emphasized - as even small delays on logical Z (~1μs) will cause aliasing artifacts due to gaps in k-space.

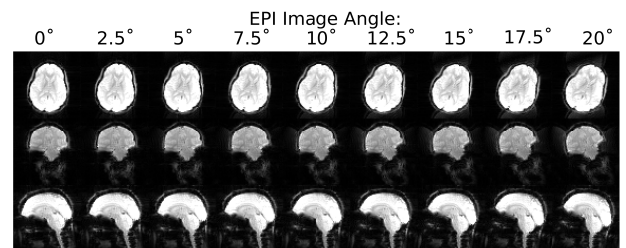


Fig 4 - Ghosting artifacts due to gradient delays. (Top) AXIAL: physical X gradient = 0.964μs and physical Z = 1.076μs (Middle) CORONAL: physical X gradient = 0.964μs and physical Y = 0μs (Bottom) SAGITTAL: physical Y gradient = 0μs and physical Z = 1.076μs

Conclusion: It is seen here that, in the event of even small amounts of rotation (< 5°), ghosting cannot be totally eliminated with the ghost correction parameters from a single EPI scan. As such, intermediate reference scans are needed for real-time imaging applications. Additionally, if any gradient delays are present in the system, post-processing methods of ghost correction, be it by a per scan or single scan basis, will not work in eliminating ghosting artifacts.

References: [1] Thesen, et al. MRM 2000. [2] Zaitsev, et al. Neuroimage 2006. [3] Reeder, et al. ISMRM 1996 [4] Nordell ISMRM 2007 **Acknowledgements:** This work was supported in part by the NIH (5R01EB002711, 5R01EB008706, 3R01EB008706, 5R01EB006526, 5R21EB006860, 2P41RR009784), the Center of Advanced MR Technology at Stanford (P41RR09784), Lucas Foundation, and Oak Foundation.