Correcting gradient-delay-induced phase errors for prospective motion correction in MRI

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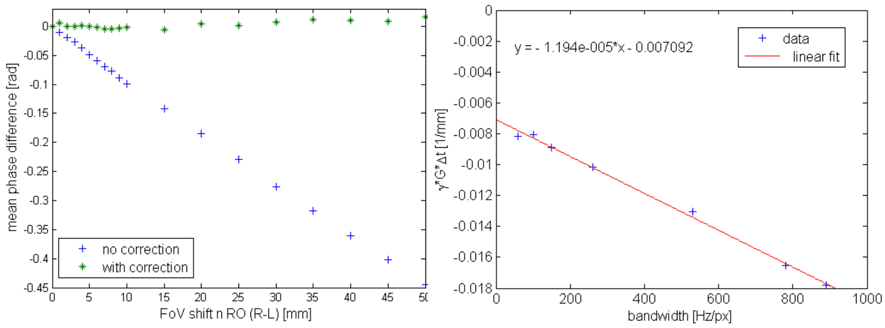
Introduction: Subject motion during an MR-scan often leads to severe image artifacts. Prospective motion correction (PMC) has been shown to vastly improve the quality of motion affected MR images [1]. To realize PMC, the imaging volume is adjusted during a measurement to maintain a constant rigid body relationship with the subject. The orientation of the imaging volume is controlled by updating the gradients, while the positioning of the volume is achieved by updating the frequencies and the phases of the ADCs and the RF-pulses. Due to various hardware imperfections, gradient delays, Δt will add a constant phase offset to the image, which under normal circumstances can be corrected for. However, under PMC, a coordinate update occurs nominally for every *k*-space line and this results in a phase offset ϕ_{off} in image space that varies throughout the acquisition according to $\phi_{off} = \gamma G \Delta t\, \Delta x + \phi_0$ [2]. In this work we investigate if ϕ_{off} must be known to correct the images accurately with PMC. Previously it has been shown that delays in the read-out (RO) direction can be corrected in the case of radial imaging [3]. For PMC, delays in the slice-encoding gradient, are also relevant. We present a procedure to determine ϕ_{off} for the RO and slice selection directions and demonstrate the feasibility of prospective phase error correction.

Methods: Measurements were performed on a MAGNETOM Tim Trio System at 3 T (Siemens Healthcare, Erlangen). A single-slice image of a cylindrical phantom was acquired using a PMC enabled GRE sequence (TE=5 ms, TR=100 ms, resolution=1 mm^2, FOV=256 mm^2, flip angle=25°) and a PMC enabled EPI sequence (TE=42 ms, TR=1000 ms, resolution=2 mm^2, FOV=256 mm^2, flip angle=90°).

Read-out: For each image, the FoV was manually shifted in the read-out direction for each measurement (from Δx=0 to Δx=50 mm) to obtain multiple images at different positions, simulating object movement. This was done for several bandwidths and for different RO-directions. In post-processing, each image was shifted back to the initial position, therefore simulating PMC. The mean phase difference in the image domain $\Delta\phi_{mean}$ between the image at Δx =0 mm and FoV-shifted image determined.



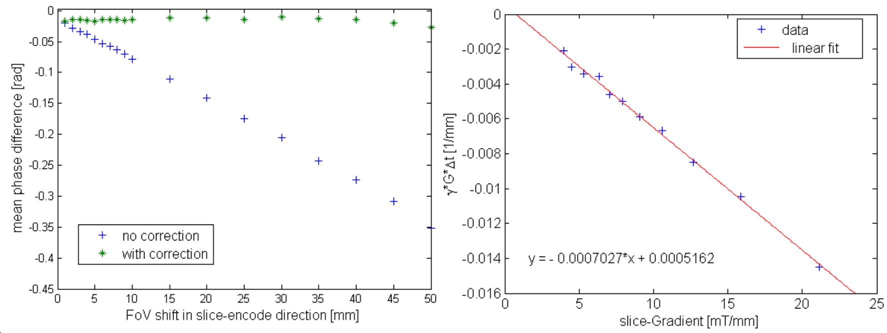


Fig. 2: a) mean phase difference vs. FoV shift (slice-encode-direction) b) slope of phase plots vs. BW

Slice-encode: In this case the measurement was altered to induce a phase differences while the actual slice position remains unchanged. To achieve this, the slice at each new position was measured first at the desired location and then following a 180° rotation around the RO direction with the slice offset therefore reflected. This simulated the effect of a negative slice-selective gradient. This was done for multiple gradient strengths G_{sl}. $\Delta\phi_{mean}$ for each image pair (the second image rotated to the initial orientation, again to simulate PMC) was calculated and divided by 2, to compensate the effective doubled change in distance.

Simulations: The effect of ϕ_{off} is difficult to visualize *in vivo*, because the volunteer would have to perform exactly the same motion twice. Therefore simulations were carried out using an artificial motion on a segmented EPI dataset (TE=22ms, TR=2000ms, resolution=1mm^2, FOV=256mm^2, flip angle=90°, EPI-factor=33). A head nodding motion with an amplitude of 10 mm was simulated. To reproduce the artifacts that would be seen from this motion each *k*-space line in the phase-encode direction of an MR image was altered with a constant phase offset according to the determined equations (see Fig. 1 b)).

Results: Figure 1 shows an example dataset for shifts of the FoV in the RO-direction (right-left in this case). Fig. 1a) displays how $\Delta\phi_{mean}$ varies linearly with the shift ('+' symbols). Fig. 1b) shows the slope of each phase plot versus the bandwidth. A linear fit of this graph yielded the equation (shown in Fig. 1b)) that was used for prospective correction, i.e. recalculating the phase of the ADC in real-time. After the correction was implemented into the sequence and applied, the data are distributed around zero ('*' symbols in Fig. a)). The results for RO-shifts along other axes were similar and are therefore not shown. Figure 2 shows the corresponding graphs for FoV shifts in the slice-encode direction. Again, $\Delta\phi_{mean}$ is linearly dependent on the shift (Fig. 2a)) and the slopes were plotted versus the gradient strength (Fig. 2b)). The equation shown in Fig. 2b) was also implemented into the sequence and resulted in prospectively adjusting the phase of the RF-pulse. The correction negates any differences in phase almost entirely as can be seen in Fig. 2a)). Figure 3 shows an MR image of a healthy volunteer without motion (a) and with the simulated artifacts due to ϕ_{off} (b). When scaled to the same contrast (red box) it can be seen that ϕ_{off} reduces contrast and increases ghosting artifacts.

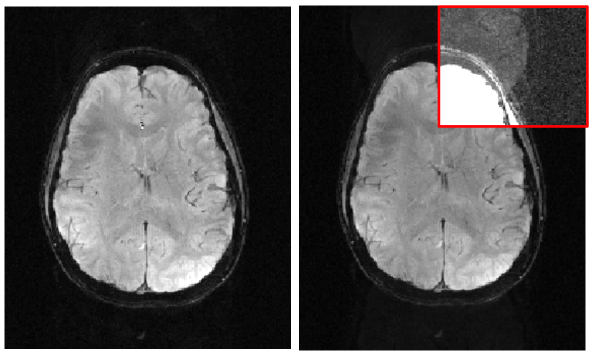


Fig. 3: a) artifact-free MR image b) MR image with phase

Discussion: The results shown in Fig. 1 and 2 imply that the phase offset ϕ_{off} depends on the bandwidth for shifts in RO, and on the gradient strength G_{sl} for shifts in the slice-encode direction. ϕ_{off} can be represented by the linear equations $\phi_{off} = A*BW + B$, and $\phi_{off} = A*G_{sl} + B$ (shown in Fig. 1 b) and 2 b)). Using these equations to adjust the phase of the ADC and the RF-pulse, it is possible to correct for existing gradient delays in real-time. This would lead to improved motion-corrected MR-images. For the several sequences tested the detected delays were between 3 and 20 µs corresponding to the typical range for hardware imperfections. The errors in the alignment of the gradients, the ADCs and the RF-pulses may not only originate from hardware imperfections as in our results, but also from inaccuracies in sequence implementation, *e.g.* *k*-space center shift of few ADC points. However such inaccuracies have no effect for normal imaging and hamper PMC insignificantly.

Conclusion: We have shown that a linear relationship exists between the phase and the translation in RO and slice encode direction, which can be calibrated and used to correct for the residual phase error in PMC. The effect of this error on MR images is, however, minor for regular motion. We also present an effective test procedure to detect sequence timing imperfections.

References: [1] M. Zaitsev et al. 2006 *Neuroimage*, 31; [2] P. Speier et al. 2005 *Prc. ISMRM*, 2295; [3] P. Speier et al. 2006 *Prc. ISMRM*, 2379