Background phase correction using k-space filters in phase contrast velocity encoded MRI

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Introduction

In phase contrast (PC) velocity encoded MRI, background phase offsets occur due to eddy currents and concomitant gradient terms. These can introduce substanial errors in the velocity measurements, and are more pronounced when imaging slow flow using large velocity encoding gradients. Pulse sequence modifications can be used to correct for these errors [1], but background phase correction is still frequently required as a post-processing step. The most common post-processing correction technique involves estimating the phase variation in stationary tissue and subtracting a fitted polynomial surface [2]. Previous work shows that a high order polynomial is desirable [3]. An alternative technique, used in susceptibility weighted imaging, is to assume that the phase inhomogeneities are low frequencies in k-space and can be filtered out using a high-pass k-space filter [4]. This work evaluates this k-space filtering technique for 2D PC velocity encoded MRI, and compares it to the surface subtraction technique.

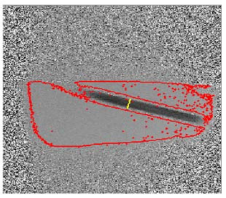

Figure 1: phantom with steady-state flow. Red lines show contours of stationary pixels and the yellow line defines the location of the flow profile calculations.

Methods

Phantom and *in-vivo* scans were performed on a 3T Allegra scanner (Siemens Medical Solutions) using a single channel head coil. A 3D PC velocity encoded MRI pulse sequence [5] was used for two directional in-plane velocity encoding. The phantom consisted of a 6 mm internal diameter silicon tube containing steady-state flow and embedded in stationary water (Figure 1). Two phantom datasets were acquired using two different mean flow velocities, 0.13 m/s and 0.25 m/s. Velocity sensitivities (Vencs) of 1 m/s and 0.5 m/s were used for the fast and slow flow rates, respectively. Other imaging parameters include TR = 6.8-7 ms; TE = 4.1-4.3 ms; flip angle = 7°; spatial resolution = 0.63×0.63 mm^2; matrix size = 256 × 256; slice thickness = 5 mm. The *in-vivo* scan was performed on a normal volunteer and consisted of a sagittal midline slice through the head showing flowing cerebrospinal fluid. Scan parameters were identical to the phantom scans except: venc = 0.05 m/s; TR = 7.2 ms; TE = 6.9 ms.

The true phase offset was estimated by repeating all of the above scans at the identical slice positions using a large phantom containing stationary water. These were then multiplied by the complex conjugate of the respective original images to remove this phase offset, thus providing reference images to be used as a gold standard.

For the surface subtraction, stationary water was identified on the phantom by thresholding in both magnitude and absolute phase (red outline in Figure 1). Polynomials of order 1-5 were fitted in a least-square sense with the magnitude data used as a weighting function.

For the high-pass k-space filtering, a Gaussian kernel was used with standard deviations of 8×8, 16×16, 32×32 and 64×64 pixels. The effect of the phase correction on the measured flow was investigated by plotting a 1D flow profile of the 2D in-plane flow magnitude $v = \sqrt{v_x^2 + v_y^2}$ (yellow line in Figure 1). The average error between the two background phase correction techniques and the reference was calculated for both the flow profile and the stationary background. The flow profile error was averaged for all flow profile pixels. The average background phase error was calculated as the absolute difference between the corrected phase image and the reference in a region of stationary pixels, and averaged between the two in-plane velocity encoded directions.

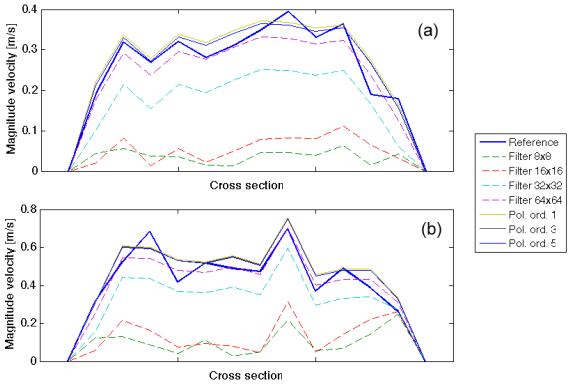

Figure 2: 1D flow profiles for (a) slow and (b) fast flow.

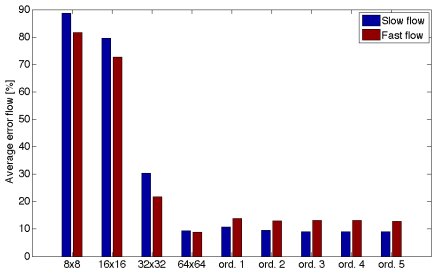

Figure 3: Average error in flow profile. Blue and red bars show slow and fast flow, respectively.

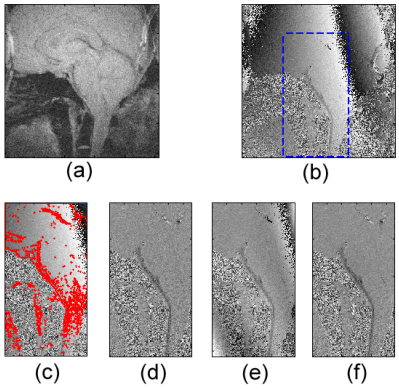

Figure 5: *In-vivo* background phase correction. (a) Magnitude image, (b) the original phase image, with the dashed blue line defining the area and (c) the red contour of stationary tissue for polynomial surface subtraction, (d) is ground truth, (e) 5th order polynomial subtraction, and (f) k-space filtering with a 64×64 pixel Gaussian kernel.

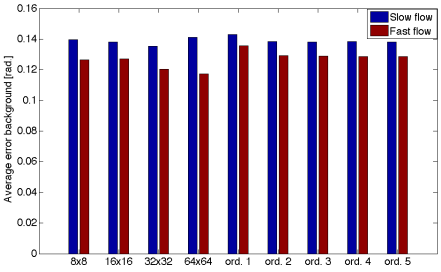

Figure 4: Average background error in radians. Blue and red bars show slow and fast flow, respectively.

Results and Conclusions

Figure 2 shows the effects of the various techniques on the flow profile. Figures 3 and 4 show the average background phase offset errors for the flowing and stationary water, respectively. These results demonstrate that k-space filtering and polynomial surface subtraction perform equally well for background phase correction. The flow profiles are corrupted by small Gaussian high-pass k-space filter sizes, but a filter size of 64×64 pixels performs equally well as a 5th order polynomial surface subtraction. The *In-vivo* data (Figure 5) also shows comparable qualitative results for the two phase correction techniques. K-space filtering is thus a suitable technique for background phase correction in PC velocity encoded MRI. In addition, k-space filtering is computationally fast, doesn't require masking of stationary pixels and can easily be extended to 3D. However, as with polynomial surface subtraction, k-space filtering will generate errors in the case of bulk motion.

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5. Markl *et al.* JMRI 2007;25:824–831.

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