Automated Phase Correction for Quantitative Phase-Contrast Flow Measurements

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Introduction: Phase-contrast (PC) imaging uses the phase-difference (PD) between two acquisitions to remove the common non-flow related sources of phase. One source that is not compensated for with this type of processing is eddy currents (EC) induced by the flow-encoding gradients themselves. Since these gradients differ between the two acquisitions, the eddy currents do not completely cancel, resulting in a non-zero spatially varying background phase. This background phase is in addition to the phase accumulated by moving spins and can cause an error in the measured flow velocity.

The background phase in the PD image can be estimated by performing a first-order (or higher) spatial polynomial fit to the image phase [1]. This fit can then be subtracted from the PD image to remove the background phase. This fitting does not distinguish between phase due to actual flowing spin and background phase. The presence of phase due to flow can perturb the results of the background phase estimation, effectively over or under-correcting for the background phase and impacting the final flow results. For this reason, this global correction is not usually applied when the data is being used for calculation of quantitative flow. An alternative approach is to manually select an ROI in a region containing stationary tissue. The average pixel value in this region or a localized polynomial fit can then be used to correct the PD data for the background phase. Another method is to use the phase map calculated from a phantom reference for an identical scan plane. In this work we describe a new automated method for calculation of the background phase in PD images that is not perturbed by the presence of flowing spins.

Methods: Background phase remains nearly constant throughout the cardiac cycle while blood exhibits significant pulsatility. This pulsatility can be estimated using the temporal frequency power spectrum. First, the PD image is spatially filtered to reduce noise (converting the phase and magnitude back to a complex value as an intermediate step). The Fourier transform is then taken along the time (cardiac phase) dimension. The result is a three-dimensional data set with the first two dimensions being the spatial pixel locations and the third dimension being the temporal frequency domain. From the temporal frequency data the power spectrum is estimated, from which the fraction of the total energy above 0 Hertz can be calculated. This value is defined here as the pulsatility index (PI) and varies between 0 and 1. A mask corresponding to the background tissue is defined by selecting all pixels below some threshold PI value (empirically 0.15 works well). The mask is then applied to the filtered PD image and a magnitude squared weighted least squares fit to a second-order polynomial is calculated for each cardiac phase. The phase calculated from this fit is then subtracted from the PD image.

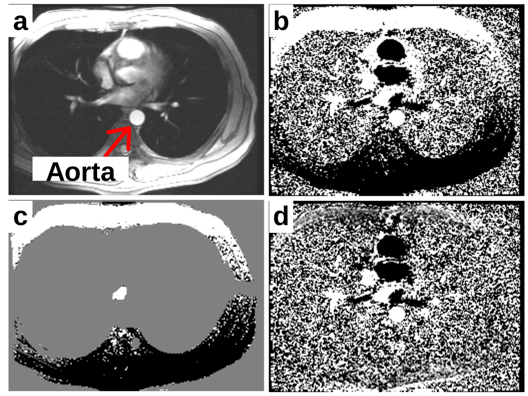

Figure 1.

To test the effectiveness of this algorithm, CINE PC scans of the descending aorta of a normal volunteer were acquired on a Signa EXCITE 1.5 T System (GE Medical Systems, Milwaukee, WI) equipped with EchoSpeed Plus gradients. Sequence parameters were: TE/TR = 3.1/6.5 msec, VENC = 100 cm/sec, oblique axial plane with flow encoding in the slice direction, 6 views per segment, heart-rate = 58 beats per minute, 32 x 24 cm FOV, 256x192 matrix, ±31.25 kHz bandwidth, 20° flip, and PD reconstruction. The effects of eddy currents were simulated by overcompensation of the digital pre-emphasis system. Three different eddy current settings were compared, i) no additional compensation, ii) 0.5% overcompensation of the Z→X cross-term, and iii) 0.5% overcompensation of the Z→Y cross-term. The time constants of all eddy currents were set equal to the TE to maximize the effect. Reference phantom data was also acquired for comparison for each of the three EC cases. All post-processing was done using Matlab (Natick, MA) software on a 1.2 GHz Pentium III personal computer. The PD data was corrected for background phase using a global correction that does not distinguish between flow and background, the new PI mask limited method, and the phantom reference correction. Flow through the descending aorta was then calculated using a region of interest around the aorta that included all pixels within 50% of the maximum magnitude pixel.

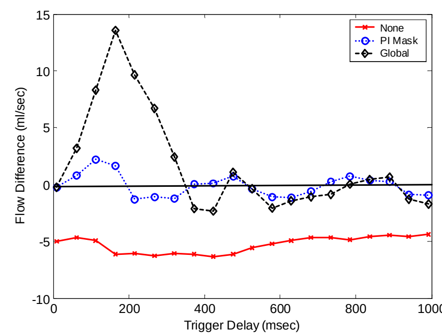

Figure 2.

Results: Figure 1 shows the image data for the case with the Y eddy current at a trigger delay of 216 msec corresponding to peak systolic flow, where a) is the magnitude image, b) is the uncorrected PD image, c) is the mask calculated from a PI value of 0.15, and d) is the corrected PD image using the PI limited mask. Figure 2 compares the difference between the measured flow for each correction techniques and the phantom reference correction for the Y eddy current case. Table 1 shows the mean and standard deviation of the flow difference for this case and the other two, X eddy current, and no additional eddy currents.

ml/sec	None	X EC	YEC
None	-1.62 +/- 0.33	1.38 +/- 0.26	-5.26 +/- 0.71
PI Mask	0.58 +/- 0.43	0.13 +/- 0.77	-0.07 +/- 0.99
Global	0.11 +/- 3.97	-0.59 +/- 4.34	1.63 +/- 4.46

Table 1.

Discussion: From Figs. 1 and 2 it is apparent that the PI mask technique effectively removes the background phase without being affected by flow-induced phase. Fig. 2 shows that the global fit method is perturbed by phase due to flow, particularly during systole when the velocity induced phase is the largest. For the cases shown here, the global correction is still better than no correction at all.

Conclusion: A new automated background phase correction method for PC has been described. This method has been shown to correct for residual background phase independent of flow related phase. The method compares favorably to a phantom based reference method but has the advantage of being completely automated.

References: [1] In Den Kleef JJE, Groen JP, DeGraaf RG. US Patent #4,870,361.