Eddy Current Corrections for Phase Contrast MRI Using Gradient Calibration

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INTRODUCTION Phase Contrast (PC) MRI is routinely used for the visualization of vessels and quantitative assessment of blood velocities. However, inaccuracies in the velocity measurements from acceleration and displacement artifacts and undesired phase offsets have hampered its clinical usage. Such phase offsets arise mainly from concomitant gradient fields, trajectory errors, and uncompensated eddy current effects and can lead to significant errors in quantitative measures such as flow [1]. Unlike concomitant field effects, trajectory errors and eddy currents are not easily described or corrected for. While the advances in gradient hardware with higher maximum gradient strengths and more rapid ramp times have facilitated shorter TR and TE times they have also lead to larger phase errors from eddy currents and concomitant gradients. Various approaches have been developed to reduce these artifacts [1-3]; however, these methods either require specialized hardware or cumbersome phantom scans, or rely on a possibly flawed segmentation. Here we propose and evaluate a novel correction technique that relies on thin-slice gradient calibrations performed immediately following actual measurements. With knowledge of the actual trajectory vs. the nominal trajectory, the phase errors can be corrected for in the reconstruction process.

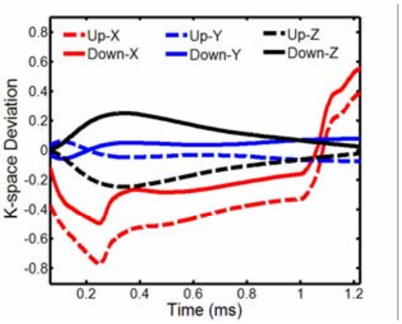

Figure 1. k-space deviations as measured for the 3D Cartesian sequence. Long time constant eddy currents remain from bipolar gradients leading to temporally changing k-space positions over the readout window.

THEORY Thin slice gradient calibrations can measure the actual k-space trajectory of any single gradient [4], which for PC MRI would require a measurement for every TR to be played out during the acquisition. In order to dramatically reduce the scan time required for the calibration measurements, we propose to assume the gradients to perform as a linear system. As such the k-space trajectory can be represented as a sum of the deviations from each gradient: $\Delta k = \sum_{i=0}^{N} a_i \Delta k_i$ where N is the total number of independent gradient waveforms, a_i is the relative amplitude of the ith gradient waveform, k_i is the measured trajectory, and Δk is the total deviation expressed as k-space sampling points. This allows us to measure only each independent gradient and determine the trajectory from each TR by combining those measurements. For example, for a fully 3D radial sequence with 3D flow encoding, 3 measurements are required for each axis: one for each lobe of the bipolar flow encoding gradients and one for the readout gradient. Any encoding during the acquisition is a linear combination of these gradient waveforms.

METHODS All imaging was performed on a 3.0T clinical scanner (Excite HD, GE Healthcare, Waukesha, WI). The calibration measurements were implemented into 2D and 3D radial and Cartesian spoiled gradient echo PC sequences and took 48 s during free breathing. To evaluate the performance of the corrections, a static phantom was imaged using 3D radial and Cartesian sequences. Exams were performed with practical clinical parameters to investigate the effect of readout bandwidth, gradient derating, and Venc selection. 3D polynomials where fit to velocity images to separate 0^{th}, 1^{st}, and 2^{nd} order phase errors. For in-vivo evaluations, pulmonary aortic flow, Qp, and systemic aortic flow, Qs, was measured in healthy volunteers (n=5) using 2D radial and Cartesian sequences with through plane flow encoding. Corrected vs. uncorrected values were compared for consistency of the Qp/Qs ratio, which is expected to be 1.05, different from unity due to the roughly 5% of coronary flow that is unmeasured.

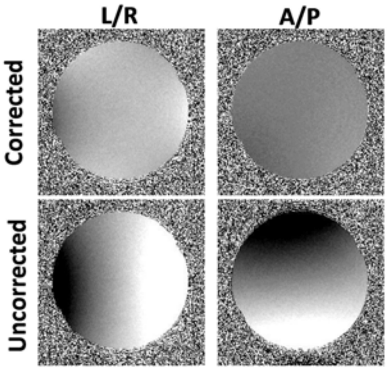

Figure 2. Example corrected and uncorrected Cartesian images showing significant reduction in linear offsets in both the L/R and A/P directions.

RESULTS An example for a measured k-space trajectory is shown in Figure 1 for the central line of a 3D Cartesian sequence (X readout direction). Long time constant eddy currents are evident, particularly along the S/I gradient. Figure 2 shows representative images with and without corrections applied. Both radial and Cartesian trajectories show significant improvement by reductions of linear phase terms in the velocity images. Polynomial fitting found that corrections effectively remove 1^{st} order phase errors and reduce 2^{nd} order phase errors; however, 0^{th} order errors are not removed. For example, with an exam with a Venc of 25cm/s corrections reduced the maximum 1^{st} order error from greater than 16% of Venc to less than 3% for Cartesian and radial exams. It was also found that gradient derating, reducing the readout bandwidth, and raising the Venc were effective methods to reduce both 0^{th} and 1^{st} order errors. In volunteers, corrections do not directly remove DC phase offsets; however, the removal of 1^{st} and 2^{nd} order offsets facilitates simple ROI measurements for corrections. Example in-vivo images are shown in Figure 3 for a radial exam. Radial images show additional reduction in image distortions caused by the Non-Cartesian trajectory. With corrections and a ROI measurement in distant static tissue, measured Qp/Qs were 1.057 ± 0.04 and 1.058 ± 0.03 for radial and Cartesian exams respectively, nearly identical to the expect value of 1.05. However without corrections, Qp/Qs values were 0.96±0.19 and 1.0±0.20 for radial and Cartesian exams respectively, showing significant deviation from the expected value.

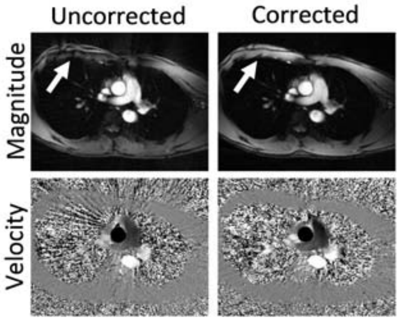

Figure 3. Example radial corrected and uncorrected in-vivo images showing improved additional reduction of distortions from trajectory errors.

DISCUSSION Gradient eddy current calibrations can be incorporated into existing PC sequences to reduce linear phase errors in velocity images. This additionally corrects gradient errors which cause 2^{nd} order errors and distortions in Non-Cartesian sequences. With correction in place, protocols can be optimized to minimize TEs and/or maximize spatial and temporal resolution, allowing for significantly more accurate quantification. 0^{th} order phase errors are likely due to eddy currents induced into the main field magnet, as such they are not corrected for through gradient calibrations. In future studies we seek to evaluate main field eddy current errors.

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