Higher order weighted least-squares phase offset correction for improved accuracy in phase-contrast MRI

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INTRODUCTION

Phase-contrast magnetic resonance imaging has the ability to accurately measure blood flow and myocardial velocities in the human body. Unwanted spatially varying phase offsets are, however, always present and may deteriorate the measurements significantly. Some of these phase offsets can be estimated based on the pulse sequence (1), but effects caused by eddy currents are more difficult to predict. A linear fit of the phase values is often estimated from either a number of manually defined areas containing stationary tissue or by semi-automatic detection of stationary tissue using the temporal standard deviation in an image sequence (2). With the advent of shorter magnets and the usage of shorter echo times, a linear model is no longer sufficient to describe the phase offset variations. In many applications, e.g. particle trace visualization and assessment of myocardial deformation, it is critical to compensate for the phase offset variations over a large region. In this study we investigate the variation of the phase offset over the complete 3D volume of interest and propose a completely automated correction approach based on a weighed fit of a high order polynomial.

METHODS

The proposed approach corrects time-resolved 3D velocity fields for phase offset variations using a weighted least-squares fit to the measured velocity data. First, the temporal standard deviation of the velocity, $SD_v(x,y,z)$, is calculated for every voxel at location x,y,z. In order to include additional information about the certainty of the estimated phase values, the standard deviation map is combined with magnitude information $m(x,y,z,t)$ as:

$$w(x,y,z,t) = \frac{m(x,y,z,t)}{SD_v(x,y,z)^p} ,$$

where p is controlling how much the standard deviation term should dominate the weighting. The magnitude image and the resulting weighting map are defined for all time frames, t, while the SD_v map will be constant for all time frames. The weighted least-squares fit is then computed by solving the minimization problem:

$$\min_{a_i} \sum_{j=1}^{n} \left(w_j \left(v_j - \mathbf{B}_{ij} a_i \right) \right)^2 ,$$

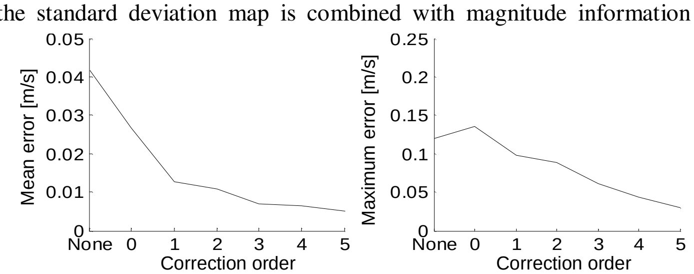

Figure 1. Mean velocity (left) and maximum velocity (right) in a stationary phantom before (none) and after correction using a first to fifth order approximation

where w is the weighting function, v is the measured velocity data, **B** is the chosen set of basis functions and n the number of voxels. The basis functions used in this study are polynomials of first (linear) to fifth order. The exponent p was set to 2. Once the coefficients, a_i, have been determined, the same basis is used to evaluate the fitted function at all locations and the obtained values are subtracted from the original velocity data.

RESULTS

A phase-contrast gradient echo pulse sequence on a clinical 1.5 T scanner (Philips Achieva, Philips Medical System, Best, The Netherlands) was used to acquire three-dimensional three-directional time-resolved velocity field data in a stationary phantom (FOV 324×324×320 mm, TR=4.8 ms, TE=2.9 ms, VENC=1.5 m/s, flip angle=8º). The mean and maximum background phase offset over the complete phantom decreased when applying a correction using a higher order polynomial (see Fig 1).

For in-vivo data, the proposed correction showed to be robust and to result in a decreased phase offset variation, as demonstrated in Figure 2 on a dataset acquired on a healthy volunteer (FOV 240×240×99 mm, TR=6.2 ms, TE=3.7 ms, VENC=1.0 m/s, flip angle=8º). Using the proposed correction method, improved particle trace visualization was obtained in several cardiovascular applications.

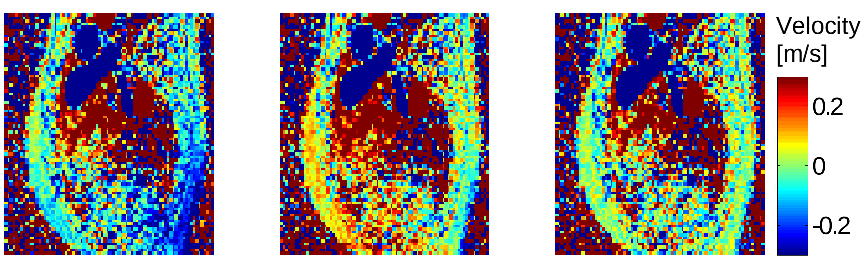

Figure 2. Through-plane velocity component in an oblique sagittal slice without correction (left), after linear correction using a mask based on the thresholded temporal standard deviation (1)(middle), and after correction using a weighed fit of a fifth order polynomial (right)

DISCUSSION

We have shown that the spatial variation of the phase offset can be far from linear. A linear correction technique may therefore be insufficient for many applications. A higher order correction puts higher demands on the amount and spatial distribution of voxels containing stationary tissue which may lead to instability and even larger errors. Using a weighted fit instead of a hard threshold for the classification of stationary tissue reduces this risk considerably.

REFERENCES

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