Offsets Correction in MR Phase Contrast Velocity Quantification within the Thorax

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Introduction

A known error in phase contrast velocity quantification is an offset caused by imperfection of the magnetic field gradients [1]. This offset depends on the spatial location and the precise setting of the gradients. Of the two main causes of this offset, uncompensated eddy currents and Maxwell concomitant terms, only the second one can be straightforward corrected [2]. In most locations in the body, stationary tissue can be found close to the vessel of interest, allowing a correction of the remaining offset. However, for the large vessels in the thorax there is hardly stationary tissue nearby. Walker et al. [3] proposed in 1993 a solution to this problem.

The purpose of this study was to verify whether this method still works on data from a modern MR scanner with improved gradient performance. Secondly, we investigated possible improvements of higher order surface fitting or other algorithms to identify stationary tissue pixels.

Methods

We selected a series of 11 clinical flow studies either within the ascending aorta (n=6) or main pulmonary artery (=5) [V_{enc} range 75-200]. Additionally, in 3 healthy subjects the flow in the ascending aorta and the main pulmonary artery was measured with three different velocity sensitivities (V_{enc} of 50, 150 and 250 cm/s). The technique used was through plane phase contrast cine spoiled gradient echo pulse sequence (TR 11 ms, TE 4.8 ms, and excitation angle 15°, receiver bandwidth 190 Hz/pixel). The spatial resolution was 1.3x1.3x8 mm^3 with an imaging matrix of around 256x208. Retrospective ECG triggering was applied. The measurements were performed on a 1.5 Tesla whole body MR system (Magnetom Sonata, Siemens, Erlangen, Germany) with a 4-element phased array receiver coil. In the reconstruction the Maxwell concomitant terms were corrected to order $1/B_0$ [2]. As the 'gold' standard for the velocity offset separate measurements in a stationary phantom were performed directly after the volunteer study, using the exact copy of the *in-vivo* parameter setting [4]. To study the effect of protocol changes besides different velocity sensitivities 'extra' series were calculated from the acquired acquisitions. First the effect of a reduced time sampling within the cardiac cycle was simulated by down sampling the data by a factor of 3 and 5. Secondly, adding white noise to the complex data simulated the effect of a reduced signal to noise ratio (SNR), with a factor of 2 and 4.

In each velocity series a mask of stationary tissue pixels was created. Taking 15% of the pixels with the lowest temporal variance of the velocity over the cardiac cycle generated this mask [3]. Besides this originally proposed algorithm we investigated also other algorithms. The time-averaged velocity values within this mask were used to fit a linear surface over the whole imaging plane for offset correction. Besides a linear fit also surfaces with polynomial terms up to the fifth power were fitted for determination of the optimum order.

In post-processing, regions of interest (ROI) were drawn in the images around the aorta or pulmonary artery and a region in the stationary tissue of the frontal thorax wall outside potential ghosting bands. The velocity offset was determined at the location of these ROI's both in the in-vivo images, the phantom images, and on the fitted surfaces. For the ROI's drawing the FLOW® software (*Medis, Leiden, the Netherlands*) was applied, whereas all other data processing was performed using custom written software in the MATLAB® (*version 6.5R13, the MathWorks, Inc.*) environment.

The data are presented as difference values between offset based on the surface fit to the mask from the *in-vivo* study and the offset determined in the phantom on the same spatial location using the same imaging protocol. The data are expressed as mean and root mean square (RMS) differences.

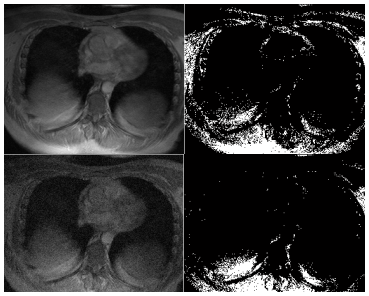

Figure 1: static tissue mask failure in case of inhomogeneous signal and limited SNR. Bottom row with artificial reduced SNR.

Results and Discussion

In the vessels of interest (aorta and pulmonary artery) a velocity offset was observed of 0.8 cm/s [-1,2 to 3.2] with RMS value of 1.3 cm/s, as determined with the phantom measurements. This RMS error is high for clinical studies. A useful accuracy of 5% in cardiac output corresponds for normal physiology to an offset below 0.6 cm/s at the location of the aorta or pulmonary artery.

First the assumption of the phantom measurement as a standard for the *in-vivo* offset determination was checked. The phantom measurement agreed to the *in-vivo* stationary tissue in the thorax wall with a mean difference of 0.11 cm/s with a RMS value of 0.5 cm/s.

The standard algorithm worked well in most cases. However, the linear fit is influenced by spatial aliasing. In a subset of 5 subjects the RMS decreased from 0.5 to 0.3 cm/s when aliased structured were removed from the mask. Another situation where the masked failed was the combination of inhomogeneous signal intensity and limited SNR (see figure 1). When corrected for spatial aliasing the mean difference in offset between fit and phantom was 0.15 cm/s with an RMS value of 0.5 cm/s. No effect was observed by reduced number of cardiac phases or change in velocity sensitivity. A weak increase of the RMS value (up to a mean value of 0.7 cm/s) was observed for the lowest SNR values.

Higher order polynomial surface fits didn't improve the accuracy of the correction, and resulted in a larger deviation of the offsets (see figure 2). Thus, a linear surface fit appeared to be the most optimum.

Other tested algorithm to locate stationary tissue pixels didn't show improved performance. First we tried other thresholds on the variance image, an absolute threshold and a threshold relative to the V_{enc}. They fail in case of limited SNR. A similar performance and similar limitations as the standard method were observed using a threshold on the first or 2nd harmonic component of the velocity power spectrum. An algorithm proposed at the ISMRM last year appeared to work well in case of large offset but failed completely in cases of limited offsets [5]. Temporal autocorrelation of the velocity data showed clear segmentation of the vessel to remove those from the mask, but failed in case of limited cardiac phases.

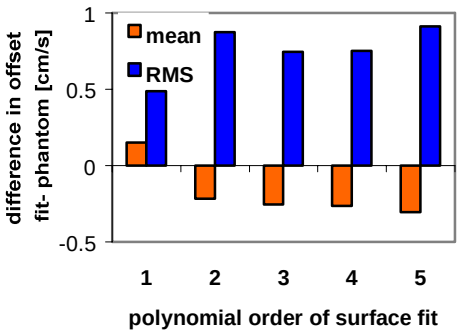

Figure 2: effect order of surface fit

Conclusions

It appeared that the proposed algorithm by Walker et al. works reasonable well on data obtained on a modern scanner. After Maxwell concomitant terms correction the remaining offset has a mainly linear spatial component. With this offset filter an accuracy is obtained within the same range as with a separate phantom measurement. Fully automatic implementation is limited because manual editing of the stationary tissue mask is necessary in case of spatial aliasing or strong spatial inhomogeneity in combination with limited SNR.

References

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