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## Chapter 7

# A multi-center inter-manufacturer study of the temporal stability of phase-contrast velocity mapping background offset errors

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**Abstract**

*Background:* Phase-contrast velocity images often contain a background or baseline offset error, which adds an unknown offset to the measured velocities. For accurate flow measurements, this offset must be shown negligible or corrected. Some correction techniques depend on replicating the clinical flow acquisition using a uniform stationary phantom, in order to measure the baseline offset at the region of interest and subtract it from the clinical study. Such techniques assume that the background offset is stable over the time of a patient scan, or even longer if the phantom scans are acquired later, or derived from pre-stored background correction images. There is no published evidence regarding temporal stability of the background.

*Methods:* This study assessed the temporal stability of the background offset on 3 different manufacturers scanners over 8 weeks, using a retrospectively-gated phasecontrast cine acquisition with fixed parameters and at a fixed location, repeated 5 times in rapid succession each week. A significant offset was defined as 0.6 cm/s within 50 mm of isocenter.

*Results:* Over the time of 5 cine repetitions, insignificant temporal drift (0.1 cm/s, 0.2 cm/s) in the baseline offset occurred on two machines, with a marginally insignificant 0.5 cm/s on the third due to an apparent heating effect. Over a longer timescale of 8 weeks, insignificant drift (0.2 cm/s) occurred on one, with larger drifts (0.9 cm/s, 0.6 cm/s) on the other machines.

*Conclusions:* During a typical patient study, background drift was insignificant. Extended high gradient power scanning with work requires care to avoid drift on some machines. Over the longer term of 8 weeks, significant drift is likely, preventing accurate correction by delayed phantom corrections or derivation from pre-stored background offset data.

## 7.1 Background

Phase-contrast velocity mapping is applied mainly to determine blood flow [1], with other applications in cerebrospinal fluid flow [2] and tissue velocity mapping[3]. Velocity images are formed by subtracting the phase images of two sequence repetitions acquired with differing velocity sensitivity. As is well known, other sources of phase difference between the two images cause stationary tissue to display an apparent non-zero velocity, known as the background offset or baseline error. This offset may vary gradually with position over the velocity image and underlies stationary and moving tissues. Maxwell (or concomitant) gradient effects are one cause which can be corrected analytically [4]. Eddy currents are another cause, and these are corrected to a large extent by activelyshielded gradient coils and pre-emphasis [5, 6]. After these corrections, the main cause of background offset is residual errors in the pre-emphasis which accumulate phase errors during the time between velocity-encoding and echo. The “residual error” here refers to the difference between pre-emphasis and the fields generated by the eddy currents. The residual error may be positive or negative and indeed change polarity during the decays of its constituent terms, and a difference between the two sequence repetitions for velocity-encoding leads to the background offset. In addition to this complicated temporal behaviour, the pre-emphasis uses the linear gradient coils whereas the fields from eddy currents may contain higher spatial orders. The background offset can sometimes be reduced by running some functions of the pulse sequence more slowly [7], but this is not always compatible with the fast sequences needed for applications such as cardiovascular breath-hold imaging.

The background offset error may reach clinical significance only in certain situations. The most typical of these is where the through-plane velocity data is integrated over a vessel cross-section, then summed again over all the cine frames of the cardiac cycle to obtain cardiac flow measurements. The clinical impact of this error may increase with the large cross-sectional area of dilated vessels, for patients with a slow heart-rate (long R-R interval) and also for measurements derived from multiple flow acquisitions, such as the

$Q_p/Q_s$  ratio. In other applications such as peak velocity assessment the background offset is usually negligible.

The background offset error may be corrected using the apparent velocity values in stationary tissue close to the vessel of interest. This correction is sufficient in most applications outside the thorax. However, in case of flow quantification in and around the heart adjacent stationary tissue is missing. This can be solved by determining the offset in stationary tissue from the whole FOV and subtract it using a fitted surface [8, 9], thereby correcting the offset at the location of the vessel of interest. This approach is limited by several requirements: sufficient stationary tissue at adequate SNR, avoidance of phaseencode FOV wraparound, avoiding steady venous flow in the stationary tissue mask, and elimination of signal far from isocenter where the error may be highly nonlinear. All of these factors hinder reliable automatic correction by this approach. Another technique avoids the need for stationary tissue by magnetic field monitoring using signals from an array of field probes [10] (a “field camera”) placed around the patient to measure the background error [11], although such technology is not yet routinely available.

A more basic method measures the background offset by repeating the flow study using a uniform stationary phantom after the patient, with identical sequence parameters [12, 13], and this approach is currently considered as the “gold standard” for correction. The phantom measurements are subtracted from the patients velocity study to correct the background offset. One manufacturer advises this correction method in its application guide for flow measurements around the heart. This approach has the drawback of requiring a stationary fluid phantom scan replicating each flow study of each patient, which is simple in principle but is inconvenient at clinical centres. This drawback might be circumvented by a pre-stored set of corrections for a sufficiently broad range of plane orientations and image parameters for the clinical through-plane flow studies that most usually require this correction. Evidently, both correction methods depend on temporal stability of the velocity offset.

This work evaluated the temporal stability of phase-contrast background or baseline velocity offsets. There appear to be no previ-

ous publications in this topic for phase-contrast imaging. An inter-scanner study of gradient amplifier response delays has shown high stability [14]. Furthermore, it can be shown that gradient anisotropy (i.e. of inter-axis delays [15]) has no impact on velocity-encoding background offset errors even for oblique image planes. For the interleaved acquisition of the two sets of phase images used in the phase-contrast technique, the effects of main field-distortion known to arise from heating of the passive shim steel should be almost exactly cancelled. However, small sequence timing differences, even including programming errors, between the two sets of phase images may exacerbate sensitivity. Probably the main factor is the temporal stability of gradient amplifier performance, and also the stability of eddy-current generation and their compensation.

The temporal stability of the background error might be questioned on three timescales. First, is it stable over a long period enabling the use of pre-stored corrections? Second, is the background error stable during the patient session, so that a phantom scan acquired at some time after the patient session is a reliable measurement of the error during a clinical flow study? Third, is the background error stable during all the images of a cine velocity map, even where it was retrospectively gated with continuous gradient waveform activity? The third question was confirmed long ago [16]. In the absence of published work answering the first two questions, we aimed to measure the temporal stability of background offset errors.

## 7.2 Methods

### MRI test phantoms

The specification was to measure apparent velocity in a stationary uniform phantom, up to 50 mm distance from isocenter as measured in the image plane. Each centre used a different test phantom of gadolinium doped gelatine, or water which was allowed to settle for 5 minutes before beginning scanning. In one instance, to confirm the fluids stability, repeated phase-contrast scans were acquired during the settling time, showing that fluid settled to  $<0.1$  cm/s.

**MRI systems and Phase contrast velocity acquisitions**

The study was limited to 1.5T whole body systems as 1.5T is currently the most widely-used main field strength for cardiac MRI. Pre-emphasis for reduction of eddy current effects and automatic correction of concomitant gradient terms<sup>4</sup> were employed as implemented by the vendor, whereas any other filtering or correction of background offset errors was turned off because this would perform unrealistically well on large uniform phantoms. Three 1.5T scanners were used, one from each of three manufacturers (GE Signa Excite, Philips Intera and Siemens Avanto) placed in random order as scanner types 1, 2 and 3. Each centre acquired the same single plane at 45° oblique between transverse and sagittal planes, with anterior-posterior phase-encoding, at isocentre. This slice orientation is used clinically, and showed in an earlier study to be sensitive to offsets [7, 14, 17]. On each machine, the same plane was repeated 5 times in rapid succession during each approximately weekly session, over a period of 8 weeks. No MRI service engineering recalibrations were performed during this time.

On six machines, two from each of the 3 manufacturers, a further study examined the effect of a high gradient-power sequence run continuously for at least 5 minutes between two sets of 5 cine studies as acquired above. This was acquired at a later date than the main 8-week data, following initial evaluation of that data, and two machines of each type were used to test the consistency of their behaviour.

Similar phase-contrast sequence parameters were used for each of the three scanner types, although we emphasise that the background offset values themselves were not being compared between scanners in this study, rather their drift with time. It was therefore essential that no changes in sequence parameters or slice locations were made during the 8 weeks. All cine phase-contrast acquisitions were retrospectively gated pulse sequences, where the phase-encode group was updated by each simulated ECG R-wave. The continuous gradient activity of this approach, with no silent gap while waiting for the next R-wave, has the advantage of a more stable background offset during the cardiac cycle [16]. Multiple signal averages were applied to ensure adequate

SNR for measurement of the small velocity offsets. Some aspects of the pulse sequence were beyond our control using standard clinical sequences, and these are listed below for each type of scanner. Unless stated below, the velocity encoding was asymmetric (i.e. it used phase-subtraction of velocity-compensated and velocity-encoded sequence repetitions). For all of the sequences, the gradient-echo was asymmetric (i.e. the gradient-echo rephased early in the ADC sampling window for short TE). The TR values stated were between the RF excitation pulses. Systems and parameters used were as follows:

GE Signa Excite at VUmc Amsterdam NL, and Addenbrookes Hospital Cambridge, UK. TR 5.9 ms TE 3.0 ms. The readout ADC bandwidth was 41.67 kHz (pixel bandwidth 326 Hz/pixel). 128 phase-encode steps, 6 segments, 6 mm slice-thickness, 1.25 mmFE  $\times$  2.5 mmPE resolution, 2 averages. This used symmetric velocity-encoding (i.e. two sequence repetitions with positive and negative velocity sensitivities around the velocity compensated waveform, also known as “balanced” velocity-encoding). The “flow analysis” flag was on, disabling a spatial high-pass filter used for phase-contrast angiography background suppression. The GE “flow optimization” control resulted in longer TE and TR than the other scanners and was therefore not used. Twenty cardiac phases were reconstructed in a 600 ms simulated R-R interval (i.e. temporal interpolation was applied by reconstruction).

Philips Intera. Both systems were at Lund University Hospital, Sweden. TR 5.7 ms, TE 3.1 ms, pixel bandwidth 355 Hz/pixel (Fat/Water Shift 0.62 pixels). 138 PE steps, 6 segments, 6 mm SLT, 288 FE samples, 1.2 mmFE  $\times$  2.4 mmPE resolution, 4 averages. The “Gradient mode” was set to “Default” and “PNS mode” to “Moderate”. The slice selective RF pulse used an asymmetric design with a late centre. The background phaseoffset correction (“LPC filter”) was obviously switched off for this study. Fifteen cardiac phases were reconstructed (i.e. temporal interpolation was not performed during reconstruction).

Siemens Avanto. Both systems were at Royal Brompton Hospital, London, UK. TR 5.8 ms, TE 2.3 ms, sampling bandwidth 355 Hz/pixel. 128 phase-encode steps, 6 segments, 6 mm SLT, 1.25



mmFE  $\times$  2.5 mmPE resolution, 2 averages. The controls for RF pulse and gradient mode, which control the use of faster and stronger RF and gradient pulses, were both set to “Normal” mode. This sequence was run ungated to enable a rapid weekly test of background errors in multiple image planes at multiple slice shifts from isocenter, i.e. it did not produce a cine for each slice. The images from all other planes except the transverse-to-sagittal slice were subsequently left unused, but the total session time for the 5 repeated scans was similar to that of the other scanners. For assessment of the temporal variability during a cine sequence, a cine version of the above sequence was run once. The cine reconstruction interpolated 20 cine frames into the 1 second simulated R-R interval.

### Image analysis

All images were analyzed using software written in Matlab (The MathWorks, Natick, USA). The software reported the velocity offset measured as the mean within circular ‘great vessel regions (30 mm diameter) [18] centered 50 mm from the centre of each velocity map, which was at the isocenter. Four such regions were placed, i.e. at 50 mm to left and right, and 50 mm above and below, the centre of the image. The velocity offsets were measured at all 4 locations, aiming to detect temporal variations occurring in any direction from image centre (assuming those of shape  $xy$  over the image columns  $x$  and rows  $y$  to be negligible).

Two timescales of variations in the offsets were considered: among the 8 weekly sessions, and during each session of five scans. For these purposes, the mean over all the frames of cine acquisitions was taken for each ROI and the standard-deviation over these frames was also recorded. The standard-deviation over the cine frames was used only to check whether the background offset variations exceeded the level of random noise and systematic errors in the retro-gated cine reconstruction.

The following procedure was adopted for the purpose of determining a significant background drift. The largest peak-to-peak differences in the offsets were measured within each group of 5 scans and also across all 40 scans. This gave the largest error which could have

occurred if the stationary phantom correction method had been used at any time during the weekly scans or the full 8 weeks of data collection. The errors were compared against the 0.6 cm/s offset which has been suggested as liable to affect some parameters derived from clinical flow measurements [17].

### 7.3 Results

The results for all 4 ROIs placed on the 5 scans of each of weeks 1-8 are plotted for the three scanners as Figure 7.1. They show random variation during and between sessions, except for scanner 2 where a drift effect occurred during each session of 5 scans. Although the three graphs are all set to the same vertical axis ( $\pm 3$  cm/s), the values themselves are not relevant in this work, rather their temporal stability.

The largest peak-to-peak differences within any 5-scan session and across all 40 scans of the 8 weeks are shown in Table 7.1. The worst-case difference within any of the weekly sessions reached 0.14 cm/s, 0.50 cm/s and 0.23 cm/s for scanners 1-3 respectively, with the larger value for scanner 2 caused by the drift effect within the 5 scans of each session. Variations over the 8 week interval were larger, reaching 0.19 cm/s, 0.86 cm/s and 0.61 cm/s for scanners 1-3 respectively. The cine temporal standard-deviations were 0.07 cm/s, 0.06 cm/s, and 0.08 cm/s for scanners 1-3 respectively (largest of the 4 ROIs).

The results before and after 5 minutes of high-power scanning are plotted for all 4 ROIs placed on the 5 scans, for the two systems of each of the three scanner types (Figure 7.2). A largely random variation occurred, except for scanner type 2 where a drift effect occurred, with some variation between machines of the same type. The largest difference in any of the ROIs after high-power scanning was 0.12 cm/s, 0.81 cm/s and 0.42 cm/s for scanner types 1, 2, 3 respectively (between the means of each 5-scan session).

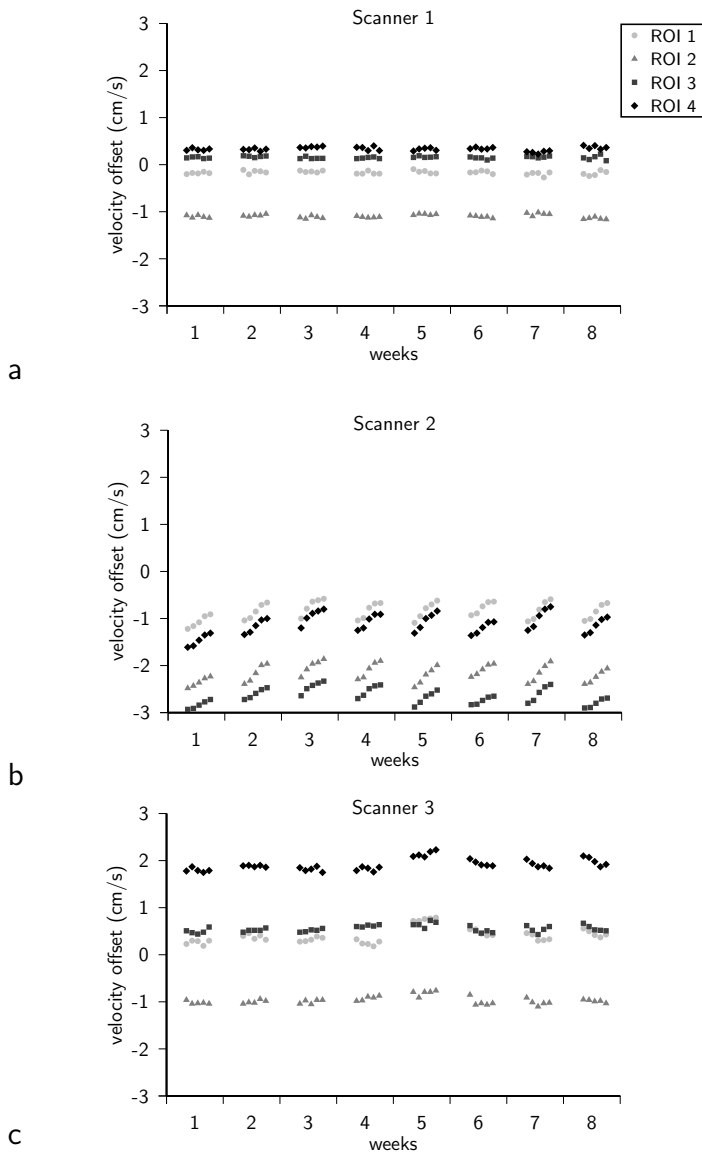


Figure 7.1: The ROI mean values on each of the 5 scans in each weekly session, for weeks 1-8.

Table 7.1: Worst-case variation that could affect the reliability of phantom correction of background offsets. The “Session” row is the largest peak-to-peak variation between the 5 scans in each week. The “Long-term” row is the largest peak-to-peak variation across all 40 scans. The 4 ROIs were placed at 4 locations consistently for each scan as explained in Methods.

Scanner		1			
ROI		1	2	3	4
Session	(cm/s)	0.12	0.08	0.14	0.10
Long-term	(cm/s)	0.18	0.15	0.14	0.20
Cine stdev	(cm/s)	0.06	0.07	0.06	0.06
Scanner		2			
ROI		1	2	3	4
Session	(cm/s)	0.47	0.48	0.40	0.50
Long-term	(cm/s)	0.64	0.62	0.60	0.86
Cine stdev	(cm/s)	0.06	0.05	0.05	0.05
Scanner		3			
ROI		1	2	3	4
Session	(cm/s)	0.19	0.21	0.19	0.23
Long-term	(cm/s)	0.61	0.34	0.30	0.48
Cine stdev	(cm/s)	0.06	0.08	0.04	0.07

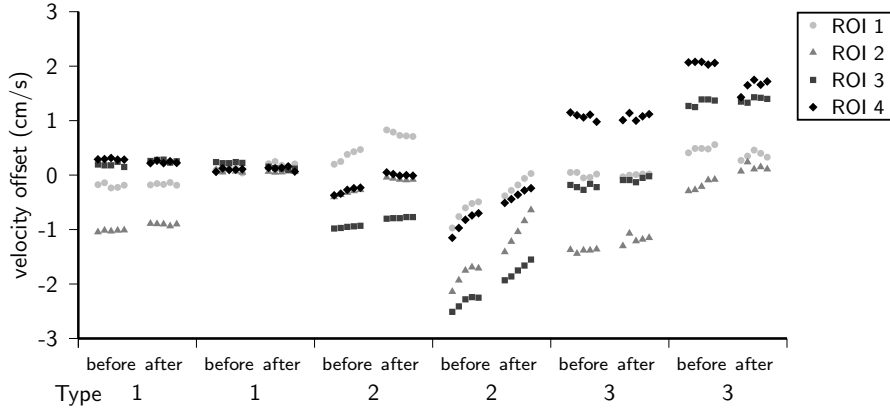


Figure 7.2: The ROI mean values from 2 scanners of each type, before and after high-power scanning. The horizontal gaps for high power scanning between the groups of five scans before and after are not to temporal scale, they were approximately as long as 20 velocity cines.

## 7.4 Discussion

Recalling the aim of reliable background offset correction for routine clinical use of phase-contrast flow measurements, we reconsider the correction methods explained in the introduction. The idea of pre-stored corrections appeared unlikely to be reliable on 2 of the 3 scanners tested, on which the peak-peak variation over 8 weeks approached or exceeded the 0.6 cm/s regarded as liable to introduce clinically significant errors in some situations. For correction using a stationary phantom after the patient, this was reliable to well below the 0.6 cm/s limit in all scanner types, although the drift observed in scanner type 2 might eventually violate this limit.

There was no significant impact of the high power scan for scanner types 1 and 3, where the difference was  $<0.6$  cm/s. However, for scanner type 2, a larger change occurred. The drift effect in scanner type 2 was perhaps related to heating of system components such as the gradient amplifiers, or alterations in the residual eddy-currents, as was shown by the effect of the high power scans.

This study implies that correction using the clinical phase-contrast scan itself is preferable, since this would annul the question of drift during the time between clinical application and a separate correction scan, but this demands further work on the automated background correction or field-camera methods mentioned in the Introduction. The results also suggest cautious use of phantom corrections as a reference for evaluating the performance of such methods.

## 7.5 Conclusion

Over the duration and activity of a typical patient study, background drift was insignificant. However, the combination of extended high gradient power scanning and work requiring background correction requires care to avoid drift on some machines. Over the longer term of 8 weeks, significant drift is likely, preventing accurate correction by delayed phantom correction scans or derivation from pre-stored background offset images.

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