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Baseline Correction of Phase Contrast Images Improves Quantification of Blood Flow in the Great Vessels

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ABSTRACT

Purpose: **Phase-contrast Cardiovascular Magnetic Resonance Imaging (CMR) generally requires the analysis of stationary tissue adjacent to a blood vessel to serve as a baseline reference for zero velocity. However, for the heart and great vessels, there is often no stationary tissue immediately adjacent to the vessel. Consequently, uncorrected velocity offsets may introduce substantial errors in flow quantification. The purpose of this study was to assess the magnitude of these flow errors and to validate a clinically applicable method for their correction.** ***Materials and Methods:*** **In 10 normal volunteers, phase-contrast CMR was used to quantify blood flow in the main pulmonary artery (Qp) and the aorta (Qs). Following image acquisition, phase contrast CMR was performed on a stationary phantom using identical acquisition parameters so as to provide a baseline reference for zero velocity. Aortic and pulmonary blood flow was then corrected using the offset values from the phantom.** ***Results:*** **The mean difference between pulmonary and aortic flow was 26 $\pm$ 21 mL before correction and 7.1 $\pm$ 6.6 mL after correction ($p = 0.002$). The measured Qp/Qs was 1.25 $\pm$ 0.20 before correction and 1.05 $\pm$ 0.07 after correction ($p = 0.001$).** ***Conclusion:*** **Phase-contrast CMR can have substantial errors in great vessel flow quantification if there is no correction for velocity offset errors. The proposed method of correction is clinically applicable and provides a more accurate measurement of blood flow.**

INTRODUCTION

Quantifying blood flow is often clinically important for guiding therapy. For example, in patients with congenital heart disease, the ratio of pulmonary to systemic blood flow (Qp/Qs) is often used to determine the need for and the timing of cardiac surgery (1). Phase-contrast CMR can noninvasively and reproducibly quantify blood flow (2, 3, 4) and has been shown to correlate well with radionuclide angiography and invasive oximetry for quantifying left-to-right shunts (5, 6).

For the most accurate blood flow quantification, phase-contrast images must be corrected for velocity offset errors that occur during image acquisition. These errors have been attributed to noncompensated eddy-current-induced fields as well as concomitant gradient field effects that are present on all commercial CMR systems (7–9). While many commercial CMR scanners perform an automatic correction of the concomitant gradient effects during phase-contrast image reconstruction (7), noncompensated eddy-current-induced fields can still introduce substantial errors in flow quantification. These errors result in a baseline shift of the velocity vs. time curve that is integrated to calculate blood flow. The magnitude of the baseline velocity shift depends on a number of imaging parameters, including where the vessel is in space (relative to the magnet isocenter), the imaging plane angles (theta and phi), and the velocity encoding gradient strength (Venc).

Small velocity offset errors often lead to much larger errors in blood flow quantification. This is because blood flow is calculated by integrating the velocity values within the cross-section of a vessel over time. The many small velocity errors sum into a larger flow error that increases linearly with the cross-sectional area of the vessel. This is illustrated in the following example. Consider the measurement of aortic flow in a patient who has an aortic cross-sectional area of 8 cm^2. If phase contrast images are acquired with a maximum velocity encoding gradient (Venc) of 200 cm/s, a velocity offset error of 1% (2 cm/sec) will result in a flow error of 16 mL per beat. If the patient has a stroke volume of

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64 mL/beat, the 1% velocity offset error results in a 25% stroke volume flow error.

To compensate for baseline velocity offsets, equipment manufacturers generally recommend that a background region of interest (ROI) be placed in stationary tissue that is immediately adjacent to the vessel of interest. Any nonzero velocity in the stationary tissue represents a baseline offset error, the magnitude of which is used to correct the flow in the vessel. Unfortunately, there is often no stationary tissue immediately adjacent to the heart or the great vessels. Using stationary tissue distant from the vessel (such as from the anterior chest wall in the case of aortic flow) is problematic because the velocity offset error often varies spatially within the image.

Others have proposed postprocessing methods to correct for velocity offset errors. One group of methods proposes estimating the velocity offset error in the vessel by examining the velocity offsets in somewhat distant stationary tissue (e.g., chest wall, liver, etc.) and spatially fitting the data using linear or higher-order interpolation (9–11). A drawback of this technique is that there may be insufficient stationary tissue to accurately determine the offset error in the vessel. Another method is to measure the phase-offset directly in the area of interest by repeating the imaging sequence on a stationary water bottle ("phantom") (12). Although this technique has the drawback of requiring additional imaging time after the patient completes the MRI study, it can be accomplished in just a few minutes.

We chose to compare pulmonary (Qp) to aortic (Qs) flow in normal volunteers to quantify the magnitude of the error on blood flow measurements, as well as to determine the effectiveness of our correction protocol. We thought this was an especially good model because of our observation that when phase contrast images are acquired separately perpendicular to the main pulmonary artery and perpendicular to the ascending aorta, the baseline error almost always resulted in an overestimation of aortic flow and an underestimation of pulmonary artery flow. This difference in effect of the baseline error of the flow accentuated the measured difference between and ratio of Qp and Qs.

METHODS

Image acquisition

All human studies were performed with the approval of an Institutional Review Board. Ten healthy volunteers (six male and four female) with no known left-to-right shunt were imaged using a 1.5 Tesla TwinSpeed scanner with version 12.0 software (GE Healthcare, Milwaukee, Wisconsin, USA). Phase-contrast images were acquired, perpendicular to the proximal ascending aorta and to the proximal main pulmonary artery, ~1 cm distal to the semilunar valves. Breath held images were acquired using the commercially resident FastCine-PC pulse sequence, which uses continuous, uninterrupted rf excitations, prospectively gated phase-encoding, and retrospectively gated image reconstruction. Continuous, uninterrupted rf excitations are beneficial for accurate flow imaging because it stabilizes the long-time constant contributions to the eddy-current background errors (13). Commercially resident FastCine-PC pulse sequence automatically compensated for the concomitant gradient affects. Acquisition parameters were as follows: TR = 7.2 ms, TE = 2.8 ms, field of view = 35 cm, slice thickness = 8 mm, matrix size = 256 × 128, bandwidth = 31 kHz, views per segment = 8, Venc = 150 cm/s, 30 reconstructed phases. Nominal temporal resolution for each image is 8 × 2 × 7.2 ms = 115 ms. A total of 6 phase contrast acquisitions were made, three for each artery. The acquisitions were made in an interleaved fashion so as to minimize the effect of any systematic errors from time-varying changes in cardiac output. After the flow images were acquired, a large bottle of water ("phantom") was then imaged with identical phase-contrast imaging parameters, to serve as the references for aortic and pulmonary artery flow. An ECG simulator set to 60 beats per minute was used to trigger image acquisition for the phantom studies.

Image analysis

Images were analyzed using the ReportCard 1.0, and 2.0 software (GE Healthcare, Milwaukee, Wisconsin, USA). Uncorrected arterial flow was quantified by using the flow analysis tool and placing a region of interest (ROI) around the artery of interest (ascending aorta or main pulmonary artery). To correct for baseline velocity offsets, a ROI of the same size and location was placed on the corresponding phantom image. Figure 1 shows an example of ROI placement. Flows in the phantom images were corrected by applying a baseline velocity shift so that flow was zero. The baseline correction for phantom images was observed to be quite reproducible from scan-to-scan provided there was no change in the acquisition parameters (including the imaging plane orientation). Flows were corrected in the aorta and pulmonary artery by applying the corresponding baseline shift that zeroed the flow in the phantom data using the ReportCard software. Subtracting an average baseline rather than performing a subtraction on an image-by-image basis avoids a 40% penalty in SNR.

Statistical analysis

Statistical analysis was performed using Microsoft Excel 2000 (Microsoft, Redmond, Washington, USA) and QuattroPro10 (Corel, Ottawa, Canada). Flows were calculated as the average of three measurements. A paired t-test was used to compare the differences in the flows of the great vessels before and after the baseline correction. In addition, the ratio of pulmonary to systemic flow (Qp/Qs) was calculated both before and after baseline correction. They were compared to the expected Qp/Qs of 1.05 using paired t-test assuming unequal variances. Since the coronary circulation in a normal person is approximately 5% of the cardiac output (14) and the measurement of the systemic blood flow (Qs) is made distal to the coronary ostia, the expected normal Qp/Qs is 1.05 when measured with this technique.

For determining the correlation between velocity offset error and vessel cross-sectional area, the values for the aorta and main pulmonary artery were averaged for each patient. Similarly, the

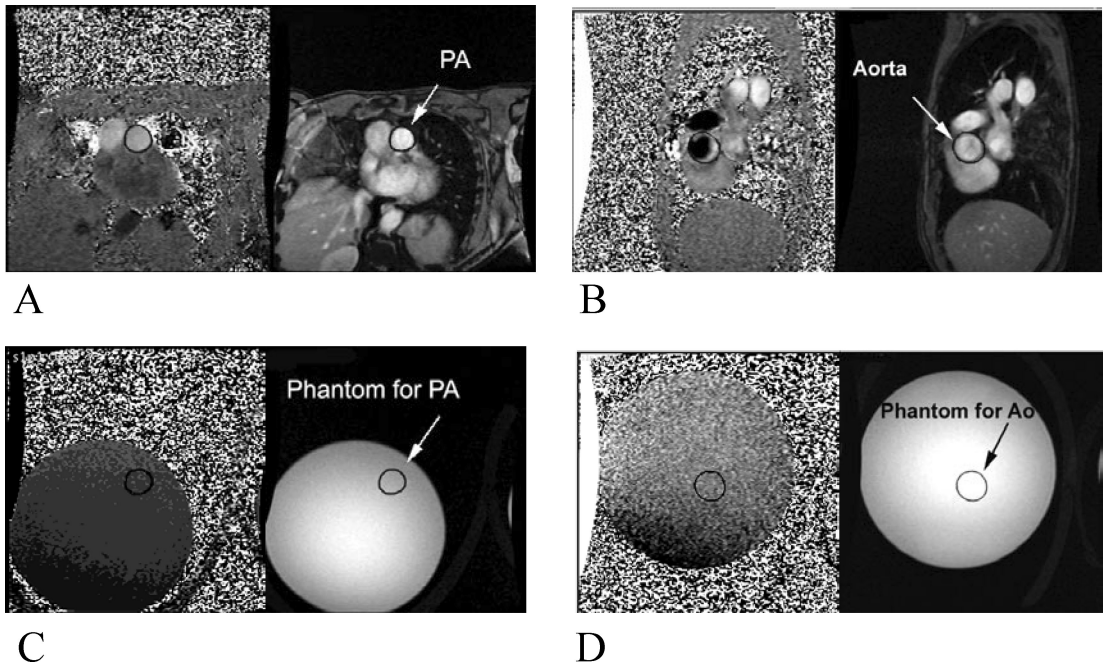

Figure 1. ROI placement is shown for measuring (A) pulmonary flow and (B) aortic flow. ROI placement for the corresponding phantom images is shown in (C) and (D). Note the shading the in both phantom phase contrast images, indicating the spatial variation of the velocity offset error.

anterior offset from magnet isocenter was calculated as the average of the anterior offset for each vessel. Multiple regression analysis was performed to determine the relative influence of vessel size and its anterior position on the velocity offset error.

RESULTS

Table 1 shows the quantitative results from the flow studies of the ten volunteers. The average uncorrected Qp/Qs is 1.25 ± 0.20, which is significantly different from the expected Qp/Qs of 1.05 ($p = 0.005$). As shown in Fig. 2, after correcting for the velocity offset error, the average Qp/Qs is 1.05 ± 0.07 ($p = 0.001$ compared to uncorrected flow), which is not significantly different from the expected Qp/Qs of 1.05 ($p = 0.45$).

Figure 3 shows that after velocity offset correction the difference between pulmonary and aortic flow decreases and the variability of this difference decreases. Specifically, the mean difference in flow decreases from 26 ± 21 mL before correction to $7.1 + 6.6$ mL after correction ($p = 0.002$).

We observed a strong direct relationship between the size of the great vessels and the magnitude of the flow correction ($r^2 = 0.91$, Fig. 4). This was expected given that flow is the product of velocity and cross-sectional area. We also observed a moderate positive linear correlation between the anterior position of the great vessels and the magnitude of the correction ($r^2 = 0.64$). However, multivariate analysis shows it was the size of the vessels and not their anterior position *per se* that was predictive of the magnitude of the velocity offset error

Table 1. Summary of measurements from phase-contrast images

Volunteer	Uncorrected aortic flow (mL)	Uncorrected pulmonic flow (mL)	Corrected aortic flow (mL)	Corrected pulmonic flow (mL)	Avg. Position anterior to iso-center (mm)	Avg. Vessel cross-sectional area (mm^2)	Total offset correction (mL)
1	93	108	101	96	38	816	20
2	84	100	91	93	20	786	11
3	92	139	85	96	80	1169	37
4	99	112	103	101	17	818	15
5	83	103	90	101	17	722	9.3
6	124	164	136	141	60	973	35
7	107	132	116	120	61	870	20
8	95	91	104	113	54	682	13
9	111	181	121	144	73	1211	48
10	61	72	67	67	42	709	11

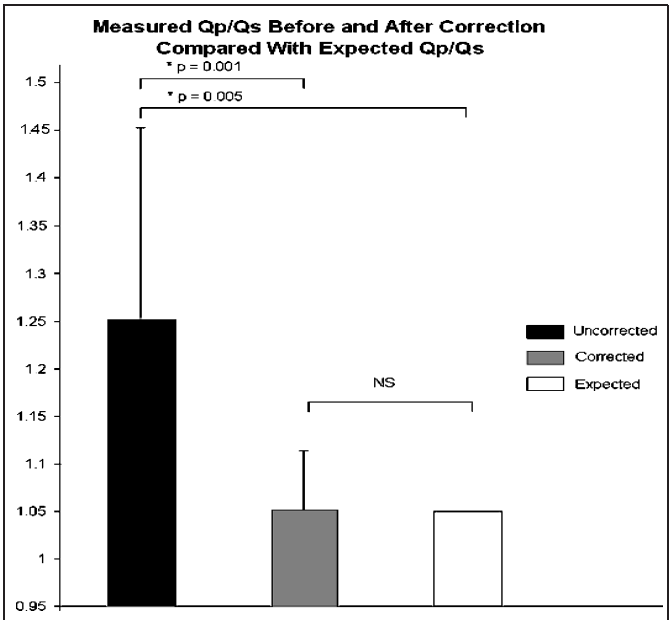

Figure 2. Qp/Qs ratio before and after velocity offset correction.

($p = 0.001$ for size, $p = 0.19$ for anterior position). In other words, the observation that larger patients needed large baseline corrections was due to the fact that they had larger vessels, rather than being further from the magnet isocenter. We could not assess the effect of offset in the superior-inferior direction because the scanner table would move the vessel to isocenter before each scan. Also, there was little variation among the volunteers in the right-left position of the vessels, so the effect of offset in this direction could not be accurately assessed.

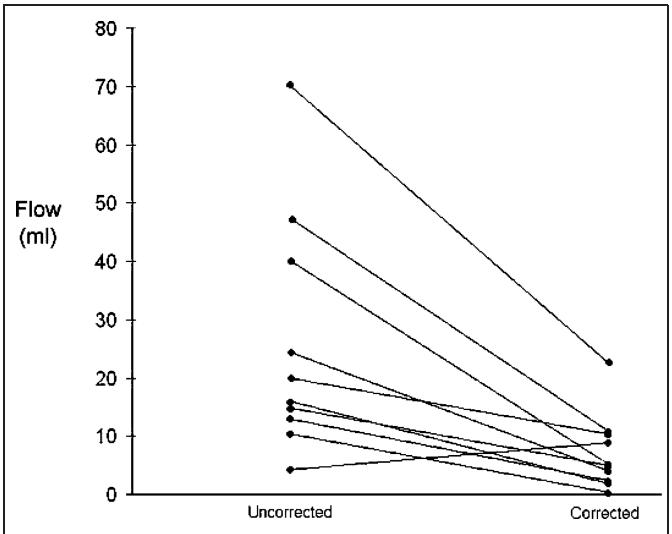

Figure 3. Effect of velocity offset correction on difference between aortic and pulmonary blood flow. In some patients, correction for the velocity offset error leads to a large change in the difference between aortic and pulmonary blood flow.

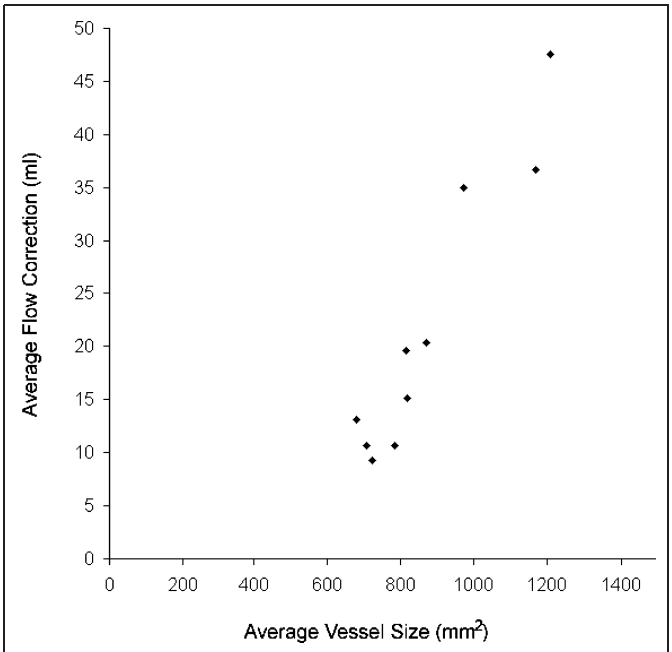

Figure 4. Average baseline flow corrections vs. average vessel size. The data show a strong direct relationship between the cross-sectional area of the vessel and the magnitude of the flow correction ($r^2 = 0.91$).

DISCUSSION

Velocity offset errors have a significant effect on the determination of blood flow in the great vessels. In our study of 10 healthy volunteers, the uncorrected Qp/Qs was 1.25 ± 0.2. In patients with congenital heart disease, this bias could cause a significant overestimation of the severity of a left-to-right shunt (or an underestimation of a right-to-left shunt). Since the decision to correct a shunt is often based on its severity (e.g., when Qp/Qs exceeds 1.5), this error could have a significant impact on clinical decision making.

Our data show a strong correlation between the cross-sectional area of the vessel and the magnitude of the offset error ($r^2 = 0.91$). For example, our normal volunteer with the largest aorta (subject #9) also had the largest uncorrected Qp/Qs (1.6). Because patients with cardiovascular disease often have enlarged vessels due to elevated and/or turbulent flow, they are at increased risk for having large and clinically significant errors in their flow measurements. While children might be expected to have smaller absolute flow errors (because of their smaller caliber vessels), the error may still be substantial when expressed as a percentage of their smaller stroke volume.

We have demonstrated that one can effectively correct for the error in blood flow by assessing the magnitude of the velocity offset in a stationary phantom and then adjusting the flow in the vessel accordingly. While this requires an additional acquisition, the total time involved is generally only a few minutes. It is most conveniently performed after all the patient images are acquired. Correction of the flow data is quick and easy using

the ReportCard software. We have incorporated this correction into the analysis of all of our clinical examinations where it is important to quantify blood flow.

Finally, we found that when flow is measured in the pulmonary artery and aorta (distal to the coronary artery origins), the corrected mean Qp/Qs is 1.05. We believe that the measured Qp/Qs differs from 1 because of coronary artery blood flow, which is $\sim$5% of the total cardiac output. Consequently, we recommend that when Qp/Qs is reported using this methodology the measured values should be divided by 1.05 to reflect coronary artery flow. In other words, if after correcting for velocity offsets, the Qp/Qs is determined to be 1.05, we recommend it be reported as 1.0 to reflect the clinician's understanding that in normal individuals Qp = Qs.

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