Colour Doppler Flow Measurements Using Surface Integration of Velocity Vectors (SIVV): Effect of Colour Flow Gain, Pulse Repetition Frequency and Number of Imaging Planes

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Abstract: Surface Integration of Velocity Vectors is a colour Doppler control volume technique for blood flow measurements. Factors such as colour flow gain (CFG), pulse repetition frequency (PRF) and the number of imaging planes may however affect the quality of measurements. Our aim was to describe how CFG and PRF affect the accuracy of Surface Integration of Velocity Vectors (SIVV) flow in *in vitro* models and to investigate the number of planes required for precise SIVV measurements in an *in vitro* model and *in vivo* at the left ventricular outflow tract. Our results show that the measured SIVV flow varied according to the gain setting while PRF had no significant effect. At least two planes were necessary to obtain <10% measurement error *in vitro*, and four planes were required for <20% measurement error *in vivo*. We conclude that CFG but not PRF had significant effects on the velocity estimate. At least two and preferably >4 imaging planes are required for precise SIVV flow measurements.

Keywords: Flow, doppler, echocardiography, cardiac output.

INTRODUCTION

Surface integration of velocity vectors (SIVV) is a colour Doppler control volume technique for non-invasive volumetric flow measurement [1-5]. Using SIVV, fluid velocity data in a spherical control volume are used, with the centre of the control volume located at the ultrasound transducer (Fig. **1**).

The measured velocity component v, is perpendicular to the control surface s, so is the one that is used for integration of the flow Q through that surface:

$$Q = \iint_S v \bullet dS$$

The size of each surface patch is dependent on the number of scan planes used, with samples from each separately recorded plane contributing to one portion of the spherical surface (Fig. **2**).

Integrating the individual perpendicular velocity components to this surface gives the volume flow, provided that the surface covers the whole flow path.

The technique offers considerable advantages over traditional Doppler flow measurement methods since multiple velocity sampling takes into account uneven blood velocity

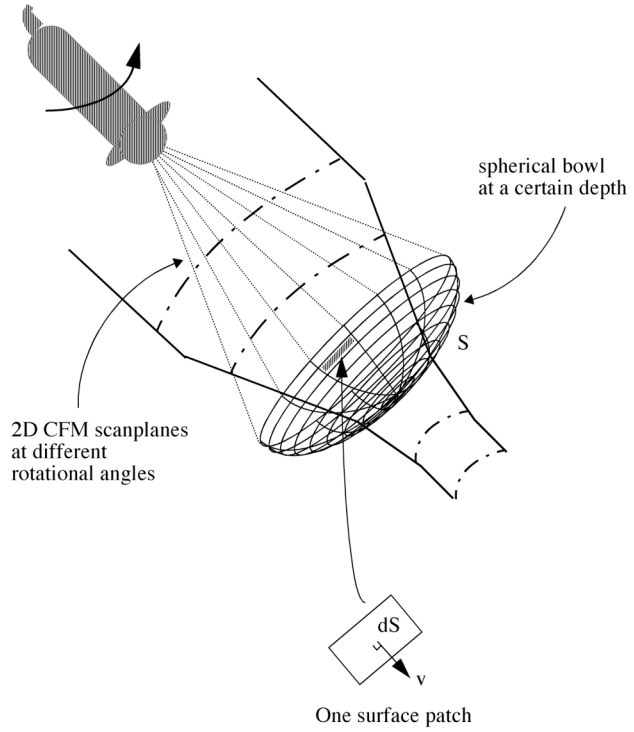

Fig. (1). Diagram showing the spherical control surface covering the flow area. S is the surface at a given depth, which is composed of many surface patches, dS, shown magnified with a corresponding velocity sample, v. The dotted lines correspond to the scanplanes used.

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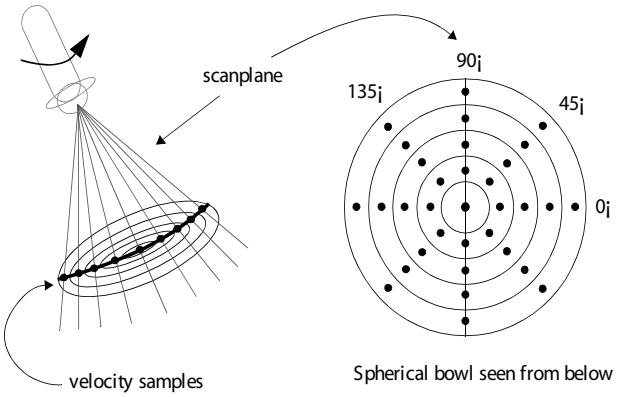

Fig. (2). Drawing illustrating one scan plane obtained with an ultrasound probe (left), and the 4 scan planes that are combined to obtain the spherical surface. Each dot has a corresponding surface patch.

profiles and non-symmetric surface area geometry. Further, an anti-aliasing algorithm has been incorporated into SIVV, allowing sampling of velocities up to 2 times the Nyquist limit [2]. Several studies have shown that methods based on this principle can accurately calculate flow in experimental and clinical settings [1-4, 6-8]. The accuracy and precision of this method may however be affected by scanner settings such as the pulse repetition frequency (PRF), colour flow gain (CFG), the Nyquist limit, frame rate, depth of imaging and the number of imaging planes.

PRF determines the range of velocities measurable by the scanner, thus may have an impact on SIVV flow measurements. This has been described as a limitation of colour Doppler flow measurement, when high PRF and low velocities are combined [9, 10], since the PRF is increased to sample higher velocities preferentially, at the expense of lower velocities. Further, PRF affects the frame rate, thus may be of importance for time resolution in pulsatile flows with rapidly changing profiles, or in the ejection phase of the cardiac cycle.

The effect of CFG on colour Doppler velocity estimates is not known. It is commonly perceived that CFG is a post-processing variable, changing the appearance of the data after it is processed and stored. This should affect the presented image quality only, therefore exerting no influence on the stored velocity estimate. While earlier studies [5, 6, 9, 11] have noted that gain may influence colour Doppler flow measurement, it is not known whether this is due to inconsistencies in defining the surface area of integration or an effect on the velocity estimate itself. We therefore wanted to investigate if gain *per se* could affect velocity measurements.

The number of imaging planes is also important since blood velocity profiles in biological settings are usually non-uniform and variations occur in the area of the cross-section. The greater the number of imaging planes, the more accurate the flow measurement will be, however this is offset by the time it takes to sample the data. Numerical simulations have shown that SIVV is capable of measuring flow through an asymmetric orifice with an error of <10% if at least two, and preferably 4 planes are used [2].

The aim of this study was therefore to test the effects of CFG and PRF on the accuracy of SIVV flow measurements *in vitro*, where the surface area of velocity integration can be kept constant. In doing so, any changes in flow measurement may be attributed to the velocity measurement itself. Further, we wanted to investigate the number of planes necessary for precise SIVV measurements in an *in vitro* pulsatile model and *in vivo* at the left ventricular outflow tract in piglets.

MATERIALS AND METHODOLOGY

In Vitro Models

The *in vitro* setup was an open system consisting of a latex test vessel of 19mm inner diameter which was immersed in a water bath. Blood mimicking fluid flowed into the vessel, controlled by a roller pump (Polystan, Værløse, Denmark) for steady state flow, or a Harvard pump (model 1423, Harvard Apparatus Co Inc, South Natick, MA, USA) for pulsatile flow. Pulsatile flow was produced using a pulse-form with systolic/diastolic time ratio 1: 2, set at 60 beats/min (www.harvardapparatus.com [12]). At the outflow of the test vessel fluid ran into a container, where it could be collected and timed, thus obtaining volume flow. The ultrasound probe was fixed and remained in the same position throughout the study, and immersed into the water bath to ensure acoustic coupling. The blood mimicking fluid used was a mixture of 0.1% corn starch, glycerol and demineralized water, with a viscosity of 4 Ns/m^2 and acoustic velocity of 1540 m/s. All images were obtained at a depth of 10 cm and a Doppler insonation angle of 40° using a commercial echocardiograph (System FiVe®, GE HealthCare, Horten, Norway) coupled to a phased array transthoracic probe (3.5 MHz, HealthCare, Horten, Norway). The centre frequencies for B-mode and colour flow scanning were 3.33 MHz and 3.23 MHz respectively. The colour flow sector varied from 30° to 60° depending on the size of the arc required to provide an enclosing surface for each image, and this was kept constant for each data series to minimize the effects of altered line density on the velocity estimates. 4 scan planes (0º, 45º, 90º, 135º) were used for each SIVV flow calculation, where each scanplane was obtained by rotation of the ultrasound probe from a marked reference point on the model. The estimated accuracy of each plane is less than ±2º. 6 steady state flow rates (1466-5052 ml/min) and 5 pulsatile flow rates (871-3095 ml/min) within physiological range were examined. In the first part of the study, three different gain levels were tested at a constant PRF of 2.5 kHz for steady state flow and 4.5 kHz for pulsatile flow. The images from the different gain levels were defined as (1) optimal - good filling of the vessel with no spillover, adjusted to the nearest multiple of 5 dB, range -25 to -35 dB (2) 5 dB below optimal, with underfilling of the vessel seen in all images and (3) 5 dB over optimal, with blooming effects seen in all images. In the second part of the study, optimal gain was maintained constant for all images while PRF was varied between 2 settings, 2.5 kHz *vs* 3 kHz for steady state and 4.5 kHz *vs* 5.0 kHz for pulsatile flows. The PRFs were chosen as that given as default by the scanner, and default + 0.5 kHz. The Nyquist limits (NL) were 0.301, 0.362, 0.542 and 0.602 m/s and the low velocity reject (LVR) levels were 0.090, 0.108, 0.163 and 0.181 m/s for PRFs of 2.5, 3.0, 4.5 and 5.0 kHz respectively; these were automatically determined by

the scanner for the set PRF. The default frame rates set by the scanner were not altered, these ranged from 11-35 fps and were consistent for the given sector angle, depth and PRF.

To study the effects of the number of imaging planes, we used both *in vitro* and *in vivo* models. The *in vitro* model consisted of a plexiglass cylindrical chamber (diameter 50mm, length 150mm) and a porcine aortic valve (diameter 28 mm) at the outflow tract. A tapered adapter was placed between the chamber and the valve to allow for differences in diameter between the two structures. Sixteen radial inlet orifices (diameter 4mm) ensured uniformly distributed flow into the main chamber from two tube connections. Pulsatile flow (systolic/diastolic time ratio 1: 2) of 0.1% corn starch water was produced by a piston pump (model 1423, Harvard Apparatus Co Inc, South Natick, Ma, USA [12]) set at 60 beats/min. Measurements were made at three flow levels (0.96 2.04 and 2.94 L/min). Time averaged flow rate for steady and pulsatile flows were measured using a measuring glass and stopwatch.

Data was acquired in 12 planes using the transthoracic transducer (3.5 MHz, GE HealthCare, Horten, Norway) with zero angle of insonation and a depth of 6.6 to 10.0 cm. The frequencies for B-mode and colour flow scanning were 5 MHz and 2.85 MHz respectively. The colour flow sector was 25°, and this was kept constant for the whole data series. Two different frame rates, 21.6 and 14.9 frames per second, with corresponding LVRs of 0.06 and 0.04 m/s were tested. SIVV flow was calculated using 1,2,3,4,6 and 12 imaging planes. Packet size was set to 12 and sample volume was fixed at 0.4 mm in all measurements. To increase the velocity dynamic range, the Nyquist limit for velocity measurements was set to allow aliasing in the centerflow resulting in a Nyquist limit of 0.47 m/s and 0.31 cm/s for high and low FPS respectively. Each image was acquired by 38 angle equidistant angle increments in a sector of 25 degrees; 110 depth samples were taken at each angle.

Each SIVV flow measurement was calculated as the instantaneous values acquired for every frame, averaged over the time required for measurement. For pulsatile flow, the average flow rate is the stroke volume multiplied by pulse rate.

In Vivo Model

The study was approved by the Danish Inspectorate for Animal Experimentation under the Danish Minsitry of Justice. Eight piglets (Danish Landrace/Yorkshire) weighing 9-17 kg were studied. After premedication with midazolam 0.1-0.5 mg/kg IM, the animals were sedated with ketamine 5-10 mg/kg IM and cannulated *via* an ear vein. Anaesthesia was induced using ketamine 1-2 mg/kg IV and maintained with fentanyl 40-100 mcg/kg/h. The piglet was intubated with a size 3.5 to 4 endotracheal tube and ventilated with a 2: 1 Nitrous Oxide: Oxygen mixture. Muscular paralysis was maintained with a pancuronium infusion of 0.25-0.75 mg/kg/hr. The pig was then moved into theatre where the right or left internal jugular vein was surgically exposed and cannulated for vascular access and measurement of central pressures. The external carotid artery was also surgically exposed and cannulated for continuous arterial pressure monitoring using the CardioMed system (CardioMed, Olso, Norway). Core temperature *via* a rectal probe and 3-lead ECGs were continuously monitored.

The chest was opened with a midline sternotomy, the pericardium divided and the heart suspended in a pericardial sling. The pericardial cavity was then filled with ultrasound gel and normal saline to maximize acoustic coupling. The pulmonary trunk was isolated and dissected free of surrounding tissue after which a suitably sized ultrasound transit time (TT) probe was applied around the vessel and cardiac output monitored continuously using the CardioMed device. Cardiac output was decreased by inhalation of 1.5-2% Isoflurane, and increased by bolus infusions of Haemaccel (250-500ml), and/or dobutamine (5 mcg/kg/min) in order to obtain a wide range of measurements.

A standoff pad (Kitecko, 3M, Melakoff Cedex, France) was applied to the surface of the exposed heart to provide acoustic coupling for a 5MHz phased array transthoracic probe operated with a System FiVe ultrasound scanner. Epicardial images of the left ventricular outflow tract simulating those obtainable using the transoesophageal or transthoracic apical view were obtained in four planes at a frame rate of 80 fps. Cineloops were downloaded into an external computer and analysed using a custom made program as previously described (4). SIVV flow was calculated using 1,2, and 4 imaging planes, using the same principles as for the *in vitro* model. Three TT measurements were registered simultaneously with the SIVV measurements and averaged to give reference flow values.

STATISTICAL ANALYSIS

Normality was tested for using the Kolmogorov-Smirnov test. X-Y plots of the data were made for each setting. The slopes and intercepts of the groups of data, with SIVV flow as the dependent variable and true flow as the independent variable were compared using multiple linear regression for 'gain' and 'PRF' separately. For 'gain' analysis, the gain setting and the type of flow (steady or pulsatile) were used as explaining factors. For 'PRF' analysis, the PRF setting and the type of flow were used as explaining factors. $p<0.05$ was considered significant. In order to study the effect of the number of imaging planes on the precision of SIVV flow measurements, we calculated the two standard deviation (2SD) of differences between SIVV and reference flow measurements [13].

RESULTS

For all flow settings, acceptable quality images based on $\leq$ 20% drops outs and/or good signal to noise ratio as determined by the operator were obtained. The upper and lower limits of flow were restricted by the pumps.

Fig. (**3**) and Table **1** show an inverse relationship between PRF and the slopes and intercepts of the data, however the difference was not statistically significant. Increasing gain increases the slopes and intercepts of the data significantly, Table **2** and Fig. (**4**). These differences were independent of the type of flow ($p=0.95$ for PRF and $p=0.52$ for gain, data not shown).

The error incurred by SIVV flow compared to true flow increases if fewer planes are used for the flow estimate. This is true for both *in vitro* and *in vivo* data (Table **3**).

Table 1. Regression Equations and Coefficients of Determination for Steady State and Pulsatile Flows when PRF was Varied. The p Value Refers to the Multiple Regression Analysis for Differences in Both Slope and Intercept, Using PRF as the Explaining Factor

Flow Type	PRF	Regression Equation	r^2	p
Steady	2.5 kHz	SIVV=1.10TF-55.22	0.94	0.20
	3.0 kHz	SIVV=0.98TF-106.58	0.95	
Pulsatile	4.5 kHz	SIVV=1.11TF-288.55	0.99	
	5.0 kHz	SIVV=1.04TF-363.77	0.99	

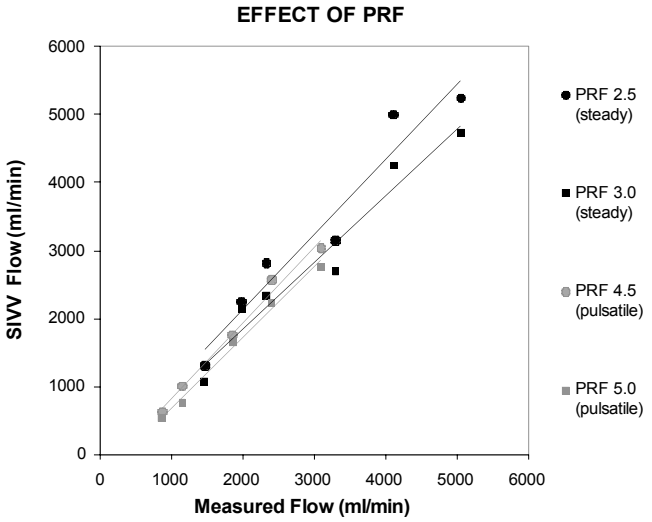

Fig. (3). XY plot of Measured *vs* SIVV flow for each PRF setting. Steady state (black symbols) and pulsatile (grey symbols) flows. See Table **1** for regression equations.

Table 2. Regression Equations and Coefficients of Determination for Steady State and Pulsatile Flows when Gain was Varied. The p Value Refers to the Multiple Regression Analysis for Differences in Both Slope and Intercept, Using Gain as the Explaining Factor

Flow Type	Gain	Regression Equation	r^2	p
Steady	Optimal-5dB	SIVV=1.03TF-457.86	0.94	<0.001
	Optimal	SIVV=1.10TF-55.22	0.94	
	Optimal+5dB	SIVV=1.13TF+397.7	0.93	
Pulsatile	Optimal-5dB	SIVV=0.60TF-19.50	0.94	
	Optimal	SIVV=1.11TF-288.55	0.99	
	Optimal+5dB	SIVV=1.11TF-4.12	0.96	

DISCUSSION

This study demonstrates that in *in vitro* settings, CFG but not PRF significantly influences colour Doppler flow measurements, here evaluated using the SIVV method. If one predefines acceptable measurement error to be less than 10% and 20% for *in vitro* and *in vivo* settings respectively, the optimal number of planes required for accurate SIVV flow calculations are at least 2 for the *in vitro* case, and at least 4 for *in vivo* measurements at the left ventricular outflow tract.

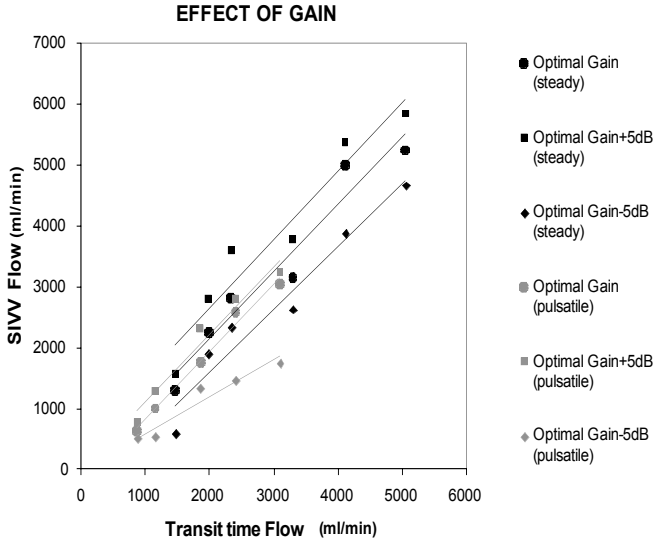

Fig. (4). XY plot of Transit-time *vs* SIVV flow for each gain setting. Steady state (black symbols) and pulsatile (grey symbols) flows. See Table **2** for regression equations.

Table 3. 2SD of the Differences Between SIVV and Measured Flows. *In Vitro* Pulsatile Flow, Using 2 Different Frame Rates and 12,6,4,3,2 and 1 Planes. *In Vivo* Pulsatile Flow Using 4,2 and 1 Planes

		12	6	4	3	2	1
in vitro pulsatile	14.9 fps	5.9%	6.0%	5.6%	7.0%	9.4%	14.0%
in vitro pulsatile	21.6 fps	2.3%	3.3%	6.8%	8.8%	8.2%	14.3%
in vivo pulsatile	14.9 fps	N/A	N/A	17.7%	N/A	23.7%	48.2%

Increases in gain level were associated with increases in the slopes and intercepts of the data. These differences were statistically significant when multiple regression analysis was applied with gain as the explaining factor. The differences were not dependent on the type of flow (steady state or pulsatile). One possible explanation for CFG dependency is that gain may affect the wall filter, causing misinterpretation of velocity data. This is particularly important during low flow conditions, where blood velocities may approximate tissue velocities, especially in areas with good myocardial contractility. Gain also affects lateral resolution, thereby causing variations in the area covered by the colour flow map and thus the surface area of integration. However, since we were able to keep the surface area of integration constant in our study, we can deduce that gain affects the velocity estimate rather than area alone. Although CFG has earlier been noted to be of influence to colour Doppler flow measurement methods [5, 6, 9, 11, 14], our study demonstrates that this is probably due to variations in the velocity estimate

itself, and not due to difficulties in defining the surface area of integration. We believe that this is a significant finding since gain is generally accepted and marketed as a post-processing variable.

There are several possible explanations for this finding. The automatic time gain compensation in the frontend of the scanner contains a non-linear function that assumes certain attenuation, and is tuned to match biological tissues. Measuring in a model with lower attenuation, which gave noisy images, would cause returned signals to be amplified erroneously. This highlights one of the difficulties in measuring *in vitro* where there is a fine line between too little gain (signal and much of the noise filtered out) and too much gain (noise falsely included as signal). This balance can be easily disturbed in the presence of low signal to noise ratios, such as those seen in our study.

Increasing PRF by 0.5 kHz in both setups resulted in a small underestimation of flow when compared to the default setting. This effect was not statistically significant with no differences found in the slopes or the offsets of the data. Ours results are in contrast to earlier studies [9, 15] which identified PRF as a limiting factor for colour Doppler flow measurements particularly with the combination of high PRF and low flow velocities. Indeed, PRF is both a theoretical and practical limitation when measuring in situations where the range of velocities exceed the dynamic range of the scanner, since increasing PRF occurs at the expense of excluding lower flow velocities. The lack of effect of PRF demonstrated in our study may be explained by the use of an anti-aliasing algorithm, since velocities up to 2*NL can be reliably recovered [2-4]. Therefore a (lower) PRF may be chosen without having to sacrifice the lower velocities, effectively increasing dynamic range. Our results provide supporting evidence for the effectiveness of the anti-aliasing algorithm. We suggest therefore that automatic anti-aliasing is used concurrently with a lower PRF in situations where a large dynamic range is required and in situations where lower velocities contribute relatively more to the flow estimate [2].

Increasing the number of imaging planes should theoretically improve the precision of SIVV flow measurements. This is however offset by image acquisition and processing time, which is of practical importance. This is particularly important in haemodynamic instability because flow needs to be relatively stable for the duration of data collection. The results here show that it is possible to substantially increase the accuracy using this approach with a few extra measurements (from 14.3% to 2.3% error using 1 and 12 planes respectively for *in vitro* pulsatile flow, from 48.2% to 17.7% error for 1 and 4 planes respectively for *in vivo* pulsatile flow, Table **3**).

In patients with severe haemodynamic instability and where geometric conditions produce complex velocity profiles, further exploration is needed to be able to find a method for increased accuracy and precision. Newer real-time 3-D approaches with data acquisition using 3-dimensional probes may be a feasible solution.

CONCLUSIONS

Colour flow gain but not pulse repetition frequency affects the accuracy of flow measurements in steady state and pulsatile flows using SIVV *in vitro*. Since the surface area of integration was kept constant, these effects may be attributed to the velocity estimate itself. The optimal number of imaging planes is dependant on the acceptable error and at least 2-4 planes should be used. Increasing the number of planes would increase precision but at the expense of time. Further studies are needed to evaluate the number of planes required at other sites of measurement, and in various clinical settings in order to assess the performance of SIVV and other colour Doppler flow measurement techniques.

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