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ORIGINAL/EXPERIMENTAL PAPER

Surface Integration of Velocity Vectors from 3D Digital Colour Doppler: An Angle Independent Method for Laminar Flow Measurements

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Background: The study was designed to test the angle independence of a dynamic three-dimensional digital colour Doppler method for laminar flow measurement. The technique acquired three-dimensional data by rotational acquisition and used surface integration of Doppler vector velocities and flow areas in time and space for flow computation.

Method: A series of pulsatile flows (peak flow 55–180 ml/s) through a curved tube were studied with reference flow rates obtained using an ultrasonic flow meter. Colour Doppler imaging was performed at three angles to the direction of flow (20°, 30°, 40°), using a multiplane transoesophageal probe controlled by an ATL HDI 5000 system. Integration of digital velocity vectors over a curved three-dimensional surface across the tube for each of the 11 flow rates at each angle was performed off-line to compute peak flow.

Results: Peak flow rates correlated closely (r=0·99) with the flow meter with the mean difference from the reference being $-0{\cdot}8 \pm 2{\cdot}4$ ml/s, $0{\cdot}9 \pm 2{\cdot}6$ ml/s, $1{\cdot}0 \pm 2{\cdot}3$ ml/s for 20°, 30° and 40° respectively. Comparison of the three angle groups showed no significant differences (P=0·15, ANOVA). When sampled obliquely, the flow area on the curved surface increased while the velocities measured decreased.

Conclusion: Surface integration of velocity vectors to compute three-dimensional Doppler flow data is less angle dependent than conventional Doppler methods.
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Key Words: echocardiography; 3 dimensional; flow quantification; Doppler

Introduction

An important application of Doppler imaging is its ability to quantify laminar blood flow. This can be used clinically to measure blood flow within the heart and great vessels, providing potential not only to determine cardiac output[1–3] but also to quantify shunt flow and regurgitant volumes on the basis of flow differences through the individual valves[4].

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Evaluation of blood flow, however, by conventional pulsed Doppler techniques has several limitations. By likening flow in the cardiovascular system to flow through a tube of constant diameter and computing it as the product of the cross sectional area and flow velocity, several assumptions have to be made. These include assumption of a circular flow area shape, a flat velocity profile and an inability to take into account dynamic changes in the flow area during the different phases of the cardiac cycle[5–7]. Also the basic Doppler equation[8],

$$v = \frac{c}{2fo} \times \frac{\Delta f}{\cos\theta}$$

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(where Δf is the difference between the frequencies of emitted and reflected signals, *fo* equals the frequency of the emitted ultrasound signal, *c* equals the velocity of sound in tissue, *v* is the velocity of blood flow and θ equals the angle of incidence between the direction of blood flow and the ultrasound signal) requires that the interrogating ultrasound beam be generally aligned parallel to the direction of flow ($\theta=0°$) or that the angle between the flow vectors and ultrasound beam is determined to correct for the difference between the measured and true velocity. At small angles (less than 20° to 25°) cosθ is often ignored as the underestimation of the velocity is small (6% at 20°)[9]. However, at an angle of incidence of 40° there would be a 23% underestimation and at 60°, a 50% underestimation in flow velocity.

Visual steering is most commonly used to align the Doppler beam parallel to the expected direction of flow and the addition of colour Doppler can assist in doing this more accurately. However, in two-dimensional studies the relationship between the flow vector and the ultrasonic beam in the orthogonal plane cannot be assessed and often the highest velocity is not necessarily recorded from the image that one would assume to be best[9]. In addition to not visualizing the orthogonal plane, the flow profile is not necessarily flat and the highest velocity may not be in the centre of the chamber or vessel, causing the magnitude of the angle θ to be misjudged. Since it may not be possible to align the beam to less than 20° or to measure θ exactly in two dimensional clinical studies, potential sources of error are introduced. A method avoiding the need for angle correction, therefore, theoretically leads to more accurate flow quantification and would have important application in the clinical setting.

Measurement of the velocity vectors over a spherical surface normal to the point of scanning theoretically avoids angle dependence. Several two dimensional studies have used the principle of surface integration of velocity vectors to compute flow rates[10–15], demonstrating its accuracy up to 60° insonation angle[14] and often using multiple planes to approximate a three dimensional approach. Haugen and his team have applied the principle to three dimensional datasets, thereby avoiding the geometrical assumptions inherent with two-dimensional scanning but the datasets were acquired using a freehand approach[16–18]. In this study we used rotational acquisition to acquire a dynamic three-dimensional digital colour Doppler dataset. The method has been previously validated by our group when scanning parallel to the direction of flow both with in vitro and in vivo models[19,20], and computes flow volumes and profiles using surface integration of the velocity vectors over a three-dimensional curved surface normal to the point of scanning. The principle is based on Gauss's theorem, which states that for any arbitrarily shaped control surface system, the flow rate through it is equal to the integral of all the velocity components normal to its surface[21]. With rotational phase array geometry, a surface that is normal to the beam direction results in the shape of a sphere. Since the flow computation uses velocity vectors perpendicular to a spherical sampling surface, no angle correction factor should be required. This study investigated the effects of varying angles of incidence on its accuracy in quantifying peak flow volumes in an in vitro model.

Methods

Flow Model (Fig. 1)

A moderately dynamic compliant latex tube (15 mm diameter, 2 mm thick), curved to alter the parabolic flow profile and deform the circular cross-section, was mounted in a water bath. A solution of water mixed with cornstarch (1% by weight) was used as the fluid medium. The model was connected to a pulsatile flow pump (Harvard Apparatus model 1423, Hallston, MA) generating 11 forward stroke volumes (peak flow rates 55–180 ml/s) at a cycle rate of 60 per min. A magnetic sensor on the pump also generated an electrocardiographic gating signal to trigger the image acquisition. Reference stroke volumes were obtained using an ultrasonic sensor positioned at the inflow of the model. It was connected to a flow meter (Transonic T201, NY) which was interfaced to a chart recorder (Beckman R611, CA). The peak flow rates recorded were used as reference data.

Colour Doppler Echocardiography

Dynamic three-dimensional colour Doppler flow imaging of the tube flow was performed by placing a 4–7 MHz multiplane transoesophageal probe in the water bath to image at angles of incidence of 20°, 30°, and 40° to the direction of flow in the curved portion of the tube. The operator aligned the rotational axis within the flow to achieve good spatial resolution for the flow field. The probe rotation was controlled by an ATL HDI 5000 ultrasound system (Philips Ultrasound, Bothell, WA). Colour gain and filters were adjusted to eliminate random colour in areas without flow. The Nyquist limit was set to avoid aliasing. The three-dimensional datasets were generated by rotational acquisition at 6° increments over 180° excursion, triggered by the gating signal. The acquired raw scanline data including digital velocity information was stored in the HDI 5000 until acquisition was complete. The HDI 5000 was networked to a Silicon Graphics Inc workstation via an Ethernet connection (Silicon Graphics Inc, Mountain View, CA). The pre-scanned data, which had been separated as velocity and a B-mode subset, were transferred to this computer workstation for analysis of flow.

Peak Flow Computation

Following transfer to the computer, flow profiles and volumes were measured using dedicated software,

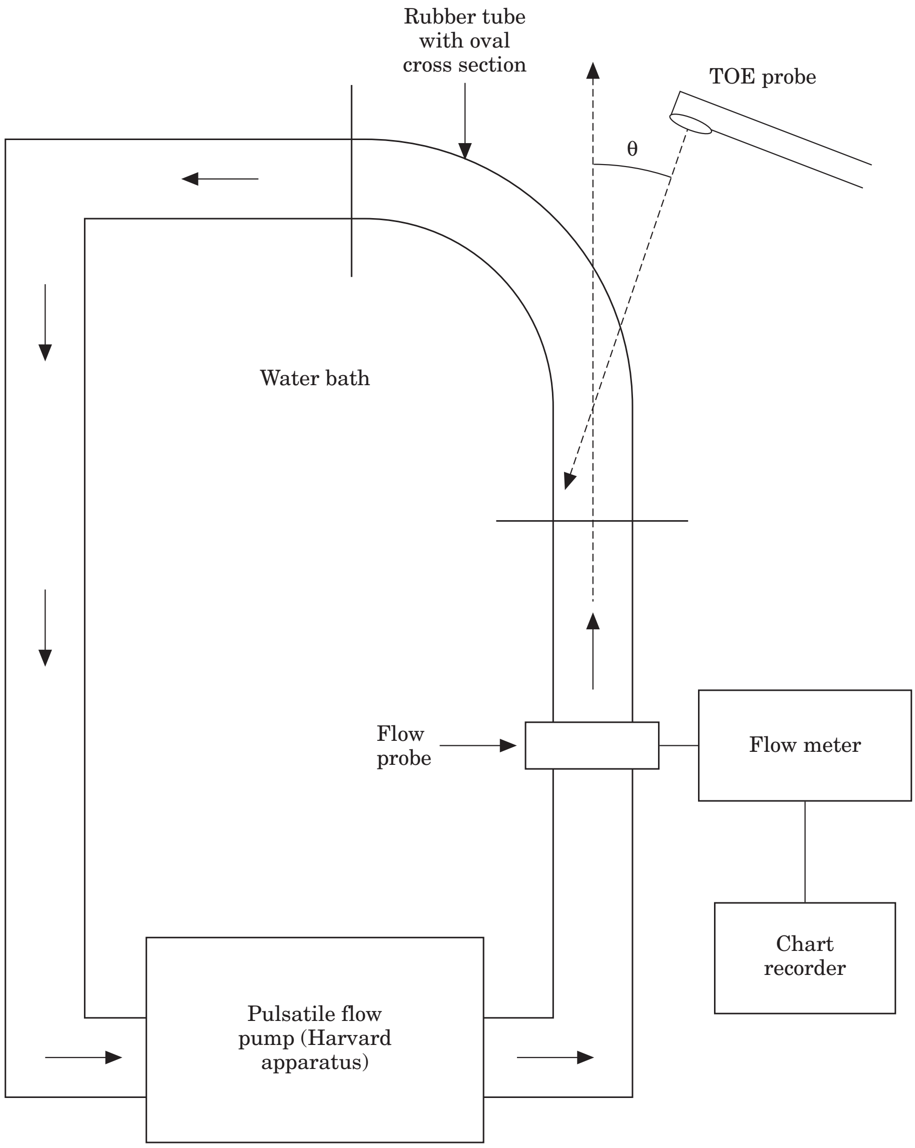

Figure 1. Diagram to illustrate the tube laminar flow model which was mounted in a water bath. The three-dimensional data were acquired by radial acquisition over 180° using a transoesophageal (TOE) probe at varying angles of incidence (=20°, 30° and 40°). Reference data were obtained from an ultrasonic flow meter.

written by our group, to compute flow volumes by applying the principle of Gauss's theorem. On a planar image, selected from the three-dimensional data, the operator placed an arc, representing a spherical sampling surface, across the tube (Fig. 2). The colour Doppler image was then reviewed over one pump cycle to check correct positioning of the sampling plate and it was repositioned if necessary. For each frame of the cycle, the flow fields on the spherical sampling surface, projected on the computer monitor as a planar image similar to a conventional cross-sectional cut, were automatically contour detected frame-by-frame from the dynamic three-dimensional digital velocity data (Fig. 3). This delineated the area inside which the velocity components would be accepted. Any colour signal or noise outside this tracing was excluded in the flow computation. The operator was able to amend the area boundaries if necessary and also select, if required, the duration of the flow event to be measured by selecting the number of frames to be included in the computation. It was also possible, using the dedicated software program, to view the profile of the velocity vectors at any time point (Fig. 4). Following delineation of the flow areas that were accepted, the velocity vectors perpendicular to each segment of the spherical surface were spatially integrated with their associated area segments to give an instantaneous flow rate at each frame. To do this the flow volume was subdivided by the software into a series of regular shapes, in this case triangles. The area of each triangle was then multiplied

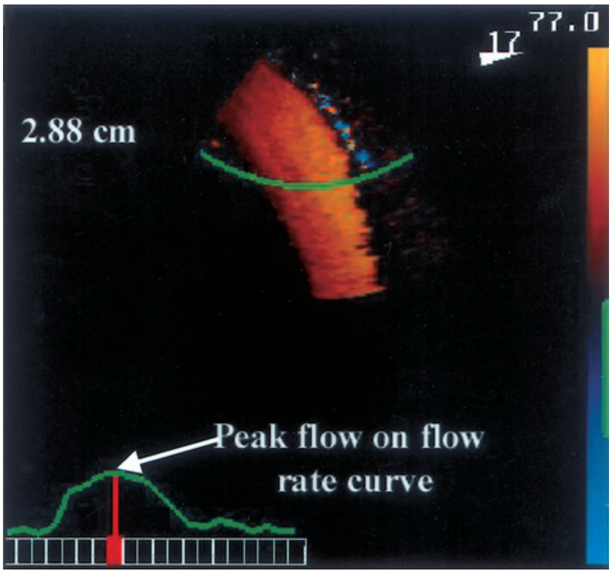

Figure 2. This figure shows the spherical sampling surface, seen here as an arc, positioned across the flow at a depth of 2·88 cm. The three-dimensional derived flow rate curve, over one pump cycle, is displayed at the bottom left.

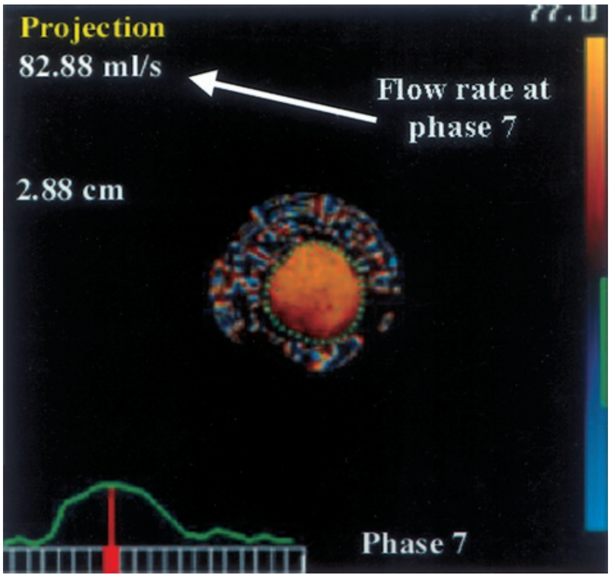

Figure 3. The figure above shows the cross-sectional flow area at the level of the sampling surface during the peak flow phase. The peak flow rate is shown at the top left.

by the area adjusted (weighted) average velocity within it. The instantaneous flow rates calculated in this way were used to compute the flow rate curve from which the peak flow rate was recorded.

Statistical Analysis

The three-dimensional derived peak flow rates are given as the average of three measurements. The correlation

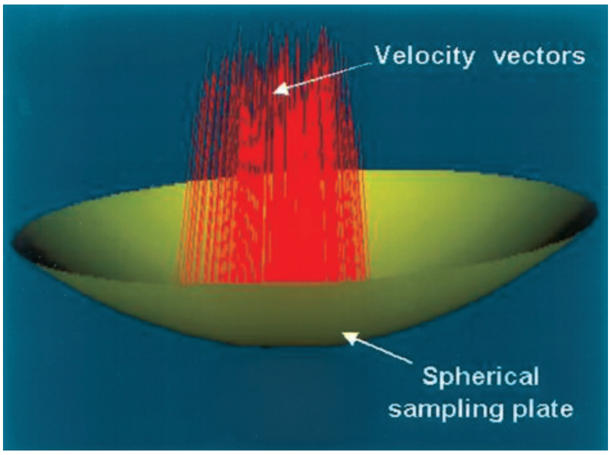

Figure 4. The sampling surface and velocity vectors in three-dimensional space at the phase of peak flow shown in Fig. 3.

between the peak flow rate computed by the three-dimensional method and that from the ultrasonic flow meter, for each angle of incidence, was determined by simple regression analysis. Limits of agreement (mean ± SD) between the three-dimensional and reference data were calculated using the method described by Bland and Altman[22]. The three angle groups were compared with each other using analysis of variance. A P value less than 0·05 was regarded as statistically significant.

Results

The three-dimensional colour Doppler peak flow rates computed for all angles of incidence are given in Table 1. These correlated well with the reference data as shown in Table 2 ($r=0{\cdot}99$ in all cases). Comparison of the three angle groups showed no significant differences between them ($P=0{\cdot}15$). Analysis of agreement between the two methods was also good as illustrated by the Bland Altman plots in Fig. 5 (angle 20° mean difference from reference $-0{\cdot}8 \pm 2{\cdot}4$ ml/s, angle 30° mean difference $0{\cdot}9 \pm 2{\cdot}6$ ml/s and angle 40° $1{\cdot}0 \pm 2{\cdot}3$ ml/s).

Discussion

This study demonstrated that peak flow, in a tube model mimicking a great vessel, was accurately determined using the dynamic three-dimensional digital colour Doppler method at varying angles of incidence, measured up to 40°.

Limitations of Conventional Doppler Techniques

The curved tube model that we used in this study was designed to mimic flow, for example, in the left or right

Table 1. Three-dimensional derived peak flow rates with corresponding reference (ultrasonic flowmeter) data for each angle of incidence.

θ=20°		θ=30°		θ=40°	
Reference PF ml/s	3D PF ml/s	Reference PF ml/s	3D PF ml/s	Reference PF ml/s	3D PF ml/s
56·7	58·6	60·0	61·9	56·7	57·2
67·7	69·0	76·7	77·4	67·7	69·7
84·2	84·3	90·0	92·6	84·2	86·9
95·2	93·0	95·0	97·1	95·2	94·2
100·7	100·8	101·7	99·6	100·7	104·4
111·7	106·6	118·3	120·4	111·7	108·6
117·2	117·6	130·0	132·6	117·2	116·8
128·2	127·6	136·7	135·9	128·2	132·4
139·2	139·6	155·0	151·6	139·2	137·3
150·0	149·8	163·3	168·8	150·0	151·5
166·7	161·2	178·0	176·9	166·7	168·8

Pump rate set at 60 per min throughout. PF, peak flow.

ventricular outflow tracts or the great vessels arising from them. As in physiological situations, the tube size, shape and location varied throughout the pump cycle. Conventional pulsed Doppler techniques assume the outflow tracts and great vessels to have a circular flow area shape for flow computation. Although it is probably reasonable to assume this for the left side, the pulmonary artery is not necessarily circular, particularly as it courses above the aorta[23]. Similarly the vessels are not rigid tubes but change size and shape during the cardiac cycle. The aortic area depends on intra-aortic pressure and may increase up to 11% during systole[24,25]. The pulmonary artery cross-sectional area may increase by even further, factors which the conventional techniques cannot take into consideration. Assumptions are also made that the flow profile is flat, when in reality it may be somewhat parabolic or asymmetric. Alterations in vessel geometry such as curvature of the aorta and asymmetry of the outflow tract may cause the location of peak velocity to shift from the centre of flow and therefore not be accurately sampled using the pulsed Doppler method[23]. Our technique, however, allows temporal characterization of flow and full spatial characterization of the flow area and velocity profile.

Table 2. Results of linear regression analysis between three-dimensional derived peak flow rates at varying angles of incidence and ultrasonic recordings, used as reference*.

Angle of incidence (θ)	Equation of linear regression	Correlation co-efficient	SEE (ml/s)	*P* value
20°	y=0·96x+3·44	0·99	2·2	<0·0001
30°	y=0·99x+2·24	0·99	2·7	<0·0001
40°	y=0·99x+1·11	0·99	2·5	<0·0001

SEE, Standard error of estimate.
*These show good correlation of the three-dimensional peak flow rates with the reference data at each angle.

The pulsed Doppler technique is also dependent on the angle correction factor. Without angle correction, the 20°, 30° and 40° angles of interrogation used in this study would be expected to result in a 6%, 13% and 23% under-estimation in peak velocity respectively. Alternatively, with angle correction, a 15° error in the visual angle correction in this range would still cause a 10% to 20% error in the velocity measurements.

There are several reasons why compensating for the angle between the ultrasound beam and flow may in fact not improve the accuracy of velocity measurement or

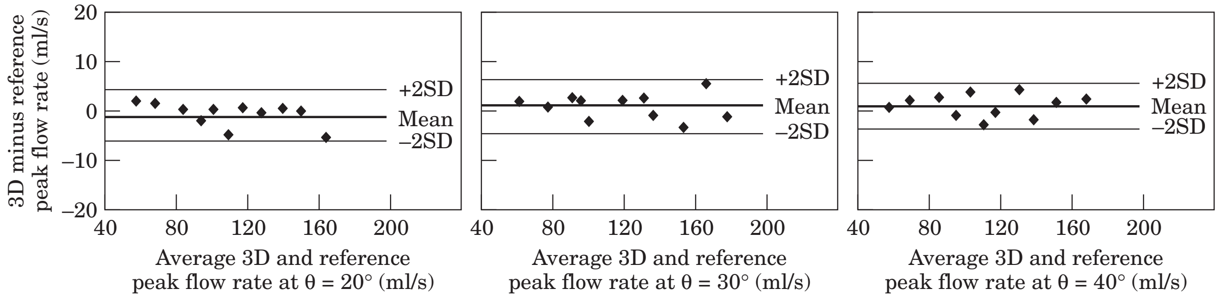

Figure 5. The Bland–Altman plots show good limits of agreement between the three-dimensional and reference data for each angle of incidence.

make it worse[26,27]. The third dimension cannot be seen using two-dimensional echocardiography and, to an extent, the direction of flow can only be assumed from the image. Flow may not be parallel to the chamber or vessel's walls and although this is more likely to be an issue in the presence of valvular stenosis, which was not addressed by this study looking at laminar flow, the image can be misleading even in cases of normal flow. Secondly, due to the fundamentals of the cosine function, an angle error when performing angle correction will produce a greater error in the velocity estimation than a similar error at 0°. For example, an angle error of 10 or 20° at an estimated angle of incidence of 30° produces a greater error in velocity than the same angle error at 0°[26]. It is often therefore recommended to attempt alignment when using conventional techniques instead of angle correction but this can sometimes prove difficult to achieve in the clinical setting. Thirdly, in the case of high velocity jets, diversion of the jet may cause the angle between the ultrasound beam and the jet to be in fact less than that suggested by the image, resulting in over estimation of the flow velocity[26].

Other Digital Colour Doppler Techniques

In view of the limitations of the pulsed wave Doppler technique, digital colour Doppler methods were subsequently developed to quantify laminar flow. One of the two-dimensional techniques, the automated cardiac output measurement method, involves positioning a rectangular region of interest in the flow while scanning parallel to the direction of flow[28–31]. The velocities within the selected region are integrated on a frame to frame basis to compute spatial and temporal flow profiles and generate flow volume data. It is less angle dependent than the conventional pulsed Doppler technique because if the Doppler beam is misaligned to flow by an angle θ, then the under-estimation of the Doppler velocity will be offset by the increase in the sampling area[30]. However, angle correction may be needed for locations away from the centre of the sample window, especially for sample windows close to the transducer or vessels with larger cross-sectional areas. Sun and his team found that for angles $\leq 30°$ measured flows averaged within 5% of true flow. With increasing angle, however, flow under-estimation occurred, reaching 30% for 50° misalignment. This method does have the advantages of taking into account the dynamically changing flow area by temporal integration frame by frame and using all the velocity information across the flow diameter. However, it is a two-dimensional technique and still makes assumptions regarding a circular shaped flow area and an axial symmetry of the spatial flow profile[30,32,33]. From a practical standpoint, it is also often difficult to maintain the axis throughout the cardiac cycle with the associated cardiac motion when using a two-dimensional technique.

Most of the studies using surface integration of velocity vectors over a surface normal to the point of scanning have been based on two-dimensional techniques. Three-dimensional colour Doppler echocardiography avoids the need for geometrical assumptions regarding laminar flow. In our method it is used for flow quantification in the setting of rotationally acquired, computer-generated three-dimensional digital colour Doppler data sets. Others have generated a three-dimensional Doppler dataset by a careful freehand fanlike sweep, which is essentially similar to a rotational acquisition but may give irregular plane spacing and risk confounding velocity measurement errors due to probe motion[16,17]. In this study we directly tested angle independence of our technique, looking at flow calculations at varying angles of interrogation.

Non-invasive acquisition of three-dimensional flow data may also be achieved using magnetic resonance imaging and the accuracy of phase velocity encoded magnetic resonance imaging for flow quantification has been demonstrated[34]. Flow measurement by this method is angle independent and direct velocity profiles can be used. However, the high cost, long acquisition times and non-portability of the equipment limit the extent of its use in clinical practice.

Theory of Reduced Angle Dependency

The flow computation method used in this study measures velocity vectors sampled over a spherical control surface. The principle for this angle-independent three-dimensional flow quantification method allows the use of any shaped surface as long as the velocity vectors that are normal to it are measurable[35]. The use of a spherical surface can be applied when Doppler velocities are measured with a radial symmetrical scan geometry such as the rotational acquisition used in this study, with the probe's pivotal centre fixed throughout the imaging.

One way to understand the angle independence is to consider a small finite area of the flow field where fluid is flowing into it with an intersecting angle. If the intersecting angle is varied then the measured velocity would decrease by the cosine of the change in angle and the finite area would increase by the cosine of the same angle, so that the effect would be negated. This is shown diagrammatically for the spherical surface as a whole in Fig. 6.

Limitations of Method

Instrument-related and technical factors may affect the flow measurements for any colour Doppler method. Our method requires a high quality colour Doppler image and, although not formally tested in this study, its accuracy would be expected to be dependent on a satisfactory signal gain. We adjusted colour Doppler gain so that the colour filled the region of interest but was not seen in the adjacent walls. It has been previously shown using the automated cardiac output measurement

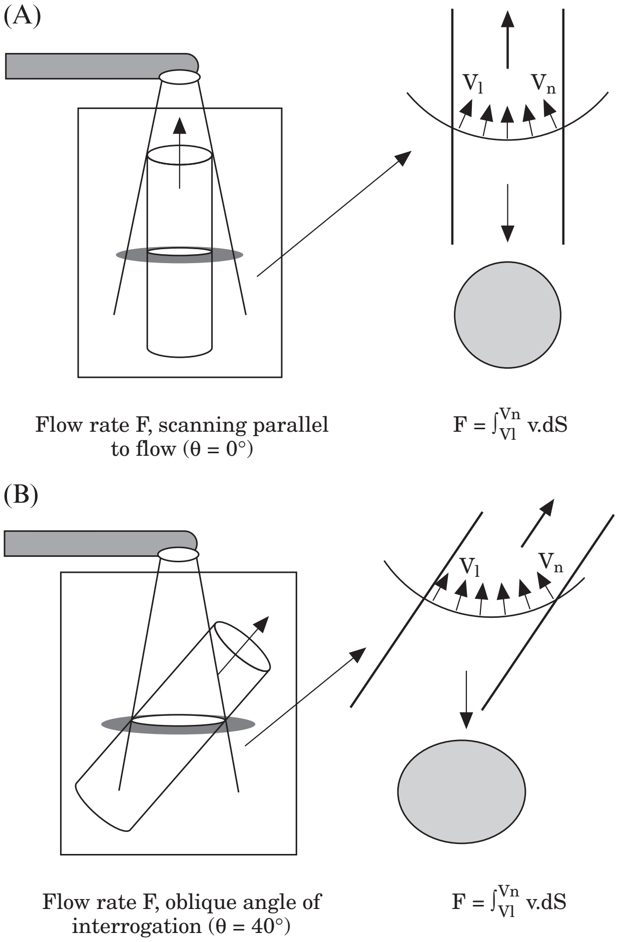

Figure 6. (A) To compute instantaneous flow rates the velocity vectors perpendicular to each segment of the spherical surface were integrated with their associated area segments to give an instantaneous flow rate at each frame. (B) The principle of flow rate calculation at an oblique angle was similar to that shown in (A). However, the decrease in the measured velocities due to the angle between the ultrasound beam and velocity vectors was offset by the increase in size of the area segments.

method that if colour gain and wall filters are selected to provide an 'acceptable' colour image quality, then within this range changes do not produce significant differences in the results[36]. Its reliance on a high quality colour image can be partly alleviated by increasing the spatial density of the image acquisition but it could still sometimes limit its application in the clinical setting. As regards lateral resolution issues, flow signals may spread outward but velocities near the boundary may be suppressed by tissue flow priority filtering (these velocities are low in magnitude and have been reported to be less problematic with laminar, parabolic shaped flow[13]). The two effects probably balance; significant over- or under-estimation of flow volumes did not occur in this study.

For accurate flow quantification the entire flow cross-section requires to be covered during the three-dimensional acquisition. For the best coverage the rotational centre should ideally be in the middle of the flow, that is, prior to acquisition the image needs to be centred as much as possible on the monitor of the scanner so that flow will be seen at each angle of the acquisition. In practice, a test rotation prior to acquisition is usually performed to ensure that the entire cross-section is covered and usually the sector width limits the amount of the oblique to cover the cross-section by colour Doppler imaging. This could be a problem at oblique angles of interrogation and means the technique is not entirely angle independent. Furthermore, colour Doppler measurement will deteriorate when the angle between the flow stream and ultrasound beam across the image plane approaches 90° and will be absent at an angle of 90°[27]. The calculation method used in this study could not be applied at angles close to 90° as the velocity vectors could not be accurately measured.

The flow computation program is for the quantification of laminar flow. To overcome any aliasing (when actual velocity exceeded the Nyquist limit), an aliasing correction algorithm could be implemented permitting accurate measurement of velocities up to twice the Nyquist limit. However, the technique would still not be able to be applied to cases of severe valvular stenosis, when divergent and eccentric jets are likely, and therefore the limitations of angle correction using the conventional pulsed Doppler technique substantial. During the computations of this in vitro study aliasing correction was not an issue.

Conclusion

This study tested the theory that angle dependency was minimized in the described dynamic three-dimensional digital colour Doppler method. At the angles of interrogation used (20°, 30° and 40°) peak flow rates were accurately determined. This technique is therefore less angle dependent than conventional methods and has potential application in the clinical setting for accurate determination of laminar flow rates.

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