EXPERIMENTAL STUDIES

Direct Computation of Multiple 3D Flow Convergence Isovelocity Surfaces from Digital 3D Reconstruction of Colour Doppler Data of the Flow Convergence Region: an in vitro Study with Differently Shaped Orifices

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Aims: Evaluation of the accuracy of direct computation of multiple three-dimensional (3D) flow convergence (FC) isovelocities by using digital reconstruction of colour Doppler data of the flow convergence region.

Methods and Results: We used a conventional ultrasound system (ATL HDI 3000) connected to a computer workstation via Ethernet link. The digital 3D datasets were directly transferred to a Silicon Graphic Imaging Octane workstation for later measurement. We generated nine pulsatile flows (20–60 ml/beat), with peak flow rates (67–180 ml/s), through three orifices (circular, rectangular and triangular, S=0·24 cm²). The 3D reconstructions of FC surface areas from multi-threshold velocities, including aliasing velocities were analysed to quantify the peak flow rate. For all orifices, linear regression demonstrated excellent correlation between the 3D calculated and electromagnetic flow meter recorded data. While there was a high correlation for 3D computation of flow rate from the single best Nyquist of 24 cm/s (r=0·97–0·98, SEE=7·75–12·58 ml/s), the ability to average three threshold velocities (15, 18 and 24 cm/s) yielded an improved correlation (r=0·98–0·99, SEE=5·70–7·73 ml/s).

Conclusions: Direct computation of multiple 3D FC isovelocities from digital reconstruction of colour Doppler data of the FC region provides the potential to accurately quantify the complex asymmetric spatial flow events at any selected velocity.

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Key Words: three-dimensional; digital; Doppler; flow convergence.

Introduction

Although non-invasive colour Doppler echocardiography is well established in the diagnosis and evaluating of valvular heart disease, we still lack a reliable echocardiographic method for accurately quantifying the severity of valvular regurgitation. Jet visualization colour Doppler methods have been widely used for evaluating the severity of regurgitation; however, they are at best only semi-quantitative due to variability introduced by both instrument settings and jet eccentricity[1–5]. Doppler laminar flow acceleration phenomena and two-dimensional (2D) flow convergence methods have been widely studied for evaluating the
severity of valvular regurgitation and stenosis. As flow approaches an orifice, it forms concentric isovelocity shells of decreasing surface area and increasing velocity. By conservation of mass, flow through any shell must pass through the orifice. Colour Doppler flow mapping can depict such shells based on the selected velocity or provide a range of isovelocity contours as a digital map. The regurgitant flow rate through the orifice is calculated as a product of surface area multiplied by the aliasing velocity which defines the surface. This approach appears to be more rational in that it eliminates the measurement of eccentric jets and is based on the organized flow driven toward the orifice, with observation less dependent on instrumentation related factors. Nonetheless, 2D flow convergence methods rely on the assumption of a hemispheric geometric shape of flow convergence region and thus need strictly selected aliasing velocities.

To date, direct flow convergence measurement from three-dimensional (3D) based datasets has been the subject of both in vivo and in vitro experimental studies. It has been demonstrated that a 3D method could probably eliminate the limitations imposed by the shape assumptions of the 2D method. However, the 3D images in these studies were derived from 2D images with a relatively convoluted colour technique that involved shifting the base line and changing the wall filter and colour gain settings to emphasize only the FCR and jet areas. Then, this flow information was transferred as video composite data and computed with a TomTec system as grey-scale 3D objects. For our current study, the new computer method we have developed provides 3D digital velocity datasets of an object occupying the spatial location of any chosen velocity threshold. Thus, multiple direct 3D surface area measurements can be made within the flow convergence zone in post processing.

The aim of this in vitro study was to test the applicability and potential value of direct computation of multiple flow convergence isovelocity surface areas by measuring different velocity thresholds of the flow acceleration region in a digitized 3D base.

**Methods**

**Flow Model**

A cylindrical two-chamber pulsatile flow model characterized by a proximal and a distal chamber (both: height=8 cm, length=24 cm) was used in this study; it has been previously described in detail. Three different orifices (circular, rectangular and triangular) with orifice areas of 0.24 cm² were mounted between the inlet and outlet chambers to produce the convergent flow toward the orifices. The model was connected to a pulsatile piston flow pump (Harvard Model 1423 Apparatus, Andover, Massachusetts) generating nine forward stroke volumes (20–60 ml/beat), peak flow rates 67–180 ml/s) at a heart rate of 60 beat/min. Each flow was driven from the top chamber through the orifice into the bottom chamber and back out to a water bath and recirculating pump. A solution of cornstarch mixed with water (1% by weight) was used as the fluid medium. Flow rate was measured by using ultrasonic flow probe (Model 16 NB272, Transonic Systems Inc.) positioned at the inflow of the model and connected with electromagnetic flow meters (Model T 106X, Transonic System Inc.) and a physiological recorder (Beckman R611, Beckman Instruments Inc.) for simultaneous recording of flow wave forms. The regurgitant volume was additionally cross-checked by measuring pump outflow collected in a graduated cylinder and timed by a stopwatch.

**Colour Doppler Echocardiography**

The flow convergence zone proximal to the orifice was imaged by using an ATL HDI 3000 ultrasound system controlling a 7.0 MHz multiplane transoesophageal probe (MPT 7–4). The probe was placed above the orifice scanning parallel to the flow. Echo scanning was accomplished at pulse repetition frequencies (PRF) of 3.0–6.0 KHz. Colour gain and wall filters were adjusted to eliminate random colour in areas without flow. System Nyquist limit was 45 cm/s, but was zero shifted to produce an aliasing velocity of 24 cm/s to define the clearest flow convergence region in the 3D data set. In addition, pulsed wave Doppler (PW) and CW Doppler interrogations were performed in the centre of the flow to check the timing of peak velocities for each pump setting.

**Digital Velocity Data and Data Transfer**

The HDI 3000 was connected to an NT workstation by a serial cable for running a program to control the TEE probe rotation during the data acquisition. It was also connected to a Silicon Graphics Inc (SGI) workstation by an Ethernet cable to allow transfer of (rθ) sector scan line data containing digital velocity information to the SGI workstation. The TEE probe was set up to step at 6°–9° increments, colour Doppler imaging was set up to display one frame per cardiac cycle and triggered to peak systolic flow using ECG gating. After the scan head had completed the 180° rotation for imaging the flow convergence region and the proximal jet area, each frame (total ± 30 frames) of the rotation was saved in the HDI’s cine loop memory and then transferred to SGI workstation. The raw velocity data, in a pre-scan converted format including all the step increments, start and end angles, were transferred to and saved onto the hard drive of the SGI computer for post-imaging processing. The digital colour 3D data could then be reconstructed using custom software. The time for the rotational 3D
data acquisition was about 20–30 s when the heart rate was 60 beat/min; total time including data transfer was approximately 60–120 s.

Flow Measurement

The flow convergence zone was clearly isolated as a different colour layer in 3D display (Fig. 1), and we could identify the individual sub-velocities towards the orifice. After digital reconstruction, these 3D velocity surfaces or surfaces at any intermediate velocity could be measured separately, since our digitized colour Doppler flow mapping method identified the surfaces numerically as a velocity threshold in 3D space. As such, the customized program used by the SGI computer could recognize any spatial location of the chosen velocity threshold. By using this unique program, we could select different velocity thresholds to measure the proximal isovelocity surface area (PISA) in the range. In this study, we chose three sequential velocity thresholds (15, 18, 24 cm/s) for each orifice at each stage. After selecting a velocity threshold, the 3D reconstructed flow convergence region (FCR) and jet area was input in a 3D space box which had already been subdivided into 200 slices evenly (Fig. 2). By using the AVS program (Advanced Visual Systems Inc, Waltham, MA, U.S.A.) we could view these images slice by slice until we determined the exact spatial location of the FCR and then initiate the measurement. For this, as seen in Figure 3, we input four to seven points to trace the selected FCR slices. Usually, 20–50 slices would be traced for each threshold velocity and the program would automatically add each measured FCR arc length sequentially and multiply by the slice thickness (also determined by the program) to obtain the actual isovelocity surface area. Finally, the peak flow rates were calculated as the product of isovelocity surface area and the threshold velocity (Q=S · V).

Interobserver Variability

To evaluate the effect of interobserver variability on the computation of multiple 3D PISA from digital

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**Figure 1.** Three-dimensional flow convergence surface areas from rectangular orifice. The different colour layers indicate the various isovelocity surfaces in 3D space accelerated toward the orifice. The yellow layer is the first aliasing border 24 cm/s; the other arrows indicate velocity thresholds of 15 cm/s and 18 cm/s.

**Figure 2.** Three-dimensional schema for direct measurement of the flow convergence region (FCR). The parallel cutting planes of the 3D reconstructed peak flow convergence surface and jet areas are shown. The program allows up to 200 cutting planes within one 3D frame.

**Figure 3.** Circular orifice, threshold 15 cm/s. An example of 3D flow convergence measurement from one cutting slice. The white dots along with the flow convergence region (FCR) are the tracing points.
reconstruction of colour Doppler data of the flow convergence region, 10 randomly selected flow conditions were analysed at different times with the same computer settings by two independent observers, each without knowledge of the results obtained by the other reader or the reference values.

**Statistical Analysis**

Data are presented as mean value ± SD. The correlation between continuous variable data was determined by linear regression analysis. Simple regression analysis was used to examine the relation between the quantified peak regurgitant flow rate and the electromagnetic flow metre-derived data. The agreement between the actual peak flow rate and 3D calculated flow rate using the digital reconstructed flow convergence method was tested according to the method of Bland and Altman. The difference of flow rate between three orifice groups were tested by a non-parametric ANOVA (Kruskal–Wallis) method. Statistical significance was defined as P<0.05.

**Results**

We first examined the results of our method to predict flow rates from a selected aliasing velocity (24 cm/s) when compared to the results obtained with averaging of three sequential isovelocity surface areas (15, 18, and 24 cm/s) within the proximal flow convergence region. For all of the nine flow conditions (peak flow rates range 67–180 ml/s) of each orifice, the linear regression shows that the calculated peak flow rates demonstrated excellent correlation with the reference data. While there was a high correlation for 3D computation of peak flow rate from the single best Nyquist limit, 24 cm/s (Circular: r=0.98, SEE=7.75 ml/s, P=0.0001; rectangular: r=0.97, SEE=12.58 ml/s, P=0.0001; triangular: r=0.97, SEE=9.82 ml/s, P=0.0001), the peak flow rate for the method which averaged three different isovelocity surface areas (three threshold velocities 15, 18 and 24 cm/s) yielded an even better correlation (circular: r=0.99, SEE=5.70 ml/s, P=0.0001; rectangular: r=0.99, SEE=7.73 ml/s, P=0.0001; triangular: r=0.98, SEE=6.98 ml/s, P=0.0001) (Fig. 4, A1, B1 and C1).

As detailed in Table 1 and the Bland-Altman plots (Fig. 4, A2, B2, C2), calculated peak flow rate from the single best Nyquist limit (24 cm/s) underestimated actual flow rate by mean difference = -11.6 ± 13.5 ml/s, percentage difference = -11.92% ± 11.84% for circular orifice; mean difference = -0.2 ± 13.31 ml/s, percentage difference = -0.44% ± 10.05% for rectangular orifice; mean difference = -11.57 ± 9.53 ml/s, percentage difference = -9.98% ± 7.66% for triangular orifice.

In contrast, the average of three velocities underestimated peak flow by mean difference = -7.99 ± 9.46 ml/s and percentage difference = -8.58% ± 8.86% for circular orifice; mean difference= -0.44 ± 10.05 ml/s, percentage difference= -1.99% ± 7.72% for rectangular orifice; mean difference= -6.38 ± 6.93 ml/s, percentage difference= -6.37% ± 7.54% for triangular orifice (Table 1). The Bland–Altman plots also demonstrated very good agreement between the three average threshold calculations and the true peak flow rate for each orifice (Fig. 4, A2, B2, C2). Using the Kruskal–Wallis test, there were no significant differences between the results for individual orifices (P=0.86 for averaged flow rate and P=0.76 for single aliasing derived flow rate.

**Interobserver Variability**

There was excellent agreement between the two independent observer measurements for the 3D calculated regurgitant peak flow rate using the digital reconstructed multiple flow convergence method (r=0.99, mean difference=1.13 ± 1.45 ml/s).

**Discussion**

Direct computation of multiple 3D flow convergence isovelocity from digital reconstruction of colour Doppler data of the flow convergence region provides the capability of accurately quantifying asymmetric spatial flow events at any selected velocity. The method not only eliminates any assumption about the shape of flow convergence, such as is required for 2D methods, but also eliminates the loss of colour Doppler information that occurs during conversion from the raw scanline data.

**Previous 2D and 3D Flow Convergence Methods**

Two-dimensional flow convergence surface area methods have been used experimentally and clinically for quantifying the severity of the valvular regurgitation[6–10,15–18,24–29]. The earliest reported work relied on geometric assumptions of the isovelocity surface shape, such as a hemisphere (single-axis method) which measured only radii from the assumed centre of the orifice to the isovelocity surface. However, the shape of the flow convergence surface depends on many factors, such as frame rate, geometry of the regurgitant valve and other physiological conditions, and this produces problems with defining flow convergence from a single radius. Additionally, it is also sometimes difficult to locate the centre of the orifice, a step which our new method avoids. Single radius flow convergence methods have been demonstrated to yield under- or over-estimation of actual flow[8,10,12,21]. A 3D-based hemielliptic method (three-axis method) was suggested, and has demonstrated improved results for flow rate compared...
Figure 4. Linear regression of circular orifice (A1), rectangular orifice (B1) and triangular orifice (C1), showing actual peak flow rates (abscissa) and calculated peak flow rates (ordinate) obtained by the 3D multi-threshold method. Graphs compared the average threshold flow rates with single best threshold (aliasing velocity 24 cm/s) flow rates. Agreement between the actual peak flow rates and calculated peak flow rates (average of three different threshold velocities) was tested according to Bland and Altman’s method (A2, B2, and C2 for circular, rectangular and triangular perspective).
to a 2D simple hemispheric (one-axis) method in a report by Shiota et al.\textsuperscript{10} The percentage difference between the actual flow rate and 3D method is $22 \pm 6.8\%$ vs. $35 \pm 12\%$ for the 2D method. That method did not actually present 3D images, but measured cut planes from a 3D acquisition of composite video signal of echo images containing colour Doppler flows and a custom map.

Three-dimensional methods for direct measurement of flow convergence appear to yield more reliable and more accurate flow data, since they do not require any geometric assumptions of flow convergence shape. Also, such methods could provide the whole image from any desired angle or cutting plane, including FCR and jet area. However, those early 3D methods required the transfer of the colour data into a black and white composite video milieu for 3D surface rendering (TomTec). During data transfer, the colour Doppler information was subject to distortion and data loss. Our direct scan line r\textsuperscript{2} transfer of data containing raw digital velocities to the computer overcomes this, but nonetheless the colour Doppler angle dependence remains, and will also affect the presentation of the flow data yielding a tendency for under estimation.

**Advantages of Current Study**

Our newly developed 3D digital multi-aliasing velocity surface area reconstruction method overcomes some of the limitations encountered in previous work. It does not require any geometric assumption and allows us to choose any desired isovelocity surface (velocity threshold) within the range of the flow acceleration towards the regurgitant orifice.

Previous work from our lab has demonstrated that 3D FC area measurement derived flow rate was superior to 2D methods in *in vitro* and *in vivo* studies even when directly measured at a single aliasing velocity\textsuperscript{11,12}. The percentage difference between actual flow rates and the two methods are $0\% \pm 7\%$ for 3D vs. $23\% \pm 8\%$ for 2D\textsuperscript{13}. In contrast, in our current study using digital 3D we have been able to measure almost any well defined isovelocity surface in addition to the ones we selected (threshold $15$ cm/s, $18$ cm/s, and $24$ cm/s) to obtain the peak flow rate at nine flow rates for three orifices. These results demonstrated that while there was a high correlation with the 3D computation of flow rate from the single best aliasing $24$ ml/s ($r=0.98$, $0.97$, $0.97$ for circular, rectangular and triangular, respectively), the ability to average three threshold velocities (including $24$ cm/s) yielded even higher correlations ($r=0.99$, $0.99$, $0.98$ for circular, rectangular and triangular, respectively). This post-processing capability will not only yield better results but also save time in the acquisition phase.

**Clinical Implications**

In current clinical practice, despite the variety device of invasive methods, including intravascular and transoesophageal echocardiography, few non-invasive methods are available for reliably quantifying the severity of valvular disease. MRI shows promise; however, its high cost, poor temporal resolution and time consuming nature have severely limited routine clinical use\textsuperscript{28}. As described in the method section, the approximate time for our 3D data acquisition is less than $30$ s, since the 3D rotation is triggered at one frame per cardiac cycle and the number of frames limited to $30$ for each acquisition by the cineloop memory. This, including file transfer time ($30$–$90$ s), equates to a total time of $60$–$120$ s per acquisition. The off-line measurement of 3D FCR normally takes between $5$–$6$ min. From these perspectives, we can see that 3D multi-threshold techniques have the potential to provide low-cost, rapid and reliable results for clinical application and offer a basis for understanding other flow events, such as shunt flows in various cardiac lesions.

**Study Limitations**

The use of this 3D multi-threshold flow convergence method requires centreline alignment of the object. In this study, we were able to place and adjust the transducer in the best position without the translational heart motion problems encountered in real clinical settings. Thus, the quality of 3D imaging will exceed that obtained from clinical settings.

Also, there is variability between each cardiac cycle due to this beat-to-beat-based 3D data acquisition. In a clinical situation, the acquisition time is dependent upon the individual’s heart rate and is limited by $30$ frames for
each 180° rotation. Therefore, the maximum 3D acquisition time should be less than 30 s and the variability between each heart beat during this short period will be minimized. The limitations inherent in colour Doppler flow mapping, such as wall filter settings, colour gain, variety of aliasing velocities and pulse repetition frequency, are carried into the 3D reconstructed flow convergence images. Likewise, the colour information at the lateral margins of the flow convergence, where velocity vectors are more perpendicular to the echo beam, are subject to a loss of colour information due to Doppler angle dependence. Thus, portions of the imaged flow convergence surface area adjacent to the orifices in this study do not represent the actual isovelocity surface area, which could contribute to our underestimation of actual flow rates. This is a separate issue from the question that would arise clinically in which the interrogating beam might not be aligned parallel to flow, which will be potentially amenable to correction by the use of multiple positions or composite image rasters.[29]

Alternatively, angle-independent methods such as MRF[28] and digital particle tracking could offer potential solutions. Additionally, the version of ATL 3000 ultrasound system’s cineloop memory that we employed did not permit the continual storage and transfer of images over an entire cardiac cycle for the 3D acquisition of all of the rotational views of colour Doppler data. Therefore, we used images at peak flow and computed peak flow rate for this study.

We expect further developments of computer technology for automatic boundary and surface area determination and advances in ultrasound equipment to eventually provide the capability of real-time acquisition and quantitation of 3D colour Doppler flow events.

Conclusions

A multiple surface 3D flow convergence area method was developed and demonstrated highly accurate quantification of flow rate. Further developments of 3D digital velocity flow imaging methods should improve flow quantitation in clinical echocardiography.

References


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