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Objectives. This study investigated the proximal centerline flow convergence region simultaneously by both color Doppler and laser Doppler velocimetry.

Background. Although numerous investigations have been performed to test the flow convergence method, to our knowledge there has yet been no experimental study using reference standard velocimetric techniques to define precisely the hydrodynamic factors involved in the accelerating flow region during steady and pulsatile flow.

Methods. Using an in vitro model that allows velocity measurements by laser Doppler velocimetry with simultaneous comparison with color Doppler results, we studied the centerline flow acceleration region proximal to orifices of various sizes (0.08 to 2.0 cm²).

Results. Agreement between theory and experimental velocities was good for large flow rates through small orifices only, and only at distances >1.2 cm from the orifice. Changing the orifice shape from circular to slitlike produced no significant changes in velocity profiles. Constraining the proximal side walls caused a significant increase in proximal velocities at distances >0.7 cm for the largest orifice only (2.0 cm²). Calculated flow rates agreed well with actual flow rates, with functional dependence on proximal distance and orifice size. Velocity profiles for pulsatile flow were similar to steady state flow profiles and could be integrated to calculate stroke volumes, which followed actual flow volumes well, although with general overestimation (y = 1.22x + 0.164, r = 0.92), most likely due to the use of all available proximal velocities.

Conclusions. The accelerating proximal flow region responds to several hydrodynamic factors that can affect flow quantitation using the flow convergence method in the clinical situation.

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the underlying physical principles for accurate quantitation of physiologic flows. However, little actual experimental data on the flow convergence region that use reference standard methods for velocity measurements exist. Such data should prove useful for comparison with available numerical and color Doppler data and should throw further light on the events involved in flow accelerating into a finite-sized orifice.

Within this context, we devised a series of steady and pulsatile flow experiments to measure the centerline velocity profile proximal to finite-sized orifices using a newly developed model that allows the simultaneous performance of ultrasound color Doppler flow mapping and the reference standard technique of laser Doppler velocimetry.

Methods

Background. The proximal hemispheric flow convergence method for computing orifice flow is based on the theoretic assumption of inviscid point-orifice flow (Fig. 1, top), which is defined as follows: 1) The orifice is represented as an infinitesimally small point—the point-orifice flow assumption; and 2) flow behavior is entirely nonviscous—the inviscid flow assumption. Under these conditions all proximal velocities are directed at the point orifice, and isovelocity contours proximal to the orifice are hemispheric in shape. Flow rate through the orifice is calculated by computing flow rate through one of these proximal isovelocity hemispheres, which, according to the principle of continuity, should equal flow through the orifice. As Figure 1 (top) shows, the radius of the hemisphere is taken as the distance from the orifice to the isovelocity contour, and flow rate through the hemispheric surface is obtained by multiplying the surface area of the hemisphere by its corresponding velocity:

\[ Q = 2\pi r^2 V_r, \quad [1] \]

Where \( r = \) distance proximal to the orifice; \( Q = \) orifice flow rate; and \( V_r = \) velocity into the orifice at distance \( r \). Equation 1 is called the simple hemispheric flow convergence equation or the proximal isovelocity surface area (PISA) equation (1-11). On color Doppler flow mapping, the isovelocity contour can be clearly visualized as the location of color change caused by blood velocity exceeding the Nyquist limit of the ultrasound system, causing a color “wraparound” or alias, and the distance from the orifice to this point of color change is taken as the radius of the isovelocity hemispheric contour, with velocity given by the Nyquist limit. The velocity magnitude at any proximal point \( r \) is obtained by solving equation 1 for \( V_r \):

\[ V_r = \frac{Q}{2\pi r^2} \left( \frac{1}{r} \right). \quad [2] \]

Use of proximal centerline velocities. The theoretic assumptions of steady inviscid flow and a point orifice obviously do not reflect the actual situation of physiologic blood flow, which involves viscosity, pulsatility and finite-sized orifices. Color Doppler flow mapping adds to these problems with its reliance on flow being parallel to the ultrasound beam for accurate velocity calculations that contain no angle-induced errors. From previous numerical and in vitro studies, it is clear that conditions in the region relatively far away from the orifice and along the axis through the center of the orifice (proximal centerline) should come closest to satisfying the theoretic assumptions because of the relative distance from any side walls that may produce viscous or constrainment effects (3,9,12). Furthermore, at these proximal “far-field” distances, the orifice should appear to be pointlike because the flow will be purely pressure driven with no effects of orifice shape or three-dimensional geometry (13,14). Therefore, if velocities along the centerline axis proximal to the orifice could be obtained accurately and with good temporal resolution, the effect of the assumptions made by the hemispheric flow convergence equation may be minimized. Color M-mode imaging is the only current color Doppler technique that has
the capability to measure proximal centerline velocities with good temporal and spatial resolution. Accordingly, we used color M-mode imaging with the beam aligned along the proximal centerline axis of the orifice to obtain color Doppler velocity data for comparison with the laser Doppler velocimetry results.

**Experimental design.** A transparent acrylic flow phantom (outer dimensions 30 cm x 30 cm x 30 cm, 1.28 cm thick) with a proximal ultrasound window was used as the test section for simultaneous laser Doppler velocimetry and ultrasound color Doppler flow convergence studies (Fig. 2, top). The cubeshaped phantom consisted of the entrance chamber, proximal chamber, orifice plate and distal chamber. Flow entering the entrance chamber was distributed around the chamber into a honeycomb section to minimize flow disturbances and through a densely perforated plate (68% forward flow area). In this way, flow entering the chamber immediately proximal to the orifice was ensured to be uniform and similar to the theoretically assumed condition of flow into a point orifice. The orifice plate (1.25-cm thick acrylic) contained a large central hole into which the orifice was mounted. Circular flat Plexiglas orifices with cross-sectional areas of 0.08, 0.16, 0.24 and 2.0 cm² were used. A rectangular orifice (8:1 major/minor axis ratio) of cross-sectional area 0.24 cm² was also used to study the effects of orifice shape. The plate could be moved to various distances (8 to 12 cm) from the ultrasound transducer by sliding it into grooves machined in the inside wall of the phantom. The dimensions of the distal chamber allowed an unconstrained jet to form on the downstream side of the orifice, and flow from this chamber was removed through a 1-in. (2.54-cm) exit and returned to the flow reservoir.

The general flow setup is shown in Figure 2, bottom. Steady flow was obtained by use of a constant-pressure flow tank with overflow, and pulsatile flow was produced by use of an oscillating piston pump (Harvard Apparatus model 1423). Steady flow rates from 2 to 6 liters/min in increments of 1 liters/min, and pulsatile flow volumes from 5 to 50 ml/beat at frequencies from 40 to 80 beats/min were used as the flow conditions. Flow rates were measured by a flow meter (Signet Scientific MK 577 Digital Flometer) calibrated using the graduated cylinder and stopwatch method. For pulsatile flow, mean stroke volume obtained by measuring volume output over a prescribed number of beats was used for calibration of the pump settings. Pressures in the chambers proximal and distal to the orifice were also measured (Validyne DP 103 Pressure Transducers calibrated against known water columns), and both flow and pressure measurements were entered into a Macintosh microcomputer using a data acquisition package (Superscope, GW Instruments).

The effects of flow rate, orifice size and orifice shape on the centerline velocities proximal to the orifice were investigated. In addition, the applicability of the simple hemispheric flow convergence equation for flow calculation was also tested on both color Doppler and laser Doppler velocimetry data. The effect of proximal walls that may constrain the flow was also studied. The flow convergence method assumes that flow approaches the orifice from all directions and in uniform fashion. However, cardiac chambers, such as left ventricular or left atrial walls, or the aortic root may alter the nature of the flow convergence region through proximal flow constraint and may thereby affect volumetric flow calculation using the proximal flow convergence method. The control condition consisted of no proximal walls upstream of the orifice, creating a very large proximal chamber that allowed flow to approach the orifice uniformly and from all directions. This condition resembled the theoretically assumed condition of inviscid flow into a point orifice and was used to compare measured centerline velocity profiles to theoretic ones and as a control to study the effect of adding proximal walls. Next, a transparent cylindrical tubelike proximal chamber 3.4 cm in diameter and 5.8 cm in length with a wall thickness of 0.5 cm was mounted proximal to the orifice. By varying orifice size, the ratio of proximal chamber diameter to orifice diameter could be changed to model various orifice–wall distances.

The fluid medium used was deionized water mixed with two types of particles to simulate scattering of red blood cells for the ultrasound Doppler studies and to provide scatter particles for the laser beam. For the laser Doppler velocimetry technique, neutrally buoyant solid nylon particles (mean diameter 5 μm) were mixed into the water to produce a 0.1% by weight suspension. For the ultrasound studies, an echocardiographic contrast agent (Albunex, Molecular Biosystems Inc.), was used to simulate red blood cell reflection. Very small doses (0.5 to 2 ml) were required to obtain excellent Doppler signals and were introduced through a port located in the chamber proximal to the orifice immediately before the measurements were carried out so that the contrast agent mixed instantly with the fluid in the test section. The contrast agent was of sufficiently small concentration that no interference occurred with the laser Doppler velocimetry measurements. The nylon particles and contrast agent microbubbles were not of sufficient quantity to change the characteristics of the fluid medium; thus, viscosity and density were the same as that for water.

**Velocity measurement techniques.** Our in vitro model allowed simultaneous velocity measurements using laser Doppler velocimetry, a multicomponent velocity measurement technique routinely used in hydrodynamics measurements, and the clinically used method of ultrasound color Doppler flow mapping. This simultaneous method has several advantages, foremost among which is the use of a reference standard technique simultaneously for point-to-point ultrasound color Doppler velocity corroboration and accurate flow measurements.

**Laser Doppler velocimetry.** Laser Doppler velocimetry is a long-established technique for accurate multicomponent velocity measurements in fluid flows and has been used previously in verifying the accuracy of pulsed wave Doppler measurements (15). There are several laser Doppler velocimetry schemes that are in current use for velocity measurements, and the optimal method depends on the nature of the studied flow (16). Because pulsatile flow was an important part of our experiments, we used the frequency tracking method, which uses a phase-locked loop circuit to follow temporally varying
velocities in positive and negative directions (17). The laser Doppler velocimetry system consisted of a 5-W argon ion laser (Spectra Physics model 2025), transmitter (Dantec 60X40), four-beam laser Doppler probe (Dantec 60X11 Fiber Flow Series), photomultiplier (Dantec 50X00), 40-kHz frequency shifter (Dantec 55N20), and frequency tracker (Dantec 55N10) for tracking velocities.

The laser produced two beams (green and blue) with wavelengths ($\lambda = 514.5 \times 10^{-10}$ cm [green] and $\lambda = 488.0 \times 10^{-10}$ cm [blue]). Each beam was also shifted 40 kHz to track velocities in both directions, resulting in a total of four beams intersecting at the point of velocity measurement. Beam separation was 0.38 cm, focal length was 1.598 cm, and intersection angle was 13.56°. Each blue and green beam was used to measure one of two orthogonal velocity components that were then vector added to produce the axial (into orifice)
component of velocity (17). The output of the laser Doppler velocimetry system was in the form of a voltage signal proportional to measured velocity that was digitized into a computer using a data acquisition system (MacADIOS data acquisition system) with a 12-bit analog/digital converter mounted on a Macintosh microcomputer. Three laser Doppler velocimetry measurement sequences of 2,000 points each were averaged to produce the velocity at each proximal centerline point. Distance proximal to the orifice was measured by a calibrated tracking system on which the laser wand could be moved axially and transversely, so that distance from the orifice for each point could be measured. Centerline velocities at distances from 0.3 to 2.1 cm proximal to the orifice were measured.

Ultrasound color Doppler flow mapping. A Vingmed CFM 750 system (scanning at 5 MHz for tissue and at 4 MHz for color and spectral Doppler) equipped with a digital output port for transfer of digital velocity, spectral and echocardiographic data into a Macintosh microcomputer was used for these studies. Two-dimensional color maps of the proximal flow region and color M-mode imaging of the proximal centerline axis were the imaging modalities used along with continuous wave tracings through the orifice for orifice velocities. A rainbow map without variance (Nyquist limit 48 cm/s) was used with gain settings optimized and kept constant through all studies. Low velocity reverts could be set on this system and were optimized to produce 3-dB roll-off at 4 cm/s for the steady flow results and at 8 cm/s for the pulsatile flow studies. For each experiment, a sequence of at least 6 s of color M-mode images from interrogations with the cursor directed through the center of the two-dimensional color flow convergence image was stored on the digital cineloop of the ultrasound system and transferred into the microcomputer in digital (raw data) format. The exact procedure for digital transfer of color M-mode velocities into an analysis computer has been described in our previous studies (12). For steady flow, 4 s of color M-mode data were averaged (800 points at 5-ms color M-mode resolution) for comparison with the laser Doppler velocimetry data. For pulsatile flow, 3 beats were averaged to produce the final velocity results. Two-dimensional color images were also transferred for any subsequently needed reference information, such as orifice position, but were not used for velocity comparison with laser Doppler velocimetry measurements.

Results

Steady flow. Proximal centerline velocity profiles. Figure 3 shows examples of proximal centerline velocities for the 0.08- and 0.24- and the 2.0-cm² orifices with no proximal walls at a steady flow rate of 6 liters/min as measured by color M-mode imaging and laser Doppler velocimetry. Centerline velocity profiles for the other flow rates were similar to those shown for 6 liters/min. The theoretic velocity curve of inviscid flow into a point orifice for each case as represented by equation 2 is also shown as the solid line.

The influence of orifice size on the centerline velocity profile measured by laser Doppler velocimetry is shown in Figure 4, and the effect of changing the orifice from a circle to a thin rectangle is shown in Figure 5. A paired Student t test showed that velocity values for the 2.0-cm² orifice were significantly different from velocities for each of the smaller orifices at proximal distances of 1.7 cm and closer for the 4-liters/min flow rate (p < 0.005) and at proximal distances of 1.9 cm and closer for the 6-liters/min flow rate (p < 0.002). Changing orifice shape from circular to rectangular did not change proximal centerline velocities significantly (p = NS) (Fig. 5).

The influence of adjacent proximal side walls on proximal centerline velocities was investigated on the 0.24- and 2.0-cm² orifices. No significant differences in centerline velocities were found for the 0.24-cm² orifice for cases with and without proximal walls. However, for the 2.0-cm² case (Fig. 6), centerline velocities with proximal side walls increased significantly, especially farther from the orifice, compared with cases without proximal constrainment (p = 0.0002 for 6 liters/min, mean increase 16.8% for distances >1.1 cm; p = 0.0037 for 6 liters/min, mean increase = 21.3% for distances >1.3 cm).

Calculation of flow rate. Flow rate calculated using the simple hemispheric equation method, normalized by actual flow rate (Calculated flow rate/Actual flow rate), is shown plotted against normalized proximal distance (Proximal distance/Orifice diameter) for a flow rate of 6 liters/min in Figure 7 (top). Such normalization allows easy comparison of results from orifices of various sizes because all units become dimensionless. Flow rate calculated using the simple hemispheric equation for each velocity–distance combination proximal to the orifice increased polynomially with increasing proximal distance for the smaller orifices and linearly for the 2.0-cm² orifice. Calculated flow rate graphs for the other conditions also exhibited similar behavior.

Pulsatile flow. Because the proximal side walls did influence the velocity profiles for the larger orifices (smaller side wall–orifice distance), and because we were interested in flow conditions dimensionally representing both valvular stenosis and regurgitation, all pulsatile flow cases were studied with adjacent walls proximal to the orifice. Previous studies have shown that regurgitation corresponds dimensionally to a situation where the immediate proximal structure dimension (Dₚ) is much larger than the orifice diameter (Dₒ) (Dₚ/Dₒ > 4.5), whereas stenosis can be simulated by a smaller ratio (Dₚ/Dₒ < 4.5) (18). Such nondimensional scaling facilitates comparative observations between “stenotic” and “regurgitant” cases when studied in vitro.

For the color M-mode velocity data, we used an alias-unwrapping algorithm previously validated in vitro (19) to extend the aliasing limit up to three times the original Nyquist velocity. Alias-unwrapped velocity measurements compared well with laser Doppler velocimetry measurements for pulsatile flow, as shown in Figure 8, which displays axial velocities measured at 0.3 cm proximal to the orifice and averaged over three beats. Our alias-unwrapping algorithm broke down after three aliases, as is evident in Figure 8 for velocities >150 cm/s,
Figure 3. Proximal centerline velocities for steady flow through three orifices (0.08, 0.24 and 2.0 cm²) for a flow rate of 6 liters/min (Lpm). Velocities were measured using laser Doppler velocimetry (LDV) and color Doppler flow mapping (CDFM) using the color M-mode technique. The color Doppler flow mapping and laser Doppler velocimetry equated well with each other, except for certain regions immediately proximal to the orifice. Also shown (solid lines) are the corresponding theoretic velocities for inviscid flow into a point orifice represented by equation 2 (see text). Overlap between theoretic and experimentally measured velocities indicates proximal points where the simple flow convergence equation would produce good results. The color Doppler flow mapping velocities overestimated laser Doppler and theoretic velocities for proximal points farther away, which results in overestimation of calculated flows using the flow convergence method. Likewise, close to the orifice, both color Doppler flow mapping and laser Doppler velocimetry fall much lower than the theoretic velocities, creating an underestimation of calculated flow at these points.

The procedure used for calculating volumetric flow rate is illustrated in Figure 9 for one flow condition. Centerline velocities at various depths during the pulsatile cycle as measured by laser Doppler velocimetry are shown in Figure 9B with its corresponding continuous wave Doppler trace shown in Figure 9A for a frequency of 40 beats/min and a stroke volume of 26 ml. To obtain the stroke volume, the following methodology was used. For each time increment, the velocity values for all proximal points were used to calculate an instantaneous flow rate based on the simple hemispheric flow convergence equation. In this way, multiple flow rates, each representing one unique velocity–distance combination, were calculated. For each time increment, the flow rates obtained from different proximal points were averaged to produce one instantaneous flow rate. These instantaneous flow rates are shown in Figure 9C, along with the standard deviations obtained from the averaging procedure. The area under the graph in Figure 9C represents the volumetric flow per beat and can be obtained by integration, producing a stroke volume of 29.44 ± 6.2 ml/beat.

For the color M-mode data, an automatic computer algorithm applied the previous procedure on all available color Doppler M-mode proximal centerline velocity data for each flow condition. This resulted in an averaging of between 15 and 40 flow rates obtained from corresponding velocity–distance combinations for every 5-ms increment of color M-mode data. These instantaneous flow rates were automatically integrated to produce a stroke volume calculation for each beat, and an
Figure 4. Effect of orifice size for all orifices studied is shown for a flow rate of 6 liters/min (Lpm) (other flow rates produced similar results). Only laser Doppler velocimetric data are shown here for simplicity—color Doppler data also produced similar results. Although the smaller orifices (0.08 to 0.24 cm$^2$) produced no significant changes in proximal velocities, velocities from the largest orifice were higher ($p < 0.05$, 6 liters/min) compared with those from the smaller orifices. The nature of the velocity increase was observed to be different for the largest orifice than for the three smaller orifices. In the general relation ($V_r \propto 1/r^3$), $\beta$ decreased from 1.5 for the 0.08-cm$^2$ orifice to 1.01 for the 2.0-cm$^2$ orifice at a flow rate of 6 liters/min. Thus, certain proximal locations will have higher velocities for the larger than the smaller orifice, although this will be reversed at the orifice.

![Figure 4](image)

average of five beats was taken to compare with actual flow rate. These results are shown in Figure 10. We found good correlation between actual and calculated flow volumes using the simple hemispheric equation to average multiple velocities for each time increment over the entire ejection cycle. Agreement also was good, but calculated flow consistently overestimated actual flow (mean overestimation 40%).

**Discussion**

The present study incorporates the use of one clinically used technique, color Doppler flow mapping, simultaneously with a recognized reference standard velocity measurement technique, laser Doppler velocimetry (16,17), allowing us to measure the flow convergence acceleration pattern and establish the accuracy of color Doppler data at the same time, providing a novel utility for in vitro flow measurements. The laser Doppler velocimetry technique has been used in various experiments, including measurement of velocity profiles through prosthetic valves and validation of pulsed wave Doppler measurements (15,20,21). Preliminary studies on the flow convergence region using laser Doppler velocimetry on steady flow models have been reported (22); however, we believe the present investigation to be the first to study the flow convergence region using both color Doppler and laser Doppler velocimetry simultaneously for steady and pulsatile flow.

**Comparison of color M-mode measurements and laser Doppler velocimetry.** Although a comprehensive treatment of the differences between laser Doppler velocimetry and color Doppler velocity measurements is beyond the scope of this report, some basic observations can be made. Color Doppler M-mode velocity data followed laser Doppler velocimetry data remarkably well for almost all experimental conditions with a few discrepancies. Color Doppler velocities consistently underestimated laser Doppler velocimetry at points immediately proximal to the orifice (0.3 to 0.9 cm) for almost all steady flow results. This is presumably a result of the averaging process inherent in color Doppler velocity estimation, which assigns velocity on the basis of an average of several pulse reflections, thereby smoothing variations in velocities within the color Doppler sample volume. Flow regions with rapid velocity changes over small distances would not be represented accurately even on color M-mode tracings. Color Doppler velocity measurements for pulsatile flow also revealed disparities during periods of rapid acceleration or deceleration when the color Doppler system could not adequately track the fast-changing velocity. Because the color Doppler autocorrelation technique assumes a time-stationary signal over the time taken to estimate the velocity (23), very rapid temporal velocity changes...

![Figure 5](image)
Figure 6. Proximal constraining walls produced no significant changes on the velocity profile proximal to the 0.24-cm² orifice (not shown) but did increase velocities for the 2.0-cm² orifice (shown here) for both 4 liters/min (Lpm) (top) (p = 0.0002) and 6 liters/min (bottom) (p = 0.0037), suggesting that orifice size and therefore transverse distance to the proximal wall play important roles in producing constraining effects on the flow convergence region. The effect of this type of flow constrainment is to increase proximal velocities, which may cause flow convergence techniques to overestimate actual flow. W/out = without.

variations would create estimation problems even when using color M-mode imaging. However, none of these disparities was enough to create any significant problems of velocity measurement, and all of them should be solved in the next few generations of color Doppler systems.

**Proximal centerline flow convergence region.** The centerline region of velocities proximal to the orifice has been previously studied to relate a nomogram of centerline velocity-distance combinations to orifice flow rate (3). This method has since been verified in vitro as a simple method for potentially assessing the severity of regurgitant lesions (12). Our investigation focused on the velocities in the proximal centerline region for three main reasons: 1) This region is the least affected by viscous effects from side walls, and so the flow in this region would approach the assumed conditions of inviscid flow; 2) comparisons to laser Doppler velocimetry measurements could be performed easily without angle-correcting the color M-mode velocities; 3) this region facilitated the use of color M-mode imaging, which is far superior to two-dimensional color Doppler for velocity measurements with good temporal and spatial resolution.

**Steady flow.** The characteristics of the flow convergence region have previously been studied using numerical techniques and color Doppler flow mapping (1-11,19). Studies that numerically solve the Navier-Stokes equations to obtain velocity fields proximal to the orifice have revealed a dependence of the isovelocity contour on distance from the orifice, with contours becoming flattened at points very close to and
Figure 7. Relation of the simple hemispheric equation method for orifice flow calculation to proximal distance at 4-liters/min (top) and 6-liters/min (bottom) flow rate. On the ordinate, a normalized flow rate value of 1 indicates perfect agreement between actual and calculated flow rates. Proximal distance was normalized by orifice diameter so that orifices of various sizes could be compared on the same graph. We observed a marked difference in flow rate calculations between the smaller orifices and the large orifice. Smaller orifices exhibited a polynomial change between flow rate and proximal distance with an inflection point around the location where calculated flow rate by flow convergence agreed with actual flow rate, between one and three orifice diameters upstream. However, the largest orifice showed a linear relation where calculated flow rate began overestimating actual flow much sooner (i.e., at regions farther than 0.5 orifice diameters). Calc = calculated.

hemielliptic far away from the orifice (4,9). This implies a breakdown in the assumptions of inviscid flow into a point orifice, which predicts hemispheric isovelocity contours for the entire proximal convergence field. Our results from both the laser and color Doppler data show that even in the most ideal experimental circumstances, the centerline velocity function does not behave as completely as that for inviscid flow into a point. We also observed an overestimation of color Doppler flow mapping velocities compared with laser Doppler velocimetry and theoretic velocities at points farther proximal to the orifice, which indicates that flow convergence calculations using color Doppler flow mapping will overestimate actual flows at these points. Closer to the orifice, color Doppler flow mapping and laser Doppler velocimetry points underestimated theoretic velocities; therefore, calculated flows at these points will also underestimate actual flow. Our results explain previous findings reporting a strong dependence of flow calculations using flow convergence on various factors, such as flow rate, orifice size and pressure gradient (2-4,6-11), by showing that many of these dependencies arise from experimentally or clinically obtained velocities that do not match the velocities prescribed by the unrealistic assumptions of inviscid sink flow.
for a majority of flow conditions, thereby causing errors in estimating actual flow rate. These relations can be further analyzed and better understood by examining the influence of flow rate, orifice size and orifice shape on the centerline velocity curve.

**Influence of flow rate.** In the first part of our study (i.e., orifices without proximal side walls), the smaller orifices (0.08 to 0.24 cm²) provided results that agreed most closely with theoretic predictions. Within this data set, flow rate produced a strong effect on how well proximal velocities agreed with corresponding theoretic values. For constant orifice area, increasing flow rate has the effect of increasing the pressure gradient through the orifice, thereby diminishing the effect of viscosity and creating better agreement between theoretic and measured centerline velocities. However, for the 2.0-cm² orifice, measured velocities did not approach the theoretic values in any significant manner for the range of flow rates studied. For larger orifices, flow rate would presumably need to be very large before any superposition between measured and theoretic velocities could occur. This explains the general underestimation that is found when flow convergence is used to calculate stenotic flow where pressure gradient is low, allowing proximal velocities to be measured only close to the orifice (10).

**Influence of orifice size.** Previous studies have also pointed out that finite-sized orifices cause a deviation in measured proximal centerline velocities from the theoretic case, but only at certain proximal locations (3,4,9,12). Our results reveal that as orifice size increased, agreement between the measured centerline proximal velocity function curve and the theoretic curve deteriorated rapidly. The nature of the velocity increase into the orifice changed as the orifice size increased, becoming less like a second-order function, as predicted by the simple hemispheric equation (v ∝ 1/r²) and more like a singular inverse function (v ∝ 1/r). For example, the exponent β in the general expression (v ∝ 1/r^α) was found to decrease from 1.5 for the 0.08-cm² orifice at 6 liters/min to 1.01 for 2.0-cm² orifice at the same flow rate. A result of such a decrease in the exponent can be seen in Figure 4, where proximal velocities for the larger (2.0 cm²) orifice are greater than those for the smaller orifices. Although for the same flow rate, the velocity at the vena contracta for the smaller orifices will be larger than the vena contracta velocity for the larger orifice because of the principle of continuity, this relation is in fact reversed at proximal points far from the orifice, indicating the strong influence of orifice size on flow acceleration immediately proximal to the orifice. This has the effect of allowing agreement between theoretic and experimental velocities only at distances relatively far from the orifice, the “far field” of the flow region, where conditions begin to resemble those assumed for derivation of the simple hemispheric equation. This is expected from theory because the large orifice “near field,” a variable that is dependent on orifice size, consists of a larger area proximal to the orifice, and because the farthest proximal distance at which measurements were taken was at only 2.3 cm, many of the data points for the larger orifices were in this “near-field” region (i.e., <1 orifice diameter), where the velocity function for finite-sized orifices falls away from the inviscid point-orifice flow prediction and becomes curvilinear into the orifice (14). Data at points farther than 2.3 cm were not taken because points farther than this are very rarely encountered in the clinical setting as a result of the finite dimensions of the ventricular and atrial chambers and interference from competing flows, such as left ventricular outflow tract flow during systole in the case of mitral regurgitation or pulmonic emptying of the veins into the left atrium in the case of mitral stenosis. This means that cases of regurgitation, where orifice sizes are usually small, need to be considered separately from those of stenosis, where, in most cases, the orifice sizes are larger, although severe stenosis can produce effective flow areas that are comparable to large regurgitant lesion areas.

**Influence of orifice shape.** There were no significant differences found in the proximal centerline profile at distances >0.3 cm between orifices that were circular and rectangular in shape but with the same cross-sectional area. The rectangular orifice that we used is almost slitlike in shape and represents the other extreme from circular for two-dimensional planar orifices. Only immediately proximal to the orifice should shape-induced viscous and pressure effects become important; farther away the flow will recognize a convective pressure gradient only, and along the proximal centerline, flow will be driven purely by this convective force. Thus, flow convergence calculations should not be affected by orifice shape so long as data points are taken far enough proximal to the orifice. For the 0.24-cm² orifice, this region was found to lie at distances >0.5 cm proximal to the orifice.

Figure 8. Comparison of laser Doppler velocimetry (LDV) and color Doppler flow mapping (CDFM) techniques of measuring velocity at a point 0.3 cm proximal to the 0.24-cm² orifice for pulsatile flow. An alias-unwrapping algorithm that extended the aliasing limit to three times the Nyquist velocity (48 cm/s) was used to unwrap the color Doppler data. However, the algorithm could not track velocities exceeding this new limit, as shown by the breakdown of color Doppler velocimetry points around 150 cm/s. Apart from this, color Doppler followed laser Doppler velocities well for the rest of the cycle.
Integrated Flow Volume = \int_{0}^{t=500\text{ms}} q(t) \, dt

Influence of proximal walls. The flow convergence method assumes that flow approaches the orifice equally from all directions. However, the finite geometry proximal to a valvular lesion will prevent the uniform approach of proximal flow and so may cause a significant breakdown in the previous assumption. Our experiment allowed us to study, in isolation, the effect of proximal walls on flow calculations using flow convergence. Our results show it is not purely the existence of proximal walls but the transverse distance from the orifice to the proximal wall that determines changes in the centerline velocity function. For smaller orifices that are farther away from proximal side walls, no significant changes in centerline velocities were observed. As orifice size increased, centerline velocities did increase when the wall was added; however, the relative increase was also dependent on flow rate, with low flow rates (2 to 4 liters/min) causing insignificant increases and higher flow rates (5 to 6 liters/min) producing significantly higher velocities. The presence of proximal walls can affect flow through two primary factors: the formation of a viscous boundary layer along the inside of the proximal walls, and flow constrainment that leads to flow being squeezed into a smaller area before being further constricted as it flows through the orifice. Previous studies on flow convergence visualization using soap film, digital particle image velocimetry and laser-induced fluorescence flow visualization techniques have shown the formation of a boundary layer along the inside of proximal side walls, the size of which was shown to depend on the ratio of proximal chamber diameter to orifice diameter (24). The additional factor of flow constrainment would lead to further localized pressure drops as flow approaches the orifice with a
concomitant increase in proximal velocities. Our results show that a noticeable increase in local proximal velocities occurs only for high flow rates (4 to 6 liters/min) and for the largest orifice (2.0 cm²), and this had the effect of causing an overestimation (average of 20% for our studies) in calculated flow. This overestimation increase is relatively modest in relation to the overestimation caused by other factors, such as proximal distance (Fig. 7); however, it should be kept in mind as a possible additional reason for overestimation of actual flow when using the simple hemispheric method.

Pulsatile flow. Centerline profiles. As stated previously, the color M-mode imaging technique could not capture the initial onset of flow acceleration for pulsatile flow, and 60 to 100 ms passed before the color M-mode profiles became similar in shape to steady state profiles. These profiles remained similar to each other until velocity decreased to a point where the flow broke up into background flow, and the centerline velocities fell away from each other.

Centerline velocities for pulsatile flow as measured by the laser Doppler velocimetry technique showed similarities to the steady flow profiles for the majority of the ejection cycle (Fig. 9B). Only during the periods of flow initiation and stoppage did the centerline velocity profiles become flatter, indicating minimal acceleration of flow into the orifice. For the case shown in Figure 9B, this is illustrated at 50 and 450 ms, where very little increase in velocity was observed as the distance to the orifice decreased. However, at all other times velocity increased rapidly as proximal distance decreased.

Flow volume calculation. Figure 9C shows the results of calculating instantaneous flow rates using the simple hemispheric equation for every velocity-distance combination and averaging over each time increment. The ability to obtain digital velocity information from the ultrasound scanner greatly facilitated this type of flow volume calculation because large amounts of data can be easily analyzed to average out beat-to-beat variations. The area under this curve represents the stroke volume and can be found by integration, producing a result of 29.44 ml, which compared favorably to the true stroke volume of 26 ml. We found that the simple hemispheric equation used to calculate instantaneous flow rates (averaged for each time increment) and then integrated over the ejection period provided good estimates of actual flow volume, although with overestimation (Fig. 10). The overestimation presumably occurs because all velocities proximal to the orifice were used for volume calculation (Fig. 7), even those velocity points where spatial acceleration into the orifice had not begun. Previous studies have also shown a marked overestimation of actual flow rate for increasing proximal distances using the proximal flow convergence method (2,3,5,9,10). This overestimation seems to overcompensate for the underestimation usually found when extremely close distances proximal to the orifice are used. Using laser-induced fluorescent particle tracking, where streamlines converging into the orifice can be visualized, we found marked changes in the isovelocity contour pattern depending on where, proximal to the orifice, measurements were made (5). Other reports have also discussed the change in the isovelocity pattern, suggesting a hemiellipsoid far away from the orifice, which leads to overestimation, and a flat, “pan-shaped” contour shape very close to the orifice that leads to underestimation in flow calculation using flow convergence (8,11,25). A more sophisticated method that selects particular proximal velocities on the basis of the spatial derivative of velocity to calculate flow has been shown to decrease error in volume calculation (19,26). Several other
methods to localize the proximal region properly for flow calculations have also been studied with good success (9,11,23,27). Because the focus of this study was on fundamental characteristics of the flow acceleration into a restrictive orifice, these other algorithms for flow calculation were not applied here.

**Study limitations.** This study was designed to answer specific questions with regard to orifice size, shape, proximal walls and pulsatility; therefore, physiologic factors, such as proximal wall geometry, orifice shape and chamber compliance, were not reproduced accurately. Water rather than a fluid equal in viscosity to blood was used, although numerical and experimental studies have found no difference in proximal centerline velocities purely related to increases in viscosity (3,4,9). The proximal cylinder used to test the influence of proximal walls was rigid and not meant to reproduce cardiac walls that move compliantly over the cardiac cycle. However, our intentions were to explore in isolation the implications of the inherent assumption of uniformly directed flow. Also, actual flow convergence isovelocity contours were not obtained by color two-dimensional Doppler because of the inaccuracy of color Doppler in measuring velocity vectors at large angles to the ultrasound beam and the large amount of time that would be required to produce isovelocity contours for all flow conditions based on a single-point velocity measurement technique, such as laser Doppler velocimetry. As a corollary, only color M-mode data were used for ultrasound color Doppler velocity measurements, and the one-dimensional nature of this technique creates problems in the clinical situation where the orifice may be difficult to locate. However, color M-mode imaging is presently the only modality that tracks pulsatile flow velocities accurately and with good temporal and spatial resolution.

**Conclusions.** Our results reveal that the flow convergence method can be used for flow quantitation provided that there is a clear understanding of the limitations and assumptions of applying such a simplified technique. Presently, this method should prove useful in the clinical situation for determining general flow magnitude or regurgitant severity. More sophisticated flow calculation algorithms that preselect proximal velocities and utilize correction factors should serve to increase the accuracy of flow convergence calculations. Because the technique is independent of measurement modality, it should also be very useful when applied on other, more accurate methods, such as magnetic resonance imaging velocity mapping, that can provide spatial velocity vectors for true isovelocity contour mapping.

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**References**

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