Validation of the Proximal Flow Convergence Method
Calculation of Orifice Area in Patients With Mitral Stenosis

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Background. It has been proposed recently that measuring the flow convergence region proximal to an orifice by Doppler flow mapping can provide a means of calculating regurgitant flow rate. Although verified in experimental models, this approach is difficult to validate clinically because there is no ideal gold standard for regurgitant flows in patients. However, this method also can be used to derive cardiac output or flow rate proximal to stenotic orifices and therefore to calculate their areas by the continuity equation (area=flow rate/velocity). Applying this method in mitral stenosis would provide a unique way of validating the underlying concept because the predicted areas could be compared with those measured directly by planimetry.

Methods and Results. We studied 40 patients with mitral stenosis using imaging and Doppler echocardiography. Doppler color flow recordings of mitral inflow were obtained from the apex, and the radius of the proximal flow convergence region was measured at its peak diastolic value from the orifice to the first color alias along the axis of flow. Flow rate was calculated assuming uniform radial flow convergence toward the orifice, modified by a factor that accounted for the inflow funnel angle formed by the mitral leaflets. Mitral valve area was then calculated as peak flow rate divided by peak velocity by continuous-wave Doppler. The calculated areas agreed well with those from three comparative techniques over a range of 0.5 to 2.2 cm²: 1) cross-sectional area by planimetry (y=1.08x−0.13, r=.91, SEE=0.21 cm²); 2) area derived from the Doppler pressure half-time (y=1.02x−0.14, r=.89, SEE=0.24 cm²); and 3) area calculated by the Gorlin equation in the 26 patients who underwent catheterization (y=0.89x+0.08, r=.86, SEE=0.24 cm²). Agreement with planimetry was similar for 22 patients with mitral regurgitation and 18 without it (P>.6), as well as for 6 in atrial fibrillation (P>.2).

Conclusions. These results validate the proximal flow convergence concept in the clinical setting and also demonstrate that it can be extended to orifice area calculation using the continuity equation. (Circulation. 1993;88:1157-1165.)

KEY WORDS • Doppler • echocardiography • color flow mapping • mitral stenosis

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validate the proximal flow convergence method in the clinical setting by demonstrating that it can be used to calculate mitral valve area by continuity in patients with mitral stenosis.

Methods

Patients

To test this method in the clinical situation, we studied 40 patients with typical rheumatic mitral stenosis selected for image quality suitable for quantification. There were 8 men and 32 women; mean age was 57 ± 21 years. Thirty-four patients were in sinus rhythm and 6 were in atrial fibrillation. (The initial 25 patients were selected to be in sinus rhythm; subsequently, patients with atrial fibrillation were also included.) Twenty-two patients had associated mitral regurgitation (12 mild, 4 moderate, and 6 moderate to severe by Doppler color flow assessment of jet extent into the left atrium\textsuperscript{18,19}). Eleven patients had aortic insufficiency, which was generally mild.

Echocardiography

Two-dimensional echocardiography, Doppler ultrasound, and color flow mapping were performed using a Hewlett-Packard 77020A or 1000 ultrasound imager equipped with a 2.5-MHz phased-array transducer and a standard velocity map. Color flow mapping of the mitral inflow was obtained from the apical window, with color gain adjusted to eliminate random color in areas without flow. From this window, the four-chamber view provided the most consistent image of the largest proximal convergence radius and allowed the proximal flow to be viewed most nearly parallel to the ultrasound beam. This view was scanned to image the largest proximal flow convergence region, and the aliasing velocity was reduced by shifting the color baseline to maximize this area on the image.\textsuperscript{2,4,7,9,11,14,20,21} Mitral inflow velocities were measured from the apex by continuous-wave Doppler, and the mitral valve was scanned in the parasternal short-axis view to image the smallest orifice area for planimetry.

Measurements and Calculations

In mitral stenosis, flow cannot approach the orifice equally from all directions but is confined by the stenotic leaflets to a proximal funnel region (Fig 2). Therefore, as an initial approximation, we treated the flow convergence region as a wedge or sector of the flow convergence region for orifices in a flat plate and empirically modified the flow calculation by an angle correction factor that expresses the restriction imposed by the inlet funnel. This factor is the funnel angle $\alpha$ divided by 180°, the inflow angle without such restriction (Fig 3). This treats the two stenotic leaflets as restricting flow most acutely in one direction—and much less so in an intersecting dimension parallel to the major axis of the orifice—as an initial approximation consistent with echocardiographic observations. This angle correction factor accounts for the fact that if the leaflets were laid out flat, they would allow flow to converge toward the orifice over an arc of 180° in any direction, whereas in mitral stenosis, flow can converge toward the orifice over an arc of only $\alpha$ degrees. A theoretical rationale for the reasonability of this factor, which corrects for the solid angle subtended by the leaflets, is provided in the “Appendix,” and recent experimental reports have confirmed its applicability in vitro.\textsuperscript{22,23} We have shown previously that flow through both elliptical and circular orifices of equal area can be calculated by the same $2\pi r^2$ formula, where the radius $r$ of flow convergence is measured along the central axis of flow at a low aliasing velocity (far from the orifice).\textsuperscript{7}

The maximal radius of the proximal flow convergence region was measured in early diastole from the first aliasing boundary to the tips of the mitral valve in a direction parallel to that of flow (Fig 2). The funnel angle $\alpha$ containing the flow convergence region was measured in the same frame using an off-line analysis system. Peak forward mitral flow rate was obtained as the product of ($2\pi r^2$) × (the angle $\alpha$/180) × (the aliasing velocity). Mitral valve area was then calculated by continuity as peak forward flow rate divided by peak inflow velocity from the continuous-wave Doppler tracing. The radius, angle, and peak velocity were measured and averaged in 5 beats for patients in sinus rhythm and 10 beats in atrial fibrillation. (In atrial fibrillation, peak velocity varied an average of only 3.3% of the mean from beat to beat, and radius varied 5.2%.) The limiting mitral orifice area was planimetered at its maximal early diastolic opening.\textsuperscript{24-27} Orifice area was also estimated by the Doppler pressure half-time method as 220 divided by pressure half-time (milliseconds).\textsuperscript{28,29} Values were averaged
from 5 beats in sinus rhythm and 10 in atrial fibrillation. The presence of mitral regurgitation was assessed by color flow mapping and graded according to penetration into the left atrium as previously described.18,19

**Catheterization**

Data from cardiac catheterization were reviewed subsequent to the echocardiographic analysis and calculations. Catheterization had been performed in 26 patients (65% of the total) during the same hospital evaluation as the ultrasound study without intervening valvuloplasty. In each patient, right and left heart pressures were obtained. Cardiac output was measured by thermodilution or by the Fick method if tricuspid regurgitation was present. Mitral valve area was calculated by the Gorlin formula.30

**Observer Variability**

Two independent observers repeated 10 measurements of the radius, angle, and peak inflow velocity measurements. Interobserver variability was calculated as the standard deviation of the differences of their measurements. Similarly, one observer repeated the measurements to determine intraobserver variability.

**Results**

**Patient Studies**

In the 40 patients studied, mitral valve area by planimetry ranged from 0.5 to 2.2 cm² with a mean of 1.2±0.4 cm². The radius of the proximal convergence region ranged from 0.6 to 1.9 cm (mean, 1.2±0.3 cm), with most of the aliasing velocities in the range of 19 to 43 cm/s. The angle formed by the mitral leaflets ranged from 80° to 158° (mean, 118±15°), and peak instantaneous flow rate ranged from 5.4 to 25.3 L/min (mean, 12.9±4.6 L/min).

Mitral valve area by the proximal flow convergence method agreed well with direct planimetry (y=1.08x−0.13, r=.91, SEE=0.21 cm²) (Fig 4 and Table). ANOVA showed no significant effect of mitral regurgitation (P>.6) or atrial fibrillation (P>.2) on the relation between calculated and planimetered areas. Correlation was also good with mitral valve area calculated from the Doppler-derived pressure half-time (y=1.02x−0.14, r=.89, SEE=0.24.
cm²). Areas calculated by the proximal flow convergence method also agreed well with the results of the Gorlin equation in the patients who underwent catheterization \( (y=0.89x+0.08, r=.86, \text{SEE}=0.24 \, \text{cm}^2) \) (Fig 5 and Table). For all three comparisons (with areas by planimetry, half-time, and Gorlin methods), the slopes and intercepts of the regression lines were not significantly different from 1.0 and 0, respectively \((P > .2)\).

**Observer Variability**

The interobserver variability was 5.7% of the mean values for \( r \), 4.52% for angle \( \alpha \), and 2.1% for peak velocity. The resulting variability in calculated orifice area was 9% of the mean. The corresponding intraobserver variabilities were 3.3%, 3.8%, and 2% of the respective mean values, resulting in a variability of 5.7% of the mean in orifice area.

**Discussion**

The advent of Doppler flow mapping has led to attempts to extract quantitative physiological information from the displayed flow fields in accordance with fluid mechanical principles. One such approach is based on the observed flow acceleration proximal to regurgitant and shunt lesions.\(^1\)-\(^4\),\(^31\)-\(^34\) It has been proposed that these laminar flows could provide a measure of regurgitant flow rate unaffected by the complexities of turbulent flow beyond the orifice. The ability to visualize such flows with existing systems has indeed correlated with moderate to severe regurgitation,\(^2\),\(^31\),\(^32\) whereas calculation of flow rates has succeeded in experimental models.\(^1\)-\(^4\),\(^6\)-\(^16\) Clinical validation, however, has been limited by the absence of an ideal gold standard for measuring regurgitation. On the other hand, flow convergence regions are seen proximal to stenotic lesions as well, where they are especially prominent because they represent the entire forward output across such valves. In particular, observations of such regions in mitral stenosis have suggested the possibility of testing these principles in the context of that disease.

**Flow Convergence and Proximal Geometry**

The results of this study validate the flow convergence concept by demonstrating that it can be used to calculate mitral orifice area by continuity in patients with mitral stenosis. The results agree well with three independent measurements of orifice area: direct planimetry, the Doppler pressure half-time method, and the Gorlin equation in patients who underwent catheterization. The results also point out the need to account for geometry proximal to the orifice in applying the flow convergence method. Without correcting for the angle of the inflow funnel, the area would have been overestimated by up to 100%. The effect of inflow angle has not been addressed previously and is also likely to be important in assessing flows through regurgitant orifices within a nonplanar leaflet geometry surrounding the orifice. (Because the same form of inflow angle correction was used for all patients, the basic \( r^2 \) dependence of the proximal flow calculation could be confirmed by the results.) The observed applicability of flow convergence principles in the nonplanar setting is also consistent with theoretical predictions based on a finite difference solution of the basic Navier-Stokes equations describing fluid flow ("Appendix"; Fig 6) as well as with preliminary in vitro reports.\(^22\),\(^23\)

**Potential Applications**

In addition to validating the flow convergence concept, this technique may provide a simple and useful alternative to calculate orifice area in mitral stenosis when the pressure half-time is affected by altered

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**Fig 3. Schematic diagram of measurements used.** The radius of the flow convergence region is denoted by \( r \); \( \alpha \) is the inlet (funnel) angle formed by the leaflets proximal to the orifice in the region containing the visualized flow convergence sector. \( LV \) indicates left ventricle; \( RV \), right ventricle.

**Fig 4. Linear regression plot of orifice area calculated from the proximal convergence method compared with planimetered mitral valve area (MVA).** Open symbols represent patients without mitral regurgitation (MR); filled symbols represent those with MR. Circles represent patients in sinus rhythm (SR); squares represent those in atrial fibrillation (AF).

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### Mitral Valve Areas by Proximal Flow Convergence

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Patients are listed in order of data acquisition. The radius is that of the proximal flow convergence region at the aliasing velocity shown. The angle is the inlet angle formed between the mitral leaflets proximal to the limiting orifice. CW indicates continuous wave; MR, mitral regurgitation; AF, atrial fibrillation; and +, present.

chamber compliance or aortic insufficiency and direct planimetry from the parasternal window is technically limited. This calculation is also independent of associated mitral regurgitation, which can adversely affect the accuracy of the Gorlin equation.

**Limitations of the Flow Convergence Method**

The small size of the proximal convergence region on the video image can limit measurement precision. Such errors can be minimized by lowering the aliasing velocity to increase the measured radius. This also provides the most hemispheric isovelocity contours for which the proximal convergence calculation is most accurate, because in the immediate vicinity of an orifice, isovelocity contours tend to flatten out, leading to flow underestimation. (In this study, the measured radii, which represented the entire forward flow through a narrowed inlet, were relatively long [1.2±0.3 cm] compared with the visualized orifice dimension, predisposing to accurate flow calculations.) Shifting the baseline...
The limited spatial resolution of color displays on standard video rasters ultimately can be improved by using a direct digital output of the flow map. The limited temporal resolution of Doppler color flow mapping can cause variability in measuring peak flow rate. In mitral stenosis, however, flow rate varies relatively slowly after its early diastolic peak, minimizing this variability. (The color frame update rates of 10 to 20 Hz used, at a typical heart rate of 80, provided approximately 5 to 10 frames during diastole to observe the flow convergence region.) Potential variations in leaflet geometry proximal to the orifice may not always be accounted for completely by the simple angle correction used. In this regard, it is worth noting that the continuity equation, in principle, predicts effective orifice area at the vena contracta, where velocity is highest, as opposed to the anatomic orifice area, which is generally somewhat larger. In this study, however, the continuity and planimetered (anatomic) values coincided well. A potential explanation for this is that milder restriction of the converging flow by the leaflets in planes perpendicular to the measured funnel angle was not taken into account, causing a mild overestimation of flow rate and predicted area. In the end, however, the empirical modification used provides good agreement with planimetered values in the clinical application studied.

FIG 5. Linear regression plot of orifice area calculated from the proximal convergence method compared with mitral valve area (MVA) from the Gorlin equation in patients who underwent catheterization.

to reduce the aliasing velocity has potential limitations; for example, if the aliasing velocity is very low relative to the Nyquist limit (fixed by the pulse repetition frequency), selective suppression of low velocities by the color wall filter will cause overestimation of the mean velocity output displayed by color Doppler and therefore overestimation of the proximal flow convergence radius. Despite this potential, moderate degrees of baseline shifting have been used in practice with accurate results.

The limited spatial resolution of color displays on standard video rasters ultimately can be improved by using a direct digital output of the flow map. The limited temporal resolution of Doppler color flow mapping can cause variability in measuring peak flow rate. In mitral stenosis, however, flow rate varies relatively slowly after its early diastolic peak, minimizing this variability. (The color frame update rates of 10 to 20 Hz used, at a typical heart rate of 80, provided approximately 5 to 10 frames during diastole to observe the flow convergence region.) Potential variations in leaflet geometry proximal to the orifice may not always be accounted for completely by the simple angle correction used. In this regard, it is worth noting that the continuity equation, in principle, predicts effective orifice area at the vena contracta, where velocity is highest, as opposed to the anatomic orifice area, which is generally somewhat larger. In this study, however, the continuity and planimetered (anatomic) values coincided well. A potential explanation for this is that milder restriction of the converging flow by the leaflets in planes perpendicular to the measured funnel angle was not taken into account, causing a mild overestimation of flow rate and predicted area. In the end, however, the empirical modification used provides good agreement with planimetered values in the clinical application studied.

FIG 6. Finite-difference simulations showing isovelocity contours for converging flow. In each case, flow rate is 44 cm/s through a 0.126-cm² orifice at the bottom. Top panels show the total velocity magnitude; bottom panels show the axial component of velocity (the component displayed by a Doppler beam directed along the axis of flow [vertical lines]). In each case, the isovelocity contours (approaching the orifice) are 2, 5, 10, 20, 50, and 100 cm/s. On the right, the velocity field is constricted by a funnel-shaped geometry that forces the isovelocity contours away from the orifice. The stepped shape of the angled walls is a consequence of the finite-difference modeling software; this feature will tend to increase boundary effects. Despite this, the angle-corrected hemispheric approach is applicable and would therefore be even more likely to hold for smoother surfaces. The solid angle subtended by such a circumferentially limiting funnel is \[2\pi(1 - \cos \tau)\], where \(\tau\) is the angle between the central axis and the funnel wall.
Other Considerations

Although each measurement of mitral valve area used for comparison has its limitations, the agreement of flow convergence results with three independent methods provides stronger support in the clinical context. Other reports also have shown initial success with this method for calculating flow across an orifice with an angled inlet in vitro,22,23 for calculating mitral valve area in patients,42 and for calculating effective regurgitant orifice area both in vitro and in vivo.43 Although the flow convergence method, like all conventional methods, currently reports only one value for area, it makes no assumption that area must be constant during the cycle or with changing pressure gradients because it uses data corresponding to the same point in the cardiac cycle to obtain a maximal area. Finally, although the hemispheric flow convergence equation for flow rate (with inlet angle modification) can be most simply derived mathematically for inviscid flow, the “Appendix” and another recent study14 show that the same formula effectively holds for viscous flow as well. (The other study showed little change even with viscosities 100-fold greater than the physiological.14) Therefore, the formula used is compatible with empirical results and can be understood in terms of theory that includes physiological viscosity.

Summary

The proximal convergence method allows accurate estimation of mitral valve area in mitral stenosis and is not influenced by superimposed mitral regurgitation. This application provides a unique opportunity for comparing predictions based on flow convergence with directly measured values—in this case, planimetered orifice areas—as well as with other independent measures of orifice area. Therefore, these results validate the proximal convergence concept in the clinical setting and also demonstrate that it can be extended to orifice area calculation using the continuity equation.

Appendix

For flow converging toward a point orifice, conservation of mass implies that velocity rises inversely with the square of the distance from the orifice. Previous work1,2 has demonstrated that, for an orifice in a flat plate, velocity accelerates with hemispheric symmetry, and the flow rate (Q) may be calculated by Q = 2πr²v, where 2πr² is the area of a hemisphere of radius r and v is the observed velocity at the radius. If the orifice is situated in a nonplanar surface such as a funnel, one would expect the proximal convergence zone to be altered because the isovelocity contours cannot span a full hemisphere. To account for this alteration, flow would be calculated as Kr²v, where K is a geometry-dependent constant less than the 2π used in the planar case. We hypothesized that Kr² should equal the area of a hemisphere (2πr²) times the fraction of a hemispheric area subtended by the proximal leaflets and therefore available for flow. K would then equal the solid angle subtended by the funnel, which equals the solid angle of a hemisphere (2πr²) times the fraction of the hemispheric area subtended by the funnel walls.

To assess the plausibility of this correction factor, we obtained solutions to the Navier-Stokes equations (the determinants of viscous flow) for planar and funnel-shaped geometries. Commercially available finite-difference software for computational fluid dynamics was used (FLUENT, Fluent, Inc, Hanover, NH). An axisymmetric domain with a 2.5-cm radius and 2.5-cm axial length was defined using 2704 discrete nodes.

At the origin of the domain, an orifice with a 2-mm radius was created through which a flow of 44 cm³/s passed. Density and viscosity were physiological (1.05 g/cm³ and 3 cp, respectively). The proposed angle correction factor was tested in a worst-case situation in which convergence is limited circumferentially by the walls of a funnel; this creates the greatest opportunity for frictional effects to occur near those walls and create boundary layers that might distort the hemispheric contours and diminish the accuracy of the simple angle correction. Four geometries were tested, with the walls surrounding the orifice diverging from the central axis to create solid angles of 33%, 55%, 69%, and 100% of a hemisphere (2π). For each simulation, the constant K was calculated as that providing the optimal estimation of true flow rate in the equation Q = Kr²v. This was then compared with the solid angle subtended by the funnel.

Results

Fig 6 demonstrates isovelocity contours for flow through a circular orifice. The top panels show the absolute velocity magnitude and the bottom panels show the axial component of velocity (the component displayed by a Doppler beam directed along the axis of flow). On the left, the orifice is in a planar surface, whereas on the right, the walls restrict flow to only 55% of a hemisphere. This constriction forces the isovelocity contours to be farther from the orifice for the same flow rate. In each panel, the isovelocity contours represent 2, 5, 10, 20, and 100 cm/s (increasing as they approach the orifice). To calculate the true flow rate, it was necessary to decrease the constant K to 55% of 2π, predicted precisely by the global geometry. For each of the four geometries studied, the calculation determined that the appropriate constant K to estimate flow by Kr²v was given by the solid angle O subtended by the funnel (y = 1.00π + 0.05, r > 99). Of note is that the boundary layers (the infoldings of the velocity contours near the walls, reflecting frictional slowing) do not distort the central contours; therefore, where r is measured, flow behaves as a portion of a hemisphere, and, in fact, just that portion dictated by the anatomic inlet angle. As flow increases, the relative importance of viscous effects becomes even lower, reinforcing the applicability of the angle-corrected hemispheric formulation. Although the finite grid size of finite-difference software cannot perfectly represent and may slightly underestimate boundary layers, it also creates a stepped shape of the angled inlet walls (Fig 6) that will tend to increase boundary effects. Nevertheless, the center of the flow field, where the flow convergence radius is measured, should be well represented by such a method. In addition, non-Newtonian effects on flow would be expected to be least in the region of concern near solid boundaries, where shear rate (dv/dy perpendicular to the wall) is greatest.

Application

For two leaflets separated in a plane by an angle of α degrees, the portion of a hemisphere available for proximal converging flow between them is α divided by 180, since 180° gives a full hemisphere. Therefore, as shown above, Q = (the solid angle subtended by the leaflets)x(r²)/(the solid angle of a hemisphere times the portion of a hemisphere available for flow)x(r²)/(α/180). This is the empirical formula used in the clinical study, which is therefore also reasonable on theoretical grounds.

Flow Pulsatility

In the clinical study, r was measured at or near the time of peak flow rate, when dv/dt = 0, so that steady-flow derivations are applicable. In pulsatile flow, boundary effects that could alter the isovelocity contours should, in fact, be even lower than in steady flow because it takes time for frictional effects to propagate from the walls toward the main flow stream where r is measured. This time can be assessed by the viscous
diffusion time (characteristic length squared of the region of interest/kinematic viscosity); for a typical mitral flow convergence region of 1 cm and viscosity of 3 cp, this can be estimated as 33 seconds, far greater than a cardiac cycle. (This can also be gauged by the Womersley number=\sqrt{\text{[viscous diffusion time \times angular frequency of pulsation]}} \times 4 \text{ at a heart rate of 60 [angular frequency}=2\pi \times 1/6].) Viscous effects are negligible if this number considerably exceeds 1.4.45 Although the Womersley number applies to pulsatile flow in long tubes, the diffusion of viscous effects should be less of a problem in flow entering through an inlet wider than the orifice than for flow in a long tube without an initially wider inlet.) Pulsatility therefore tends to reinforce the applicability of the above angle-corrected hemispheric formulation.

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