Application of Color Doppler Flow Mapping to Calculate Orifice Area of St Jude Mitral Valve

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Background—The effective orifice area (EOA) of a prosthetic valve is superior to transvalvular gradients as a measure of valve function, but measurement of mitral prosthesis EOA has not been reliable.

Methods and Results—In vitro flow across St Jude valves was calculated by hemispheric proximal isovelocity surface area (PISA) and segment-of-spheroid (SOS) methods. For steady and pulsatile conditions, PISA and SOS flows correlated with true flow, but SOS and not PISA underestimated flow. These principles were then used intraoperatively to calculate cardiac output and EOA of newly implanted St Jude mitral valves in 36 patients. Cardiac output by PISA agreed closely with thermodilution ($r=0.91$, $\Delta=-0.05 \pm 0.55$ L/min), but SOS underestimated it ($r=0.82$, $\Delta=-1.33 \pm 0.73$ L/min). Doppler EOAs correlated with Gorlin equation estimates ($r=0.75$ for PISA and $r=0.68$ for SOS, $P<0.001$) but were smaller than corresponding in vitro EOA estimates.

Conclusions—Proximal flow convergence methods can calculate forward flow and estimate EOA of St Jude mitral valves, which may improve noninvasive assessment of prosthetic mitral valve obstruction. (Circulation. 1998;98:1205-1211.)

Key Words: mitral valve ▪ prosthesis ▪ echocardiography

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sessment of prosthetic heart valve function with current techniques is imprecise. Transvalvular gradients obtained by catheterization or Doppler echocardiography are indicative of prosthetic valve obstruction but are highly flow-dependent. Furthermore, pressure recovery occurs in some prosthetic designs, resulting in discrepancies between catheter and Doppler pressure gradients.

Analogous to native valve area, prosthetic orifice area is a flow-dependent measure of obstruction. Unfortunately, the Gorlin formula may be unreliable in this setting, and the pressure half-time method, widely used in native mitral stenosis, has not been validated for prosthetic valves. Pulsed-wave Doppler has been used to calculate prosthetic aortic valve area but is more problematic for prosthetic mitral valves.

Analysis of the proximal flow convergence region on color flow mapping can quantify mitral regurgitant severity as well as low-velocity flow across relatively large orifices such as stenosed native mitral valves and atrial septal defects. However, the utility of the proximal flow convergence to measure prosthetic forward flow and effective orifice area is unclear.

The aim of this study was to investigate, in flow models and in the operating room, the feasibility and accuracy of measuring flow across St Jude valves and calculating their EOA by analyzing the flow convergence region proximal to the prosthesis by use of color Doppler mapping.

Methods

Theoretical Background

The theory of the proximal flow convergence method is well described. Briefly, flow converges on a small orifice as concentric hemispheric shells of decreasing surface area and increasing velocity. For a specific contour of velocity $v_a$ and radius $r$ from the orifice (highlighted as the blue-red aliasing boundary on color Doppler), the instantaneous flow rate $Q$ is given by $Q=2\pi r v_a$. However, relatively low-velocity flow across large orifices such as prosthetic valves causes flattening of the isochromes and underestimation of flow, but this can be countered by multiplying $Q$ by $v/(v-v_a)$, where $v$ is the peak transorifice velocity.

$$Q = 2\pi r v_a v/(v-v_a).$$ (1)

The isovelocity surface area can also be calculated by the segment-of-spheroid (SOS) method using the chord (p) from the zenith of the contour to its outer edge (Figure 1):

$$Q = \pi p v_a.$$ (2)

EOA then is given by $Q/v$, where $v$ is the transorifice velocity. Although localized high velocities and pressure recovery have been shown in the central orifice of the St Jude valve, we recently demonstrated that pressure recovery across the lateral orifices of mitral prostheses is limited. Because most flow passes through the side orifices, these were subsequently used in the EOA and stroke volume calculations.
In Vitro Models

Steady-Flow Model
This in vitro flow model was described in detail previously10: a Plexiglas model with 2 chambers (proximal, $28 \times 20 \times 8.5$ cm; distal, 90 cm long) divided by a septum with a mount for prosthetic heart valves. St Jude heart valves (3 each of 23, 25, 27, 29, and 31 mm) were mounted with the leaflets oriented vertically to eliminate gravity effect. Flow, a 1% to 2% aqueous suspension of cornstarch, entered the proximal chamber from an upper reservoir whose height could be adjusted to vary transorifice pressure and flow rate and then passed through the mounted prosthetic valve under constant hydrostatic pressure causing steady flow. Flow rate was measured by the average of 3 to 5 timed collections. At least 3 different flow rates were examined for each valve size. Measurement variability was expressed as mean percent error for all flow rates studied, given by the SD of a set of timed collection measurements divided by their mean. Flow rates were chosen to simulate normal mitral transprosthetic velocities.

Pulsatile-Flow Model
To examine the effects of flow pulsatility on our method, St Jude heart valves (23, 25, 27, 29, and 31 mm) were studied in triplicate in the mitral position of a left heart pneumatic pump model. Cardiac output was measured by timed collections, with variability expressed as mean percent error. Four cardiac outputs were examined for each prosthesis, with heart rate constant at 70 bpm.

Echocardiographic Study
A Hewlett-Packard Sonos 1500 system was used with 2.5- or 3.5-MHz phased-array transducers held by an adjustable-clamp system to yield an imaging plane perpendicular to the leaflets to show 3 distinct orifices. Flow velocities across the center and side orifices were obtained by CW Doppler during suspended respiration, and the pressure half-time ($t_{1/2}$) of the E wave was measured. Images of the proximal convergence zone were obtained with a color aliasing velocity between 17 and 26 cm/s, reducing sector size to maximize frame rate (generally 18 to 22 frames per second), and stored on optical disks and/or VHS videotape.

Clinical Study
Patients
Patients in regular rhythm without significant tricuspid regurgitation who were undergoing St Jude mitral valve replacement were considered for the study. Cardiac output was measured in triplicate by thermodilution. The size of the implanted prosthetic heart valve was recorded.

Echocardiographic Examination
Intraoperative transesophageal echocardiography was performed with HP Sonos 1500 or Acuson 128XP with 3-MHz probes when patients were stable after weaning from cardiopulmonary bypass. Images of the prosthesis showing 3 distinct orifices were obtained at a depth of 6 to 10 cm. Velocities across the center and side orifices were obtained by CW Doppler during suspended respiration, and the pressure half-time ($t_{1/2}$) of the E wave was measured. Images of the proximal convergence zone were obtained with a color aliasing velocity between 17 and 26 cm/s, reducing sector size to maximize frame rate (generally 18 to 22 frames per second), and stored on optical disks and/or VHS videotape. Thermodilution cardiac output was obtained simultaneously.

Data Analysis and Calculations

Steady-Flow Model
We selected 5 frames with a clear blue-red aliasing boundary to measure $r$ and $p$, assuming the valve orifice to be at the prosthetic annular level. Prosthetic EOA was given by $Q_c/v$ for the hemispheric proximal isovelocity surface area (PISA) method and by $Q_p/v$ for the SOS method.

Pulsatile-Flow Model
Stroke volume was calculated by multiplying $Q_c$ or $Q_p$ by the time-velocity integral normalized by $v$. Cardiac output was given by the stroke volume $\times 70$ bpm. Prosthetic EOA was given by (1) $Q_c/v$ for the PISA method, (2) $Q_p/v$ for the SOS method, and (3) the modified Gorlin equation:

$$EOA = \frac{cardiac\ output/\ heart\ rate/diastolic\ filling\ period}{51.6 \sqrt{\Delta p}}.$$  

where $\Delta p$ is the mean gradient across the St Jude side orifice by CW Doppler.

Clinical Study
Forward stroke volume across the prosthetic orifice was calculated by multiplying $Q_p$ or $Q_c$ by the time-velocity integral normalized by $v$. The product of stroke volume and heart rate yielded cardiac output. Prosthetic EOA was calculated by (1) $Q_c/v$ (PISA), (2) $Q_p/v$ (SOS), (3) $220/t_{1/2}$ (pressure half-time), and (4) the modified Gorlin equation above. In both the clinical and in vitro studies, the localized
high velocities in the small central orifice were disregarded and the side orifice velocities used in the calculations. 25

Statistical Analysis

In Vitro Models
Flow rates and cardiac outputs calculated by (1) PISA and (2) SOS were each compared with timed collections by linear regression, with the difference between calculated and measured flow (\(DQ\)) expressed as mean ± SD. These 3 measurements of flow were also compared by repeated-measures ANOVA. Calculated in vitro EOAs were reported as mean ± SD for each size, and the triplicate prostheses were compared by ANOVA. Center and side orifice velocities were compared by paired Student’s \(t\) test. To evaluate the impact of aliasing velocities on flow estimation in the pulsatile model, \(DQ\) for 3 ranges of \(v_a\) (≤20, 21 to 29, and ≥30 cm/s) were compared by ANOVA, with \(DQ\) for PISA and SOS in each range compared by paired Student’s \(t\) test. \(DQ\) by PISA and SOS were also correlated with \(v_a\) by linear regression.

Clinical Study
Cardiac outputs calculated by (1) PISA and (2) SOS were each compared with thermodilution by linear regression, with the difference between Doppler and thermodilution cardiac output expressed as mean ± SD. These 3 cardiac output measurements were also compared by repeated-measures ANOVA. In vivo EOA (reported for each valve size as mean ± SD) calculated by (1) PISA, (2) SOS, and (3) pressure half-time method were each compared with Gorlin calculations and the geometric orifice area. Center and side orifice velocities were compared by paired Student’s \(t\) test, with the difference expressed as mean ± SD. Statistical significance was defined as a 2-tailed \(P<0.05\).

Interobserver and Intraobserver Variability
Ten randomly selected color Doppler images and continuous Doppler recordings from the in vitro and clinical studies were used to assess interobserver and intraobserver variability in measurement of

### TABLE 1. EOA (Mean±SD) of St Jude Valves in the Steady Flow Model

<table>
<thead>
<tr>
<th>Valve Size, mm</th>
<th>EOAPISA, cm²</th>
<th>EOASOS, cm²</th>
<th>GOA, cm²</th>
</tr>
</thead>
<tbody>
<tr>
<td>23</td>
<td>1.5±0.07</td>
<td>1.33±0.03</td>
<td>2.55</td>
</tr>
<tr>
<td>25</td>
<td>2.09±0.22</td>
<td>1.87±0.19</td>
<td>3.09</td>
</tr>
<tr>
<td>27</td>
<td>2.56±0.35</td>
<td>2.17±0.32</td>
<td>3.67</td>
</tr>
<tr>
<td>29</td>
<td>2.73±0.29</td>
<td>2.52±0.43</td>
<td>4.41</td>
</tr>
<tr>
<td>31</td>
<td>3.12±0.21</td>
<td>3.12±0.3</td>
<td>5.18</td>
</tr>
</tbody>
</table>

\(EOAPISA\) indicates effective orifice area by the hemispherical proximal convergence method; \(EOASOS\), effective orifice area by the segment-of-spheroid method; and \(GOA\), geometric orifice area supplied by the manufacturer.

### TABLE 2. EOA (Mean±SD) of St Jude Valves in the Pulsatile Flow Model

<table>
<thead>
<tr>
<th>Valve Size, mm</th>
<th>EOAPISA</th>
<th>EOASOS</th>
<th>EOAGorlin</th>
<th>GOA</th>
</tr>
</thead>
<tbody>
<tr>
<td>23</td>
<td>1.13±0.06</td>
<td>0.93±0.06</td>
<td>1.04±0.02</td>
<td>2.55</td>
</tr>
<tr>
<td>25</td>
<td>1.64±0.19</td>
<td>1.53±0.19</td>
<td>1.37±0.1</td>
<td>3.09</td>
</tr>
<tr>
<td>27</td>
<td>2.22±0.38</td>
<td>2.21±0.38</td>
<td>1.84±0.33</td>
<td>3.67</td>
</tr>
<tr>
<td>29</td>
<td>2.58±0.35</td>
<td>2.39±0.33</td>
<td>2.19±0.23</td>
<td>4.41</td>
</tr>
<tr>
<td>31</td>
<td>2.61±0.51</td>
<td>2.41±0.33</td>
<td>2.35±0.24</td>
<td>5.18</td>
</tr>
</tbody>
</table>

\(EOAGorlin\) indicates effective orifice area by the Gorlin equation. Other abbreviations as in Table 1. All values are in cm² unless specified.
the radius $r$ and chord $p$ of the proximal convergence. Variability was expressed as the ratio of the difference between the 2 measurements to their mean.$^6$

**Results**

**In Vitro Models**

**Steady-Flow Model**

Flow rates ranged from 218 to 406 cm$^3$/s, with mean error in the timed collections of 1.7±1%. The peak velocities through the center orifices (153.9±39.6 cm/s; range, 82.9 to 245 cm/s) were significantly higher than that through the side orifices (133.1±32.4 cm/s; range, 73.5 to 190 cm/s, $\Delta v=20.9\pm10.9$ cm/s, $P<0.001$). The ratio of the side to center velocities was 0.87±0.04. PISA and SOS flow rates correlated well with the timed collections (Figures 2 and 3), but compared with PISA, SOS underestimated true flow rate ($\Delta Q=-29.8\pm42.3$ versus 0.7±28.2 cm$^3$/s, $P<0.001$), also significant by repeated-measures ANOVA ($P<0.0001$). PISA and SOS EOAs ($y$) correlated with but underestimated the geometric orifice area ($x$) significantly (PISA: $y=0.52x+0.43$, $r=0.87$, $P<0.001$, $\Delta y=-1.47\pm0.50$ cm$^2$; SOS: $y=0.61x-0.09$, $r=0.87$, $P<0.001$, $\Delta y=-1.6\pm0.45$ cm$^2$, Table 1). There was no significant variation in EOAs for the 3 sets of prostheses by ANOVA.

**Pulsatile-Flow Model**

Cardiac output ranged from 1.6 to 8.6 L/min, with a mean of 5.9 L/min and variation in the timed collections of 2.2±1.3%. The ratio of side to center orifice velocity was 0.83±0.11. The cardiac output calculated by PISA agreed closely with timed cardiac output ($y=0.997x-0.09$, $r=0.95$, $P<0.001$, $\Delta Q=-0.11\pm0.59$ L/min, $P=NS$), but SOS showed poorer correlation ($y=0.81x+0.52$, $r=0.88$, $P<0.001$) and significant underestimation of flow ($\Delta Q=-0.55\pm0.82$ L/min, $P<0.001$), significantly worse ($P<0.0001$) than the PISA calculations. PISA and SOS EOAs correlated with but underestimated geometric orifice areas ($r=0.79$ and 0.78, respectively, Table 2), without significant variation among the 3 sets of prostheses by ANOVA.

By SOS, $\Delta Q$ was $-0.88\pm0.7$ L/min for aliasing velocities $\leq 20$ cm/s, $-0.49\pm0.9$ L/min for 21 to 29 cm/s, and $-0.28\pm0.8$ L/min for $\geq 30$ cm/s ($P=0.07$ by ANOVA). SOS $\Delta Q$ was significantly worse than PISA $\Delta Q$ for $\leq 20$ cm/s ($-0.05\pm0.4$ L/min, $P<0.001$) and 21 to 29 cm/s ($0.12\pm0.3$ L/min, $P=0.007$) but not for $\geq 30$ cm/s ($-0.38\pm0.8$ L/min, $P=NS$). There were opposite but nonsignificant linear trends between aliasing velocity and $\Delta Q$ for PISA ($r=-0.24$, $P=NS$) and SOS ($r=0.24$, $P=NS$).

**Clinical Study**

The study population comprised 36 patients (26 women, 58±11 years old). At the time of study, 24 were AV paced, with the remainder in sinus rhythm. Heart rate was 92±10 bpm.

The peak and mean velocities through the center orifice (161±29 and 97±16 cm/s, respectively) were significantly higher than through the side orifices (137±23 and 86±14 cm/s, respectively, $\Delta_{peak}=23.6\pm11$ cm/s, $P<0.001$ and $\Delta_{mean}=11\pm6.5$ cm/s, $P<0.001$), with a side-to-central velocity ratio of 0.86±0.05 (Figure 4).

Cardiac output by thermodilution ($x$) was 5.2±1.2 L/min (3.3 to 8.1 L/min), with close agreement by PISA ($y=1.018x-0.21$, $r=0.91$, $\Delta CO=-0.05\pm0.55$ L/min, Figure 5) but underestimation by SOS ($y=0.79x-0.21$, $r=0.82$, $\Delta CO=-1.33\pm0.73$ L/min, Figure 6), $P<0.001$ for the significance of this difference by repeated-measures ANOVA ($P<0.001$).
Table 3 and Figure 7 summarize EOA calculated by the different methods. EOA by the modified Gorlin equation was significantly smaller than the geometric orifice area provided by the manufacturer \( y = 0.37x + 0.24, r = 0.64, P < 0.001, \Delta = -2.5 \pm 0.5 \text{ cm}^2, P < 0.001 \). PISA EOA agreed closely with Gorlin valve area \( y = 0.78x + 0.7, r = 0.75, P < 0.001, \Delta = 0.3 \pm 0.28 \text{ cm}^2, P < 0.001 \) and also underestimated the geometric area \( y = 0.48x + 0.05, r = 0.8, P < 0.001, \Delta = -2.15 \pm 0.43 \text{ cm}^2, P = 0.001 \). Similarly, SOS EOA agreed with Gorlin area \( y = 0.64 + 0.41, r = 0.68, P < 0.001, \Delta = -0.23 \pm 0.3 \text{ cm}^2, P < 0.001 \) and underestimated the geometric area \( y = 0.42x - 0.23, r = 0.76, P < 0.001, \Delta = -2.7 \pm 0.46 \text{ cm}^2 \). Orifice area calculated by pressure half-time showed very poor (inverse) agreement with both Gorlin \( y = 4.9 - 1.01x, r = -0.44 \) and geometric valve area \( y = -0.44x, r = -0.33 \).

Table 3. In Vivo EOA (Mean ± SD) of St Jude Mitral Valves

<table>
<thead>
<tr>
<th>Valve size, mm</th>
<th>EOAPISA</th>
<th>EOASOS</th>
<th>EOAT1/2</th>
<th>EOAGorlin</th>
<th>GOA</th>
</tr>
</thead>
<tbody>
<tr>
<td>25 (n=4)</td>
<td>1.52±0.11</td>
<td>1.12±0.18</td>
<td>3.92±0.9</td>
<td>1.38±0.16</td>
<td>3.09</td>
</tr>
<tr>
<td>27 (n=10)</td>
<td>1.75±0.14</td>
<td>1.28±0.19</td>
<td>3.35±0.86</td>
<td>1.53±0.26</td>
<td>3.67</td>
</tr>
<tr>
<td>29 (n=14)</td>
<td>2.25±0.18</td>
<td>1.61±0.29</td>
<td>2.79±0.74</td>
<td>1.9±0.28</td>
<td>4.41</td>
</tr>
<tr>
<td>31, 33 (n=8)</td>
<td>2.46±0.42</td>
<td>1.95±0.26</td>
<td>2.93±0.99</td>
<td>2.09±0.43</td>
<td>5.18</td>
</tr>
</tbody>
</table>

EOA T1/2 indicates effective orifice area by the pressure half-time method. Other abbreviations as in previous tables. All valve areas are in cm² unless specified.

**Interobserver and Intraobserver Variabilities**

In vitro interobserver variabilities in the measurement of proximal convergence radius and chord were 3.3 ± 3.8% and 3.9 ± 2.9%, respectively, with intraobserver variabilities of 3.5 ± 1.4% and 2.1 ± 1.1%, respectively. In the clinical study, interobserver variabilities for r and p were 3.6 ± 3.2% and 3.2 ± 2.4%, respectively, with intraobserver variabilities of 4.1 ± 3.4% and 5.3 ± 2.9%, respectively.

**Discussion**

The proximal convergence method, a recently developed color Doppler technique for quantification of mitral regurgi-
Effective Prosthetic Orifice Area

All prosthetic valves are inherently mildly stenotic. Analogous to stenotic native valve area, prosthetic orifice area, incorporating flow and pressure gradient, gives a more flow-independent measure of prosthetic resistance. Although the continuity equation and pressure half-time methods\(^{8,29}\) are accepted Doppler techniques for native mitral and aortic stenosis, effective prosthetic orifice area has been more elusive.\(^{30}\) Although Doppler has been used in flow models to calculate effective prosthetic valve areas\(^{31}\) and the continuity equation has been used clinically for prosthetic aortic valves,\(^{32}\) a reliable noninvasive method to estimate prosthetic mitral orifice area clinically is still lacking.

The present study demonstrated that the pressure half-time method has little value in measuring effective prosthetic orifice areas, actually showing an inverse relation to valve size. We have previously shown that the pressure half-time method is unreliable immediately after balloon valvuloplasty\(^{33}\) because of abrupt changes in transvalvular gradient and chamber compliance. In the present study, the pressure half-time was measured immediately after mitral valve replacement, when sudden changes in net chamber compliance and transvalvular gradient might have contributed to the inaccuracy of the pressure half-time method for effective prosthetic valve area.

The effective prosthetic valve areas in our clinical study are consistent with some published studies\(^{31}\) but smaller than others.\(^{34-36}\) Baumgartner et al\(^{31}\) calculated EOA in a pulsatile-flow model with the Gorlin formula and Doppler gradients and obtained values similar to our in vivo EOA. However, in the same study, EOAs were significantly larger when catheter gradients were used with the Gorlin formula. Similarly, Yoganathan and his group obtained significantly larger EOAs with the catheter-based Gorlin formula both in vitro\(^{37}\) and in vivo,\(^{38}\) related to the overestimation of catheter gradients by Doppler\(^{39}\) due to the phenomenon of pressure recovery.\(^{25}\)

Interestingly, Baumgartner et al observed larger EOAs in their steady in vitro model than in the pulsatile one. This is supported by a preliminary study by Trujillo et al\(^{37}\) who showed that for rigid circular orifices, Doppler continuity EOAs during steady flow are consistently larger than in pulsatile-flow conditions. Further preliminary data show flow dependency of EOA in pulsatile conditions.\(^{37,38}\) In our clinical study, the relatively fast heart rate and therefore lower stroke volume observed just after cardiopulmonary bypass may contribute to smaller calculated EOAs. In addition, the geometry surrounding the prosthesis in vivo and the presence or absence of the mitral subvalvular apparatus may also have important effects on the calculated EOA.

Limitations

The major limitation in all studies examining effective prosthetic orifice area is the lack of a proper, universally accepted “gold standard.” Various references have been used, including geometric, Gorlin, continuity, and half-time EOAs. However, true in vivo EOA may vary for a given prosthetic depending on physiological conditions, shown by the range in reported EOA in our and prior studies. In vitro results suggest that flow pulsatility affects EOA. Our data showed consistent results across 3 valves of each size, suggesting relatively little actual interprosthesis variance.

Clinical Application

Although we would not recommend this technique for routine assessment of mitral prostheses (because of the need for transesophageal echocardiography), we have found it quite useful in relatively subtle or questionable situations of prosthetic obstruction. When a leaflet is completely stuck by thrombus or pannus, this is usually evident by direct 2-dimensional imaging and a high transprosthetic gradient in the presence of a low-output state. In contrast, the PISA EOA is helpful for assessing nonobvious obstruction or, conversely, ruling it out when a high gradient occurs in a high-output state.

Conclusions

This is the first study to demonstrate that mitral prosthetic EOA can be measured noninvasively by proximal flow convergence methods. We examined the St Jude prosthesis because it is the mostly commonly implanted prosthetic valve; because its hydrodynamic profile is among the most complex, this technique is likely to be applicable to other types of prostheses. This may improve the management of patients with mitral prostheses by allowing accurate detection and assessment of prosthetic valve obstruction independent of transvalvular flow.

Acknowledgments

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References


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