Doppler Echocardiographic Assessment of the St. Jude Medical Prosthetic Valve in the Aortic Position Using the Continuity Equation

Edward R. Chafizadeh, MD, and William A. Zoghbi, MD

To test whether the continuity equation can be applied to the noninvasive assessment of prosthetic aortic valve function, Doppler echocardiography was performed in 67 patients (mean age, 58±14 years) within 10±6 days after valve replacement with St. Jude Medical valves. All patients were clinically stable and without evidence of valve dysfunction. Valve size ranged from 19 to 31 mm, and ejection fraction ranged from 30% to 75%. With the parasternal long-axis view, the left ventricular outflow diameter measured just proximal to the prosthetic valve correlated well with valve size (r=0.92). Doppler-derived maximal gradients ranged from 9 to 71 mm Hg. Effective prosthetic aortic valve area by the continuity equation ranged between 0.73 cm² for a 19-mm valve and 4.23 cm² for a 31-mm valve. With analysis of variance, effective orifice area differentiated various valve sizes (p<10⁻¹⁴) better than did gradients alone (p=0.003) and correlated better with actual valve orifice area (r=0.83 versus -0.40). A Doppler velocity index, the ratio of peak velocity in the left ventricular outflow to that of the aortic jet, averaged 0.41±0.09 and was less dependent on valve size (r=0.43). Thus, the continuity equation can be applied to the assessment of prosthetic St. Jude valves in the aortic position. By accounting for flow through the valve, it provides an improved assessment over the sole use of gradients in the evaluation of prosthetic valve function. (Circulation 1991;83:213–223)

Ultrasound imaging, using M-mode or two-dimensional echocardiography, has been limited in the evaluation of prosthetic valve function. Recent Doppler echocardiography has significantly improved the noninvasive evaluation of prosthetic valves with the estimation of gradients across the valve using the simplified Bernoulli equation. However, the sole use of gradients is limited in the assessment of prosthetic valve function because gradients are dependent on flow in addition to valve type and size. This explains in part the significant overlap observed among blood velocities and gradients across normally functioning valves.

Recently, the use of the continuity equation, which accounts for flow through the valve, has improved the assessment of stenotic native valves by Doppler echocardiography with the estimation of a valve area. We therefore undertook this study to test whether the continuity equation can be applied to the evaluation of normal prosthetic valves in the aortic position by the estimation of an effective orifice area. Results were compared with actual orifice areas of the prosthetic valves. The left ventricular outflow (LVO) was used as the site for determination of flow as mean blood velocity multiplied by the cross-sectional area of the LVO. Because of the potential errors and limitations inherent to the measurement of LVO area in the presence of a prosthetic aortic valve, LVO area was determined by two approaches: first, from echocardiographic measurements, and second, by using the sewing ring area of the prosthetic valve as the LVO area. The relative merits of these two approaches for calculation of effective orifice areas were assessed. Furthermore, a Doppler velocity index, the ratio of peak velocity in the LVO to that of the aortic jet, was derived. A major advantage of this index is that it does not rely on determination of LVO area.

Methods

Patient Population

The patient population comprised 78 patients who were admitted to The Methodist Hospital between
1984 and 1988 and underwent an aortic valve replacement with a St. Jude Medical valve and two-dimensional and Doppler echocardiographic studies within 1 month after surgery. The St. Jude Medical valve was chosen because over the past few years, this type of valve was inserted in the aortic position in the majority of cases in our institution. Exclusions from the study included patients with technically difficult echocardiographic and/or Doppler studies (n=6) or previous or concurrent mitral valve replacement (n=5). The remaining 67 patients constituted the study population. All patients were hemodynamically stable and without clinical or Doppler evidence of valve dysfunction. Twenty of the 67 patients had mild prosthetic valve regurgitation as assessed by pulsed, continuous wave Doppler and, after 1986, by color flow Doppler.22-26 There were 41 men and 26 women, with an age range of 24–81 years (mean age, 58±14 years). Forty patients were in sinus rhythm, and 27 were in atrial fibrillation. The average time from valve replacement to Doppler echocardiographic study was 10±6 days (range, 3–29 days). Left ventricular ejection fraction as measured by two-dimensional echocardiography ranged from 30% to 75% (mean, 54±6%).27

Two-Dimensional Echocardiographic Measurements

Two-dimensional echocardiographic studies were obtained using either a Hewlett-Packard or an Ultrasonographic sector scanner equipped with 3.5- or 2.5-MHz transducers. Studies were recorded on ½-in. videotape and were reviewed on an off-line station equipped with internal calipers and interfaced with the video signal (Digisonics EC500, Houston). A search module allowed bidirectional frame-by-frame playback of the video images. All echocardiographic and Doppler measurements were performed without knowledge of the prosthetic valve size. Using the parasternal long-axis view, measurement of the aortic annulus diameter was performed in early systole using the inner-edge-to-inner-edge method previously described by researchers from our laboratory for normal or stenotic native aortic valves (Figure 1).19,28 Care was taken to exclude the bright echoes that occasionally trail in the LVO from the prosthetic valve. For each patient, an average of the largest three diameters was made and used to derive the cross-sectional area of the LVO by echocardiography (πD²/4).

Pulsed Doppler Measurements

Pulsed Doppler studies were performed with the same imaging systems, which are equipped with movable cursors and adjustable sample volume sizes. Blood flow velocity in the LVO was recorded from the apical window as previously described, using a sample volume size of 5 mm and a sweep of 100 mm/sec.19 Similar to native aortic stenosis, the sample volume was placed into the prosthetic valve leaflets and gradually moved apically until a clear spectral display was observed. This usually occurred approximately 0.5 cm upstream from the valve. As expected, further apical placement of the sample volume resulted in a decrease in velocity.

The systolic velocity curves were digitized along the external contour of the darkest portion of the spectral display as previously described.19,28 These
curves were used to derive LVO time–velocity integral (TVI jet) (cm) and peak flow velocity (PKV LVO) (m/sec) (Figure 1). Heart rate was determined from the preceding R-R interval on the electrocardiogram. Cardiac output by Doppler was derived as the product of cross-sectional area of the LVO, time–velocity integral, and heart rate. An average of three cardiac cycles was used for patients in sinus rhythm, whereas an average of at least five cardiac cycles was used in atrial fibrillation.

Continuous Wave Doppler Measurements

Continuous wave Doppler recordings of the jet velocity through the prosthetic valve were made using a 2-MHz nonimaging transducer equipped with audio and spectral displays. In each patient, recording of the jet velocity was attempted from multiple windows including the apical, right parasternal, suprasternal, and subcostal approaches. The window providing the highest jet velocity was used for measurements because it implies the least angle between the Doppler beam and the jet. Cardiac cycles with the highest peak velocities were used for analysis. The spectral display was digitized along its outer border, from which the following measurements were derived: 1) peak jet velocity (PKV jet) (m/sec); 2) maximal gradient (mm Hg) derived using the simplified Bernoulli equation as 4(PKV jet² − PKV LVO²) and also simply as 4(PKV jet), ignoring the correction from the usually low subvalvular velocity; 3) mean gradient; and 4) TVI jet.

An average measurement from three cardiac cycles in sinus rhythm and at least five cycles in atrial fibrillation was obtained. Furthermore, in patients with atrial fibrillation, the preceding R-R intervals of the continuous wave Doppler tracings used for quantitation were matched to those of the pulsed Doppler cycles so the heart rates of pulsed and continuous wave Doppler recordings were within 10% of each other. This was performed to further decrease the errors introduced from mismatch of flow during an irregular rhythm.

Specifications of St. Jude Medical Valve

The St. Jude Medical valve is an all-pyrolytic carbon bileaflet valve (Figure 2). The valve size denotes the sewing ring diameter, which is larger than the actual orifice diameter. Table 1 lists the reported dimensions and measured areas of the St. Jude valve from the 19-mm to the 31-mm valve size. The actual orifice diameter is approximately 4.5 mm smaller than the sewing ring diameter (4.3 mm for a 19-mm valve and 5.0 mm for a 31-mm valve) (Table 1). Because of the cross-sectional area of the metallic leaflets and other structures, the actual orifice area is slightly smaller than that derived from the actual orifice diameter assuming a circular geometry.
TABLE 1. Valve Specifications and Echocardiographic and Doppler Data in 67 Normal St. Jude Medical Valves in the Aortic Position

<table>
<thead>
<tr>
<th>Valve size (mm)</th>
<th>Actual orifice diameter (mm)</th>
<th>D_{LVO} (cm)</th>
<th>PkV_{jet} (m/sec)</th>
<th>Mean gradient (mm Hg)</th>
<th>Velocity index</th>
<th>EOA_{ECHO} (cm²)</th>
<th>EOA_{SR} (cm²)</th>
<th>AOA (cm²)</th>
</tr>
</thead>
<tbody>
<tr>
<td>19</td>
<td>9</td>
<td>14.7</td>
<td>1.85±0.07</td>
<td>3.0±0.6</td>
<td>17±7</td>
<td>0.37±0.07</td>
<td>0.99±0.20</td>
<td>1.04±0.19</td>
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<tr>
<td></td>
<td></td>
<td>(1.77–1.97)</td>
<td>(2.2–4.2)</td>
<td>(8–30)</td>
<td>(0.26–0.47)</td>
<td>(0.73–1.32)</td>
<td>(0.73–1.32)</td>
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<tr>
<td>21</td>
<td>14</td>
<td>16.7</td>
<td>2.00±0.04</td>
<td>2.7±0.3</td>
<td>14±5</td>
<td>0.40±0.06</td>
<td>1.25±0.21</td>
<td>1.38±0.22</td>
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<tr>
<td></td>
<td></td>
<td>(1.90–2.07)</td>
<td>(2.5–3.5)</td>
<td>(9–27)</td>
<td>(0.29–0.55)</td>
<td>(0.92–1.73)</td>
<td>(1.02–1.91)</td>
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<tr>
<td>23</td>
<td>16</td>
<td>18.5</td>
<td>2.10±0.12</td>
<td>2.8±0.5</td>
<td>16±6</td>
<td>0.37±0.06</td>
<td>1.28±0.31</td>
<td>1.52±0.26</td>
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<td></td>
<td></td>
<td>(1.97–2.33)</td>
<td>(2.0–3.8)</td>
<td>(8–29)</td>
<td>(0.28–0.51)</td>
<td>(0.93–2.20)</td>
<td>(1.15–2.14)</td>
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<tr>
<td>25</td>
<td>14</td>
<td>20.4</td>
<td>2.32±0.15</td>
<td>2.6±0.5</td>
<td>13±6</td>
<td>0.42±0.08</td>
<td>1.80±0.41</td>
<td>2.08±0.41</td>
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<td></td>
<td></td>
<td>(2.0–2.53)</td>
<td>(1.8–3.7)</td>
<td>(6–28)</td>
<td>(0.27–0.61)</td>
<td>(1.20–2.76)</td>
<td>(1.35–3.00)</td>
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<tr>
<td>27</td>
<td>6</td>
<td>22.3</td>
<td>2.58±0.20</td>
<td>2.2±0.5</td>
<td>11±5</td>
<td>0.46±0.10</td>
<td>2.43±0.63</td>
<td>2.65±0.58</td>
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<td></td>
<td>(2.27–2.80)</td>
<td>(1.8–3.1)</td>
<td>(6–20)</td>
<td>(0.33–0.60)</td>
<td>(1.54–3.15)</td>
<td>(1.89–3.42)</td>
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<tr>
<td>29</td>
<td>5</td>
<td>24.1</td>
<td>2.63±0.06</td>
<td>2.0±0.1</td>
<td>7±1</td>
<td>0.49±0.04</td>
<td>2.66±0.26</td>
<td>3.23±0.30</td>
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<tr>
<td></td>
<td></td>
<td>(2.57–2.73)</td>
<td>(1.8–2.1)</td>
<td>(6–9)</td>
<td>(0.43–0.55)</td>
<td>(2.26–2.97)</td>
<td>(2.82–3.61)</td>
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<tr>
<td>31</td>
<td>3</td>
<td>26.0</td>
<td>2.83±0.06</td>
<td>2.1±0.6</td>
<td>10±6</td>
<td>0.49±0.19</td>
<td>3.08±1.09</td>
<td>3.72±1.40</td>
</tr>
<tr>
<td></td>
<td></td>
<td>(2.80–2.90)</td>
<td>(1.5–2.7)</td>
<td>(5–16)</td>
<td>(0.31–0.69)</td>
<td>(2.07–4.23)</td>
<td>(2.36–5.18)</td>
<td></td>
</tr>
</tbody>
</table>

Values are given as mean±SD; ranges of values are in parentheses.

D_{LVO}, diameter of left ventricular outflow measured by echocardiography; PkV_{jet}, peak velocity of aortic jet; EOA_{ECHO} and EOA_{SR}, effective orifice areas using continuity equation, where cross-sectional area of ventricular outflow was measured by echocardiography or determined from sewing ring area, respectively; AOA, actual orifice area.

Assessment of Prosthetic Valves by Continuity Equation

The application of the continuity equation\textsuperscript{19,20} to prosthetic valves in the aortic position yields

EOA×V_{jet}=CSA_{LVO}×V_{LVO} \tag{1}

where EOA is the effective orifice area of the prosthetic valve (cm²), CSA_{LVO} is the cross-sectional area of the LVO tract (cm²), and V_{LVO} and V_{jet} are the mean spatial blood velocities (m/sec) in the outflow tract and through the prosthetic valve, respectively. The equation is valid provided that the flow velocity profile in the LVO is similar to that of the prosthetic valve.\textsuperscript{20}

Because of the potential errors and limitations inherent to measurement of CSA_{LVO} by echocardiography, especially in the presence of a prosthetic aortic valve, several approaches to the use of the continuity equation were tested.

Doppler velocity index. Rearranging the continuity equation and applying the simplified peak velocity method, previously validated in our laboratory\textsuperscript{19} for native aortic valves, the Doppler velocity index was derived as

\text{Doppler velocity index=}

\frac{PkV_{LVO}}{PkV_{jet}} = \frac{EOA}{CSA_{LVO}} \tag{2}

This peak velocity ratio is dimensionless and does not rely on determination of cross-sectional area of the LVO.

Effective prosthetic valve area. By integrating Equation 1 over the duration of systole, the effective orifice area can be calculated as

EOA=CSA_{LVO} \times \frac{TVI_{LVO}}{TVI_{jet}} \tag{3}

where TVI_{LVO} and TVI_{jet} are the time–velocity integrals at the LVO and of the aortic jet, respectively. In addition, valve area was determined by the simplified peak velocity method\textsuperscript{19} as

EOA=CSA_{LVO} \times \frac{PkV_{LVO}}{PkV_{jet}} \tag{4}

The cross-sectional area of the LVO in Equations 3 and 4 was determined by two approaches: first, by two-dimensional echocardiography as described above under echocardiographic measurements, and second, by assuming that the sewing ring area of the prosthetic valve is the cross-sectional area of the LVO.

Interobserver and Intraobserver Variability

To assess the interobserver and intraobserver variability of the various measurements, determinations of LVO diameter, pulsed and continuous wave Doppler parameters, were repeated by the same and another independent observer in a subgroup of 13 patients. Variability was expressed as mean percent error, calculated as the absolute difference between the two observations divided by the mean of the observations and expressed as percent.

Statistical Analysis

Values are given as mean±SD. Correlations between valve sizes and echocardiographic and Doppler parameters were performed using linear and/or second-degree polynomial regression analysis where appropriate. Mean values of echocardiographic parameters were compared using paired t tests. Analysis of variance was used to determine whether the Doppler
echocardiographic variables were different among various valve sizes. If the \( F \) value was significant, Duncan’s multiple comparison test was performed to assess differences between individual groups.\(^{30}\)

**Results**

Table 1 lists the distribution of valve sizes and the results of echocardiographic and Doppler measurements. In the 67 patients studied, the size of the St. Jude valves ranged from 19 to 31 mm and included all intermediate sizes. The 20 patients with mild aortic insufficiency were proportionally distributed among the various valve sizes (\( \chi^2 = 5.5; p=NS \)).

**Measurement of Left Ventricular Outflow Diameter**

A measurement of the LVO diameter just below the insertion of the prosthetic valve was possible in all patients in whom overall imaging quality was adequate. Reverberations from the prosthesis interfered minimally with this determination because they usually occurred within the areas of the aortic root and left atrium when imaging from the parasternal window. The LVO diameter ranged from 1.9 to 2.9 cm. A good correlation was observed between measured LVO diameters and known sewing ring diameters with a correlation coefficient of 0.92 (Figure 3). LVO diameters were generally smaller than sewing ring diameters (mean, 2.21±0.29 versus 2.36±0.32 cm, respectively; \( p=0.0001 \)). This finding was also reflected in the plot and regression equation in Figure 3.

**Pulsed Doppler Measurements**

In the LVO tract, the peak blood flow velocity ranged between 0.82 and 1.37 m/sec (mean, 1.04±0.11 m/sec), and the time-velocity integral ranged between 9.4 and 26.8 cm (mean, 16.5±3.4 cm). Results were similar in patients with and without mild aortic insufficiency. Cardiac output by Doppler averaged 5.34±1.80 l/min.

**Doppler-Derived Maximal and Mean Gradients**

An adequate recording of the aortic jet velocity through the prosthetic valve was obtained in all patients. In the majority of cases (85%), the best jet recordings were obtained from the apical window. The peak velocity of the jet ranged from 1.53 to 4.21 m/sec; its distribution by valve size is given in Table 1. As expected from the observed normal range of velocities in the LVO, the maximal gradients derived without the correction for LVO velocity were only an average of ±1 mm Hg (range, 2–8 mm Hg) higher than those with correction for subvalvular velocity. Table 1 and Figure 4 show the derived maximal gradients without correction for LVO velocity and the calculated mean gradients for the various valve sizes. Maximal and mean gradients were similar in patients with and without mild aortic insufficiency. Although gradients tended to be smaller with larger valve size, a significant heterogeneity within each valve size and overlap in maximal as well as mean gradients were seen among the various valve sizes (Figure 4). A weak inverse linear relation was observed between valve size and maximal or mean gradient (\( r=-0.46 \) and \(-0.40 \), respectively). The relation between maximal gradient and valve size was not improved when correction for LVO velocity was performed (\( r=-0.45 \)). With analysis of variance, only maximal gradients of the 29- and 31-mm valves and

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**Figure 3.** Plot of relation between valve size or sewing ring diameter and left ventricular outflow diameter measured by echocardiography (D\(_{LVO}-\text{Echo} \)). Solid and dotted lines represent regression and identity lines, respectively. Numbers in parentheses are number of observations in each group.

**Figure 4.** Plot of distribution of maximal gradients (left panel) and mean gradients (right panel) grouped by size of St. Jude Medical valve. Bars, mean±SD of observations; ANOVA, analysis of variance.
mean gradients of 29-mm valves were significantly lower than those of other valves. The significance level of the discrimination was \( p = 0.003 \) for maximal gradients and \( p = 0.002 \) for mean gradients.

**Doppler Velocity Index**

The ratio of peak velocity in the LVO to the peak velocity of the aortic jet ranged between 0.26 and 0.69 (mean, 0.41±0.09). Individual data of this index, grouped by valve size, are plotted in Figure 5 and given in Table 1. With analysis of variance, a difference in the velocity index among various valve sizes was detected at \( p = 0.009 \). The velocity indexes of the 27-, 29-, and 31-mm valves were larger than those of valves 25 mm or smaller. A significant correlation was observed between the Doppler velocity index and valve size \( (r = 0.43, p = 0.0002) \).

**Effective Orifice Area by Doppler**

Using the diameter of the LVO measured by echocardiography, the effective orifice area \( (EOA_{ECHO}) \) derived with the continuity equation ranged between 0.74 and 4.28 cm\(^2\) and between 0.73 and 4.23 cm\(^2\) with the simplified peak velocity method. The smallest area was observed in a 19-mm valve, and the largest in a 31-mm valve. An excellent correlation was found between results by the simplified peak velocity method and those by the original continuity equation: \( r = 0.97 \), and \( y = 0.99x + 0.11 \) (Figure 6). Because of these results and the simplicity of the peak velocity method, the values reported subsequently will be those of the peak velocity method. The normal range and mean±SD of effective orifice areas grouped by valve size are shown in Table 1 and Figure 7. A similar mean±SD and range were observed for the various valve sizes when patients in atrial fibrillation were excluded. With analysis of variance, the determination of effective orifice areas discriminated well between various valve sizes \( (p < 0.01) \) (Figure 7). With linear regression analysis, effective orifice area correlated well with actual orifice area and, as expected, was always smaller \((r = 0.83; EOA_{ECHO} = 0.62 AOA - 0.10)\). The correlation was not improved \( (r = 0.83) \) when a curvilinear fit to the regression equation was attempted.

Effective orifice area derived using the sewing ring area \( (EOA_{SR}) \) and the peak velocity method ranged between 0.73 and 5.18 cm\(^2\). Analysis of variance revealed a very good discrimination between valve sizes \( (p < 10^{-17}; \text{Figure 7}) \). With linear regression analysis, effective orifice area correlated well with actual orifice area \( (r = 0.87, EOA_{SR} = 0.77 AOA - 0.29) \). The correlation coefficient was similar when a curvilinear fit to the regression analysis was attempted \( (r = 0.87) \). As expected from the relation of sewing ring diameters to measured LVO diameters, \( EOA_{SR} \) values were larger than \( EOA_{ECHO} \) values (mean, 1.87±0.83 versus 1.63±0.70 cm\(^2\); \( p < 0.0001 \)).

**Interobserver and Intraobserver Variability**

The results of intraobserver and interobserver variability for the various echocardiographic and Doppler measurements, expressed as mean percent error, are shown in Table 2.

**Discussion**

The present study demonstrates that the continuity equation can be applied to the assessment of St. Jude Medical valves in the aortic position and allows the differentiation of various sizes of normally functioning valves. This can be performed by either measuring the LVO diameter by echocardiography or using the sewing ring diameter of the valve. As in the case of native aortic valves, the simplified peak velocity method for the determination of effective areas yielded results similar to those obtained using planimetry of the time–velocity curves. A Doppler velocity index, derived as the ratio of maximal velocity upstream to that downstream from the valve, is less dependent on valve size and may...
add to the overall evaluation of prosthetic valve function by the Doppler technique, especially if the valve size is not known.

Doppler-Derived Gradients

Several studies have demonstrated that the application of the simplified Bernoulli equation to flow velocity measurements by Doppler across prosthetic valves or various forms of obstruction provides accurate estimates of pressure gradients. More recently, a simultaneous Doppler and catheter study in patients with various types of prosthetic valves comparing the simplified Bernoulli equation with dual-catheter measurement revealed very good correlations between gradients measured by Doppler and those measured by catheter. However, flow velocity as well as gradients across the valve depend on several factors, including valve size and type, and flow through the valve. Several studies have demonstrated that although an inverse relation exists between valve size and gradients, a significant overlap is present between gradients of a particular valve type and size as well as among various sizes of the same valve. The present Doppler study involving a large series of normal St. Jude valves in the aortic position further demonstrates the significant overlap of gradients between valves. With milder degrees of valve stenosis, as seen in the smaller-size prostheses, more overlap of gradients occurred, probably because of the greater dependence of gradient on flow conditions. In an effort to ensure normality of prosthetic valve function, we selected patients studied very soon after valve replacement. Therefore, many of these patients were still under the influence of a hyperdynamic state induced by the recent unloading of the left ventricle and the possible contribution of anemia. This may account for some of the high gradients observed, particularly in the smaller-size prostheses. However, the gradients observed in the present study are similar to those previously reported in aortic St. Jude valves within a similar time interval after valve replacement and are on the average 6 mm Hg or less higher than those in patients studied at a later time after surgery. Because of the normal range of blood velocity observed in the LVO, the difference in the present study between maximal gradients derived with and without correction for LVO velocity was small. However, with higher LVO velocities, a correction for upstream velocity in the derivation of maximal gradients by the Bernoulli equation is necessary for accurate results.

Application of Continuity Equation

The determination of an effective orifice area improved the discrimination between sizes of normal St. Jude valves in the aortic position. LVO diameters measured by echocardiography were on the average smaller than sewing ring diameters. Although this finding could represent an underestimation of actual size due to methodology (e.g., inner-edge-to-inner-edge measurement, interference of prosthesis shadows, or foreshortening of LVO tract), the variability

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**TABLE 2. Variability of Doppler Echocardiographic Parameters in Normal Aortic St. Jude Medical Valves**

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Intraobserver (%)</th>
<th>Interobserver (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>D&lt;sub&gt;LVO&lt;/sub&gt;</td>
<td>4±5</td>
<td>7±6</td>
</tr>
<tr>
<td>PK&lt;sub&gt;LVO&lt;/sub&gt;</td>
<td>6±3</td>
<td>14±11</td>
</tr>
<tr>
<td>PK&lt;sub&gt;jet&lt;/sub&gt;</td>
<td>6±4</td>
<td>8±8</td>
</tr>
<tr>
<td>Mean gradient</td>
<td>14±8</td>
<td>14±11</td>
</tr>
<tr>
<td>Velocity index</td>
<td>12±4</td>
<td>12±8</td>
</tr>
<tr>
<td>EOA&lt;sub&gt;echo&lt;/sub&gt;</td>
<td>13±11</td>
<td>18±12</td>
</tr>
<tr>
<td>EOA&lt;sub&gt;SR&lt;/sub&gt;</td>
<td>11±4</td>
<td>12±8</td>
</tr>
</tbody>
</table>

Values are given as mean±SD.

D<sub>LVO</sub>, diameter of left ventricular outflow measured by echocardiography; PK<sub>LVO</sub>, peak velocity of left ventricular outflow; PK<sub>jet</sub>, peak velocity of jet; EOA<sub>echo</sub> and EOA<sub>SR</sub>, effective orifice areas using continuity equation, where cross-sectional area of ventricular outflow was measured by echocardiography or determined from sewing ring area, respectively.
observed may well depict an actual variable relation of the size of the LVO to that of the sewing ring. This could represent differences in how tightly the prosthesis is seated in position as well as the variable degree of septal hypertrophy below the aortic anulus. However, the substitution of sewing ring area for flow area in the continuity equation slightly improved the discrimination among various sizes of normal valves. This finding coupled with the interobserver variability in the determination of LVO area would favor the use of this method for calculation of effective orifice area if the valve size is known. The lowest value of an effective orifice area by Doppler was 0.73 cm² in a 19-mm valve. This as well as all effective orifice areas by Doppler should not be equated to values of native aortic valves but instead should be referenced to normal values of the particular valve type and size. Further clinical studies are needed to determine the severity of a prosthetic valve obstruction based on effective area calculations.

The Doppler velocity index, derived as the ratio of the peak velocity in the outflow tract to that through the prosthesis, does not rely on measurement of flow area in the ventricular outflow; rather, it depends on the ratio of effective orifice area to LVO area. Because of the relation of LVO area to valve size, the velocity index is proportional to the ratio of effective area to sewing ring area, defined in the hemodynamic literature as the performance index of a prosthetic valve.\textsuperscript{13} The weak relation observed between the velocity index and valve size reflects in part the higher performance of larger valves, previously reported using the hydraulic formula.\textsuperscript{17} Ideally, the Doppler velocity index of a valve should be referenced to normal values of a particular valve size. The smallest velocity index observed in this study was 0.26 for a 19-mm valve. However, a velocity index of less than 0.23 should cause suspicion of an obstruction of a St. Jude aortic valve of any size. This cutoff value represents 2 SDs below the mean velocity index of all patients studied and is less than 2 SDs below the mean of each valve size group (Table 1). Because the Doppler velocity index is less dependent on valve size, it may provide a good and simple screening test for valve obstruction, particularly if the valve size is not known. Whether the Doppler velocity index and effective orifice area calculation have similar accuracies in the detection of prosthetic valve obstruction remains to be determined.

**Comparison With Previous Hemodynamic Studies**

Effective orifice areas of prosthetic valves have usually been derived using modifications of the hydraulic formula relating effective area to flow and mean pressure gradient across the valve.\textsuperscript{13,17,33} In the hydraulic equation, the discharge coefficient is a measure of how well a valve uses its actual orifice. Knowing the actual orifice area of a valve, the discharge coefficient is calculated as the effective orifice area divided by the actual orifice area and is equal to or less than 1. Table 3 compares effective orifice areas and discharge coefficients obtained in the present study by Doppler using the continuity equation with previously reported results in vitro and in patients for normal aortic St. Jude valves using the hydraulic equation.\textsuperscript{14,15,17,18,32} With analysis of variance, there were no differences among discharge coefficients of various valve sizes derived with the continuity equation using either echocardiographic or sewing ring measurements. Overall, effective orifice areas and discharge coefficients by the hydraulic equation were larger than those obtained by Doppler, even if the sewing ring areas were used as flow areas. Possible explanations for this observation include overestimation of valve area by the hydraulic equation when the mean gradient approaches zero, combined with errors in flow determination, particularly in the clinical setting. In some clinical studies, this has resulted in the calculation of effective orifice areas larger than actual areas and thus discharge coefficients of more than 1 (Table 3).\textsuperscript{14,32,34} Forcing some investigators to use the actual orifice area in this situation.\textsuperscript{18} As to possible reasons for smaller effective orifice areas by the Doppler technique,

<table>
<thead>
<tr>
<th>Valve size (mm)</th>
<th>AOA (cm²)</th>
<th>$EOA_{echo}$ (cm²)</th>
<th>$Cd_{echo}$</th>
<th>$EOA_{SR}$ (cm²)</th>
<th>$Cd_{SR}$</th>
<th>In vivo*</th>
<th>In vitro†</th>
</tr>
</thead>
<tbody>
<tr>
<td>19</td>
<td>1.63</td>
<td>0.99±0.20</td>
<td>0.61±0.12</td>
<td>1.04±0.19</td>
<td>0.64±0.12</td>
<td>1.48±0.52</td>
<td>0.91±0.32</td>
</tr>
<tr>
<td>21</td>
<td>2.06</td>
<td>1.25±0.21</td>
<td>0.61±0.10</td>
<td>1.38±0.22</td>
<td>0.67±0.11</td>
<td>2.39±0.37</td>
<td>1.16±0.66</td>
</tr>
<tr>
<td>23</td>
<td>2.55</td>
<td>1.28±0.31</td>
<td>0.50±0.12</td>
<td>1.52±0.26</td>
<td>0.60±0.10</td>
<td>3.19±1.53</td>
<td>1.25±0.60</td>
</tr>
<tr>
<td>25</td>
<td>3.09</td>
<td>1.80±0.41</td>
<td>0.58±0.13</td>
<td>2.08±0.41</td>
<td>0.67±0.13</td>
<td>2.82±1.03</td>
<td>0.91±0.33</td>
</tr>
<tr>
<td>27</td>
<td>3.67</td>
<td>2.43±0.63</td>
<td>0.66±0.17</td>
<td>2.65±0.58</td>
<td>0.72±0.16</td>
<td>...</td>
<td>3.62</td>
</tr>
<tr>
<td>29</td>
<td>4.52</td>
<td>2.66±0.26</td>
<td>0.59±0.06</td>
<td>3.23±0.30</td>
<td>0.71±0.07</td>
<td>...</td>
<td>...</td>
</tr>
<tr>
<td>31</td>
<td>5.18</td>
<td>3.08±1.09</td>
<td>0.59±0.21</td>
<td>3.72±1.40</td>
<td>0.72±0.27</td>
<td>...</td>
<td>...</td>
</tr>
</tbody>
</table>

Values are given as mean±SD.

*Pooled mean and SDs from References 14, 18, and 32; †mean values from References 15 and 17, where available.

AOA, actual orifice area; $EOA_{echo}$ and $EOA_{SR}$, effective orifice areas using echocardiographic and sewing ring area, respectively (see text); $Cd_{echo}$ and $Cd_{SR}$, discharge coefficients using echocardiographic and sewing ring area, respectively; Cd, discharge coefficient.
these include a consistent underestimation of velocity in the outflow tract and, more likely, an overestimation of mean spatial velocity of the aortic jet because of a change in flow profile through the valve. Recent studies have demonstrated that the flow profile through metallic valves is altered. In the case of a St. Jude valve, the flow is central with a modified parabolic profile. With continuous wave Doppler, the velocity contour that can be traced for quantitation is that of the maximal velocity, which in this case will overestimate the mean spatial velocity of the jet and thus underestimate effective valve area. In support of this hypothesis are recent preliminary observations in vitro demonstrating in St. Jude compared with Hancock valves a consistent overestimation of Doppler-derived gradients compared with those derived by catheter. This is thought to be the result of localized gradients at the valve level and pressure recovery distal to the valve. This phenomenon may depend on the type of valve and its inherent flow profile. In a recent study by Kapur et al involving 11 patients with a variety of prosthetic mechanical aortic valves, the correlation coefficient between effective areas by the Gorlin and continuity equations was 0.76, without a consistent trend for overestimation or underestimation of effective areas. Further studies comparing the continuity equation by Doppler with the hydraulic equation in the same setting are needed to substantiate the potential discrepancies observed in the present study and elucidate their underlying mechanisms.

Clinical Case Studies

To test whether the use of the continuity equation can discriminate between obstructed and nonobstructed valves, the various Doppler parameters were determined in five patients who had surgically documented St. Jude aortic valve dysfunction as well as echocardiographic and Doppler examinations within 3 days before surgery (Table 4). All patients had symptoms of congestive heart failure. Three patients had obstruction of the prosthetic valve due to pannus formation restricting valve motion, and two had severe paravalvular insufficiency secondary to a large paravalvular leak, also documented by contrast angiography and Doppler studies. The last two patients (patients 4 and 5) also had studies during a hyperdynamic state, a few days after repair of the paravalvular leak.

Although the gradients across the severely obstructed valves were exceedingly large, they were also increased in the majority of cases of valvular regurgitation and in normal valves with hyperdynamic ventricular function, despite correction for LVO velocity. However, the application of the continuity equation by either the Doppler velocity index, calculation of an effective orifice area, or discharge coefficient well differentiated the obstructed from the nonobstructed valves (Table 4). Furthermore, it helped elucidate that the high gradients observed in patients 4 and 5 were secondary to increased flow through the valve. Whether the high flow through the valve is secondary to prosthetic valve insufficiency or merely a hyperdynamic state should be assessed in the same setting by an integrative approach using pulsed, continuous wave, and color Doppler techniques.

**Table 4. Clinical and Doppler Findings in Patients With St. Jude Aortic Valve Dysfunction**

<table>
<thead>
<tr>
<th>Patient</th>
<th>Diagnosis</th>
<th>Valve size (mm)</th>
<th>Age (yr/sex)</th>
<th>SVLVO (mL)</th>
<th>PkVlVO (m/sec)</th>
<th>Maximal gradient (mm Hg)</th>
<th>Corr max gr (mm Hg)</th>
<th>Mean gradient (mm Hg)</th>
<th>Corr max velocity index</th>
<th>VOAECCHO (cm²)</th>
<th>CdsR</th>
<th>EOAASR (cm²)</th>
<th>CdsR</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Obstruction</td>
<td>19</td>
<td>39/F</td>
<td>59</td>
<td>1.1</td>
<td>100</td>
<td>95</td>
<td>63</td>
<td>0.22</td>
<td>0.50</td>
<td>0.31</td>
<td>0.62</td>
<td>0.38</td>
</tr>
<tr>
<td>2</td>
<td>Obstruction</td>
<td>23</td>
<td>74/F</td>
<td>35</td>
<td>0.9</td>
<td>121</td>
<td>118</td>
<td>80</td>
<td>0.16</td>
<td>0.33</td>
<td>0.13</td>
<td>0.68</td>
<td>0.27</td>
</tr>
<tr>
<td>3</td>
<td>Obstruction</td>
<td>19</td>
<td>77/F</td>
<td>66</td>
<td>0.9</td>
<td>100</td>
<td>97</td>
<td>57</td>
<td>0.18</td>
<td>0.46</td>
<td>0.28</td>
<td>0.51</td>
<td>0.31</td>
</tr>
<tr>
<td>4</td>
<td>Paravalvular</td>
<td>21</td>
<td>39/M</td>
<td>104</td>
<td>1.6</td>
<td>88</td>
<td>78</td>
<td>46</td>
<td>0.34</td>
<td>1.07</td>
<td>0.52</td>
<td>1.18</td>
<td>0.57</td>
</tr>
<tr>
<td>5</td>
<td>S/P repair</td>
<td>21</td>
<td>39/M</td>
<td>93</td>
<td>1.4</td>
<td>64</td>
<td>56</td>
<td>36</td>
<td>0.35</td>
<td>1.10</td>
<td>0.53</td>
<td>1.21</td>
<td>0.59</td>
</tr>
<tr>
<td>6</td>
<td>Paravalvular</td>
<td>23</td>
<td>67/M</td>
<td>137</td>
<td>1.5</td>
<td>46</td>
<td>37</td>
<td>26</td>
<td>0.44</td>
<td>1.68</td>
<td>0.66</td>
<td>1.83</td>
<td>0.72</td>
</tr>
<tr>
<td>7</td>
<td>S/P repair</td>
<td>23</td>
<td>67/M</td>
<td>138</td>
<td>1.45</td>
<td>52</td>
<td>44</td>
<td>27</td>
<td>0.40</td>
<td>1.53</td>
<td>0.60</td>
<td>1.67</td>
<td>0.65</td>
</tr>
</tbody>
</table>

SVLVO, stroke volume determined by Doppler at left ventricular outflow; PkVlVO, peak velocity in left ventricular outflow; Corr max gr, Doppler-derived maximal gradient with correction for LVO velocity; VOAECCHO, effective orifice area using continuity equation, where cross-sectional area of ventricular outflow was measured by echocardiography; CdECHO, discharge coefficient using echocardiographic area; EOAASR, effective orifice area using continuity equation, where cross-sectional area of ventricular outflow was determined from sewing ring area; CdsR, discharge coefficient using sewing ring area.

**Limitation and Sources of Error**

There are several sources of error that arise in the determination of an effective area by Doppler. Measurement of the LVO diameter may be more difficult in the presence of a prosthetic valve, accounting in part for the higher interobserver variability compared with that observed in native aortic valves. An angle of more than 20° between the ultrasound beam and flow would result in significant underestimation of velocity. Varying the apical position of the transducer for recording of velocity in the outflow tract by pulsed Doppler and attempting to record the jet velocity from multiple windows with the selection of the highest velocities will help reduce this error. Although we have not performed calculations of effective areas by both hydraulic and continuity equations in the same patients, the variability in the determi-
nation of effective areas and discharge coefficients (Table 4), assessed as coefficient of variation (SD/mean), appears to be greater for the Gorlin equation than with Doppler. This may reflect errors in cardiac output measurements and, more importantly, that in the hydraulic equation, when the mean gradients are small, a small variability in these measurements would impart large errors in area calculation.

In conclusion, the continuity equation can be applied to the evaluation of St. Jude Medical valves in the aortic position and provides additional quantitative parameters for the overall assessment of prosthetic valve function. Whether these concepts can be applied to other prosthetic valves in various positions remains to be determined. The dependence of normal values of these indexes on valve size emphasizes the need to know the valve type and size when evaluating possible obstruction of a prosthetic valve by Doppler echocardiography.

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