Validation of continuous-wave Doppler echocardiographic measurements of mitral and tricuspid prosthetic valve gradients: a simultaneous Doppler-catheter study

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ABSTRACT For patients with stenotic native valves, the modified Bernoulli equation ($\Delta P = 4v^2$) may be applied to Doppler-measured transvalvular velocities to yield an accurate estimate of transvalvular gradients. Although it would be useful if the same approach could be used for those with stenotic prosthetic valves, no previous study has validated the Doppler technique in this setting. We therefore recorded simultaneous continuous-wave Doppler flow profiles and transvalvular manometric gradients in 12 catheterized patients in whom all atrial and ventricular pressures were directly measured (transseptal left atrial catheterization and transthoracic ventricular puncture) were performed where necessary. A total of 13 prostheses were studied: 11 mitral (seven porcine, three Starr-Edwards, and one Björk-Shiley) and two tricuspid (one porcine and one Björk-Shiley). The Doppler-determined mean gradient was calculated as the mean of the instantaneous gradients ($\Delta P = 4v^2$) at 10 msec intervals throughout diastole. The correlation of simultaneous Doppler (DMG) and manometric mean gradients (MG) for the whole group ($n = 13$) demonstrated a highly significant relationship ($MG = 1.07DMG + 0.28; r = .96, p = .0001$). The correlation was equally good for porcine valves alone ($n = 8$) ($MG = 1.06DMG + 0.55; r = .96, p = .001$) and for mechanical valves alone ($n = 5$) ($MG = 1.06DMG - 0.04; r = .93, p = .02$). In a subset of patients without regurgitation ($n = 8$), prosthetic valve areas were estimated by two Doppler methods originally described by Holen and Hatle, as well as by the invasive Gorlin method. As expected from theoretical considerations, a close correlation was not demonstrated between results of the Gorlin method and those of either Hatle’s Doppler method ($r = .65, fp = NS$) or Holen’s method ($r = .14, p = NS$). Comparison of the results of the two Doppler methods yielded a somewhat closer correlation ($r = .73, p = .05$). These results suggest that in patients with disk-occluder, ball-occluder, and porcine prosthetic valves, Doppler estimates of transvalvular gradients are virtually identical to those obtained invasively.

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IN 1960, Starr and Edward1 and Harken et al.2 implanted the first intracardiac prosthetic valves in humans, and introduced a new era in cardiology. Unfortunately, subsequent experience has demonstrated that prosthetic valves may become regurgitant or pathologically stenotic for a variety of reasons.3 Detection of such abnormalities may pose a diagnostic challenge because the symptoms of valvular dysfunction are frequently nonspecific and the signs are difficult to detect on routine clinical examination.

The identification of prosthetic valve stenosis is particularly problematic. Although noninvasive diagnostic testing in the form of phonocardiography, fluoroscopy, and M mode and cross-sectional echocardiography4,5 has been applied with limited success, to date the mainstay of investigation for prosthetic valve stenosis has been cardiac catheterization with careful manometric measurements of gradients and cardiac output. This form of investigation, however, is complicated by the risks associated with crossing prosthet-
ic valves, and patients often require direct cardiac chamber puncture (transthoracic or transseptal). In the presence of multiple prosthetic valves, the problem is compounded.

A noninvasive method capable of reliably assessing prosthetic valve function would therefore be invaluable. In recent years, studies have shown that Doppler echocardiography is capable of measuring the flow velocity across native valves. These velocities may be used to calculate transvalvular pressure gradients by the modified form of the Bernoulli equation ($\Delta P = 4v^2$). In theory, this general Doppler approach should also be applicable to prosthetic valves. The situation is complicated, however, since flow patterns across prosthetic valves are fundamentally different from those found normally, with further variation between types and subtypes of prostheses.

Studies of porcine valves in vitro suggest that their flow stream is central and similar to that through human aortic valves. Flow beyond mechanical valves, however, is considerably different, with semicentral flow demonstrated in tilting-disk valves and peripheral flow in ball valves.

Holen et al. have demonstrated an excellent correlation between the Doppler-estimated mean gradient and the simultaneously recorded manometric mean gradient across the mitral valve in nine patients with Björk-Shiley and in eight patients with Hancock valves. These patients were studied soon after valve replacement and presumably had normally functioning prostheses. More recent data, however, suggest that the correlation between Doppler- and catheter-determined gradients may, in fact, be less good. Thus, limited and conflicting data are available concerning the accuracy of the Doppler measurements of transvalvular gradients in patients with prostheses in the mitral or tricuspid position, and virtually nothing is known about the accuracy of these measurements in the group in which they are most important (i.e., those with prosthetic dysfunction). In this study, therefore, we examined a group of patients with clinical evidence of prosthetic valve dysfunction to determine whether the mean valve gradients measured by continuous-wave Doppler echocardiography in disk-occluder, ball-occluder, and porcine heterograft prostheses, in either the mitral or tricuspid position, correlate with those simultaneously measured manometrically. In addition, because hemodynamic and Doppler data have both been previously used to calculate valve orifice areas in native and prosthetic valves, we also compared these derived variables with one another for our patient group.

**Methods**

**Study group.** The study group consisted of 12 patients (three men and nine women) ranging in age from 58 to 75 years (mean 65). Each of the 12 patients had a prosthesis in the mitral position and two had an additional prosthetic tricuspid valve. In one patient (No. 7, table 1), only the tricuspid valve gradient was measured directly (no transseptal catheterization) and the mitral valve data were therefore not included. Of the 11 mitral prostheses studied seven were porcine (four Hancock, three Carpenter-Perrin-Edwards), three were Starr-Edwards, and one was a Björk-Shiley. One of the tricuspid valves was porcine (Hancock) and one was a Björk-Shiley. Additional prosthetic valves and surgical procedures are listed in table 1. The cardiac rhythm was normal sinus in five patients and atrial fibrillation in seven. Flow velocity data from prostheses in the aortic position were not measured in this study due to the limitations of time that exist when patients have multiple catheters (including transseptal and transthoracic) in position.

All patients presented with clinical evidence of atrioventricular prosthetic valve dysfunction: that is, pulmonary edema of recent onset, a significant decrease in exertional ability, recent breathlessness on exertion, or right heart failure in the case of those patients with tricuspid prostheses. In all these patients with complex cardiac lesions (table 1), the suspicion of prosthetic valve dysfunction at the mitral or tricuspid site was considered strong enough to justify a detailed invasive examination.

**Doppler examination.** All Doppler studies were performed in the catheterization laboratory, with continuous-wave Doppler flow velocity and transvalvular pressure gradients recorded simultaneously. Transvalvular flow velocity was measured with an Irex Exemplar (Johnson & Johnson Ultrasound, Ramsey, NJ) or a Hewlett-Packard model 77020AC (HP Andover, MA) velocity meter with a dedicated 2.0 mHz Pedof/Irex or 2.5 mHz Hewlett-Packard continuous-wave Doppler transducer. Patients were examined while they were supine on the catheterization table. Before the insertion of the catheters, a screening Doppler study was performed to identify the position on the chest wall where the maximal flow velocity could be recorded. This position was marked to facilitate the subsequent study. For patients with mitral prosthetic valves, this was usually located 1 to 3 cm superomedial to the apex; for those with tricuspid prostheses, the sampling position was medial to that used for mitral interrogation. The position was, however, unique to each patient and required careful searching in multiple windows. To confirm that the correct valve was being recorded by Doppler echocardiography, especially in patients with both tricuspid and mitral prostheses, the left or right atrial catheter was flushed with saline during the continuous-wave Doppler examination. This produced marked signal enhancement and confirmed the correct sampling position.

**Cardiac catheterization.** All patients were sedated with diazepam (5 mg orally) and chlorpromazine (25 mg orally). One patient was mechanically ventilated because of severe pulmonary edema. In all mitral valve prostheses studied ($n = 11$), catheterization was performed by the transseptal technique and a direct left atrial pressure was measured. The left ventricular pressure was simultaneously measured by either retrograde cannulation across the aortic valve or by left ventricular transthoracic puncture (in the three patients with additional aortic valve prostheses). In the two patients with tricuspid prostheses, the tricuspid valve gradient was measured by right atrial catheterization via the femoral vein and either a simultaneous transthoracic right ventricular puncture (in the patient with a Björk-Shiley prosthesis) or, in the case of the Hancock tricuspid prosthesis, by crossing the prosthesis with a second catheter. Manometric measurements were all performed with No. 7F or
TABLE 1
Summary of Doppler and catheterization data from 12 patients with prosthetic valves

<table>
<thead>
<tr>
<th>Patient No.</th>
<th>Age (yr)</th>
<th>Sex</th>
<th>Valve type</th>
<th>Size</th>
<th>Other lesions</th>
<th>Heart rate (bpm)</th>
<th>Cardiac output (l/min)</th>
<th>P_{man} (mm Hg)</th>
<th>P_{Dop} (mm Hg)</th>
<th>MR</th>
<th>Gorlin Half-time (s)</th>
<th>EOA (cm²)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>70/F</td>
<td>F</td>
<td>Starr-Edwards MVR</td>
<td>—</td>
<td>Native AS</td>
<td>99</td>
<td>3.3</td>
<td>12.14</td>
<td>—</td>
<td>1.2</td>
<td>1.87</td>
<td>1.23</td>
</tr>
<tr>
<td>2</td>
<td>61/M</td>
<td>M</td>
<td>Hancock MVR</td>
<td>No. 27</td>
<td>Single CABG</td>
<td>92</td>
<td>4.8</td>
<td>11.0</td>
<td>—</td>
<td>1.2</td>
<td>1.5</td>
<td>1.49</td>
</tr>
<tr>
<td>3</td>
<td>67/F</td>
<td>F</td>
<td>Björk-Shiley MVR</td>
<td>No. 27</td>
<td>Asthma, tricuspid annulo</td>
<td>68</td>
<td>5.6</td>
<td>3.45</td>
<td>—</td>
<td>3.0</td>
<td>2.6</td>
<td>1.75</td>
</tr>
<tr>
<td>4</td>
<td>58/F</td>
<td>F</td>
<td>Björk-Shiley MVR</td>
<td>No. 29</td>
<td>Björk-Shiley No. 27 AVR, coarctation repair, PH</td>
<td>103</td>
<td>4.8</td>
<td>6.81</td>
<td>+</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>5</td>
<td>70/F</td>
<td>F</td>
<td>Carpenter-Edwards MVR</td>
<td>No. 33</td>
<td>—</td>
<td>128</td>
<td>5.5</td>
<td>10.28</td>
<td>—</td>
<td>1.4</td>
<td>2.1</td>
<td>1.95</td>
</tr>
<tr>
<td>6</td>
<td>60/M</td>
<td>M</td>
<td>Hancock MVR</td>
<td>No. 31</td>
<td>COPD</td>
<td>68</td>
<td>2.4</td>
<td>5.31</td>
<td>—</td>
<td>1.7</td>
<td>1.2</td>
<td>0.62</td>
</tr>
<tr>
<td>7</td>
<td>67/F</td>
<td>F</td>
<td>Carpenter-Edwards MVR</td>
<td>No. 31</td>
<td>Starr-Edwards AVR</td>
<td>78</td>
<td>4.9</td>
<td>15.06</td>
<td>—</td>
<td>0.8</td>
<td>0.3</td>
<td>1.23</td>
</tr>
<tr>
<td>8</td>
<td>66/F</td>
<td>F</td>
<td>Carpenter-Edwards MVR</td>
<td>No. 29</td>
<td>Hancock AVR</td>
<td>81</td>
<td>2.5</td>
<td>6.59</td>
<td>+</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>9</td>
<td>60/F</td>
<td>F</td>
<td>Björk-Shiley MVR</td>
<td>No. 27</td>
<td>Björk-Shiley No. 21 AVR, PH</td>
<td>97</td>
<td>5.6</td>
<td>3.80</td>
<td>—</td>
<td>1.2</td>
<td>2.29</td>
<td>1.92</td>
</tr>
<tr>
<td>10</td>
<td>63/F</td>
<td>F</td>
<td>Carpenter-Edwards MVR</td>
<td>No. 29</td>
<td>PH, LV dysfunction, tricuspid annulo</td>
<td>56</td>
<td>4.4</td>
<td>4.0</td>
<td>—</td>
<td>1.5</td>
<td>1.9</td>
<td>1.82</td>
</tr>
<tr>
<td>11</td>
<td>63/M</td>
<td>M</td>
<td>Starr-Edwards MVR</td>
<td>3M</td>
<td>—</td>
<td>71</td>
<td>3.5</td>
<td>6.61</td>
<td>+</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>12</td>
<td>75/M</td>
<td>M</td>
<td>Carpenter-Edwards MVR</td>
<td>No. 31</td>
<td>—</td>
<td>82</td>
<td>8.4</td>
<td>10.36</td>
<td>+</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

P_{man} = mean manometric transvalvular gradient (mm Hg); P_{Dop} = mean Doppler-determined transvalvular gradient (mm Hg); MR = mitral regurgitation; MVR = mitral valve replacement; TVR = tricuspid valve replacement; AS = aortic stenosis; CABG = coronary artery bypass graft; tricuspid annulo = tricuspid annuloplasty; AVR = aortic valve replacement; PH = pulmonary hypertension; COPD = chronic obstructive pulmonary disease.

Valve area (cm²) results with three methods (Gorlin, half-time, effective orifice area [EOA]; see text equations 3, 2, and 1, respectively).

8F fluid-filled catheters connected to Hewlett-Packard 1280C pressure transducers. In each case, simultaneous superimposed atrial and ventricular pressure tracings were recorded at the same attenuation on a Hewlett-Packard 7760A physiologic recorder. Once stable manometric recordings were obtained, the Doppler maximal velocity was again located and simultaneous pressure and velocity data were recorded (figure 1). Immediately after the simultaneous measurements, the cardiac output was determined by either the Fick or green-dye method, with the thermodilution method used in only two patients (both of whom had no demonstrable tricuspid or pulmonary valve regurgitation on prior pulsed Doppler study). The presence or absence of prosthetic valve regurgitation was determined by contrast angiography immediately after the hemodynamic recordings were completed.

Data analysis. The manometric tracings were analyzed by an observer who determined the mean gradient by planimetry. The mean gradient was determined from the Doppler spectral velocity tracings by a second independent observer. The outer border of the continuous-wave Doppler spectral profile was digitized and the instantaneous maximal pressure gradient was calculated at 10 msec intervals with the modified Bernoulli equation \( \Delta P = 4v^2 \). The mean \( \Delta P \) was calculated as the arithmetic mean of the instantaneous maximal pressure gradients (figure 2). No angle correction was used in the calculation since flow was assumed to be parallel to the Doppler beam. The square of the flow velocity proximal to the prosthetic valve in each case was considered to be negligible relative to the square of the flow velocity distal to the valve; accordingly, the maximal velocity recorded was used in the calculation. For both hemodynamic and Doppler data sets, 10 cycles were analyzed for patients in atrial fibrillation and five were analyzed for those in normal sinus rhythm and the mean of these measurements was calculated.

Estimation of valve area. The Doppler spectral tracings were used to estimate a prosthetic valve orifice by two methods: (1) the pressure half-time method described by Holen et al., \( \frac{220}{t_{\frac{1}{2}}} \) and (2) the method described by Holen et al., where \( \hat{Q} \) is transprosthetic valve flow volume per minute (ml/min) taken from the measured cardiac output (Fick, green-
**FIGURE 1.** Superimposed continuous-wave Doppler spectral and manometric tracings obtained simultaneously during catheterization in patient 2. Time (horizontal scale) is calibrated at 40 msec intervals. Pressure and velocity calibrations (vertical scale) are indicated. LA = direct left atrial pressure; LV = left ventricular pressure; PA = mean pulmonary arterial pressure.

dye, thermodilution), \( \bar{v} \) is the mean transprosthetic valve flow velocity taken from the Doppler tracings, and \( T_d \) is the diastolic filling period measured on the Doppler tracings. The Doppler mean velocity \( (\bar{v}) \) was calculated by dividing the planimetered area under the spectral profile curve (the velocity integral) by the diastolic filling period \( (T_d) \). This equation may be simplified to \( Q/VI \) (flow divided by the velocity integral).

The prosthetic valve area was also estimated by the method described by Gorlin and his colleagues\(^2\) with the use of invasively determined variables as

\[
\text{Area} = \frac{F}{38\sqrt{\Delta P}}
\]

where 38 is the empiric constant for mitral valves and \( F \) is the volume rate of flow across the orifice in milliliters per second (derived by dividing the measured cardiac output by the diastolic filling period). Holen’s method (equation 2) measures the effective orifice, that is the contracted area of the physical orifice through which the flow jet actually passes. Gorlin’s method (equation 3) measures the anatomic orifice area. For purposes of direct comparison, Holen et al.\(^2\) have suggested that the effective orifice area be divided by a theoretic correction factor of 0.775 to approximate the physical orifice. In this study, the Doppler and hemodynamic data were compared with and without this correction. Since Hatle’s method (equation 1) was developed to equate a pressure half-time of 220 to a Gorlin orifice of 1 cm\(^2\), it also estimates a physical orifice area. Therefore, the correction factor should also be applied in comparing results from equation 1 and equation 2.

Only data from patients with competent prosthetic valves were included in valve area comparisons, since both the Holen and Gorlin equations assume that forward flow at the valve in question is the same as the cardiac output, an assumption that breaks down in the presence of valvular regurgitation when the flow at the examined valve is the sum of the regurgitant volume plus the forward cardiac output. Excluding those patients demonstrated at angiography to have prosthetic valve regurgitation reduced the study group to two patients with tricuspid prosth-
ses and six patients with mitral prostheses, for a total of eight prostheses (table 1).

Statistical methods. All data are expressed as the mean ± 1 SD. The correlation between valve gradients simultaneously recorded by the Doppler method and manometry and between results of various methods of estimating prosthetic valve orifice was by linear regression analysis (least squares method). The effect of valve type (porcine vs mechanical) was assessed by comparison of the results of linear regression analyses for the two groups by a t-test for the intercepts and slopes. The statistical power of the comparisons of slope and intercepts was calculated from the beta error of each regression analysis.

Results

Continuous-wave Doppler echocardiography. The maximal velocities recorded ranged from 1.42 to 2.24 m/sec (mean 1.97 ± 0.29). The Doppler-calculated peak pressure gradients ranged from 8.1 to 20.1 mm Hg (mean 15.8 ± 4.4) and Doppler-calculated mean transvalvular gradients ranged from 3.13 to 15.08 mm Hg (mean 7.6 ± 3.73). The values for porcine valves (range 3.45 to 15.08 mm Hg, mean 7.87) were similar to those for mechanical valves (range 3.80 to 12.14 mm Hg, mean 7.18).

Catheterization. The manometric mean gradients ranged from 3 to 15 mm Hg (mean 8.4 ± 4.17). Cardiac output ranged from 2.4 to 8.4 liters/min (mean 4.7 ± 1.6). Cardiac index was from 1.5 to 4.0 liters/min (mean 2.8 ± 0.77). Heart rate ranged from 56 to 128 beats/min.

Correlation of simultaneous Doppler- and catheter-measured mean pressure gradients. An excellent correlation was obtained between the Doppler-determined mean prosthetic valve gradients (Dop. grad.) and the simultaneously obtained mean manometric gradients (cath. grad.) (cath. grad. = 1.07 Dop. grad. + 0.28; r = .96, p <.0001) (figure 3). The correlation, slope, and intercept were not statistically different from those of the line of identity. When porcine valves (figure 4) and mechanical valves (figure 5) were examined separately, the correlations with the manometric gradients were similar, with no significant differences in slope or intercept. For porcine valves (n = 8), Cath. grad. = 1.06 Dop. grad. + 0.55 (r = .96, p <.001), and for mechanical valves (n = 5) Cath. grad. = 1.06 Dop. grad. −0.04 (r = .93, p = .02). Because of the small number of patients in the two subgroups (n = 8 and n = 5), a considerable difference in the slopes or intercepts would have been necessary before a statistically significant difference could have been detected. Further analysis according to the site of prosthetic valve insertion (tricuspid vs mitral) or type of mechanical valve (Starr vs Björk-Shiley) was not possible because of the small numbers in these subsets. It was found, however, that the slope of the line drawn with data from the two patients with prostheses in the tricuspid position (slope 1.12) was similar to that drawn with data from patients with mitral prostheses (slope 1.06) (figure 3).

Orifice measurements. As calculated from the Hatle equation (equation 1), the orifice areas ranged from 0.3 to 2.60 cm² (mean 1.6 ± 0.77). With the uncorrected effective orifice area equation (Holen, equation

\[
\text{CATH} + 0.07 \text{DOP} + 0.28
\]

\[
r = 0.96
\]

\[
p < 0.001
\]

\[
\text{MITRAL}\quad \triangle \quad \text{TRICUSPID}
\]

FIGURE 3. Correlation of simultaneous mean Doppler-determined (x axis) and manometric (y axis) gradients for 13 prosthetic valves. Prostheses in the mitral position (n = 11) are identified by squares, and those in the tricuspid position (n = 2) by triangles. The valve types are eight porcine, three Björk-Shiley, and two Starr-Edwards prostheses. Cath = manometric gradient; Dop = Doppler-derived gradient.

\[
\text{CATH} = 1.06 \text{DOP} + 0.55
\]

\[
r = 0.96
\]

\[
p < 0.001
\]

\[
\text{MITRAL}\quad \triangle \quad \text{TRICUSPID}
\]

FIGURE 4. Correlation of simultaneous mean Doppler-determined (x axis) and manometric (y axis) gradients for eight porcine valves. Cath = manometric gradient; Dop = Doppler-derived gradient.
2), orifice areas ranging from 0.61 to 1.95 cm² (mean 1.50 ± 0.44) were calculated.

The application of Gorlin's equation (equation 3) to the hemodynamic data yielded estimates of the valve orifice areas ranging from 0.80 to 3.0 cm² (mean 1.50 ± 0.66 cm²).

Correlation of methods of estimating valve orifice area. Comparison of the orifice areas calculated by the Gorlin and pressure half-time methods (n = 8) failed to yield a significant correlation (r = .65, p = NS; y = 0.55x + 0.61, where y is Gorlin area and x is that derived with the pressure half-time method). Similarly, the uncorrected effective orifice area and that calculated by the Gorlin method were not significantly correlated (r = .14, p = NS). The use of the theoretical correction factor suggested by Holen did not, as expected, improve the correlation with results of the Gorlin method (y = 0.16x + 1.19, where y is the Gorlin calculated orifice area and x is the corrected area calculated according to the method of Holen).

Comparison of results obtained with the two Doppler-dependent methods, however, demonstrated a significant correlation (r = .73, p ≤ .05). When the theoretical correction factor was applied to the areas calculated by the Holen equation to allow a direct comparison between the two Doppler methods, a slope of 0.54 and an intercept of 1.69 was demonstrated.

Discussion

A number of studies have now established that in native valves, transvalvular flow velocities measured by continuous-wave Doppler echocardiography can be accurately converted to transvalvular pressure gradi-

This equation predicts the gradient to be four times the square of the maximal velocity measured distal to the valve, provided the velocity proximal to the valve is low and can be ignored. A similar noninvasive estimate of the pressure gradient across prosthetic valves would also be useful in assessing prosthetic valve function. However, the modified Bernoulli equation is valid only under certain conditions that may not be satisfied in the various hydrodynamic states created by different prostheses. The equation is applicable if three conditions are fulfilled: (1) flow must be incompressible, i.e., the density of blood does not change, (2) flow must be effectively frictionless, i.e., the pressure loss due to viscous resistance is negligible, and (3) the flow velocities used in the equation must be measured along streamlines of flow, which are lines drawn tangentially to the direction of flow at a given point in the flow field. The first condition, that of incompressible flow, is an assumption considered to be valid in cardiac applications and should be independent of valve type. However, it is not immediately obvious that the last two requirements are satisfied for all prosthetic valves. With respect to the second condition, flow across prosthetic valves may occur through irregularly shaped orifices unlike those of a central stenosis, leading to the potential for an additional gradient due to viscous resistance. With respect to the third condition, the velocities measured by the Doppler beam may not be parallel to the streamlines of flow throughout the flow period, particularly for valves with moving and tilting elements. Thus, it is necessary to demonstrate that, in the presence of prosthetic valves, the shortened Bernoulli equation provides an accurate measure of the transvalvular pressure gradient.

In an extension of the work in vitro of Holen et al. using a model with a single circular orifice,20 Teirstein et al.23 have recently used a model to show that the continuous-wave Doppler method can accurately measure gradients in irregular, multiple, and tunnel-like obstructions. Their study, in a model with dimensions consistent with those found in prosthetic valves, suggests that the condition of frictionless flow may be effectively fulfilled in prosthetic valves and should be independent of the position of the valve within the heart. Unfortunately, however, relatively little and conflicting clinical data are available relating transvalvular pressure to flow velocity across prosthetic valves. In those studies15, 16 of normally functioning valves, a good correlation was demonstrated between data obtained by Doppler and manometric methods. In the single preliminary report addressing Doppler ve-
locities and transvalvular gradients in malfunctioning valves, however, a good correlation was not found.17

The present study validates the mean transvalvular pressure gradients obtained by applying the Bernoulli equation to continuous-wave Doppler measurements in patients with prosthetic mitral and tricuspid valves. In the patients examined, the Doppler-estimated gradients were essentially identical to those simultaneously measured manometrically for a variety of prostheses representative of the basic valve types currently in clinical use. This relationship held when the study group was divided into those with mechanical (Björk-Shiley and Starr-Edwards) and those with porcine (Hancock and Carpentier-Edwards) valves. However, the conclusion that the identical relationship holds for each subgroup should be viewed with caution because of the small number of patients in the mechanical and porcine groups.

Although the number of patients in the study with tricuspid prostheses was too small for meaningful statistical analysis, the virtually identical slopes suggest that the method should be equally effective in those with mitral and tricuspid prostheses.

We believe that the excellent correlation demonstrated in this study is dependent on a number of important features of study design. First and most important, the Doppler data were recorded simultaneously with the manometric data. Since the gradient developed at any point of obstruction to flow changes dynamically in relation to flow volume, major changes in the gradient can result from changes in the cardiac output. Thus, gradients may vary significantly due to normal physiologic changes in flow. This effect has been well documented in two previous published reports.10,24 Second, manometric gradients were measured by dual-catheter systems of large bore, and the tracings were superimposed. Pull-back gradients were not used since they have reduced accuracy because of normal physiologic beat-to-beat variations in normal forward flow. Third, only patients in whom direct left atrial pressure measurements were recorded via transseptal catheters were included in this study when the mitral valve was examined. Indirect measurement of left atrial pressure by the pulmonary arterial wedge technique can introduce a significant error.25 Wedge estimates of the left atrial pressure may be higher than those measured directly and if used in place of the direct left atrial pressure would result in an overestimation of the transmitial gradient. Although such direct pressure measurements may be difficult to obtain (e.g., ventricular puncture was required in four valves studied), their use in any comparison is essential. Fi-

nally, the comparison made was between the calculated mean gradient from the continuous-wave Doppler tracings and the mean pressure gradient measured at catheterization. The comparison of the more easily obtained peak instantaneous Doppler gradient with the traditionally recorded mean or peak-to-peak manometric gradient is a comparison of different measurements and would not be expected to yield a close correlation.

A further error in calculating the mean gradient from Doppler measurements can be introduced if the fundamental relationship between pressure gradient and velocity is not observed at each instant throughout the flow period. Since at each point in time the pressure gradient is related quadratically to the velocity, the calculated mean pressure gradient must be derived from the instantaneous pressure gradients calculated throughout the diastolic period, as in figure 2. If the mean velocity for the flow period (here the diastolic filling period) is simply introduced into the simplified Bernoulli equation (4v2), a mean pressure gradient analogous to the manometric pressure gradient is not produced. This substitution error will generally result in an underestimation of the gradient and may account for low “mean” gradients observed in a previously published report on Doppler examination of St. Jude prostheses.26

The measurement of pressure alone to quantify the degree of obstruction at a valve (whether native or prosthetic) is clinically limited, because the transvalvular gradient is, itself, a function of the flow volume passing through the orifice. Since the orifice area is less affected by flow,27,28 a measure of the area through which flow is occurring would be more useful than a simple gradient. The continuity equation, which expresses the conservation of mass, provided blood is incompressible, can be applied to this problem. In its simplest form this equation,29 Q = v · A, where Q is volume rate of flow, v the temporal mean velocity of flow, and A the cross-sectional area through which flow is occurring, allows this area to be derived as:

\[ A = \frac{Q}{V} \]  

(4)

and defines the relationship of flow volume to flow velocity. The Doppler application of this equation assumes that flow velocity is uniform across the effective orifice area (plug flow), so that the peak velocity measured by the continuous-wave Doppler method is equal to the spatial average of velocity across the effective orifice or vena contracta (see below). If this velocity is integrated over time to produce a temporal-spatial mean, this value can be directly inserted into equation 4, which constitutes Holen’s method.20 In its basic
form, equation 4 applies generically to all orifices, irrespective of their inlet geometry or shape. The equation can, therefore, be expected to apply equally well to native and prosthetic valves.

Flow through an orifice will consist of a central core in which flow will be streamline and frictionless (the vena contracta) and the peripheral boundary layer around the margins of the physical orifice. As the orifice becomes smaller, the boundary layer will encroach further on the central stream until eventually flow will no longer be streamline and significant energy losses will result from viscous resistance (Poisson’s equation). Prosthetic valve flow can be demonstrated both theoretically and in vitro to have a significant streamline element, up to the extremes of prosthetic valve dysfunction. The continuity equation (equation 4) should, therefore, measure the area through which the central flow core is passing. This effective area will represent a fraction of the total physical orifice and its relative size will depend on the inlet geometry and the flow across the obstruction. The dimensions of the boundary layer in relation to the central layer are described by the coefficient of contraction (Cc). Cc will vary for each valve type and for any rate of flow. Furthermore, this relationship is not linear, and hence cannot be adequately described by a single linear constant for all situations.

The invasive Gorlin method was also derived with the use of the continuity equation (equation 4). Since at the time of the derivation (as now), velocity could not be measured easily in the catheterization laboratory, this component was derived from the pressure drop across a valve (a quadratic relationship). There are, however, two fundamental differences between this method and Holen’s use of the continuity equation. First, Gorlin’s method introduces a linear constant that empirically attempts to correct the equation for the Cc. Hence, this method, as devised, measures the physical orifice dimension. Second, the velocity term in the Gorlin equation is approximated by the term \( \sqrt{\Delta P} \) rather than the theoretically more correct term \( \sqrt{\Delta P} \).

In situations in which there is a large pressure drop (i.e., severe mitral stenosis), the approximation will likely result in a small difference; however, the error increases with smaller gradients. For these reasons, the Gorlin equation may not be satisfactory for prosthetic valves with differing inlet geometries and flow rates than stenotic native mitral valves, the situation for which it was designed. This potential inaccuracy has been pointed out by Herman et al. and is supported by data from Yellin et al. Nevertheless, the method represents the de facto standard, and normal ranges for a number of different prostheses have been published.

Hatile’s method (equation 1), using the pressure half-time relationship, was originally derived from invasive manometric methods. It can only arrive at a valve area by the use of a constant that empirically relates the observed pressure half-time of 220 to an invasively determined orifice area of 1 cm² (equation 3). Nevertheless, the method may be applicable to prosthetic valves since the pressure half-time itself is a physiologic measure of obstruction and does not require assumptions about inlet geometry or flow rates.

Given these theoretical considerations, a direct comparison between these three methods of calculating orifice areas would not necessarily be expected to yield close correlations. Accordingly, this study fails to show a satisfactory correlation of results obtained by either Doppler method with the invasive measurements of prosthetic valve area. The possibility of a close correlation may have been reduced, in part, by the small number of patients studied and by the narrow range of valve areas over which the comparisons were made. Holen and Nitter-Hauge found a more satisfactory correlation (r = .8) between their Doppler measurement and an invasively determined valve area in a group of 10 patients with disk valves. However, this correlation was achieved by use, not of the Gorlin equation, but of an invasive method using a different constant (51.7 vs 38) and the theoretically more correct pressure velocity term (\( \sqrt{\Delta P} \)) vs \( \sqrt{\Delta P} \). That results obtained with the Gorlin equation probably do not represent an ideal gold standard for comparison is suggested by our data also. Since our pressure data were recorded simultaneously and the same measure of cardiac output (Q) was used in both the equations (equations 2 and 3), any difference in derived area must be related to a difference in the denominator of each equation (i.e., the velocity term). The error then may be in the Doppler measurement of flow velocity (a direct measurement), or in the approximation of flow information from the pressure data. We have already demonstrated that the mean Doppler and manometric pressure data correlate very closely; confirming that the Doppler measurement of flow velocity is accurate. Therefore, the data suggest that the lack of a close correlation between areas calculated by the Gorlin and Holen equations may reflect an error in the Gorlin term 38\( \sqrt{\Delta P} \).

Similarly, better correlation of valve areas derived by the two Doppler methods (equations 1 and 2) likely reflects the more precise assessment of the transvalvular flow velocity that is possible by continuous-wave...
Doppler echocardiography compared with that derived from manometry. These data, and the theoretical and clinical considerations discussed, suggest that the Doppler methods may provide a satisfactory means of assessing prosthetic valve area, the poor correlation with the Gorlin valve area notwithstanding. The measure of the effective orifice, i.e., the area through which streamline or frictionless flow is actually occurring, as opposed to the anatomic orifice, is a measure of the flow obstruction and can be applied regardless of valve type or size. Normal values for both Doppler methods have been published for a number of different types and sizes of valve prostheses in patients presumed to have normally functioning valves.42-44

This study includes two patients with prosthetic valves positioned at the tricuspid site (see table 1). Very little information on the Doppler examination of tricuspid prostheses has been published. A previous case report45 demonstrated a satisfactory agreement of the Doppler-determined mean transvalvular gradient with the mean manometric gradient measured subsequently in a stenotic Starr-Edwards prosthesis in the tricuspid position. Our data support this observation. An important reason for the paucity of data concerning tricuspid prostheses is the lack of a satisfactory way of recording the right ventricular pressure without transthoracic puncture of the ventricle or crossing the prosthetic valve with a catheter. The noninvasive Doppler method promises to allow this problem to be avoided.

In conclusion, this study validates the continuous-wave Doppler method of calculating gradients across mitral and tricuspid valve prostheses by demonstrating that the gradients it provides are virtually identical to those measured by manometry. Because the method was equally accurate with mechanical and porcine prostheses, the Doppler technique should become widely used as a clinical method of assessment of prosthetic valves. Calculations of orifice area in the subset of patients without valvular regurgitation showed a less satisfactory correlation with hemodynamically calculated valve areas, but the source of this poor correlation may have been in part the invasive standard to which the Doppler data were compared.

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