Ventricular Stroke Work Loss: Validation of a Method of Quantifying The Severity of Aortic Stenosis and Derivation of an Orifice Formula

DAVID C. SPRIGINGS, MA, MRCGP, JOHN B. CHAMBERS, MA, MD, MRCGP, THOMAS COCHRANE, BSc, PhD,* JOHN ALLEN, MB, MRCPath, GRAHAM JACKSON, MB, FRCP

London and Sheffield, England

Because aortic stenosis results in the loss of left ventricular stroke work (due to resistance to flow through the valve and turbulence in the aorta), the percentage of stroke work that is lost may reflect the severity of stenosis. This index can be calculated from pressure data alone.

The relation between percent stroke work loss and anatomic aortic valve orifice area (measured by planimetry from videotape) was investigated in a pulsatile flow model. Thirteen valves were studied (nine human aortic valves obtained at necropsy and four bioprosthetic valves) at stroke volumes of 40 to 100 ml, giving 57 data points. Valve area ranged from 0.3 to 2.8 cm² and mean systolic pressure gradient from 3 to 66 mm Hg. Percent stroke work loss, calculated as mean systolic pressure gradient divided by mean ventricular systolic pressure x 100%, ranged from 7 to 68%. It was closely related to anatomic orifice area with an inverse exponential relationship and was not significantly related to flow (r = -0.15). An orifice formula was derived that predicted anatomic orifice area with a 95% confidence interval of ±0.5 cm² (orifice area [cm²] = 4.82 [2.39 x log percent stroke work loss], r = -0.94, SEE = 0.029).

These results support the clinical use of percent stroke work loss as an easily obtained index of the severity of aortic stenosis.

(J Am Coll Cardiol 1990;16:1608-14)

Quantification of the severity of aortic valve stenosis is central to the appropriate selection of patients for valve replacement surgery and to the assessment of the results of balloon valvuloplasty (1). Aortic stenosis is characterized by a systolic pressure gradient across the valve, but this gradient is an inaccurate index of the severity of stenosis because of its dependence on flow (2-5). Gorlin and Gorlin (2) proposed a formula for the orifice area of a stenotic valve that relates the pressure gradient to flow. This formula has been widely adopted (6), although it has recognized limitations that are due in part to the imprecision of methods of measuring systolic flow across the valve (6,7).

As aortic stenosis results in the loss of left ventricular stroke work (due to resistance to flow through the valve and turbulence in the ascending aorta) (3,4), the percentage of stroke work that is lost has been proposed (8) as an index of the severity of the stenosis. This index has the important advantage over the Gorlin formula of not requiring measurement of cardiac output for its calculation (as described under Theory). However, there are no data validating percent ventricular stroke work loss against directly observed aortic valve orifice area. We therefore compared percent stroke work loss and anatomic orifice area in a model with pulsatile flow and derived an orifice formula.

Theory

Total left ventricular stroke work (SWtotal) is equal to the product of left ventricular pressure during ejection (P_{LV}) and the volume of blood ejected (V) integrated over the ejection period (T):

\[ SW_{\text{total}} = \int_0^T P_{LV} \, dV \, dt. \]  

[1]
where $\frac{dV}{dt}$ is the infinitesimal volume, $dV$, ejected in the infinitesimal interval, $dt$. As a first approximation, this can be simplified to:

$$SW_{hau} = \text{mean } P_{LV} \times SV.$$  \hfill (2)

where $SV$ is the stroke volume (9).

In valvular aortic stenosis the static pressure measured at the exit plane of the valve is less than that in the left ventricular outflow tract, reflecting energy losses due to inertia of the cusps and resistance to flow through the valve and also the increase in velocity of flow through the narrowed orifice (3,4). Downstream to the valve, the static pressure rises (pressure recovery) as flow decelerates (10,11). However, pressure recovery may be offset by loss of energy due to turbulence. The effective stroke work delivered to the circulation ($SW_{eff}$) can be estimated as the product of ascending aortic pressure during ejection ($P_{aorta}$) and the volume of blood ejected ($V$) integrated over the ejection period ($T$):

$$SW_{eff} = \int_0^T P_{aorta} \frac{dV}{dt} \cdot dt.$$  \hfill (3)

Again, this can be simplified to:

$$SW_{eff} = \text{mean } P_{aorta} \times SV.$$  \hfill (4)

This ignores the small contribution of dynamic pressure (given by blood density, $\times$ velocity$^2$) to the total pressure in the ascending aorta. Assuming a blood density of 1 g/cm$^3$ and a mean velocity in the ascending aorta (beyond the point where pressure recovery has occurred) of 1 m/s, the dynamic pressure will account for <4 mm Hg of the total pressure head.

**Percent left ventricular stroke work loss** (the difference between the total and effective stroke work expressed as a percentage of total stroke work) can therefore be calculated from equations (5) and (4) as:

$$\text{Percent left ventricular stroke work loss} = \left( \frac{\text{mean } P_{LV} \times SV - \text{mean } P_{aorta} \times SV}{\text{mean } P_{LV} \times SV} \right) \times 100\%.$$  \hfill (5)

The stroke volume terms cancel out, giving the expression:

$$\text{Percent left ventricular stroke work loss} = \left( \frac{\text{mean aortic valve gradient}}{\text{mean LV systolic pressure}} \right) \times 100\%.$$  \hfill (6)

where $LV = \text{left ventricular}$.

**Methods**

Aortic valves. Thirteen aortic valves were studied (nine human aortic valves obtained at necropsy and four bioprosthetic valves). Clinical and pathologic data are summarized in Table 1. Valves 1 to 3 were obtained from patients in whom a diagnosis of severe aortic stenosis had been made during their lifetime. In valves 1 and 3 the stenosis was due to degenerative-calcific disease; valve 2 was a calcified bicuspid valve. Valves 10 and 11 were explanted Carpenter-
Figure I. Schematic diagram of the Sheffield pulse duplicator. A = aorta; A/D = analog to digital converter; Ap = aortic pressure transducer; At = atrium; AV = aortic valve; DA = digital to analog converter; EMF = electromagnetic flow meter; FCV = flow control valve; MV = mitral valve; Pc = piston position control; S = flow straighteners; V = ventricle; VC = video camera; VP = ventricular pressure transducer.

Edwards porcine prostheses: valve 10 was explanted at operation because of dysfunction with stenosis and valve 11 had functioned normally.

The human aortic valves were prepared for mounting in the pulse duplicator as follows: the valve, together with a cuff of aortic root and left ventricular outflow tract, was excised and stored in a saline/antibiotic solution at 4°C. It was then trimmed and sewn to a base ring with stents to support the aortic root. After study the valves were radiographed in an axial projection and the degree of calcification of the cusps graded on a semiquantitative scale (12).

Pulsatile flow model. A computer controlled positive displacement pulse duplicator was used (Fig. 1) that has been described before (13). The cuboidal atrium was constructed of Plexiglas and was open to atmosphere. It contained a cylindrical Plexiglas ventricular chamber, closed at one end by the piston of the drive unit and at the other by the aortic valve. Miniature strain gauge pressure transducers (type 3EA: Gaeltec) were located in the ventricular wall 25 mm upstream from the plane of the valve sewing ring and in the aorta 100 mm downstream. Aortic flow was measured by an electromagnetic flowmeter (Gould SP2202: Spectramed) 200 mm downstream to the valve. The precision of the pressure transducers was ±1 mm Hg and of the flowmeter, ±5 ml/s.

The cycle rate was constant at 70 cycles/min (systolic duration 280 ms). Systole was defined as the period from the time at which ventricular pressure exceeded aortic pressure until the end of forward flow. Valves were studied at three to five flow rates with stroke volumes of 40 to 100 ml (cardiac output 2.8 to 7 liters/min) giving 57 data points. Each run consisted of 20 dummy cycles followed by 10 cycles during which pressure and flow data were recorded at 5 ms intervals. Mean ventricular systolic pressure, mean systolic pressure gradient across the aortic valve and mean systolic flow were calculated electronically and averaged over 10 cycles.

The system was filled with normal saline solution allowing videotape of valve motion to be recorded simultaneously with hydrodynamic data. Anatomic orifice area was measured by planimetry at the free edges of the valve cusps when maximally open, using the monitor and analysis software on an ultrasound imaging system (Hewlett-Packard model 77020A).

Gorlin formula. The Gorlin formula (2) relates the orifice area at the exit of a stenotic valve (OA) to the mean flow rate through the valve (Q) and the square root of the mean pressure gradient (MPG):

$$\text{OA} = \frac{Q}{C \times 44.3 \times (\text{MPG})^{1/2}},$$

where C is an empiric constant incorporating the coefficients of orifice contraction and flow velocity of the valve (assumed to be constant for a given valve site), a conversion factor and other unknown factors (2). This formula can be rewritten:

$$\text{OA} = K \times \frac{Q}{(\text{MPG})^{3/2}},$$

where K is the reciprocal of (C x 44.3).

Statistical analysis. Linear regression analysis was used to study the relations between anatomic valve orifice area and 1) percent ventricular stroke work loss (after logarithmic transformation) calculated from equation (6), and 2) the ratio of mean systolic flow divided by the square root of the mean systolic pressure gradient (equation (8)).

Agreement between predicted orifice area derived from the regression equations and anatomic orifice area was quantified by calculation of confidence intervals (14).

Results (Table 2)
Planimetry-derived valve orifice area ranged from 0.3 to 2.8 cm² and was directly related to flow, although the relation showed substantial variation among valves. The normal human valves and the bovine pericardial bioprostheses opened fully to give a round orifice, whereas the orifice of the porcine bioprostheses was trefoiled. The valves with severe degenerative-calcific stenosis showed markedly reduced cusp motion resulting in a triradiate orifice.

Pressure gradient and ventricular stroke work loss. Mean systolic pressure gradient ranged from 3 to 84 mm Hg (Fig. 2). Percent stroke work loss was 7% to 68% and showed an inverse exponential relation to anatomic orifice area (Fig. 3). It was not significantly related to flow (r = -0.15, p = 0.25). Linear regression of anatomic orifice area on the logarithm of percent stroke work loss is shown in Figure 4.
Table 2. Valve Orifice Area and Hydrodynamic Data

<table>
<thead>
<tr>
<th>Stroke Volume</th>
<th>Orifice Area</th>
<th>Mean Syst. Flow</th>
<th>Mean Syst. Gradient</th>
<th>Stroke Work Loss</th>
</tr>
</thead>
<tbody>
<tr>
<td>(ml)</td>
<td>(cm²)</td>
<td>(ml/s)</td>
<td>(mm Hg)</td>
<td>(mg)</td>
</tr>
<tr>
<td>1</td>
<td>40</td>
<td>0.31</td>
<td>127</td>
<td>31.1</td>
</tr>
<tr>
<td>55</td>
<td>0.37</td>
<td>176</td>
<td>44.5</td>
<td>26.4</td>
</tr>
<tr>
<td>70</td>
<td>0.42</td>
<td>219</td>
<td>57.0</td>
<td>29.0</td>
</tr>
<tr>
<td>85</td>
<td>0.41</td>
<td>266</td>
<td>70.3</td>
<td>31.7</td>
</tr>
<tr>
<td>100</td>
<td>0.41</td>
<td>317</td>
<td>83.8</td>
<td>34.6</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Stroke Volume</th>
<th>Orifice Area</th>
<th>Mean Syst. Flow</th>
<th>Mean Syst. Gradient</th>
<th>Stroke Work Loss</th>
</tr>
</thead>
<tbody>
<tr>
<td>(ml)</td>
<td>(cm²)</td>
<td>(ml/s)</td>
<td>(mm Hg)</td>
<td>(mg)</td>
</tr>
<tr>
<td>2</td>
<td>40</td>
<td>0.10</td>
<td>133</td>
<td>13.5</td>
</tr>
<tr>
<td>55</td>
<td>0.29</td>
<td>190</td>
<td>24.2</td>
<td>42.3</td>
</tr>
<tr>
<td>70</td>
<td>0.06</td>
<td>239</td>
<td>27.0</td>
<td>45.3</td>
</tr>
<tr>
<td>85</td>
<td>0.93</td>
<td>287</td>
<td>35.2</td>
<td>48.3</td>
</tr>
<tr>
<td>100</td>
<td>0.96</td>
<td>340</td>
<td>41.0</td>
<td>53.1</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Stroke Volume</th>
<th>Orifice Area</th>
<th>Mean Syst. Flow</th>
<th>Mean Syst. Gradient</th>
<th>Stroke Work Loss</th>
</tr>
</thead>
<tbody>
<tr>
<td>(ml)</td>
<td>(cm²)</td>
<td>(ml/s)</td>
<td>(mm Hg)</td>
<td>(mg)</td>
</tr>
<tr>
<td>3</td>
<td>40</td>
<td>0.51</td>
<td>136</td>
<td>35.5</td>
</tr>
<tr>
<td>55</td>
<td>0.81</td>
<td>182</td>
<td>40.1</td>
<td>28.6</td>
</tr>
<tr>
<td>70</td>
<td>0.69</td>
<td>234</td>
<td>50.4</td>
<td>31.6</td>
</tr>
<tr>
<td>85</td>
<td>0.72</td>
<td>274</td>
<td>63.2</td>
<td>34.5</td>
</tr>
<tr>
<td>100</td>
<td>0.82</td>
<td>326</td>
<td>75.4</td>
<td>37.3</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Stroke Volume</th>
<th>Orifice Area</th>
<th>Mean Syst. Flow</th>
<th>Mean Syst. Gradient</th>
<th>Stroke Work Loss</th>
</tr>
</thead>
<tbody>
<tr>
<td>(ml)</td>
<td>(cm²)</td>
<td>(ml/s)</td>
<td>(mm Hg)</td>
<td>(mg)</td>
</tr>
<tr>
<td>4</td>
<td>40</td>
<td>0.70</td>
<td>134</td>
<td>13.8</td>
</tr>
<tr>
<td>55</td>
<td>1.30</td>
<td>240</td>
<td>20.6</td>
<td>52.9</td>
</tr>
<tr>
<td>70</td>
<td>1.12</td>
<td>337</td>
<td>33.6</td>
<td>58.1</td>
</tr>
<tr>
<td>85</td>
<td>1.48</td>
<td>124</td>
<td>8.4</td>
<td>46.1</td>
</tr>
<tr>
<td>100</td>
<td>1.45</td>
<td>340</td>
<td>36.5</td>
<td>66.0</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Stroke Volume</th>
<th>Orifice Area</th>
<th>Mean Syst. Flow</th>
<th>Mean Syst. Gradient</th>
<th>Stroke Work Loss</th>
</tr>
</thead>
<tbody>
<tr>
<td>(ml)</td>
<td>(cm²)</td>
<td>(ml/s)</td>
<td>(mm Hg)</td>
<td>(mg)</td>
</tr>
<tr>
<td>5</td>
<td>40</td>
<td>0.94</td>
<td>132</td>
<td>9.9</td>
</tr>
<tr>
<td>55</td>
<td>—</td>
<td>—</td>
<td>—</td>
<td>—</td>
</tr>
<tr>
<td>70</td>
<td>1.04</td>
<td>236</td>
<td>19.5</td>
<td>53.5</td>
</tr>
<tr>
<td>85</td>
<td>1.12</td>
<td>337</td>
<td>33.6</td>
<td>58.1</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Stroke Volume</th>
<th>Orifice Area</th>
<th>Mean Syst. Flow</th>
<th>Mean Syst. Gradient</th>
<th>Stroke Work Loss</th>
</tr>
</thead>
<tbody>
<tr>
<td>(ml)</td>
<td>(cm²)</td>
<td>(ml/s)</td>
<td>(mm Hg)</td>
<td>(mg)</td>
</tr>
<tr>
<td>6</td>
<td>40</td>
<td>1.48</td>
<td>124</td>
<td>8.4</td>
</tr>
<tr>
<td>55</td>
<td>—</td>
<td>—</td>
<td>—</td>
<td>—</td>
</tr>
<tr>
<td>70</td>
<td>1.62</td>
<td>236</td>
<td>14.1</td>
<td>62.9</td>
</tr>
<tr>
<td>85</td>
<td>1.76</td>
<td>341</td>
<td>20.4</td>
<td>75.4</td>
</tr>
<tr>
<td>100</td>
<td>1.28</td>
<td>338</td>
<td>20.0</td>
<td>75.5</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Stroke Volume</th>
<th>Orifice Area</th>
<th>Mean Syst. Flow</th>
<th>Mean Syst. Gradient</th>
<th>Stroke Work Loss</th>
</tr>
</thead>
<tbody>
<tr>
<td>(ml)</td>
<td>(cm²)</td>
<td>(ml/s)</td>
<td>(mm Hg)</td>
<td>(mg)</td>
</tr>
<tr>
<td>7</td>
<td>40</td>
<td>0.98</td>
<td>133</td>
<td>9.6</td>
</tr>
<tr>
<td>55</td>
<td>—</td>
<td>—</td>
<td>—</td>
<td>—</td>
</tr>
<tr>
<td>70</td>
<td>1.20</td>
<td>236</td>
<td>15.3</td>
<td>60.2</td>
</tr>
<tr>
<td>85</td>
<td>1.28</td>
<td>338</td>
<td>20.0</td>
<td>75.5</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Stroke Volume</th>
<th>Orifice Area</th>
<th>Mean Syst. Flow</th>
<th>Mean Syst. Gradient</th>
<th>Stroke Work Loss</th>
</tr>
</thead>
<tbody>
<tr>
<td>(ml)</td>
<td>(cm²)</td>
<td>(ml/s)</td>
<td>(mm Hg)</td>
<td>(mg)</td>
</tr>
<tr>
<td>8</td>
<td>40</td>
<td>1.25</td>
<td>136</td>
<td>5.6</td>
</tr>
<tr>
<td>55</td>
<td>1.32</td>
<td>190</td>
<td>6.7</td>
<td>73.4</td>
</tr>
<tr>
<td>70</td>
<td>1.55</td>
<td>234</td>
<td>8.2</td>
<td>82.1</td>
</tr>
<tr>
<td>85</td>
<td>1.65</td>
<td>287</td>
<td>10.5</td>
<td>88.6</td>
</tr>
<tr>
<td>100</td>
<td>1.77</td>
<td>352</td>
<td>13.0</td>
<td>97.8</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Stroke Volume</th>
<th>Orifice Area</th>
<th>Mean Syst. Flow</th>
<th>Mean Syst. Gradient</th>
<th>Stroke Work Loss</th>
</tr>
</thead>
<tbody>
<tr>
<td>(ml)</td>
<td>(cm²)</td>
<td>(ml/s)</td>
<td>(mm Hg)</td>
<td>(mg)</td>
</tr>
<tr>
<td>9</td>
<td>40</td>
<td>1.01</td>
<td>139</td>
<td>6.7</td>
</tr>
<tr>
<td>55</td>
<td>1.03</td>
<td>182</td>
<td>8.7</td>
<td>61.9</td>
</tr>
<tr>
<td>70</td>
<td>1.10</td>
<td>220</td>
<td>11.2</td>
<td>65.9</td>
</tr>
<tr>
<td>85</td>
<td>1.20</td>
<td>272</td>
<td>15.7</td>
<td>69.7</td>
</tr>
<tr>
<td>100</td>
<td>1.28</td>
<td>334</td>
<td>18.5</td>
<td>77.1</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Stroke Volume</th>
<th>Orifice Area</th>
<th>Mean Syst. Flow</th>
<th>Mean Syst. Gradient</th>
<th>Stroke Work Loss</th>
</tr>
</thead>
<tbody>
<tr>
<td>(ml)</td>
<td>(cm²)</td>
<td>(ml/s)</td>
<td>(mm Hg)</td>
<td>(mg)</td>
</tr>
<tr>
<td>10</td>
<td>40</td>
<td>0.68</td>
<td>130</td>
<td>11.5</td>
</tr>
<tr>
<td>55</td>
<td>0.73</td>
<td>185</td>
<td>19.7</td>
<td>41.8</td>
</tr>
<tr>
<td>70</td>
<td>0.78</td>
<td>231</td>
<td>29.0</td>
<td>43.2</td>
</tr>
<tr>
<td>85</td>
<td>0.82</td>
<td>290</td>
<td>42.8</td>
<td>44.3</td>
</tr>
<tr>
<td>100</td>
<td>0.87</td>
<td>338</td>
<td>56.8</td>
<td>44.8</td>
</tr>
</tbody>
</table>

The regression equation for orifice area (OA) is OA (cm²) = 4.82 - (2.39 × log percent stroke work loss) (r = -0.95, SEE = 0.029, p < 0.00001). Stroke work loss data predicted anatomic orifice area with a 95% prediction interval of ±0.5 cm².

Gorlin formula. Mean systolic flow divided by the square root of the mean systolic pressure gradient: Syst. = systolic.

The regression equation for orifice area (OA) is:

\[ OA \text{ (cm}^2) = 0.02 + 0.020 \times \frac{\text{mean systolic flow}}{\sqrt{\text{mean pressure gradient}}} \]

Figure 2. Plot of valve orifice area against mean systolic pressure gradient. Stroke volume was varied from 40 to 100 ml. There is substantial scatter, most marked with orifice areas <1.0 cm², reflecting the flow dependence of the relation.
Figure 3. Plot of valve orifice area against percent ventricular stroke work loss. An exponential curve has been fitted to the points.

\( r = 0.93, \text{ SEE } 0.033, p < 0.00001 \). The regression equation predicted anatomic orifice area with a 95% prediction interval of \( \pm 0.5 \, \text{cm}^2 \).

**Discussion**

The results obtained in this pulsatile flow model suggest that percent ventricular stroke work loss may be a clinically useful index of the severity of aortic stenosis. Percent stroke work loss showed a close inverse relation to anatomic aortic valve orifice area in both normal and stenotic human valves and in bioprostheses and was not significantly affected by changes in stroke volume over the range of 40 to 100 ml.

Comparison with previous studies. There have been few reports on the measurement of ventricular stroke work loss in patients with aortic stenosis and no previous studies in which valve orifice area could be measured directly. Tobin et al. (8) compared percent stroke work loss with orifice area calculated by the Gorlin formula in 49 patients aged 2 months to 73 years and found an inverse relation \( r = -0.79 \). Percent stroke work loss was relatively unaffected by changes in cardiac output brought about by exercise or anesthesia.

Tobin et al. (8) also gave data on percent ventricular stroke work loss in two open chest dogs in which aortic stenosis was simulated by supravalvular constriction of the aorta. Cardiac output was varied by opening one or two femoral arteriovenous fistulas giving a total of 24 data points. Linear regression of aortic area on the logarithm of percent stroke work loss yields a regression equation closely similar to that obtained from the data of this study \( y = 4.59 - 2.31 x; r = -0.83, \text{ SEE } 0.144; p < 0.00001 \). The slopes of the regression lines are closely similar (difference between the slopes -0.08, 95% confidence interval -0.67 to +0.51). This result is strong evidence that the inverse exponential relation observed between percent stroke work loss and aortic orifice area is of biologic significance and not an artifact of the model.

**Prediction interval of the orifice formula.** The empirically derived orifice formula predicted anatomic orifice area with a 95% prediction interval of \( \pm 0.5 \, \text{cm}^2 \); this interval is wide but is equal to the 95% prediction interval of the Gorlin formula in this model. The size of these intervals may reflect the inherent limitations in the accuracy with which any formula can predict anatomic orifice area from hemodynamic data (15). As the results show, anatomic orifice area in aortic stenosis is flow dependent; thus, full characterization of a valve requires measurement of orifice area over a range
of flows. Additionally, the relation between the area through which flow occurs and the anatomic orifice area (expressed in the coefficient of orifice contraction) depends on the geometry of the orifice (16,17), which differs according to the cause of aortic stenosis (18) and among native, bioprosthetic and mechanical valves. Orifice geometry also influences the degree of pressure recovery and hence the pressure gradient across the valve (11,19).

Possible advantages over the Gorlin formula. The Gorlin formula has been widely adopted as the method of choice for the invasive measurement of valve orifice area (6). In our model anatomic orifice area bore a close linear relation to mean systolic flow divided by the square root of the mean systolic flow divided by the square root of the mean systolic pressure gradient, as predicted by the Gorlin formula. In clinical practice, however, inaccuracy in the measurement of mean systolic flow is an important source of error in valve areas calculated from the Gorlin formula. The Fick method is regarded as the reference standard for measurement of cardiac output but may be unreliable: duplicate determinations of cardiac output in 61 patients showed a median error of 8.6%, but in 8 patients (13%) the error was >19% (20).

There are limited data on the reproducibility of valve area calculated by the Gorlin formula in patients with aortic stenosis. In 28 elderly patients who underwent cardiac catheterization on two separate occasions (with cardiac output measured by the Fick method), the agreement between valve area calculations in the two studies was poor (r = 0.42), although the reasons for this were not established (21). The major practical advantage of percent stroke work loss as an index of severity of aortic stenosis is that it can be calculated from pressure data alone.

Potential limitations of this study. Although the model does not reproduce all features of the normal or pathologic human circulation, the size and shape of the ventricular outflow tract and aorta are similar to those in the adult human and the flow waveform generated by the ventricle is quasiphysiologic. To allow simultaneous recording of valve motion and hydrodynamic data, the model was filled with saline solution, which has a lower viscosity than that of blood. We do not believe that this difference had a significant effect: studies in in vitro models have shown that the pressure gradient is effectively independent of fluid viscosity over the range spanning that of saline solution and blood (22,23). We defined anatomic valve orifice area as maximal rather than mean orifice area. Because the total time taken for the aortic valve to open and close is <40 ms (24), with a systolic duration of 280 ms maximal and mean orifice areas could have been similar.

Conclusions and clinical applications. In this model percent ventricular stroke work loss can be used to predict anatomic orifice area in native aortic valves and bioprostheses with an accuracy of ±0.5 cm². Further studies are needed to establish the accuracy and reproducibility of the method in patients and the clinical correlates of different degrees of stroke work loss. Many catheterization laboratories have the computing capability to allow on-line calculation of percent stroke work loss from left ventricular and aortic pressure measurements. This capability and the relative flow independence of the index would make it a particularly useful measurement in aortic balloon valvuloplasty when serial estimation of the severity of stenosis is required under changing hemodynamic conditions.

References

16. Segal J, Lerner DJ, Miller DC, Mitchell RS, Alderman EA, Popp RL. When should Doppler-determined valve area be better than the Gorlin


