JACC Vol. 21, No. 4
March 15, 1993:1018–25

Effect of Stenosis Geometry on the Doppler-Catheter Gradient Relation In Vitro: A Manifestation of Pressure Recovery

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Objectives. This study investigated the effect of stenosis geometry on the Doppler-catheter gradient relation.

Background. Although gradient estimation by Doppler ultrasound has been shown to be accurate in various clinical and in vitro settings, there have also been reports of substantial discrepancies between Doppler and catheter gradients. These conflicting results may be due to differences in geometry and hemodynamic characteristics of flow obstructions.

Methods. Stenoses of various geometry were simultaneously studied with continuous wave Doppler and catheter technique in a well controlled pulsatile flow model.

Results. Doppler and catheter gradients correlated very well regardless of stenosis geometry and site of distal catheter measurement (r = 0.98 to 0.99, SEE = 1.8 to 5.3 mm Hg). When the catheter was pulled back through the stenosis, the highest gradients were found in or close to the stenosis. When these catheter gradients were compared with Doppler gradients, the agreement between the two techniques was excellent regardless of stenosis geometry (slope 0.97; mean difference 0.6 ± 2.0 mm Hg). However, when distal pressures were measured 10 cm downstream from the stenotic segment, the slope of the regression line, and therefore the agreement between Doppler and catheter gradients, differed for the different stenosis types (slopes from 0.98 to 1.69). In stenoses with abrupt narrowing and abrupt expansion, agreement was acceptable. Doppler gradients were only slightly greater than catheter gradients (mean difference 4.5 ± 5.2 mm Hg). In stenoses with a gradually tapering inlet and outlet, the Doppler-catheter gradient relation was dependent on the outflow angle. Good agreement was found for an angle of 60° (mean difference 0.6 ± 1.8 mm Hg). In stenoses with a 40° outflow angle, Doppler gradients exceeded the catheter gradients by 13% on average; for stenoses with a 20° outflow angle, Doppler gradients exceeded catheter gradients by 46 ± 11.4%, with differences as great as 65 mm Hg. These results were identical for stenoses gradually tapering outward to the distal tubing diameter and those with abrupt expansion after 2 cm of gradual expansion. The results were also not affected by changing the inflow angle from 20° to 60°. However, an abrupt narrowing instead of a tapering inlet significantly altered the Doppler-catheter gradient relation ($p < 0.001$); Doppler gradients exceeded the catheter gradients by 34 ± 10% for this stenosis type.

Conclusions. Doppler gradients accurately reflect the highest gradients across flow obstructions that occur in the vena contracta. However, these gradients may be significantly greater than catheter gradients that are measured further downstream, as is usually the case in clinical catheterization studies. These discrepancies are due to pressure recovery. The magnitude of pressure recovery is highly dependent on the stenosis geometry, which therefore significantly affects the Doppler-catheter gradient relation. It is the outflow geometry that predominantly influences this relation, but the shape of the inlet may affect the results as well. Although pressure recovery occurs even in stenoses with abrupt narrowing and abrupt expansion, the phenomenon is most likely to become clinically relevant in stenoses with a gradually tapering inlet and outlet with an outflow angle ≤20°.

(J Am Coll Cardiol 1993;21:1018–25)

The pressure drop across a flow obstruction is a key variable in the assessment of various cardiovascular disorders such as valvular stenoses, prosthetic valves, intraventricular obstructions, shunts and vascular stenoses. Doppler echocardiography has been used as a noninvasive technique for estimating gradients from velocity measurements across the stenoses using the simplified Bernoulli equation (1,2). The accuracy of this technique has been evaluated in numerous studies. Many of these studies (3–10) have reported excellent agreement between Doppler and catheter gradients. In some settings such as prosthetic valves (11,12), hypertrophic obstructive cardiomyopathy (13) and aortic coarctation (14), however, there have also been reports of substantial disagreement between Doppler and catheter gradients. Pressure recovery—the increase of pressure distal to the stenosis—has recently been shown to be a potential cause of the discrepancy between Doppler and catheter gradients (12,15). Differences in stenosis geometry may be the major

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Manuscript received May 20, 1992; revised manuscript received September 3, 1992, accepted September 8, 1992.

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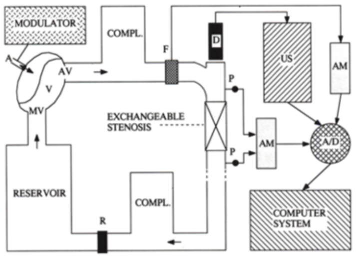

Figure 1. Diagram of the pulse duplicator. A = pressurized air; A/D = analog to digital converter; AM = amplifier; AV = aortic valve; COMPL. = compliance chamber; D = Doppler probe; F = flow probe; MV = mitral valve; P = pressure transducer; R = resistance; US = ultrasound device; V = ventricle.

reason that this phenomenon becomes clinically important in some settings but usually does not cause significant disagreement between Doppler and catheter gradients in others. These differences may result in different flow characteristics and a different magnitude of pressure recovery. Although significant pressure recovery has been demonstrated for discrete membranelike stenoses, it may be particularly important in tunnel-like stenoses (15,16). However, the exact stenosis geometry in which it can be expected to cause clinically relevant differences between Doppler and catheter gradients has not been well defined.

To investigate the effect of stenosis geometry and catheter location on the Doppler-catheter gradient relation, we studied stenoses of various geometry simultaneously with continuous wave Doppler and catheter techniques in a well controlled pulsatile flow model. The study attempts to define basic geometric stenosis variables that may predict clinically relevant differences between Doppler and catheter gradients due to pressure recovery.

Methods

Flow model (Fig. 1). The flow model consists of a ventricle, Lucite tubing, compliance chambers and a reservoir (Fig. 1). The ventricle (Vienna elliptic heart type, 100 ml) is pneumatically driven and passively filled from the reservoir. This pulse generator has successfully been used in previous Doppler in vitro studies (17,18). Stroke volume is adjustable from 0 to 100 ml, ejection pressure from 0 to 300 mm Hg and pulse rate from 40 to 120 beats/min. The ejection time can be varied from 100 to 700 ms. Inlet and outlet valves are Bjork-Shiley prostheses (size 23 mm).

The test section has been designed to allow interposition of various stenotic segments and optimal alignment of the Doppler beam with the flow across the stenosis. The inner

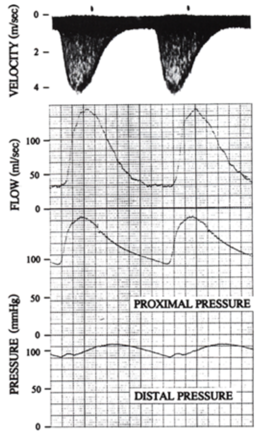

Figure 2. Examples of simultaneous Doppler velocity, flow and pressure recordings.

diameter of upstream and downstream tubings is 2.4 cm. Pressure taps may be connected to fluid-filled catheters but also allow insertion of tubing for measurement of pullback pressures. Pressures can be adjusted by varying proximal and distal compliance and resistance.

The test fluid in the present study was a 70% water-30% glycerol solution with 10 g/liter cornstarch and 4 g/liter sodium chloride (viscosity 3.5 centipoise).

Flow was measured with an electromagnetic flow meter (Cliniflow, Carolina Medical Electronics Inc.) that was calibrated with a geared pump. The flow probe was attached between the proximal compliance chamber and test section.

Pressures were measured with fluid-filled catheters of matched length and electronic pressure transducers (monitoring set, Peter van Berg).

Continuous wave Doppler measurements were performed with a Vingmed CFM 750 (Vingmed Sound A/S) using a Duplex probe (3.5-MHz imaging, 2.5-MHz continuous wave Doppler transducer). The ultrasound probe was carefully adjusted to record the highest Doppler velocities and could be fixed in place with a clamp system.

Pressure transducers and the flow meter were connected to a four-channel physiologic recording system (Hellige GmbH) and Doppler velocities were recorded on paper (Videographic Printer YP 1810) and videotape. A typical example of pressure, flow and Doppler recordings is shown in Figure 2. For further calculation, the Doppler signals and the analog signals from the differential pressure amplifier and

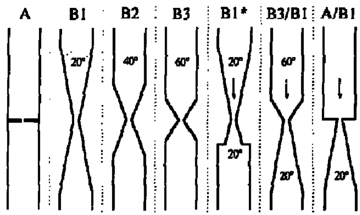

Figure 3. Types of stenosis studied (arrows indicate the flow direction). See text for definitions.

electromagnetic flow meter were fed to an analog to digital converter and transferred to a computer system (Macintosh IIci, Apple Computer GmbH). Peak catheter gradients, peak flow and peak Doppler gradients were calculated. Peak catheter gradient was defined as the maximal instantaneous difference between the proximal and distal pressure. Doppler gradients (Δp) were calculated from the maximal instantaneous ultrasound velocity (v) with the simplified Bernoulli equation ($\Delta p = 4v^2$). The proximal velocities as calculated from flow rate and tubing size ranged from 0.07 to 0.39 m/s and were therefore neglected.

Each set of measurements was obtained by averaging the calculations of three consecutive beats.

Stenoses. The different types of stenoses are shown in Figure 3. Stenoses with a diameter of 0.76, 0.55 and 0.34 cm were built. The corresponding areas were 0.45 cm^2 (90% stenosis), 0.24 cm^2 (95%) and 0.09 cm^2 (98%).

Membranelike (type A) and hourglasslike stenoses with inflow and outflow angles of 20° (type B_1) were built in all three sizes. The other types were only studied with a diameter of 0.55 cm. Type B_2 and B_3 stenoses had angles of 40° and 60°, respectively. In addition to these symmetrically shaped hourglasslike stenoses, asymmetric stenoses with an inflow angle of 20° and outflow angle of 60° (type B_1/B_3), an inflow angle of 60° and outflow angle of 20° (type B_3/B_1) and an inflow angle of 180° and outflow angle of 20° (A/B_1) were studied. Type B_1* stenoses were type B_1 stenoses modified by opening the outflow segment after a length of 2 cm abruptly to the distal tubing diameter of 2.4 cm.

Test protocol. *1) In type A, B_1, B_2 and B_3 stenoses with a 0.55-cm diameter,* pressure gradients were measured between a proximal end-hole catheter (3 cm upstream from the stenotic segment) and a side-hole catheter (Lectrocath PE, Vygon, Ecouen, France) that was pulled through the stenosis. The side-hole catheter was initially moved to the level of the end-hold catheter, making sure that identical pressures were measured with both catheters. The side-hole catheter was then pulled back through the stenosis to define the location of the catheter port where the highest obtainable catheter gradient was found. The catheter was left at this site and the flow rate was varied in five steps from 70 to 120 ml/s peak flow. At each flow rate, Doppler and catheter gradients were simultaneously measured.

2) All stenoses were studied at eight different flow rates, while taking pressures at taps 3 cm upstream and 10 cm downstream from the stenotic segment. This distance was chosen to allow complete pressure recovery (19). Depending on the stenosis diameter, peak flow rates ranged from 30 to 175 ml/s. Proximal pressure was maintained between 125 and 205 mm Hg systolic pressure, 65 and 125 mm Hg diastolic pressure and 100 and 135 mm Hg mean pressure. Distal pressure was maintained between 70 and 135 mm Hg systolic pressure, 50 and 105 mm Hg diastolic pressure and 60 and 115 mm Hg mean pressure. Pulse rate was maintained at 60 beats/min, with an ejection time of 350 ms. At each flow rate, Doppler and catheter gradients were simultaneously measured.

Statistical analysis. The Doppler-catheter gradient relation was assessed by linear regression analysis and Pearson correlation coefficients were calculated. The hypotheses about two regression lines was tested with a two-tailed t test. To assess the agreement between Doppler and catheter gradients, mean differences and SD were also calculated.

Results

Catheter pullback measurements. When the distal catheter was pulled back through the stenosis, the highest gradient was found in the stenosis itself for hourglasslike stenoses and approximately 2-cm distal to the membranelike stenosis. With further pull back of the catheter, the greatest increase in pressure distal to the stenosis was found in type B_1 stenoses, reflected by a 32% decrease in gradient at 10 cm compared with the highest gradient at the stenosis. The gradient decreased only slightly by 13% in type B_2 stenoses when the catheter port was moved from the stenosis to a 10-cm distance and even less in type B_3 and A stenoses (8% and 7%, respectively).

With the catheter port at the site of the highest obtainable catheter gradient, excellent agreement between Doppler and catheter gradients was found regardless of the stenosis geometry (Fig. 4). The mean difference between Doppler and catheter gradients was 0.6 ± 2 mm Hg. The slope of the regression line was not statistically different from 1.

Relation between Doppler and catheter gradients (distal pressure measured 10 cm downstream from the stenotic segment). For all stenosis geometries, Doppler and catheter gradients correlated very well (r = 0.98 to 0.99; SEE = 1.8 to 5.3 mm Hg). However, the relation between Doppler and catheter gradients differed substantially for the various types of stenoses as shown by a variation of slopes from 0.98 to 1.69 (Table 1).

Type A stenoses. In membranelike stenoses, acceptable agreement between Doppler and catheter gradients was found. Doppler gradients were only slightly greater than catheter gradients, with a mean difference of 4.5 ±

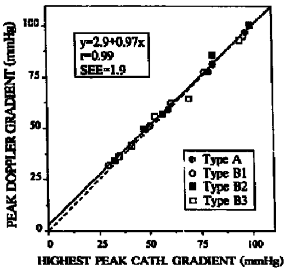

Figure 4. Correlation of the highest obtainable pressure gradient (catheter [CATH.] pullback) with the Doppler gradient for type A, B_1, B_2 and B_3 stenoses (**dashed line** represents the line of identity).

5.2 mm Hg. The slope of the regression line was not statistically different from 1 (Fig. 5).

Type B stenoses with symmetric shape. In hourglasslike stenoses, the relation between Doppler and catheter gradients was dependent on the outflow angle (Fig. 5). In type B_3 stenoses with an outflow angle of 60°, Doppler and catheter gradients differed by only 0.6 ± 1.8 mm Hg (slope not statistically different from 1). In type B_2 stenoses with an angle of 40°, Doppler gradients slightly exceeded the catheter gradients by 13 ± 4% (mean difference 7.9 ± 2.9 mm Hg). The results were statistically different from those for the type B_3 stenosis ($p < 0.01$). In type B_1 stenoses with angles of 20°, the greatest differences between Doppler and catheter gradients were found (Doppler gradient = 1.6 × catheter gradient − 5.5 mm Hg). Doppler gradients exceeded catheter gradients by 46 ± 11% (27.6 ± 18 mm Hg), with differences as great as 65 mm Hg. The differences increased with higher gradients. In the modified B_1* stenoses that opened abruptly after 2 cm of gradual expansion (outflow angle 20°), the Doppler-catheter gradient relation was not statistically different from that in original B_1 stenoses ($p = 0.1$). Doppler

Table 1. Correlation Between Doppler Gradients and Catheter Gradients 10 cm Distal to the Stenosis

Type	r Value	SEE (mm Hg)	Slope	y Intercept
A	0.98	5.3	0.99	5.5
B_1	0.99	3.1	1.60	−5.6
B_2	0.99	2.0	1.07	3.7
B_3	0.99	1.8	0.98	1.2
B_1/B_3	0.99	2.1	1.00	3.7
B_3/B_1	0.99	2.3	1.55	−4.8
A/B_1	0.99	3.2	1.19	9.3
B_1*	0.99	4.7	1.69	−10.9

See text for definitions of types of stenoses.

gradients exceeded catheter gradients by 48 ± 10% (32.9 ± 19.4 mm Hg).

Type B stenoses with asymmetric shape: effect of inflow geometry. For a given outflow angle, the Doppler-catheter gradient relation was not affected by variation of the inflow angle from 20° to 60° (Fig. 6, Table 1). For the B_3/B_1 stenoses (outflow angle 20°), the results were not statistically different from those for B_1 stenoses ($p > 0.5$) and Doppler gradients exceeded catheter gradients by 46 ± 7% (30.3 ± 15.7 mm Hg). For the B_1/B_3 stenoses (outflow angle 60°), the results were not statistically different from those for the B_3 stenoses ($p = 0.2$). Doppler and catheter gradients differed by only 4.3 ± 1.9 mm Hg. Therefore, for stenoses with a tapering inlet, the inflow angle did not affect the results. However, an abrupt narrowing instead of a gradually tapering inlet (A/B_1 stenoses) altered the Doppler-catheter gradient relation (Fig. 6). Doppler gradients differed significantly less from catheter gradients as compared with the results for B_1 and B_3/B_1 stenoses (Doppler gradient = 1.19 × catheter gradient + 9.3 mm Hg, $p < 0.001$). Doppler gradients still exceeded catheter gradients by 34 ± 10% (24.7 ± 7.3 mm Hg).

Reynolds numbers. To assess the importance of viscous effects, Reynolds numbers (Re) were calculated with the equation:

$$Re = \frac{\text{Fluid density} \times \text{Velocity} \times \text{Vessel diameter}}{\text{Fluid viscosity}}.$$

The peak Reynolds numbers calculated on the basis of the stenosis diameters ranged from 2,900 to 10,200. In the region distal to the stenosis, they ranged from 12,100 to 47,000 (calculated on the basis of continuous wave Doppler velocities). All Reynolds numbers were >2,300, indicating turbulent flow conditions. Therefore, viscous losses can be expected to be quite low (16,20).

Discussion

Pressure gradient estimation by Doppler ultrasound. Gradient estimation by Doppler ultrasound using the Bernoulli equation has been shown to be fairly accurate in various clinical and in vitro settings, including valvular stenoses (3–5), prosthetic valves (7,8,21,22), hypertrophic obstructive cardiomyopathy (9) and aortic coarctation (10). Substantial discrepancies between Doppler and catheter gradients have only been encountered under rare circumstances. Two potential sources of error by Doppler ultrasound were usually proposed to explain these discrepancies. First, underestimation by Doppler technique can be the result of malalignment of Doppler beam and blood flow because the angle in the Doppler equation is assumed to be zero. Second, overestimation by Doppler ultrasound may occur when the velocity proximal to the stenosis is high (significantly >1 m/s) because this velocity has generally been neglected using the most simplified modification of the Bernoulli equation (Δp =

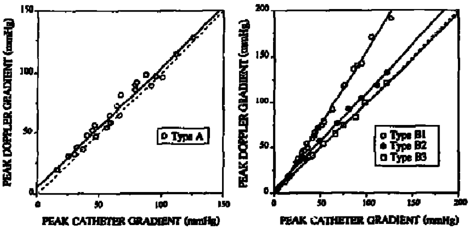

Figure 5. Correlation between Doppler and catheter gradients (10 cm distal to the stenotic segment) for symmetric stenoses (dashed line represents the line of identity). **Left panel,** Type A stenoses (abrupt narrowing and abrupt expansion). **Right panel,** Type B stenoses (hourglasslike stenoses with inflow and outflow angles of 20° [B_1], 40° [B_2] and 60° [B_3]).

$4v^2$). Nevertheless, overestimation by Doppler technique that cannot be explained by this mechanism has been reported in several studies in settings such as prosthetic valves (11,23–25), aortic coarctation (14) and hypertrophic cardiomyopathy (13). In the present study, the Doppler-catheter gradient relation varied significantly for the different types of stenoses. For some, good agreement between Doppler and catheter gradients was found, whereas slight or even substantial overestimation by Doppler ultrasound was observed for others. With use of an in vitro model, proximal velocities were known to be very low. This source of error could therefore be excluded.

Recent studies (12,15,16) have revealed the importance of the spatial variability of pressure fields as a cause of the discrepancy between Doppler and catheter gradients. The site of distal pressure measurement has been shown to be of particular importance in this situation. Spatial variability of pressure fields can be due to the complex three-dimensional geometry of the flow obstruction. This, for example, is the case in the St. Jude mechanical prosthetic valve, where pressures are significantly lower in the central orifice between the two leaflets than in the two larger side orifices (12). The second main cause of spatial variation of pressure is the increase in pressure with increasing distance from the stenosis (12,15,16,20). Continuous wave Doppler ultrasound records the highest velocity along the line of interrogation and therefore provides the highest pressure gradient along the path of the beam. In contrast, when catheters are used to assess pressure gradients, distal pressures are measured only at the site of the sampling port. Assuming a spatial variation of pressure, the measured gradient will be influenced by the site of the distal pressure measurement and may differ from the Doppler gradient. The magnitude of this difference between Doppler and catheter gradients will depend on the extent of spatial pressure variation.

Principles of pressure recovery. The increase in pressure and the concomitant decrease in the pressure gradient at a distance from the stenosis are due to pressure recovery (20). Pressure recovery is based on the physical principle of the conservation of energy. As fluid is forced to flow through a stenosis, flow accelerates and kinetic energy, therefore, increases. Because the total amount of energy is constant, there has to be a corresponding decrease in potential energy (that is, lateral pressure). Where the velocity is highest (that is, in the vena contracta), the pressure will be lowest, resulting in the highest pressure gradient. Downstream from the stenosis, flow velocity decreases with resultant reconversion of kinetic energy to potential energy. In an ideal system in which viscosity would be zero and no flow separation at the stenosis would occur, kinetic energy downstream from the stenosis would be completely reconverted to potential energy and pressure would fully recover. In reality, however, the extent of pressure recovery will be significantly reduced because viscosity needs to be considered and turbulences and some conversion of kinetic energy to heat do occur.

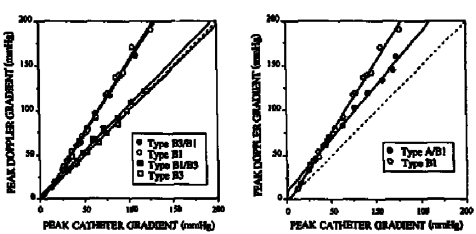

Figure 6. Correlation between Doppler and catheter gradients (10 cm distal to the stenotic segment) for asymmetric stenoses (dashed line represents the line of identity). **Left panel,** Type B_3/B_1 stenosis (inflow 60°, outflow 20°) and Type B_1/B_3 stenosis (inflow 20°, outflow 60°). Symmetric stenoses with angles of 20° (B_1) and 60° (B_3) are shown for comparison. **Right panel,** Type A/B_1 stenoses (abrupt narrowing and gradual expansion with an outflow angle of 20°). Symmetric stenoses with 20° inflow and outflow angle (B_1) are shown for comparison.

Clark (19,26) demonstrated the occurrence of pressure recovery in vitro and in an animal model of aortic stenosis many years ago. Nevertheless, good agreement between Doppler and catheter gradients has been reported (3–5) for aortic stenosis, although the potential difference between the highest gradient in the vena contracta (that is, Doppler gradient) and a gradient based on distal measurement of a more or less recovered pressure has been ignored. An explanation may be that in aortic stenosis, the magnitude of pressure recovery is usually too small to cause clinically significant differences. In the present study, distal pressures were taken 10 cm downstream from the stenotic segment. At this site, pressure should have recovered to its full extent (19). Nevertheless, the agreement between Doppler and catheter gradients for type A stenoses was still acceptable for clinical purposes, although Doppler gradients were slightly greater than catheter gradients (mean difference = 4.5 mm Hg), suggesting some pressure recovery. The actual extent of pressure recovery for this type of stenosis depends on flow rate, fluid density and the ratio of stenosis flow area to downstream flow area (19,26).

In the present study, catheter pullback measurements confirmed that Doppler gradients indeed reflected the highest pressure gradient (that is, in the vena contracta). For the type A stenoses, the catheter gradient farther downstream decreased by 7% as a result of pressure recovery. This decrease compares favorably to the 7% to 8% decrease predicted by fluid mechanics equations for this scenario (26). Assuming that Doppler ultrasound measures the gradient in the vena contracta, whereas catheter gradients taken 10 cm downstream from the stenosis are based on a recovered distal pressure, the observed decrease in the initial gradient due to pressure recovery was 7.6% on average for all type A stenoses studied. In concordance with fluid mechanics theory, the extent of pressure recovery differed for the three stenosis diameters and was greatest (15.9%) for the largest orifice (theoretically predicted decrease in gradient assuming a discharge coefficient of 0.85 would be 14.7%) (26). Because larger orifices cause a smaller pressure decrease, the magnitude of pressure recovery in absolute terms remained small even for this type of stenosis, and the overall agreement between Doppler and catheter gradients appeared to be acceptable. The same may happen in clinical studies. In addition, pressure recovery in a clinical setting will be further reduced by the eccentricity of jets. Thus, although pressure recovery occurs to some extent even in discrete stenoses with abrupt narrowing and abrupt expansion, its magnitude is usually not clinically relevant.

Effect of stenosis geometry on pressure recovery. With discrete obstruction to flow, as in valvular stenoses (type A in the present study), the sudden expansion results in turbulent mixing and the loss of kinetic energy by dissipation to heat, limiting pressure recovery. If the expansion is gradual rather than abrupt, the occurrence of turbulences and frictional losses is reduced and pressure can recover to a greater extent. This principle has long been recognized in the field of fluid mechanics and led to the design of streamlined obstruction flow meters for insertion into pipelines (the Venturi meter has an outflow taper angle of 14° to maximize pressure recovery) (20). When this hydrodynamic principle is applied to the human circulatory system, it becomes clear that the magnitude of pressure recovery will to a large extent depend on the geometry of a stenosis.

The purpose of this study was to determine the magnitude of pressure recovery for a variety of stenosis geometries to define configurations in which differences between Doppler and catheter gradients are likely to become clinically relevant. Our findings demonstrate that Doppler gradients accurately reflected the highest obtainable catheter gradient, which is the pressure decrease in the vena contracta. When distal catheter pressures were measured downstream from the stenosis, the agreement between Doppler and catheter gradients was indeed highly dependent on the stenosis geometry, the predominant variable being the outflow angle of the stenosis. In stenoses with sudden or relatively fast expansion such as membranelike stenoses and hourglasslike stenoses with an outflow angle of 60°, acceptable agreement between Doppler and catheter gradients was found. However, with a further decrease of the outflow angle, Doppler gradients significantly exceeded catheter gradients. The discrepancies became substantial when this angle reached 20°. In this case, Doppler ultrasound "overestimated" the catheter gradients by approximately 50%. This Doppler-catheter gradient relation was found whether the outlet expanded gradually to the final vessel diameter or expanded abruptly after 2 cm of gradual expansion. These results demonstrate that pressure recovers within a short distance distal to the stenoses. Therefore, gradual expansion over short distance may be sufficient to cause significant differences between Doppler and catheter gradients due to pressure recovery.

In the present study, the Doppler-catheter gradient correlation was also affected by the inflow geometry. Greater discrepancy was observed for a gradually tapering inlet regardless of whether the inflow angle was 20° or 60°. Keeping the outflow angle constant, the differences between Doppler and catheter gradients were smaller for a stenosis with abrupt narrowing. This may be the result of significant flow contraction in the latter, resulting in a greater extent of flow separation (27). The effect of flow contraction on the Doppler-catheter gradient relation may deserve further study.

All stenoses in the present study were axisymmetric and stenoses that are not axisymmetric may differ in the precise magnitude of pressure recovery. However, considering the results obtained with models without axisymmetric stenoses in the past (15), the magnitude of pressure recovery is primarily determined by the outflow taper angle, whereas axisymmetry seems to be of minor importance.

Comparison with previous studies. Levine et al. (15) have reported significant pressure recovery in an in vitro study of hypertrophic obstructive cardiomyopathy. In their study, the magnitude of pressure recovery was greatest for a tunnel-like stenosis gradually tapering outward. However, they did not specify the exact geometry, studied steady flow conditions only and did not present any Doppler data for comparison. Yoganathan et al. (16) found clinically relevant pressure recovery in an in vitro model of subvalvular pulmonary stenosis. The sequence of a stenotic segment and a bioprosthetic valve allowed a rather streamlined reexpansion of flow, resulting in significant pressure recovery. The same investigators (16) have reported substantial overestimation of catheter gradients by Doppler ultrasound in a model of ventricular septal defect tunnels. They hypothesized that these differences were due to pressure recovery downstream from the tunnel because they did not measure the pressure immediately distal to the tunnel. However, because these tunnels opened abruptly to a large chamber, it seems unlikely that pressure recovery distal to the tunnel would occur to the suggested extent. Furthermore, in another study of similar tunnel-like stenoses (6), no significant pressure recovery and no overestimation by Doppler ultrasound were found. That study (6) even reported an underestimation of catheter gradients across tunnel obstructions by Doppler ultrasound as a result of viscous resistance. However, flow dynamics in tunnel obstructions are certainly more complex. Such geometries were not included in the present study and deserve further study.

Clinical implications. In clinical catheterization studies, pressure gradients across flow obstructions are usually measured with the distal catheter port in a position where pressure will have recovered to some extent. The agreement between this gradient and the Doppler gradient that reflects the highest pressure gradient in the vena contracta will depend on the actual magnitude of pressure recovery. The extent of this pressure recovery will vary with the geometry of the stenosis, which, therefore, significantly affects the Doppler-catheter gradient relation. The outflow geometry predominantly influences this relation, but the shape of the inlet may affect the results as well. Because Doppler gradients provide the highest local gradient rather than the net pressure decrease that reflects the hemodynamic effect of a flow obstruction, one should be aware that the Doppler technique may considerably overestimate the hemodynamic relevance of a stenosis with significant pressure recovery.

Valvular stenoses are usually discrete, with abrupt narrowing and abrupt expansion. Although pressure recovery has to be considered in such lesions, it is usually slight and of no great clinical relevance. Pressure recovery is highly dependent on the ratio of the stenotic flow area to the downstream flow area. This ratio will almost always be unfavorable for stenoses of atrioventricular valves as well as for the majority of aortic and pulmonary stenoses. Significant discrepancy between Doppler and catheter gradients may occur when this ratio is more favorable, as in patients with hypoplastic great vessels or with mild stenoses with relatively high initial gradients due to high flow rates, as seen with concomitant severe regurgitation.

Pressure recovery and discrepancies between Doppler and catheter gradients are most likely to become clinically relevant in stenoses with a gradually tapering outlet. The outflow angle has to be relatively shallow (≤20°) to cause substantial differences between Doppler and catheter estimates. Such geometry may be found in subvalvular or supravalvular stenoses, heart valve prostheses (between the two leaflets of bileaflet valves), hypertrophic cardiomyopathy and, in particular, vascular stenoses such as aortic coarctation.

We thank Gerald Maurer, MD for helpful discussions during this work.

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