Continuous-wave Doppler velocities and gradients across fixed tunnel obstructions: studies in vitro and in vivo

AJIT P. YOGANATHAN, PH.D., LILLIAM M. VALDES-CRUZ, M.D., JAN SCHMIDT-DOHNA, M.S., A. JIMOH, M.S., CHARLES BERRY, PH.D., T. TAMURA, M.S., AND DAVID J. SAHN, M.D.

ABSTRACT The simplified Bernoulli relationship appears to be quite accurate for predicting gradients across discrete valvular obstructions. Controversy exists about how accurately it predicts the severity of disease in longer segment obstructions. In this study we constructed a pulsatile model of subvalvular pulmonary stenosis in vitro to study nine custom-made subvalvular tunnels 2, 4, and 7 mm in length with flow cross sections of 0.5 to 1.5 cm² and with the stenotic segment proximal to a nonstenotic bioprosthetic valve, and a pulsatile model in vitro of a 16 mm long tunnel-like ventricular septal defect (VSD) of varying cross-sectional area (0.20 to 0.64 cm²). We also compared the observations in vitro with those in an open-chest dog preparation with a tunnel-like interventricular communication. In the subpulmonic stenosis model, for each individual tunnel, 10 instantaneous peak gradients between 15 to 105 mm Hg were available. The pressure gradients across the tunnel alone, measured in the subvalvular area, were consistently higher than the measured gradients across the tunnel plus valve, suggesting some relaminarization of flow (i.e., a decrease in velocity) and pressure recovery (i.e., an increase in pressure) distal to the obstruction. Continuous-wave Doppler velocities across the 4 and 7 mm tunnels for the highest gradients were slightly lower than for the 2 mm tunnel at the same gradients, and it was only for the 0.5 cm² cross section, 4 and 7 mm tunnels that there was a suggestion of minor viscous energy loss. For all the subvalvular tunnels studied, the Bernoulli relationship accurately predicted the results of the pressure drop across the tunnel only, while the gradient across tunnel plus valve was consistently lower. For the VSD tunnel model in vitro, the Doppler-derived gradients were approximately 40% higher than the measured gradients. The findings for the subvalvular and VSD tunnels in vitro and similar findings in the open-chest dogs with VSD suggest that relaminarization of flow and recovery of pressure occurred distal to the tunnel orifice, whereas continuous-wave Doppler findings correlate with the highest instantaneous gradients measured in the lowest pressure areas at the vena contracta of the tunnel.


ONE OF THE UNQUESTIONED contributions of the Doppler technique to clinical cardiology has been its use as a noninvasive method for estimating gradients across discrete stenotic orifices by use of a combination of the Doppler measurement of the maximal velocity and a simplification of the Bernoulli relationship.1, 2 Since Holen et al.1 first proposed and validated this method, numerous investigators in the United States and Europe have revalidated their results in many forms of valvular and vessel stenoses.3–6 The controversies that have arisen regarding the validity of this equation have been mostly related to the need for angle correction for correct velocity determination and more fundamentally, the applicability of the Bernoulli relationship itself in complex flow systems like dual or irregular orifices, or very small long tunnel-like obstructions.

Subvalvular pulmonary stenosis is a common form of congenital heart disease found either alone or, most commonly, in combination with ventricular septal defects (VSDs), pulmonary valve stenosis, tetralogy of Fallot, and subaortic stenosis.7 Subpulmonary obstructions can be either dynamic or fixed since they can result from hypertrophy of normal right ventricular bands, from fibrous membranes immediately below the pulmonary valve, accessory tricuspid valve tissue, aneurysms of the ventricular septum, or from congenital

From the Cardiovascular Fluid Dynamics Laboratory, School of Chemical Engineering, Georgia Institute of Technology, Atlanta, and the Division of Pediatric Cardiology, University of California, San Diego.

Supported by the American Heart Association, and by a grant from the Whitaker Foundation.

Address for correspondence: David J. Sahn, M.D., UCSD Medical Center, Division of Pediatric Cardiology, 225 West Dickinson St. H814A, San Diego, CA 92103.

Received April 1, 1986; revision accepted May 14, 1987.
stenosis of the os infundibuli. The length of the stenosis can vary from a discrete circumferential narrowing to longer, tunnel-like lesions of up to several centimeters in length. Fixed subaortic obstructions are also of clinical significance and they vary in length substantially and may have a progressive natural history.8

The purpose of this study was to design and investigate (1) a pulsatile model of subvalvular obstruction of the right ventricular outflow tract in vitro to study the effect on Doppler velocities of flow area and/or length of fixed subpulmonary tunnel-type obstructions, and (2) a pulsatile model in vitro of a tunnel-like interventricular communication (VSD). Our goal was to evaluate the accuracy of the simplified Bernoulli relationship for the prediction of gradients across these more complex flow systems. The findings would be referable to some types of fixed subvalvular left ventricular outflow tract obstructions, as well as to tunnel orifices in general, such as those between the two ventricles in the presence of muscular VSDs. Additionally, since the Bernoulli method is being applied in clinical cardiology in an attempt to use Doppler velocities to predict the gradients across VSDs,9 and to provide an opportunity to test our observations related to the pressure velocity relationships governing tunnel orifices in a setting in vivo, we also undertook the study of an open-chest animal preparation with a variable sized interventricular communication.

Materials and methods

Pulsatile flow apparatus for subvalvular pulmonary tunnels. The pulsatile flow apparatus used was designed to duplicate the flow and the pressure waveforms observed on the right side of the human heart. The apparatus consists of a physiologically shaped glass model of the pulmonary artery into which one of nine tunnel-like obstructions is inserted proximal to the pulmonary valve, three electromagnetic flow probes (Carolina Medical model EP680), pressure taps, two gate valves, two pulmonary pressure wave control (PPWC) sections on both the right and left pulmonary arteries, a PPWC air reservoir, an atrial reservoir, a straight valve section, a right ventricular bulb, a ventricular pressure wave control (VPWC) section, a VPWC air reservoir, an immersible centrifugal pump (Little Giant), and a plastic bucket. The latter two items are used for filling and emptying the pulse duplicator. A detailed description of the design and function of the pulse duplicator and pulmonary artery model has been published elsewhere.10

The right heart pulse duplicator was operated under the following physiologic conditions: (1) heart rate of 70 beats/min, (2) systolic time interval of 320 to 350 msec, (3) mean pulmonary artery pressure of 15 to 30 mm Hg, and (4) cardiac outputs from 1.0 to 7.5 liters/min. An aqueous glycerine solution containing 2% (by volume) cornstarch particles (10 μm in diameter) was used as the blood analog fluid and adjusted to a physiologic viscosity of 3.5 cp at 22°C.

A nonstenotic bioprosthetic pericardial valve with a flow area of 3.0 cm² was used to represent a mildly deformed or minimally stenotic natural "pulmonary" valve. The tunnel-like obstructions were inserted proximal to the valve such that the center of the length of the tunnel was 10 mm from the valve anulus, as shown in figure 1. Nine tunnels were custom made to measure 2, 4, and 7 mm in length, with flow cross-sectional areas of 0.5, 1.0, and 1.5 cm². The 4 and 7 mm long tunnels had smooth hourglass shapes, while the 2 mm obstructions represented discrete short stenoses. The combination of a physiologically realistic pulsatile flow system, a minimally stenotic bioprosthetic valve, and subvalvular tunnels that were similar in length and flow area to those potentially found in the human heart resulted in a model closely approximating the physiology encountered in forms of congenital and acquired heart disease.

Pressures immediately distal to the tunnel, between the tunnel and the valve, and immediately distal to the pulmonary valve were measured with fluid-filled side-hole catheters (less than or equal to 5 cm in length). The pressure proximal to the tunnel was monitored via a wall pressure tap in the model itself. The pressure gradients across the tunnel and the pressure gradients

![FIGURE 1](https://circ.ahajournals.org/doi/abs/10.1161/CIRCULATIONAHA.117.031972)

The dimensions of the model, especially for the area proximal to the valve, are shown. The positions of the pressure measurements (from position 1 proximal to the tunnel to position 2 between the tunnel and valve, or to position 3 downstream from the valve) are also shown. The center of each tunnel was 10 mm proximal to the valve ring.
across the tunnel plus valve were measured with a Validyne differential pressure transducer system. The differential pressure transducer was connected to the upstream wall pressure tap and the distal fluid-filled catheters. The cardiac output in the model was monitored by a Carolina Medical model EP680 cannulating-type electromagnetic flow probe and a model 501 flowmeter. The analog signals from the Validyne differential pressure amplifier and the electromagnetic flowmeter were fed to an analog-to-digital converter that was interfaced to an Apple II Plus microprocessor for on-line data collection and analysis. Therefore, it was possible to obtain pressure and flow information in real time, on a beat-to-beat basis. The pressure and flow waveforms were also monitored on a Tektronix (T912) dual-beam storage oscilloscope. For each tunnel the cardiac output was varied to produce instantaneous peak gradients from the right ventricle to the pulmonary artery of 15 to 105 mm Hg.

**Pulsatile flow apparatus for VSD model.** An alteration of the above-described flow apparatus was used to duplicate the flow and pressure waveforms observed in the left and right ventricles. The apparatus incorporated the left and right heart Georgia Tech pulse duplicator systems, which have been described previously. The chambers immediately distal to the left and right ventricular bulbs were interconnected by a 16 mm long, 9 mm inner diameter cylindrical nylon prosthesis.

The lumen of the prosthesis contained a very thin latex membrane that could be inflated with air or water via a small tygon tube, thus constricting the lumen of the “interventricular communication.”

The left and right heart pulse duplicators were operated under the following conditions: (1) synchronized heart rate of 70 beats/min, (2) systolic duration of 300 to 320 msec, (3) left heart cardiac outputs of 2.0 to 5.5 liters/min, and (4) gradual narrowing of the VSD tunnel by filling of the latex membrane. Pressures were measured with short (less than or equal to 5 cm in length) fluid-filled side-hole catheters in the left and right ventricles, about 25 mm proximal to the inlet and 25 mm distal to the outlet of the VSD tunnel. The catheters were connected to the Validyne differential pressure transducer system. Cardiac output was measured by the electromagnetic flowmeter. As previously described, the analog signals from the differential pressure transducer and the electromagnetic flowmeter systems were analyzed in real time by the Apple II Plus microprocessor.

The cross-sectional area of the VSD tunnel and the left heart cardiac output were varied to produce instantaneous peak systolic gradients from the left ventricle to the right ventricle of 15 to 105 mm Hg.

**Ultrasound Doppler method.** For each instantaneous peak gradient obtained, a simultaneous continuous-wave Doppler velocity recording was made with a dual-frequency (3.5 MHz imaging, 2.0 MHz Doppler) phased-array scanner (IREX Meridian) that has a nonsteerable line of sight within the sector image along which continuous-wave Doppler information can be obtained. The ultrasound transducer was coupled with gel to the model, and for the subpulmonary stenosis model it was placed at the bifurcation of the glass pulmonary artery model aiming directly toward the jet visualized through the glass by virtue of the scattering cornstarch suspension as it emerged from the pulmonary valve. For the study of the VSD tunnel, the transducer was placed on the right ventricular chamber such that it was aimed parallel to the jet emerging from the tunnel. In the continuous-wave Doppler sampling mode without simultaneous imaging, the Doppler system we used could resolve velocities up to 10 m/sec. Doppler outputs were available as an audio signal, a spectral display derived from a Chirp Z algorithm, and an analog display of the maximal and mean velocities. For the purpose of this study, the maximal velocities were measured only from the outer envelope of the spectral display. The Doppler sampling line position was altered by fine adjustments of transducer angulation as in clinical studies, with the use of both the audio and the spectral outputs, until spectral curves with clearly defined and smooth peak velocity envelopes were obtained (figure 2). The ultrasound Doppler velocity information was recorded on a strip chart as well as in real time on videotape. The maximal instantaneous ultrasound velocities were measured from the maximal shift obtained on the spectral display from at least six consecutive cycles.

**Animal preparations.** To evaluate an animal preparation in which a longer segment, tunnel-like orifice could be studied in vivo, we used a modification of our previously described open-chest animal preparation of a variable-sized VSD. Five 20 to 30 kg mongrel dogs were anesthetized with sodium pentobarbital (30 mg/kg), intubated, and ventilated with a standard volume respirator. A midline sternotomy was performed and the pericardium was opened. The ascending aorta and main pulmonary artery were both cleaned of adventitia and fat, and Gould-Statham SP2204 electromagnetic flow probes were placed on each vessel. A cylindrical cutter was then introduced through a right atrial purse-string suture and under venous inflow occlusion it was advanced across the tricuspid valve and forced through the ventricular septum to create a muscular or perimembranous defect.

Next, a 12 mm outer diameter, 9 mm inner diameter, 16 mm long, flanged, cylindrical nylon prosthesis was advanced over the tissue cutter and placed into the interventricular defect. The lumen of the prosthesis contained a latex membrane that could be inflated via a small polyvinyl tube exteriorized through the atrial purse string. When inflated, the latex membrane could be expanded to vary the size of the VSD. Pressure recordings were obtained with short No. 8F, end-hole Dacron catheters 5 mm in length attached to Statham P23Db pressure transducers. The catheters were sewn in place through the myocardium to reduce catheter motion. The left ventricular catheter was advanced through the apex and placed near the left side of the VSD prosthesis under echocardiographic imaging guidance. The right ventricular catheter was placed through a puncture incision in the right ventricular outflow tract with the tip 2 to 3 cm downstream from the orifice of the VSD. The shunt through the VSD and the pressure gradient were varied by changing the amount of saline in the latex membrane of the prosthesis. Shunts were produced with pulmonary-to-systemic flow ratios (QP/QS) of 1:4 to 1:6. The peak instantaneous pressure gradients across the defects in 40 steady-state recordings for the five dogs ranged from 4 to 80 mm Hg.

**Ultrasound Doppler method for animal studies.** For the study in vivo the transducer of an Irex System IIIB phased-array scanner (3.5 MHz imaging, 2 MHz continuous-wave Doppler) was placed directly on the right ventricular surface to image the right septal orifice of the VSD prosthesis. Continuous-wave Doppler sampling was undertaken with the transducer placed firmly on the heart. The two-dimensional imaging was then frozen to allow continuous-wave Doppler interrogation of the maximal velocity of flow, along with a reference electrocardiogram. Fine transducer angulations were performed with use of the audio Doppler output and the spectral waveforms to record maximal flow velocity, which was measured by one observer as an average of 4 to 6 beats during a sequence in which high-quality Doppler spectra were recorded, along with stable right and left ventricular pressures, with the animal’s respiration suspended. The peak instantaneous gradient between the right and left ventricles in systole was calculated from the pressure waves. Bernoulli calculation of the Doppler-predicted interventricular gradient was performed with the use of maximum velocities ranging from 180 to 610 cm/sec.

**Statistical analysis.** For both studies in vitro all readings of...
the ultrasound peak instantaneous velocities were made at separate settings by two independent investigators who at the time were unaware of each others results or of the catheter gradients. The ultrasound peak instantaneous velocities were converted to instantaneous peak gradient estimates with the simplified Bernoulli equation ($\Delta P_{\text{peak}} = 4 V^2_{\text{peak}}$) and compared with the actual instantaneous peak gradients measured across (1) tunnel, (2) tunnel plus valve, and (3) the VSD tunnel, by least squares regression analysis. Intraobserver and interobserver variabilities were analyzed by the paired two-tailed Student’s t test. For the animal study, gradients across the interventricular orifice were compared by linear regression to continuous-wave Doppler waveforms measured by a single observer blinded to the pressure gradient results.

**Results**

**General.** For each of the two tunnel models in vitro, 10 individual right ventricular–to–pulmonary arterial gradients or left ventricular–to–right ventricular gradients (for the VSD tunnel) between 15 and 105 mm Hg (in increments of 10 mm Hg) were studied, corresponding to Doppler-measured velocities of 1.9 to 6.1 m/sec. Ten data samples were obtained at the lowest and highest gradient settings (i.e., 15 and 105 mm Hg) for each of the 10 tunnels and for the VSD prosthesis, while at the remaining eight intermediate gradients only two data samples were obtained. Figure 2 shows examples of the continuous-wave Doppler spectra obtained through 2 and 4 mm subvalvular tunnels, with settings resulting in low and high gradients, respectively. For both models in vitro the peak instantaneous velocities read from the ultrasound Doppler tracings by two independent observers were statistically identical (p>.05). A similar p value was obtained for the intraobserver variability. Peak instantaneous velocities measured by the same observer on two separate occasions had a mean difference of 2.8% from the first observation ($r = .99$, SEE = 0.10 m/sec). Similarly, measurements made by two independent observers had a mean difference of 4.0% ($r = .99$, SEE = 0.18 m/sec).

**Subpulmonary stenosis model**

**Effects of tunnel length.** Figure 3 serves to summarize the effect of tunnel length on the gradient calculated with Doppler velocities (i.e., gradient estimated from
the Bernoulli equation) for selected 0.5, 1.0, and 1.5 cm² tunnels. For all three cross-sectional areas studied, results for the 4 and 7 mm length tunnels of matched cross section were statistically identical (p > 0.05 by two-tailed t test). The velocities and Doppler-calculated gradients across the 2 mm long models were slightly but not significantly higher than for the 7 mm tunnels (p > 0.10) for any given true gradient.

**Effects of tunnel area.** The effect of tunnel area on gradients calculated with Doppler velocities is also summarized in figure 3. At any given gradient and tunnel length, no differences (p > 0.10) could be demonstrated for the effect of tunnel flow area on the measured Doppler velocities.

**Transtunnel vs tunnel plus valve gradients.** For all nine tunnels studied and for any given flow state, the measured pressure difference across the tunnel only (i.e., transtunnel gradient) was higher than the gradient across the tunnel plus valve (i.e., right ventricle-to-pulmonary arterial gradients), by 20% to 33% (see examples given in table 1).

Figures 3 and 4 show plots of catheter-measured pressure gradients vs the corresponding Doppler-derived gradients. For all nine tunnels, the Doppler-derived gradients correlated very well with the transtunnel catheter gradients (r = .97 to .99, SEE = 3.9 to 9.3 mm Hg), with slopes close to unity. Doppler-derived gradients were, however, invariably significantly higher than the catheter-measured gradients across the tunnel plus valve, as shown by the slopes of 1.29 to 1.46 in figure 4.

**VSD tunnel gradients.** Figure 5, top, shows a plot of catheter-measured pressure gradients vs the corresponding Doppler-derived gradients for the VSD tunnel. A linear relationship was obtained between the Doppler-predicted gradient and the measured gradient (r = .99, SEE = 3.5 mm Hg). It can be clearly seen that the Doppler-derived estimates of interventricular gradients were consistently higher than the catheter-measured gradients (Doppler gradient = 1.42 × measured gradient = 2.2 mm Hg).

**Effects of Reynolds number.** The Reynolds number was calculated with the equation:

\[
Re = \frac{\text{fluid density} \times \text{velocity} \times \text{vessel diameter}}{\text{Fluid viscosity}}
\]

(fluid density = 1.05 g/cm³; viscosity = 0.035 dynes/cm²; vessel diameter = 2.54 cm.) The peak
systolic Reynolds numbers in the region distal to the tunnels and proximal to the valve (calculated based on the maximum measured continuous-wave velocities) ranged from 17,800 to 57,200 for all nine subpulmonary tunnels studied. The peak systolic Reynolds numbers immediately distal to the VSD tunnel in the right ventricle ranged from 33,200 to 93,600 (calculated based on a right ventricular diameter of 5.08 cm and the maximum measured continuous-wave velocity). Based on these high Reynolds numbers, the flow fields would definitely be turbulent, and the viscous losses would be expected to be quite low.\textsuperscript{13, 14} For example, for flow in a circular tube a Reynolds number in excess of 2200 is associated with turbulent flow conditions.

\textbf{Animal study.} Forty Doppler-predicted gradient values were available for comparison with measured gradient. The linear relationship between Doppler-predicted and measured gradient showed a correlation coefficient of $r = .86$ (SEE = 18.1 mm Hg). As shown in figure 5, bottom, the slope of the relationship was significantly greater than 1 for the regression equation with measured gradient on the abscissa and Doppler gradient on the ordinate (i.e., Doppler gradient = 1.58 × measured gradient + 4.5 mm Hg), suggesting

\begin{table}[h]
\centering
\caption{Examples of peak instantaneous pressure drops measured across the various tunnels}
\begin{tabular}{cccc}
\hline
Tunnel area (cm$^2$) & Tunnel length (mm) & Cardiac output (liters/min) & Peak instantaneous pressure drop across tunnel plus valve (mm Hg) & Peak instantaneous pressure drop across tunnel (mm Hg) \\
\hline
1.5 & 2 & 3.3 & 25 & 33.5 \\
1.5 & 2 & 6.9 & 105 & 129.2 \\
1.5 & 4 & 3.0 & 15 & 20.1 \\
1.5 & 4 & 7.9 & 55 & 73.2 \\
1.5 & 7 & 3.7 & 15 & 20.0 \\
1.5 & 7 & 6.4 & 45 & 60.0 \\
1.0 & 2 & 2.3 & 35 & 46.2 \\
1.0 & 4 & 2.0 & 15 & 18.5 \\
1.0 & 4 & 5.8 & 75 & 99.0 \\
1.0 & 7 & 2.8 & 15 & 18.0 \\
1.0 & 7 & 6.6 & 85 & 102.9 \\
0.5 & 2 & 1.2 & 15 & 18.2 \\
0.5 & 2 & 2.9 & 95 & 129.2 \\
0.5 & 4 & 1.3 & 25 & 30.0 \\
0.5 & 4 & 3.1 & 105 & 139.7 \\
0.5 & 7 & 1.1 & 15 & 19.4 \\
0.5 & 7 & 3.0 & 95 & 125.4 \\
\hline
\end{tabular}
\end{table}

\begin{figure}
\centering
\includegraphics[width=\textwidth]{figure4}
\caption{Regression relationships for measured right ventricular-pulmonary arterial pressure gradients (abscissa) compared with Doppler-predicted gradients (ordinate) are shown for the same 2 and 7 mm tunnels as in figure 3.}
\end{figure}
that the Doppler method significantly overestimated the measured interventricular gradients.

Discussion

The pulsatile flow apparatus used in this investigation has various control features that permitted fairly accurate control of the fluid mechanics of the left and right sides of the human heart. It was therefore possible to obtain highly repeatable settings so that the nine subvalvular pulmonary tunnels and the VSD tunnel could be studied at reproducible flow (i.e., pressure gradient) conditions. This differs from some other flow simulation studies in vitro used to investigate tunnel-like obstructions with Doppler ultrasound that did not adequately reflect the pressure and flow waveforms of the heart.15

The results of our study show that at equivalent pressure gradients, changing tunnel flow area from 0.5 to 1.5 cm² does not appear to affect the measured continuous-wave Doppler velocities. In addition, for equivalent pressure gradients, the continuous-wave Doppler-measured velocities were slightly lower with the 4 and 7 mm long tunnels than with the 2 mm obstructions. However, this effect of length on Doppler velocity determinations is minimal and of only borderline statistical significance.

The peak Reynolds numbers distal to the nine subvalvular tunnels and the VSD tunnel were always in

![Regression relationships for measured left ventricular-right ventricular pressure gradient (abscissa) compared with Doppler-predicted gradients (ordinate) are shown for the VSD tunnel model. Top, in vitro study. Bottom, in vivo animal study (the dotted lines show the 95% confidence limits for the regression relationship). Note that in the top panel, the lowest and highest values each represent an average of 10 data samples (see text for details).](image-url)
excess of 15,000. Therefore, the effects of viscous friction on the simplified Bernoulli equation would be negligible, compared with the “convective” or kinetic energy term, for the tunnel geometries and flow conditions used in this investigation.

The results also indicate that the highest pressure drops appear to occur across the tunnel itself, with pressure recovery and a drop off in measured pressure gradient when it is measured distal to the nonstenotic bioprosthetic valve for the subvalvular pulmonary tunnels (table 1). Observations of the Venturi flow phenomena indicate that the maximal pressure drop (i.e., highest distal velocity and lowest distal pressure) occurs at the venacontracta. The venacontracta is defined as the narrowest cross section of the jet immediately distal to an obstructing orifice. As the flow travels further downstream the velocity decreases and the pressure increases, leading to lower pressure drops. This indicates that for the subvalvular tunnel geometries studied, the venacontracta occurs immediately distal to the tunnels (i.e., between tunnel and valve), as schematically illustrated in figure 6. Studies conducted with the nonstenotic bioprosthetic valve only yielded peak Doppler velocities of 1.2 to 3.0 m/sec over a cardiac output range of 1.5 to 6.5 liters/min. It is therefore reasonable to expect that the velocities would decrease as the blood analog fluid passes through the valve, leading to higher measured pressures distal to the valve (i.e., lower pressure drops). In the case of the VSD tunnel geometry studied in vitro, the venacontracta once again occurred immediately distal to the tunnel, leading to lower pressure gradients further downstream within the ventricle (i.e., the location at which the right ventricular pressure was measured).

Because of the Venturi phenomena described above, the Doppler-derived pressure gradients were larger compared with the measured instantaneous peak gradients across the tunnel plus valve or the measured instantaneous peak gradients between the left and right ventricles. The continuous-wave Doppler samples the maximal velocity in the flow system as long as the beam traverses the jet. Therefore, the simplified Bernoulli relationship will reflect the maximal gradient in the system (i.e., the gradient across the tunnel only, to the position of the vena contracta). This gradient would be larger than the measured gradient across the tunnel plus valve or the measured gradient across the left and right ventricles, which is lower due to a decrease in velocity (i.e., increase in pressure) of the flow jet as it travels downstream of the obstruction.

Tierstein et al. also reported that until studying their smallest tunnels, akin to coronary atherosclerotic

![Diagram](http://circ.ahajournals.org/doi/abs/10.1161/CIRCULATIONAHA.117.033139)

**FIGURE 6.** Diagram showing that the maximal velocity is reached between the tunnel and the valve, and that the flow field recovers pressure with a loss of velocity and with relaminarization distal to the valve.
obstructions, the Bernoulli relationship allowed accurate gradient prediction without the necessity for consideration of viscous factors (i.e., consideration of frictional loss equations for gradient prediction). If viscous loss were to occur and were not taken into consideration, the result would be an underestimation of pressure gradient by the simplified Bernoulli relationship.

The obstructions we studied in terms of their lengths and their severity were similar to the types encountered both in the subpulmonary region and in those patients with discrete or slightly longer segment fibromuscular subaortic stenosis. Our experience in animal preparations, when we studied a population of Newfoundland retrievers, also suggested that in the more severe range of subaortic obstruction, the simplified Bernoulli relationship worked accurately if we measured the gradient only across the subaortic obstruction on slow pullback distal to the membrane and proximal to the valve to calculate a peak “instantaneous” gradient. Especially for the animals with more severe obstruction, the Bernoulli relationship overestimated left ventricular body to aortic root gradient. It must be remembered that the prediction of pressure gradient obtained with the Bernoulli method is always an instantaneous gradient and specifically represents the gradient across the area where the highest velocity jet is formed.

The preparation of VSD in vivo in animal studies likewise demonstrated that in a restrictive tunnel orifice, in this case a 16 mm long tunnel prosthesis, the Bernoulli relationship and ultrasound Doppler—measured velocities could predict the gradient between left and right ventricles, although with a tendency for overestimation. In the animal preparation there was also significant variability in the prediction relationship, which we believed was secondary to the variable shape of the tunnel in the flanged nylon tube of the VSD prosthesis as the latex membrane was inflated with saline, since many more manipulations of the tunnel itself were used in the study in vivo than in the study in vitro. The overestimation of gradient by the Doppler method with the slope of the relationship significantly greater than 1 occurred, we believe, for reasons similar to those in the study in vitro (i.e., in this tunnel-like VSD, the downstream pressure measured within the body and outflow tract of the right ventricular cavity was obtained at a significant distance from the orifice of the tunnel). Most of the clinical studies of prediction of VSD gradient have involved short membranous VSDs in a thinner portion of the septum with right ventricular orifices closer to the usual site of right ventricular pressure measurement.

Clinical experience with very long segment tunnel orifices such as tunnel-like muscular VSDs or long segment tortuous coarctations has varied. While the Doppler technique has in fact overestimated coarctation gradients in some studies, other experience with coarctations, especially those of long segments where one might expect viscous factors to be manifest and cause an underestimation of gradients by the Bernoulli method, have in fact reported underestimation. However, these results were not attributed to viscous factors, but to poor jet formation and tortuosity altering alignment of the emerging jet from the coarctation, as judged angiographically. For straight jets, even with fairly long segment stenosis, the Bernoulli relationship has worked fairly well and has not required adjustments for frictional loss. Requarth et al. studied a system in vitro that featured unphysiologically long obstructions of 4 and even 8 cm in length and 82% and 91% reductions of area and concluded that there was a need to consider frictional loss, but their model represented an exaggeration of the obstructions encountered clinically.

In summary, in the clinical setting with good alignment of the jet, high velocities can be recorded in patients with subvalvular obstruction. We have investigated whether, when using clean velocity recordings in subvalvular obstructions, the velocities can be used along with the simplified Bernoulli relationships to predict a gradient. Our results have shown that for subvalvular tunnel obstructions of clinically relevant length and severity, providing one is measuring the appropriate gradient, that is the peak “instantaneous” gradient just across the tunnel from its proximal end to a pressure recorded in close proximity to its distal end, the simplified Bernoulli equation provides accurate prediction of pressure gradient results without the need for consideration of viscous factors.

References

6. Teirstein P, Yeager M, Yock PG, Popp LR: Noninvasive deter-
Continuous-wave Doppler velocities and gradients across fixed tunnel obstructions: studies in vitro and in vivo.
A P Yoganathan, L M Valdes-Cruz, J Schmidt-Dohna, A Jimoh, C Berry, T Tamura and D J Sahn

Circulation. 1987;76:657-666
doi: 10.1161/01.CIR.76.3.657

Circulation is published by the American Heart Association, 7272 Greenville Avenue, Dallas, TX 75231
Copyright © 1987 American Heart Association, Inc. All rights reserved.
Print ISSN: 0009-7322. Online ISSN: 1524-4539

The online version of this article, along with updated information and services, is located on the World Wide Web at:
http://circ.ahajournals.org/content/76/3/657

Permissions: Requests for permissions to reproduce figures, tables, or portions of articles originally published in Circulation can be obtained via RightsLink, a service of the Copyright Clearance Center, not the Editorial Office. Once the online version of the published article for which permission is being requested is located, click Request Permissions in the middle column of the Web page under Services. Further information about this process is available in the Permissions and Rights Question and Answer document.

Reprints: Information about reprints can be found online at:
http://www.lww.com/reprints

Subscriptions: Information about subscribing to Circulation is online at:
http://circ.ahajournals.org//subscriptions/