The accuracy of Doppler ultrasound measurement of pressure gradients across irregular, dual, and tunnellike obstructions to blood flow

PAUL S. TEIRSTEIN, M.D., PAUL G. YOCK, M.D., AND RICHARD L. P O P P, M.D.

ABSTRACT The accuracy of Doppler-estimated pressure gradients in the setting of irregular, multiple, and tunnellike stenoses was investigated. An in vitro model of the left ventricular outflow tract was designed to allow pulsatile flow of red cells in saline across valve orifices from 0.01 to 2.5 cm². Simultaneous pressure gradients were estimated by both Doppler and direct-pressure manometer techniques. Gradients obtained by the two methods correlated well for valve areas in the range of clinical stenoses at pressure gradients of 10 to 150 mm Hg (r = .97 to .99). Model valves were constructed with a large orifice (0.75 to 1.25 cm²) placed beside a small orifice (0.02 to 0.25 cm²) in the same outflow tract. A distinct jet was recorded when the Doppler transducer was aligned with each orifice. Doppler-estimated gradients for each pair of large and small orifices were identical and correlated well with those measured by manometer (r = .97 to .99). Irregularly shaped orifices also provided good correlation between the two methods (r = .98 to .99). Pulsatile flow was generated through long tunnellike obstructions with cross-sectional areas varying from 0.06 to 1.25 cm². Tunnel length varied from 0.1 to 4 cm. Tunnel areas above 0.25 cm² gave good Doppler-to-manometer correspondence at all tunnel lengths. Doppler underestimated manometer-determined values in the 0.25 cm² tunnel by 8% at 3 cm and by 15% at 4 cm. In the 0.06 cm² tunnel, Doppler underestimated manometer gradients by 12%, 15%, 32%, and 42% at lengths of 1, 2, 3, and 4 cm, respectively. We conclude that accurate Doppler estimation of transvalvular pressure gradients is possible in the setting of irregular or multiple valve orifices. Doppler-estimated pressure gradients across tunnellike obstructions are accurate under most geometric conditions seen in noncoronary stenoses. However, decreasing tunnel cross-sectional area and increasing tunnel length cause underestimation of manometer-derived values by Doppler. This underestimation becomes significant at tunnel areas below 0.25 cm² and tunnel lengths above 3 cm.


THE PRESSURE GRADIENT across an obstruction to blood flow can be estimated noninvasively by measuring the maximum velocity (V) beyond the obstruction with Doppler echocardiography and a simplified Bernoulli equation (pressure gradient = 4V²). The accuracy of this simplified Bernoulli equation has been validated both in vivo and in vitro with models of valvular stenosis. These models used discrete, round orifices of varying diameters to simulate stenotic lesions. Clinical obstructions to blood flow, however, are often irregular and non discrete. For example, calcific or rheumatic aortic valve deformities are usually asymmetric. Severe stenoses often contain multiple small orifices that produce multiple jets of blood. Prosthetic valvular stenoses can also produce irregular or multiple orifices if a thrombus forms between the valve struts and the ball or disc. In other lesions, non discrete or tunnellike obstructions to blood flow are created. These include subpulmonic or infundibular pulmonic stenosis, some ventricular septal defects, and coarctation of the aorta. Coronary arterial stenoses represent another kind of tunnellike obstruction to blood flow. The adequacy of coronary angioplasty can be assessed by documenting the reduction of a pressure gradient across a coronary stenosis. Catheter-tipped Doppler ultrasonography has been suggested as an alternative method for measuring these changes in pressure gradient. In light of these geometric variations in clinically observed stenoses, we asked the following question: Does the simplified Bernoulli equation as

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Supported in part by grant 5 T 32 HL 07526 from the National Institutes of Health.
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Received Feb. 5, 1985; revision accepted June 20, 1985.
applied to blood velocities measured by Doppler ultrasound accurately predict pressure gradients across irregular, multiple, and tunnellike orifices?

Methods

A model was designed to create pulsatile blood flow through stenoses of varying diameters and lengths in vitro. A pulse of blood was generated with a 60 ml plastic syringe (diameter 2.8 cm). The syringe tip was removed and the barrel connected directly to a lucite inlet chamber (figure 1). The inlet chamber was bolted to an outflow chamber of the same dimensions. Interchangeable lucite plates with a variety of orifice shapes and sizes were inserted between the inlet and outflow chambers as illustrated in figure 1. Multiple lucite plates were stacked in series to create tunnel obstructions. Blood leaving the outflow chamber entered a 500 ml reservoir. Circulation from the reservoir back to the inlet chamber was provided by a centrifugal pump (Biomedical Inc., Minnetonka, MN).

Continuous-wave Doppler flow signals were recorded with a 2 MHz nonimaging transducer and an Irex Exemplar (Irex, Ramsey, NJ) ultrasonograph. This transducer has a beam diameter of 0.5 cm at its focal point of 4 cm. The beam diverges as it leaves its focal point. At 7 cm the beam diameter is 0.8 cm; at 11 cm it is 1.1 cm. The Doppler transducer was inserted through a rubber gasket into the inlet chamber and directed downstream toward the orifice. When directed at a single orifice the distance between the transducer and orifice was 6.5 cm. This distance was either 6.0 or 7.0 cm when the sound beam was moved across a dual-orifice plate. (Because of the angle inherent in the transducer head, when it is moved 3 cm horizontally it must also be moved 1 cm forward to maintain parallel alignment with the jet.)

While the centrifugal pump circulated blood at a constant velocity, the Doppler transducer was precisely aimed along the jet of blood flowing through the orifice. We found that continuous flow allowed accurate transducer alignment parallel to the jet. Position of the transducer was adjusted by both the audio and spectral signal of the frequency shift to obtain the highest blood velocities, indicating an angle of near zero degrees between the ultrasound beam and the flow stream caused by the stenosis. When the highest velocity was obtained, the transducer was clamped in position, the pump was stopped, and pulsatile flow was generated by manually depressing the syringe plunger.

Simultaneous direct pressure measurements were obtained from pressure ports provided both in the inlet and outflow chambers. The pressure ports were positioned 5 cm on either side of the obstruction. Pressure transducers (Microdot Inc., Industrial City, CA) were connected to these ports and recordings were made on a physiologic recorder (Hewlett-Packard, Palo Alto, CA). In addition to continuous recordings of inlet and outflow chamber pressures, a separate signal subtraction channel directly displayed the pressure difference between the two chambers. Pressure transducers were calibrated with a mercury manometer. The accuracy of pressure transducer measurement was initially verified by simultaneous measurement with manometer-tipped catheters (Millar Instruments, Houston), placed directly into the inlet and outflow chambers. Pressure waveforms generated during pulsatile flow were virtually identical in both measurement systems. Moving the manometer-tipped catheter to a variety of positions within the outflow chamber did not significantly alter the pressure measurements.

ABO- and Rh-compatible, outdated, banked human red blood cells were used as Doppler targets. Packed cells at room temperature were diluted with 0.9% saline. Blood was anticoagulated with 3000 U of heparin/liter. Hematocrits were measured with a Coulter Coulter Counter (Coulter Electronics, Hialeah, FL) and maintained between 33% and 36%.

Orifice preparation. Single-orifice obstructions were prepared by drilling holes in the center of thin lucite plates (thickness 1 mm). Orifice area varied from 0.01 to 2.5 cm². Sets of dual parallel orifices were prepared by drilling one large and one small hole in a single lucite plate. The holes were positioned equidistant from the center of the plate and separated by 3.0 cm (figure 1). Other irregular orifices, with a variety of shapes, were cut in lucite discs with a jeweler’s saw. Tunnel obstructions were prepared by drilling holes through a stack of lucite plates (thickness 3 mm each). Tunnel cross-sectional area was 0.06, 0.25, 0.75, or 1.25 cm². The plates were aligned precisely between the inlet and outflow chambers with three guiding bolts to ensure a smooth tunnel surface (figure 1). Serial addition of plates provided tunnel lengths from 1 mm to 4 cm.

Measurements. Peak blood velocity (V) in meters per second was obtained from the hard copy of the Doppler waveform.

![FIGURE 1. Diagram of experimental apparatus. Orifice plates are interchangeable. (A plate containing parallel dual orifices is shown at the top). Pulsatile flow is generated with the syringe mounted above the inlet (right). The Doppler transducer measures blood velocity while the pressure gradient is simultaneously recorded as the difference of the two manometers.](http://circ.ahajournals.org/content/95/3/577/F1.large.jpg)
LABORATORY INVESTIGATION—DOPPLER ECHOCARDIOGRAPHY

The highest coherent portion of the spectral display was designated as the maximum velocity (figure 2). Doppler-estimated pressure gradients (mm Hg) were calculated from the simplified Bernoulli equation (pressure gradient = 4V^2).1 Direct manometer-derived maximal pressure gradients (mm Hg) were obtained from the subtraction channel of the physiologic recorder (paper speed 50 mm/sec). For each orifice approximately 10 pulses of blood flow were analyzed. The pulses were generated by hand injections and the volume and speed of each injection were varied to provide a range of pressure gradients between 10 and 150 mm Hg.

Statistics. Statistical analysis was performed by a linear regression analysis. Correlation coefficient, standard deviation, y intercept, and slope of the regression line were obtained. A t test was performed to test the degree of doubt that the slope was not equal to 1. In the case of parallel dual orifices, the t test established the degree of doubt that the two slopes were not equal.

Results

Single round orifices (table 1). An example of simultaneous manometer and Doppler recordings is provided in figure 2. Of note, despite the use of continuous-wave Doppler, the tracings have a narrow spectral density. This indicates that most of the Doppler targets are moving at the same speed. Each pulse of blood is

FIGURE 2. Example tracings of simultaneous manometer (PGm) and Doppler (PGd) pressure gradients with extreme variation in contour to demonstrate the nearly identical waveform pattern obtained with both techniques. Orifice area is 0.06 cm². Paper speed is 50 mm/sec. V = velocity.
TABLE 1

<table>
<thead>
<tr>
<th>Orifice area (cm²)</th>
<th>y intercept</th>
<th>SD mm Hg</th>
<th>r</th>
<th>Slope</th>
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<td>.97</td>
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*p value (indicating degree of doubt that the slope is not equal to 1) is <.5 for each orifice tested.

generated by manual depression of a 60 cc plastic syringe. The jagged contour of some of the tracings is intentional. These nonphysiologic waveforms demonstrate the excellent correlation between Doppler and manometer measurements. In the first tracing, the Doppler signal shows a peak velocity of 3.8 m/sec, yielding a Doppler-estimated maximum pressure gradient of 58 mm Hg. The maximum manometer-derived gradient for the same pulse was 57 mm Hg. A representative plot of Doppler vs manometer gradients for a valve area of 1 cm² is shown in figure 3. The two methods correlated well at low pressure gradients (<50 mm Hg) with orifices of 0.01 to 2.5 cm² (r = .98 to .99, slope = .95 to 1.06). Correlation was good at high gradients (50 to 152 mm Hg) with orifices of 0.06 cm² or greater but 1.5 cm² or less. High gradients could not be generated across the 2.0 or 2.5 cm² orifices with our injection system. High gradients across smaller orifices (0.01 and 0.02 cm²) were difficult to interpret because of dispersion of the Doppler signal (see Discussion).

Parallels dual round orifices (table 2). Parallel dual round orifices were separated from one another by 3 cm. The Doppler transducer produces a beam diameter of 8 mm at 6 to 7 cm (−60 dB). This provides ample resolution of the two distinct jets. Directing the transducer across the face of a plate with two orifices allowed recording of two distinct jets with an intervening area of low velocity in all cases. Each set of dual orifices had a large orifice (either 0.75 or 1.25 cm²) and a small orifice (0.02, 0.06, 0.13, or 0.25 cm²). Eight sets of dual orifices were tested. Doppler and manometer gradients correlated well with each orifice (r = .97 to .99, slope = .94 to 1.07).

Irregular orifices (figure 4). The irregularly shaped orifices (see table 3) all demonstrated good Doppler-to-manometer correlation (r = .98 to .99, slope = .99 to 1.06).

Tunel obstructs (table 3). Tunnels with cross-sectional areas of 1.25 and 0.5 cm² resulted in good Doppler-to-manometer correlation at tunnel lengths of 1 mm to 4.0 cm (r = .98 to .99, slope = .94 to 1.07).
two methods correlated well with the 0.25 cm² tunnel area at tunnel lengths of 1 mm to 2 cm (r = .99, slope = .98 to 1.05); t values for all the above tunnels indicated that the slope was not statistically different from 1. The 3 and 4 cm tunnel lengths demonstrated a slope of 1.08 (r = .99) and 1.15 (r = .99), respectively, indicating an 8% and 15% underestimation of manometer-derived gradients by Doppler; t values indicated the slope was statistically different from 1. The 0.06 cm² tunnel area demonstrated striking underestimation of manometer-derived gradients by Doppler that increased as tunnel length increased. The r values were .99 for all tunnel lengths. The slope was 1.12, 1.14, 1.30, and 1.42 for the 1 mm, 1 cm, 2 cm, and 4 cm tunnel lengths, respectively (figure 5). This indicates that Doppler underestimates manometer gradients by 42% at the 4 cm tunnel length.

Discussion

It is important to emphasize the nonphysiologic nature of this model of flow through a stenosis. For example, the orifice is not a valve and there is no leaflet motion. Moreover, no attempt is made to mimic the contractile characteristics of the left ventricle. However, this simplified model does provide an ideal setting to evaluate the accuracy of the simplified Bernoulli equation. Previous verification of Doppler estimated pressure gradients in vitro has been limited to stenoses with discrete, round orifices. Holen et al. used a model somewhat different from ours, found good correlation between Doppler-derived and manometer-derived pressure gradients for orifice areas greater than 0.10 cm². Requarath et al. also found good correlation between Doppler-estimated vs manometer-measured gradients for valve areas ranging from 0.3 to 0.85 cm². Our experiments confirm the accuracy of Doppler gradient determinations with single round orifices. Doppler and manometer gradients were essentially identical for most orifices tested over a large range of pressure gradients. One unanticipated finding was that high gradients (>50 mm Hg) across small orifices (<0.06 cm²) resulted in signal dispersion. The Doppler waveform lost definition and differentiation between signal and noise became difficult. This signal dispersion probably occurs as flow becomes non-

<table>
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<th>Orifice Shape</th>
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<th>r</th>
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FIGURE 4. Irregular orifices used in the apparatus of figure 1 are shown at left (approximate cross-sectional area of each orifice is 1.0 cm²). Parameters of linear regression analysis of Doppler-estimated vs manometer-measured pressure gradients are listed for each orifice. The p value (indicating degree of doubt that the slope is not equal to 1) is <.5 for each orifice tested, indicating excellent correspondence between the two techniques.

<table>
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*p value (indicating degree of doubt that the slope is not equal to 1) is <.01. All other p values are <.5.
laminar and turns into a spray. Studies by Requarth et al., using flow systems with direct visualization of postobstructive jets, suggest that “spray” formation occurs at small orifice diameters. Spray formation could cause dispersion of the Doppler signal if blood flow becomes turbulent and no longer moves parallel to the transducer.

Our study is the first to demonstrate that the Doppler gradient technique can be applied to irregular stenoses, multiple orifices, and tunnel obstructions. We found Doppler measurement of jet velocities produced by various irregularly shaped orifices provided accurate estimations of pressure gradient. We also tested a more specific form of irregular stenosis, the presence of dual orifices.

When a smaller, secondary orifice was placed parallel to the large orifice, Doppler measurements remained accurate. Blood velocities were identical in both the small and large jets and both accurately reflected the pressure gradient. One might intuitively expect a higher velocity jet through the small orifice. This does not occur because the increased resistance offered by the small orifice in effect causes a shift of blood flow toward the low-resistance large orifice. The result is a net pressure drop between the upstream and downstream chambers, regardless of the geometry of the orifice. These results are best understood if one views the velocity of blood moving through an orifice as the direct result of the pressure gradient across that orifice. If the pressure gradients across two orifices are the same (and they must be if the two orifices are in parallel), the blood velocities generated by those pressure gradients must also be the same, as dictated by the Bernoulli equation. This should not be confused with the absolute quantity of blood flow (measured in ml/min), which is, of course, larger through the larger orifice. The velocity (measured in m/sec) refers to the speed of red blood cells as they move through the orifice. Thus more blood flows through the larger orifice but the velocity of red blood cells moving through each orifice is the same.

These results indicate that in severely distorted native and prosthetic valvular stenoses, where irregular flow or multiple jets are produced, Doppler estimation of pressure gradients is accurate. If the transducer is aimed parallel to one jet, the presence of other jets in different planes should not interfere with determination of Doppler pressure gradient.

Little data are available regarding the accuracy of Doppler measurements in tunnel obstructions in vivo. Our results in vitro indicate that Doppler-measured gradients are accurate within certain anatomic limitations. Tunnels with cross-sectional areas of 0.5 cm² or greater yielded accurate Doppler measurements, even at tunnel lengths of 4 cm. However, at tunnel areas of 0.25 cm² we found that Doppler under-
estimated pressures by 15% at the 4 cm tunnel length. This underestimation became severe in the 0.06 cm² tunnel area, where Doppler underestimation of the pressure gradient increased as the tunnel length increased. At the 4 cm length, Doppler underestimated manometer-measured gradients by as much as 42%.

One explanation for this underestimation is found in the Bernoulli equation:\(^\text{12}\):

\[ \text{PG} = 4 \frac{V^2}{L} + \text{flow acceleration} + \text{viscous resistance} \]

where PG is the pressure gradient and V is the maximum velocity.

The simplified Bernoulli equation omits the acceleration and resistance terms because in most clinically encountered stenoses they are negligible relative to the convective acceleration represented by the term 4V².\(^\text{12}\) This is supported by our data above. However, the pressure drop caused by viscous resistance can be calculated as described by Poiseuille’s law:\(^\text{13}\):

\[ \text{PGr} = \frac{8 \times V \times \mu \times L}{r^2} \]

where PGr is the pressure drop along a tunnel caused by viscous resistance, V is the Doppler-measured velocity of blood, \(\mu\) is the viscosity of blood (10⁻² Pascal sec),\(^\text{13}\) L is the tunnel length, and r is the tunnel radius.

It is apparent from this equation that with increases in tunnel length and decreases in tunnel diameter, viscous resistance produces a larger pressure drop. The pressure drop (mm Hg) attributable to viscous resistance can be calculated for each of our tunnel obstructions (see Appendix). For example, given a blood velocity of 5 m/sec through a 0.06 cm² tunnel that is 4 cm in length, the underestimation predicted by the Poiseuille equation is 38.7%. This is reasonably close to our observed 42% underestimation. Some differences are expected both because of assumptions in the calculation about blood viscosity and because of the application of the Poiseuille equation (based on constant flow) to our model (which uses pulsatile flow).

Our results indicate that Doppler-estimated pressure gradients are accurate in most clinically encountered stenoses (>0.25 cm² in cross-sectional area and <1.0 cm in length). Perhaps the most common clinical situation in which Doppler-estimated pressure gradients may prove inaccurate is in the setting of epicardial coronary artery disease. A 50% stenosis in the diameter of 6 mm coronary vessel results in a cross-sectional area of 0.068 cm², whereas a 75% diameter reduction gives an area of 0.018 cm². Thus coronary stenoses are in the range where our tunnel experiments demonstrate underestimation of pressure gradients by Doppler.

In summary, we found that Doppler ultrasound ac-curately estimates pressure gradients across a wide variety of clinically encountered obstructions to blood flow. As studied here, Doppler measurements are unaffected by eccentric or multiple orifices. The pressure gradient across most tunnel stenoses can be measured by Doppler but underestimation occurs in long, very narrow obstructions. Pressure drops along tunnels with cross-sectional areas less than 0.25 cm² are subject to significant underestimation by Doppler. This validates the accuracy of Doppler measurements in most clinical situations but raises serious questions about its use in determining gradients across coronary arteries.

Addendum

Since submission of this work, Wong et al.\(^\text{14}\) have reported similar findings in a dual orifice flow model.

We appreciate the excellent technical assistance of Bruce Director, Cecil Profitt, Eben Kermit, James Johnston, Ph.D., and Arthur Casper, M.S. We also thank Helen Kramer, Ph.D., for her help with the statistical analysis and Gretchen Houd for her preparation of the manuscript.

Appendix

Calculation of the pressure drop caused by viscous resistance using Poiseuille’s law

Using Poiseuille’s equation to calculate the expected viscous resistance, we made the following assumptions: (1) Blood viscosity = 10⁻² Pascal sec.\(^\text{13}\) We did not measure the actual blood viscosity in our system. However, we suspect the viscosity will be somewhat less than 10⁻² Pascal sec because we used only red blood cells diluted with saline. (Other components contributing to blood viscosity, i.e., platelets, white cells, and plasma proteins, are excluded). (2) The Reynolds’ number is less than 2000, indicating laminar flow. (3) Pousille’s equation, derived for continuous flow, is applicable to pulsatile flow.

Poiseuille’s equation:

\[ \text{PG} = \frac{8 \times V \times \mu \times L}{r^2} \]

where PG is the pressure drop (in Pascals), V is the blood velocity (m/sec), \(\mu\) is the blood viscosity (Pascal sec), L is the tunnel length (m), and r is the tunnel radius (m).

Examples:

1. Tunnel area = 0.06 cm² (radius = 0.0014 m)
   Tunnel length = 1.0 cm (0.01 m)
   Blood velocity = 5 m/sec

\[ \frac{8 \times 5 \text{ m/sec} \times 10^{-2} \text{ Pascal sec} \times 0.01 \text{ m}}{0.0014^{2} \text{m}^{2}} = 2105.26 \text{Pascals} \]

Therefore PG = 15.78 mm Hg.

2. For the same conditions as in example 1, but with a tunnel area of 0.25 cm² (radius = 0.0028 m), PG = 3.85 mm Hg.

3. For the same conditions as in example 1, but with a tunnel area of 0.5 cm² (radius = 0.004 m), PG = 1.88 mm Hg.

4. For the same conditions as in example 1, but with a tunnel area of 1.25 cm² (radius = 0.0063 m), PG = 0.76 mm Hg.

In each of the above examples, blood velocity was 5 m/sec. With the simplified Bernoulli equation (PG = 4V²), the expect-
ed pressure drop is 100 mm Hg. Our calculations, however, indicate that there is an additional pressure drop caused by viscous resistance. Without taking these resistive forces into account, the simplified Bernoulli equation will provide an underestimation of the actual pressure drop. In the 1.25 and 0.5 cm² tunnels, this additional pressure drop is small (0.76 and 1.88 mm Hg/cm tunnel length, respectively). Calculations for the 0.25 cm² tunnel indicate a pressure drop of 3.85 mm Hg/cm tunnel length. Therefore, in this tunnel, at the 3 and 4 cm lengths, we would expect underestimations of 11.55 and 15.4 mm Hg, respectively. When the tunnel cross-sectional area is reduced to 0.06 cm², the expected pressure drop caused by viscous resistance becomes quite significant. Here the calculated underestimation is 15.78 mm Hg/cm tunnel length. Therefore, in the 4 cm tunnel the Poiseuille equation predicts an underestimation of 63.12 mm Hg. This means that a pressure gradient of 163.12 mm Hg is required to generate a velocity of 5 m/sec (instead of the 100 mm Hg pressure gradient predicted by the simplified Bernoulli equation: \( PG = 4V^2 \)). By not taking the resistive forces into account, the simplified Bernoulli equation has underestimated the actual pressure gradient by \((163.62-100)/163.12\) or 38.7%. Our experimentally observed underestimation was 42%.

Therefore, the resistive forces as calculated by the Poiseuille equation offer one possible explanation for the underestimation of pressure gradients by Doppler ultrasonography in long, narrow tunnel-like obstructions.

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The accuracy of Doppler ultrasound measurement of pressure gradients across irregular, dual, and tunnellike obstructions to blood flow.

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Circulation. 1985;72:577-584
doi: 10.1161/01.CIR.72.3.577

Circulation is published by the American Heart Association, 7272 Greenville Avenue, Dallas, TX 75231
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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:
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