A pulsed Doppler echocardiographic method for calculation of pulmonary and systemic flow: accuracy in a canine model with ventricular septal defect

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ABSTRACT This study was designed to assess the accuracy of a quantitative, two-dimensional, range-gated, Doppler echocardiographic method for estimating systemic and pulmonary flows by comparing them with simultaneous electromagnetic (EM) flow-meter determinations in an open-chest canine model with a ventricular septal defect, the size of which was varied. With each step-by-step variation in defect size, systemic and pulmonary flows were estimated by the Doppler method and matched to simultaneous pulmonary and aortic EM flow-meter recordings. The Doppler value for systemic blood flow was obtained by measuring the size of the aorta on a two-dimensional echocardiographic image and sampling blood flow velocity in the ascending aorta. Because of high flow velocities and spectral broadening on the pulmonary artery Doppler flow curves in the presence of ventricular septal defects, pulmonary blood flow was measured as pulmonary venous return through the mitral valve orifice, with the size of the mitral orifice measured on the short axis by two-dimensional echocardiography and with transfow curves obtained from apical views. For a total of 32 left-to-right shunt magnitudes obtained in 10 dogs, excellent correlations were found between ascending aortic Doppler and EM flow-meter values ($r = 0.91$, SEE = 0.11), between Doppler mitral valve flow and EM flow-meter main pulmonary artery flow values ($r = 0.93$, SEE = 0.32), and between Doppler and EM flow-meter determinations of the ratio of pulmonary to systemic flow ($r = 0.93$, SEE = 0.52). Our study demonstrates the accuracy of this two-dimensional echocardiographic Doppler method for measuring pulmonary and systemic flows and their ratios in the presence of ventricular septal defects. *Circulation* **68**, No. 3, 597–602, 1983.

RECENTLY PUBLISHED STUDIES have demonstrated that estimates of cardiac output by two-dimensional echocardiographic Doppler methods are reasonably accurate in the presence of normal or nearly normal circulatory physiology.1-3 Other preliminary reports have suggested that even in the presence of small intracardiac shunts, Doppler determinations of systemic and pulmonary flows may also be close to those measured by simultaneous invasive techniques.4

The purpose of our study was to assess the accuracy of systemic and pulmonary flow values measured with a quantitative two-dimensional, range-gated, echocardiographic Doppler flow meter by comparing them with simultaneously recorded electromagnetic (EM) flow-meter readings in an open-chest canine model with a ventricular septal defect of variable size.

Methods

Surgical technique and animal model. Ten mongrel dogs weighing 20 to 30 kg 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 cleaned of fat and adventitia, and appropriately sized EM flow probes (Gould-Statham SP 2204) that were previously calibrated against a mechanical roller pump were placed around each vessel 2 cm above the distal end of the sinuses of the semilunar valves. A cylindrical cutter was then introduced through a right atrial purse-string suture and forced through the ventricular septum to create a muscular or perimembranous defect. A 12 mm outer diameter, 9 mm inner diameter, 16 mm long flanged cylindrical nylon prosthesis was advanced over the cutter and placed snugly into the defect. The lumen of the prosthesis contained a latex membrane that was inflated with
saline via a small vinyl tube exteriorized through the atrial purse string. When inflated, the latex membrane constricted circumferentially, thereby varying the size of the interventricular communication.

EM flow-meter recordings served as the reference standard for comparison with Doppler-calculated flows. Adequate contact of both EM flow-meter cuffs with the vessels was verified by recording phasic aortic and pulmonary flow traces (figure 1). After each step-by-step variation in defect size, the animal was allowed a period of at least 2 min once the EM flow-meter readings stabilized at the new flow rates. Doppler measurements of systemic and pulmonary flows were calculated from RR matched beats, as described below, for each size defect, and were matched to the simultaneous EM flow-meter recordings to permit calculation of the ratio of pulmonary to systemic flow (QP:QS) by both methods.

Ultrasound and Doppler methods. Ultrasound imaging and Doppler studies were performed with a commercially available, range-gated, pulsed Doppler unit (Electronics for Medicine/Honeywell). The unit contains a 3.5 MHz single-element transducer mechanically swept through a 30 to 75 degree arc to achieve real-time two-dimensional echocardiographic imaging at 30 frames/sec. The scanner could be stopped along any line within the image and a Doppler sample volume could be positioned at any depth along that line; this permitted precise localization of the sample volume and determination of the angle of Doppler sampling relative to the direction of flow within the plane of imaging. The sampling angle relative to the direction of flow within the elevational or azimuthal plane, that is, the plane perpendicular to the plane of imaging, could not be determined; however, small deviations from sampling exactly parallel to flow (angles = 0 or 180 degrees) were of no practical importance since the cosine of the sampling angle would still be close to unity (see equation 1). Sample volume length varied between 2 mm and 2 cm and was usually 5 mm in these studies. Sample volume width in a water tank at 6 dB was ± 2 mm at 4 to 8 cm depth. The operational mode of the scanner could be switched from real-time to Doppler imaging in less than 1/10 sec. In Doppler mode, signals were sampled at a pulse repetition frequency of 19,500 samples/sec when the signal was obtained from a depth less than 4 cm, resulting in a maximal nonambiguously detectable velocity of ± 220 cm/sec; at a depth of 4 to 8 cm, signals were sampled at 9750 samples/sec, resulting in a maximal nonambiguously detectable flow velocity of 110 cm/sec at a 0 degree sampling angle. Two outputs of the Doppler frequency shift were available: an audio signal and a fast Fourier transform spectral analysis of the Doppler shift sampled at 200 times/sec. The fast Fourier output was converted automatically by the scanner to flow velocity (in cm/sec) with the equation

\[
\text{Flow velocity} = \frac{\text{frequency shift} \times \text{velocity of sound in medium}}{2 \times \text{transmitted frequency} \times \cos \theta}
\]  

(1)

Correction for \( \cos (\theta) \), that is, the angle between the Doppler beam and the direction of flow, was not applied automatically by the unit, but rather was done manually with equation 2 (below).

Freeze frames of the two-dimensional images and the Doppler velocity outputs obtained at a paper speed of 100 cm/sec were recorded on strip-chart page prints and/or in real time on videotape along with lead II of the electrocardiogram.

Doppler measurements of blood flow volumes. Pulmonary and systemic blood flow volumes were calculated from the two-dimensional images and the flow velocity curves with the general equation

\[
\text{Blood flow/min} = \frac{\text{mean flow velocity} \times \text{cross-sectional area} \times 60 \text{ sec/min}}{\cos \theta}
\]  

(2)

where mean flow velocity throughout the cardiac cycle, uncorrected for angle, is in centimeters per second and cross-sectional area is in square centimeters.

The sampling angle \( \theta \), that is, the angle of incidence between direction of flow and the Doppler sample volume, was determined manually with a protractor directly from the freeze frame of the two-dimensional image that showed the sample volume position relative to the imaged cardiac structures (figure 2). Correction for angle \( \theta \) was done manually with equation 2 rather than with equation 1.

Aortic flows. In our animal model systemic blood flow was measured in the ascending aorta. A two-dimensional image of the ascending aorta was obtained by lightly positioning the

FIGURE 1. Example of an EM flow-meter curve derived from the main pulmonary artery demonstrating increasing pulmonary blood flow as the occluder membrane is deflated, thereby opening the defect.

FIGURE 2. Echocardiographic image of the ascending aorta (ASCAO), aortic arch, and descending aorta with the Doppler sample volume (SV) positioned along the line of flow within the ascending aorta. Doppler velocity output derived from that sampling site is shown below.
transducer directly over the ascending aorta proximal to the EM flow meter. The Doppler sample volume was placed within the central ascending aorta and adjusted so that no valve leaflets crossed it. Once the optimal two-dimensional image of the ascending aorta was obtained and the sample volume confirmed to be as parallel as possible to the apparent direction of flow (angle $\theta = 0$ degrees), the two-dimensional image and the fast Fourier–Doppler output were recorded on a strip chart and videotape (figure 2). Mean temporal flow velocity was obtained by digitizing and integrating the area under the Doppler wave form over complete cardiac cycles with a minicomputer (Numonics). Aortic cross-sectional area was calculated from maximal systolic inner diameter measurements obtained from the two-dimensional echocardiographic images and was not corrected for variation in size during the cardiac cycle.

**Pulmonary flow.** Initially, as in our previous work, we attempted to measure pulmonary blood flow by Doppler sampling of flow velocity in the main pulmonary artery beyond the pulmonic valve from a parasternal short-axis view. However, spectral broadening of the Doppler flow signals and velocities higher than the Doppler Nyquist limit were present in pulmonary artery flow curves when the ventricular septal defects were open and this precluded quantification of the Doppler-derived pulmonary artery flow traces. Pulmonary flow was therefore measured as pulmonary venous return, i.e., flow through the mitral valve orifice.5

**Transmitral flows.** To accomplish this, a two-dimensional image of the mitral valve was obtained by positioning the transducer at the cardiac apex so as to obtain a two- or four-chamber view. The Doppler sample volume was then placed within the inflow tract of the left ventricle and adjusted so as to be inferior to the mitral leaflets and lateral to the left ventricular outflow tract. Once the optimal two-dimensional image and Doppler velocity waveforms were obtained and the sample volume confirmed to be as parallel as possible to the apparent direction of flow, the two-dimensional image and the fast Fourier spectral output of the Doppler flow velocity were recorded on a strip chart and videotape (figure 3, A). Transmitral Doppler flow curves were digitized over complete cardiac cycles with a minicomputer (Numonics) to calculate mean temporal mitral valve flow velocity. To determine the mitral valve flow area, the transducer was placed directly over the mitral ring scanning in a short-axis plane and a gated freeze frame was recorded at maximal orifice opening. Maximal area was determined by digitizing the orifice through the inner contour of the leaflet echoes. Since the mitral valve is not maximally opened throughout diastole, a
Correction factor for its phasic diastolic motion was calculated as the ratio of mean to maximal leaflet separation traced through the middle of the leaflets on a derived M mode trace. When the maximal orifice on the short axis, as measured by two-dimensional echocardiography, was multiplied by the correction factor, it resulted in the effective mean cross-sectional mitral valve orifice area during diastole (figure 3, B).

**Digitizing methods — calculation of mean temporal flows.** Mean flow velocities for the aorta or the mitral valve were calculated by digitizing and integrating the areas under three consecutive RR matched Doppler flow velocity curves. To accomplish this, we traced the middle of the most dense portion of the gray-scale spectral display of the Doppler velocity curve, that is, the modal velocity shift, which is the shift most frequently present in the returning signal. The minicomputer divided the velocity time integrals for the 3 beats by the time of the 3 beats to yield mean aortic or mitral flow velocity with respect to time.

For the mitral valve recordings, the sample volume had been moved laterally to avoid recording flow in the left ventricular outflow tract. For all Doppler recordings a low reject setting was used in recording the Doppler signals, which resulted in low-velocity noise when phasic flow was not occurring, as in systole for the mitral valve and in diastole for the aorta. These low-velocity signals and any flows below the zero line, when present, were ignored during the tracing of the curves. As such, during digitizing the aortic velocity curves were traced along the zero line through diastole to the beginning of the next systole, and the mitral valve curves were traced along the zero line during systole to the following diastole, as shown in figures 2 and 3. The inclusion or exclusion of these low-velocity signals had a minimal effect on mean temporal velocities calculated but, in our previous studies, resulted in greater reproducibility of the data and good agreement with EM flow determinations.1, 5

**Repeatability of measurements.** To determine repeatability, all measurements were made in duplicate. To test interobserver variability, all measurements and calculations were made by two investigators who were unaware of each other’s results, or of the simultaneous EM flow-meter readings.

**Statistical analysis.** Linear correlation was used to compare pulmonary and systemic flows and QP:QS ratios obtained by Doppler echocardiography to those recorded with the EM flow meter. The paired t test was used to analyze measurement repeatability and interobserver variability.

**Results**

A total of 32 left-to-right interventricular shunts were obtained from the 10 experimental animals. In each animal a minimum of two and a maximum of eight shunt magnitudes were recorded. Aortic flows determined by the EM flow meter ranged from 0.4 to 1.66 l/min (mean = 0.97, SE = 0.05). Pulmonary flows ranged from 1.3 to 4.8 l/min (mean = 2.9 ± 0.16), and the QP:QS ratio ranged from 1.4:1 to 6:1 (mean = 3.2:1 ± 0.22).

One ascending aortic Doppler flow measurement was discarded because the two-dimensional aortic image for diameter measurement was not adequate and one Doppler mitral flow determination was discarded because of an inadequate two-dimensional image of the valve orifice. This resulted in 31 Doppler systemic flow values ranging from 0.6 to 1.5 l/min (mean = 0.99 ± 0.1), 31 Doppler pulmonary flow values ranging from 1.5 to 4.4 l/min (mean = 2.9 ± 0.16), and 30 Doppler QP:QS ratio determinations ranging from 1.4:1 to 7:7:1 (mean = 3.3:1 ± 0.25). Peak aortic Doppler flow velocities ranged from 30 to 143 cm/sec, with a mean spectral width (6 dB) of 15.3 ± 0.8 cm/sec. Mitral valve peak flow velocities ranged from 35.8 to 125 cm/sec, with a mean spectral width of 12.0 ± 9.0 cm/sec. Thus, pulmonary flow measured at the mitral orifice resulted in modal velocities within the Nyquist limit of the ultrasound system (220 cm/sec).

There was good correlation between Doppler and EM flow readings for the ascending aorta (figure 4, A; r = .91, SEE = 0.11), the pulmonary artery and mitral flows (figure 4, B; r = .93, SEE = 0.32), and the QP:QS ratios (figure 4, C; r = .93, SEE = 0.52).

**Interobserver variability and errors of repeatability.** All determinations were highly repeatable. The standard error of the mean testing repeatability was less than 5% when comparing duplicate measurements by one observer on a given record. Interobserver variability was also less than 5%.

**Discussion**

Our previous study results,1-5 and those of other investigators,6-11 suggest that Doppler velocity measurements can be used to quantitate cardiac output. Our present results demonstrate that range-gated, two-dimensional Doppler echocardiography with quantitative fast Fourier transform spectral output allows accurate estimation of pulmonary and systemic blood flows and QP:QS ratios even in the presence of large shunts through ventricular septal defects in an open-chest animal model.

Our previous studies with Doppler flow measurements in humans and in open-chest animals with intact circulations have shown that the greatest source of error in the Doppler methodology lies in the measurement of vessel radius from a two-dimensional image.1-4, 5 Since the measured vessel diameter is raised to the second power to calculate vessel “flow” area, small errors are compounded and can result in significant inaccuracies in flow calculations. These problems were exaggerated in our earlier studies since the prototype equipment had not been optimized for near-field imaging; therefore, transducer artifacts often made the vessel wall difficult to recognize in a stop frame. In our present study, a second generation system was used in which technical modifications improved near-field imaging. As a result, we were able to measure aortic diameter with acceptable accuracy even with direct transducer placement over the aortic arch.
Variations in Doppler power, reject threshold, and tracing method in our previous studies have produced little error\textsuperscript{1-5} in application of the Doppler technique to the calculation of volume flow. Nonetheless, in the presence of very low volume flow, the profile distal to the aortic or mitral valve may not be flat, with most flow occurring in the central lumen. Under these circumstances, the Doppler method may overestimate volume flow, as is suggested by our results and those of previous investigators.\textsuperscript{6-9} According to our data, the intercepts of both the mitral and aortic regression curves (figure 4, A and B) are slightly above the origin, although not statistically different from zero.

The mitral valve orifice method, previously described as an alternative for measuring cardiac output in patients with intact circulations,\textsuperscript{5} allows the calculation of pulmonary blood flow if the atrial septum is intact, even in the presence of high-velocity and disorganized flow within the pulmonary artery. Since mitral flow velocities were measured from within the valve orifice itself, that is, proximal to the ventricular septal defect, we hypothesized that these flows represent pulmonary venous return. The excellent correlation of Doppler results to flows measured with the EM flow meter placed around the main pulmonary artery served to confirm our hypothesis. A note of caution is appropriate with respect to sample volume placement. We did not examine flow patterns within the left ventricle in depth; however, in our animal studies and in our initial clinical applications of the Doppler method\textsuperscript{12} sample size and position were adjusted to maximize diastolic flow toward the transducer and to avoid the left ventricular outflow tract and the area around the ventricular septal defect itself. As such, the sample volume was placed as laterally as possible within the mitral inflow.
The mitral valve orifice can be examined and measured directly by two-dimensional echocardiography and our previous observations suggested that the valve expands to accommodate increased flow. Therefore we believe that mitral orifice size determined by two-dimensional echocardiographic images accurately reflects flow area. Since the mitral valve orifice is not maximally opened throughout diastole, but rather varies with phasic ventricular filling, a correction factor based on leaflet separation calculated from the derived M mode trace was used to arrive at an effective mitral valve orifice area. This method resulted in excellent correlations between the area and a strict reference standard for flow both in our previous studies and in our present one. The model of ventricular septal defect used in our study ensures that a surgically created defect will remain open, permitting evaluation of several shunt magnitudes in the same experimental animal.

In conclusion, our study demonstrates the accuracy of this two-dimensional echocardiographic Doppler method for measuring pulmonary and systemic flows and QP:QS ratios in the presence of ventricular septal defects. Constancy of heart rate and flows in our model, as in most closely controlled animal experiments, and the open-chest method we used, minimized technical difficulty in the two-dimensional echocardiographic Doppler method. However, results of this study and of our preliminary studies in which we applied this method to children with congenital heart disease suggest that this technique may prove useful in the evaluation and management of children with simple left-to-right shunting ventricular septal defects.

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