Combined Doppler and Phased-array Echocardiographic Estimation of Cardiac Output

PAUL A. MAGNIN, B.S.E., JAMES A. STEWART, M.D., SONDRA MYERS, R.D.M.S., OLAF VON RAMP, PH.D., AND JOSEPH A. KISSLO, M.D.

SUMMARY  The capability of a pulsed Doppler flowmeter combined with a phased-array imaging system to measure volume flow was tested in vitro and in patients undergoing cardiac catheterization. The Doppler-phased-array system (DPA) was used to determine vessel diameter and a superimposed cursor was used to locate the range and angle of the Doppler sample volume. DPA estimates of continuous flow through tubing in a water tank correlated strongly (r = 0.99) with measured flow corresponding to physiologic ranges from 3-12 l/min. For pulsatile flow in a water tank, a correlation of r = 0.86 with measured flow was obtained, whereas DPA estimates of cardiac output as compared with Fick estimates in the 11 patients produced a correlation of r = 0.83. These data indicate that estimates of cardiac output are possible using the DPA approach.

A NONINVASIVE and quantitative measurement of cardiac output using ultrasound would provide a means of obtaining serial evaluations over a long period of time, which would be useful in identifying and projecting trends in cardiac function.

Accurate measurement of volume flow using a pulsed Doppler flowmeter requires determination of the angle of the Doppler beam with respect to the vector representing flow velocity, the cross-sectional area of the vessel, and the flow profile at the cross-section being interrogated. By combining a pulsed Doppler flowmeter with a phased-array scanner, an image of a vessel is produced with a cursor line, corresponding to the path of the Doppler sound beam, superimposed on it. This facilitates measurement of the angle between the sound beam and the vessel and the cross-sectional area of the vessel. Using such a system to measure continuous flow in a water tank, Schwartz and de Cristofaro found that estimates of flow velocity varied significantly with both the hematocrit and angle of the Doppler sound beam with respect to the flow vector. Calibration curves were determined for both hematocrit and angle.

In this preliminary study, we tested the hypothesis that volume flow can be measured over a physiologic range of flow velocities without hematocrit or angle calibration curves. We used a combined Doppler and phased-array (DPA) system to estimate continuous and pulsatile flow in vitro and cardiac output in patients undergoing cardiac catheterization. This ultrasonic technique provides a noninvasive, nonionizing method for measuring cardiac output and is suitable for serial evaluations.

Methods

Patients

Sixteen patients undergoing cardiac catheterization for reasons other than valvular or congenital heart disease were examined. Eleven patients, ages 36–65 years (average 58 years), who had ultrasonic data of suitable quality for analysis, were included in this study. All patients had coronary artery disease and were without evidence of valvular or congenital impairment at catheterization. The DPA estimates of cardiac output were made within 1 day of the Fick estimates collected at catheterization. Fick determinations were made immediately after catheter insertion and before angiography. Heart rates for each patient were measured at the time of the Fick and at the time of the DPA determinations and varied less than 5%, implying similar physiologic conditions for the two determinations.

Transducer Assembly

The transducer from a commercially available 3-MHz pulsed Doppler flowmeter (Advanced Technology Laboratories Model #500A) was mounted on the side of a focused, phased-array transducer (Duke University) using an angle-sensing device to digitally encode the angle between the Doppler transducer and the axis of the phased-array display circuitry (fig. 1). A cursor line corresponding to the position of the Doppler beam was written on the display monitor over the phased-array tomographic image. The depth of the sample volume was similarly digitized and written as a bright spot on the cursor line (fig. 2).

Scan Technique

The Doppler transducer was located in the fourth intercostal space at the left sternal border. The Doppler sample volume was located in the aortic outflow tract just above the valve leaflets and the phased-array plane was oriented parallel to and intersecting the axis of the vessel. The range cursor was positioned within the vessel at the point that had the greatest
analog-net-flow (ANF) output to ensure that it was centered in the vessel. Angles from 40–68° were used and care was taken to ensure that the aortic valve did not enter the Doppler sample volume (fig. 2).

Doppler Estimates

Estimates of continuous flow were made in plastic tubing with a 3-mm inner diameter suspended in a water tank. The scattering medium consisted of a suspension of Dupont microspheres (4 μ in diameter) in water. The concentration of the microspheres was made sufficiently high to produce strong flow signals and remained fixed for all of the in vitro experiments. Changes in the microsphere concentration had no noticeable affects on the flow signal. Flow velocity was varied by changing the level in a constant-pressure-head tank. Although the velocities measured in vitro were equal to those in patients with cardiac outputs ranging from 3–12 l/min., the actual volume flow was much smaller due to the difference in the cross-sectional areas of the plastic tubing and the aorta. Seven flow determinations were made at each of five different angles and were correlated with flow measured simultaneously using a graduated cylinder for collection and a stopwatch.

Estimates of pulsatile flow duplicated those of continuous flow except that a variable-speed Travenol roller pump was used to create the flow. The rollers were pressed tightly against the outer cylinder. After each forward pulse, a short burst of reverse flow would occur when the leading roller left the cylinder edge, thereby mimicking the biphasic flow found in the aorta. The flow rate was varied by changing roller speed.

Graphic information from the Doppler flowmeter was displayed and recorded on an Irex 101 Continu- trace recorder and consisted of an ANF output, the M-mode echocardiogram, the ECG signal and both the forward and reverse flow signals (fig. 3). The area between the ANF output and a "zero-flow" baseline was then integrated using a Science Accessories Corporation graf/pen (interfaced with a PDP 11/70 computer) over 10–15 pulsatile cycles (or 10–15 seconds in the case of continuous flow) measured from the maximum flow point on the first cycle to the maximum flow point on the last cycle. This integral was then divided by the total time elapsed during the integration and by the cosine of the angle between the vessel axis and the Doppler beam to obtain an average flow per second as measured by the DPA system. The ratio of the average flow per second measured using the stopwatch and graduated cylinder to that measured using the DPA technique then determined the calibration constant:

$$FC = FL/\left(\frac{1}{T \times \cos(\theta)}\right) \int_{0}^{T} ANF \, dt$$
where \( FC \) = flow constant, \( FL \) = average flow/second measured using a graduated cylinder and a stopwatch, \( \theta \) = angle between the vessel axis and the Doppler beam, and \( T \) = integration time.

All gain and threshold settings on the pulsed Doppler unit remained unchanged throughout the studies. The far gain and slope were set at 10, the near gain and delay at 0, the threshold at 8 and the overall gain at 6. The estimate of the flow was then made using the calibration constant in the following equation:

\[
FL = \frac{FC \times D_a^2 \times \frac{1}{T} \int ANF \, dt}{D_t \times \cos(\theta)}
\]

where \( D_a \) = internal diameter of aorta and \( D_t \) = internal diameter of plastic tube used in continuous and pulsatile studies. Determinations at angles greater than 70° were discarded due to the increased sensitivity to small errors caused by a small \( \cos(\theta) \) term in the denominator.

Results

For continuous flow in the water tank, the DPA estimate correlated extremely well with the actual flow \((r = 0.99)\) over flows corresponding to 3–10 L/min and angles of 47–66°. The continuous flow data are shown graphically in figure 4. The root mean square (RMS) error from the regression line is 4.0 ml/min. Flow at angles approaching 70° was noticeably less accurate than flow at more acute angles.

Estimates of pulsatile flow using the DPA technique also showed good linear correlation \((r = 0.86)\) with actual flow (fig. 5). The angle, as measured by the

**Figure 3.** M-mode echocardiogram of aortic outflow tract displaying sample volume (SV), aortic outflow root (AoR), ECG, analog-net-flow (ANF), forward flow (FF), zero-flow line (ZFL) and reverse flow (RF). The analog-net-flow signal has been shifted upward from the zero-flow line for the purpose of presentation; the zero flow line (not shown) would also be shifted positively.

**Figure 4.** The relationship between continuous flow measured using a stopwatch and a graduated cylinder and the Doppler phased-array system. Each point represents the average of three determinations of flow at a given angle and flow velocity. Seven determinations were taken at each of five angles and a least-squares regression line is superimposed. A unique scaling factor was determined for each angle to minimize systematic error caused by angle measurement.

**Figure 5.** The relationship between pulsatile flow measured using a stopwatch and a graduated cylinder and the Doppler phased-array system. Each point represents the average of three determinations of flow at a given angle and flow velocity. Seven determinations were taken at each of five angles and a least-squares regression line was superimposed.
DPA system, between the Doppler beam and the vessel was used in determining a single calibration constant for all of the angles. Systematic error in the flow estimate associated with the angle was particularly evident in the $65^\circ$ and $68^\circ$ determinations. Multiple determinations of flow at a particular point and angle showed only small variation. Less accuracy was observed at large angles. The RMS error from the regression line is 8.7 ml/min. High flows were measured more consistently than low flows.

For the 11 patients studied, a correlation coefficient of $r = 0.83$ with the Fick method estimates was found (fig. 6). However, there was a significant offset from a line through the origin. The slope of the regression line was 0.60. Determinations from different transducer locations on the same patient varied considerably, although multiple determinations at a single transducer location and angle showed little variation. Measurements that were as small as half the size of measurements made in other transducer locations were found in three of the 11 patients. Examination of video tapes showed only small variation in the diameter of the aortic outflow tract over the cardiac cycle.

Discussion

Previous attempts at measuring cardiac output non-invasively,11,12 such as thoracic impedance techniques, have resulted in unacceptably large variations. Constant-infusion radioisotopic dye techniques13 have correlated reasonably ($r = 0.72$) with the Fick and the indicator-dilution methods, but measurements cannot be repeated in a short period of time, owing to background build-up of radioisotopic dye. As a consequence of the long computation time, this method can provide only a time-averaged estimate of cardiac output. The Fick and indicator-dilution approaches were compared in 105 patients in whom both methods were performed simultaneously.14 Eighty-four percent of the duplicate determinations were within 10%, and 98% were within 25%, of each other. These data indicate the clinically acceptable variations in results for methods in common use. Using the DPA method it is possible to measure beat-to-beat cardiac output as often as desired with no change in the signal-to-noise ratio.

A previous attempt at noninvasive measurements of cardiac output using continuous-wave Doppler in the descending aorta15 was limited by the variations in the cross-sectional area of the vessel along its axis and by the inability to measure the angle between the Doppler beam and the flow vector. The pulsed Doppler unit does not have this problem, as it senses flow in a small sample volume rather than along a line.

In all DPA volume flow determinations, it is necessary to assume one dimensional, cylindrically symmetrical flow. In the absence of flow profile information, an estimate of the mean velocity is sufficient. It is also assumed, for the sake of simplicity, that the vessel interrogated is perfectly rigid and that the ANF output from the Doppler flowmeter does not vary with differing tissue absorptions and hematocrits. Attempts to measure cardiac output using ultrasound suffer from the limitations of these assumptions. The DPA estimate of volume flow in vitro indicates that, carefully calibrated, such a system would provide acceptable estimates of cardiac output. The applicability of this approach for estimating cardiac output in patients must remain in question despite the encouraging results of this initial study. Several limitations of this combined approach are apparent. The non-zero offset in figure 6 is most likely a manifestation of the differing acoustic properties of the blood and the microsphere suspension. The correlation coefficient ($r = 0.83$) for the 11 patients is encouraging because it accounts for the variation from actual cardiac output of both the DPA estimate and the Fick estimate. The correlation coefficient compares favorably with those relating the Fick method to indicator-dilution methods.

Errors in angle measurement may have occurred due to refraction of the Doppler beam and inaccurate alignment of the plane of the phased-array transducer with the vessel axis. In this system little can be done about refraction-induced errors. Inaccuracies in the alignment of the two-dimensional plane with the vessel axis would result in artificially small estimates of cardiac output and could have been the cause of the large variations found with different transducer locations. This alignment is particularly difficult in short, curved vessels like the ascending aorta. Because the cosine of the angle is used in estimating flow, angle errors have a more detrimental affect on the flow estimate as angles approach $90^\circ$ than for smaller angles. This may

![Figure 6. Cardiac output in 11 patients determined by the Fick estimate is compared with the Doppler phased-array estimate. Each point represents the average of three determinations from each of two, and when possible three, transducer locations.](http://circ.ahajournals.org/)

- The correlation coefficient of $r = 0.83$ with the Fick method estimates was found (fig. 6).
- There was a significant offset from a line through the origin. The slope of the regression line was 0.60.
- Measurements that were as small as half the size of measurements made in other transducer locations were found in three of the 11 patients.
- The non-zero offset in figure 6 is most likely a manifestation of the differing acoustic properties of the blood and the microsphere suspension.
- Errors in angle measurement may have occurred due to refraction of the Doppler beam and inaccurate alignment of the plane of the phased-array transducer with the vessel axis.
in part explain why, even in the in vitro experiments, less accuracy was found at angles larger than 65°.

Despite the care taken to avoid having the aortic valve leaflets enter the Doppler sample volume, this may have occurred in some cases and caused artificially large ANF outputs. Although the Doppler shift frequency is not inherently affected by the acoustic density of the scattering medium, the zero-crossing frequency detector used in this particular pulsed Doppler unit does have a finite threshold, which results in a nonlinear response. Careful attention to either the A-mode or audio signal helps minimize this source of error. The A-mode and B-scan also indicate whether the vessel changes caliber significantly during the cardiac cycle. If the amount of dilatation is great, it should be accounted for in the flow calibration.

The flow profile in the aortic outflow tract was assumed to be uniform. No experiments to test this assumption were made, but little change in the ANF output was noticed as the sample volume was moved across the vessel. All vessels were assumed to be cylindrical. Care must be taken to measure the diameter of the vessel accurately, because cardiac output has a squared dependence on this variable. In this study, the vessel diameter was measured on the two-dimensional display; however, the M-mode or the A-mode from the Doppler unit can be used if the angle between the vessel axis and the Doppler beam is known. The two-dimensional measurement was used to ensure that the diameter was estimated at the location of the Doppler sample volume.

Errors may have been due to variations of the ANF output with hematocrit, with depth of vessel and with nonlinearities within the pulsed Doppler unit. No corrections were made for these sources of error. Calibration curves can be made for each source and errors can be further reduced, although variations in the ANF signal amplitude with differing tissue absorptions would still exist due to the finite threshold of the different components of the system. The errors introduced by the angle measurement and the graph/pen integration are believed to be small by comparison with other sources of error.

These data can also be adjusted for variations in cardiac output caused by different respiratory rates. It is doubtful that any such adjustment would significantly improve the Fick correlation, as the ANF was integrated over 10–15 seconds, which would include at least one respiratory cycle in most patients. The DPA estimate, being a time-averaged determination, would be expected to correlate more closely with a Fick estimate (due to the time-averaged nature of the Fick method) than with a single-cycle angiographic determination.

The large variations in cardiac output estimates from different transducer locations limit the clinical usefulness of the DPA system. To reduce the expected error of the estimate it is necessary to perform a number of output determinations at different transducer locations and average these values. Aside from the time involved in obtaining and processing multiple determinations, it is frequently difficult to find more than two suitable acoustic windows. This variation with transducer location is presumably caused by errors associated with angle, tissue depth (and therefore tissue absorption) and acoustic noise caused by rib reflections. Future studies can be done to calibrate for attenuation of the ANF signal caused by tissue absorption coefficient.

More patient studies need to be performed to establish the clinical acceptability of this method. The current study considers data from a small number of patients. Further, evaluation of the Doppler method simultaneously with Fick is required to establish the validity of the Doppler approach more precisely. In view of the many difficulties that have yet to be overcome, the technique provided reasonable estimates of cardiac output in 11 patients and showed promise for the noninvasive measurement of cardiac output. Studies in locations other than the aortic outflow tract and further refinements of the DPA technique will certainly provide more accurate estimates of cardiac output.

References

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P A Magnin, J A Stewart, S Myers, O von Ramm and J A Kisslo

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