The Effect of Variations on Pulsed Doppler Sampling Site on Calculation of Cardiac Output: An Experimental Study in Open-chest Dogs

DANIEL C. FISHER, M.D., DAVID J. SAHN, M.D., MARK J. FRIEDMAN, M.D., DOUGLAS LARSON, M.S., LILLIAM M. VALDES-CRUZ, M.D., SUZANA HOROWITZ, B.S., STANLEY J. GOLDBERG, M.D., AND HUGH D. ALLEN, M.D.

SUMMARY We measured aortic flow by two-dimensional Doppler echocardiography in an open-chest dog model to examine how variations in Doppler sample volume length and position influence aortic hemodynamic flow calculations. Fourteen dogs underwent right-heart bypass, in which venous return from the vena cavae drained by gravity to a reservoir. A variable-speed roller pump returned the blood to the pulmonary artery, fixing left-sided cardiac input and output.

Echo Doppler measurements were performed using a 3.5-MHz transducer placed directly on the aortic arch to determine internal aortic cross-sectional area. The transducer was then directed to image the aortic arch for Doppler velocity measurements and the various sampling sites were investigated. Doppler cardiac output could then be determined for each of the various sample volumes over a range of known roller pump settings.

Doppler velocity was analyzed using fast Fourier transform spectral analysis. Mean velocity over the cardiac cycle was obtained by planimetry of the area under the Doppler velocity curve with a minicomputer. Doppler-derived determinations of cardiac output achieved a correlation of \( r = 0.98-0.99 \) to values obtained by the roller pump over a range of cardiac outputs from 0.75–5 l/min. The standard error of the estimate was 0.2 l/min. In this laminar flow model, there was no difference between the predictive accuracy of any of the sampling sites over the range of roller pump flows.

Our study shows that Doppler velocity measurements can be used to quantify aortic flow over a clinically useful range and that variations of sample length and position did not produce significant differences in calculated flows.

SEVERAL studies suggest that Doppler echocardiography can be used to determine cardiac output noninvasively.1,5 Recently, pulsed Doppler two-dimensional echo scanners have been available with quantitative Doppler outputs. The range gate or sample volume obtained in pulsed Doppler echocardiography is a region determined in depth and in location from within a two-dimensional scan over which information is processed for the Doppler shift during the return cycle of an echo pulse. In these new scanners, the Doppler information is calculated by Fourier transform spectral analysis performed rapidly using digital electronics, which provides a linear and quantitative method for determining the Doppler shifts present within the sample volume. An important question for the application of pulsed Doppler technology to flow characterization concerns how sample volume size and location influence the recorded velocities. This question has assumed greater importance because many hemodynamic predictions are being made on the basis of noninvasively derived Doppler flow information from the ascending aorta and other locations.5,4,6,7 In this study, we examined aortic pulsatile flow in an open-chest dog model using Doppler echocardiography while varying sample volume depth and size.

Methods

Surgical Techniques and Animal Model

Fourteen mongrel dogs that weighed 20–25 kg were anesthetized with pentobarbital. They were intubated and ventilated after which a midline sternotomy was performed. The aorta and its branch vessel were dis-
sected and cleaned of fat and adventitia. The azygos
vein was ligated and the pericardium opened. Stab
incisions were made in the lateral right atrial wall and
the right atrial appendage. Retrograde cannulation of
the venae cavae was accomplished with ¼-inch tubing
fastened to the atrial wall by pursestring sutures. This
tubing drained by gravity to a 5-liter reservoir from
which a ¼-inch tube passed through a mechanical
roller pump back to the right atrium. The return tube
from the pump was sutured to the right atrial free wall
and passed through the tricuspid valve into the pulmo-

ary artery. Heparin (5000 U) was given in a bolus and
the reservoir primed with 1 liter of Ringer’s lactate.
Ligatures around the venae cavae and the pulmonary
artery were tightened and right-heart bypass was insti-
tuted. The roller pump was calibrated by emptying the
reservoir into a graduated cylinder and could then be
set to achieve and maintain left-heart cardiac output
within strict limits. When venous return did not keep
up with forward flow, Ringer’s lactate was added to the
reservoir. Cardiac output, 0.74–5/min, was varied in
increments of 0.5 l/min. Additional measurements of
aortic flow were also available from precalibrated 16-
or 18-mm-diameter electromagnetic flow probes
(Gould-Statham SP2204) placed 2 cm distal to the
aortic valve. These measurements were often available
(n = 30) simultaneously with the roller pump mea-
surements as a further cross-check for both. To com-
pare descending aortic flow with right-heart roller
pump volumes, the aortic branch vessels were clamped
during the flow measurements, but released afterward
to minimize cerebral and forequarter necrosis.

Ultrasound and Doppler Method

A prototype two-dimensional sector scanner with
range-gated Doppler (Electronics for Medicine/Hon-
eywell Co.) was used for imaging and Doppler studies.
The scanner has a 3.5-MHz transducer that is oscillat-
ed mechanically through an angle of 60–75°. The
transducer can be stopped along any line of the sector
to sample Doppler signals along that direction. The
region from which the Doppler signal is obtained (sam-
ple volume) can be selected to have an axial length
from 2 mm to 2 cm, and the depth of the sample
volume is variable to a maximum of 16 cm from the
transducer. Doppler shift is detected only from region
specified by the sample volume. Flow velocities are
analyzed within the instrument using fast Fourier
transform spectral analysis to provide quantitative ve-
locity determination. Doppler signals were sampled at
a rate of 19,500 samples/sec at 0–4 cm, 9750 samples/
sec at 4–8 cm, and 6500 samples/sec at 8 cm. The
instrument can be switched quickly from Doppler sam-
pling to real-time imaging to check sampling position.

Two-dimensional images of the aortic arch were
obtained by holding the transducer directly on the aorta
(fig. 1). Once adequate images were obtained and a
sample volume placed, a movable vector cursor was
positioned in the aorta on the sample volume along the
visually estimated direction of flow. This angle (vec-
tor) cursor was recorded on stop frames so that the
angle between visually estimated direction of flow and
the direction of the interrogating beam from the trans-
ducer could be determined, either by the system or
manually with a protractor. Correction for the sam-
ping angle was applied during minicomputer analysis
of the recorded Doppler wave forms, since the velocity
output of the system is uncorrected for angle. The
angle of sampling relative to the direction of flow
across the azimuthal plane (the plane at right angles to
the plane of imaging) could not be estimated. A two-
dimensional stop-frame image of the arch was ob-
tained in systole and the internal diameter of the arch
was measured directly from the image to serve as the basis
for calculating the internal aortic cross-sectional area
(A = πD/2)². Because the system is not optimized for
near-field imaging and with the transducer placed on
the aorta, artifacts from the transducer and oil-path
reverberation from within the scan head sometimes
made imaging of the internal diameter of the aorta
difficult, especially for the superior aortic wall. There-
fore, these vessel areas were occasionally checked
against external aortic circumference with direct mea-
surements obtained using a string and aortic wall thick-
ness determined using a nonbeveled needle. Visualiza-
tion of the descending aorta was achieved by directing
the sector posteriorly from its position on the arch.

Variations of Sample Volume Size and Position

Doppler sampling was performed over three sample
volume areas, each with an axial length of 0.5 cm and
one sample volume combining the three smaller
lengths for both the ascending and descending aorta
(fig. 2). These sizes were chosen on the basis of anat-
omic size in a 25-kg dog aorta to obtain three adjacent
samples that did not overlap or extend past the level
of the aortic valve or superior aspect of the arch. More-
over, we had to leave some flow unsampled immedia-
tely near the valve and in the superior arch to avoid
the perivalvular vortex area and to preclude the possi-
bility of wall motion artifact from positional changes
of the arch.

The lateral beam characteristics estimated in a water
bath for this system suggest that a 6-db Doppler sample
volume is ± 2 mm wide at 2–4 cm depth. Thus,
position changes in the axial plane were used to bring
the sample volume closer to the center stream rather
than attempts at lateral steering. The ECG and Doppler
flow were recorded simultaneously on LS-8 hard copy
at a speed of 100 mm/sec after 5 minutes of stabiliza-
tion after changes in roller pump speeds.

Data Analysis

Doppler velocity flow wave forms from three se-
quential beats were digitized using a programmable
graphics analyzer (Numonics) to obtain a mean veloc-
ity of flow for the entire cardiac cycle. Although the
scanner provides spectral analysis for quantitating all
the velocities within the sample volume, its printout
records the number of times it senses a particular ve-
locity through the darkness of the lines during any 5-
msec sampling period. At any moment, we could de-
termine a modal velocity on the printout with a line
drawn to lie between the dark modal velocities and the
less frequently received lighter velocities, but closer to the darker areas. Once this velocity line was obtained, the area under the curve was digitized over time to provide a mean velocity throughout the cardiac cycle. All records were analyzed by one of us, who was blinded with regard to the cardiac output at the time of sampling and were subsequently reviewed by another blinded observer.

A range-gated, Doppler-derived stroke volume (RGD-SV) was determined using the formula

$$\frac{V \times CSA \times RR}{\cos \theta}$$

where $V =$ mean velocity (cm/sec) calculated by planimetry of the velocity curves over the full cardiac cycle before correcting for angle, $RR$ interval = time between each QRS complex (seconds) and $\cos \theta =$ the angle between the Doppler beam and the blood flow. Cross-sectional area of the aorta was measured in cm$^2$, and sample angle was obtained from gated stop-frame page printouts at each cardiac output.

**Statistical Analysis**

Doppler cardiac output derived from the mean velocity from the eight sample sites were compared with roller pump cardiac output by linear regression. When electromagnetic flowmeter and roller pump values were simultaneously obtained, the roller pump value was used for statistical analysis. Differences in mean velocity between the sample sites were analyzed using univariate analysis of variance.

**FIGURE 1.** (A) Ascending aorta. (B) Descending aorta. Doppler sampling sites (top panels) and representative flow records (bottom panels) are shown. The dotted lines on the flow curves show how they were traced. $\theta =$ angle between sampling vector and estimated direction of flow; TAA = thoracic ascending aorta; RPA = right pulmonary artery.
Results

One hundred forty-two Doppler velocity samples from the different sampling sites were obtained from 11 dogs. Not all sites were sampled at every flow value in every dog. The range of flows and the number of samples is shown in Table 1. Doppler signals were sampled from the ascending aorta for 95 determinations and from the descending aorta for 47. Roller pump values ranged from 0.75 to 5.0 l/min. The roller pump and electromagnetic flow values were always within 5% of each other. Angles of Doppler interrogation to flow were 0–30° (Figs. 1 and 2) with a mean of 15°, the larger angles obtained from the descending aorta and the smaller angles from the ascending aorta. Visual estimation of flow direction and sampling angle is difficult in this portion of the aortic arch; nonetheless, since the correction for angle is a cosine function, it is a minimal source of error of angles of less than 20°. Doppler mean flow velocities corrected for sampling angle (V/cos θ) ranged from 4.2 to 27.7 cm/sec. Angle-corrected peak flow velocity (that is, peak flow velocity/cos θ) ranged from 21.3 to 160 cm/sec. Doppler-calculated stroke volume ranged from 5.9 to 35 ml/beat and Doppler cardiac output ranged from 0.75 to 5.1 l/min.

Location and Size of Sample Volume

None of the eight locations (four in the ascending aorta and four in the descending aorta) differed statistically in corrected mean velocity (V/cos θ) at identical cardiac outputs in the same animal; neither did the derived Doppler cardiac outputs based upon the mean velocity from eight sample sites differ significantly. Table 1 and figure 3 show that Doppler cardiac output derived from the various sample locations achieved correlations of r = 0.98–0.99 and an r² of 0.96–0.98 to the roller pump values. Because fewer data points were obtained in the descending aorta, and since no statistical difference existed between these sites, data for sites 5–8 were pooled for presentation in Table 1. The standard error of the estimate for calculated Doppler cardiac output compared to roller pump flow in the ascending aorta ranged from 0.184 to 0.226 l/min. Bias was analyzed for the lower (0.75–2.5 l/min) and higher (2.5–5.0 l/min) ranges of roller pump output to determine if the technique was more or less accurate over a particular range. The mean percent error in the lower range was consistently at least twice the mean percent error in the higher range, suggesting that no range bias exists. This lack of range bias was statistically confirmed by finding no correlation between either the absolute or percent difference of roller pump and Doppler using univariate analysis of variance.

Mean Aortic Cross-sectional Area

Within each dog, cardiac output and aortic diameter (area) varied directly, but the correlations were not of predictive magnitude (r = 0.57–0.78).

Spectral Dispersion

No significant differences in spectral frequency dispersion between the eight sample sites was observed in the model. All curves showed less than 10 cm/sec spectral width at all locations and outputs during systole. This narrow spectral width allowed excellent reproducibility of determinations of mean velocity by two observers, who separately traced 25 velocity curves and never differed by more than 8% for mean velocity determinations.

Discussion

In previous studies of flow quantitation using Doppler techniques, aortic flow has been examined using pulsed Doppler ultrasound analyzed by the time-interval histogram.¹,² Unlike time-interval histogram, which has a nonlinear response to spectral broadening and does not weigh the various amplitudes of the
Doppler back-scattered frequencies present within the Doppler sample volume, fast Fourier transform spectral analysis in the newest Doppler devices records the frequency using a gray-scale display. In the fast Fourier transform output, the more often-occurring frequencies are assigned a darker gray-scale representation and the less-often-occurring frequencies are assigned a lighter gray-scale representation. Although the velocity curves are less clean than an electronically calculated mean velocity, it is possible from the gray scale display with a Fourier transform to obtain information about the mean and modal velocities. The actual velocities measured may or may not be representative of the flow cross section; the effect of sampling site has been largely unexplored. Doppler sample volume size has both lateral and axial determinants. The lateral component is determined by crystal size, focus, depth from the transducer and tissue refraction. Axial determinants are a function of sample volume size, interrogation frequency and ultrasound pulse and range gate duration. The latter variable may be user-altered in many instruments.

Results obtained from this study suggest that axial sample volume size and location as defined in this protocol do not significantly affect Doppler hemodynamic predictions. Detailed flow profiles of the aorta as a function of time were not obtained in this study; rather, flow was examined over the entire cardiac cycle. Differences in the timing of a flow and skewing of velocity profiles exist at different sites in the aorta. When the velocities over an entire cardiac cycle are digitized, however, it is not surprising that regional and temporal skewing of flow becomes less obvious. Our data indirectly support the lack of importance of temporal skewing in that mean velocity during the cardiac cycle did not differ between sampling sites.

Due to lateral beam characteristics and anatomic constraints in open-chest dog flow recording, the sample sites did not precisely define cylindrical laminae or flow streamlines within the aorta (center stream vs...
adjacent to walls). The most peripheral sample sites (4 and 6), for instance, necessarily included some of the flow within the central stream because the lateral beam profile would predict a small amount of overlap. For the most part, however, the sample sites were distinct from each other. Nonetheless, our study supports the concept that center stream flow velocities integrated over the imaged cross-sectional area accurately predicts volume flow. These findings are consistent with the work of both Seed and Wood and McDonald. McDonald found that pulsatile flow in large arteries displayed fairly flat profiles, unlike the more clearly parabolic flow profiles seen in smaller vessels, in which viscous forces predominated. By injecting dye and recording at high frame speeds, he showed that a true parabolic profile was not formed in the aorta at any time. The fact that the center flow core moves as a mass helps to explain the lack of difference between the sample sites in this study. In the present study, we did not attempt to sample from the most peripheral portions of the aorta, where the viscous drag forces predominated; the sample volumes remained mostly within the central 80% of the aorta.

Seed and Wood investigated velocity distributions in the dog aorta using hot-film constant temperature probes. The hot-film technique works on the principle that the rate of heat transfer from a small metallic element is measured electrically and is related to the fluid velocity. Modifications of this technique permit not only flow velocity determination at various points within the aorta, but also the direction of flow to be measured. They found that at a site near the aortic valve (3 cm proximal to the brachiocephalic artery), the velocity profiles within the proximal aorta were fairly uniform during early systole, but during late systole velocity was greater near the anterior aortic wall. As the blood traveled up the aorta, higher velocities were measured posteriorly. They concluded that in the ascending aorta, a center line velocity measurement gives a good approximation to the cross-sectional mean in the core despite small differences in regional velocity flow changes. Moreover, they showed that velocity profiles at all levels in the aorta were, for the most part, blunt.

Peronneau examined instantaneous flow profiles in the aorta of an open-chest dog model using Doppler echocardiography analyzed by time-interval holography. He examined specific time periods at 140 msec and 300 msec into a 400-msec cardiac cycle and confirmed that velocity profiles varied as a function of aortic location and timing within the cardiac cycle. This confirmed the capability of pulsed Doppler for profiling instantaneous blood flow within the aorta, but did not consider mean velocity over time for various sample sites.

We did not measure flow at the level at the aortic valve, for which profiles are not established. The visual analysis of spectral dispersion of the sample site closest to the valve did not indicate a different velocity or greater turbulence than in more distal sample sites. There is a vortex shed phenomenon reported to occur over a variable length in the perivalvular inlet area in which turbulence is produced in early systole, usually persisting less than 50 msec. This effect was not noticeable in the sample closest to the aortic valve.

The ranges of cardiac outputs achieved by our model simulates those seen clinically. Higher values could not be obtained using our roller pump technique without precipitating acute left ventricular failure. To determine both how well Doppler predicts cardiac output and how sample size affects flow prediction for the higher range of cardiac outputs, we used isoproterenol stimulation with thermodilution. Preliminary results (n = 10) for thermodilution cardiac output values of 5.0–8.6 l/min correlate quite well (r = 0.94) with Doppler cardiac output. No differences between sample sites are apparent at these higher outputs.

Previous studies in humans and in open-chest animals have found that the greatest source of error in Doppler flow calculation is the determination of vessel dimension by M-mode or two-dimensional measurement. In our study, it was not always obvious, especially in the near field, which echoes were the true aortic inner dimension. This was especially true for the aortic wall just under the transducer, where transducer artifact and reverberations often made the endothelium difficult to recognize within the echo stop frame (fig. 1). This was even more marked in our open-chest images with the transducer placed directly on the aorta because the system is not optimized for near-field imaging. Nonetheless, our recent clinical experience suggests that distortion in imaging the size of vessels does affect the accuracy of flow determinations. Our method of determining the internal circumference by string and nonbeveled needle for each roller pump setting allowed us to find the correct and corresponding echo dimension on the images. This is a significant factor in accounting for our excellent correlation with flow measurements by roller pump. Checking dimensional measurements under clinical conditions is not possible if there is distortion within the ultrasound image. Our experience in this regard suggested that while a clinically acceptable range of accuracy for Doppler flow determinations in children could be obtained for Doppler aortic flow (r = 0.91) compared with indicator-dilution curves, the standard error of the estimate for this regression was reduced by 40% if the angiographic dimension of the aorta was used instead of the echo dimension. Use of methods that provide direct cross-sectional measurement of the flow area that are not diameter-dependent and not raised to the second power in calculation of flow should lessen the effect of these difficulties.

Our study confirms the ability of Doppler echocardiography to determine cardiac output by measuring aortic flow velocities. Previous studies have used thermodilution and indicator dilution techniques as the standard against which Doppler has been assessed. Under the best of circumstances, thermodilution techniques are accurate within 5–10%. By developing the roller pump technique as a standard, both absolute cardiac output prediction and possible differences in
regional aortic velocity flow differences could be explored. Our study suggests that a centrally placed sample volume in the ascending aorta 0.5 cm long for children or -1.5 cm for adults should allow adequate Doppler sampling. Used in conjunction with Fourier transform spectral analysis and aortic measurement by echo, this technique should allow a reasonably accurate noninvasive determination of cardiac output in humans.

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