

Cardiac Output Monitoring by Echocardiography: Should We Pass on Swan-Ganz Catheters?

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Transesophageal echocardiography offers a noninvasive technique for the continuous monitoring of cardiac performance. The combination of 2-dimensional echocardiography and Doppler velocitometry provide assessment of cardiac anatomy, valve function and, ventricular loading conditions. Although transesophageal echocardiography has become accepted for perioperative monitoring, it is typically used in conjunction with Swan-Ganz catheterization. To supplant Swan-Ganz catheters, an echocardiographic technique to monitor cardiac output is necessary. Despite considerable effort to achieve this goal, a satisfactory technique has been difficult to realize. This paper discusses the role of cardiac output monitoring in perioperative care and critically examines echocardiographic techniques for cardiac output monitoring.

Swan-Ganz (S-G)^b catheterization provides a sophisticated assessment of cardiac performance not previously attainable in the clinical setting. Measurements of ventricular preload, mixed venous oxygen content, and CO have proven useful for diagnosis and to guide therapy. Although noninvasive monitoring modalities have experienced startling growth over the past decade, S-G catheters remain widely used in the operating room and in critical care settings.

Despite the utility of S-G catheter monitoring, the invasive nature of the technique, as well as recent concerns of the validity of derived measurements, has fostered dissatisfaction with the technique [1]. Owing to the inherent hazards, the technique requires trained personnel, is expensive, and thus can be justified in only a select group of critically ill patients. The validity of measurements derived from S-G catheter data in many settings has been questioned. Two-dimensional echocardiographic (2-D echo) studies have shown pulmonary capillary wedge pressure to be an inaccurate guide to left ventricular preload [2, 3, 4, 5]. As for monitoring myocardial ischemia, van Daele et al. [6] among others have shown that S-G catheter-based indicators, such as acute increases in pulmonary artery wedge pressures, are neither sensitive nor specific. Swan-Ganz measures of CO using the thermodilution technique have an established variability of 10 to 20% and can provide only intermittent assessment of cardiac function [7]. The recent appreciation of these shortcomings has pushed development of alternative approaches to cardiac assessment. While diverse techniques, which include transthoracic bioimpedance, arterial pulse contour, and electrocardiographic R-T ratios, have been examined, it is echocardiography which appears best suited to supplant the method of Swan and Ganz.

Echocardiography can provide noninvasive assessment of cardiac anatomy, valve function, ventricular preload, and blood flow. This has led to extensive use of the technique by cardiologists. Advances in instrumentation have enabled the development of esophageal probes, which contain both 2-D echo and Doppler capabilities. Transesophageal echocardiography has become accepted for monitoring cardiac function

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^bAbbreviations used: S-G, Swan-Ganz; CO, cardiac output; CSA, cross-sectional area; 2-D echo; two-dimensional echocardiograph.

in the operating room, usually in conjunction with S-G catheters. To replace S-G catheter monitoring, however, an echocardiographic technique to monitor CO is necessary. Despite considerable effort towards this goal, it has been difficult to realize. This paper will briefly discuss the need for direct measures of CO in perioperative care and examine the various echocardiographic approaches to noninvasive CO monitoring.

THE CASE FOR CO MONITORING

The importance of determination of CO to manage hemodynamic status is well recognized. Cardiac output reflects upon the heart's performance as a pulsatile pump as modified by vascular tone. Tissue oxygen delivery is the product of CO and arterial blood oxygen content. The clinical utility of CO monitoring is well established. Even experienced clinicians have difficulty in assessing CO. Connors et al. [8] showed that in an intensive care setting a physician's clinical judgement is inadequate to differentiate between high and low CO states. Eisenberg et al. [9] confirmed these results and found that, in the majority of patients, planned therapy was altered by the information obtained from S-G catheterization (Figure 1).

Additional studies have examined the role of CO output monitoring to aid diagnosis and guide therapeutic interventions. A study of vascular surgery patients showed that pre-operative maximization of cardiac performance through CO monitoring resulted in significant reductions in postoperative cardiac complications [10]. Rao et al. [11] established that sophisticated cardiac assessment, including CO monitoring, was valuable in the management of surgical patients who had suffered a recent myocardial infarction. Their data shows such monitoring to reduce the incidence, morbidity, and mortality rate of perioperative reinfarction. When optimization of CO and oxygen delivery was set as the therapeutic goal, a prospective study demonstrated that outcome improved in critically ill patients [12]. Thus, development of a continuous, noninvasive technique for assessment of CO would be of considerable clinical significance.

PRINCIPLES OF DOPPLER ULTRASOUND

Ultrasound has rapidly advanced to the forefront of diagnostic imaging. It is capable of both real time imaging of cardiac structures and measuring blood flow within the heart

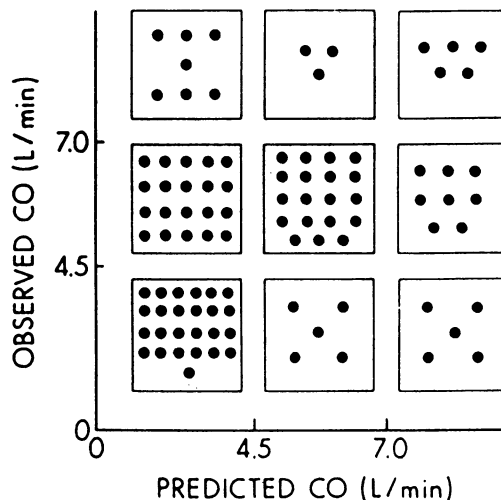


Figure 1. Discrepancies between a clinician's prediction of CO and that measured by thermodilution.

and great vessels. Consisting of high frequency sound waves, ultrasound easily penetrates skin and other body tissues, making it suitable for noninvasive applications. A fraction of an emitted ultrasound signal is reflected as it encounters tissues of differing acoustic density. The frequency, time delay, and amplitude of the ultrasonic backscatter or "echoes" can be interpreted to create a 2-D image of the tissue or, using Doppler techniques, can estimate blood flow velocity.

Clinical devices use sound wave frequencies of 1 to 10 million Hz to achieve an axial resolution of less than one millimeter. A tradeoff between tissue penetration and imaging resolution dictates choice of signal frequency. High frequency signals provide superior image resolution but penetrate only short distances. For example, a 5 MHz signal provides an axial resolution of 0.5 mm but is limited to distances of 30 cm. When the power emission is kept below 100 mWatts/cm, ultrasound has not demonstrated any injurious effects and is approved for fetal imaging [13].

Blood flow velocity measurements using ultrasound are based on the Doppler principle. When ultrasound is reflected from a stationary object, the reflected signal is of the same frequency as that transmitted. Objects moving towards the transmitted signal cause reflections of higher frequency. This alteration in frequency, the Doppler shift, is described by the equation:

$$\Delta F = 2 f_t v \frac{\cos\theta}{s}$$

where ΔF = Doppler shift, f_t = transmitted frequency, v = velocity, s = speed of ultrasound in tissue, θ = the angle of incidence between the ultrasound beam and the direction of the object's motion. The equation can be rearranged to solve for velocity:

$$v = \frac{2 \Delta F}{f_t \cos\theta}$$

Thus, the velocity of an object is related to two variables, the Doppler frequency shift and the angle of incidence between the direction of the object's motion and the transmitted ultrasound signal (Figure 2).

The measurement of blood flow with Doppler ultrasound takes advantage of the large ultrasound reflections caused by red blood cells. The Doppler shift from these reflections can be used to quantitate blood flow velocity. To determine volumetric flow (L/min^{-1}), however, these velocities are multiplied by the cross-sectional area (CSA) of the flow channel, that is:

$$Flow = (CSA) \int v dt$$

The accuracy of the Doppler technique is related to both velocity and area measurements.

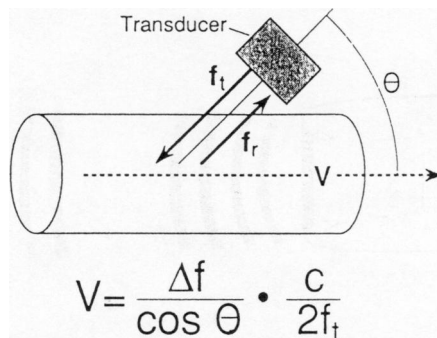


Figure 2. Doppler technique for determination of flow velocity.

DOPPLER INSTRUMENTATION

The technique for Doppler CO measurement requires an ultrasound transducer to transmit and receive the ultrasound signal and a computer to convert the Doppler frequency shift to flow velocities and CO. Piezoelectric crystals which mechanically oscillate when a voltage is applied generate the ultrasound signal. Conversely, when a reflected ultrasound signal impacts upon these crystals, an electric potential is generated. Two types of Doppler transducers, continuous wave and pulsed wave, are widely employed for cardiac monitoring. Continuous wave Doppler instruments employ two crystals, one which continuously transmits an ultrasound signal and the other which acts as a receiver for reflected signals. The pulsed-wave system employs a single crystal, which acts as both the ultrasound transmitter and receiver. The crystal emits short bursts of ultrasound at regular intervals and pauses between bursts to receive the reflected signal (Figure 3.). The frequency of the pulsed ultrasound emissions is called the pulse repetition frequency. As the speed of ultrasound propagation through tissue is fairly constant (1540 cm/sec),

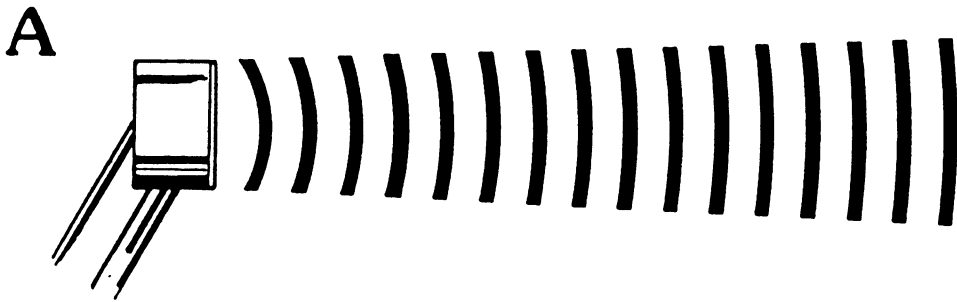


Figure 3a. Continuous wave Doppler transducer.

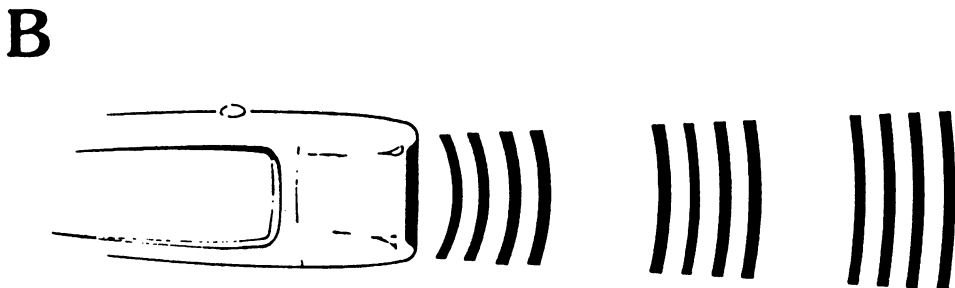


Figure 3b. Pulsed wave Doppler transducer.

the delay between transmission and reception of the reflected signal is dictated by the distance of the reflector from the transducer. By regulating the waiting period from signal transmission to reception, the pulsed wave transducer can select the exact distance from which the reflected signals originated. This capacity, called "range gating", allows the operator to sample blood flow from a specific location in a cardiac chamber or major blood vessel and is a major advantage to pulsed-wave Doppler. The disadvantage of this approach is that the pulsed technique limits the maximum velocities that can be measured to one half the pulse repetition frequency. This maximum frequency shift is known as the Nyquist limit. At frequencies above the Nyquist limit the returning signal becomes distorted ("aliasing"), making the velocity indeterminable. As the depth of the sample volume increases so does the pulse repetition frequency and, thus, the maximum velocities which can be measured becomes limited. A comparison of the advantages of pulsed wave versus continuous wave Doppler measurement techniques is presented in Table 1.

Both pulsed wave and continuous wave Doppler can be used in conjunction with 2-D echo so that the Doppler signal can be directed precisely (Figure 4). This allows the operator to place the beam parallel to blood flow, and, with the pulsed wave technique, permits the selection of a sample volume within the aorta or valve orifice.

SOURCES OF ERROR IN DOPPLER CO DETERMINATION

Although the Doppler approach is theoretically sound and has been validated in experimental flow systems, several potential sources of error become apparent when the technique is applied to the clinical setting [14, 15]. Inaccuracies in the measurement of velocity and CSA can result in erroneous flow measurements.

Errors with cross sectional area determination. Determination of cross sectional area is most commonly derived from equations based on measurements of valve or vessel diameter. This requires an assumption of the geometric shape of the flow chamber. For the aortic and pulmonary valve area measurements, the use of a circular πr^2 model appears acceptable. However, the geometry of the mitral and tricuspid valves would be best modeled as an ellipsoid and, thus, require orthogonal 2-D echo images for CSA estimation. In circular or ellipsoid models, accuracy in diameter measurements are critical because CSA is an exponential product of diameter. For example, a 2 mm error in measurement of a 25 mm aorta results in a 17% error in CSA. To obviate the need for an echocardiographic measurement of aortic CSA, some devices employ a nomogram to estimate aortic CSA based on the patient's height, weight, and age. While such nomograms may correlate well in large population studies, they may be invalid for any given patient.

The second weakness of using fixed diameter measurements for calculating CSA is that the method does not account for variations in vessel size during the cardiac cycle and with changes in pressure. This would be a only a minor source of error for aortic mea-

Feature	Continuous-Wave	Pulsed-Wave
Quantitatively Measure High Velocities	Yes	No
Select Specific Sample Volume	No	Yes
Use in Combination with 2-D Imaging	Yes	Yes

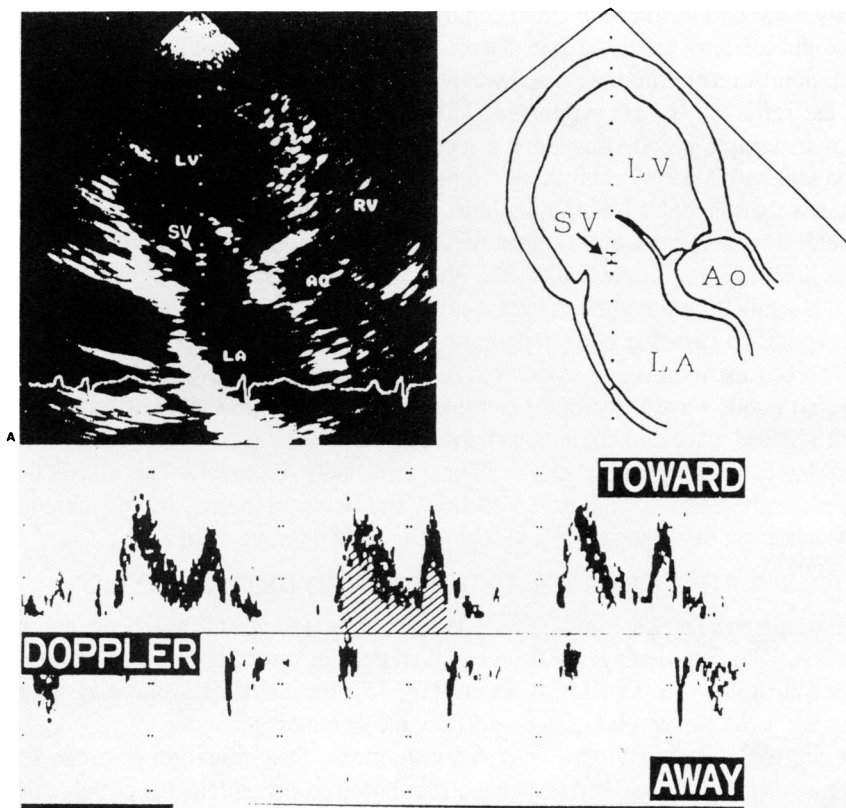


Figure 4. 2-D echo used to position the sampling volume of a pulsed wave Doppler signal at the mitral valve annulus. Derived velocity tracings are shown below.

surements, where the systolic variation is small but can be significant for measurements of the pulmonary artery and mitral valve. The large variations in pulmonary artery size (up to 20%) during systole may account for the poor correlations of Doppler CO measurements at this site [16].

An alternative approach to CSA estimation is to use 2-D echo along a short axis plane. The utility of this approach is hampered by the technical difficulty in obtaining a short axis view and errors related to the imprecision of plane positioning. Also, a short axis view lies perpendicular to the plane required for optimal Doppler measurements. Recent advances in transducer technology have resulted in biplane and multiplane transesophageal transducers which may permit more precise beam orientation and greater accuracy in CSA estimation.

Errors in velocity determination. Continuous wave Doppler measurements estimate velocities along the entire path of the ultrasound beam, and require the assumption that the maximal velocities registered equate with the mean blood velocity at the site of CSA estimation. With pulsed wave systems, the velocities can be precisely sampled at the site of CSA determination. Both techniques, however, rely on sample sizes that are a small fraction of the total CSA (typically less than 10%) and, therefore, require extrapolation of the measured velocity to calculate mean flow velocity for the vessel. Such an

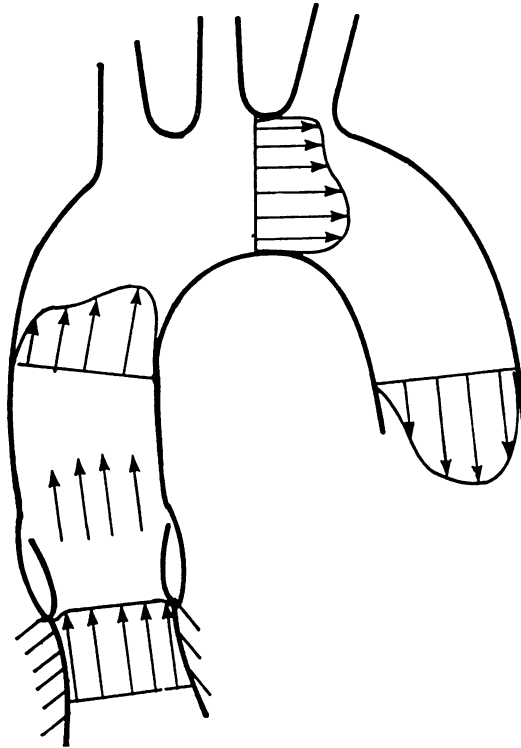


Figure 5. Blood flow velocity profile in the aorta. The blunt profile resulting from the acceleration of blood in the left ventricular outflow tract becomes parabolic and asymmetric in aortic arch.

extrapolation assumes that blood flow velocities are equal across the vessel, i.e., a "blunt" flow profile. A blunt profile in the left ventricular outflow tract and across the semilunar valves results from the acceleration and convergence of blood flow. Turbulent flow, seen in conditions of aortic valvular disease and high output states, disrupts the velocity profile. Alterations in the vessel geometry, such as curvature of the ascending aorta or asymmetry of the outflow tract, will also cause deviations from a blunt profile and erroneous Doppler measurements (Figure 5).

SITES FOR DOPPLER CO MONITORING

Aortic flow estimates of CO. Most work on determination of CO with Doppler ultrasound has focused on aortic velocity measurements. This approach is appealing, in that many of the conditions required for accurate Doppler measurements are present. The aorta is a concentric, low compliance structure. Additionally, blood flow in the left ventricular outflow tract and proximal aorta has been shown to have a blunt flow pattern [17, 18].

Doppler signals derived from ascending aortic flow can be reliably obtained from a transcutaneous Doppler probe placed at the suprasternal notch (Figure 6). When combined with a 2-D echo measurement of aortic diameter, CO can be calculated. This approach has been validated by several investigators. However, manipulations required to obtain adequate signals are time consuming and the technique provides only intermittent measurements. These limitations have led to the development of alternative techniques

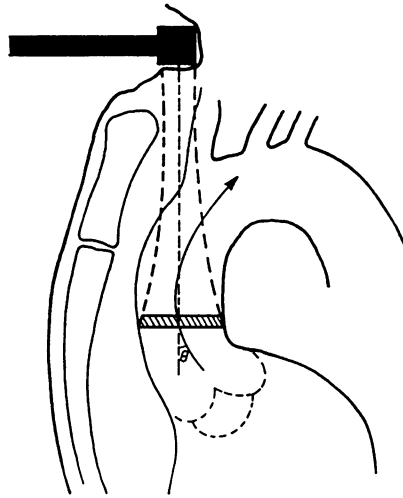


Figure 6. Schematic of the suprasternal window to ascending aortic flow.

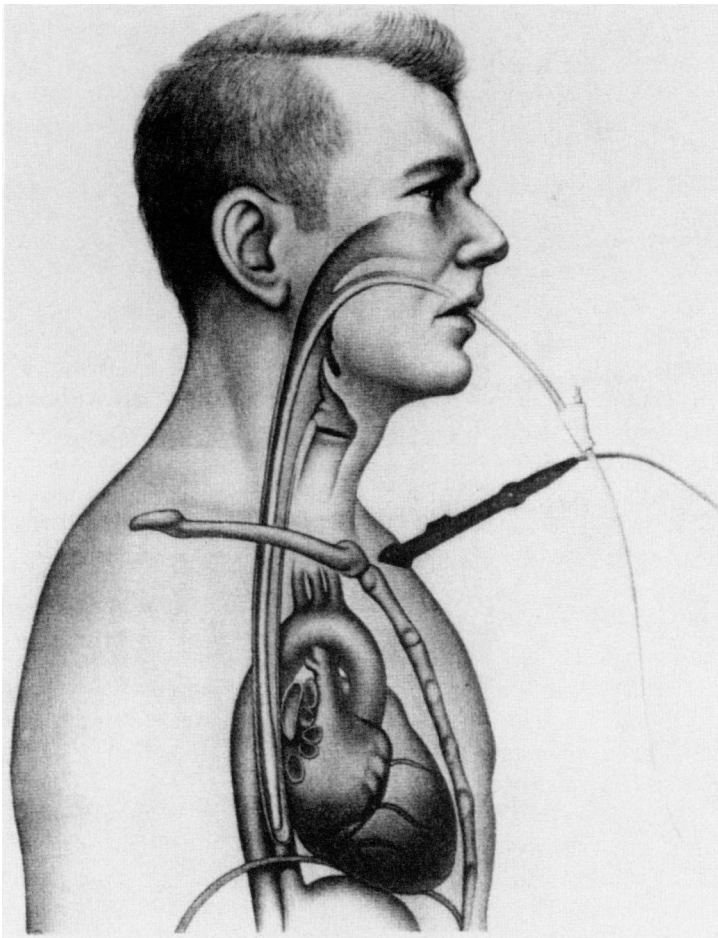


Figure 7. Technique for transesophageal monitoring of descending aortic flow.

which are better suited to monitoring purposes.

Transesophageal echocardiography employs an ultrasound transducer mounted on the tip of a steerable esophageal probe. The heart lies anterior to the esophagus and, thus, the transesophageal approach avoids interference from ribs and lung tissue which hampers transthoracic imaging. The close proximity of the esophagus to the cardiac structures enables high resolution images to be obtained. An acoustic window for imaging the descending aorta can be reliably obtained by positioning the probe to a depth of approximately 30 cm (Figure 7). With the transesophageal approach, the esophagus stabilizes the probe position allowing continuous monitoring. Moreover, obtaining optimal Doppler signals requires only rotation of the probe and adjustments in the depth of insertion.

Determination of transesophageal Doppler estimates of CO using descending aortic flow measurements requires a three step process. First, aortic CSA is estimated by nomogram or a 2-D echo measurement. Second, the esophageal probe is positioned so as to maximize descending aortic flow signals. Third, the descending aortic flow signals are calibrated against a measure of ascending aortic flow. Such a calibration is necessary to account for blood flow distributed to the head and upper extremities and for the significant angle of incidence between the Doppler signal and descending aortic flow. The calibration procedure determines the proportionality constant, K-factor, which is the ratio of ascending aortic to descending aortic flow. Once calibrated, the transesophageal Doppler signals continuously monitor CO based on descending aortic flow.

Non-imaging transesophageal Doppler devices became commercially available in the 1980's. Miniaturization of the transducer allowed for an esophageal probe of 12 FR in diameter including a port for stethoscopy. As anesthesiologists routinely place esophageal stethoscopes for monitoring cardiorespiratory function during general anesthesia, transesophageal CO monitoring does not require additional procedures to the patient.

In clinical trials, transesophageal Doppler measurements have given inconsistent results. Initial studies by Mark et al. [19] in patients undergoing aortocoronary bypass surgery, found a poor correlation between Doppler and thermodilution measurements ($r = 0.43$). Their data showed that calibration of the esophageal probe was the major source of

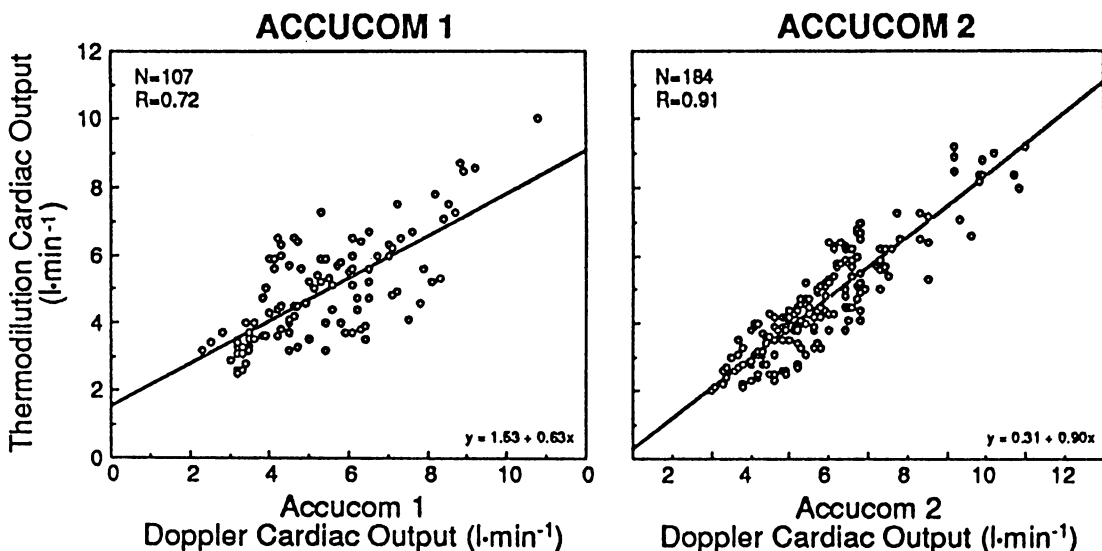


Figure 8. A second generation transesophageal Doppler device (Accucom 2) shows significant improvement in performance as compared to prior technology.

error. Freund [20] reported variable performance in a comparison of Doppler to thermodilution CO in 23 surgical patients. The authors concluded that the operator-dependent nature of the Doppler technique contributed to the inconsistent results. Spahn et al. [21] studied 35 patients and judged that the variability between Doppler and thermodilution measurements (S.E. 1.4 L/min) was unacceptable for clinical monitoring.

Advances in Doppler technology have produced devices which have demonstrated significant improvement in clinical trials. In an intraoperative evaluation of non-cardiac surgery patients, Perrino et al. [22] reported that real-time display of the Doppler velocity wave form and wide-beam esophageal transducers resulted in a marked reduction in the measurement error of the Doppler technique (Figure 8). The improved accuracy of the new generation Doppler devices has been confirmed by Schmid [23].

A source of error specific to the transesophageal approach to determination of CO is the use of the proportionality constant or K-factor. The K-factor is determined during the initial calibration procedure and is usually not adjusted during the operation. Several studies have suggested that the K-factor can fluctuate during surgery as a response to changes in sympathetic tone, arterial blood pressure, and alterations in anesthetic depth [24, 25, 26, 27].

Recently, a transesophageal approach has been described for measurement of ascending aortic flow. This technique uses transesophageal echocardiography to obtain an apical long axis view of the left ventricle and proximal aorta. This view is achieved by

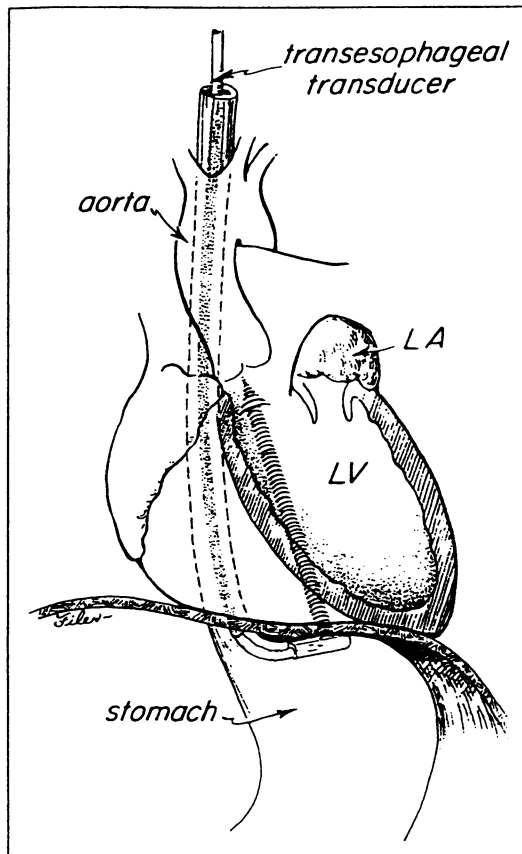


Figure 9. Transesophageal echo probe positioned to interrogate ascending aortic flow.

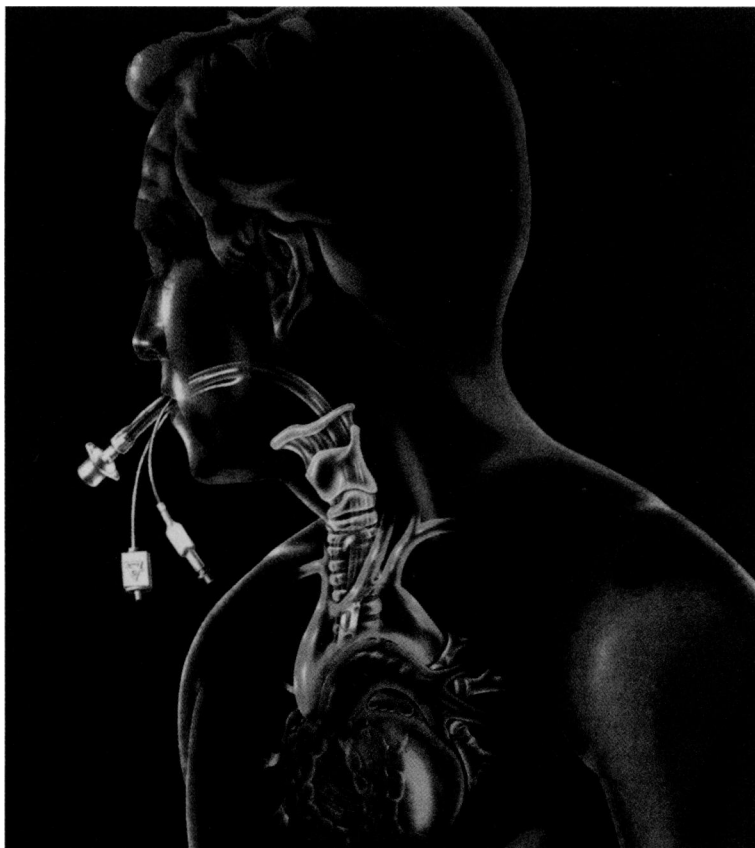


Figure 10. With the ultrasound transducer mounted at the tip of an endotracheal tube, transtracheal Doppler monitors ascending aortic flow.

anteroflexion of the probe after it is positioned in the stomach (Figure 9). After 2-D echo measurement of aortic diameter, a pulsed wave sample volume is placed across the aortic valve using 2-D echo guidance. Katz and colleagues [28] reliably obtained aortic flow signals with this approach and derived CO measurements which correlated with thermodilution measurement. In light of the simplification this technique promises, further evaluation of the approach is warranted.

Transtracheal Doppler is a novel approach for monitoring ascending aortic flow. This technique employs a non-imaging pulsed-wave Doppler crystal mounted to the tip of an endotracheal tube. Positioning at the mid-trachea provides an acoustic window to the ascending aorta (Figure 10). The transtracheal approach has several potential advantages for CO monitoring. By incorporating the Doppler crystal into the endotracheal tube, no additional procedures are required. As opposed to the transesophageal approach, no calibration is needed as ascending aortic flow is directly monitored. Also, the pulsed wave signal is used not only to determine blood flow velocity measurements but also to determine the diameter of the flow channel. Thus, an independent echocardiographic measurement of aortic dimensions is unnecessary when using this technique.

Initial work by Abrams et al. [29] showed that transtracheal Doppler CO measurements were highly correlated with thermodilution measurements. Perrino et al. [30], in a

prospective evaluation of the technique in 27 patients undergoing noncardiac surgery, also demonstrated a good correlation with thermodilution measurement ($r = 0.80$) and found that transtracheal Doppler measurements tracked sequential changes in CO with precision. Transtracheal Doppler performance was found to be dependent on operator experience and the stability of the received Doppler signal. A limitation to the transtracheal approach is that patient positioning and the surgical manipulations associated with open heart surgery significantly increase the error of the technique [31]. Further development of this innovative approach may yield a satisfactory device for clinical use.

Transmitral and transpulmonic flow estimates of CO. Transmitral flow measurement has also been examined as a method for CO determination. Cineangiographic studies by Lynch and Bove [32] showed a flat flow profile at the opening of the mitral valve. Cross sectional area measurement is complicated by the ellipsoid shape of left ventricular inflow tract and the significant variations in mitral valve area during diastole. Fisher et al. [33] described a quantitative technique for CO determination using transmitral flow measurements in an animal model. The flow velocity was sampled at the tips of the mitral valve leaflets and the maximal opening area of the mitral valve seen on a 2-D echo short axis image was corrected for diastolic variations in dimensions M-mode tracings of the leaflets. They accurately estimated CO by this method. Stewart et al. [34] confirmed the need for dynamic assessment of mitral valve area in an experimental preparation. Clinical validation of this approach has followed but difficulties in obtaining a clean, short axis image of the thin pliable mitral leaflets limit its usefulness [35].

The shortcomings of the Fisher method have led to the search for a simplified technique. Meijboom et al. [36] reported a technique based on 2-D echo and Doppler measurements from a single four-chamber view. Their data in both dogs and children showed this method to be on par with the more complex method of Fisher. Holt [37] described a technique using the apical four chamber view where M-mode tracings of the mitral valve tips were used to estimate the mean mitral valve orifice area. They showed that these estimates closely correlated to 2-D echo measures of mitral valve area and that Doppler CO using this approach accurately estimates thermodilution CO. The above studies confirm the validity of the transmitral approach and, by relying on a single imaging plane, greatly simplify the technique.

The success of single plane mitral flow estimates of CO obtained by transthoracic Doppler has led to its application *via* transesophageal echocardiography. An initial report by LaMantia et al. [38] found a strong correlation between Doppler derived CO measured at the mitral valve leaflets and thermodilution. Ryan et al. [39] also found transesophageal Doppler measurements of mitral flow to provide accurate determination of CO. Other reports have not been as encouraging. Muhiudeen et al. [40], in an intraoperative study of transesophageal Doppler, measured mitral CSA at the annulus and velocities from the leaflet tips. Poor correlations with thermodilution were obtained. Further, the error could not be attributed solely to errors in CSA estimation. Shimamoto et al. [41] using both area and velocity measurements obtained at the level of the annulus, also found poor agreement between Doppler and thermodilution CO measurements. Clearly, the utility of mitral flow measurements of CO, as determined by transesophageal Doppler estimations, is not resolved. Accurate measurements appear to require a dynamic assessment of the mitral orifice throughout diastole. The recent development of multiplane echocardiography may also offer improvements in the accuracy of both flow and area measurements.

Doppler measurements of PA blood flow are complicated by systolic variations in vessel CSA and technical imaging difficulties. The transesophageal probe must be ante-flexed at the basal short axis view to obtain pulmonary artery measurements. Muhiudeen

et al. [40] reported that they were unable to obtain either adequate 2-D echocardiography or Doppler signals of the pulmonary artery in 20 of 140 measurements (14%). They found only a modest correlation between Doppler and thermodilution-determined CO estimates ($r = 0.65$). Further, sequential changes in Doppler-defined CO were in the opposite direction to changes in thermodilution measurements in 20 of 65 events (31%). More recently, Savino et al. [42] used transesophageal echocardiography to determine CO in 33 cardiac surgery patients. They found pulmonary artery Doppler-defined estimates to be highly correlated with thermodilution measurements, although they were unable to obtain adequate signals in 24% of patients. This study supports the view that, with careful technique and proper selection of area and velocity measurements, accurate estimates of CO are feasible using data obtained from the pulmonary artery. Its utility for clinical monitoring, however, remains in doubt.

CONCENTRIC BEAM DOPPLER

The clinical application of Doppler instrumentation for CO monitoring has been hindered by errors introduced by the unknown angle of incidence between the ultrasound beam and blood flow. This error is important as it effects both CSA and velocity measurements. Use of concentric beam Doppler measurements represents a novel approach to reducing such errors.

The concentric beam Doppler measurement technique consists of a non-imaging pulsed-wave Doppler system which is capable of transmitting and receiving two parallel channel ultrasound beams (Figure 11). One of the Doppler emissions consists of a wide

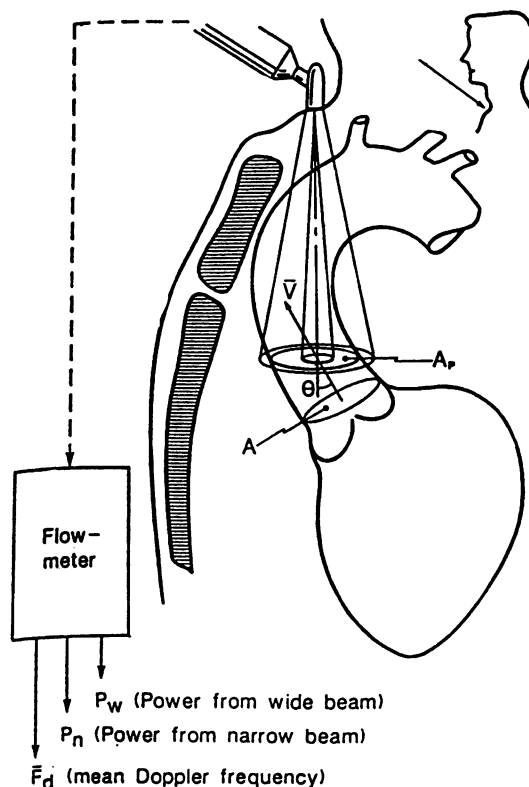


Figure 11. Concentric beam Doppler utilizes information from both a wide and a narrow ultrasonic beam for CO determination.

ultrasound beam (30 mm diameter). This technique thus allows complete interrogation of the ascending aorta. The second, or narrow beam, has a small sample volume within the core of aortic flow. The details of the measurement technique are as follows: Mean aortic blood velocity is determined from the wide beam. The width of this beam allows measurement of the mean velocity of all blood passing through the aorta. Thus, errors associated with extrapolation from a small sample volume of aortic flow are avoided and a blunt flow profile does not have to be assumed.

The CSA of the aorta is determined from the intensity (power return) of the received Doppler signals from the two beams. The intensity of the each received Doppler signal is proportional to the number of red blood cells within the beams path. The power of the wide beam (P_w) equals the projected aortic area (A_p) multiplied by an attenuation constant (K_a) as follows:

$$P_w = A_p K_a$$

The true CSA of the aortic flow channel is the projected area divided by $\cos \theta$. The power return from the narrow beam is equal to the known area of this beam (A_n) multiplied by the same attenuation constant. Therefore, the true CSA of aortic flow can be determined by the ratio of the power returns from the two beams, as follows:

$$A = A_n \cos \theta \frac{P_w}{P_n}$$

Thus, aortic CSA can be determined solely from the Doppler measurements. This obviates the need for 2-D echo instrumentation. Further, estimates are dynamic assessments of the flow channel rather than the anatomic CSA.

Calculation of CO is the product of the velocity and CSA measurements and heart rate:

$$CO = D \frac{\Delta F}{\cos \theta} A_n \frac{P_w}{P_n} HR$$

where D is a precalibrated attenuation constant. Because the cosine function cancels out of the equation, the technique is angle independent, a major advantage.

Initial reports by Taylor and Silke [43] in cardiac catheterization patients and by Looyenga et al. [44] in patients in intensive care units showed the concentric beam Doppler device measurements to correlate well with thermodilution measurements. In contrast, subsequent reports have not been favorable, with standard error of the estimates of approximately 1.5 L/min^{-1} [45, 46, 47]. These errors are clearly unacceptable for a clinical monitor and are larger than those obtained with conventional Doppler measurement devices. The varied results from use of the concentric beam Doppler technique can be attributed to both theoretical and practical considerations. In the clinical setting, the operator cannot be assured that the wide beam encompasses the entire aortic flow channel. The diameter of the beam used in the above studies was 30 mm which approximates the diameter of a normal aorta. However, eccentric placement of the ultrasound beam would lead to incomplete interrogation of aortic flow and, thus, underestimation of CSA. Secondly, the operator cannot be assured that the narrow beam will remain focused at the depth appropriate for aortic flow velocity measurements. This could result in additional inaccuracies in the CSA measurement. Further, most authors have noted technical difficulties in properly positioning the concentric beam device using the suprasternal approach [43, 45]. Taylor and Silke [43] note that proper aiming of the transducer is critical for reliable performance and these authors recommend that performance of 100 examinations is needed for an operator to become proficient with its use. Niclou et al. [45],

however, did not find improvement with experienced operators. Refinements in the operational features of concentric beam devices appear warranted and may resolve the difficulties with probe positioning.

CONCLUSION

Transesophageal echocardiography is uniquely suited to provide continuous, noninvasive assessment of cardiac function. The combination of 2-D echo imaging with Doppler ultrasonography has proven useful in the perioperative period for monitoring left ventricular preload, myocardial ischemia and valvular function. An echocardiographic technique for monitoring CO, however, has been more difficult to realize. Although the soundness of the Doppler technique has been validated, it is in the clinical application where substantial measurement errors have been found to occur. Precise measurement technique is required to avoid errors in both area and velocity measurements. Rapid advances in transducer technology and electronics, including multiplane transducers, automated border detection and concentric beam Doppler, promise further improvements to the technique. Until such improvements are realized, the thermodilution technique remains the standard for perioperative CO monitoring.

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