EXPERIMENTAL STUDIES

Instantaneous and Continuous Cardiac Output Obtained With a Doppler Pulmonary Artery Catheter

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A newly developed, flow-directed, Doppler pulmonary artery catheter, capable of measuring instantaneous and continuous cardiac output, was evaluated in both an in vitro pump model and an animal model. Quantitative flow was calculated with use of the instantaneous, space-average velocity (obtained from the velocity profile) and the instantaneous area (obtained from the vessel diameter) and compared with electromagnetic flow. Additionally, simultaneous thermodilution flow measurements were obtained.

Doppler catheter-determined flow was highly predictive of electromagnetic flow in both continuous and pulsatile pump models \( r^2 = 0.98, \text{m} = 1.04, \text{SEE} = 0.44; \) and \( r^2 = 0.97, \text{m} = 1.04 \) and \( \text{SEE} = 0.33, \) respectively. Thermodilution was less predictive and appeared to underestimate electromagnetic flow in both the continuous and the pulsatile model \( r^2 = 0.99, \text{m} = 0.91, \text{SEE} = 0.20 \) and \( r^2 = 0.95, \text{m} = 0.84 \) and \( \text{SEE} = 0.34, \) respectively.

In the animal model, Doppler catheter-determined cardiac output appeared to modestly underestimate electromagnetic flow \( r^2 = 0.80, \text{m} = 0.79, \text{SEE} = 0.72. \) However, Doppler determinations of flow remained more accurate than did simultaneous thermodilution measurements \( r^2 = 0.73, \text{m} = 0.79, \text{SEE} = 0.72. \)

Accurate, continuous and instantaneous cardiac output measurements appear possible with use of a flow-directed, Doppler pulmonary artery catheter. This catheter system also provides instantaneous diameter measurements and mapping of instantaneous velocity profiles within the main pulmonary artery and may lead to more accurate Doppler-derived assessment of cardiac output in humans.

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Traditional methods used to measure cardiac output in humans include thermodilution, dye dilution, Fick oxygen consumption and radionuclide or radiocontrast angiography. The most commonly utilized method of measuring cardiac output is the thermodilution technique, in which measurements are obtained through the use of a multilumen, Swan-Ganz type catheter with the thermistor located in the pulmonary artery (1-3). The accuracy of these measurements depends on several factors including temperature of the injectate bolus, technique of injection, adequacy of mixing, method of sampling and phase of the respiratory cycle during injection (4-7). The reported accuracy of thermodilution varies widely, ranging from ±3% to 30%, depending on the thermodilution technique used and the method of flow measurement under comparison (4-14). In addition, measurements are performed only on an intermittent basis and continuous or instantaneous measurement of cardiac output is not possible.

More recently, noninvasive methods for measuring cardiac output have been developed that utilize Doppler ultrasound in combination with two-dimensional, A- or M-mode echocardiography (15-19). These noninvasive methods measure maximal blood flow velocity in either the ascending aorta or the pulmonary artery and calculate cardiac output as the product of time-average mean velocity (assuming a flat velocity profile) and estimated cross-sectional area. Several studies (16-19) have documented the correlation of noninvasive, Doppler-determined cardiac output with either thermodilution or Fick cardiac output. User variability, anatomic variation, non-flat velocity profiles and difficulty in obtaining accurate diameter and Doppler measurements combined with an unknown angle of incidence between the ultrasound beam and flow have limited the clinical utility of these noninvasive methods, especially in the critically ill patient (20,21).
Newer approaches to measure cardiac output utilizing ultrasound have included transesophageal Doppler measurements (22–24) and intraoperatively placed Doppler transducers sewn to the ascending aorta (25,26). However, these methods are useful only in intubated patients or patients undergoing open heart surgery, or both. Also, some studies (22,24,25) have demonstrated that the measurements obtained with these methods correlate poorly with thermodilution cardiac output measurements.

We have developed a new method for measuring both instantaneous and mean cardiac output that utilizes Doppler-determined flow in the main pulmonary artery obtained from a catheter-mounted transducer. In addition, simultaneous measurement of right ventricular, pulmonary artery and pulmonary capillary wedge pressure may be obtained. Flow measurements obtained with use of this Doppler catheter were initially compared with timed flow, electromagnetic flow and thermodilution flow measurements in vitro, continuous and pulsatile blood perfusion models. Then, Doppler catheter-determined cardiac output was compared with thermodilution and electromagnetic cardiac output in an animal model.

**Methods**

**Doppler Catheter**

All studies were performed using a newly developed, continuous cardiac output, pulmonary artery catheter (Cardiometrics Inc.) (Fig. 1). This catheter utilized a 10 MHz ultrasound transducer, 1 mm in diameter, located 8 cm proximal to the distal catheter tip. The catheter was preformed into a bent configuration and straightened by an internal guide wire mechanism before insertion. The catheter was inserted into the internal jugular vein through a percutaneous sheath and floated into the pulmonary artery in the usual manner. Transducer location within the main pulmonary artery was confirmed by fluoroscopy and monitoring of pressure waveforms obtained from the transducer lumen and right ventricular lumen. Once the catheter was properly positioned within the main pulmonary artery, the guide wire was retracted and the catheter assumed its preformed shape, placing the ultrasound transducer against the vessel wall (Fig. 2). Angled configuration of the catheter's distal tip along with circular cross-sectional configuration of the pulmonary artery assured aiming of the ultrasound beam through a central vessel diameter.

**Doppler Flow Measurements**

**Instantaneous space-average velocity.** This was obtained with use of a multirange-gate pulsed Doppler velocimeter (DOPCOM Cardiac Output Monitor, Cardiometrics Inc.). This pulsed Doppler velocimeter maps a one-dimensional velocity profile by establishing eight individual range gates spanning the vessel diameter. The sample volume length for each range gate is approximately 1 mm. Instantaneous blood velocity at each of the eight gates was calculated with use of the following relation:

\[
V_i = \frac{F_d \times C}{2 F_0 \times \cos \theta}
\]

for range gate \(i = 1\) through 8. \[1\]

where \(V_i\) = velocity at range gate \(i\), \(F_d\) = Doppler shift frequency, \(F_0\) = transmission frequency, \(\theta\) = the incidence angle between the ultrasound beam and blood flow vector and \(C\) = transmission speed of ultrasound in blood.

Transmission frequency was 10 MHz ± 0.01% and velocity of ultrasound in blood was assumed to be 1,570 cm/s. Incidence angle (\(\theta\)) was determined by measurement of the beam profile for each transducer with use of a three-dimensional acoustic test system with determination of the
Figure 2. Doppler catheter in place in the pulmonary artery. Catheter configuration places the ultrasonic transducer against the vessel wall with the ultrasound beam aimed through the central vessel diameter.

The instantaneous velocity at each of the eight range gates was calculated and stored by computer at a sampling rate of 50 Hz. A custom-designed computer program was used to reconstruct and display the instantaneous velocity profile or calculate an average profile representing a typical cardiac cycle (Fig. 3). Profile averages were constructed with use of 10 s of instantaneous profile data with phase relation determined with use of the simultaneous electrocardiogram (ECG).

Vessel diameter. This was determined with use of an automatic diameter detection scheme developed in our laboratory. This scheme for dynamic range gating and diameter detection utilized backscattered power from the returning signal at a point known to be within the vessel lumen. This signal was compared with returning signals distal to that point. Location of the far wall was determined to be the point where the reflected acoustic power exceeded, by some threshold value, the level seen at the reference point within the vessel. This increase in signal amplitude should occur at the blood-tissue interface of the opposite vessel wall. True vessel diameter was calculated from this “slant distance” with use of the following formula (Fig. 4).

\[ D = D_s \sin \theta + D_c, \text{ with} \]

Figure 3. Average velocity profile obtained from Doppler ultrasound measurement of instantaneous velocity in eight range gates distributed across a central vessel diameter. The flow profile represents a 10 s average obtained from the main pulmonary artery of the sheep at a mean cardiac output of 4.38 liters/min.

Figure 4. Calculation of vessel diameter from ultrasound transit time \((T_1)\) of pulsed Doppler signal to opposite vessel wall, \(t_1\) = transit time to vessel wall; \(t_2\) = return transit time from vessel wall to transducer; \(T\) = total transit time; \(C\) = speed of ultrasound in blood; \(D_s\) = slant distance between transducer and opposite vessel wall; \(D\) = vessel diameter; \(D_c\) = catheter diameter and \(\theta\) = angle of incidence between catheter and ultrasound beam.
\[ D = \frac{C T_1}{2} \]

where \( D \) = vessel diameter, \( D_s \) = slant distance from transducer to opposite vessel wall, \( \theta \) = incidence angle (previously defined), \( D_c \) = catheter diameter, \( T_1 \) = transmit time for the ultrasound beam to reach the opposite vessel wall and return to the transducer, and \( C \) = velocity of ultrasound in blood (previously defined).

A feedback loop was used to continually track the "wall edge" throughout the cardiac cycle. Spacing between the eight sample volumes was adjusted, using this wall tracking information, so that all eight range gates were placed with equal spacing throughout the vessel lumen. This instantaneous diameter \( (D) \) measurement was then used to calculate the vessel cross-sectional area \( (CSA) \), with use of the following formula:

\[ \text{CSA} = \frac{\pi D^2}{4} \]

**Volume flow.** Instantaneous space-average velocity for volume flow calculations was obtained by weighing the instantaneous velocity in each of the eight range gates by the percentage of the total surface area occupied by the half anulus centered at each range gate location (Fig. 5). This model assumes hemisymmetrical, three-dimensional flow profiles.

The product of the space-average velocity (obtained from the instantaneous velocity profiles) and the cross-sectional area (obtained from the diameter) was used to calculate the instantaneous volumetric flow in the vessel every 20 ms, with use of the following formula:

\[ Q = \frac{\pi D^2}{4} \left( \frac{1}{81} \sum (V_i + V_j) + 8(V_2 + V_3) - 4(V_4 + V_5 + V_6) + 4(V_7 + V_8) \right) \]

where \( Q \) = instantaneous volume flow and \( V_i \) = velocity in range gate \( i \), for \( i = 1 \) through 8.

**Experimental Model**

**Continuous flow.** A continuous flow model was set up with use of PVC tubing approximately 16 mm in diameter and a centrifugal pump used to move bovine blood at 37°C through the circuit. An electromagnetic blood flow probe (Carolina Medical Electronics) was placed into the circuit. The electromagnetic flowmeter was calibrated against timed flow and the proper probe factor established. A calibrated thermistor (Thermometrics) was placed into the Doppler catheter, approximately 4 cm from the distal tip. Iced saline solution at 0°C to 5°C was injected at approximately 4 ml/s and the flow rate indicated on the thermodilution cardiac output computer was recorded along with Doppler and electromagnetic flow. Three thermodilution, Doppler and electromagnetic flow measurements were obtained at each flow rate.

**Pulsatile flow circuit** (Fig. 6). A pneumatically powered, variable rate, pulsatile pump was constructed with use of a 300 ml latex balloon and two 30 mm porcine heterograft valves. Afterload and preload were adjusted by varying the height of the blood reservoir. An electromagnetic flow probe was placed into the circuit as illustrated and...
Figure 6. In vitro pulsatile flow model. Doppler flow measurements are obtained from the ultrasound transducer placed within the bovine aorta.

calibrated against timed flow. The Doppler catheter again contained a calibrated thermistor, approximately 4 cm from the distal catheter tip. Iced saline solution was injected approximately 20 cm upstream from the thermistor and thermodilution flow measurements were obtained by using a thermodilution cardiac output computer with computation constant set to the value previously established in the continuous flow circuit. The Doppler catheter was positioned in a 10 cm long section of fresh bovine aorta and the vascular test section was again submerged in a heated water bath.

Fresh bovine blood at 37°C was pumped through the circuit. With use of the computer controller, pneumatic pulsation of the circuit was established such that the electromagnetic flowmeter read mean flows of 1.0 to 8.0 liters/min. Pulsation rate was maintained between 36 and 100 pulses/min; 10 ml of iced saline solution was injected into the model at approximately 4 ml/s and three thermodilution cardiac output measurements were obtained at each flow rate along with mean electromagnetic and Doppler flow measurements. Instantaneous electromagnetic and Doppler flow measurements were also recorded at the various flow rates.

Animal Model

Experimental preparation. All animal studies were performed conforming to the “Position of the American Heart Association on Research Animal Use.” Three male sheep weighing between 25 and 30 kg were anesthetized with intravenous thiopental, 20 to 25 mg/kg body weight. The sheep were intubated and mechanically ventilated with use of 1% halothane in oxygen. The right femoral artery and right internal jugular vein were dissected free of surrounding tissues and isolated. A 3F arterial cannula was placed into the right femoral artery and used for
continuous arterial pressure monitoring. An 8F venous sidearm sheath was placed into the right internal jugular vein and advanced under fluoroscopic guidance into the animal's right atrium.

The Doppler pulmonary artery catheter was inserted into the right jugular sheath and advanced under fluoroscopic and pressure guidance into the distal pulmonary artery. Baseline right ventricular, pulmonary artery and pulmonary capillary wedge pressure tracings were obtained along with simultaneous ECG and femoral artery tracings. Once the ultrasound transducer was properly positioned within the main pulmonary artery, as confirmed by fluoroscopy and proximal pulmonary artery pressure tracings, the position activation wire was retracted allowing the catheter to assume its preformed configuration. The ultrasound transducer could be seen fluoroscopically to remain in contact with the pulmonary artery wall. Continuous monitoring of mean cardiac output was displayed on the Doppler cardiac output monitor, with instantaneous cardiac output recorded on a multichannel recorder.

Electromagnetic flow probe around aorta. A median sternotomy was performed; the sternum was retracted and the pericardium, heart and great vessels were exposed. The ascending aorta was dissected free of surrounding tissues and a previously calibrated electromagnetic flow probe was placed around it. The electromagnetic flow probe was placed around the aorta rather than the main pulmonary artery to prevent possible restriction and alteration of the flow profile within the main pulmonary artery.

Measurements and recordings. Pulmonary artery and pulmonary capillary wedge pressure tracings along with ECG and femoral artery tracings were obtained. These were recorded along with simultaneous Doppler-determined cardiac output and electromagnetic cardiac output. Thermodilution measurements were obtained with use of the precalibrated thermistor contained within the pulmonary artery catheter. Iced saline solution injections were performed with use of 10 ml of normal saline solution at 0°C to 5°C, injected over 4 s into the right internal jugular sheath. Cardiac output was varied over the range of 1.0 through 7.0 liters/min with use of an intravenous dopamine infusion. Three thermodilution cardiac output measurements along with mean electromagnetic and Doppler measurements were made at each stable (<1 liter/min change in flow over 5 min) level of cardiac output. Extremely low cardiac outputs (<1 liter/min) were obtained in the animal by bleed-out performed from the right femoral artery cannula at the conclusion of the experiment.

Statistical analysis. For all flow experiments, analysis of data was performed by standard linear regression with calculation of \( r^2 \), slope, intercept and SEE (27). Correlations of Doppler flow and thermodilution with electromagnetic flow were compared with use of the significance of the difference between dependent correlation coefficients with determination of \( t \) and \( p \) values (28).

Results

In Vitro Flow Measurements

Given the excellent correlation of electromagnetic flow measurements (QEM) with timed flow (Q) in both the continuous model (QEM = 1.09Q - 0.18; \( r^2 = 1.00, \) SEE = 0.11) and pulsatile models (QEM = 0.93Q - 0.06; \( r^2 = 1.00, \) SEE = 0.10), electromagnetic flow was used as the standard to which all other flow measurements were thereafter compared.

Continuous flow model. Doppler catheter-determined flow (QD) was highly predictive of electromagnetic flow (Fig. 7A): (QD = 1.04 QEM - 0.20; \( r^2 = 0.98, \) SEE = 0.44). Correlation with thermodilution flow (QTD) was also excellent (Fig. 7B): (QTD = 0.91 QEM + 0.08; \( r^2 = 0.99, \) SEE = 0.20). However, thermodilution appeared to slightly underestimate electromagnetic flow (slope = 0.91) over the flow range tested. These results are well within the reported accuracy of the thermodilution technique (4-14).

Pulsatile model. Similar results were obtained with the in vitro pulsatile model. Figure 8 illustrates instantaneous Doppler and electromagnetic flow data along with instantaneous diameter measurements. Figure 9 depicts plots of
electromagnetic flow versus Doppler catheter and thermodilution flow over the range of 0.5 through 8 liters/min. Linear regression analysis yields: \( Q_D = 1.04 Q_{EM} - 0.12; r^2 = 0.97, \text{SEE} = 0.33 \) and \( Q_{TD} = 0.84 Q_{EM} + 0.53; r^2 = 0.95, \text{SEE} = 0.34 \).

Once again, Doppler catheter flow tended to predict electromagnetic flow more accurately than did thermodilution. However, these differences in ability to predict electromagnetic flow were not statistically significant (\( p = 0.10 \)).

**In Vivo Flow Measurements**

**Doppler catheter versus electromagnetic flow.** Instantaneous femoral artery, pulmonary artery and pulmonary capillary wedge pressure tracings along with electromagnetic flow from the ascending aorta and Doppler catheter flow from the main pulmonary artery are illustrated in Figure 10. A phase lag (40 ms) and contour difference in flow waveform are notable between the instantaneous electromagnetic flow and Doppler catheter-derived flow. These differences were anticipated because electromagnetic cardiac output was obtained from the ascending aorta, whereas Doppler catheter flow was obtained from the main pulmonary artery.

**Doppler catheter versus thermodilution cardiac output.** Linear regression analysis of mean Doppler catheter cardiac output versus electromagnetic flow for one of the sheep studied is illustrated in Figure 11: \( Q_D = 0.94 Q_{EM} - 0.08; r^2 = 0.94, \text{SEE} = 0.29 \). Correlation of thermodilution cardiac output with electromagnetic flow for this sheep (Fig. 11) was also excellent. However, thermodilution tended to be less accurate than the Doppler catheter in predicting electromagnetic flow: \( Q_{TD} = 0.86 Q_{EM} + 0.79; r^2 = 0.95, \text{SEE} = 0.36 \). Combined data for all three sheep (Figs. 12A, B) reveal an excellent correlation between Doppler-determined flow and electromagnetic flow: \( Q_D = 0.87 Q_{EM} + 0.33; r^2 = 0.80, \text{SEE} = 0.61 \). Thermodilution less accurately predicted electromagnetic flow: \( Q_{TD} = 0.79 Q_{EM} + 0.50; r^2 = 0.73, \text{SEE} = 0.72 \). Overall, Doppler catheter-determined cardiac output appeared statistically better than the thermodilution-determined cardiac output at predicting electromagnetic flow (\( p < 0.005, t = 2.95 \)).

**Discussion**

**In Vitro Flow**

Electromagnetic flow was shown to have excellent correlation with timed flow in both the continuous and pulsatile flow circuits and was used as the standard to which all other flow measurements were compared. Use of electromagnetic flow in the pulsatile circuit also allowed comparison of instantaneous electromagnetic and Doppler catheter data.

**Doppler catheter versus thermodilution flow.** Both Doppler catheter-determined flow and thermodilution-determined flow appeared to correlate excellently with electromagnetic flow in the continuous and pulsatile flow models. Thermodi-
Figure 10. Instantaneous recordings in the sheep of pulmonary artery, femoral artery and pulmonary capillary wedge pressure tracings along with pulmonary artery diameter and Doppler and electromagnetic cardiac outputs. EKG = electrocardiogram.

In Vivo Flow

Mean flow data. In the animal model, Doppler catheter-determined cardiac output appeared to more accurately predict electromagnetic flow than did simultaneous thermodilution-measured output for combined data for all three animals (Fig. 12). These differences in the abilities of Doppler versus thermodilution to accurately predict electromagnetic flow were statistically significant.

Instantaneous flow data. Instantaneous animal data revealed minor differences in waveform and phase between electromagnetic flow and Doppler catheter-determined flow (Fig. 10). These differences were likely due to differences in the site of measurement: Doppler catheter flow was obtained from the main pulmonary artery, whereas electromagnetic flow was obtained from the ascending aorta. The electromagnetic flow probe was not placed around the main pulmonary artery because of the possibility that it might cause vessel restriction and alter the flow profile. Differences in right versus left ventricular ejection, acceleration, pressure and distal impedance would also account for variation between Doppler and electromagnetic flow waveforms. Time-average mean flow values, taken over several respiratory cycles (10 s), should provide similar cardiac output data from both the aorta and the pulmonary artery. However, cardiac output measured from the ascending aorta should be less than that measured from the main pulmonary artery by an amount equal to coronary flow, which may represent nearly 5% of the total cardiac output.

Effect of velocity profile on flow calculation. Previously described methods of obtaining cardiac output using Doppler-determined maximal velocity and estimates of aortic or pulmonary areas have assumed flat velocity profiles and constant cross-sectional areas (16-21). The Doppler catheter system continuously maps a one-dimensional profile with use of the multirange-gated system described and dynamically tracks intravascular diameter. The calculated instantaneous space average velocity and instantaneous area are...
then used to calculate a true instantaneous flow value with use of equation 5. Examination of data obtained from our animal experiments suggests that the velocity profile within the main pulmonary artery may be quite parabolic during peak systole (Fig. 3). Such a profile would result in gross overestimation of mean flow if maximal time-average velocities rather than true space-average velocities are used to calculate cardiac output. Such overestimation of cardiac output by Doppler technique compared with the value obtained with thermodilution or Fick technique has been reported (17).

Effect of diameter change on flow calculation. Additionally, our data indicate (Fig. 12) up to a 10% change in internal diameter for the main pulmonary artery during the cardiac cycle (dependent on cardiac output and pulmonary artery pressure). This change would result in a 21% variation in calculated surface area based on circular cross-sectional geometry. Thus, significant variation in cardiac output calculations may occur, dependent not only on location of diameter determination within the ventricular outflow anatomy, but also on the timing of static diameter measurements.

Limitations

Assumption of a symmetrical flow profile. There are a number of potential sources of error in our experimental protocol. Doppler catheter-derived calculation of space-average velocity is based on the assumption of hemisymmetrical three-dimensional flow profiles within the vessel of interest. Although this assumption is likely valid for the long, straight section utilized in our in vitro studies, curved vessels such as the main pulmonary artery should produce somewhat skewed flow profiles. The degree of deviation from three-dimensional symmetry is critically dependent on radius of curvature. Relatively short vessels with large radii of curvature, such as the main pulmonary artery, would be expected to exhibit fairly symmetrical flow profiles (Fig. 3). However, the assumption of a truly symmetric flow profile will result in error in any flow calculation using space-average velocity obtained from measurements in only one plane.

Stability of catheter transducer against vessel wall. Doppler-determined flow velocities are critically dependent on the angle of incidence between acoustic beam and flow. The Doppler catheter used in our experiments utilized a small acoustic transducer constructed with a fixed angle to the main catheter body (Fig. 1). This angle was measured utilizing a three-dimensional acoustic test system to measure beam profiles for each transducer. This measured angle was then used in all subsequent Doppler flow calculations. This system, however, assumes that the bent catheter configuration fixes the transducer in a stable position against one vessel wall. Should the transducer not be held in such a position, underestimation of vessel diameter and over- or underestimation of the incidence angle between the ultrasound beam and flow would occur. Fluoroscopy of the catheter during in vivo testing revealed a relatively stable position for the transducer. However, some variation with pulsation of the pulmonary artery is likely to occur.

Catheter position in relation to the pulmonary artery. The bent catheter configuration increases the likelihood that the catheter will span the main pulmonary artery across its major dimension (diameter). This positioning assures aiming of the acoustic beam through a central vessel diameter. Should the catheter ‘torque’ in relation to the vessel, the acoustic beam would transect a chord rather than the true vessel diameter, resulting in an underestimation of vessel diameter and error in calculation of the velocity profile. In both the in vitro flow models and the animal model, the catheter was manually torqued in both directions during continuous monitoring of vessel diameter to determine whether the maximal dimension was obtained with the catheter in its neutral position (after retraction of the guide wire). In both models, the distance between the transducer and the opposite vessel wall was maximal when the catheter was in this neutral position, thus suggesting that the acoustic beam transected the vessel diameter in this position. All
quantitative Doppler measurements were obtained with use of this neutral catheter position.

Disturbances of flow profile: turbulent flow. The presence of the catheter body both across the pulmonary artery and proximal to the ultrasound beam may disturb the flow profile, thereby introducing error into the calculation of the space-average velocity and flow. Fluid dynamic theory predicts that the effect of such a "wake" disturbance or separation layer will be negligible beyond a distance equal to approximately five times the diameter of a cylindrical obstacle (the size of the catheter) placed within a flow stream (the size of the main pulmonary artery) (29). All Doppler flow measurements were obtained beyond this distance. Additionally, laminar flow patterns will be maintained within the blood vessel if the ratio of obstruction height (c) (catheter diameter) to vessel radius (r) is <20% of the square root of the mean Reynolds number (30). For a catheter diameter of 2.0 mm, pulmonary artery diameter of 25 mm and mean pulmonary artery velocities of 25 to 50 cm/s (measured during systole), this c/r ratio is well below that required to create a turbulent flow profile.

Experimental errors in determination of thermodilution flow. Thermodilution flow measurements tended to underestimate electromagnetic flow in both the in vitro continuous and pulsatile models. Such underestimation may be accounted for by inadequate mixing of injectate or by increased heat exchange between the vascular test section and heated water bath. Heat exchange in the pulmonary bed, which surrounds the pulmonary artery with air, would be less and should result in less underestimation (31). Other potential sources of experimental error in determination of thermodilution flow include: inaccurate determination of thermodilution computation constant, errors in method or site of indicator injection or slow rate of indicator injection. Additionally, despite attempts to obtain mean electromagnetic and Doppler flow measurements simultaneously with thermodilution during stable hemodynamic conditions, some variation in dynamic cardiac output may have occurred. Our overall accuracy of thermodilution flow measurements, however, remains within the wide range reported by others (4–14).

Additionally, a larger number of simultaneous data points were obtained using on-line, simultaneous measurement of Doppler and electromagnetic flow than was possible with thermodilution. This difference was due to the time required to obtain thermodilution measurements. This discrepancy may have contributed to the superiority noted for the Doppler catheter versus thermodilution in prediction of electromagnetic flow in portions of our statistical analysis.

Conclusions

We have developed a flow-directed, pulmonary artery catheter with a Doppler ultrasound transducer, that is capable of measuring instantaneous cardiac output in the intact animal. Instantaneous velocity profile maps, obtained in a single plane, along with instantaneous measurements of vessel diameter were obtained with this system. Doppler flow calculated by using instantaneous space-average velocity (obtained from the velocity profile) and instantaneous area (obtained from vessel diameter) compared favorably with electromagnetic flow measurements. In addition, continuous monitoring of instantaneous and mean cardiac output appears possible with use of this Doppler catheter system (Fig. 10). Currently available pulmonary artery catheters provide continuous measurement of pressure and mixed venous oxygen saturation only. Mixed venous oxygen saturation reflects flow but may be affected by variables other than cardiac output.

A Doppler catheter system that provides continuous monitoring of cardiac output may provide the physician with an early warning of acute changes in cardiac output due to ischemia, hypovolemia or increases in pulmonary or systemic vascular resistance. In addition, the ability to accurately measure instantaneous and mean cardiac output should provide useful information concerning hemodynamic variables such as peak flow, acceleration, deceleration, stroke work, pulmonary resistance and impedance. Further studies are required to confirm the accuracy of this method for measuring cardiac output in humans.

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References


