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VELOCITY OF BLOOD FLOW AND STROKE VOLUME OBTAINED FROM THE PRESSURE PULSE *

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There are several methods available for obtaining the velocity of blood flow from the aorta (1-5). The bristle flowmeter and the electromagnetic flowmeter are two such methods which are satisfactory in some circumstances; however, both require an open chest preparation and isolation of the aorta. More recently it has been suggested that if suitable restrictions are imposed on the Navier-Stokes equations, the axial aortic pressure gradient, $\partial p/\partial x$, can be related approximately to the instantaneous blood velocity, u , by the equation:

$$-\frac{\partial p}{\partial x} = \rho \frac{du}{dt} + au, \quad 1)$$

where x is the axial coordinate of the tube, t is time, ρ is the blood density, and "a" is a "blood friction" constant. This equation can be instantaneously and continuously solved for u , the blood velocity, by simple analog computer technics if the pressure gradient, $\partial p/\partial x$, is known (6, 7). The pressure gradient can be estimated by measuring the difference in lateral pressure between two points several centimeters apart along the aorta. This pressure difference may be obtained with a suitable double lumen catheter. Consequently, this method has a great advantage over previously described technics since it may be used in the intact animal or man.

This method has the great disadvantage, however, of requiring a large catheter (usually necessitating the sacrifice of an artery) and two exactly balanced manometers. Therefore, a modification permitting the use of a single catheter and a single manometer would be advantageous. It will be the purpose of this report to suggest a further restricting assumption which permits use of a single

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lumen and to present data that lend support to the validity of this assumption.

THEORY

If a pressure wave travels undistorted and without reflection down an elastic tube, then the instantaneous pressure gradient, $\partial p/\partial x$, is related to the time derivative of pressure, $\partial p/\partial t$, at a point by

$$-\frac{\partial p}{\partial x} = \frac{1}{s} \frac{\partial p}{\partial t}, \quad 2)$$

where s is the pressure pulse wave velocity down the tube. Substituting Equation 2 into Equation 1 gives:

$$\frac{\partial p}{\partial t} = \rho s \frac{du}{dt} + asu. \quad 3)$$

This equation may be instantaneously and continuously solved for the velocity, u , by the same computer technics previously mentioned if $\partial p/\partial t$ is measured.

METHODS

Eight dogs were anesthetized with Nembutal® and tracheotomized for positive pressure artificial respiration. The left chest was opened. A large cannula connected to a reservoir of blood was placed in the left atrium so that the atrial pressure could be varied as desired.

Pressures from the ascending aorta were obtained through a PE 250 Odman Ledin catheter attached to a Statham P23D pressure transducer. An eight channel Electronics for Medicine¹ recorder was used. The frequency response of the system consisting of the catheter, pressure transducer and strain gauge amplifier was checked and determined to be flat to 50 cycles per second. The output from the strain gauge amplifier was fed into another amplifier through an RC differentiator¹ so as to obtain the first time derivative of aortic pressure. This differentiator is reliable up to 80 cycles per second. The signal from the amplifier carrying the first derivative of the aortic pressure pulse was led through the analog computer described by Fry, Mallos, Casper and Noble (6, 7). This computer has an adjustment which provides for variations in the frictional and inertial factors of the

¹ Made by Electronics for Medicine, Inc., White Plains, New York.

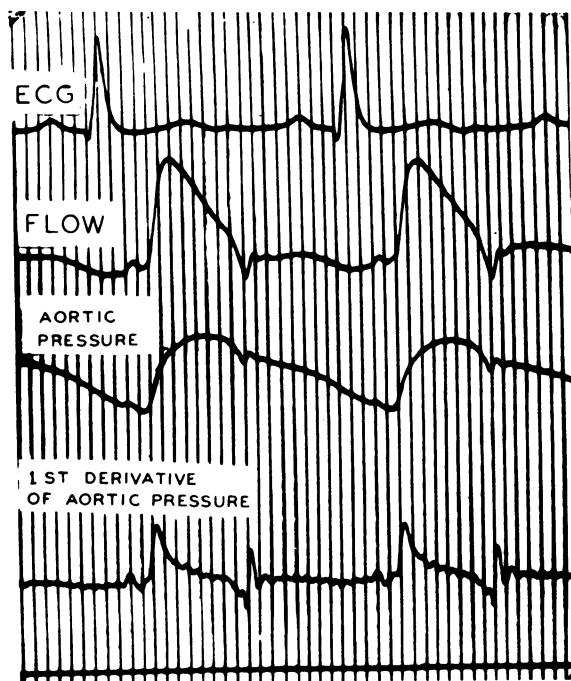


FIG. 1. A REPRESENTATIVE TRACE SHOWING THE ECG, THE BLOOD VELOCITY CURVE, THE AORTIC PRESSURE PULSE AND THE FIRST TIME DERIVATIVE OF THE AORTIC PRESSURE PULSE

The paper speed is 100 mm. per second and the time lines are 0.04 second.

Navier-Stokes equation. The resulting signal is led into a third amplifier and recorded as the velocity curve.

The frictional and inertial factor computer adjustment was set so that the velocity curve obtained was on the baseline during the last one-third to one-half of diastole. This setting was made by eye and is necessarily somewhat arbitrary. However, once the setting was made on each dog, no further changes were made during the course of the experiment.

In each of the eight dogs used, the cardiac output was varied by changing the atrial reservoir level. In six of the dogs, the cardiovascular status was further varied by infusions of first epinephrine and then methoxamine.

Multiple dye dilution cardiac output determinations were made on each dog, using Cardio-Green² and the Enesco Oximeter.³ While the dye dilution curve was being obtained on one recorder, simultaneous aortic pressure pulse, first time derivative of the aortic pressure pulse, aortic blood velocity and electrocardiogram were obtained on a second recorder (Figure 1). These simultaneous determinations were made under widely varying conditions of heart rate, cardiac output and blood pressure.

² Supplied by Hynson, Westcott and Dunning, Inc., Baltimore, Md.

³ Manufactured by ENSCO Engineering Specialty Co., P.O. Box 19, Sugar House Station, Salt Lake City, Utah.

RESULTS

The data were calculated as follows: The minute systolic area under each velocity curve was determined planimetrically. This was done by measuring from five to 15 consecutive complexes, determining the average and multiplying by the heart rate. A constant was determined for each dog to convert the area under the velocity curve into cubic centimeters of blood by dividing the second dye dilution cardiac output by the cor-

TABLE I
*Comparison of cardiac outputs estimated by dye dilution and blood velocity methods**

	I	II	I	II
Dog No. 1	866 1,477 1,970 2,166	1,037 1,477 1,799 1,973	886 1,040 1,770 2,181	902 1,040 1,376 1,792
Dog No. 6	1,180 1,668	1,119 1,668	Dog No. 6	1,452 1,130 1,074
Dog No. 2	2,000 3,030	2,074 2,465	Dog No. 6	1,274 1,579 1,083
Dog No. 3	436 596 827	443 596 737	Dog No. 7	2,044 2,213 1,310
Dog No. 4	560 800 1,230 1,330 1,070	497 800 1,168 1,361 1,092	Dog No. 7	1,637 2,140 3,231 2,900 2,839
Dog No. 4	870 1,010 1,420 1,290 420 446 720 400	1,107 1,361 1,527 1,135 373 377 558 408	Dog No. 8	2,388 2,320 2,085 3,063 3,399 2,933
Dog No. 5	1,380 1,970 2,680 2,380 2,440 1,980 1,820 1,950 1,805 2,300 2,820 2,360 1,025	1,613 1,970 2,418 2,207 2,150 1,631 1,386 1,660 1,877 2,291 2,859 2,235 1,094	Dog No. 8	1,900 1,877 1,445 1,472 1,127 1,334 1,277 1,422 2,071 1,663 1,369 810

* Column I is the cardiac output in cubic centimeters per minute as determined by dye dilution. Column II is the corresponding area under the blood velocity curve adjusted to cubic centimeters per minute with a constant determined by dividing the second cardiac output in each dog experiment by the corresponding area under the velocity curve.

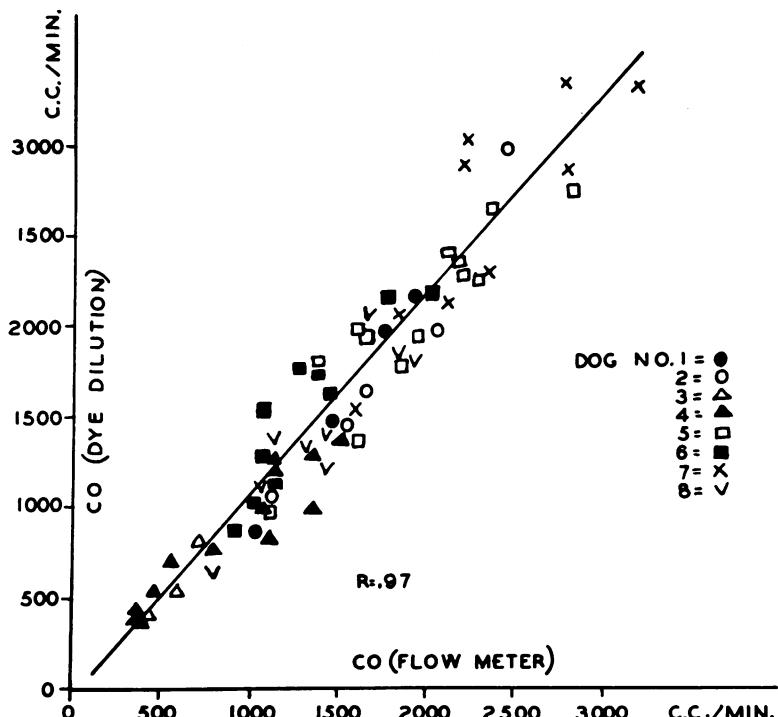


FIG. 2. COMPARISON OF CARDIAC OUTPUTS BY DYE DILUTION AND BLOOD VELOCITY METHODS

The ordinate is the cardiac output as determined by dye dilution. The abscissa is the corresponding area under the aortic velocity curve converted to cubic centimeters per minute by a constant determined by dividing the second cardiac output in each dog experiment by the corresponding area under the velocity curve. The correlation coefficient is 0.97. The average deviation is 10 per cent and the maximum deviation is 35 per cent. Individual correlation coefficients are from dogs 1 to 8: $R = 0.99, 0.96, 0.99, 0.94, 0.92, 0.91, 0.85$ and 0.92 , respectively, ($y = -45 + 1.1x$).

responding minute systolic area. The second determination in each dog was used because the left atrial pressure was near normal at this time. The results are presented in Table I, Column II, and graphically in Figure 2. A correlation coefficient of 0.97 is obtained.

DISCUSSION

In this study, the data demonstrate the close correlation of the area under the flow curve to stroke output. The velocity curve obtained is, from mathematical considerations, proportional to blood velocity (centimeters per second). This is proportional to flow (cubic centimeters per second) provided the size of the vessel does not change appreciably. This would appear to be the case because no significant correlation was found

between the mean ejection blood pressure and the ratio of cardiac output by the flowmeter method to the cardiac output by the dye method. If changes in aortic size were an important cause of the scatter noted in Figure 2, this method of analysis should have detected it. The maximum variation of mean ejection pressure was found in Dog No. 4 and was from 50 to 249 mm. Hg.

No evidence of the validity of the form of the velocity curve was obtained. However, a simultaneous comparison with a bristle flowmeter has been found to give curves of similar contour. Fry, using his method, has demonstrated his velocity curves to have the same contour as those obtained with an electromagnetic flowmeter (7).

In a previous communication from this laboratory, it has been pointed out that the spatial aortic pressure differential and the first time derivative

of pressure are similar in timing and contour. This relationship lends support to the basic assumption in this method that there is relatively little reflected wave energy in the ascending aorta.

From the mathematical analysis, it is apparent that variations in the rate of pulse transmission (S) will affect the recorded blood velocity. From the data presented, this variation appears to be insignificant within the conditions of the experiment.

The primary advantage of this method of application of the Navier-Stokes equation is that a single lumen catheter is used. This permits the use of a single catheter-manometer system which has a good frequency response. To obtain comparable results with the two-lumen method, both manometer systems must be set at the same sensitivity and have the same frequency response. These conditions are difficult to obtain. However, it must be pointed out that the presence of significant reflected pressure waves will theoretically distort the recorded velocity curve using the time derivative. This objection does not apply to the use of the spatial pressure differential. To illustrate, if pressure is measured at a single point, the computer is fed the same signal whether a given component of the pressure pulse is traveling toward or away from the aortic valve, but with the two-manometer system the signal fed the computer is opposite in phase if a given component is traveling upstream as a reflected wave. The calibration method used for stroke output (a comparison with a dye dilution cardiac output) is advantageous in that the size of the aorta is not a factor.

The method described appears to be a relatively simple method of obtaining beat-by-beat stroke volume and blood velocity in any situation where an accurate pressure pulse can be recorded. Obtaining pressure pulses from the human ascending aorta is practical using a small radiopaque catheter passed retrograde from the brachial artery by the Seldinger percutaneous technic (8). Since a

small catheter is suitable, stroke volume can be obtained without permanent damage to the brachial artery.

SUMMARY

A method for using a single lumen catheter in recording a continuous blood velocity in the aorta is described. The system is based on the Navier-Stokes equation and uses an electronic computer. The cardiac output as determined by the area under the velocity curves is compared with cardiac outputs determined simultaneously with the dye dilution technic. A correlation coefficient of 0.97 is obtained.

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