Sequential Velocity Development in the Ascending and Descending Aorta of the Dog

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ABSTRACT

A conical hot-film probe was used to measure instantaneous velocities in the ascending and descending aorta of anesthetized open-chest dogs. From these point measurements, the radial distributions of velocity were obtained over one cardiac cycle. In general, the hot-film measurements confirmed the observation that the velocity profiles tend to be flat with the highest rates of shear confined to the region of the wall. There were, however, significant variations in the detail from one dog to another. These variations in the shape of the profiles probably are a consequence of many geometrical factors, which include valve inlet geometry, configuration and orientation of the valve plane, and aortic curvature and branching in the descending aorta.

KEY WORDS
hot-film probe  aortic blood flow  velocity profile  cardiac output  pulsatile flow  rates of shear

At the current time most instrumentation for hemodynamic studies measures the instantaneous volume of time-averaged bulk flow and does not permit detailed investigation of pulsatile phenomena. Yet it is apparent that there are a number of important questions in the field of circulatory physiology which can be answered only by detailed in vivo measurements of pulsatile blood flow. One would like to know, for example, the magnitude of the local shearing stresses in the arterial system, for it is widely believed that when these stresses are sufficiently large the cellular elements in the blood are irreparably damaged. Endothelium lining the blood vessels may similarly be subjected to damage by these same stresses acting at the walls. The physiological significance of pulsatile motions on the blood vessel walls, particularly the aorta, is also of interest.

These walls contain both active (muscle) and passive (connective) tissues which cause the vessel to change dimensions during the flow period and thus modulate the flow developed initially by the heart. One might also inquire about the flow phenomena which contribute to the opening and the closing of the heart valves or about the effect of branching in the vessels on the shape of the flow pulse. These questions and many more have been addressed in the past. However, since detailed flow measurements are unavailable, the answers necessarily are incomplete.

To make in vivo measurements of instantaneous point velocities, a probe is needed which (1) is small compared to the size of the containing vessel, (2) possesses a reasonably high frequency response and (3) is strong. One probe having these characteristics is the hot-film probe used successfully for a number of years in engineering studies of more general fluid motions. Recent studies (1-7) show that these devices can be adopted for use in open-chest surgery when the probe is introduced through the wall of the vessel or for use as catheter probes introduced into the vessels under less severe clinical procedures.

The purpose of this paper is to present measurements of sequential velocity profile.
VELOCI TY DEVELOPMENT IN AORTA

Development in the ascending and descending aorta. First, however, it will be of value to review briefly our current knowledge of pulsatile flow phenomena in tubes. This review will facilitate the interpretation of the in vivo phenomena.

Analytical solutions to certain classes of laminar pulsatile flow problems date back to about 1930 (8, 9), but the most general solutions for axisymmetrical laminar flow in straight, rigid-walled tubes were developed more recently by Womersley (10) and Uchida (11). Both of these men linearized the Navier-Stokes equations of motion by neglecting the inertial terms and solved them for the case of a Newtonian fluid driven by a pressure gradient having an arbitrary wave form: the only requirement for a solution was that the wave form be amenable to Fourier representation. The total velocity or bulk flow rate resulting from the arbitrary pressure gradient was simply that obtained from a linear combination of the appropriate series representation. Womersley (12) subsequently generalized this work to include the case of pulsatile flows in distensible tubes.

Womersley’s solution pertained to flows in tubes of infinite length (long-tube theory), and entrance effects, such as might be found in the ascending aorta or at positions just downstream from bifurcations, were not considered. Szymanski (13) obtained a solution to the problem of a fluid started impulsively from rest in an infinite tube. The pressure wave was a step function, and the limiting solution was Poiseuille flow. Chang and Atabek (14) obtained a solution for oscillatory flow in the entrance region of a semi-infinite rigid tube: this solution was generalized recently by Kuchar and Ostrach (15) to include the case of distensible tubes. The solutions were all similar in that they resulted from linearizations of the Navier-Stokes equations of motion. Unlike the long-tube solutions cited earlier, however, the inertial terms in these equations were retained. Chang and Atabek (14) gave the dimensionless inlet length as \( L(R_0, Re)^{-1} = 0.16 \), although this value depended on Womersley’s dimensionless parameter, \( \alpha^2 = (Re/\omega/v) \), and on the phase angle in the wave motion (14, 15). Taking the total length of a large canine aorta measured from the aortic valve to about the renal artery as 25 cm, the mean diameter as 1 cm, and the mean flow Reynolds number as 500, then \( L(R_0, Re)^{-1} = 0.100 \). From this we concluded that the whole of the aortic flow lies within the entrance flow region. Such a calculation, of course, fails to take into account the curvature of the aorta, its taper, or the effect of branching. Further, the theory is based on a uniform entrance flow at the aortic valve. The influence of these factors has yet to be determined, but it probably is considerable.

Symbols and Parameters

- \( K \) = Calibration constant defined in Eq. 1.
- \( L \) = Entrance length.
- \( n \) = Calibration constant defined in Eq. 1.
- \( R_0 \) = Radius of aorta.
- \( r \) = Radial coordinate.
- \( T, t \) = Time.
- \( U \) = Instantaneous velocity at a point in aorta.
- \( U_0 \) = Uniform entrance velocity.
- \( U_{max} \) = Maximum instantaneous velocity at some point in aorta.
- \( U_{Cl, max} \) = Maximum instantaneous velocity on center line of aorta.
- \( \bar{U} \) = Velocity of mean flow.
- \( V \) = Voltage output of anemometer.
- \( V_0 \) = Voltage output of the anemometer at zero flow.
- \( X, x \) = Axial coordinate.
- \( \nu \) = Fluid kinematic viscosity in cm/sec².
- \( \tau \) = Duration of one cardiac cycle.
- \( \omega \) = Angular frequency of velocity (or pressure) wave.
- \( \Re_p \) = Peak Reynolds number = \( 2U_{Cl, max}R_0/\nu \).
- \( \Re \) = Reynolds number for the mean flow = \( 2\bar{U}R_0/\nu \).
- \( \alpha \) = Pulsatile parameter = \( R_0(\omega/v)^{1/2} \).

Methods

The hot-film probe and traverse assembly used in this study are diagramed in Figure 1. The probe was similar to commercially produced probes used for other purposes except that the lower shaft had a smaller distance (1.5 mm), and the conical tip was sharper to facilitate piercing the aortic wall. The probe was different from those used by other investigators (1, 3, 6, 7) for...
The hot-film probe is attached to the probe clamp and mounted on the anterior collar attached to the graduated shaft.

measuring blood velocity; its frequency response was flat to 30 kHz (16). The response was probably better than that of the probes used by other investigators, although this fact cannot be established with certainty since the other investigators did not report the frequency response of their probes. On the other hand, our probe was less suitable for monitoring flow reversal, since the reverse flow could only be measured by reversing the direction in which the probe faced.

The flow sensor was a vacuum-sputtered nickel film located near the base of the conical tip. Electrical insulation and abrasion resistance were obtained with an overlay of quartz, also deposited by vacuum sputtering. Both films were very thin (at most a few thousand Angstroms) so that thermal inertia was minimized.

The probe traversing assemblies were made with collars of various sizes to accommodate vessels of different diameter. The upper shaft of the probe was attached to the probe clamp which was free to move along the external traverse scale: the position of the probe was thus determined by noting the position of the clamp along the scale. When properly inserted, the lower shaft of the probe lay inside and parallel to the wall of the aorta and faced into the on-coming flow. Reverse flow was measured by rotating the probe 180°.

The nickel film was heated by an electric current to a temperature a few degrees above the temperature of the fluid whose velocity was to be detected. A DISA 55D01 linearized constant-temperature anemometer was used as a feedback amplifier system to keep the nickel film at the fixed temperature. Heat was lost from the film to the blood at a rate which depended on the velocity of the fluid in the immediate vicinity of the film. Variations in flow velocity appeared on the anemometer as variations in voltage. The relationship between the local velocity and the voltage was determined by calibration in the laboratory with flows whose velocities were accurately known.

The probes were calibrated in water and in heparinized blood in a small tow tank (13 x 13 x 144 inches). A tow cart was used to pull the hot-film probe through the quiescent liquid at a known velocity. Mechanical relays at fixed distances apart were tripped by the cart as it moved along the tank. The time required to traverse this distance was measured accurately with a frequency counter and an oscillator, and from these data the velocity of the probe was determined. Calibration curves for the probes immersed in blood and water are shown in Figure 2. The relationship between velocity, U, and voltage, V, is

\[ V^2 - V_0^2 = KU^{1/n}, \]

where K and n are parameters which depend on the properties of the fluid and the characteristics of the hot-film probe. \( V_0 \) is the voltage signal at zero flow. The curves for blood and water were parallel but differed in position by a factor of 1.22, primarily because the two fluids have different physical properties.

Studies were made on 11 dogs weighing 48-83 lb. They were anesthetized with sodium pentobarbital. Their tracheas were intubated, and the dogs were ventilated with room air, using a Harvard respiratory pump. The right femoral artery and vein were isolated and catheterized for subsequent measurements of cardiac output by hydrogen-infusion techniques (17). The chest was opened using standard surgical techniques (mid-sternal thoracotomy), and the heart was kept in its usual anatomical orientation by a pericardial cradle. The ascending aorta was dissected free of enough connective tissue to fit the traverse collar; the probe, attached to the probe clamp, was inserted into the aorta. An appropriately sized traverse assembly was then fitted around the aorta and the probe clamp attached. The anterior and posterior walls of the aorta were located with reference to the traverse scale, and the probe was positioned at various radial stations for short periods of time. Ten or more pulses were recorded at each station before the probe was moved.

One of the difficult problems associated with the use of the hot-film anemometer (and the electromagnetic flowmeter or the acoustical flowmeter, for that matter) was the establishment of the zero-flow condition. This was important since the absolute values had to be fed into the
The hot-film probe was calibrated in water and in heparinized blood by towing the probe through the fluid at known speeds. After each experiment on a dog, the calibration was repeated; no change was observed over the course of the experiments. The calibrations are linear on a log-log scale from 2-200 cm/sec. \( V_0 \) is the voltage corresponding to the zero-velocity condition; its value must be established in vivo before each run for the determination of absolute velocities. \( V \) = voltage, \( U \) = velocity of flow; \( n \) = parameter dependent on property of the fluid.

A standard electrocardiogram made with needle electrodes was recorded simultaneously with all measurements. The R wave provided a reference time base for synthesizing the instantaneous velocity profiles from the recorded velocity waves.

Aortic or left ventricular pressure, or both, were measured using a well-flushed, short, wide-bore cannula attached to a Statham P23Db transducer. This system was calibrated and found to have a natural frequency in excess of 150 Hz; its response was flat to 75 Hz.

All of the above data plus a voice log describing events and velocity probe positions were recorded on either a Honeywell 7000 or a Hewlett-Packard 3960 FM tape system for later playback and analysis.

Finally, for comparative purposes, measurements of the instantaneous volumetric flow rates were made using a Biotronix BL-610 pulsed-logic electromagnetic flowmeter, and cardiac output was measured by the hydrogen-infusion technique (17).

Results

Figure 3 shows the velocity waves measured along the center line of the ascending aorta of
one of the dogs at a point about 4 cm above the valve. Also included in this figure are simultaneously measured aortic root pressure and ECG. Two velocity traces taken at different times are needed to describe the total flow wave because the conical hot-film probe is direction sensitive, i.e., it must always point into the flow to give meaningful information. This direction sensitivity is evident in the two traces: the regurgitant flow is hardly noted in the trace for the forward flow, and the forward flow is barely noticeable in the trace for the regurgitant flow.

To obtain a total flow wave it was necessary to point the probe towards the valve to obtain the forward flow and to point it away from the valve to obtain the regurgitant flow. The total flow wave illustrated in Figure 4 was obtained by combining the meaningful portions (i.e., the forward flows with respect to the probe) of the two flow waves after first superposing the corresponding ECG traces recorded with each velocity measurement. The velocity profiles were developed from traces like those in Figure 3 made at 1-mm intervals across the aorta.

Figures 4-9 show such profiles in the ascending and descending aorta for a number of dogs. In these figures each profile pertains

**FIGURE 4**

Velocity profiles in the ascending aorta for dog 6. Left: Accelerating flow. Right: Decelerating and regurgitant flow. The velocity, $U$, at each point is made dimensionless by dividing by the maximum velocity, $U_{\text{max}}$. The radial position, $r$, is normalized by dividing by the aortic radius, $R_o$. Zero is the tube center line. Time is normalized by dividing the time, $t$, by the cardiac cycle, $T$. Each profile pertains to a different time in the cardiac cycle. For this dog $R_o = 0.6$ cm, $U_{\text{cl, max}} = 47$ cm/sec, and $t = 0.33$ seconds.

**FIGURE 5**

Velocity profiles in the ascending aorta of dog 8. Left: Accelerating flow. Right: Decelerating flow. Each profile pertains to a different time in the cardiac cycle. The curves are normalized as described in Figure 4. For this dog $R_o = 0.68$ cm, $U_{\text{cl, max}} = 52$ cm/sec, and $t = 0.34$ seconds.

**FIGURE 6**

Velocity profiles in the ascending aorta for dog 10. For this dog $R_o = 0.66$ cm, $U_{\text{cl, max}} = 103$ cm/sec, and $t = 0.45$ seconds.

**FIGURE 7**

Velocity profiles in the ascending aorta for dog 11. For this dog $R_o = 0.65$ cm, $U_{\text{cl, max}} = 64$ cm/sec, and $t = 0.37$ seconds.
The velocity profiles contained in Figures 4-7 were measured in the ascending aorta at a point three to four aortic diameters distal from the aortic valve. Those in Figures 8 and 9 were measured in the straight segment of the descending aorta at a point about 6 cm distal from the left subclavian artery. Figures 7 and 9 contain measurements in the two aortic segments of a single dog. In the ascending aorta of this dog, $\alpha = 14$; in the descending segment, $\alpha = 9.5$. Examination of these data reveals that the shapes of the profiles changed with position along the aorta, with time at a given position, and, perhaps more significantly, from one dog to another.

Although regurgitant flow was observed in the ascending aorta of all the dogs, it is shown only in Figure 4. It is not shown in Figures 5-7 either because it was not measured (Figs. 5, 6) or because it was so small that the velocities were less than the estimated error of the measurements (Fig. 7). No significant regurgitant flow was observed in the descend-
ing aorta of either of the dogs referred to in Figures 8 and 9.

The velocity waves from each of the dogs were examined for signs of turbulence. In every case, however, the flow was laminar. Disturbances of course were observed, but they never degenerated into turbulence.

The velocity wave trains measured along the center line are compared in Figure 10 for the forward flow in the ascending and the descending aorta of the same dog. Once again, since the measurements were made at different times, the comparison was obtained by overlaying the ECG traces. Left ventricular pressure recorded simultaneously in each instance showed no significant change over the course of the experiment.

The vital statistics for all the dogs are contained in Table 1. Also tabulated are the instantaneous peak velocity along the center line of the aorta, $U_{CL, \text{max}}$, the area-averaged velocity, $\bar{U}$, the corresponding Reynolds numbers, $Re_p$ and $Re$, and the values for $\alpha$, all for the ascending aorta.

**Discussion**

The center-line velocity wave form shown in Figure 3 is typical of those obtained in the ascending aorta of all the dogs and is more or less similar in form to a volumetric flow curve obtainable with an electromagnetic flowmeter. The wave forms differed in detail from one dog to another, however, for various physiological reasons which subsequently will be discussed. Comparing the velocity wave with the pressure wave measured simultaneously in the aortic root, the peak of the back flow occurred at the same time that the dicrotic notch appeared in the pressure trace. Since it is reasonably well established that the dicrotic notch is associated with valve closure, this helps to establish the credibility of the measurements. This is important because there are no independent, widely accepted measurements against which those contained in this paper may be compared.

Figures 4–7 contain the developing velocity profiles measured about three aortic diameters downstream from the aortic valve for four different dogs whose characteristics are given in Table 1. Most of the profiles were more or less blunt with no hint of flow reversals near the wall during periods of net forward flow. Furthermore, each dog possessed a distinctive set of profiles. For example, for the dog of Figure 4, the forward flow was reasonably symmetrical and there was substantial regurgitant flow. The dog of Figure 5, however, showed a symmetrical flow with rather large peak velocities near the wall. The two other

**Table 1**

<table>
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<tr>
<th>Dog</th>
<th>Wt (lb)</th>
<th>Heart rate (beats/sec)</th>
<th>$2R_o$ (cm)</th>
<th>$a$</th>
<th>$U_{CL, \text{max}}$ (cm/sec)</th>
<th>$\bar{U}$ (cm/sec)</th>
<th>$Re_p$</th>
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</table>

$a = Re (\omega / \rho)^{1/2}$, where $Re$ is the aortic radius, $\omega = 2\pi$ times the first harmonic of pulse frequency, and $\nu$ = kinematic viscosity or $3.8 \times 10^{-4}$ cm$^2$/sec. $Re_p = (2U_{CL, \text{max}} R_o) / \nu$, where $U_{CL, \text{max}}$ = maximum velocity along the tube center. $Re = 2U \bar{U} / \nu$, where $U$ is the velocity of the mean flow. $U$ and $Re$ are uncorrected for regurgitant flow.
dogs (Figs. 6, 9) possessed skewed but otherwise uniform profiles. Measurements on the other dogs listed in Table 1 confirmed the variations from dog to dog. The reasons for the variations within the ascending aorta are not well understood, but some of them probably are attributable to the way in which the valve opens and closes. For example, the three leaflets may not function evenly to produce flows that are axisymmetrical. The fact that the valve plane is slightly oblique to the axis may also contribute to the skewness of the profiles. Finally, we note that the uniform entrance condition was but an analytical simplification which has no present basis in fact.

Postmortem studies on the valves did not reveal any pathological damage. The examination did show that the route of traversing from anterior to posterior wall was approximately 10° to the left of the junction of the noncoronary valve cusp with the right coronary valve cusp in all the dogs. This put the route of the traverse approximately across the center of the left coronary cusp on the posterior side of the artery and more or less along a juncture on the anterior side. As indicated above, all ascending traverses were made about three aortic diameters downstream from the valve. Based on some measurements (18), this was barely sufficient for entrance disturbances to be damped; based on entrance flow theory (13, 15), it was far from sufficient.

Another problem which could contribute to the distortion of the measured profiles was the formation of clots of blood on the sensing surface of the probe. When clots formed on this surface, the amplitude and the frequency responses of the probe fell off dramatically, and much lower velocities were indicated. Clearly if this happened in the course of a traverse, the measured profiles would take on a skewed appearance, and the velocity on the side where the traverse was started would be much higher than that on the side where it was terminated. Clotting on the sensor, however, was easily detected by retracing the traversing steps and comparing the amplitudes of the stored signals. The phenomenon was also detectable when the velocity waves were monitored on an oscilloscope, for the rate of clot development, once started, was fast. Where clotting occurred the data were discarded, the probe was withdrawn from the artery, cleaned, and reinserted, and the measurements begun anew.

Clotting was rare with the quartz-coated probes but was a continuous problem in earlier work where the metallic sensor was covered with a silicone grease. Some evidence indicated that even the quartz-coated probes lost their resistance to the clotting mechanism with repeated use. Possibly this was due to a roughening or scratching of the quartz layer with repeated penetrations of the aortic wall. It is well established that such surface blemishes will accelerate the incidence of clot formation. Usually, before insertion, the probe was dipped in heparin to minimize clot formation. Another approach would be to heparinize the whole animal, but this was not done.

It should be noted that the general lack of symmetry and the flatness in the profiles was not consistent with the measurements of Ling et al. (1, 18) but compared with measurements reported by Schultz et al. (3) and Tunstall-Pedoe (4). The same comment could be made about the scatter in the data. Ling et al. carefully selected the beats over which they averaged (a second stationary monitoring probe was employed for this purpose), selecting only those beats "in which the mean deviation of velocity over the beat as well as the variation in the period of the beat were within 5% of the mean." As indicated earlier, the averaging operation was carried out over a sequence of beats in this work. This certainly explains the differences in the scatter of the data between the two works but not necessarily the differences in the shapes of the profiles.

The scatter in the present data resulted from a smoothing of the flow waves, from difficulty in ascertaining the exact position of the probe in the aorta, and from computational errors which largely were due to the
transfer of data from the tapes to the computer. Of these, the second mentioned was perhaps the most serious. Except in a few instances the deviations from the smoothed curves drawn through the data were less than about 3% of full scale. For this reason the curves drawn through the data points in the figures were considered to be justified.

Although the primary objective of this paper was the presentation of measurements of local velocities in the aortas of dogs, it was of interest to compare these measurements, insofar as possible, with theoretical expectations. For the ascending aorta, the measurements were made at a dimensionless distance from the aortic valve (i.e., the entrance) of \( x = 0.01 \). This was well within the entrance region \( (x < 0.16) \), and thus the theories of Chang and Atabek (14) and of Kuchar and Ostrach (15) apply.

Computed velocity profiles were presented by Kuchar and Ostrach (15), for the position \( x = 0.01 \), where \( \alpha = 1 \) and 10 and the magnitude of inlet flow was given by \( U = U_0 (1 + 0.5 \cos t) \). The case for \( \alpha = 4 \) was found elsewhere (14). In this notation \( U_0 \) is the time-mean (uniform) entrance velocity and \( t \) is the dimensionless time \( (\equiv \omega t) \). \( t = 0 \) corresponds to the time of maximum forward flow. In the ascending aortas of the dogs at this position, the observed conditions were, to a first approximation, \( \alpha = 15 \) and \( U = U_0 (1 + 2 \cos t) \). Thus the main difference between the two flows (i.e., with \( \alpha = 10 \) and 15) was that in the aortic flow, \( \alpha = 15 \), the oscillatory component was dominant, whereas in the computed cases, \( \alpha = 10 \), the mean flow was dominant. Although a close match of the two cases, particularly with respect to inlet conditions, is more desirable for comparing theory with observation, the closest available theoretical case is \( \alpha = 10 \).

In the computed case, the velocity profiles were noticeably blunt with large gradients in the velocity confined to the region \( r/R_0 > 0.6 \) for all values of \( t \). Most notable, however, was the absence of flow reversal for all values of \( r/R_0 \) and \( t \). This, of course, was a consequence of the small amplitude chosen for the oscillating component. From the form of the analytical solutions it may be surmised that a larger oscillatory component will produce net flow reversal in the core of the tube when \( t > \pi \) and local flow reversals near the wall even when \( t < \pi \), that is, even when the net flow is still forward. Finally, by comparing the computed cases for \( \alpha = 1, 4, \) and 10 (14, 15), it is evident that the profiles become blunter as \( \alpha \) increases.

Insofar then as the profiles contained in Figures 4-7 are quite flat in the core region over most of the flow time, the experimental results are consistent with entrance flow theory. The absence of measured reverse flow near the wall for all times must be construed as a serious discrepancy between theory and experiment. However, others have noted this discrepancy for flows in flexible tubes far from any entrance effects (18).

Ignoring for the moment that the aorta is curved, that there are branches in the aorta which drain off significant amounts of the flow, and that the aorta is tapered, the flow in the descending aorta is still well within the theoretical entrance length. Thus the profiles in this region can be expected to have the form they do in the ascending aorta except perhaps for a little more rounding. This seems to be the case with the data contained in Figures 8 and 9. Nevertheless, the fact is that curvature and branching do play a role in shaping the profiles in the descending aorta and this cannot be ignored. The measurements of Ling et al. (18) on pulsatile flows in elastic tubes show that branch flows greatly disturb the main flow, and there is some analytical indication that tapering the tubes shortens the entrance length. Much more work is needed in these areas, however, before their effect will be known with any reasonable degree of certainty.

The absence of flow reversal in the vicinity of the tube walls in this work cannot, of course, be given undue weight because of the size of the probe. Its diameter was about 1.5 mm compared with an average diameter of 14 mm in the ascending and 10 mm in the descending aorta. Clearly this was too large.
for precise measurement near the wall. In fact it was for this reason that the curves in the figures were drawn with dashed lines in this region.

Finally, the measured profiles have been compared in each instance with theoretical results based on a single harmonic wave; it might easily be argued that such a comparison was at least inaccurate if not incorrect. In principle, the real flow could be formulated as a harmonic series and the equations of motion solved for each harmonic component. Since the theories are linear the total solution is just the sum of the individual solutions. A full formulation is, however, a most formidable problem. Preliminary estimates using the computer indicated that four to ten terms were needed in the series to fit just the center-line velocity wave over just the period of systole. Many more certainly would be required to fit the full cardiac cycle. It is easy to see that harmonic analysis would generate a series of a parameters of ever-increasing magnitude which, if the coefficients of the Fourier components were all positive, would tend to further flatten the profiles, thereby giving better agreement with the measurements. However, many of the coefficients will be negative, so the final result is not intuitively obvious. Such a calculation requires excessive computer time and probably is not justifiable in light of preceding discussions.

Figure 10 contains two velocity waves, one measured along the center line in the ascending aorta and the other along the center line in the descending aorta. The wave for the ascending aorta showed higher accelerations and decelerations during systole than did the wave for the descending aorta, and the peak velocity in the ascending aorta was greater (~60 cm/sec) than that observed in the descending aorta (~40 cm/sec). Both observations were predicted using a model flow (19). Ling et al. (18), on the other hand, found that the center-line velocity increased with distance from the aortic valve in one dog. The discrepancy may be attributed to the geometry of the respective arteries, to the differences in the amount of blood leaving the main artery through the various branches between the two measuring stations, or to the rearrangement of the velocity profile. Certainly, if the outlets were closed, the center-line velocity would increase because of the narrowing of the trunk artery, assuming, of course, that the impedance of the arterial tree did change drastically and that the shapes of the profiles were not altered drastically. A small shift in the wavelength was noted between the waves in the ascending and the descending aorta. Since the same shift was observed in the ECG, the shift was not considered significant.

As indicated earlier, cardiac output was measured using the hydrogen-infusion technique (17) and instantaneous volumetric flow was measured in selected cases with an electromagnetic flowmeter. In six cases the flow rates computed from the hot-film anemometer data ranged from 4 to 19% (mean 11%) lower than the results obtained by the hydrogen-infusion technique and, where the profiles were symmetrical, the results were about the same for the electromagnetic flowmeter. This gives some indication of the absolute velocities. Considering the extensive integrations (time and space) required to obtain cardiac output from the hot-film data, the agreement is reasonable. It should be recognized, however, that the hot-film anemometer is more suitable for localized measurements than for such gross measurements as volumetric flow rates, and its use for the latter measurements is not recommended.

It is tempting to infer the shear stress at the wall from the slopes of the velocity profiles. This should not be done, however, because the data points nearest the wall are of limited accuracy. These points have not been corrected for the proximity of the wall to the probe. Because of the relatively large size of the probe, 1.5 mm in diameter, this effect could be significant. Further, the sensing element, i.e., the metallic film, was wrapped around the entire probe tip. Near the walls the velocity gradients were steep, and the probe necessarily averaged. Even when the probe touched the wall, zero flow was seldom indicated because one side remained in the
flow field. Away from the walls, however, the velocity gradients were less severe and this type of error was small.

Based on the measurements reported in this paper, it seems reasonable to conclude that velocity profiles in the aorta of dogs are generally more blunt than is predicted by steady oscillatory flow theory. In fact the shapes of the profiles in both the ascending and descending aorta, at least during early systole, are more consistent with the theory for a flow started impulsively from rest. This should not be too surprising since the blood is at rest for about a third of the cardiac cycle just prior to the onset of systole.

Although the profiles are more or less blunt, deviations are common, and each dog appears to have a distinctive set of them. The variations in the profiles probably are a consequence of geometrical factors, for example, valve movement and configuration in the ascending aorta and the branching in the descending aorta.

In some dogs considerable asymmetry in the flow was observed. This has an important clinical implication especially in the area of electromagnetic flow measurement where cardiac output can be misleading when the flow is significantly asymmetrical (20).

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References

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