Hemodynamic Effects of Acute Experimental Aortic Coarctation in the Dog

By David L. Newman, Robin K. Walesby, and Nelson L. R. Bowden

ABSTRACT
Pressure and flow pulses were recorded immediately proximal and distal to a temporary stenosis of the thoracic aorta of the dog. The severity of the stenosis was varied, and the magnitude of the pulse changes was correlated with the degree of stenosis expressed as the area ratio, $A'/A''$, where $A'$ is the cross-sectional area of the stenotic section and $A''$ that of the normal vessel. The harmonic amplitudes of the pressure and flow waves during the stenotic period were obtained by Fourier analysis and normalized to their values when the stenosis was released. It was found that as $A'/A'' \to 0$ the proximal pressure amplitude increased and the distal pressure and the proximal and distal flow amplitudes decreased. These changes were such that the proximal fluid impedance calculated from the corresponding pressure and flow amplitude ratios increased as $A'/A'' \to 0$; the distal impedance remained relatively unaffected. These findings can be interpreted in terms of reflection of the pulse waves at the stenosis. The normalized proximal pressure amplitudes approximate to $1 + R$ and the distal pressure and both the proximal and the distal flow vary according to $1 - R$, where $R$ is the reflection coefficient. The variation of $R$ with $A'/A''$ can be determined from the proximal fluid impedance changes; we found that $R$ increased relatively slowly for moderate stenoses, varying from 0 at $A'/A'' = 1.0$ (no stenosis) to $\sim 0.2$ at $A'/A'' = 0.2$. At more severe stenoses, it increased more rapidly, approaching 1.0 at $A'/A'' = 0$. The generation of turbulence was most marked at $A'/A'' \sim 0.2$.

KEY WORDS: blood pressure, turbulence, aortic stenosis, pulse wave reflections, blood flow, fluid impedance, reflection coefficient

A number of authors have investigated the effects of experimental arterial stenoses in attempts to elucidate the underlying mechanisms of diseases such as atherosclerosis and aortic coarctation. These studies have contributed a great deal to an understanding of the hemodynamic effects of stenoses, but they have not provided adequate information regarding the changes in both pressure and flow immediately proximal and distal to the stenosis. For example, May et al. (1) studied the effect of common iliac stenosis on both pressure and flow but were concerned only with changes in the mean values of these parameters. Shenk et al. (2) recorded both pressure and flow after experimental coarctation of the thoracic aorta produced a few months earlier. However, the observations that they made were qualitative and were not correlated to the severity of the stenosis. Recently, using a test line of distensible tubing, Kim and Corcoran (3) investigated the effect of various insert sizes on the amplitude ratio and the phase difference of the distal and proximal oscillatory pressure components. Linear and nonlinear models were used to analyze the results, and it was concluded that the linear model was applicable only to small obstructions at low flow rates for which the flow is laminar.

It has been shown previously (4–6) that valuable quantitative information can be obtained by momentarily occluding a vessel and comparing the oscillatory amplitudes of pressure and flow during the occlusion with the normal amplitudes immediately prior to the occlusion. By this means information can be obtained concerning the immediate hemodynamic effects of the occlusion without the complication of compensatory mechanisms such as collateral development and vasoconstriction which alleviate to some degree the effect of the occlusion. The oscillatory components of pressure and flow are of particular interest in this type of study, because the amplitude ratios and the phase difference of their Fourier components enable calculation of the fluid impedance which, apart from the importance of its analogous relationship to electrical impedance, leads directly to values for reflection coefficients from the occlusion (6).

The purpose of this paper is to describe the
changes in pressure, flow, and impedance proximal and distal to acute stenoses of various degrees of severity applied to the thoracic aorta of the dog.

Methods

To investigate the hemodynamic effects of acute aortic stenoses, a series of measurements was carried out on eight greyhounds weighing between 28 and 34 kg. The stenosis-producing device was similar to that used by Fineburg and Wiggers (7), it consisted of a braided nylon tape 0.35 cm in width, which was looped around the vessel. With one end of the tape clamped, the other end was attached to a piston so that the loop could be gradually tightened to various degrees of severity. A schematic diagram and a radiograph of the device partially occluding an excised thoracic aorta are shown in Figure 1.

The site chosen for the stenosis was the descending thoracic aorta distal to the left subclavian artery, a common site for postductal coarctation in humans. The dogs were anesthetized with acetylpromazine (0.1 mg/kg body weight, iv) and subsequently maintained with a halothane-oxygen mixture (~1%) using a positive-pressure respirator. The thoracic aorta was exposed by a left thoracotomy through the fourth intercostal space. Proximal and distal pressure and flow velocity recordings were made as close as possible to the stenosis site following ligation of adjacent intercostal space.

The pressure measurements were made by direct puncture of the vessel with 21-gauge hypodermic needles connected by saline-filled catheters to semiconductor strain-gauge pressure transducers. The two pressure systems were calibrated before and after each experiment. The dynamic amplitude response was investigated with a sine-wave pressure generator and found to be flat to at least 30 Hz with linear phase shift with frequency. The flow recordings above and below the stenosis were made with electromagnetic cuff probes (SE Labs SEM 230) of a size to fit snugly around the vessel. The probes were connected to a two-unit flowmeter system (SE Labs SEM 275). The manufacturer’s calibration figures were used for the flowmeters, since they have been found to be correct to ± 5% (6). To obtain an adequate frequency response, a bandwidth setting of 40 Hz was used; this setting provided a flat frequency response to 70% of that figure, i.e., 28 Hz. The phase relationship between the pressure and flow channels was compared by a previously described method using a viscous gelatin-saline mixture (6). The outputs from the pressure transducers and the flowmeters were connected to a suitable chart recorder so that recordings could be made in the normal vessel and in vessels with stenoses of various degrees of severity. The experimental procedure was as follows. First, recordings were made in the normal vessel. Then the nylon loop was tightened until the required stenosis was established, and recordings were made for approximately 5-10 heart beats. Next, the catheters on the pressure lines were interchanged, and the stenosis recordings were repeated. The total stenosis time did not usually exceed 20 seconds. Finally, the nylon loop was loosened, and an additional set of normal recordings was obtained.

This procedure allowed us to establish that the condition of the dog was stable throughout the experiment and to reduce possible errors in the pressure recording. The pressure and flow wave forms were subjected to Fourier analysis to obtain the amplitude and the phase of the first ten harmonics by sampling pressure and velocity values over four successive pulses at a rate of approximately 50 sample points/pulse. The sampling was with an electromagnetic follower (d-Mac Ltd.) interfaced to an IBM 029 card punch. The analysis computations were carried out on an IBM 360 computer.

The degree of stenosis was expressed as the area ratio, $A' / A^*$, where $A'$ is the internal cross-sectional area of the vessel in the stenotic region and $A^*$ is the normal cross-sectional area. The area ratio was obtained by a postmortem radiographic method. The thoracic aorta with the stenosis device left in place was excised and restretched to its in vivo length. The vessel was inflated to the in vivo mean blood pressure with radiopaque dye, and a series of radiographs was taken for the normal and the stenosed conditions. Assuming circular cross sections, the area ratio was determined from these radiographs by measuring the appropriate diameters with vernier calipers. The area ratios obtained by this postmortem radiographic technique agreed with in vivo radiographic measurements made on a few selected dogs following the injection of radiopaque dye into the aortic arch.

Results

To demonstrate the effects of a gradually increasing stenosis on the proximal and the distal pressure and flow through the vessel, the nylon loop was gradually tightened in one of the dogs. The recordings are shown in Figure 2. Marks
Continuous recordings of pressure and flow wave forms with gradually increasing stenosis (A'/A" — 0). After momentary complete occlusion (A'/A" — 0), the nylon loop was gradually released (A'/A" — 1.0).

Changes in the mean levels of pressure or flow did not take place until an area ratio of less than 0.1 (> 90% occlusion) was reached. Up to this degree of stenosis, the respective increase and decrease in the proximal and the distal pulse pressure took place only gradually. A similar gradual reduction occurred for the oscillatory flow through the stenosis until an area ratio of less than 0.1 was reached, at which point the flow diminished more abruptly.

An ultraviolet recorder (SE Labs SE 2009) was used to permit greater recording sensitivities and faster run-out speeds than the pen recorder used for Figure 2 so that we could look in more detail at the changes in the pulsatile components of the pressure and flow wave forms which occurred at stenoses less severe than those which provided major changes in mean levels. A typical set of wave forms from one dog is shown in Figure 3. Both the proximal and distal flow wave forms (Vt and Vd, respectively) were progressively reduced in amplitude with increasing stenosis. An increase in the proximal pulse pressure (Pt) and a corresponding decrease in the distal pulse pressure (Pd) also occurred. No significant changes occurred in the pressure and flow wave forms until an area ratio of 0.4 was reached. It should also be noted that the heart rate remained constant throughout these measurements. To examine these changes quantitatively, the harmonic amplitudes of the pressure and flow wave forms were obtained using Fourier analysis. Since the absolute values of pressure and flow differed slightly from dog to dog, the changes in the pressure and flow amplitudes are expressed in terms of a ratio of the amplitude during the stenosis to that during the normal unstenosed state. Thus, if the proximal and distal pressure and flow amplitudes during stenosis are given as Pts, Pds, Vts, and Vds and the corresponding amplitudes during the normal condition as Ptn, Pdn, Vtn, and Vdn, then the ratios of interest are Pts/Ptn, Pds/Pdn, Vts/Vtn, and Vds/Vdn.

The frequency dependence of the pressure and flow ratios was not marked, as is shown in Figure 4 in which the ratios are plotted against harmonic number for three dogs in which A'/A" = 0.1. The frequency independence of these ratios permits calculation of mean values over the first six harmonics; these values are plotted against the corresponding area ratios in Figure 5. The analysis was not extended beyond the sixth harmonic, because the amplitudes of higher harmonics corresponded to chart deflections of less than 0.1 cm. The proximal and distal pressure ratios, Pts/Ptn and Pds/Pdn, showed symmetrical variation about the normal value. In the case of the flow ratios, the variation of Vts/Vtn was similar to that of Vds/Vdn and also to that of the distal pressure ratio Pds/Pdn.

Oscillatory pressure (Pt and Pd, respectively) and flow (Vt and Vd, respectively) wave forms proximal and distal to stenoses of four degrees of severity. The normal unstenosed condition is represented as A'/A" = 1.0.
The presence of a stenosis increased the proximal impedance modulus from a normal value of $Z_0$ to $Z^*_0$, but the distal impedance modulus was relatively unaffected, i.e., $Z^*_D \sim Z_D$. Figure 6 illustrates these findings by showing the variation with the area ratio of the impedance modulus ratios $Z^*_0/Z_0^n$ and $Z^*_D/Z_D^n$ (these values are the mean of the first six harmonics). It should be noted that marked changes in the proximal impedance modulus did not take place until an area ratio of 0.2 was reached. Below this area ratio, it increased rapidly, approaching infinity at $A^*/A^n = 0$, since there is then zero oscillatory flow at the stenosis site. The distal impedance modulus on the other hand remained relatively unaffected by increasing stenoses, although it did appear to increase slightly at area ratios of less than 0.1.

The impedance phase angles at both sites were unchanged until an area ratio of 0.1 was reached. Below this value, there were some variable and modest changes at the proximal site.

**Discussion**

The major contribution of previous in vivo experimental coarctation studies (1, 8) has been the observation that severe stenoses are required before there is a significant change in the mean pressure or flow. At the so-called critical stenosis there is a rapid fall in the distal pressure and flow. Kindt and Youmans (8) have shown that the
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Area Ratio , A'/A

0.2 0.4 0.6 0.8 1.0

Impedance modulus ratios (mean values over six harmonics) proximal (Z_v/\bar{Z}_v) and distal (Z_n/\bar{Z}_n) to stenoses of various area ratios. Solid and open symbols refer to the proximal and distal sites, respectively.

Critical stenosis depends to a limited extent on the length of the stenosis, and for a stenosis similar in length to that used in the present study the critical stenosis for mean pressure and flow was found to lie between area ratios of 0.20 and 0.10. Figure 2 indicates that in the present study dramatic changes in the proximal and distal mean pressures and flows did not occur until area ratios of less than 0.10 were reached. Changes in the mean levels were not reported in detail in the present study because of the difficulty in obtaining accurate quantitative information about these parameters, particularly in the case of flow for which location of an accurate zero level is difficult. From the results presented, it is apparent that changes in oscillatory pressure and flow and fluid impedance began to occur at an area ratio of about 0.40, suggesting that the oscillatory components of pressure and flow (or flow velocity) are a more sensitive index to the effect of stenosis than are the mean values.

The variation with area ratio of the oscillatory pressure and flow components is of particular interest because of the similarity between the variation in the distal pressure ratio and those in the proximal and distal flow ratios. In addition, these variations are the symmetrical inverse of the variation in the proximal pressure ratio (Fig. 6). A quantitative interpretation of the results can be made in terms of reflections due to the abrupt reduction in lumen size which typifies the stenosis.

In the case of the normal unstenosed vessel, the incident pressure wave can be written as:

\[ P_i = A_i \exp(\imath t - z/c). \]

If the vessel is stenosed, then the reflected wave from the stenosis can be written as:

\[ P_r = A_r \exp(\imath t + z/c), \]

where \( A_i \) and \( A_r \) are the coefficients of the incident and reflected waves, respectively, \( \imath = \sqrt{-1} \), \( n \) = angular frequency, \( t \) = time, \( z \) = axial coordinate, and \( c \) = propagation velocity.

The resultant pressure wave immediately proximal to the reflection site \((z = 0)\) is, therefore,

\[ (P_i + P_r)/P_i = (A_i + A_r)/A_i. \]

We can define a reflection coefficient, \( R \), as:

\[ R = A_r/A_i, \]

and hence, since the proximal measurements were made close in the stenosis,

\[ (P_i + P_r)/P_i = P_r/P_i = 1 + R. \]

In general therefore, the presence of the stenosis leads to the production of a partial standing wave with an antinode of pressure immediately proximal to the stenosis. This analysis assumes that there are (1) no substantial reflections from other parts of the arterial net, particularly the peripheral beds, and (2) no significant effects due to reflection at the heart of backward-going waves from the stenosis. It has been suggested (9, 10) that waves at frequencies \( \geq 2 \) Hz reflected from the peripheral beds are not normally found in the upper aorta, since apart from attenuation processes they are subject to high reflection at junctions on their return trip toward the heart and also tend to be randomly distributed in phase. The experimental data in this study support this conclusion, since, apart from the fact that peripheral reflection was probably low because of the vasodilatory effects of halothane, it was found that \( P_i/P_c \to 2.0 \) for near total occlusion conditions. It has been shown in a previous study (4) that such a finding indicates the
absence of peripheral reflections. In addition, the impedance modulus, $Z$, of the normal unstenosed vessel was largely independent of harmonic number (Fig. 7), representing a frequency range of 2.1 Hz to 15.0 Hz. The spectrum is similar to that found by other workers (11, 12) and indicates, as does the similar trend in propagation velocity found by Nichols and McDonald (13), that the upper aorta is relatively free of peripheral reflections at frequencies greater than ~ 2 Hz.

It is more difficult to assess the effect of reflection at the heart. However, in one of the dogs, the stenosis site was moved to the abdominal aorta to increase the stenosis-heart distance. Results substantially similar to those described previously in this paper were obtained, indicating that reflections at the heart did not contribute significantly to the changes observed for the thoracic aorta site.

The stenosis-heart-stenosis distance was ~30 cm and, although some attenuation does occur over this distance particularly for the higher frequency components, it is unlikely that this explanation is sufficient. The heart might only partially reflect the waves and in addition the branches between the stenosis and the heart might also play an important role in diminishing the effect of reflections at the heart.

In the case of the transmitted pressure wave, there will be a corresponding reduction in amplitude of $1 - R$ compared with the normal value; thus,

$$P_D^i/P_D^o = 1 - R.$$  (7)

The reflection coefficient for the flow pulses will be equal to $-R$, there will be a similar partial standing wave system but with a node at both proximal and distal sites, and the amplitude of the flow pulses will be decreased by $1 - R$; thus,

$$V_U^i/V_U^O = V_D^i/V_D^o = 1 - R.$$  (8)

The dependence of the pressure and flow ratios on the value of $R$ as indicated in Eqs. 7–9 leads to the symmetry of their variation with $A^1/A^o$ shown in Figure 5. Thus, if $R = 0.4$,

$$P_U^i/P_U^O = 1.4$$

and

$$P_D^i/P_D^o = V_U^i/V_U^O = V_D^i/V_D^o = 0.6,$

and so on.

The experimental values of $R$ were obtained from the well-known equation:

$$R = (Z_U^i - Z_U^o)/(Z_U^i + Z_U^o).$$  (9)

$R$ can therefore be calculated from the impedance data; these experimental values are shown in Figure 8 together with the predicted curve of $R$ derived from the infinite line model of Womersley (14). It appears that the increase in the proximal impedance modulus was almost entirely due to the increased reflections from the stenosis rather than to any decrease in compliance that might have
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resulted from the slight changes in mean pressure which occurred over the range of partial occlusions used.

The impedance distal to the stenosis remained unaffected by the severity of the stenosis, i.e., \( Z_p/Z_n \sim 1.0 \) (Fig. 6), since there was a similar reduction in both \( P_p/P_n \) and \( V_p/V_n \). This fact confirms the expected result that the distal impedance value is determined by the cardiovascular state of the arterial system below the stenosis. This situation does not change significantly during the course of the experiment provided that stenoses are not maintained for long periods, i.e., > 20 seconds. There is a suggestion in Figure 6 that the distal impedance increased slightly at very low area ratios, < 0.1; this phenomenon may be the result of peripheral changes due to the sudden drop in mean pressure, the effects of turbulence, or both. The pressure effect is unlikely, since a pressure reduction would be expected to cause a reduction in impedance. Some degree of turbulence is indicated in Figure 4; the distal flow trace \( (V_p) \) became irregular compared with the continued smoothness of \( V_n \) at low area ratios. Turbulent flow is just possible in the unstenosed condition of these experiments but probably is not significant since it would occur only at the very peak of systole. However, as the area ratio decreases and assuming for the moment that \( V_n \) remains constant, turbulence would increase significantly since \( Re > Re_c \), the critical Reynolds number. In practice however, turbulence is made less likely since the flow velocities are reduced by the stenosis. Calculating the true \( Re \) using the velocity reduction factors obtained from the experiment, it can be seen (Fig. 9) that \( Re \) does not increase as dramatically as would be expected, barely reaching a value of 4,000 at the peak of systole. Indeed, for a severe stenosis, \( Re \) may actually decrease making turbulence less likely. This finding is in agreement with those of previous workers (15, 16) who have found a reduction in systolic murmur with severe stenosis.

It is important to note that the acute hemodynamic effects described in this paper are those occurring immediately adjacent to the stenotic site. A modification of the pressure and flow amplitude changes found would obviously occur if measurements were made further from the stenotic discontinuity or if the stenosis were chronic. In addition, other factors such as the influence of the vascular state of the animal, the length of the stenosis, and the location of the stenosis also require additional investigation.

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D L Newman, R K Walesby and N L Bowden

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