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The Relationship between Ultrasonic Pulsatility Index and Proximal Arterial Stenosis in a Canine Model


SUMMARY Although the ultrasonic pulsatility index (PI) is used as a test of arterial inflow, the relationship between proximal arterial stenosis and PI has not been fully evaluated. In a series of experiments on dogs, over 200 measurements of pressure, flow (using an indwelling electromagnetic flowmeter), and PI were made distal to implanted arterial stenoses of varying length (0.5-9 mm) and cross-sectional area (36-94% reduction in area). It was shown that, whereas the reduction of PI correlates broadly with stenosis severity, the scatter of results is wide for all but the tightest stenoses. This scatter is due at least in part to the variability of the vascular bed distal to the site of measurement. Circ Res 46: 470-475, 1980

ULTRASONIC Doppler blood velocimeters are now in wide use for the assessment of peripheral vascular disease. Their applications include use as simple flow detectors for measuring blood pressure in conjunction with occlusive cuffs, use as flowmeters, and use as the basis of vessel imaging devices. It is with their use as flowmeters, or more strictly as velocity-measuring devices, that this paper is concerned. In this application, use is made of the fact that the Doppler shift on a beam of ultrasound scattered from the blood corpuscles within a blood vessel is directly proportional to their velocity. Be-
cause neither the angle between the blood vessel of interest and the Doppler probe nor the actual size of the vessel is known accurately, thus making it difficult to convert blood velocity to volumetric flow, it has become normal practice to analyze the shape of the blood velocity waveform rather than to attempt to derive flow in absolute terms.

Blood velocity waveforms recorded from sites distal to severe narrowings are in general more damped than those recorded from undiseased arteries, and this has led to the introduction of various indices which purport to quantify these changes. The most widespread techniques are those based on the pulsatility index (PI) introduced by Gosling and co-workers (1971). This index was originally defined as the sum of the energies in the first and subsequent Fourier harmonics of a velocity waveform, divided by the energy in the zeroth harmonic; however this has been superseded by a similar but simpler PI calculated merely by dividing the maximum vertical excursion of the waveform (P to p) by its mean height (Gosling and King, 1974) (Fig. 1).

The PI has been shown to correlate with arteriographic findings and with ankle systolic pressure (Harris et al., 1974; Johnston and Taraschuk, 1976; Fitzgerald and Carr, 1977) and also with the outcome of femoropopliteal saphenous vein bypass grafts (Charlesworth et al., 1975). There has, however, been no systematic evaluation of the technique in an animal model. Whereas such a model cannot hope to simulate the conditions found in a diseased artery, the geometry of any stenoses used can be precisely controlled, enabling quantitative measurements to be made on the relationship between PI and the degree of narrowing of the artery.

**Methods**

The stenoses used for these experiments were designed to allow rapid known changes to be made in the reduction of cross-sectional area. They consisted of five separate parts, two "adaptors," a "stenotic insert," and two short connecting pieces of transparent polythene tubing (Fig. 2). The adaptors were designed to be inserted into the artery at the start of the experiment and remain there while various inserts were fitted between them. The inserts were available in eight diameters (ranging from no reduction in area to 94% area reduction) and in four lengths, (0.5, 3.0, 6.0, and 9.0 mm) and were held in place by means of the tubing, it being transparent to aid the correct approximation of the adaptors and inserts. This "interchangeable stenosis" technique combined the advantages of the external constriction technique (Mann et al., 1938; May et al., 1963) in that it was possible to examine the effects of more than one stenosis in each dog, and the implanted stenosis technique (Mann et al., 1938; Young et al., 1975) in that the dimensions of each stenosis were precisely known. This last point is very important since mild arterial stenoses have little effect on blood flow, and yet it is difficult to estimate accurately the degree of stenosis produced by a tight external constriction.

Experiments were performed on a total of 10 greyhounds. In each case the dog was starved for 24 hours preoperatively and received premedication of 2 mg of acepromazine 1 hour prior to surgery. Anesthesia was induced with thiopental and maintained with oxygen, nitrous oxide, and pentobarbital. Using electrocautery, a midline abdominal incision was made and extended across the right inguinal ligament to the thigh. The abdominal aortic trifurcation was exposed and the right iliac and femoral arteries displayed.

The size of the stenosis assembly required was estimated by measuring the circumference of the proximal third of the external iliac artery in which the stenosis was to be inserted. Assemblies of only two sizes were found necessary, those having unstressed internal diameters of 4 and 5 mm, respectively.

The dog was heparinized (approximately 200 units/kg) and an 18-gauge side hole cannula was inserted through a branch of the internal iliac artery and advanced so as just to protrude into the aorta, where it was secured by tying a suture round branch and cannula. A similar cannula with an end hole was inserted into the femoral artery via a muscular branch just distal to the site earmarked for the
stenosis. Both cannulas were attached to Elcomatic EM 751 pressure transducers, which were in turn connected to Hewlett-Packard 8805C pressure preamplifiers. All other branches of the iliac and femoral arteries were then ligated.

Two transverse slits 3 cm apart were made in the iliac artery, between the aortic trifurcation and the first branch of the artery. The interchangeable stenosis assembly, containing a non-stenosed middle section, was inserted and secured. The ties encircling the two adaptors were left long and held together with arterial forceps to prevent them springing apart. A Statham cannulating blood flow probe of the appropriate size was similarly inserted distal to the stenosis and the two pressure lines (Fig. 3).

Once the preparation was complete, the dog was allowed to stabilize for a period of at least 30 minutes. The non-stenosed insert was replaced by one of reduced area and the preparation again allowed to stabilize for 10 minutes. Measurements then were made of the pressure proximal and distal to the stenosis, flow through the stenosis, and of the Doppler waveform distal to the stenosis. Between 11 and 25 substitutions of stenosis were made on each dog so that, in the 10 experiments, a total of 211 separate series of measurements were made. The flow and pressure measurements were recorded both on a Brush 260 chart recorder and on a Tanberg 115 FM tape recorder.

The Doppler velocity signal was measured by means of a modified Sonicaid BV380 Doppler unit. The output from this was recorded on a Uhre 4400IC A.M. tape recorder for later use, and analyzed in real time on a modified Honeywell SAI-5IC spectrum analyzer. The output from this was displayed on a Tekronic 604 variable persistence display monitor and visually recorded on a Medelec For-4 fiber optic recorder (Fig. 4). The modification of the spectrum analyzer was to allow a sweep of the first 100 frequency bins every 20 msec, rather than a complete sweep of all 200 frequency bins every 40 msec. Figure 5 shows two typical sonagrams recorded in this way, the first one recorded distal to a mild stenosis, the second distal to a severe one.

Results

The PE of each sonagram was calculated as described in the introductory section, and the results are presented plotted against percentage stenosis in Figures 6 and 7. The results from the 3-, 6-, and 9-mm lengths have been plotted on the same diagram as they were almost identical. The results from the 0.5-mm stenosis, although similar, were more scattered, and have therefore been plotted separately. In both cases the PE's distal to tight stenoses are low, whereas those distal to moderate stenoses take on a wide range of values, suggesting perhaps that PE is dependent on some factor other than proximal stenosis.

One obvious possibility is peripheral resistance, and indeed it is well known that PE falls following exercise, at a time when peripheral resistance is low. To investigate the effect of peripheral resistance on PE, the stenoses were divided into groups of varying severity and the PE's of each group plotted against peripheral resistance. Unfortunately, it was impossible to measure this directly, and so its approximate value was calculated from the “apparent peripheral resistance” found by di-
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FIGURE 5  Sonagrams recorded distal to mild (A) and severe (B) arterial stenoses.

viding the mean pressure distal to each stenosis by the mean flow through the stenosis. True peripheral resistance is given by the mean difference between arterial pressure and venous pressure, divided by total limb blood flow. Venous pressure normally can be disregarded but, unfortunately, the flow through the stenosis does not correspond to total limb blood flow, there being an additional component \( Q_c \) due to flow through collateral vessels. Under physiological conditions, the contribution of \( Q_c \) is usually ignored, and this results in a slightly inaccurate quantity which will be termed apparent peripheral resistance \( R_a \).

The size of the error introduced by measuring the flow through the stenosis \( Q_s \), rather than the total limb flow \( Q_t \) is simply calculated. The flow through the stenosis may be written

\[
Q_s = Q_c R_s + R_s R_c
\]

where \( R_c \) is the collateral resistance and \( R_s \) the resistance of the stenosis, and therefore the apparent peripheral resistance \( R_a \) may be written

\[
R_a = R_p (R_s + R_c)/R_c
\]

where \( R_p \) is the true peripheral resistance.

If \( R_c \) is large compared with \( R_s \), then \( R_a \) and \( R_p \) are similar. Although under physiological conditions \( R_c \) is the resistance of the main artery and can be ignored, when the artery is stenosed the distinction between apparent peripheral resistance and peripheral resistance cannot be disregarded.

The values of apparent peripheral resistances found from the dog experiments were converted into corrected readings using Equation 2. The resistance of each stenosis was found by dividing the pressure difference across it by the flow through it, while the collateral resistance of each limb was estimated from the pressure-flow relationship in the presence of different stenoses under standardized conditions of hyperemia (see Appendix).

Figures 8 and 9 are plots of PI against corrected peripheral resistance for stenoses of 50–80% area reduction, and for stenoses of greater than 88%, respectively. It can be seen that in each case PI is
dependent on peripheral resistance, but that this is more pronounced for the less severe stenoses.

**Discussion**

The results presented in the previous section indicate, perhaps not surprisingly, that PI is dependent on the vascular beds both proximal and distal to the site of measurement. It appears that, although PI is dependent on peripheral resistance for a given proximal stenosis, this dependency is much more marked when the proximal stenosis is mild, and this explains why tight proximal stenoses always gave rise to a low PI while a wide range of values were found distal to mild stenoses.

In this animal model, very severe stenoses (90% for the 0.5-mm stenoses and 86% for the longer stenoses) were required to decrease PI by a diagnostically significant amount. However, these measurements were made under operative conditions and, whereas every effort was made to preserve stable preparation, both the anesthetic and surgery itself may interfere with the normal functioning of the peripheral bed. As will be seen from Figures 8 and 9, the separation between the results obtained from different stenoses is better in the presence of higher peripheral resistances, which may well pertain in the undisturbed circulation. Furthermore it might be argued that the peripheral resistances of patients with diseased limbs tend to be less than those with normal limbs (Strandess and Sumner 1975), thus enhancing the differences between the PI's of the two groups. However, the position is far from clear, as there is little doubt that the peripheral resistance of patients undergoing Doppler tests will vary widely depending on such factors as the ambient temperature and how recently they have walked or smoked cigarettes. Even more important is that the resistance of the vessels distal to the site of measurement is likely to be modified by the disease process itself.

Since the normal PI of the blood flow in the hindlimb of a greyhound is different from (less than) that of healthy human subjects, the absolute figures from the experiments described here cannot be related directly to figures obtained from diseased human arteries. However, it is a reasonable assumption that the changes produced by narrowings in the arteries proximal to the site of measurement will affect PI in broadly the same manner.

In conclusion, although PI has been advocated as a measure of arterial inflow, it is in fact influenced by both inflow and peripheral resistance. Since peripheral resistance is governed by many factors, PI alone cannot be relied on to distinguish between stenoses of different severity. In this animal model, only a very tight stenosis (>86%) produced a PI low enough to be diagnostic of inadequate inflow.

**Appendix 1**

**The estimation of collateral resistance**

As there is no simple way to measure collateral resistance, it was in each case calculated from measurements of the pressure-flow relationship of the limb under different conditions. To make this calculation, it was necessary to assume that the peripheral resistance always dropped to a similar value ($R_x$) in any one dog following the period of arterial occlusion required to change the arterial stenosis (that is to say, the response of the peripheral bed to a stenosis of infinite resistance was always similar). The validity of this assumption was confirmed by the consistency of the collateral resistance calculations.

The true value of $R_x$ for each dog was calculated by averaging all the values of apparent peripheral resistance obtained during the peak of hyperemia.
following the insertion of a mild stenosis. This is valid since peripheral resistance and apparent peripheral resistance are almost identical when the resistance of the stenosis is low (see Eq. 2).

Since the real value of $R_x$ for each dog was known, it was then possible to calculate the value of the collateral resistance from the apparent values of $R_x$ in the presence of various severe stenoses. Rearranging Equation 2:

$$R_c = \frac{R_p R_s}{R_a - R_p}.$$  

Under these conditions, $R_p$ may be replaced by $R_x$, $R_s$ by $\Delta P/Q_s$ where $\Delta P$ is the mean pressure drop across the stenosis, and $R_c$ is of course the apparent value of $R_x$. An example of the calculation for one dog is given below:

| Stage One – Estimation of true value of $R_x$ |
|-----------------|-----------------|-----------------|
| % Stenosis | $R_x$ | $R_a$ |
| 0 | 0.004 | 0.14 |
| 36 | 0.006 | 0.18 |
| 52 | 0.009 | 0.15 |
| 68 | 0.016 | 0.14 |

In each of these cases, $R_c$ can be regarded as small compared with $R_a$ (see final answer), and thus the value $R_x$ measured is the correct value (mean 0.15).

| Stage Two – Estimation of $R_c$ using $R_x$ |
|-----------------|-----------------|-----------------|
| % Stenosis | $R_x$ | $R_a$ | $R_c$ |
| 85 | 0.122 | 0.19 | 0.46 |
| 92 | 0.255 | 0.21 | 0.64 |
| 94 | 0.446 | 0.30 | 0.45 |

In these cases, $R_x$ is significant and thus modifies the measurement, and $R_c$ can be calculated from the amount by which $R_a$ differs from $R_x$. In this example, $R_c$ has a mean value of 0.52 and this value has been used to correct the measured values of peripheral resistance in the following tabulation:

<table>
<thead>
<tr>
<th>% Stenosis</th>
<th>$R_x$</th>
<th>$R_a$</th>
<th>$R_p$</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>—</td>
<td>0.603</td>
<td>0.603</td>
</tr>
<tr>
<td>36</td>
<td>0.004</td>
<td>0.618</td>
<td>0.613</td>
</tr>
<tr>
<td>52</td>
<td>0.005</td>
<td>0.574</td>
<td>0.569</td>
</tr>
<tr>
<td>68</td>
<td>0.010</td>
<td>0.584</td>
<td>0.573</td>
</tr>
<tr>
<td>78</td>
<td>0.024</td>
<td>0.684</td>
<td>0.654</td>
</tr>
</tbody>
</table>

The following points are worth noting. (1) The values of $R_p$ are lower than in the previous tabulations because of the nonlinear relationship between flow through and pressure drop across a stenosis. (2) Only when the stenosis exceeds 80% is the difference between $R_x$ and $R_p$ of consequence. (3) The resistance of the peripheral bed appears to rise as proximal stenosis severity increases, unless an allowance is made for collateral flow.

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