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PULSE WAVE REFLECTIONS FROM ARTERIAL DISCONTINUITIES

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Summary

The true or characteristic velocity of pulse waves along different arterial segments in both animals and man often permits the determination of the elastic state of the vessels in health and disease. In this study, the extent to which the measured, or apparent pulse wave velocity (PWV) suffers from reflection effects is investigated. An *in vitro* test-rig is used to demonstrate that PWV, measured in an elastic tube model, will oscillate about the value characteristic of the vessel wall, as distance progresses close to a discontinuity-generating reflection. These variations are shown to be markedly reduced if the segment length is longer than $\lambda/4$ for the dominant harmonics in the pulse waves. Harmonic analysis of the pulse waves recorded from the elastic tube and from two human volunteers also show that phase velocities oscillate with increasing amplitude as reflection increases.

Résumé

La vraie vitesse ou la vitesse caracteristique des ondes du pouls le long des différents segments des artères tant chez les animaux que chez les humains permet souvent de déterminer l'état élastique des vaisseaux sanguins dans la santé ainsi que dans la maladie. Dans cette étude, nous avons cherché à savoir jusqu'à quel point la vitesse mesurée ou apparente des ondes du pouls est affectée par les effets de la réflexion. On s'est servi d'un appareil à tester *in vitro* pour démontrer que la vitesse mesurée des ondes du pouls dans un modèle de tube élastique oscille à peu près jusqu'à la valeur caractéristique de la paroi du vaisseau à mesure que la distance progresse en se rapprochant des réflexions qui engendrent la discontinuité. On voit que ces variations sont considérablement réduites si la longueur du segment depasse $\lambda/4$ pour l'harmonique dominante dans les ondes du pouls. L'analyse harmonique des ondes du pouls faite avec une tube élastique et avec deux personnes volontaires montre aussi que les vitesses des phases oscillent avec une amplitude croissante suivant la montée de la réflexion.

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Introduction

The arterial system consists of distensible and progressively branching visco-elastic tubes. It stems from the root of the aorta where there is a non-return valve separating it from the left ventricle of the heart. Blood is intermittently, but regularly, forced through this valve by ventricular contraction and each ejection generates pressure and flow waves which travel through the system. Because the elastic modulus of blood is several orders of magnitude higher than the volume elastic modulus of the arterial wall (10^9 cf 10^5 Nm^{-2}), these pressure/flow waves propagate with a finite velocity largely dependent on the vessel wall. Investigations in both animals and man have shown that as the waves travel towards the periphery, they change shape due to vascular reflection and damping (Remington, 1963; Luchsinger *et al.*, 1964; Ruch & Patton, 1965; O'Rourke, 1967; Mills *et al.*, 1970; Laogun, Newman & Gosling, 1978). Reflections arise where there are geometrical or structural discontinuities in the vessel wall (McDonald, 1974). A geometrical discontinuity may be due to a change in the calibre of the vessel, as at a point of branching, or it may be due to the tapering of the vessel towards the periphery. The aortic bifurcation in particular has been identified to be an important source of pressure pulse reflection, as its area ratio changes with age and disease (Gosling, Newman & Bowden, 1971; Newman *et al.*, 1973; Laogun, Ajayi & Lagundoye, 1979). Structural discontinuities on the other hand may be due to a change in the elastic properties of the wall, often caused by vascular diseases.

The combination of a reflected, with an incident wave determines both the amplitude and phase of the resultant oscillation. Maxima and minima may thus occur at points along the system spaced approximately a quarter wavelength apart. For the production of absolute nodes and antinodes in a system, complete reflection and freedom from attenuation would be required (McDonald & Taylor, 1959). In the arterial system, the retrograde wave is normally of smaller modulus than the advancing one and the cancellation required for the production of absolute nodes may not be possible. Maxima and minima may therefore occur as a result of partial cancellation, but because of attenuation, the ratio between them would change as one moves away from the reflection site.

Amplitudes of pressure harmonics measured as a function of location in the systemic, arterial tree in dogs have been found to exhibit a series of maxima and minima due to reflections (McDonald & Attinger, 1965). Maxima and minima have also been found in the arterial input impedance measured in the dog (Taylor, 1966; O'Rourke & Taylor, 1966) and in man (Mills *et al.*, 1970). From an experiment on a length of rubber tube occluded at one end, Taylor (1957) studied the variation of apparent phase velocity c (measured at finite intervals) with distance from the occluded end. He found that like pressure amplitudes and impedance, c depends on the distance of the point of observation from the reflection site and also on the reflection coefficient, m, which is complex and should be expressed in terms of a ulus $|m|$ and phase angle ϕ, i.e. $m = |m|\, e^{j\phi}$.

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a position x in front of a discontinuity in the rial pathway, the effect of the combination he forward-going incident wave, from the .t, and the reflected wave, from the discon- iuity, is to create a resultant pressure wave with a measurable propagation velocity c', sometimes called the 'apparent' pulse wave velocity (PWV). This is related to the velocity of the forward-going wave c, often called the 'characteristic' PWV (McDonald & Taylor 1959; McDonald, 1974). For a lossless line, c and c' are related by the equation

$$c' = c \left[\frac{A_f e^{-j2\pi x/\lambda} + A_b e^{j2\pi x/\lambda}}{A_f e^{-j2\pi x/\lambda} - A_b e^{j2\pi x/\lambda}}\right] \quad (1)$$

where A_f and A_b are the amplitudes of the forward going wave and the reflected wave respectively, at the discontinuity. *Lambda* (λ) is the wavelength. Thus,

$$c' = c \left[\frac{1 + \dfrac{A_b}{A_f} e^{j4\pi x/\lambda}}{1 - \dfrac{A_b}{A_f} e^{j4\pi x/\lambda}}\right] \quad (2)$$

The bracketed quantity in equations (1) and (2) is inversely proportional to the pressure gradient of the resultant wave. This pressure gradient determines the rate of blood flow at any point x in the arterial pathway (Taylor, 1957). As x varies from 0, at the discontinuity, through values of $4x/\lambda = n$, where n is any integer, $e^{j4\pi x/\lambda} = \pm 1$, depending as n is even or odd. Thus c' oscillates about the value of c at $\lambda/4$ intervals in front of the discontinuity. Further, since there is not only absorption, but change of phase at any reflection, one may write $A_b/A_f = m$, the amplitude coefficient of reflection as $m = |m|\, e^{j\phi}$. If the reflection occurs at an abrupt complete occusion in the line then $\phi = 0$, if at a sudden open termination then $\phi = \pi$. For these two extremes $m = \pm |m|$. Thus for the first case c' is greater than c at the discontinuity and less than c at $\lambda/4$ away, whilst the opposite is true for the second. If ϕ lies between 0 and π then amplitudes of the oscillations of c' are reduced. The effect of considering attenuation in the line is to further reduce the oscillations of c' as x increases.

If c' is measured close to the discontinuity, ($x \ll \lambda$) the effects of attenuation are negligible and the complex terms in equation (2) are unity. From equation (2) it would appear that this condition is also fulfilled whenever $x = n\lambda/4$ (where $n = 1, 2, 3 \ldots$). However in practice, pressure and flow pulses in animals are not monochromatic, i.e. the waveforms have several components (with several values of λ). Thus it is only when

$x \ll \lambda$ that the resultant complex term for the whole composite waveform is effectively unity, then c' may be written as

$$c' = c \left[\frac{1+m}{1-m}\right] \tag{3}$$

However m is complex and one requires a further condition that ϕ be close to 0, then

$$c' = c \left[\frac{1+|m|}{1-|m|}\right] \tag{4}$$

$$\text{and } |m| = \frac{c'-c}{c'+c} \tag{5}$$

Thus, for a closed-end type stenosis with $\phi \approx 0$, the characteristic and apparent PWVs, close to the stenosis, permit calculation of $|m|$.

The experimental study of reflected waves reported in this paper was made to test the extent to which theory held true in:

(1) an *in vitro* test-rig with known discontinuities, and

(2) the aortic, iliac and leg arterial pathways in human volunteers.

Materials and methods

Effect of reflections and segment length on PWV measured in an elastic tube

The test rig (Fig. 1) used in this study consists of a Pauls' colostomy elastic-tubing (Precision Dippings Ltd) of length 5·0 m, internal radius 0·008 m and wall thickness 0·0005 m. One end of the tubing was connected to a pulsatile blood pump (Harvard Apparatus Co., Model 1423), which provided flows of physiological form. A reflecting discontinuity was created at the other end by connecting the elastic tubing to a 4·0 m rigid silicone tubing of slightly smaller internal radius (wall thickness, approximately 0·002 m) leading the fluid (1 pint of milk to 1 gal of water) through the colostomy tubing into the pump-priming reservoir. Milk was used in this study because it contains particles which reflect ultrasonic radiations like the red cells in the blood.

Doppler-shift frequency recordings were carried out along the colostomy tubing by means of a Doppler data-collection unit, comprising of two ultrasonic blood velocimeters working in conjunction with a Sony stereo tape-recorder. Each velocimeter had a pen-torch ultrasonic detector, containing two piezo-electric crystals, the transmitter and receiver in the same assembly. Using two 5·0 MHz detectors Doppler-shift frequency signals were recorded simultaneously in pairs along the elastic tubing at regular intervals of 0·2 m. Measurements were later repeated using a larger interval of 0·5 m. The signals recorded on tape in both cases were spectral analysed by a real-time, continuous, two channel spectrascribe (Coghlan, Taylor & King, 1974). The sonogram traces were produced on photosensitive paper (Kodak Linagraph Direct Print Paper, type 1895). The traces of the signal from

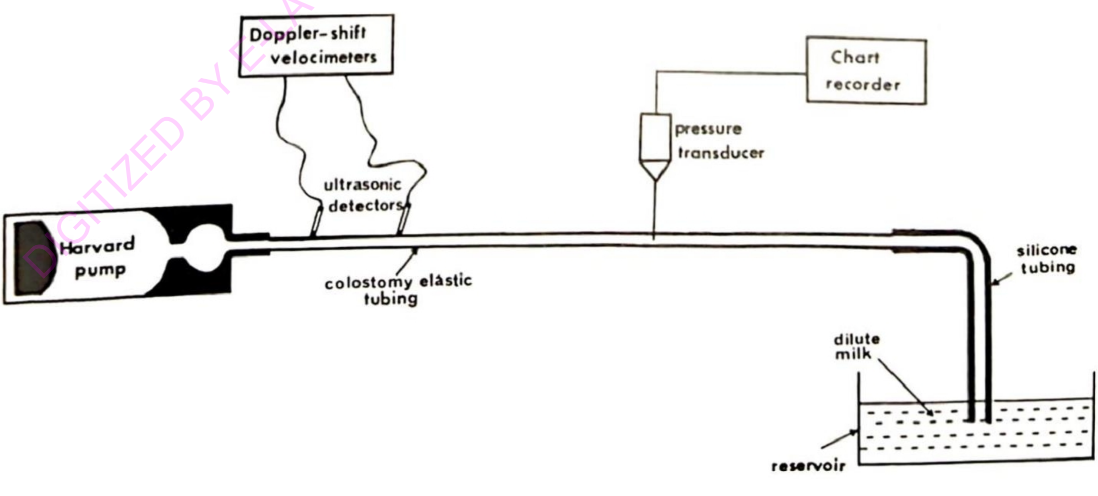

FIG. 1. *In vitro* elastic tube test-rig (return silicone-tubing, forming a closed system between pump and reservoir; not shown).

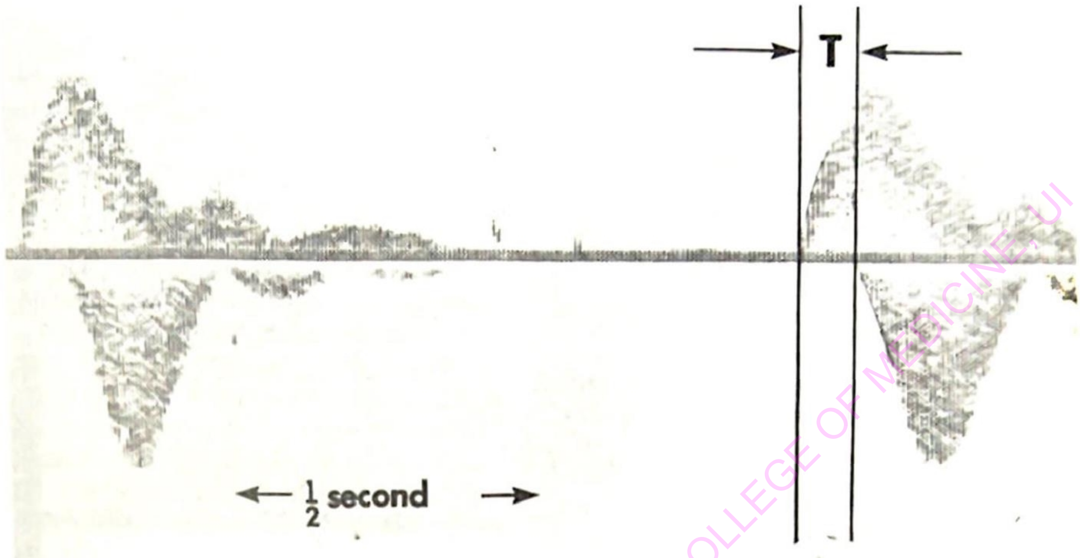

FIG. 2. Hard copy of recorded trace of sonogram signals showing measurement of time delay, T.

the distal sites were delayed compared to those of the proximal sites (Fig. 2) and the transit time T was measured from the foot-to-foot of the sonogram waveforms using an autograf digitizer. The apparent PWV at the successive locations along the tubing was then determined for each segment from T and path length 1.

Variation of reflection with degree of stenosis

In order to study changes in local reflection occurring from varying degree of stenosis, the same type of elastic tubing (Precision Dippings Ltd) used previously was employed but in this case the length of the tubing was approximately 17 m in order to provide a reasonable reflection-free line. One end of the tubing was connected to the Harvard pump and the other end to the reservoir. The experimental arrangement was as given previously. Five connectors, each of length 5·0 cm and diameter 0·8, 1·0, 1·2, 1·42 and 1·6 cm respectively, were made from plastic tubings and each was connected in turn into the elastic tubing at a point approximately 5·0 m from the pump. The degree of stenosis at each connector was expressed in terms of area ratio. Apparent PWV was measured as a function of distance from the stenosis using the continuous wave Doppler-shift technique. These measurements were made at regular intervals of 0·2 m between the connector and the Harvard pump. In order to demonstrate reflection effects in this experiment, a Fourier analysis was carried out on some of the signals recorded from certain locations along the tubing and for certain area ratios of stenosis.

Effect of reflection on PWV measured from specific arterial segments (aorta, iliac and leg) in man

Proximal and distal pulse wave recordings were made in the aorta, iliac and leg arterial segments in two human volunteers, ages 14 and 25 years respectively, using the Doppler-shift ultrasonic technique (Laogun, 1977; Laogun & Gosling, 1979). The length, l, of each segment was determined by measuring distances between detector sites to within a few millimetres. Sonogram traces from each of the segments were produced on photosensitive papers and the maximum frequency (f_{max}) envelopes traced onto transparent graph papers. Four pulses were used in each case because of possible slight pulse-by-pulse variation. The waveforms were digitized at a sampling rate of about 70–80 points per pulse on a D-mac electromagnetic pencil follower interfaced to an 029 IBM card punch

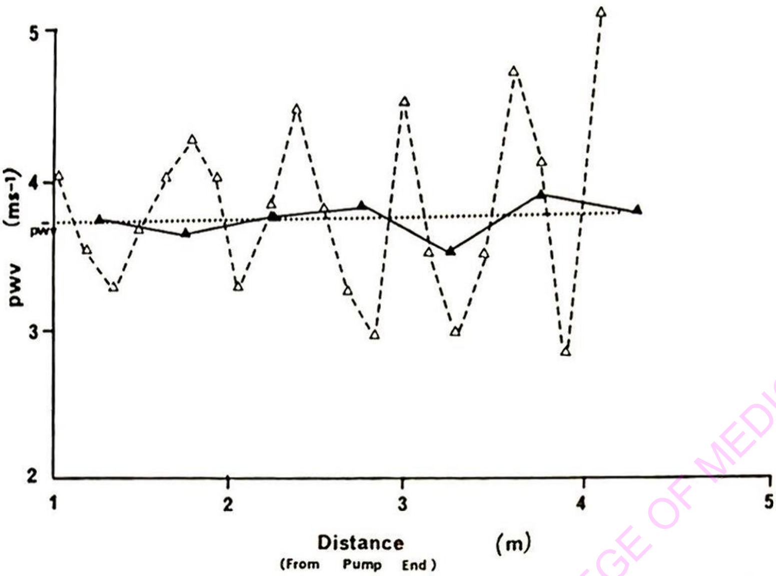

FIG. 3. Pulse wave velocity (PWV) measured in an elastic tube over successive constant segment lengths proximal to a discontinuity at about 5·0 m from the pump. △, 0·20 m segments; ▲, 0·5 m segments; . . . . . ., characteristic PWV for the tubing.

and the Fourier analysis of the data was carried out on an IBM 360 Computer. The variation of the phase velocity with harmonics (1–10) for the three segments has been presented in Fig. 6.

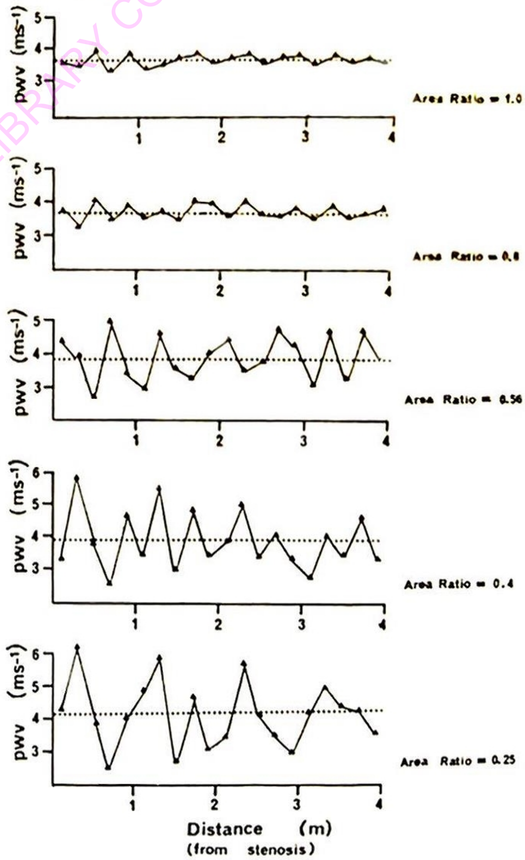

FIG. 4. Pulse wave velocity (PWV) measured over successive 20 cm segments and plotted against distance proximal to 5·0 cm long stenoses of area ratios as shown (area ratio = cross-section of daughter/cross-section of parent tube).

Results

Figure 3 shows the variation of PWV with distance from the pump to the position of the discontinuity at the other end of the tubing. The curve represented by broken lines shows the PWV/distance variation for measurements over 0·2 m segment lengths, and the curve in continuous line shows the variation for the 0·5 m segment length. The dotted line represents the mean value of the propagation velocity or the characteristic PWV of the tubing. Its value for the colostomy tubing used was found to be 3·76 ± 0·12 ms^{-1}. From Fig. 3, it may be observed that propagation velocity exhibits maxima and minima about the mean value, over both segment lengths with a much wider amplitude of oscillation for the 0·2 m path than for that of 0·5 m.

Figure 4 shows PWV measured over 0·2 m segments and plotted against distance from varying sizes of stenosis at the reflection site. In this figure, dotted lines represent the calculated mean propagation velocities over the length of the tubing, close to the site of the stenosis. From Fig. 4, it may be deduced that the smaller the area ratio at the discontinuity, the greater the

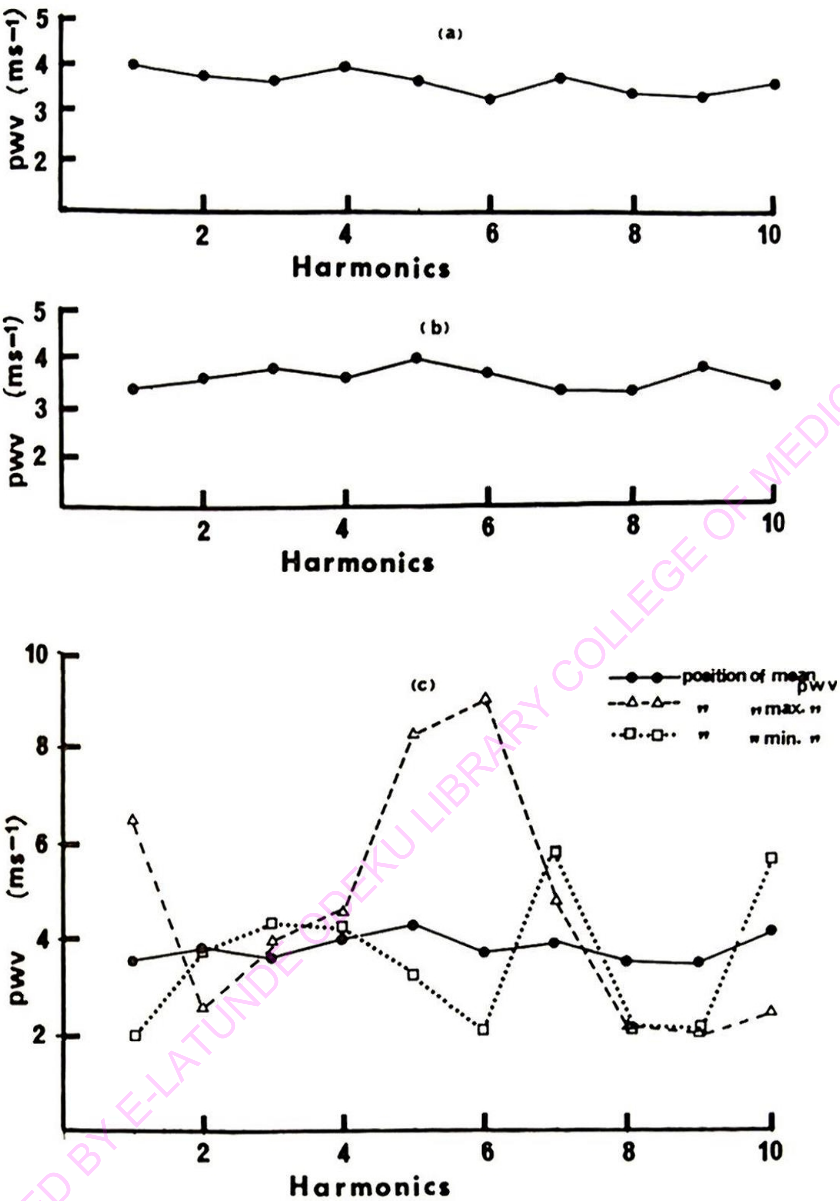

FIG. 5. Variation of phase velocity for the first 10 harmonics in flow pulses proximal to stenoses of area ratios; (a) 1·0 (b) 0·8 (c) 0·25. In (c) three sites were analysed as indicated.

·on in the apparent PWV/distance char- :.

der to explore further the relationship the apparent and the characteristic in the presence of reflections, the sono- veforms were Fourier analysed for some ls recorded at certain locations on the ping for stenosis of area ratio 1·0, 0·8 respectively. Result of this analysis is presented in Fig. 5 where variation of the phase PWV for up to 10 harmonics is shown. For area ratios 1·0 and 0·8, the phase velocities are fairly constant over the harmonic range 1–10, indicating low reflections. Case (c) in Fig. 5 represents Fourier analysis at three distinct positions of measurement in front of the stenosis with area ratio 0·25, i.e. positions of maximum, minimum and mean PWV respectively. At the position

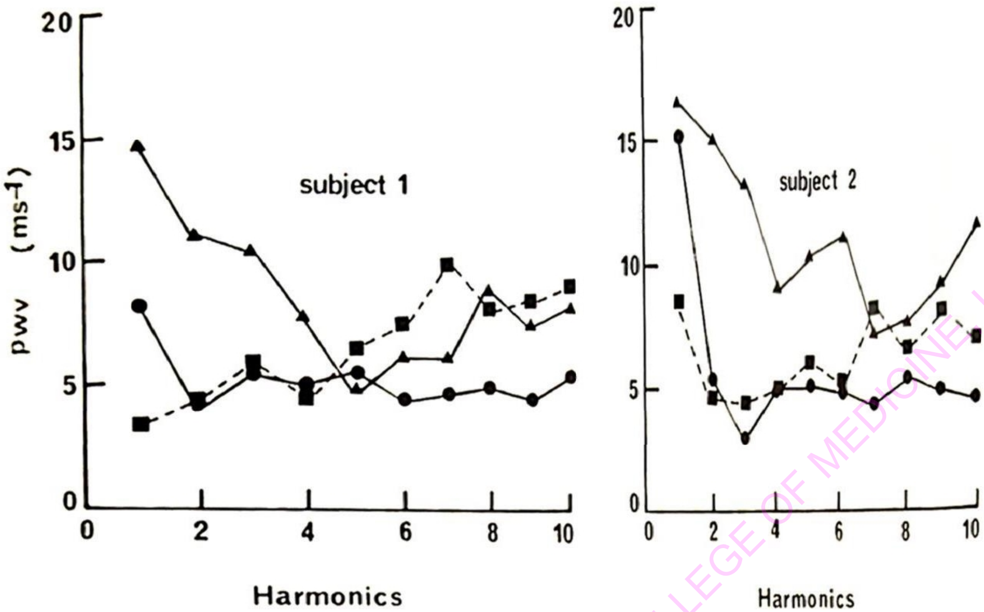

FIG. 6. Typical harmonic variation of phase velocity in the aorta, iliac and leg arterial segments in two human volunteers. ●, Aorta; ■, iliac; ▲, leg.

where the apparent PWV was close to its characteristic value for the tubing, harmonic analysis showed little fluctuation in phase velocity. On the other hand, there was a wide fluctuation in phase velocity with harmonics for positions of maximum and minimum PWV respectively, and this was indicative of the presence of reflections as predicted by theory, particularly since measurements were made over small intervals (0·2 m). Figure 6 showed typical variation of phase velocities with harmonics in the aorta, iliac and leg arterial segments in two human volunteers. It may be observed that the aortic segment showed little fluctuation in phase velocity. However, the iliac and the leg arteries showed greater fluctuation than the aorta particularly in the lower harmonic range.

Discussion

It is known that, due to reflections, the amplitude of pressure waves increases towards a closed termination at which there is an antinode of pressure and a node of flow. The converse also holds for an open end where ϕ, the phase change on reflection, is 180° for pressure and 0° for flow in contrast to 0° pressure and 180° flow for the closed end (Taylor, 1957). Partial reflection of both pressure and flow waves occurs at a constriction or stenosis usually producing an increase in oscillatory pressure and a decrease in oscillatory flow (McDonald, 1974; Newman, Batten & Bowden, 1977). It has been shown, theoretically, that pulse propagation velocity measured by flow or pressure, proximal to such partial or total occlusion, will oscillate about the characteristic value. This has been practically demonstrated in the experiments reported in this paper. In the elastic tubing, the PWV has been shown to exhibit maxima and minima, proximal to a reflection site. The amplitude of oscillation has also been found to be largely dependent on the position of the segment over which the pulse propagation velocity is measured as well as the segment length. Figure 3 shows that the amplitude of oscillation increases towards the reflection site. It also demonstrates the 'averaging' effect of measuring with a segment length longer than one-quarter wavelength of the dominant harmonics in the pulse waveform. In all these experiments, the pulse waveform was of constant frequency of 1·25 Hz and physiological shape, with the third harmonic dominant. Since the characteristic flow pulse wave velocity for the colostomy elastic tube was $3{\cdot}76 \pm 0{\cdot}12$ ms^{-1}, the fundamental wavelength, λ, was 3·0 m. Hence, $\lambda/4$ for the third harmonic was 0·25 m. As may be seen in Fig. 3, the amplitude of oscillation of the propagation velocity is wider for the 0·2 m segments than for those of 0·5 m. These findings are in agreement with previous suggestions by McDonald (1974) that larger segment lengths provide better averaging for the estimation of the characteristic value of PWV.

Relating this to man, it may be expected that pulse propagation velocity measured in the leg arteries (from the common femoral to the posterior tibial) and in the aorta would be closer to the characteristic value than that measured in the iliac arteries, where the segment length is small compared with the pulse wavelength. It is however known that wave propagation in the aorta is not often affected by reflection from the periphery (Peterson & Gerst, 1956). On the other hand, the iliac and leg arteries are closer to the peripheral beds and therefore witness greater pulse wave reflection. Also, because of the pro nounced tapering and continuous branching of the vessels in those segments, the area ratio decreases and the reflection modulus increases. The amplitude of oscillation of the pulse propagation velocity above, and below the characteristic value thus increases as observed in this study.

In conclusion, when the effect of pulse wave reflection in a conducting elastic tube (e.g. the arterial system) is small, the apparent propagation velocity determined by measuring the foot-to-foot transit time or phase lag of flow pulses will be approximately equal to the PWV characteristic of the vessel wall. However, the presence of marked reflections causes an oscillation of the apparent PWV about the characteristic value. This oscillation may be quite pronounced, particularly in cases where short segments ($l \ll \lambda$) are considered.

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