Effect of Age on the In Vitro Reflection Coefficient of the Aortoiliac Bifurcation in Humans

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The local reflection coefficient (r) at the aortoiliac junction was estimated in vessels removed at autopsy from 15 females and 31 males aged 2 months to 88 years by calculating the characteristic impedance ($Z_c$) of the abdominal aorta and the two common iliac arteries. $Z_c$ was evaluated for each vessel by measuring cross-sectional area from radiographs and propagation velocity of an isolated pressure impulse generated by a solenoid-driven piston connected to the distal end of the abdominal aorta, and detected at several sites a known distance from the junction. Attenuation coefficients in the aorta were estimated from the peak amplitude of the impulse at these several sites. We observed significant decreases with age in abdominal aortic attenuation coefficient ($\gamma=0.053-5\times10^{-5}\times$age [yr], $r=-0.42, p<0.005$), and the area ratio (AR) of the junction (sum of the iliac cross-sectional areas/aortic cross-sectional area) (AR=$0.93-0.002\times$age [yr], $r=-0.31, p<0.05$). Bifurcation angle (Angle) and aortic propagation velocity ($c_{aorta}$) increased significantly (angle=$40.2+0.26\times$age [yr], $r=0.41, p<0.01$; $c_{aorta} \text{ (ms}^{-1}) = 7.59+0.175\times$age [yr], $r=0.69, p<0.0001$). No significant association between age and iliac propagation velocity was found. On the basis of area measurements alone, it has been argued that the observed decrease in area ratio with age causes r to become more positive. The lack of age-related changes in iliac propagation velocity (and hence stiffness), an observation confirmed by most of the few reports in the literature, and the large increase in abdominal aortic stiffness, however, greatly outweighed the area ratio changes and caused a significant decline in reflection coefficient from +0.3 in early life to −0.3 in old age. (Reflection coefficient=0.30−0.0065\times$age [yr], $r=-0.68, p<0.0001$). Between the ages of 30 and 60 years, its absolute value was less than 0.1, confirming in vivo work on subjects within this age range. In children, a significant positive reflection at the distal end of the aorta will amplify the pulse wave in this region. As this vessel becomes stiffer with increasing age, amplification will increase, whereas the increasingly negative value of r will partially offset this rise, reducing pulse pressure on either side of the junction. These processes might help to minimize the adverse effects of increased vascular stiffness due to aging and disease in this part of the circulation. (Circulation 1990;82:114–123)

The magnitude of measured pressure and flow waves in the cardiovascular system is determined by the interaction of incident waves generated by the heart and one or more reflections due to junctions and, in diseased vessels, occlusive or aneurysmal lesions. Constricted muscular vessels in the peripheral circulation can cause additional reflections. Thus, to measure the magnitude of reflection coefficients, it is necessary to resolve the measured pressure and flow wave into its generated component traveling away from the heart, a reflected wave traveling back toward the heart and, possibly, one or more rereflected waves traveling in either direction.

The different approaches to this problem described in the literature can be divided into two types.1 First, those in which the magnitude of all reflections distal to the measurement site are determined.2–8 These methods do not, in general, give any information about the location of particular sites of reflection but allow the calculation of a global reflection coefficient together with propagation constants (phase velocity and attenuation coefficient) of the vessel in which the measurements are made. The second class of measurements is based on the prop-
property that reflections occur at discrete sites such as a junction at which there is a change in the local characteristic impedance ($Z_c$) and require that $Z_c$ be measured at at least two points in the region of the putative reflection site. Because $Z_c$ is defined as the fluid impedance in the vessel in the absence of any reflections, it is often not easy to estimate its value directly from pressure and flow measurements at sites where reflections are indeed present. In several studies, it has been estimated from the higher harmonics of the input impedance spectrum, using the assumption that any reflections of waves above a certain frequency, for example, 5 Hz, are attenuated to negligible amplitudes by the time they have reached the measurement site. Although this might be satisfactory in in vivo studies in experimental animals in which it is possible to perform simultaneous measurements of pressure and flow, it is technically and ethically difficult to adapt these methods to studies on large numbers of human vessels.

$Z_c$ can be calculated from measurements of true propagation velocity and cross-sectional area, and the true propagation velocity can be estimated from the higher harmonics of its spectrum or from foot-to-foot time delays of pressure waves measured at two or more sites a known distance apart. In general, measurements of this type, involving fewer invasive procedures, are more suitable for studies on living human subjects.

Propagation, attenuation, and reflection coefficients can be measured directly by generating a single pulse of flow and measuring the resultant pressure wave. If this pulse is of sufficiently short duration, it can be observed in isolation from its own reflections. In living animals, the pulse will be superimposed on the normal pressure wave; whereas, in dead specimens or in rubber tube models of the circulation, the pulse and its reflections will be seen in isolation.

In recent years, impulse measurements have been performed on the aorta and the femoral artery in experimental animals but no reports of measurements of this type in humans have been found. In 1980, we reported the results of postmortem propagation velocity, attenuation, and reflection measurements on the human aorta of five human subjects. We observed that within the age range of 26–48 years, attenuation remains constant and the reflection coefficient of the aortoiliac junction in vessels free of occlusive disease is close to zero. Some in vivo impedance measurements, however, suggest that this junction is a major reflection site, whereas others suggest that stronger reflections occur near the inguinal ligament and the renal arteries and that the aortoiliac junction is well matched. It is important that the behavior of the pulse wave in this region of the circulation is understood, not only to explain its shape but also to clarify the effect of abnormal pressure amplitudes and shear stresses associated with the vascular lesions that commonly occur in the abdominal aorta and iliac arteries.

For these reasons, we have made similar impulse measurements on many subjects and report here the effect of age on wave propagation in the region of the aortoiliac junction.

**Methods**

Human aortas were obtained within 48 hours of death. The lengths of the abdominal segment and both common iliac vessels were measured in situ as was the angle of the aortoiliac bifurcation. This was defined, for cadavers with their legs together, as the angle subtended at the midpoint of the junction (assessed by eye) by lines drawn between this point and the midpoints of the internal-external iliac junction. The vessels were dissected free of fat and connective tissue, and as many branches as could be identified were cut, wherever possible, at least 1 cm from their origin to allow for later ligation. They were flushed with 0.9% saline, wrapped in damp paper tissue, and stored at 4°C for periods not exceeding 24 hours until used. Specimens were mounted in an adjustable rig, stretched to their in situ length, and the bifurcation angle was set to its in situ value.

The impulse generator, which consisted of a solenoid driving a Teflon piston in a glass cylinder, was connected by a polyvinyl chloride manifold to the proximal end of the specimen (Figure 1). The piston was withdrawn a known distance (typically, 2 mm, corresponding to a volume of ~0.1 ml), and the impulse was generated by the impact of the moving core of the solenoid. The typical impact speed of approximately 2 msec$^{-1}$ produced an impulse mea-

![Figure 1](http://circ.ahajournals.org/DownloadedFrom/)
sured at the outlet manifold that lasted about 1 msec. The position and velocity of the piston were monitored by an infrared-emitting diode, an optical wedge, and a phototransistor (Figure 1). The two iliac vessels were connected to a pressure head adjusted to give a mean pressure in the vessels of 100 mm Hg. All major branches were tied with string. The system was filled with a mixture of 32% glycerol (by volume) in 0.9% saline, giving a viscosity close to that of blood (≈30 mp). When the system was pressurized, many small leaks were observed. These were clamped and tied off with silk thread. At this stage, the specimen would normally continue to weep through numerous branches too small to tie off individually. Typically, this would constitute a total leak rate of less than 35 ml/min. The leaked fluid was returned to the reservoir with a roller pump.

Pressure was measured with a catheter-tipped transducer (16 CT, Gaeltec Ltd., Dunvegan, Isle of Skye, Scotland) passed into the system through a side arm on the impulse generator. The output of the transducer was amplified and passed through a 12-bit analog-to-digital converter to an Apple IIgs computer, sampling at a rate of 5 kHz.

For each impulse, the solenoid was triggered by a push button connected to a relay. This action was also used to initiate sampling of the pressure and piston displacement signal. The stroke of the piston was set to give an impulse amplitude of 20 mm Hg. Typically, this was achieved by injecting approximately 0.1 ml fluid into the system. Recordings were made with the transducer positioned at 2-cm intervals down the aorta, starting at 2 cm from the outlet of the generator and, in turn, down each iliac branch at 1-cm intervals. Data for each run were stored on disks for later analysis.

The internal radii of the aorta and iliac vessels at 100 mm Hg were obtained by measuring a radiograph of the system, having filled it with a 50% suspension of barium sulfate. The film used was Kodak Industrex CX (Eastman Kodak, Hemel Hempstead, Hertfordshire, England) exposed at 40 kV and a dose of 140 mA.

The arrival time of the peak of each impulse was defined by the maximum of each pressure-time trace. The computer was programmed to detect this automatically, by sampling the stored waveforms within a time interval selected by the operator. In this way, it was possible to detect a local maximum corresponding to the appropriate peak rather than the overall maximum attained during the entire sampling period. (See, for example, the 18-cm trace in Figure 2.) Waveforms with no clearly defined local maximum (e.g., 22 cm in Figure 2) were not measured. Because the signal was sampled at 5 kHz, it was possible to estimate the arrival time of the peak to the nearest 0.2 msec. Peak heights were calculated as the difference between the amplitude of the local maximum and the mean pressure sampled for 500 msec before the impulse was triggered. Successive traces were synchronized from the point that the piston began to move after the impact of the solenoid core.

For the aorta and both iliacs, straight lines were fitted to plots of the time taken for each impulse to travel from the generator against the distance it had traveled along each segment. Propagation velocity was taken as the gradient of this line.

The attenuation was expressed as follows: \( A_i/A_0 = e^{-\gamma d_i} \), where \( A_i \) is the impulse amplitude at site \( i \), distance \( d_i \) is from the site nearest the generator, and \( A_0 \) is the amplitude at this position. Hence, \( \gamma \) was evaluated as the gradient of a line fitted to a plot of log \( (A_i/A_0) \) against \( d_i \). The reflection coefficient, \( r \), was calculated from the following relation\(^20\):

\[
R = (Y_A - Y_L - Y_R)/(Y_A + Y_L + Y_R)
\]

where \( Y \), the admittance, is given by the following: \( Y = 1/Z = a/\rho c \), for a tube of cross-sectional area \( a \),

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**FIGURE 2.** Pressure time traces for a 37-year-old woman. Numbers at the side of each trace are the distances in centimeters of the transducer from the generator. Arrow marked B indicates position of the bifurcation. The impulse reflected from the bifurcation is indicated by the arrowheads.
propagation velocity \( (c) \), filled with fluid of density \( (\rho) \). The subscripts refer to the aorta (A) and the left (L) and the right (R) common iliac arteries.

For an equifurcation, the reflection coefficient can be expressed in terms of the area and propagation velocity ratios of the junction, as follows:

\[
r = \frac{1 - \lambda}{1 + \lambda}
\]

where \( \lambda \) is given by

\[
\left( \frac{2A_{\text{ILIAC}}}{A_{\text{AORTA}}} \right) \cdot \left( \frac{c_{\text{AORTA}}}{c_{\text{ILIAC}}} \right)
\]

Hence, as the cross-sectional area of the aorta increases with respect to the sum of the iliac areas, \( \lambda \) is reduced and the reflection coefficient becomes more positive; whereas, if the aorta becomes stiffer than the iliacs, \( \lambda \) increases, and the reflection coefficient becomes more negative. It should be noted that because we did not assume that the two iliac vessels had the same cross-sectional area, Equation 1 rather than Equation 2 was used to calculate \( r \) in all cases.

The effect of changes in mean and impulse pressure on propagation velocity in the aorta and iliac arteries was investigated on six specimens. After the initial set of measurements under standard conditions, two of these six vessels were subjected to mean pressures of 50 and 150 mm Hg and impulses obtained throughout the system as before. The next two were maintained at a mean pressure of 100 mm Hg, and the pulse pressure was varied between 10 and 40 mm Hg. In the final two specimens, recordings were made at various combinations of mean and pulse pressure. In three additional specimens, the area ratio measurements were made from radiographs exposed at mean pressures in the range of 50–150 mm Hg.

The effect of time after death and excision was studied on two bovine carotid arteries and one bovine aorta removed immediately after death, and one human vessel that was obtained 20 hours after death. The experiment was repeated on all four vessels at 24-hour intervals up to 5 days after excision. Between runs, the vessels were removed from the rig, moistened with isotonic saline, and stored in sealed containers at 4\(^\circ\) C.

**Results**

Of the 46 specimens used in this study (Table 1), 15 were from females (average age, 52.6 years; range, 6 months to 86 years) and 31 were from males (average age, 53.1 years; range, 2 months to 88 years). Because the males and females were not age-matched, results for both sexes have been combined and no assessment based on sex differences has been attempted. Vessels with marked aneurysmal dilations were excluded from the study as were those with any segment that appeared from the radiographs to be more than 50% occluded by radius (>75% by area). Also excluded were any specimens that contained radiographically detectable plaques within 10 mm of the apex of the aortoiliac junction. Hence, all area ratio measurements were made on radiographically unoccluded junctions. The approximate degree of calcification was estimated from the radiographs and

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<th>Cause of Death</th>
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</tr>
<tr>
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<td>Respiratory failure</td>
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<td>F</td>
<td>Lymphoma</td>
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</tr>
<tr>
<td>26</td>
<td>M</td>
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</tr>
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<td>M</td>
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</tr>
<tr>
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</tr>
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*Estimated from the radiographs by expressing summed length of plaques as a fraction of the total length of vessel.

0, 0–24% involvement; 1, 25–49% involvement; 2, 50–74% involvement; 3, 75–100% involvement.
graded as shown in Table 1. No further analysis of these observations was attempted.

Figure 2, a representative set of recordings from one subject, shows a marked attenuation and broadening of the impulse as it travels away from the generator. The double peak seen at the more distal sites and the subsequent broader oscillations are reflections from the rigid terminations connected to the iliac arteries, which are rerereflected from the generator piston. These are not seen with the vessel in situ. In this specimen from a 37-year-old female, a small positive reflection (r=0.27) at the bifurcation (arrow B, Figure 2) gives rise to a detectable backward-traveling impulse (arrowheads).

Figure 3, a distance-versus-time plot, shows a slight increase in aortic propagation velocity toward the distal end of the aorta (square symbols). In approximately half the specimens, no such increase was observed, although there was no apparent relation between this distal velocity increase and age. In those that did show such an increase, a second-order polynomial was fitted to the distance time points (in all cases, the correlation coefficient was >0.95), and the velocity near the junction was calculated from the first derivative of the polynomial. The velocity values obtained in this way did not differ significantly from those obtained with a linear fit (paired t test, p>0.1). For consistency in all cases, the velocity used to calculate the reflection coefficient was obtained from the gradient of a straight line fitted to the points for each segment.

The propagation velocities of the fresh bovine arteries declined steadily as the time between measurement and death increased. In all cases, however, the gradient of the regression of aortic-to-iliac velocity ratio on time since death did not differ significantly from zero. Increasing the pulse pressure from 10 to 40 mm Hg gave a mean relative change in velocity ratio (i.e., \( VR_{40 \text{ mm Hg}} / VR_{10 \text{ mm Hg}} \)) of 1.03±0.14 (SD). When the mean pressure was increased from 50 to 150 mm Hg, the mean relative changes in area and velocity ratios were 1.09±0.06 and 1.04±0.05, respectively.

The results of dimensional and radiographic measurements are shown in Table 2, together with velocities and attenuation coefficients derived from peak amplitudes and time delay measurements as shown in Figures 2 and 3. The correlation coefficients between each variable and age, and their degree of statistical significance, are shown in the bottom three rows of the table. In summary, we have found significant decreases with increasing age in the aortic propagation velocity, together with the velocity ratio and the area ratio of the junction. The reduction in the area ratio with age is also shown in Figure 4, where the data from this study (solid symbols) are compared with those of Gosling et al. (open symbols). In general, the bifurcations were not symmetrical about the long axis of the aorta. The angle, as defined in “Methods,” increased significantly with age. No further analysis of bifurcation geometry was performed.

Aortic propagation velocity increased significantly with age, whereas no significant association between age and left, right, or mean iliac propagation velocity was observed. Left and right iliac propagation velocities did not differ significantly (left, 13.2±6.0 msec\(^{-1}\); right, 12.4±4.9 msec\(^{-1}\); p>0.1, paired t test).

The regression line in Figure 5 (Reflection coefficient=0.30–0.0065×age, \( r=-0.68, p<0.0001 \)) shows that the reflection coefficient calculated from Equation 1 falls from a value of approximately +0.3 in early life to --0.3 in old age. Between the ages of 30 and 60 years, its absolute value is less than 0.1.

**Discussion**

The major advantage of the impulse methods used in this study is simplicity. Past experience with experimental animals and rubber tube systems has shown that successive impulses are repeatable. It is therefore possible to obtain propagation velocity readings and attenuation values by performing successive measurements using only one transducer at several different positions within the system.

The use of tissue excised after death can cause several potential problems. Although postmortem changes did cause a steady decline in aortic and iliac propagation velocities, the velocity ratio measured in
TABLE 2. Geometric and Propagation Constants for Each Subject

<table>
<thead>
<tr>
<th>Age</th>
<th>Sex</th>
<th>Angle*</th>
<th>AR</th>
<th>$\gamma$ (cm$^{-1}$)</th>
<th>PWV$_A$</th>
<th>PWV$_I$</th>
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Gradient 0.26 -0.002 -5x10^{-4} 0.175 -0.054 0.02
Intercept 40.20 0.93 0.053 7.59 15.57 0.42
Correlation 0.41 -0.31 -0.42 0.69 -0.24 0.70

*Data for five subjects not obtained. †Mean value for both iliac vessels.

AR, area ratio; $\gamma$, attenuation coefficient; PWV, pulse-wave velocity (units, msec$^{-1}$); VR, velocity ratio; area ratio, sum of iliac cross-sectional areas/aortic area; velocity ratio, aortic PWV/mean PWV of both iliac arteries.

Gradient 0.26 -0.002 -5x10^{-4} 0.175 -0.054 0.02
Intercept 40.20 0.93 0.053 7.59 15.57 0.42
Correlation 0.41 -0.31 -0.42 0.69 -0.24 0.70

*Data for five subjects not obtained. †Mean value for both iliac vessels.
one human aorta obtained less than 24 hours after death and the three fresh bovine vessels did not change significantly during the subsequent 72 hours. This confirms our previous observations in dogs\(^1\) in which the aortoiliac reflection coefficient obtained in vivo was similar to the excised value obtained 24 hours after death.

The effect of increasing mean system pressure was, as expected, an increase in the propagation velocity of all segments. Because the velocity ratio (and the area ratio) of the junction changed little in the pressure range of 50–150 mm Hg, however, we found that the reflection coefficient was similarly independent of pressure. For the same reasons, the calculated reflection coefficient was insensitive to changes in pulse amplitude.

The absence of peripheral vascular beds originating from upstream branches such as the renal arteries and the need to tie these branches gave rise to additional possible reflection sites. In many specimens, these appeared as an amplification of the pulse near the midpoint of the abdominal aorta (Figure 2). There was no indication, however, that they perturbed the position of the impulse peak, or its amplitude at distal sites.

In previous studies,\(^1\)\(^5\),\(^1\)\(^6\),\(^2\)\(^2\) we have shown that Fourier analysis of impulses produced by several different methods makes it possible to measure propagation, attenuation, and reflection coefficients in the frequency range of 4–100 Hz. We have found that in experimental animals, the propagation velocities of the Fourier components of the pulse are independent of frequency in this range, that the attenuation per unit wavelength is constant, that, in rubber tube models of vascular junctions and lesions, the reflection coefficient is a real number, and that it also does not change with frequency.

In the experiments described here, reflections from the terminations of the iliac vessels (where they are connected to the perfusion system) perturb the low-frequency components of the measured pulses. These can be removed by interpolation\(^2\)\(^3\); however, uncertainties in the value of the baseline and the true shape of the tail of the impulse (i.e., as it would appear in a reflectionless system) suggest that this procedure is of doubtful value. Therefore, in this study, we have not used Fourier analysis but have calculated c from peak-to-peak time delays and \(A_i/A_o\), as the ratio of pulse amplitudes. In impulse experiments on the canine aorta, however, propagation velocities measured from peak-to-peak time delays are close in value to those measured from phase differences in the Fourier coefficients obtained from pressure measurements at different sites.\(^1\)\(^3\) Reflection coefficients calculated in the manner used here correlate closely with values obtained by mea-

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**Figure 4.** Plot showing effect of age on area ratio of the aortoiliac junction. □, data from Reference 21; ■, data from this study, mean for ages 0–19 years, with remainder grouped into decades.

**Figure 5.** Plot showing effect of age on reflection coefficient of the aortoiliac junction. Solid line is least-squares linear fit to the points shown.
measurement of the peak height of incident and reflected impulses. We assume, therefore, that the estimates of c, λ, and r in this study that, being based on peak impulse measurements and calculation of characteristic impedance, are confined to the upper part of the impulse frequency range (approximately 100 Hz),\textsuperscript{15} are representative of their values in the physiological frequency range.

Because the attenuation coefficient depends strongly on frequency, however, the values we have obtained (at an indeterminate frequency) cannot be compared with others in the literature. Nevertheless, they clearly show, we believe for the first time, a decline in aortic pulse-wave attenuation with age. This decline is perhaps related to a decrease in wall viscosity due to a reduction in smooth muscle content and an increase in collagen, although there is no evidence of an age-related decrease in the viscous component of the dynamic elastic modulus in the abdominal aorta.\textsuperscript{24}

The effect of the bifurcation angle on turbulence and flow separation has been discussed in several papers,\textsuperscript{25–27} and an association between lower-than-normal values and atherosclerosis has also been reported.\textsuperscript{28} In these studies and others,\textsuperscript{29,30} no breakdown of the results by age of the subject was reported, so comparisons with our results are probably not meaningful. We have observed a steady increase in bifurcation angle with age, although Laogun et al\textsuperscript{42} and Bargeron et al\textsuperscript{32} found no such correlation. A possible reason for this discrepancy is that our results are based on in situ measurements of deflated vessels, whereas the measurements of Laogun et al\textsuperscript{42} were performed on radiographs obtained in vivo, and Bargeron et al\textsuperscript{32} used a more sophisticated technique in which the angle was determined from a computerized analysis of radiographs and more closely reflects the true angle close to the bifurcation. An increase in branching angle with age would increase the reflection coefficient. This effect is thought to be small,\textsuperscript{33,34} however, and it has not been considered in our calculation of reflection coefficient.

Few studies concerning the geometry of vascular junctions have reported the effect of age on area ratios. Beales and Steiner\textsuperscript{55} found no age-related changes in area ratio, although their measurements on living patients were confined to subjects who were 12–50 years old. Gosling and colleagues\textsuperscript{21} found a significant decrease in the area ratio of the aortoiliac junction with age. Their results based on measurements in living normal subjects less than 50 years old are similar to ours obtained postmortem (Figure 4). Laogun et al\textsuperscript{31} observed a similar decline with age in Nigerians. The ratio remained close to 0.6 in atherosclerotic patients who were 40–80 years old.\textsuperscript{36,37} We have also observed little change in older subjects, although the mean for the age range of 50–88 was 0.78. In this study, however, the subjects were not selected for atherosclerosis, and those with radiographically detectable occlusive disease near the aortoiliac junction were excluded. A reduction in the area ratio of atherosclerotic specimens has also been observed by Sauvage et al,\textsuperscript{38} but ages were not reported. We conclude that in vitro radiographic measurements accurately reflect those obtained from angiograms of living subjects, that the area ratio of the aortoiliac junction declines from birth to the end of the fourth decade, and changes little after the age of 50 years.

In common with many other studies, we observed an increase in aortic propagation velocity with age (see, for example, Reference 39), although our values appear to be higher than those observed in living subjects. In eight men and one woman with a mean age of 42 years, Latham et al\textsuperscript{10} obtained a value of 9.2 msec\textsuperscript{−1} from foot-to-foot time delays in measurements of pressure waves in the abdominal aorta (compare with 15 msec\textsuperscript{−1} in this study). Similarly, the regression equation derived from measurements on urban Chinese\textsuperscript{39} gives a value of 10 msec\textsuperscript{−1} for subjects who are 42 years old, although this figure applies to the entire aorta and is expected to be lower than values for the abdominal segment alone. This discrepancy is unlikely to be related to differences between living and dead vascular tissue because Learoyd and Taylor\textsuperscript{24} calculated a value of 8.8 msec\textsuperscript{−1} obtained from elasticity measurements in the excised abdominal aortas of six subjects with an average age of 45 years. In all these studies, the mean pressure at which measurements were made was less than 100 mm Hg, whereas our figures were obtained from peak-to-peak determinations at which the pressure was approximately 120 mm Hg. We believe that this accounts for the higher velocity values observed in this report. The calculated reflection coefficient, however, depends on the aortoiliac velocity ratio rather than the absolute values, and this ratio was found to be independent of mean pressure in the range of 50–150 mm Hg.

Although it is generally believed that true phase velocity increases as the pulse wave moves from the heart to the periphery, and there is good evidence that this is true in dogs,\textsuperscript{40} the few reported results of studies in humans suggest that this is only true in young people. Butcher and Newton\textsuperscript{41} showed that the abdominal aorta taken from the cadaver of a 15-year-old subject was twofold to threefold more distensible than the common iliac, a difference that was maintained in the pressure range of 100–150 mm Hg. In three subjects who were 23–58 years old, this difference was small, and in a 70-year-old subject, the volume distensibility of the common iliac was approximately 50% greater than that of the abdominal aorta. Learoyd and Taylor\textsuperscript{24} also observed a small decline in iliac distensibility in six subjects more than 35 years old, when compared with six subjects younger than this age. A similar reduction in propagation velocity (and hence, elastic modulus), on going from the abdominal aorta to the iliac artery, was also observed by Latham et al\textsuperscript{10} in nine subjects with a mean age of 42 years and an age range of 35–61 years. Similarly, Laogun and Gosling\textsuperscript{42} have shown
that although the propagation velocity of the aorta declines between the ages of 10 and 65 years, its value in the iliac artery is independent of age. Furthermore, in women more than 50 years old, iliac values are lower than aortic values. In men, this change appears to occur at the approximate age of 60 years.

We have observed a pronounced age-dependent increase in aortic/iliac velocity ratio (Table 2). This is due to a large increase in aortic propagation velocity with age and no significant change in iliac values. The only report we can find of an age-related increase in iliac artery stiffness\(^\text{43}\) can be questioned on technical grounds because the vessels were mounted vertically and allowed to increase in length during inflation.

In summary, in dogs and in young humans, the abdominal aorta is less stiff than the iliac artery; whereas in older people, the abdominal aorta becomes stiffer than the iliac due to a greater rate of increase with age.

It is worth noting that a pronounced age-related decline in propagation velocity is seen in the femoral\(^\text{24}\) and brachial arteries\(^\text{44}\) if the measurements are made at a fixed pressure. These observations suggest that increasing stiffness as a result of aging is an effect confined only to the aorta and the pulmonary arteries, and that the distensibility of the iliac artery, which is intermediate in structure between the "elastic" aorta and the more muscular distal vessels, changes little with age. These observations are also consistent with data presented by O'Rourke and colleagues.\(^\text{45,46}\) They reported reduced pulse-wave amplification with increasing age, although they argue that this is due to changes in peripheral vascular reflection rather than a reduction in regional elasticity differences.

The precise locations and, indeed, the existence of the major reflection sites in the human aorta have been questioned in several recent publications. Mills et al.\(^\text{18}\) and Murgo and colleagues\(^\text{19}\) have argued that the major reflection site in the abdominal aorta is located at the aortoiliac bifurcation, whereas more recently, a careful study by Latham et al.\(^\text{10}\) on subjects in the age range of 35–61 years has shown that no significant reflections occur between the renal arteries and the bifurcation. Our results confirm this latter observation, at least in specimens 30–60 years old. Although in younger subjects we observe a significant positive reflection, in older subjects, a negative value is seen.

The steady decline in the reflection coefficient with age was an unexpected observation. By calculating a local reflection coefficient from estimations of characteristic impedance (as we have done), Li\(^\text{2}\) has shown that the elasticities of the vessels on either side of the aortoiliac trifurcation in the dog are similar, that the area ratio is close to the theoretical optimum, and that the reflection coefficient is therefore close to zero. In the absence of any evidence to the contrary, it has been assumed that in humans, the velocity ratio similarly remains close to 1.\(^\text{47}\) Hence, on the basis of area ratio measurements alone, it has been argued that the junction becomes progressively more mismatched with age and the reflection coefficient becomes more positive.\(^\text{21}\) Our measurements confirm that the area ratio does decline with age; however, the concomitant increase in aortic stiffness and hence aortoiliac velocity ratio (see Equation 3) more than compensates for this.

Li's proposition that geometric effects predominate over elasticity in determining local reflection coefficients\(^\text{34}\) is based only on the form of the expression for characteristic impedance in which the radius term is squared and the elasticity term is a square root. Nevertheless, if the changes in elasticity are great enough, as shown in this study by the large increase in velocity ratio with age, then the magnitude of the local reflection coefficient will be dominated by changes in elasticity rather than geometry.

The hemodynamic effects of these changes in the aortoiliac reflection coefficient are complex. On the basis of modeling studies,\(^\text{48}\) it has been argued\(^\text{46}\) that reflections from peripheral sites, when measured near the heart, appear to originate from a single lumped site near the aortoiliac bifurcation. Together with the observation that vasodilator agents abolish or reduce the reflection,\(^\text{8,49}\) this argument has been used to justify the assertion that local reflection at the bifurcation is minimal. Attenuation of a reflected wave and its retrograde passage through numerous junctions along the aorta, however, would greatly reduce its amplitude. Thus, even in children and the aged in whom the absolute magnitude of this local reflection approaches 0.3, a reflected component would probably not be detectable near the heart. The local effect of such a wave cannot, however, be negligible. In the young, the positive reflection will amplify the measured wave as it approaches the junction. It is important to realize that this increased pulse pressure will also be propagated into the iliac arteries. In the old, the greatly increased stiffness of the abdominal aorta will amplify the pulse in its own right, whereas the negative value of \(r\) will partially offset this increase and reduce pulse pressure on either side of the junction. These processes might help to minimize the adverse effects of increased vascular stiffness in this part of the circulation due to age alone, chronic hypertension, and atherosclerosis.

To assess the individual effects of these factors will require measurements of local reflection coefficients in more subjects.

References
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KEY WORDS • impedance • propagation velocity • attenuation
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S E Greenwald, A C Carter and C L Berry

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