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Pulsatile Diameter and Elastic Modulus of the Aortic Arch in Essential Hypertension: A Noninvasive Study

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A noninvasive evaluation of the aortic arch diameter was performed in 16 subjects with sustained essential hypertension and in 15 normal subjects of similar age, gender and body surface area. In all subjects, measurements were obtained of brachial mean arterial pressure and pulse pressure, cardiac mass (judged on echocardiography) and carotid-femoral pulse wave velocity together with ultrasound determinations of aortic arch diastolic and systolic diameter (suprasternal window). For each subject, pulsatile change in aortic diameter, strain and aortic arch elastic modulus were calculated.

Compared with normal subjects, the hypertensive subjects showed an increase in aortic arch diameter (diastolic diameter 29.6 ± 1.0 versus 25.4 ± 1.0 mm, p < 0.01), in elastic modulus (1.071 ± 0.131 versus 0.526 ± 0.045 10^-5 N m^-2, p < 0.001) and pulse wave velocity (11.8 ± 0.5 versus 8.9 ± 0.3 m/s, p < 0.001). In the study group, a positive correlation was observed between diastolic aortic arch diameter and mean arterial pressure (r = 0.54, p < 0.01) and between elastic modulus and cardiac mass (r = 0.60, p < 0.01). Elastic modulus and age were positively correlated (r = 0.73, p < 0.01) in hypertensive but not in normal subjects (r = 0.08, NS).

This study is the first to demonstrate noninvasively that both the aortic arch diameter and the elastic modulus are increased in patients with sustained uncomplicated essential hypertension. These findings suggest that the increase in elastic modulus could influence the development of cardiac hypertrophy, and that both age and blood pressure act independently as factors that alter the arterial wall of subjects with sustained essential hypertension.

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Experimental studies (1–4) have shown convincingly that morphologic changes of large arteries occur in hypertension and affect arterial distensibility. However, clinical assessment of such changes in large vessels is difficult in hypertensive humans. Until now, elastic properties of the human aorta have been studied either postmortem or in vivo with invasive angiographic techniques (5–8). Noninvasive assessment of global rigidity of an arterial segment is also possible by measuring pulse wave velocity (9–11). The recent development of ultrasound devices has provided a new approach for evaluating arterial diameter and distensibility (12,13). Several studies using this technique have been reported but they have been limited to the study of straight superficial arteries such as the common carotid or the brachial arteries. Also, the pulsatile changes in arterial diameter have been poorly studied in hypertensive humans, especially when noninvasive techniques have been used (14,15). The aim of this study was to analyze the dimensions of the human thoracic aorta and to evaluate the pulsatile changes in aortic arch diameter with use of ultrasound track in normotensive and hypertensive subjects. With such measurements, it has been possible to evaluate the aortic arch elastic modulus in patients with essential hypertension.

Methods

Study groups (Table 1). This study was carried out on 31 subjects (24 male, 9 female) aged 20 to 52 years. Their mean age was 36 ± 2 years (±1 SEM). Fifteen subjects (11 male, 4 female) were in the normotensive group and 16 (13 male, 3 female) were in the hypertensive group. In the normotensive subjects, the systolic pressure was constantly ~140 mm Hg and the diastolic pressure <90 mm Hg. Nine of the 16 hypertensive subjects had never been treated; the remaining 7 had discontinued treatment at least 3 weeks before the study. All the hypertensive subjects had a diastolic pressure >95 mm Hg on at least three occasions during the untreated...
ambulatory period. Subjects with evidence of secondary causes of hypertension were excluded on the basis of a thorough clinical and biologic screening, as previously described (12,13). No signs, symptoms or history of heart and renal failure, coronary insufficiency or major diseases other than hypertension were present.

**Study design.** After giving informed consent to a detailed description of the procedure, the subjects were studied during a day's hospitalization, after a 20 min rest period in the supine position. The following measurements were then performed: blood pressure, carotid-femoral pulse wave velocity, systolic-diastolic variation of aortic arch diameter and echocardiographic variables.

**Arterial blood pressure** was measured with the right arm held at the midhilaracic level with use of a mercury sphygmomanometer. The mean value of at least three measurements was used. Diastolic pressure was evaluated as phase 5 of Korotkov sounds. Pulse pressure was calculated as the sum of diastolic pressure and one-third of pulse pressure.

**Determination of carotid-femoral pulse wave velocity.** Pulse wave velocity is a variable derived from measurements of pulse transit time and the distance traveled by the pulse wave between two recording sites (9–11). One pulse transducer probe (Siemens Elema AB) was fixed on the skin over the most prominent part of the right carotid and femoral arteries successively. The pulse transducer probe was connected to a recording device (Siemens Mingograph 4). The carotid and femoral waves and electrocardiogram (ECG) were recorded on paper at a speed of 100 mm/s. The time between the top of the QRS complex and the foot of each wave, which contains the high frequency information, was determined. The foot, identified as the point at which the sharp systolic upstroke began, was defined by the point of separation between the tangent of the upward part and the upward part of the wave itself. The time delay was averaged over at least one respiratory cycle of 10 beats. Estimation of the distance between the position of transducers was given by surface measurements of the distance between the two recording sites. Pulse wave velocity was the ratio between the distance and the time delay.

In this study, the variability of the method was assessed by measuring pulse wave velocity before and after administration of placebo in 11 healthy volunteers. Measurements were performed at 9 AM and 12 AM, the placebo having being given at 9:15 AM. Mean arterial pressure and pulse wave velocity did not change significantly, their respective pre- and postplacebo values being 83 ± 2 and 85 ± 2 mm Hg and 9.7 ± 0.5 and 9.1 ± 0.5 m/s (p = NS).

**Evaluation of changes in pulsatile aortic arch diameter.** An ultrasound M- and B-mode apparatus (CGR Sonel 3000) was used for these measurements. A 3.0 MHz transducer was placed on the suprasternal notch, with the patient resting in supine position with the head in slight extension (16,17). The aortic arch was primarily studied in the B-mode. Then the beam was adjusted to be in a plane perpendicular to the two edges of the vessel (16,17). The systolic-diastolic movements of the aortic arch walls were recorded on paper (Tektronix reprograph) during a period of 8 to 10 beats at three different times. The ECG was recorded simultaneously allowing the changes in diameter with the cardiac cycle to be plotted. Ten measurements of diastolic and systolic diameter were averaged and the diameter change was calculated as the difference between the systolic and diastolic diameter averages. For technical reasons, these measurements were not made on five hypertensive patients (24%) and one control subject (7%), who were thus excluded from the study.

**The reproducibility of this method was studied by testing different techniques in five normotensive and five hypertensive subjects.** First, the intraobserver reproducibility was expressed as the standard deviation divided by the average of 10 measurements. The results were 2.2 ± 0.5% for diastolic diameter, 1.6 ± 0.2% for systolic diameter and 16.3 ± 2% for the pulsatile change in diameter. Intraobserver reproducibility was evaluated 3 h after a first measurement, and expressed as the difference between the first and the second value divided by the first value. The results were 3.6 ± 0.5% for diastolic diameter, 2.4 ± 0.5% for systolic diameter and 23 ± 5% for the pulsatile change in diameter. For the analysis of echocardiographic recordings, the interobserver reproducibility was 1.2 ± 0.4% for diastolic diameter, 1.5 ± 0.2% for systolic diameter and 11.2 ± 2.3 for the pulsatile change in diameter. For analysis of the overall recordings, the study was performed by two physicians without knowledge of the subjects' blood pressure.

**Aortic strain** was then calculated as the ratio between the change in systolic-diastolic diameter divided by diastolic diameter (dD/DD) and expressed as a percentage. The pressure strain elastic modulus was defined as dP × DD/dD
femoral (PWVcf) pulse wave velocities were measured in 33 (8,15,18-20). Because pulse pressure was measured in the (where \(dP = \text{pulse pressure}\)) expressed in \(10^5 \text{N.m}^{-2}\) (8,15,18-20). The brachioradial (PWVhr) and carotid femoral (PWVcf) pulse wave velocities were measured in 33 normotensive and 93 hypertensive subjects in a similar age range (20 to 50 years). With the use of a simple transmission line model, values of pulse wave velocity in the aorta and brachial artery were used to estimate the amplification of the pulse due to nonuniform arterial elasticity (11). Indeed the “water hammer” formula (20) gives the relation between pulse wave velocity (PWV), characteristic impedance (ZO, impedance in the absence of wave reflection) and blood density (\(\rho\)) as:

\[\text{PWV} = \frac{Z_0}{\text{dist}}.\]

For a lossless transmission line that alters its characteristic impedance along its length, the ratio of proximal (Pp) to distal (Pd) pressure amplitude is proportional to the square root of the characteristic impedance (21): \(\frac{P_p}{P_d} = \sqrt{Z_0 \text{prox}}/Z_0 \text{dist.}\) Hence, in terms of pulse wave velocity:

\[\frac{P_p (\text{aorta})}{P_d (\text{brachial})} = \sqrt{\text{PWVcf/PWVhr}}.\]

This ratio was 0.96 ± 0.03 and 0.95 ± 0.04, respectively, in normotensive and hypertensive subjects (\(p = \text{NS}\)) suggesting that within the age range, the aortic pulse pressure could be approximated to the brachial pulse pressure for noninvasive measurements performed under baseline conditions.

Echocardiography. M-mode echocardiography was performed in 15 normotensive and 12 hypertensive subjects with the same device that is used for assessing the aortic diameter. Each patient was studied in the left lateral position to obtain good visualization of the left ventricular internal diameter, left interventricular septal thickness and left ventricular posterior wall thickness. The transducer was placed in the third or the fourth left parasternal interspace near the left sternal edge. All diameter and thickness measurements were made in each tracing with use of the leading edge technique and following the usual recommendations of the American Society of Echocardiography (22), as previously described (23). Echocardiographic left ventricular mass was estimated according to classical formula (24). The readings were performed by two observers in a double-blind fashion. Agreement between the two and reproducibility of the readings by the same observer were within 1 mm, as previously described (23,25).

Statistical study. Results were expressed as mean values ± 1 SEM (26). For basal values, differences in means were assessed by the Student’s \(t\) test and the unpaired \(t\) test was used to compare these values. Partial and total correlation coefficients were calculated by standard analytic methods.

### Results

Aortic arch diameter (Table 2). Table 2 shows that the sigmoid aortic diastolic diameter was not significantly different in the two groups of subjects, but aortic arch diastolic and systolic diameters were significantly higher in hypertensive subjects (\(p < 0.01\) and \(p < 0.05\)). For diastolic diameter, the hypertensive diameter reached a value about 1.2 times the normotensive value (29.6 ± 1.0 versus 25.4 ± 1.0 mm, \(p < 0.01\)). Mean diastolic diameter was the same when hypertensive subjects were classified into two groups: those who had never been treated and those whose treatment had been discontinued ≤3 weeks before treatment. In normotensive subjects, diastolic diameter and body surface area were positively \((r = 0.66)\) and significantly \((p < 0.01)\) correlated, but there was no significant correlation observed in the hypertensive subjects (Fig. 1). However, values for 13 of the 16 hypertensive subjects were on or above those on the normotensive curve. Diastolic diameter was positively and significantly correlated with age in both normotensive \((r = 0.66, p < 0.01)\) and hypertensive \((r = 0.59, p < 0.02)\) subjects; in 13 of the 16 hypertensive subjects, diastolic diameter was on or above the normotensive curve. Figure 2 shows a positive correlation between diastolic aortic arch

### Figure 1. Relations of diastolic aortic diameter with body surface area. For the aortic diameter-body surface area relation, a significant correlation was observed only in the normotensive subjects (open circles, \(p = <0.01, r = 0.66, n = 13\)). Thirteen of the 16 hypertensive subjects (solid circles) were on or above the normotensive curve.

### Table 2. Geometric and Functional Characteristics of the Aorta in the Subject Groups

<table>
<thead>
<tr>
<th></th>
<th>Normotensive Group (n = 15)</th>
<th>Hypertensive Group (n = 16)</th>
<th>(p) Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sigmoid diastolic aortic diameter (mm)</td>
<td>29 ± 1.2</td>
<td>31.7 ± 1.3</td>
<td>NS</td>
</tr>
<tr>
<td>Aortic arch</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Diastolic diameter (mm)</td>
<td>25.4 ± 1.0</td>
<td>29.6 ± 1.0</td>
<td>&lt;0.01</td>
</tr>
<tr>
<td>Systolic diameter (mm)</td>
<td>28.2 ± 1.0</td>
<td>31.9 ± 1.0</td>
<td>&lt;0.05</td>
</tr>
<tr>
<td>Strain (%)</td>
<td>11.6 ± 1.0</td>
<td>8.0 ± 0.6</td>
<td>&lt;0.01</td>
</tr>
<tr>
<td>Elastic modulus ((10^5 \text{N.m}^{-2}))</td>
<td>0.526 ± 0.065</td>
<td>1.071 ± 0.131</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>Pulse wave velocity ((\text{ms}^{-1}))</td>
<td>8.9 ± 0.3</td>
<td>11.8 ± 0.5</td>
<td>&lt;0.001</td>
</tr>
</tbody>
</table>

Values are mean ± SEM.
Diameter and mean arterial pressure in the overall study group (normotensive and hypertensive subjects) \( (r = 0.54, p < 0.01) \).

**Aortic strain and elastic modulus** (Table 2). Aortic strain was significantly lower in hypertensive than in normotensive subjects \( (p < 0.01) \). The hypertensive elastic modulus was significantly increased and reached a value that was about twofold the normotensive value \( (1.071 \pm 0.131 \text{ versus } 0.526 \pm 0.045 \text{ N m}^{-2}; p < 0.001) \). Elastic modulus was significantly correlated with age in hypertensive \( (r = 0.73, p < 0.01) \) but not in normal subjects \( (r = 0.08) \). As figure 3 indicates, at any given age, elastic modulus was higher in hypertensive than in normotensive subjects.

**Carotid-femoral pulse wave velocity** (Table 2). This variable was significantly increased in the hypertensive group \( (p < 0.001) \) and was approximately 1.3 times the normotensive value \( (11.8 \pm 0.5 \text{ versus } 8.9 \pm 0.3 \text{ m/s}) \). Elastic modulus and pulse wave velocity were positively \( (r = 0.45) \) and significantly \( (p < 0.02) \) correlated.

**Cardiac mass** (Table 3). As previously reported (23,27,28), cardiac mass was increased in hypertensive subjects. In the overall study group, a positive correlation was observed between cardiac mass index and mean arterial pressure \( (r = 0.79, p < 0.001) \), pulse wave velocity \( (r = 0.62, p < 0.01) \), diastolic diameter \( (r = 0.48, p < 0.02) \) and elastic modulus \( (r = 0.60, p < 0.01) \) (Fig. 4). The latter correlation was observed even for constant age, mean arterial pressure and pulse wave velocity.

**Discussion**

**Limitations.** The principal limitation of this investigation was the quality of the echocardiographic signal, which was not sufficient in six subjects (see Methods) and caused exclusion from the study. The accuracy of measurement with use of this device was within about 0.2 mm and the reproducibility of measurements of systolic and diastolic diameter was good and never >4%. However, the reproducibility of the change in diameter was not as good, (range 16 to 23%) because the value for pulsatile change in diameter is only about 10% of that for systolic or diastolic diameter and is calculated as the difference between these diameters. Indeed, similar kinds of errors are obtained when invasive techniques are used (5-8).

**Increased aortic arch diameter in hypertension.** The most important finding of this study was the increase in aortic arch diameter observed in the hypertensive group, in contrast with the near normal values observed for the sigmoid aortic diameter (23). Comparable results have been reported by other observers using angiographic or M-mode echocardiographic techniques (16,29,30). However, with noninvasive methods, the larger population of subjects studied allowed a

| Table 3. Echocardiographic Variables in the Two Subject Groups |
|-------------------------|-------------------------|-----------------|-----------|
|                         | Normotensive Group      | Hypertensive Group | p Value   |
| Interventricular septal thickness (mm) | 9.0 ± 0.3 | 10.9 ± 0.4 | <0.001 |
| Posterior wall thickness (mm)     | 8.7 ± 0.3 | 10.4 ± 0.4 | <0.001 |
| Left ventricular diastolic diameter (mm) | 50 ± 1 | 50 ± 1 | NS |
| Left ventricular systolic diameter (mm) | 32 ± 1 | 33 ± 1 | NS |
| Fractional shortening (%)      | 35 ± 1 | 35 ± 2 | NS |
| Left ventricular mass (g)       | 185 ± 12 | 372 ± 14 | <0.001 |
| Left ventricular mass index (g m⁻²) | 103 ± 5 | 192 ± 7 | <0.001 |

Values are mean ± SEM.
more accurate assessment of the increased aortic diameter in hypertensive subjects. Because, in normal subjects, aortic diameter is positively correlated with body surface area, it was important to demonstrate that, at any given value of body surface area, the aortic diameter was already increased in hypertensive subjects (Fig. 1). In addition, although aortic diameter is known to increase with age (18), this factor did not seem to be the only explanation of the increased arterial diameter seen in hypertensive subjects. Finally, a positive relation was observed between aortic diameter and mean arterial pressure (Fig. 2). Because mean aortic diameter was the same in hypertensive subjects who had never been treated and in those who had discontinued treatment ≤3 weeks before the investigation, it does not seem likely that the positive relation was due to the short-term increase in blood pressure during the 3 weeks of nontreatment. It seems more likely that the mechanical effect of the long-term elevation of blood pressure was the most important factor contributing to the increase in arterial diameter in subjects with hypertension. However, the elevation of mean blood pressure (from 85 ± 6 mm Hg in normotensive to 121 ± 2 mm Hg in hypertensive subjects) was much larger than the increase in arterial diameter, suggesting that active intrinsic modifications of the arterial wall might occur, thus preventing a larger increase in arterial diameter in hypertensive subjects (18,31).

**Role of changes in the arterial wall.** In an attempt to resolve this problem, several investigators (8,37) have studied aortic abnormalities in hypertensive patients before and after normalization of blood pressure after the administration of sodium nitroprusside. The fact that nitroprusside normalized all the hemodynamic abnormalities suggested that they were related to a smooth muscle-mediated increase in vascular tone (32). However, such findings are difficult to interpret, because nitrates are known to produce not only arteriolar dilation and blood pressure reduction but also active increase in arterial diameter (33). Perhaps the most important arguments in favor of intrinsic modifications of the aortic arch in hypertensive subjects result from two observations. First, structural and functional modifications of the aorta have been described for a long time in animal hypertension (1–4). Second, it has been previously shown that intrinsic modifications of the arterial wall develop in several regional circulations in hypertensive humans. For the same mean arterial pressure (12,13), brachial artery diameter is slightly increased, whereas common carotid artery diameter invariably remains normal in hypertensive persons. However, in this study on the aortic arch, it cannot be proven that the findings are due to changes within the arterial wall, and the exclusive role of the elevated blood pressure itself cannot be excluded.

**Aortic strain, elastic modulus and pulse wave velocity.** In this study, the differences between normotensive and hypertensive subjects for strain as well as for elastic modulus were much greater than the reproducibility limits of the technique used. These differences cannot be easily explained by changes in cardiac function and stroke volume because ventricular diameters and percent fractional shortening were similar in the two groups. Thus, the modifications of strain and elastic modulus in hypertensive subjects indicate modifications within the arterial wall. All studies on the assessment of elastic modulus, whether in vivo or in vitro, describe the technical difficulties of such measurements (8,15,18,19,34–36). The variable often used to evaluate arterial elastic properties is the “incremental” modulus or Young’s modulus, which necessitates the evaluation of the arterial wall thickness (34–36), a variable that is practically impossible to measure in living human beings. For this reason, we used the elastic modulus proposed by Peterson et al. (19). Elastic modulus is linearly related to Young’s modulus by the radius-wall thickness ratio, which varies with age and pressure. The different methods for assessing regional elasticity of a given arterial segment make comparisons among reported studies difficult.

There are few data (7,8,35) available on the thoracic aorta elastic modulus in living human subjects. In 43 normotensive subjects, Gozna et al. (7) measured pulsatile change in the thoracic aorta using angiography and found a value that, when matched with that of subjects in the age range of our study group, was 0.618 10⁵ N·m⁻². This value corroborates the present finding (0.526 10⁵ N·m⁻²), which was obtained using a noninvasive technique. Our values of strain and elastic modulus were also close to those given by Barnett et al. (37), who reported, respectively, 17% and 0.466 10⁵ N·m⁻² in the ascending aorta of men. With the use of angiography, Merillon et al. (8) found 12 and 8.4% for strain and 0.33 10⁵ N·m⁻² and 0.69 10⁵ N·m⁻², respectively, in normotensive and hypertensive patients. Their values of strain were close to our own results but the values of elastic modulus were slightly lower, although we reported the same ratio (of about 2) between hypertensive and normotensive subjects. The discrepancies in absolute values of elastic
modulus can be partly explained by the technique used for blood pressure measurement. We used the auscultatory method to determine pulse pressure, whereas Merillon et al. (8) measured it directly using an intraaortic catheter. The differences in the pressure contour in the aorta and the brachial artery are well known and it is possible that we slightly overestimated the value of aortic pulse pressure. However, we have shown under Methods that brachial pulse pressure gives an acceptable approximation of intraaortic pulse pressure because the square root of the brachial-aortic pulse wave velocity ratio did not differ in hypertensive and normotensive subjects and was close to 1 (11).

With this taken into consideration, our study was the first to use an ultrasound method to measure the viscoelastic property of the thoracic aorta in living human beings. Imura et al. (15) also used an ultrasound device to measure the elastic modulus of the abdominal aorta and found a value about 1.19 10^5 Nm^-2, when extrapolated to the age of subjects in our study group. Studies in dogs (34-36) show that abdominal aortic elastic modulus is about twice the value of thoracic aorta elastic modulus. The value reported by Imura et al. (15) at the abdominal level is also approximately twofold the value we found at the thoracic level. This observation probably explains the low correlation coefficient of the relation that we observed between elastic modulus and pulse wave velocity. Whereas elastic modulus was assessed exclusively in the aortic arch, the carotid-femoral pulse wave velocity was a reflection of an arterial elasticity coefficient involving both the thoracic and the abdominal aorta.

**Elastic modulus, age and cardiac mass.** Because the study was performed in subjects within a narrow age range, no significant correlation was observed between elastic modulus and age in normotensive control subjects. In contrast, a strong positive correlation was observed in the hypertensive subjects, suggesting that the elevated blood pressure accelerated the effects of age on the rigidity of the arterial wall (18). Such a role of the aging process might have consequences not only on large vessels but also on cardiac structure and function. As previously reported (8, 23, 38), the degree of cardiac hypertrophy in hypertension is affected independently by the level of blood pressure and vascular resistance on the one hand and by the reduction in aortic distensibility on the other hand. In this investigation, the finding of a positive correlation between elastic modulus and cardiac mass is further confirmation of this possibility (Fig. 4).

**Conclusions.** This study has shown that both the geometry and the function of the aortic arch are significantly modified in patients with sustained untreated essential hypertension. Such modifications not only reflect significant alterations in the buffering function of hypertensive large vessels, but also may have important effects on the structure and function of the heart.

References


