Altered Dependence of Aortic Pulse Wave Velocity on Transmural Pressure in Hypertension Revealing Structural Change in the Aortic Wall
Nicholas R. Gaddum, Louise Keehn, Antoine Guilcher, Alberto Gomez, Sally Brett, Philipp Beerbaum, Tobias Schaeffter, Philip Chowienczyk

Abstract—Aortic pulse wave velocity (aPWV), a major prognostic indicator of cardiovascular events, may be augmented in hypertension as a result of the aorta being stretched by a higher distending blood pressure or by a structural change. We used a novel technique to modulate intrathoracic pressure and thus aortic transmural pressure (TMP) to examine the variation of intrathoracic aPWV with TMP in hypertensive (n=20; mean±SD age, 52.1±15.3 years; blood pressure, 159.6±21.2/92.0±15.9 mm Hg) and normotensive (n=20; age, 55.5±11.1 years; blood pressure, 124.5±11.9/72.6±9.1 mm Hg) subjects. aPWV was measured using dual Doppler probes to insonate the right brachiocephalic artery and aorta at the level of the diaphragm. Resting aPWV was greater in hypertensive compared with normotensive subjects (897±50 cm/s versus 784±43 cm/s; P<0.05). aPWV was equal in hypertensive and normotensive subjects when measured at a TMP of 96 mm Hg. However, dependence of aPWV on TMP in normotensive subjects was greater than that in hypertensive subjects (9.6±1.6 versus 3.8±0.7 cm/s per mm Hg increase in TMP, respectively, means±SEM; P<0.01). This experimental behavior was best explained by a theoretical model incorporating strain-induced recruitment of stiffer fibers in normotensive subjects and fully recruited stiffer fibers in hypertensive subjects. These results explain previous contradictory findings with respect to isobaric aPWV in hypertensive compared with normotensive subjects. They suggest that hypertension is associated with a profound change in aortic wall mechanical properties possibly because of destruction of elastin leading to less strain-induced stiffening and predisposition to aortic dissection. (Hypertension. 2015;65:362-369. DOI: 10.1161/HYPERTENSIONAHA.114.04370.)

Key Words: aortic dissection ■ collagen ■ elastin ■ pulse wave analysis ■ vascular stiffness
Methods

Subjects

Patients with essential hypertension were recruited from the hypertension clinic at Guy's and St Thomas' Hospital. Hypertension was diagnosed as an office blood pressure >140/90 mm Hg on ≥3 occasions and ambulatory blood pressure >130/85 mm Hg. Patients on treatment for hypertension were included if office blood pressure on treatment was >140/90 mm Hg. Exclusion criteria included intercurrent illness, pregnancy, significant systemic disease other than mild hypertensive nephropathy, echocardiographic evidence of left ventricular ejection fraction <50%, and rhythm other than sinus rhythm. Subjects were also excluded if high-quality Doppler waves could not be recorded from the aorta. Healthy volunteers on no regular medication of a similar age and sex distribution to hypertensive subjects were recruited by advertisement from the local community.

The study was approved by the London Westminster Research Ethics Committee, and written informed consents were obtained from all participants. Subject characteristics are summarized in the Table. Drug treatment in hypertensive subjects consisted of angiotensin-converting enzyme inhibitors/angiotensin receptor antagonists (5/20 subjects), calcium channel antagonists (7/20), diuretics (4/20), β-adrenergic receptor antagonists (3/20), and α-adrenergic receptor antagonists (2/20).

Doppler Ultrasound Measurement of Intrathoracic aPWV During Modulation of ITP

aPWV was measured using a dual probe, continuous wave, Doppler ultrasound system developed at our institute. Four megahertz and 2 MHz bidirectional probes (VP4-HS and OP2-HS; Huntleigh Healthcare Ltd, Cardiff, UK) were used to insonate the right brachiocephalic artery and aorta at the level of the diaphragm (Figure 1). Beat-to-beat transit time between the 2 sites was measured as described below, path length estimated from the distance from sternal notch to xiphisternum and beat-to-beat aPWV calculated from path length/transit time.

Mouth pressure (which under conditions of zero flow equals intrapulmonary pressure) was varied by asking subjects to perform Valsalva and Mueller maneuvers (Figure 1) with their mouth sealed around a mouthpiece (Spiroguard, Air Safety Limited, Morecambe, UK) connected to an occluded airway with a small leak, to ensure that mouth pressure during the maneuver was generated in the thoracic cavity. Mouth pressure was measured by a pressure sensor (Omega, Irlam, Manchester, UK) connected just proximal to the occluded airway.12 Although ITP on the adventitial surface of the aorta differs by some recordings were thus of higher quality than others. One of each test. The Finapres recording was then calibrated according to a linear interpolation in time between these 2 points.

The maneuvers, notably Valsalva, tended to displace the descending aorta affecting the signal amplitude of the distal waveform, and some recordings were thus of higher quality than others. One of each of the 3 recordings (free breathing, Valsalva, and Mueller maneuvers) was therefore manually selected based on the amplitude and quality of the velocity signals by an observer blinded to the final results. This allowed an aPWV versus TMP profile to be compiled for each subject.

Signal Processing

Doppler ultrasound signals and pressure signals were acquired using an analog-to-digital conversion card (USB-6211; National Instruments, Austin, TX), and subsequent signal processing was done with software developed at our institute written in MATLAB (The MathWorks, Natick, MA). Successive discrete Fourier transforms were performed on 20-ms windows of the raw Doppler data (sampling frequency 8000 Hz) applied in a sliding sample fashion with 40% overlap of window length. A Hamming window filter was applied to each discrete window sample to minimize spectral leakage.16 This approach optimized the trade-off between temporal resolution of velocity profiles and signal-to-noise ratio and provided a temporal resolution of 8 ms. Velocity waveforms were filtered with a Savitzky-Golay filter (which maintains high frequency components of the waveform) before identifying the foot of the waveform at end diastole using an early systolic least squares method to minimize transit time measurement error and variability.17 TMP was obtained from the difference between MAP and ITP, and aPWV values averaged over 10 mm Hg ranges of TMP to provide an aPWV versus TMP profile for each subject and mean aPWV versus TMP profiles for the

Table. Characteristics of Normotensive and Hypertensive Subjects

<table>
<thead>
<tr>
<th>Measure</th>
<th>Normotensive</th>
<th>Hypertensive</th>
</tr>
</thead>
<tbody>
<tr>
<td>Number (males)</td>
<td>20 (7)</td>
<td>20 (9)</td>
</tr>
<tr>
<td>Age, y</td>
<td>55.2±11.1</td>
<td>52.1±15.3</td>
</tr>
<tr>
<td>Height, cm</td>
<td>171.2±10.4</td>
<td>170.5±10.0</td>
</tr>
<tr>
<td>Weight, kg</td>
<td>66.4±12.7</td>
<td>77.3±15.2</td>
</tr>
<tr>
<td>SBP, mm Hg</td>
<td>124.5±11.9</td>
<td>159.6±21.2*</td>
</tr>
<tr>
<td>DBP, mm Hg</td>
<td>72.6±9.1</td>
<td>92.0±15.9*</td>
</tr>
<tr>
<td>MAP, mm Hg</td>
<td>89.9±8.6</td>
<td>114.6±16.8*</td>
</tr>
<tr>
<td>Total cholesterol, mmol/L</td>
<td>5.3±0.8</td>
<td>5.6±1.3</td>
</tr>
<tr>
<td>HDL cholesterol, mmol/L</td>
<td>1.8±0.6</td>
<td>1.6±0.4</td>
</tr>
<tr>
<td>Triglycerides, mmol/L</td>
<td>1.0±0.4</td>
<td>1.4±0.8</td>
</tr>
</tbody>
</table>

Values are means±SD. DBP indicates diastolic blood pressure; HDL, high density lipoprotein; MAP, mean arterial pressure; and SBP, systolic blood pressure.*P<0.0001 compared with controls.
normotensive and hypertensive groups. Heart rate for each subject was averaged over the same 10 mm Hg ranges of TMP.

**Statistical Methods**

Subject characteristics are presented as means±SD and results as means±SEM. Variation of aPWV with TMP was assessed by a multiple regression model incorporating age, sex, hypertension (as a dummy variable), and an interaction between TMP and hypertension to test for differing aPWV versus TMP gradients in hypertensive and normotensive subjects. All results were adjusted for subject age. All analysis was performed in SPSS (version 19). All tests were 2-tailed, and \( P < 0.05 \) was considered significant.

**Theoretical Model**

To explain observations in the clinical data, the relationship between aPWV and TMP was analyzed theoretically. This incorporated the classical stress-strain relationship applied to a thin-walled cylinder relating TMP to distension, then an elastic modulus model for aortic wall tissue, and finally the Bramwell-Hill equation to relate aPWV to mechanical properties. The following provides a brief description of the theoretical relationship between aPWV and TMP.

The hoop stress (\( \sigma \)) in the aortic wall caused by TMP induces an elastic strain (\( \varepsilon \)) according to,

\[
\sigma = E\varepsilon, \tag{1}
\]

where \( E \) is the Young modulus of the aortic wall (for a composite elastic and collagen or homogeneous elastic material). Applying this to a thin shell cylinder model, the following relationship between TMP and aortic diameter \( \Omega \) can be derived:

\[
\text{TMP in } \frac{c_0}{h} \left( \frac{1}{2} \varepsilon_0 - \varepsilon_0 \right) = \frac{\varepsilon_0}{h}, \tag{2}
\]

where \( h \) is the wall thickness and \( \varepsilon_0 \) is the unloaded diameter (TMP of 0).

Depending on the assumed structure of the aortic wall, either a composite nonlinear elastance (NLE) model or a linear elastance (LE) model may be assumed. The NLE model assumes that resistance to aortic wall strain is provided by elastin and incrementally recruited collagen fibers, whereas the LE model accommodates aortic wall strain by a homogeneous elastic tissue with a constant Young modulus.

In the NLE model, aortic stiffness (\( E \)) comprises the composite stiffness of the elastin/collagen in the media and intima. The stiffness of a 2-species composite material (\( E_{\text{inc}} \)) is evaluated as the area fraction–weighted sum of each species’ stiffness. Assuming homogeneity along the vessel length, this area fraction can be replaced by a through-thickness fraction,

\[
E_{\text{inc}} = \frac{h}{h} E_r + \frac{h}{h} E_c = pE_r + (1-p)E_c, \tag{3}
\]
where \( h_1 \) and \( h_2 \) are the proportions of the effective cumulative thicknesses of elastin fibers and collagen fibers, respectively, divided by the thickness \( h \) (ie, \( h_1 + h_2 = 1 \)), and \( p = h_1 / h \). Within the range of aortic wall strain assessed, we assumed the stiffness of elastin to be constant and the stiffness of collagen to be incrementally recruited as modeled by Bank et al\(^2\) (Figure 2). This assumes that \( E_c \) increases linearly with strain as previously reported\(^2\) (\( E_c \))

\[
E_{inc} = pE_e + (1 - p)E_c \left( \frac{\varnothing - \varnothing_0}{\varnothing_0} \right).
\]

In the LE model, \( E_{inc} \) is treated as constant and therefore independent of diameter (ie, \( p = 1 \)).

In Bramwell and Hill original development of the Moens-Korteweg equation, \( aPWV \) is derived in terms of the relationship between changes in blood volume per unit length and changes in arterial pressure.\(^2\) This can be simply modified to relate changes in diameter to changes in arterial pressure,

\[
aPWV = \sqrt{\frac{\frac{dT}{MP} d(TMP)}{\rho \varnothing}} = \sqrt{\frac{32E_h}{\varnothing_0}} \frac{dT}{MP} \frac{dT}{\varnothing}.
\]

Expressions for \( \varnothing \) and \( d(TMP)/d\varnothing \) can be found by rearranging, and then differentiating Equation 2, with respect to \( \varnothing \), respectively, using either the NLE or the LE models. Substituting these into Equation 5, \( aPWV \) for a composite NLE, \( (PWV_{NLE}) \), aorta model can be expressed as,

\[
aPWV_{NLE} = \sqrt{\frac{\gamma h \left[ 4 \left( 3E \varnothing_0 - 2E_c \left( \varnothing - \varnothing_0 \right) + 2E_c \gamma \right) \right]}{15\varnothing_0^2 \rho}}.
\]

where,

\[
\gamma = \frac{32E_h}{32E_c},
\]

\[
\varnothing_0 = \frac{32E_h \varnothing - 24E_h + 15}{64E_h^2 + 5\varnothing_0(E_c(TMP))}
\]

Then using the LE model (\( E_{inc} \) =constant), \( aPWV \), \( (PWV_{LE}) \), can be expressed as,

\[
aPWV_{LE} = \sqrt{\frac{3(TMP)\varnothing_0 + 8E_{inc}h}{6\varnothing_0 \rho}}.
\]

To examine the potential influence of altered elasticity of elastin and collagen, the same geometric parameters (\( h, \varnothing_0 \)) were applied to both LE and NLE; \( h=2.5 \) mm and \( \varnothing_0=1.5 \) mm. Young moduli were then selected to indicate flexible elastin (\( E_e = 0.1 \) MPa), a stiffer collagen (\( E_c = 1.5 \) Mpa) for the NLE model; and a moderately stiff network of extended collagen, residual elastin, and extracellular matrix (\( E = 0.475 \) MPa) for the LE model.

**Results**

**Clinical Study**

Resting \( aPWV \) (at a TMP equal to resting MAP) was greater in hypertensive compared with normotensive subjects (897±50 cm/s versus 784±43 cm/s; \( P<0.05 \)). Examples of beat-to-beat variation of \( aPWV \), MAP, ITP, and TMP during a Mueller maneuver of gradually increasing intensity for a typical normotensive and hypertensive subject are shown in Figure 3. Changes in MAP were small in comparison with those of ITP, and therefore, changes in TMP were determined mainly by changes in ITP. In normotensive subjects, as ITP decreased and thus TMP increased, \( aPWV \) increased (Figure 3, left). However, in hypertensive subjects, there was little change in \( PWV \) with TMP (Figure 3, right). The dependence of mean, age-adjusted, \( aPWV \) on TMP in the normotensive and hypertensive groups is shown in Figure 4. When measured at fixed TMP of 96 mm Hg, \( aPWV \) was equal in hypertensive and normotensive subjects. However, dependence of \( aPWV \) on TMP was significantly greater in normotensive compared with hypertensive subjects (9.6±1.6 cm/s per mm Hg and 3.8±0.7 cm/s per mm Hg in normotensive and hypertensive subjects, respectively; \( P<0.01 \)), leading to an intersection of the \( aPWV \) versus TMP relationships for hypertensive and normotensive subjects. Thus, isobaric \( aPWV \) was higher, equal, or lower in hypertensive subjects compared with normotensive subjects, depending on the level of TMP at which it was assessed (Figure 4). There was a modest change in heart rate with modulated TMP (3.1±0.70 bpm per 10 mm Hg). However, the difference in dependence of \( PWV \) on TMP between hypertensive and control subjects
was similar, irrespective of whether results were or were not adjusted for heart rate.

Because the wide range of measured aPWV and resting blood pressures (and hence resting TMP) within both normotensive and hypertensive groups might reduce the dependence of aPWV on TMP, we examined variation from each individual’s resting state, that is, the variation from resting aPWV and resting TMP (Figure 5). This revealed greater discrimination between hypertensive and normotensive groups in dependence of aPWV on TMP: dependence of aPWV on TMP in the hypertensive group (0.12±0.17% per mm Hg) was <20% of that in the normotensive group (1.16±0.14% per mmHg; P<0.01).

**Theoretical Model**

Assuming a nonlinear composite stiffness aortic wall model, as described in Equation 6, variation in elastin stiffness (Ee) or collagen through thickness fraction does not allow for an intersection of aPWV versus TMP profiles, as observed in the clinical study. Instead, increasing the portion of collagen (p) or stiffness of the elastin (Ee) displaces the aPWV-TMP profile along the aPWV axis (Figure 6, top and middle). However, modeling the hypertensive case as a homogeneous linear structure, such as a network of extended collagen, residual elastin, and extracellular matrix, allows the intersection observed clinically to be reproduced (Figure 6, bottom).

**Discussion**

Whether hypertension is associated with an alteration in the intrinsic elasticity of the aorta or results in a stiffening of the aorta merely through a greater transmural stretching force is a fundamental question which has received surprisingly little attention. Previous studies to address this issue have examined the relationship of arterial pressure to diameter of the common carotid artery throughout the cardiac cycle with extrapolation to a common pressure. This approach suggests that, when compared at the same operating pressure, stiffness of the common carotid artery, as represented by Young elastic modulus (E) is similar in hypertensive and normotensive subjects. However, the interpretation of such studies is limited by many factors. Comparison of distensibility at the same pressure relies on an overlap of arterial pressure in the 2 groups (ie, a measurement taken close to diastolic in the hypertensive group and close to systolic in the normotensive group). The interpretation of the result is, therefore, critically dependent on assumptions regarding the form of the pressure/diameter relation during the cardiac cycle and whether this differs in hypertensive and normotensive subjects. Calculation of E requires measurement of wall thickness, an assumption that the through-wall elasticity is constant and that intima-medial thickness (used to estimate the relevant wall thickness) is representative of the part of the wall that determines its overall elastic behavior. In reality, increased intima-medial thickness in hypertension may result from both intimal and medial thickening.24 Finally, although carotid stiffness is closely correlated to aortic stiffness, it is of less prognostic impact than aPWV.25

**Figure 4.** Dependence of aortic pulse wave velocity (aPWV) on transmural pressure in normotensive and hypertensive groups.
the interpretation of this work is limited by the range over which blood pressure can be modulated and possible effects of the drugs on arterial stiffness.

To our knowledge, this is the first study to have examined the relation between aPWV and TMP in vivo in man using a direct mechanical approach to modulate TMP, and the results reveal structural change in the properties of the aorta in hypertensive compared with normotensive subjects that cannot be appreciated from simple resting measures of aPWV. In normotensive subjects, the results demonstrate an increase in aPWV with increasing TMP, replicating results observed during mechanical manipulation of blood pressure in the canine aorta.6 By contrast, in hypertensive subjects, aPWV is seen to show little variation with TMP. Therefore, the higher potential for increased aortic wall stiffening with increased distending pressure in normotensive subjects, when compared with hypertensive subjects, means that a comparison of isobaric aPWV (ie, aPWV measured at a fixed TMP) will depend on the applied TMP. For example, if a TMP is applied, which is below the usual operating pressures in normotensive subjects, then a hypertensive subject is likely to have a higher isobaric aPWV compared with normotensive subjects, and at a TMP above the usual operating pressure in hypertensive subjects, the reverse is true. These results explain inconsistent findings with regard to isobaric stiffness in earlier studies.7–10

Figure 5. Change in aortic pulse wave velocity (aPWV) from resting values with change in transmural pressure (TMP) from rest in normotensive and hypertensive groups.

Increased aortic stiffening with increased TMP is usually attributed to progressive recruitment of collagen fibers, necessary to prevent overextension and rupture of the aortic wall. Our theoretical model of collagen recruitment (NLE model) reflects this increasing trend of wave speed, therefore stiffness, with TMP. Furthermore, the NLE model shows that as the proportion of collagen to elastin (p) or the Young modulus of elastin (E_e) increases, the aPWV versus TMP trend translates along the aPWV axis (a similar translation is also seen with variation of aortic wall thickness and diameter). The distinction we observed in hypertensive compared with normotensive subjects did not display this upward trend but rather rotation of the trend, indicating that the composite stiffness model of elastin and fractionally recruited collagen may not accurately represent our hypertensive group. Instead, a linear stiffness hypertensive model (LE) offered a better reflection of our clinical observations, indicative of a group whose artery wall maintains a constant stiffness over the range of distension examined in this study. In this case, a network of extended collagen, residual elastin, and extracellular matrix would operate mechanically more like a stiff, homogeneous material.

Figure 6. Theoretical variation in aortic pulse wave velocity (aPWV) vs transmural pressure (TMP) profiles in a nonlinear elastance (NLE) aorta model with varied elastin/collagen portions (top), and varied elastin stiffness, E_e (middle). Applying a composite NLE model as normotensive, and linear elastance model as hypertensive, an aPWV vs TMP intersection is theoretically possible. Model Parameters used were \( \Omega = 15 \) mm (unstretched), \( h = 2.5 \) mm, \( P = 0.6 \) elastin, and 0.4 collagen (measured in dogs). In the NLE model, \( E_e = 0.1 \text{ MPa} \) and \( E_c = 1.5 \text{ Mpa} \) and in the LE model, \( E = 0.475 \text{ MPa} \). See text for definition of parameters.
Irrespective of the exact model invoked to explain the observed relation of aPWV with TMP, our results strongly suggest that hypertension is associated with a structural change in the aortic wall away from a composite NLE model leading to higher stiffness at normal operating pressures. Furthermore, there is a failure of stiffening to increase with increased TMP in hypertension, which may predispose to dissection of the wall when subject to a surge in pressure above the usual operating pressure. Such a structural change could result from mechanical rupture of elastin and digestion of elastin through increased expression of matrix metalloproteinases. However, we cannot exclude the possibility that the structural change precedes or indeed is causal to the development of hypertension.

Several important limitations of our study should be noted: the majority of our subjects had long-standing hypertension. It is likely that results may differ in subjects with new-onset hypertension. Most subjects were on treatment that might have influenced vascular smooth muscle tone in the aorta. However, previous studies have indicated that vascular smooth muscle tone in elastic arteries has little direct influence on thoracic aortic stiffness. The respiratory maneuvers used to induce a controlled change in TMP are known to influence cardiac (contractility and stroke volume) and autonomic function over the course of the maneuver. Although theoretical considerations show aPWV to be independent of stroke volume, it is possible that changes in cardiac function could have influenced aPWV. Against this is experimental evidence that aPWV is not influenced by inotropic stimulation with dobutamine (other than through an influence on MAP) and the observation in the present study that aPWV tracked TMP throughout the respiratory maneuver (during which time cardiac output varies widely). aPWV is known to depend on heart rate. However, the dependence is relatively weak and the size of the heart rate effect observed in previous studies would result in relatively minor confounding, which would be equal in normotensive and hypertensive subjects.

**Perspectives**

The present findings explain previous contradictory findings with respect to isobaric aPWV in hypertensive compared with normotensive subjects. Differing dependence of aPWV on TMP in normotensive and hypertensive subjects means that differences in isobaric aPWV depend on the TMP at which this is assessed. They suggest that hypertension is associated with a profound change in aortic wall mechanical properties with less strain-induced stiffening in hypertension compared with normotension that may reflect destruction of elastin and predisposition to aortic dissection.

In conclusion, we describe a novel method for examining the relation of aPWV to TMP based on modulation of ITP through respiratory muscle activity that provides measures of isobaric aPWV and pressure dependence of aPWV. The relation of aPWV to TMP differs in hypertensive and normotensive subjects, implicating a change in mechanical properties of the aorta in hypertension.

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**Disclosures**

King’s College London has formed a spin-out company to develop technology in central blood pressure measurement (not used in the present study). P. Chowienczyk has an interest in this company.

**References**


What Is New?
- A novel method for examining the relation of aortic stiffness to transmural pressure providing measures of aortic pulse wave velocity (aPWV) at fixed pressure (isobaric aPWV) and pressure dependence of aPWV.
- Pressure dependence of aPWV differs in hypertensive and normotensive subjects explaining previous contradictory findings with regard to isobaric aPWV.

What Is Relevant?
- Hypertension is associated with less strain-induced stiffening compared with normotension that may reflect destruction of elastin and predisposition to aortic dissection.

Novelty and Significance

Measurement of aPWV during modulation of intrathoracic pressure by varying respiratory muscle activity allows the relationship of aPWV to transmural pressure to be assessed. Differing dependence of aPWV on transmural pressure in hypertensive and normotensive subjects means that differences in isobaric aPWV between hypertensive and normotensive subjects depend on the transmural pressure at which this is assessed. They suggest that hypertension is associated with a profound change in aortic wall mechanical properties with less strain-induced stiffening in hypertension compared with normotension that may reflect destruction of elastin and predisposition to aortic dissection.
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