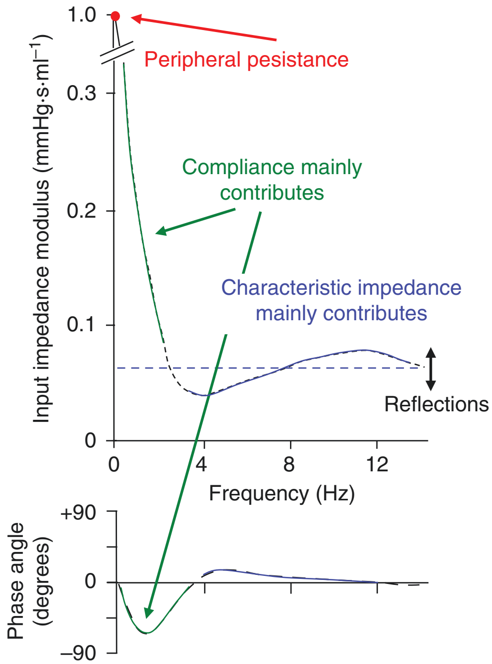
Chapter 24
Arterial Input Impedance



Input impedance completely and comprehensively describes an arterial system. Here the input impedance of the human systemic arterial tree is shown in schematic form. The ratio of the mean arterio-venous pressure drop and mean flow is systemic vascular resistance or peripheral resistance, R_p. To obtain information about the oscillatory aspects of the arterial system wave shapes of pressure and flow are used. To derive this information sinusoidal pressures and flows as obtained by Fourier analysis (see Appendix 1) are related. The amplitude ratio and the phase difference

N. Westerhof et al., *Snapshots of Hemodynamics*,
https://doi.org/10.1007/978-3-319-91932-4_24

of the sine waves of pressure and flow are calculated, giving the modulus and phase angle of the impedance (application of Ohm's law). The impedance modulus and phase are plotted as a function of the frequency, with at zero frequency systemic vascular resistance. For intermediate frequencies the impedance modulus decreases precipitously and the phase angle is negative. This shows the contribution of (total) arterial compliance, C. For high frequencies the modulus approaches a constant value and the phase angle is close to zero. This is the contribution of the aortic characteristic impedance, Z_c, which is a real quantity for large vessels. Thus the three elements R_p, C, and Z_c together give a good description of the input impedance. Without reflections in the arterial system, the input impedance would be equal to aortic characteristic impedance and the pressure and flow would have the same wave shape. For low frequencies the reflections at the many discontinuities, 'diffuse reflections', add and cause the impedance to be high. At high frequencies local, reflected waves return with different phases and contribute less causing the impedance to be close to the characteristic impedance. The oscillations about the characteristic impedance are related to distinct reflections, and depend on local geometry. The frequency of the minimum of the impedance modulus and the zero crossing of the phase angle have been used to calculate the effective reflection site (the quarter wave length rule). This calculation is often inaccurate. Input impedance can be derived for any (sub) section of the arterial system, but the present chapter mainly concentrates on the systemic arterial system as a whole.

24.1 Description

24.1.1 Definition of Impedance

Impedance is the relation between the pressure difference over and flow through a linear system, for sinusoidal signals. Impedance completely describes the whole system at its entrance, and therefore often called input impedance. Input impedance derivation is performed from the pulsatile pressure difference and pulsatile flow by applying of Fourier analysis and calculation of their harmonics (Appendix 1). Inversely, when the impedance is known, a given flow allows for the calculation of pressure in terms of magnitude and wave shape and vice versa. Systemic arterial and pulmonary arterial input impedance are a comprehensive description of the systemic and pulmonary arterial tree. Input impedances of organ systems may be derived as well.

24.1.2 Derivation of Input Impedance

In the calculations of input impedance, Z_{in}, we use both the mean values and the pulsatile part of the pressure and flow waves. We apply Fourier analysis of the aortic pressure and flow because impedance calculations can only be performed for sinusoidal signals (in the frequency domain). Fourier analysis is only permitted for a full beat, or a series of full beats. The details and limitations of Fourier analysis are discussed in Appendix 1. To derive impedance Ohm's law is applied as is done for the calculation of resistance from mean pressure and mean flow (Chap. 6). As discussed in Chap. 6, Ohm's law may only be applied to full beats and this also holds for the calculation of impedance. For each pair of sine waves of pressure and flow (called harmonics) we calculate the ratio of the amplitudes (impedance modulus) and the difference in phase angle (impedance phase angle). It is only allowed to perform these calculations if the system is in the steady state, time-invariant, and linear. Steady state means that the arterial system may not vary in time over the study period, e.g., vasomotor tone should be constant. Linearity in this context implies that if a sine wave of pressure is applied a sine wave of flow results. The calculation of impedance of a time-varying system does not lead to interpretable results. An example is the coronary circulation, where resistance and arterial compliance are time-varying over the heartbeat. For a nonlinear system the calculation of impedance also does not lead to interpretable results. An example is the calculation of 'impedance' from ventricular pressure and aortic flow, where the aortic valves make the system nonlinear, and the result is not meaningful. The arterial system is not perfectly linear but the variations of pressure and flow over the heartbeat are sufficiently small so that linearity is approximated and the derived input impedance is a meaningful description. The limitations of linearity and time-invariance also hold for the calculation of peripheral resistance, see Chap. 6. It has been shown that some of the scatter of the input impedance data (Fig. 24.1) results from nonlinearity [1, 2].

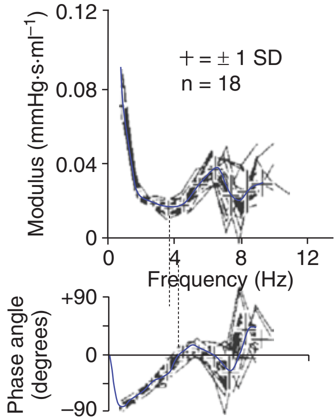


Fig. 24.1 Input impedance calculated from 18 heart beats. The scatter in the input impedance is caused, in part, by noise on the pressure and flow signals especially affecting the small amplitudes of high harmonics. Non-linearity of the arterial system also contributes to the scatter. The blue line is drawn through the averaged values. The vertical dashed lines indicate that the modulus minimum and phase zero crossing are not at the same frequency, implying that a single tube with real, in phase reflection is not a good model of the entire arterial tree and the quarter wave length rule may not be applied. Impedance of a patient with type A beat. (Adapted from Ref. [1], by permission)

Fourier analysis and subsequent calculation of the input impedance only gives information at frequencies that are multiples of the Heart Rate, i.e., the harmonics (Appendix 1). Because the information contained in the signals for high frequencies is small, the higher harmonics (Appendix 1) are subject to noise, so that the impedance at high frequencies often scatters considerably (Fig. 24.1). Analyzing multiple heartbeats and averaging the results decreases noise [1]. By pacing the heart at different rates or by analyzing long stretches of data in a steady state, the frequency resolution can be increased.

In the systemic circulation venous pressure may be neglected (Chap. 6), so that Fourier analysis of aortic pressure and flow gives a sufficiently accurate approximation of the input impedance. However, in the analysis of the pulmonary circulation venous pressure cannot be neglected (Chap. 28).

In Appendix 2 the basic hemodynamic elements are discussed. For a resistor, it holds that the sine waves of pressure and flow are in phase, i.e., the phase angle is zero. For compliance the flow is advanced with respect to pressure. This is seen as −90° in the impedance phase angle. For inertance flow is delayed, and shows as +90° for the impedance phase. The modulus of the impedance decreases with frequency, as $1/\omega C$ for compliance, and increases with frequency, as ωL, for the inertance, respectively. In Chap. 12 characteristic impedance is discussed. It is shown that in a large arteries, like the aorta, the mass effects and compliance effects interact in such a way that sinusoidal pressure and flow waves are in phase, and their ratio is constant, this interaction implies that characteristic impedance is a frequency-independent number. This means that the amplitude ratio of pressure and flow is the same for all frequencies and the phase angle is zero. It should be remembered that characteristic impedance is not a simple Ohmic resistor: it is non-existent at zero Hertz and no energy is lost in it. Thus, when modeling characteristic impedance with a resistor this limitation must be kept in mind.

24.1.3 Explanation of Input Impedance

The peripheral resistance, total arterial compliance and characteristic impedance together describe the main features of the input impedance (see Figure in the Box).

24.1.4 Input Impedance and Wave Transmission

From the principles of travelling and reflected waves we can explain input impedance. For a reflection-less system the input impedance equals aortic characteristic impedance. Inversely, the difference between input impedance and characteristic impedance results from reflections. The reflected waves resulting from reflections at bifurcating arteries and other discontinuities return to the proximal aorta. The reflected waves add and result in an impedance that differs from aortic characteristic impedance. For low frequencies, the return times of the reflected waves a small fraction of the period of the wave, all waves add ‘in phase’ and total reflection is large: input impedance differs from aortic characteristic impedance. For high frequencies, the return times of waves are considerable fraction of the period of the waves and they arrive with different phases, and cancel each other out so that the arterial system appears reflection-less: input impedance differs little from aortic characteristic impedance, i.e. constant modulus and phase angle close to zero. Also damping is stronger for the high frequencies and reduces the magnitude of the reflections. It has been suggested that in the human arterial system distinct (relatively strong) reflections may occur at the aortic bifurcation or at the level of the renal arteries. These distinct reflection sites are possibly leading to the secondary rise in systolic pressure and the pressure augmentation (type A beat, Chap. 22). The ratio of reflected and forward running wave amplitudes (reflection magnitude) is related to the magnitude of the oscillations in the impedance modulus [1].

24.1.5 Impedance and Windkessel

The description of the impedance given in the Box suggests three important parameters describing input impedance. These three parameters form the basis of the Windkessel models (Chap. 25). The original two-element Windkessel, proposed by Frank, consists of peripheral resistance, R_p, and total arterial compliance, C. From the information on input impedance, which became available in the 1960s, the idea of aortic characteristic impedance, Z_c, as third Windkessel element appeared [3]. However, we should keep in mind that when the characteristic impedance is modeled with a resistor the mean pressure over mean flow will be $R_p + R_c$, while it should be R_p only. Although this error is not large for the systemic circulation where R_c is about 7% of R_p, it leads to errors when, for instance the three-element Windkessel is used to estimate total arterial compliance. To correct for these shortcomings, a fourth element, the total arterial inertance (see Chap. 25) was introduced [4].

24.1.6 Effective Length of the Arterial System

The frequencies where the minimum in the input impedance modulus and the zero crossing of the phase angle occur have been used to estimate the ‘effective length’ of the arterial system (See Chap. 26). The effective length of the arterial system is used as a conceptual description to determine at what distance from the ascending aorta the major reflection arises. In this concept, it is assumed that the arterial system behaves like a single tube, the aorta, with a single resistance, the peripheral resistance or impedance at its distal end. Based on single tube models the ‘quarter wave length’ principle was introduced. With a distal resistance the frequency of the modulus minimum and the zero crossing of the phase angle are found for a tube

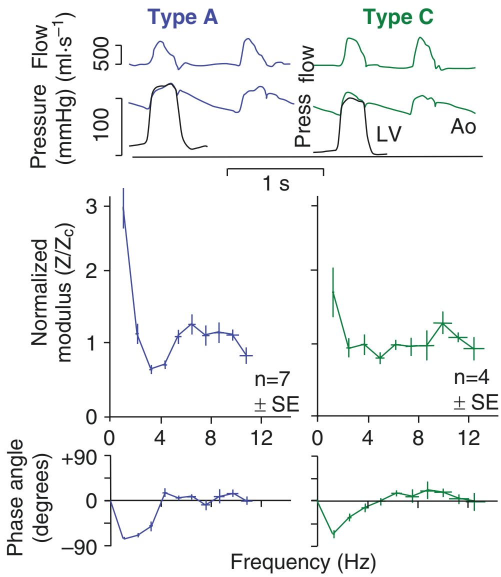


Fig. 24.2 Types of beats relate to input impedance. In subjects with Type A beats a distinct reflection returns in systole and pressure augmentation is high. The impedance oscillates about the characteristic impedance. In Type C beats, reflections are smaller augmentation is negative and the impedance oscillates less than in the Type A beat. (Adapted from Ref. [1], by permission)

length $l = c_{phase}/(4f_{min}) = \lambda/4$. Quantitatively we describe this phenomenon as follows. The wave speed, c_{phase}, equals wavelength, λ, times frequency, f, thus, $c_{phase} = \lambda \cdot f$. When wave speed c_{phase} = 600 cm/s, and the impedance minimum is found at 4 Hz (Fig. 24.1), the wave length $\lambda = c/f$ = 150 cm and the 'effective length' of the tube (aorta) is a quarter wave length, $l = \lambda/4$ = 38 cm. Figures 24.1 and 24.2 show that the zero-crossing of the phase angle differs from the impedance minimum. The assumption of a single tube, mimicking the aorta loaded with the peripheral resistance as model of the systemic arterial tree is too simple and often unrealistic [5, 6]. Tube models and their limitations are discussed in Chap. 26.

24.1.7 External Power

The power produced by the heart equals $(1/T)\int P(t)\cdot Q(t)\cdot dt$, with integration over the heart period T, and is called total external power (see Chap. 16). When mean aortic pressure and Cardiac Output are multiplied the mean power is obtained. The difference between total and mean power is called oscillatory or pulsatile power, which is about 15% of total power. Using Ohm's law, mean power can also be calculated from $CO^2 \cdot R_p$. In the frequency domain, we can perform the following calculations. Oscillatory power must be calculated for each harmonic (i) separately as $P_i \cdot Q_i = Q_i^2 \cdot |Z_{in,I}| \cdot \cos\varphi_i$ and then added as: Power $= \Sigma Q_i^2 \cdot |Z_{in,i}| \cos\varphi_i$, with $|Z_{in,i}|$ the impedance modulus and φ_i the impedance phase angle, and i harmonic number. It has been suggested that the minimum of the modulus of the input impedance should relate to Heart Rate since power was then thought to be minimal. However, we see that not the modulus $|Z_{in,i}|$ but the real part of the input impedance (modulus times cosine of the phase angle) determines power. Often the real part of the impedance has no clear minimum and if it does, it is not found at the same frequency as the minimum of the impedance modulus.

How power is related to the forward and reflected pressure and flow waves was shown in Chap. 22. In the frequency domain mean forward and reflected power equal $Q_f^2 R_p$ and $Q_r^2 R_p$, respectively, with their ratio $Q_f^2/Q_r^2 = \Gamma_q^2 = \Gamma_p^2$. With a reflection coefficient at zero Hz the $\Gamma = (R_p - Z_c)/(R_p + Z_c) \approx 0.87$ (Chap. 12), the ratio of forward and reflected mean power is 0.77. Thus about 0.77/(1 + 0.77) = 44% of the mean power reflects. For the oscillatory components, the ratio equals $\Gamma^2(\omega)$ for each harmonic, and these need to be summed. The total oscillatory power is about 15% of total power.

24.1.8 Impulse Response

Conceptually, it is rather awkward that while pressure and flow are functions of time, the input impedance is expressed as a function of frequency. There exists a characterization of the arterial system in the time domain. This characterization is the so-called impulse response function, which is the pressure that results from an impulse of flow. The impulse should be a short-lasting flow, i.e., short with respect to all travel times and characteristic times of the arterial system, and typically about 1–5 ms in duration. Because the impulse has a height with dimension ml^{-1} and the duration is in seconds, the area under the impulse is ml. The pressure response resulting from this impulse is normalized with respect to the volume of the impulse and the units of the impulse response are therefore mmHg/ml. The calculation of the impulse response function from measured pressure and flow is complicated but straightforward [7]. When the measured flow is broken up in a number of short impulses following each other, the proper addition of the impulse responses leads to the pressure as a function of time.

The input impedance and impulse response function form a 'Fourier pair'. Fourier analysis of the impulse response function leads to the input impedance and inverse transformation of input impedance leads to the impulse response function [7].

If the impulse response is short in duration with respect to the time constant of variation of the time varying system, it may be used to obtain a characterization of that system as a function of time. For example, if the duration of the impulse response is less than 100 milliseconds, and the system under study varies with a typical time of a few hundred milliseconds, the system can be characterized by the impulse response. In this way input impedance of the coronary arterial system was derived in systole and diastole [8].

24.2 Physiological and Clinical Relevance

Although the input impedance gives a comprehensive description of the arterial system its practical use is limited. As discussed above, derivation of input impedance requires Fourier analysis of pressure and flow waves and the data show scatter

(Fig. 24.1), and interpretation requires models. Thus, routine clinical applications are seldom carried out. However, impedances calculated in the human and animals have led to a much better understanding of arterial function. For instance, the input impedance of the arterial system, when normalized to peripheral resistance, is similar in mammals [9]. This explains, in part, why aortic pressures and flows are of similar in shape in mammals (Chap. 32).

Also, derivation of arterial compliance from impedance is of limited accuracy since the modulus is already decreased to near characteristic impedance values for the first or second harmonics (Figure in the Box). Nevertheless, the knowledge of the function the arterial system has greatly helped its function and can serve as quantitative tests of arterial models (Chaps. 25 and 26).

The arterial system can be described in terms of Windkessel models and distributed models. The main arterial parameters describing input impedance are peripheral resistance, total arterial compliance and aortic characteristic impedance (Figure in the Box). Arterial function is often easier to understand when the three Windkessel parameters are given as description of the arterial system. With the modern computing techniques, the calculation of the parameters of the Windkessel model can be performed rapidly and gives directly interpretable results (Chap. 25). For instance, the effect of a change in total arterial compliance on aortic pressure, which is often not obvious from the impedance plot, can be obtained directly by means of the Windkessel, and the impedance calculation can be avoided.

24.2.1 Characteristic Differences in Pressure Wave Shapes

The examples shown in Figs. 24.2 and 24.3, make clear that, although the pressure and flow waves result from the interaction of the heart and arterial load, major features of the pressure wave shape arise from the arterial system and can therefore be related to aspects of the input impedance.

In older subjects, where arterial compliance is decreased the Pulse Wave Velocity and reflections are increased. The larger compound reflected wave adds to the forward pressure wave resulting in an increase in Pulse Pressure, Augmentation Index and systolic pressure, a Type A wave (Chap. 22). As a result of the increased reflection the input impedance oscillates around the characteristic impedance. In subjects with small reflections, the pressure shows an early maximum, with negligible or negative augmentation, and the impedance oscillates only little.

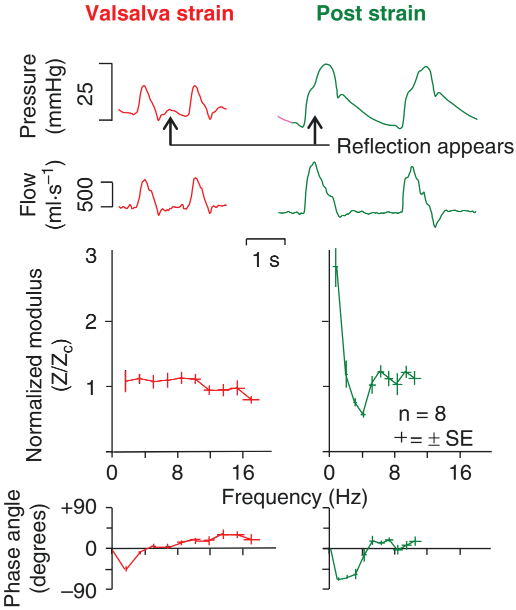


Fig. 24.3 Valsalva strain increases intra-thoracic and intra-abdominal pressures. The thus lower transmural pressure increases arterial compliance and lowers pulse wave velocity. Reflections decrease. An almost reflection-less situation appears where pressure and flow resemble each other (left, top) and input impedance is close to aortic characteristic impedance. In the release phase the reverse is true, reflections are large, pressure and flow wave shapes differ strongly and input impedance oscillates. (Adapted from Ref. [10], by permission)

24.2.2 Changes in Reflection

In the Fig. 24.3 (left side) we see that during the Valsalva maneuver aortic pressure resembles aortic flow in terms of wave shape [10]. During the Valsalva maneuver, intra-thoracic and intra-abdominal pressures increase. The transmural pressure of the aorta decreases and the aortic compliance increases leading to a decreased pulse wave velocity [11]. Waves from different reflection sites arrive more at random in the proximal aorta and reflections smaller. As a result pressure and flow become similar in shape and the input impedance is close to the characteristic impedance of the aorta. After the release of the Valsalva maneuver, cardiac filling and transmural pressure are increased, wave speed is increased as well and reflections increase, and a large augmentation in the pressure is seen (Fig. 24.3 right side). It should be mentioned that during the Valsalva Maneuver arterial stiffness is decreased but peripheral resistance is somewhat increased and not decreased, from 1045 ± 48 to 1342 ± 141 dyne·s·cm^{-5} [10]. From this experiment we have to conclude that reflections at the periphery contribute little to overall reflections, but it is the (many) local reflections at all discontinuities in the major conduit arteries that are of primary importance.

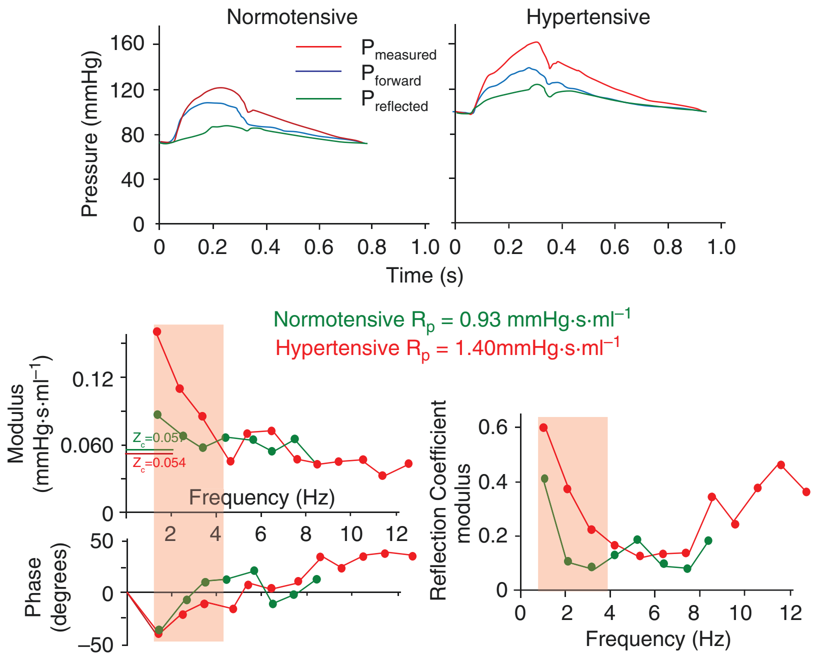


Fig. 24.4 Hemodynamics of hypertension. In the top panel the aortic pressure with its forward and reflected waves of a normotensive and a hypertensive subject are shown. Data in the other panels are averaged, normotensives (n = 14) and hypertensives (n = 12). The mean pressure is increased due to the resistance increase. The reflection magnitude is increased from 0.43 to 0.63. Group-averaged input impedance modulus and phase at low frequencies differ considerably as a result of the decreased in arterial compliance. The reflection coefficient is also increased for the low frequencies. The relative increase in resistance and decrease in compliance are about 50%. Colored areas indicates harmonics 1–3 the major determinants of wave shapes. (Based on data by Ref [12])

24.2.3 Hypertension

Figure 24.4 shows an example of the changes in pressure, input impedance and reflection coefficient. Peripheral resistance is increased by about 50%, and arterial compliance is decreased about 50% [12]. Mean pressure is higher because peripheral resistance is increased. The Pulse Pressure is mainly increased as a result of the decrease in compliance. It may be seen that for the first three harmonics the input impedance modulus is larger in hypertension than normotension, implying a decrease in compliance and higher reflection coefficient.

With increasing age systolic pressure increases and diastolic pressure may even decrease somewhat. Figure 24.5 schematically shows the changes in input impedance with age. Resistance increases in aging but the magnitude of the increase is relatively small. In aging the wave speed increases strongly up to a factor two, and aortic compliance may decrease by a factor 3–4. Characteristic impedance is

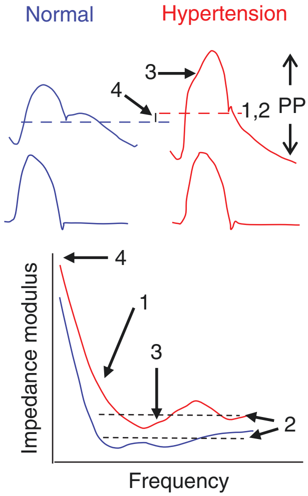


Fig. 24.5 In hypertension peripheral resistance and thus mean pressure is increased are increased with a limited amount (4). Compliance is decreased resulting in a larger pulse pressure, PP, and in a less rapid decrease in impedance modulus with frequency (1), and a higher characteristic impedance (2). Reflections are increased, the impedance oscillates more about the characteristic impedance (3), and the wave is augmented (3). (Adapted from Ref. [13], by permission)

increased due to the aortic stiffening. The decrease in aortic compliance is the main reason why pulse pressure increases.

In essential hypertension both resistance and compliance change by similar amounts, but in old age hypertension the main parameter of the arterial system that changes is compliance.

References

1. Murgo JP, Westerhof N, Giolma JP, Altobelli SA. Aortic input impedance in normal man: relationship to pressure wave forms. Circulation. 1980;62:105–1116.
2. Stergiopulos N, Meister J-J, Westerhof N. Scatter in the input impedance spectrum may result from the elastic nonlinearity of the arterial wall. Am J Phys. 1995;269:H1490–5.
3. Westerhof N, Elzinga G, Sipkema P. An artificial system for pumping hearts. J Appl Physiol. 1971;31:776–81.
4. Stergiopulos N, Westerhof BE, Westerhof N. Total arterial inertance as the fourth element of the windkessel model. Am J Phys. 1999;276:H81–8.
5. Sipkema P, Westerhof N. Effective length of the arterial system. Ann Biomed Eng. 1975;3:296–307.
6. Westerhof BE, van den Wijngaard JP, Murgo JP, Westerhof N. Location of a reflection site is elusive: consequences for the calculation of aortic pulse wave velocity. Hypertension. 2008;52:478–83.
7. Sipkema P, Westerhof N, Randall OS. The arterial system characterized in the time domain. Cardiovasc Res. 1980;14:270–9.
8. Van Huis GA, Sipkema P, Westerhof N. Coronary input impedance during the cardiac cycle as obtained by impulse response method. Am J Phys. 1987;253:H317–24.
9. Westerhof N, Elzinga G. Normalized input impedance and arterial decay time over heart period are independent of animal size. Am J Phys. 1991;261:R126–33.
10. Murgo JP, Westerhof N, Giolma JP, Altobelli SA. Manipulation of ascending aortic pressure and flow reflections with the Valsalva maneuver: relationship to input impedance. Circulation. 1981;63:122–32.

11. Latham RD, Westerhof N, Sipkema P, Rubal BJ, Reuderink P, Murgo JP. Regional wave travel and reflections along the human aorta: a study with six simultaneous micromanometric pressures. Circulation. 1985;72:1257–69.
12. Ting CT, Chen JW, Chang MS, Yin FC. Arterial hemodynamics in human hypertension. Effects of the calcium channel antagonist nifedipine. Hypertension. 1995;25:1326–32.
13. O'Rourke MF. Pulsatile arterial haemodynamics in hypertension. Aust N Z J Med. 1976;6(Suppl 2):40–8.