Increase in pulse wavelength causes the systemic arterial tree to degenerate into a classical windkessel

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Mohiuddin MW, Laine GA, Quick CM. Increase in pulse wavelength causes the systemic arterial tree to degenerate into a classical windkessel. *Am J Physiol Heart Circ Physiol* 293: H1164–H1171, 2007. First published May 4, 2007; doi:10.1152/ajpheart.00133.2007.—Two competing schools of thought ascribe vascular disease states such as isolated systolic hypertension to fundamentally different arterial system properties. The “windkessel school” describes the arterial system as a compliant chamber that distends and stores blood and relates pulse pressure to total peripheral resistance ($R_{\text{tot}}$) and total arterial compliance ($C_{\text{tot}}$). Inherent in this description is the assumption that arterial pulse wavelengths are infinite. The “transmission school,” assuming a finite pulse wavelength, describes the arterial system as a network of vessels that transmits pulses and relates pulse pressure to the magnitude, timing, and sites of pulse-wave reflection. We hypothesized that the systemic arterial system, described by the transmission school, degenerates into a windkessel when pulse wavelengths increase sufficiently. Parameters affecting pulse wavelength (i.e., heart rate, arterial compliances, and radii) were systematically altered in a realistic, large-scale, human arterial system model, and the resulting pressures were compared with those assuming a classical (2-element) windkessel with the same $R_{\text{tot}}$ and $C_{\text{tot}}$. Increasing pulse wavelength as little as 50% (by changing heart rate +33.3%, compliances -55.5%, or radii +50%) caused the distributed arterial system model to degenerate into a classical windkessel ($r^2 = 0.99$). Model results were validated with analysis of representative human aortic pressure and flow waveforms. Because reported changes in arterial properties with age can markedly increase pulse wavelength, results suggest that isolated systolic hypertension is a manifestation of an arterial system that has degenerated into a windkessel, and thus arterial pressure is a function only of aortic flow, $R_{\text{tot}}$, and $C_{\text{tot}}$.

The classical (2-element) windkessel model, quantified by Otto Frank more than a century ago, was the first mathematical model that successfully related global systemic arterial system properties to aortic pulsatile pressure and flow (32). This early model described large arteries as a single compliant compartment distending and storing blood in systole. In diastole, the aortic valve closes and the recoil of the large arteries propels blood continuously to the rest of the system. Although explicitly based on the assumption of conservation of mass, the model had an implicit assumption that pressures in the large arteries rise and fall simultaneously. As a result, all storage of blood is determined by the total arterial compliance ($C_{\text{tot}}$) and all resistance to blood flow is determined by the total peripheral resistance ($R_{\text{tot}}$) (20, 32). The resulting model predicts an exponential fall in pressure in diastole, which roughly approximates measured aortic diastolic pressures (18). The implicit assumption that pressures in the large arteries rise and fall simultaneously is equivalent to the assumption of infinite pulse-wave velocity (20) or, more generally, infinite pulse wavelength. This assumption has been considered to be a flaw in the windkessel model (1).

Spatially distributed arterial system models and importance of pulse-wave reflection. The transmission-line description of the arterial system was developed to address the inherent limitations of the windkessel model. Based on linear approximations to the Navier-Stokes equation, this approach includes the effects of blood inertia, and arterial length arises as an explicit parameter (43). The resulting equation of motion not only predicts a finite pulse wavelength but also that the pressure pulse consists of waves traveling in both forward and backward directions (6, 42). Pulses generated at the heart travel to the periphery and are reflected whenever there is a change in the local impedance, which occurs at bifurcations or in regions where the geometry or stiffness of vessels change (13). On the basis of this description, numerous large-scale distributed arterial system models were developed to capture the realistic topology of the mammalian systemic arterial tree. The model proposed by Westerhof et al. (38, 39) relates aortic input impedance to the radii, lengths, and compliances of 121 arterial segments and has formed the basis for several models with increasing levels of complexity (2, 25, 33). This model yields a realistic root aortic pressure when coupled with a realistic root aortic flow and has helped establish that the arterial system has significant pulse-wave reflection (38, 39).

Isolated systolic hypertension. The example of isolated systolic hypertension (ISH), a condition where systolic pressure is elevated and diastolic pressure remains normal, exposes a conflict between the approaches to determining arterial system properties from measured aortic pressure and flow (i.e., the “inverse problem”). One school of thought has focused on the distributed nature of the arterial system. Assuming a spatially distributed transmission-line model, some studies suggested that changes in the magnitude of pulse-wave reflection from the periphery and the timing of the return of the reflected wave (19, 41) are responsible for ISH. The other school of thought focused on the storage properties of the arterial system. Assuming a lumped windkessel model, these studies suggested that decreased $C_{\text{tot}}$ (5) and peripheral compliance (3) are responsible for ISH. Both schools have identified specific changes in estimated parameters with ISH to support their claims. The conclusions drawn by each school of thought are directly related to the assumptions inherent in each school’s choice of modeling approach.
Reconciling windkessel and transmission descriptions. The classical windkessel has an implicit assumption that pressures in the large arteries rise and fall simultaneously, which is equivalent to the assumption of infinite pulse wavelength (20). On the other hand, the transmission description has no such assumption, although pulse wavelengths are typically longer than the arterial system in mammals (22). Recently, Quick et al. (26) reconciled the windkessel and transmission descriptions of the arterial system. The knowledge that pulse wavelength increases with decreased frequency and compliance (25, 26) has led to two insights. First, individual vessels degenerate into classical windkessels at low frequencies (4, 26). Second, an entire arterial system can degenerate into a classical windkessel when arterial compliance is low (28). These two particular cases support the general assertion made decades earlier that the systemic arterial system becomes more like the classical windkessel as the pulse wavelength increases (20).

Rationale. Although these recent advances have provided much-needed insight, the reconciliation of windkessel and transmission descriptions has yet to be exploited to solve three interrelated problems. First, focusing on the frequency domain, transmission descriptions have yet to be exploited to solve three much-needed insight, the reconciliation of windkessel and radiations affect aortic pulse pressure. Second, focusing on interrelated problems. First, focusing on the frequency domain, that the arterial system degenerates into a windkessel when arterial compliance is low (28). These two particular cases support the general assertion made decades earlier that the systemic arterial system becomes more like the classical windkessel as the pulse wavelength increases (20).

METHODS

Characterizing the arterial system independent of the heart with input impedance. Pulsatile pressure and flow depend not only on the arterial system but also on the properties of the heart. Because the aortic pressure-flow relationship is generally linear (26, 37), the concept of input impedance is typically invoked to characterize the arterial system independent of the heart (23). Whereas pulsatile pressure and flow are typically expressed in the time domain, the pulsatile load to the ventricle formed by the arterial system can be conveniently described by aortic input impedance \( Z_{in} \), the relationship of input pressure \( (P) \) to input flow \( (Q) \) expressed in the frequency \( (\omega) \) domain (20).

\[
Z_{in} = \frac{P(\omega)}{Q(\omega)}
\]  

(1)

Equation 2 completely describes hemodynamics in individual vessels and can be applied to any vessel within a branching arterial network. \( Z_{in} \), then, is the input impedance at the end of the vessel representing the load formed by the rest of the arterial tree.

Relationship of arterial mechanical properties to hemodynamic factors. Ultimately, the key factors affecting pulsatile phenomena (characteristic impedance and pulse wavelength) depend on mechanical properties of arteries as well as the blood. Two basic approaches have been used to derive this relationship. First, Noordergraaf (20, 21) took an ad hoc approach to arrive at equations of motion and continuity. Assuming that the axial pressure gradient in a vessel is equal to a pressure drop due to the inertia of a column of blood and viscous effects described by Poiseuille’s Law, vessel ineratance (\( L' \)) and resistance (\( R' \)) per unit length were characterized. Assuming a vessel with linear elasticity, vessel compliance per unit length \( (C) \), the change in cross-sectional area with change in pressure was characterized. The resulting characterizations of \( R' \), \( L' \), and \( C' \) can alternatively be derived by a more rigorous approach based on linearizing the Navier-Stokes equation (20, 43). Briefly, all nonlinear and second-order terms were neglected, as well as rotational flows. In larger vessels, this method yields equivalent values for \( L' \), \( R' \), and \( C' \) expressed in terms of blood density \( (\rho) \), vessel radius \( (r) \) and blood viscosity \( (\mu) \).

\[
R' = \frac{8\mu}{\pi r^4} \quad L' = \frac{\rho}{\pi r^4} \quad C' = \frac{dA}{dP}
\]  

(3)

where \( A \) is the cross-sectional area, equal to \( \pi r^2 \). As a result, \( Z_{in} \) can be expressed as a function of \( \rho \), \( \mu \), \( r \), and \( C' \) (20). In the larger arteries, viscous effects become negligible, and \( Z_{in} \) simplifies

\[
Z_{in} = \sqrt{\frac{j\omega L' + R'}{j\omega C'}} = \sqrt{\frac{\rho}{\pi r^4} + \frac{8\mu}{\pi r^4 j\omega C'}} \approx \frac{\rho}{\pi r^2 C'}
\]  

(4)

The other parameter of importance in Eq. 2 is the pulse wavelength \( (\lambda) \) which is a function of \( r \), \( C' \), \( \mu \), \( \omega \), and \( \rho \). In larger vessels, the viscous effects become negligible and the normalized wavelength of a pulse pressure at the heart rate (HR) simplifies

\[
\lambda = \frac{j\omega}{L} = HR \cdot L \cdot \sqrt{j\omega C' (j\omega L' + R')}
\]  

\[
= \frac{j\omega}{HR \cdot L \cdot \sqrt{j\omega C' (j\omega L' + R')}} = \frac{8\mu}{\pi r^4 j\omega C'}
\]  

(5)

Equations 4 and 5 are commonly used to describe the characteristic impedance and pulse wavelength in the large arteries of mammals (7, 20, 30).

Windkessel arterial system model. The windkessel model is much simpler than the transmission description described above, because pressure and flow are assumed to vary slowly enough to make inertial effects negligible. As a corollary, pulses are assumed to have an infinitely long wavelength, which leads to simultaneous rise and fall of pressure within the arterial system. Assuming conservation of mass, the resulting windkessel input impedance \( Z_{in} \) is a function of \( C_{tot}, R_{tot}, \) and \( \omega \)

\[
Z_{in} = \frac{R_{tot}}{1 + j\omega R_{tot}C_{tot}}
\]  

(6)

Quick et al. (25–28) illustrated both theoretically and experimentally that the arterial system degenerates into a windkessel at low
Distributed arterial system model. The arterial system model used in the present work consists of 121 major arterial segments representing the large conductance vessels of the human systemic arterial system. The vascular parameters characterizing each vessel segment \( r, C', L \) were first published by Westerhof et al. (38, 39). Using the transmission-line equations described above, vessel inductance, resistance, compliance per unit length were calculated (Eq. 3). These hemodynamic values were then used to characterize the pressure-flow relationships in the 121 vessel segments. The proximal end of this large-scale model starts from the root of the ascending aorta and terminates with resistive elements in the periphery. Because the resistances in the 121 vessel segments were very small, almost all resistance in the system resided in these terminal resistances. Any changes in the radii of these conducting arterial segments therefore have negligible effects on mean pressure. Comparisons to later measurements of systemic arterial compliance (31) led Stergiopulos et al. (34) to suggest that the original model compliances were much too low to represent normal human values. Like in models reported previously (25, 29, 35), we thus increased all of the original arterial segment compliances of the Westerhof model by 50% to represent the normal or "control" case. The input to this large-scale model assumed to be a typical root aortic flow pulse of a healthy human at rest (Fig. 1) originally reported elsewhere (34). It has a period of 1 s, a peak flow of 480 ml/s, and a mean value of 5 l/min. Because the input flow was experimentally derived and included measurement error, there were very small oscillations in flow throughout diastole. The output of the model was input impedance and root aortic pressure (Fig. 1A).

Windkessel as a predictive model. The same flow pulse is used as an input for the classical (two-element) windkessel model (Fig. 1B). \( C_{tot} \) for the windkessel model was taken as the sum of all of the compliances of the distributed model. \( R_{tot} \) for this model was set to match the resistance calculated from the average pressure and flow at the ascending aorta of the distributed model. The output of the model is the windkessel approximation: the aortic pressure and input impedance expected if pulse wavelength is long enough to cause the arterial system to degenerate into a windkessel.

Comparison of human data with the windkessel approximation. Nichols et al. (17) reported a substantial change in aortic pressure waveforms and input impedance with age. For comparison purposes, we digitized the reported pressures and flows for two extremes of age. The input impedance was then calculated with Eq. 1 after pressure and flow were transformed into the frequency domain by using the Fast Fourier Transform (9). \( C_{tot} \) was estimated for both cases by using the technique described in Quick et al. (25). Briefly, the apparent compliance was calculated from input pressure and flow, and the lowest frequency available was taken as the best estimate of \( C_{tot} \). \( R_{tot} \) was estimated by dividing average pressure by average flow. In this case, measured pressure and flow data are used in place of the distributed arterial system model in Fig. 1A. The input impedance of a windkessel with the same \( C_{tot} \) and \( R_{tot} \) was predicted, as well as the pressure waveform resulting from this windkessel given the same measured input flow (Fig. 1B). The predicted windkessel input impedance and the pressure waveform were then graphed with the experimental data (17).

Quantifying the deviation of distributed model and windkessel pressures. To quantify the deviation of the windkessel model from the realistic distributed model, a simple correlation coefficient was calculated. These models were considered similar when \( r^2 = 0.99 \) and difference in pulse pressure was <3 mmHg, signifying the degeneration of the arterial system into a classical windkessel.
RESULTS

Figure 1 illustrates the pressures resulting from inputting a measured human aortic flow into the distributed arterial system model (Fig. 1A) and into the classical windkessel (Fig. 1B). Only if diastolic flow is exactly zero will diastolic pressure predicted from the windkessel fall exponentially. The experimentally measured flow used as an input for both the distributed and windkessel models, however, was not exactly zero throughout diastole. This resulted in slight oscillations appearing in diastolic pressure that deviated from an exponential decrease.

Figure 2, A and B, compares pressures and input impedances of the distributed model with those of a classical (two-element) windkessel model under normal physiological conditions. There was considerable deviation between the two predicted pressures ($r^2 = 0.95$), particularly in systole. Similarly, there was considerable deviation between the two predicted input impedances, particularly at higher frequencies. These results serve as baseline for comparing the three cases described below.

Figure 2, C and D, illustrates pressures and input impedances from both models with a lower heart rate. A 33.3% decrease in heart rate yielded an increase in pulse wavelengths of 50%. Neither pulse pressures ($r^2 = 0.99$) nor input impedances exhibited much deviation. Figure 2, E and F, illustrates the pressures and input impedances from both models with decreased compliance. A 55.5% decrease in compliance yielded an increase in pulse wavelengths of 50%. The pressure pulses were highly correlated ($r^2 = 0.99$). Input impedances were also similar, especially at lower frequencies. Figure 2, G and H, illustrates that the correlation of pressures ($r^2 = 0.99$) and input impedances increased when radii were increased in the realistic distributed model. A 50% increase in radius yielded an increase in pulse wavelengths of 50%. Pressures in
diastole also exhibited little deviation. In each case, the results from the distributed model were more similar to the windkessel model than the baseline cases in Fig. 1.

Figure 3, A and B, illustrates measured pressures and input impedances originally reported by Nichols et al. (17). For the 28-yr-old subject, \( R_{\text{tot}} \) was estimated to be 0.66 \( \text{mmHg} \cdot \text{ml}^{-1} \cdot \text{s}^{-1} \) and \( C_{\text{tot}} \) was estimated to be 2.20 ml/mmHg. The pressures predicted assuming a windkessel with the same values of \( R_{\text{tot}} \) and \( C_{\text{tot}} \) approximated the measured pressure. For the 68-yr-old subject (Fig. 3, C and D), \( R_{\text{tot}} \) was estimated to be 1.08 \( \text{mmHg} \cdot \text{ml}^{-1} \cdot \text{s}^{-1} \) and \( C_{\text{tot}} \) was estimated to be 0.96 ml/mmHg. The pressure predicted assuming a windkessel with the same \( R_{\text{tot}} \) and \( C_{\text{tot}} \) closely approximated the measured pressure. The older subject exhibited less deviation between the measured data and the model results assuming degeneration into a classical windkessel \( (r^2 = 0.93 \text{ vs. } r^2 = 0.99) \).

Figure 4 illustrates how rapidly the human systemic arterial system degenerates into a windkessel with changes in heart rate, vascular compliances, and radii. In each case, the correlation between pressures resulting from the distributed and windkessel models increased as pulse wavelengths increased 50% above normal values. Decreasing heart rate by 33.3%, decreasing arterial compliance by 55.5%, or increasing vessel radii by 50% caused the correlation coefficient to increase to 0.99.

Figure 5 describes the effect of increased pulse wavelength on the degeneration of the arterial system into a windkessel. In each case (altered heart rate, vascular compliances, and vessel radii) the correlation coefficient increased with increases in pulse wavelength. In general, the relationship of pulse wavelengths to the resulting correlation coefficient was insensitive to the particular parameter that was altered.

**DISCUSSION**

This work illustrates with a realistic, large-scale model that the human arterial system degenerates into a windkessel when pulse wavelength increases as little as 50%. Wavelengths become long relative to the lengths of vessels when heart rate is low, vessel compliances are low, or vessel radii are large \( (Eg. \text{ Eq. } 5) \). Although investigators have maintained that long wavelengths make an arterial system degenerate into a windkessel (20), this postulate has never been tested, in part because of the inherent inability to independently alter heart rate, compliances, and radii experimentally. This is the first time that increases in wavelength have been shown to result in increased “windkesselness,” both in terms of aortic pressure and input impedance (Figs. 2, 4, and 5). Furthermore, the present work quantifies how much and how quickly an arterial system degenerates into a windkessel.

**Insight from case studies.** This study suggests that an arterial system can degenerate into a windkessel when pulse wavelengths increase as little as 50% \( (\text{Fig. } 5) \). With insight into the factors that increase pulse wavelength, three physiologically relevant cases are explored.

**Case 1: bradycardia.** Both cardiac and noncardiac abnormalities can lower heart rate to very low levels. With bradycardia, heart rate drops to a range of 30–50 bpm. If mean arterial pressure is maintained, pressure-dependent vascular wall properties would remain normal. In this case, the model predicts that pulse wavelengths would increase by a factor of 1.5, the diastolic pressure would fall exponentially \( (\text{Fig. } 2C) \), and the input impedance would fall monotonically \( (\text{Fig. } 2D) \). Indeed, the error between the distributed system and windkessel approximation becomes negligible as heart rate falls below 30 bpm \( (\text{Fig. } 4A) \). These model results are consistent with the exponential decrease in pressure in diastole reported in the literature \( (16) \), and therefore the hemodynamic changes reported with bradycardia can be attributed to increased wavelength and therefore increased windkesselness.
Case II: exercise-induced hypertension. When healthy individuals exercise, the heart rate increases to provide the necessary increase in cardiac output to meet the higher metabolic demand. In response to the attending increase in mean arterial pressure, there are minor increases in radii and rather substantial decreases in compliances of the conductance arteries (15). Interestingly, compliances and heart rate can change significantly, but they balance and result in relatively small changes in pulse wavelength, as suggested in Eq. 5. In this case, the system does not degenerate into a windkessel; diastolic pressures do not fall exponentially, nor does input impedance fall monotonically.

Case III: age-related changes in arterial structure. As humans age, there is a tendency for pulse pressure to increase, even when both heart rate and mean pressures remain normal. In this case, the aging process causes changes to the arterial wall composition (18), leading to both significantly decreased compliances and increased radii of the large conductance vessels. This causes pulse wavelength to increase (Eq. 5). In this case, diastolic pressure falls exponentially and input impedance falls monotonically, as expected for a windkessel.

Identifying the causes of ISH. These case studies provide an essential insight into the interaction of heart rate, arterial compliances, and radii in normal individuals: changes in parameters in normal conditions tend to balance and keep wavelength constant. This suggests a homeostatic principle that acts to moderate pulse pressure. It is interesting to note that if this homeostatic principle is violated and pulse wavelength increases, the arterial system can begin to behave like a windkessel. In this case, \( C_{\text{tot}} \) and \( R_{\text{tot}} \) are the only arterial system mechanical properties that effect pressure and input impedance. Although increasing the radius of the vessels in the Westerhof model can cause it to degenerate into a windkessel (Fig. 2, G and H), the required increase (\( \sim 50\% \)) is not physiological. However, the reported changes in arterial compliance with ISH ranges from \(-57\%\) to \(-75\%\) (5, 12, 14, 44). Changing compliance in our model by the same amounts resulted in \( r^2 = 0.995-0.998 \), indicating that this change in arterial compliance is enough to cause an arterial system to degenerate into a classical windkessel. Similarly, estimated \( C_{\text{tot}} \) of the older subject was 44% less than that of the younger subject (Fig. 3). From model results (Fig. 2, E and F) and analysis of human data (Fig. 3), it can be surmised that ISH is a manifestation of the degeneration of the systemic arterial system into a classical (2-element) windkessel. Although changes in pulse pressure have been observed in patients with increased pulse-wave velocity and the magnitude of the reflected wave (19, 41), the present...
study suggests that pulse pressure in ISH is directly related to only two global mechanical properties of the arterial system: \( C_{\text{tot}} \) and \( R_{\text{tot}} \).

Calculating reflection when a system behaves like a windkessel. It is well established that pulse-wave propagation and reflection effects in a windkessel are poorly defined (20). Because pulse wavelengths are long, there is very little delay between the arrivals of pressure pulses from different locations throughout the system. In this case, pulses are reflected and re-reflected many times in a single cardiac cycle (6), making it impossible to identify a discrete set of reflection sites (30). In fact, the magnitude and phase of reflection becomes a function of \( C_{\text{tot}}, R_{\text{tot}}, \) and \( \text{HR} \) (26). Although the pressure pulses can be analyzed into forward and reflected waves, the composite reflection at the aortic root cannot be attributed to any particular arterial system property such as arterial system length, changes in reflection site, or magnitude of local reflection (28, 30). The fact that arterial systems degenerate into windkessels explains why it becomes much easier to estimate \( C_{\text{tot}} \) in hypertension (25): \( C_{\text{tot}} \) is the dominant determinant of pulsatile pressure.

Windkessel can be used as a predictive model. The use of the windkessel in the present work represents a major departure from convention. Typically, a version of the windkessel model is used to solve the inverse problem. That is, parameters such as \( C_{\text{tot}} \) are unknown, and its value is carefully chosen to best fit the measured data. Here, the process is reversed. Instead of retrieving information from fitting the parameters of the three-element windkessel (40), four-element windkessel (36) or five-element windkessel (10) to data, information is retrieved by analyzing the deviation of the classical (2-element) windkessel from measured data. In this study, \( C_{\text{tot}} \) is first determined from low-frequency apparent compliance (25) and then taken to be the value of compliance in the windkessel model. Determining \( C_{\text{tot}} \) in this way does not (forcibly) fit the model to measured pressure and flow. Instead, windkessel pressure is predicted from input flow, \( C_{\text{tot}} \), and \( R_{\text{tot}} \) (Fig. 1B). Therefore, there is a much greater deviation of the model from the data at intermediate and high frequencies. Thus, in the present study, the windkessel is used as a predictive model, and deviations are used as a means of quantifying the similarity of an arterial system to a windkessel.

Critical role of mathematical modeling and model-based analysis of empirical data. The present work developed two complimentary approaches to test the hypothesis that the systemic arterial system degenerates into a classical windkessel when pulse wavelengths increase sufficiently. The first approach used a novel analytical technique to quantify the windkesseness of an arterial system (Fig. 1). The second approach used a realistic, large-scale human systemic arterial system model to quantify how changes in each parameter affect the degree of windkesseness (Fig. 4). Together, these two approaches led us to the conclusion that it is the pulse wavelength (independent of particular heart rate, compliances, or radii) that determines whether an arterial system behaves like a windkessel (Fig. 5). These results could not be obtained by experimental means because it is not possible to independently change heart rate, compliances, or radii of the large conductance arteries in an experimental model. For instance, unless peripheral resistance is significantly manipulated, altering heart rate affects mean pressure and thus compliances and radii of the large conductance arteries (12, 35). Similarly, attempts to decrease aortic compliance by encausting the aorta in plaster (24) or banding it (11) do not mimic naturally occurring diseases and can only be accomplished surgically. There is no safe method of altering aortic compliance or radius in humans. Our novel index of windkesseness provides a simple, minimally invasive means to characterize hemodynamic changes in an individual over time or relative differences in experimental groups. Requiring only measured pulsatile pressure and flow (e.g., Fig. 3), this index provides a much-needed tool to explore how the mechanical properties of the arterial system change with naturally occurring diseases and is amenable for use in a clinical setting. Mathematical modeling and model-based analysis of empirical data have provided us the missing tools to reveal that ISH is a manifestation of an arterial system that has degenerated into a windkessel, and thus arterial pressure in ISH is a function only of aortic flow, \( R_{\text{tot}} \), and \( C_{\text{tot}} \).

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