A New Model-Based Method of Reconstructing Central Aortic Pressure from Peripheral Arterial Pressure

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Abstract: We have shown in our previous study that the transfer function between central aortic pressure and tonometric radial arterial pressure can be modeled as a pure elastic lossless tube terminated with a modified Windkessel. We hypothesized, using the model-derived radial arterial flow, that central pressure could be reconstructed by adding the time-shifted forward and backward pressure components (Stergiopoulos *et al.*: Am J Physiol 274: H1386–H1392, 1998). In eight patients (age 16–75), central micromanometric and radial arterial tonometric pressure were measured simultaneously. We imposed measured tonometric pressure to the terminal modified Windkessel to estimate radial arterial flow, with which tonometric pressure was separated into forward and backward components. These components were then appropri-

ately time shifted, and summed to central pressure. We used average parameter values for the terminal impedance, but individualized the transmission delay. The poor correlation (r^2) between tonometric and central pressure (0.264–0.765) was improved by both central pressure reconstruction methods (generalized transfer function: 0.887–0.974, model-based method: 0.849–0.979). The sensitivity analysis indicated that the key model parameter in reconstructing central pressure was the transmission delay. We conclude that our model-based method was capable of reconstructing central pressure as precisely as the generalized transfer function method, and also capable of individualizing the transfer function by changing the transmission delay. [Japanese Journal of Physiology, 51, 217–222, 2001]

Key words: tonometry, wave transmission, wave reflection, sensitivity analysis.

Although blood pressure is one of the oldest and the most widespread ways to examine the vital status of animals, the precise waveform was available only by invasive methods until the tonometric device was developed. Obviously sphygmomanometry is not suitable for following the rapid changes in pressure, e.g., in patients with arrhythmia, to detect information available only from waveform and even to calculate exact mean pressure. Although the waveform recorded by radial artery tonometry can relatively faithfully reproduce intra-arterial pressure [1], the

waveform is distorted and delayed compared with central aortic pressure, which is required for a certain type of hemodynamic measurement. Several groups have tried to correct this distortion by a generalized (average) transfer function with some success [2–5]. Stergiopoulos *et al.* [6] approached this with another method. They showed that by separating the peripheral arterial pressure into forward and backward components, central pressure was reconstructed by adding the appropriately time-shifted forward and backward pressure waves. Although the latter method is superior

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Abbreviations: Z_c , characteristic impedance; T_d , wave transmission delay; C , peripheral capacitance; R , peripheral resistance; $P_{f(AO)}$, forward aortic pressure wave; $P_{b(AO)}$, backward aortic pressure wave; $P_{f(RA)}$, forward radial arterial pressure wave; $P_{b(RA)}$, backward radial arterial pressure wave; $P_{(AO)}$, aortic pressure; $P_{(RA)}$, radial arterial pressure; $Q_{(RA)}$, radial arterial flow; Δ , delay operator; Γ , reflection coefficient; ω , angular frequency; j , imaginary unit; Z , terminal impedance; P_d , pressure at the point distal to Z_c ; r^2 , coefficient of determination; TF, generalized transfer function method.

because we need not assume that the transfer function is uniform among subjects, it does need the measurement of radial arterial flow.

We have shown in our previous study [7] that the transfer function between central aortic pressure and tonometric radial arterial pressure can be approximated with a model made of a pure elastic lossless tube terminated with a modified Windkessel. With the known model, because it is possible to estimate the radial arterial flow from the radial arterial pressure, we hypothesized that central aortic pressure could be reconstructed by the method of Stergiopoulos *et al.* [6] without the direct measurement of radial arterial flow. Furthermore, the approximation of the transfer function with a model might provide the opportunity to individualize the transfer function by changing parameters in the model. For this purpose, we also studied the sensitivity to parameter changes of the model in reproducing central pressure.

MATERIALS AND METHODS

Model. The vascular system was modeled with a pure elastic lossless tube terminated with a modified Windkessel vascular model (Fig. 1) [7]. The elastic tube corresponds to the arterial tube between the aorta and the radial artery, and the modified Windkessel represents the vascular system distal to the radial artery. As shown in Fig. 1, the model is characterized by four parameters: characteristic impedance of the tube (Z_c), wave transmission delay along the tube (T_d),

Fig. 1. Schema illustrating the model of the vascular system used to approximate the transfer function between central aortic and radial arterial pressure. The model is characterized by 4 parameters: characteristic impedance of the pure elastic lossless tube (Z_c), wave transmission delay along the tube (T_d), capacitance (C), and resistance (R) of the terminating impedance (Z). The forward pressure wave ($P_{f(Ao)}$) travels along the tube with pure delay (Δ) until it reaches the end of the tube ($P_{f(RA)}$) and reflects according to the reflection coefficient (Γ), which arises from the mismatch between Z_c and Z . The backward wave ($P_{b(RA)}$) travels back with delay Δ ($P_{b(Ao)}$). The sum of these forward and backward waves form the actual pressure waves ($P_{(Ao)}$ and $P_{(RA)}$). P_d , pressure at the point distal to Z_c of the terminal impedance.

peripheral capacitance (C), and peripheral resistance (R). Of these, only 3 parameters (Z_c/R , $C \cdot R$, and T_d) are observable by analysis of the transfer function between pressure waveforms.

The model transfer function can be obtained as follows. The incident (forward) pressure wave ($P_{f(Ao)}$) travels along the tube with delay (Δ), but with no deformation until it reaches the end of the tube ($P_{f(RA)}$). At that point, the pressure wave reflects backward according to the reflection coefficient (Γ), which arises from the mismatch between the characteristic impedance of the tube and the terminal impedance. The reflected (backward) wave ($P_{b(RA)}$) now travels with the same delay (Δ) with no deformation ($P_{b(Ao)}$), but toward the aortic root. The sum of these forward and backward pressure waves form the actual pressure waves at the aorta and at the radial artery, respectively. From the above, the transfer function can be obtained by

$$H(\omega) = \frac{1 + \Delta^2 \Gamma}{\Delta + \Delta \Gamma}$$

where Δ is the delay operator and equals $\exp(-\omega \cdot T_d \cdot j)$, Γ is the reflection coefficient spectrum and equals $(Z - Z_c)/(Z + Z_c)$, ω the angular frequency, j the imaginary unit, and Z the terminating impedance. We have already shown that the high-resolution transfer function (see below) can be approximated with the model transfer function well up to 8 Hz [7]. The average model parameters were $C \cdot R$: 1.33 ± 0.55 s, Z_c/R : 0.0318 ± 0.0300 , and T_d : 31.9 ± 4.1 ms [7].

Subjects. The study was performed in eight patients who underwent cardiac catheterization for diagnostic purposes. All patients gave written informed consent for both the diagnostic and investigational parts of the catheterization before the study. There were 2 males and 6 females, and the ages ranged from 16 to 75 (mean $\pm$ SD, 53 ± 19). Five were suffering from rheumatic valvular heart disease, two were patients with cardiac sarcoidosis, and the remaining patient had Ebstein's anomaly. All patients were arrhythmic. In all patients with valvular heart disease, cardiac rhythm was atrial fibrillation. In patients with sarcoidosis, one had complete atrioventricular block, and the other exhibited frequent atrial and ventricular premature beats. The patient with Ebstein's anomaly had frequent ventricular premature beats. We intentionally selected patients with arrhythmia for technical reasons. Without the irregular cardiac rhythm, we could not obtain wide-band high-resolution transfer function between pressures [8, 9]. An irregularity of pressure wave form was sufficient to determine transfer function precisely up to 7 Hz in these patients as evidenced

by high coherence values (figure 3 in Sugimachi *et al.* [7]). Four of the 5 patients with valvular heart disease were taking digitalis to control ventricular response rate, which was well controlled (60 to 80 beats/min) in all patients with atrial fibrillation during catheterization.

After diagnosis was made by catheterization, pressure was measured with a 3-French micromanometric (Model 420, Camino, San Diego, CA) catheter, the tip of which was advanced to the aortic root by way of a 7-French thin-walled or 8-French pigtail catheter inserted via the femoral artery. Simultaneously, blood pressure was also measured with a tonometric pressure recorder (Model CBM7000 Nihon-Colin, Komaki, Japan). This device records radial arterial pressure by compressing the radial artery between the radius and the pressure sensor. The compressing force is adjusted not to obstruct the artery, but to make steady contact with the artery on a flat surface of the sensor so that the pressing force reflects the intravascular pressure. These pressure waves and electrocardiograms were recorded on an 8-channel digital-audio-tape (DAT) data recorder (Model RD-111T, TEAC, Tokyo, Japan) for off-line analysis. The frequency response of the recorder was far beyond the concerned frequency range. We recorded signals for at least 2 min while arrhythmia naturally perturbed the cardiac cycle length and pressure.

Analysis.

Central pressure reconstruction based on the model. In reference to Fig. 1, radial arterial flow was obtained by dividing the pressure gradient across the Z_c of the terminal impedance by the Z_c value. Although the absolute value of Z_c is unavailable and only Z_c/R is, the following derivation is not affected by the absolute parameter values. Therefore we expressed the flow in arbitrary units by setting R as unity. The pressure (P_d in Fig. 1) at the point distal to Z_c was calculated by integrating the flow entering into and exiting from C .

From these pressure and flow waves, we separated the forward and backward pressure waves according to Westerhof *et al.* [10] as

$$P_{f(RA)} = \frac{P_{(RA)} + Z_c \cdot Q_{(RA)}}{2}$$

$$P_{b(RA)} = \frac{P_{(RA)} - Z_c \cdot Q_{(RA)}}{2}$$

where $P_{(RA)}$ and $Q_{(RA)}$ are the radial arterial pressure and flow waves, respectively. These pressure waves were then time shifted by T_d , backward and forward, respectively, to calculate $P_{f(Ao)}$ and $P_{b(Ao)}$. $P_{(Ao)}$ was es-

timated by adding $P_{f(Ao)}$ and $P_{b(Ao)}$.

For $C \cdot R$ and Z_c/R values, we used the average values (see the Model section) obtained in our earlier study [7]. T_d values were estimated for each subject by the time difference between the onset of upstroke of radial pressure and that of aortic pressure (foot-to-foot method). We selected to use the average $C \cdot R$ and Z_c/R , but the individually estimated T_d based on the sensitivity analysis (see below).

Central pressure reconstruction by the average transfer function. The two pressure wave forms, central aortic pressure recorded by the micromanometer and radial arterial pressure recorded by tonometry, were replayed by DAT recorder. These were sampled by a personal computer (PC-9801FA7, NEC, Tokyo, Japan) equipped with a 12-bit analog-to-digital converter (AD12-16D(98), Contec, Osaka, Japan) at a sampling frequency of 200 Hz. We then divided the data into two portions. The data for about 20 s at the last portion were saved and used exclusively for testing the performance of the central pressure reconstruction. The rest of the data was used to determine the transfer function to reconstruct the central pressure.

The way to determine high-resolution transfer function from pressure data under irregular beats has been described in detail [8, 9, 11]. In brief, we divided data into 10.24-s segments, removed the linear trend, and applied the Blackman-Harris window (4 term, -92 dB) for each segment. By using Fourier transform, we calculated power and crosspower and ensemble averaged them across the segments. Transfer function was calculated by dividing ensemble-averaged crosspower by ensemble-averaged input (tonometric pressure) power.

We averaged the transfer function over 8 subjects to reconstruct central aortic pressure. The modulus and the phase of the transfer function were averaged separately at each frequency. We convolved the tonometric pressure with the impulse response (inverse Fourier transformed transfer function) corresponding to the average transfer function. Central pressure reconstruction was performed for data spared exclusively for this test, which repeated for 10.24-s data segment until no available data could be found (2 to 4 segments).

Sensitivity analysis. We studied the sensitivity to parameter changes of the capability of the model to reproduce central pressure, to determine the key parameter to individualize the transfer function. We increased or decreased one of the parameters ($C \cdot R$, Z_c/R , or T_d) in a stepwise manner and repeated this to reconstruct the central aortic pressure. Starting with the average parameter values for $C \cdot R$, Z_c/R , and T_d

Fig. 2. Left: Representative tracings of measured central aortic pressure (top row), measured tonometric radial arterial pressure (second row, none [i.e., without reconstruction]), central pressure reconstructed with the generalized transfer function (third row, TF), and with the model-based method (bottom row, model). Right: Scatter plots between measured and reconstructed central aortic pressure for each of the reconstruction methods (from top, none, TF, model).

(reference), we altered each parameter, one by one, from a fourth to four times the reference value in 40 steps forming a geometric series. The precision of the central pressure reconstruction was expressed by the coefficient of determination (r^2) between measured and reconstructed central aortic pressure.

RESULTS

Shown in Fig. 2 are representative tracings of measured central aortic pressure (top row), central pressure reconstructed with the average transfer function (third row), and our model-based method (bottom row). Both methods succeeded in reconstructing central pressure waves faithfully. The scatter plot (Fig. 2, right panels) between measured and reconstructed pressure further supported this.

In Fig. 3 the coefficient of determination (r^2) between measured and reconstructed pressure is shown for all subjects. With no correction, the tonometric and the central aortic pressure correlated poorly, as quantified with low r^2 values (left column, none). The r^2 values ranged from 0.264 to 0.765 (median: 0.631). With reconstruction by the average transfer function (middle column, TF), r^2 increased dramatically and ranged from 0.887 to 0.974 (median: 0.943). Almost the same r^2 values were obtained in the model-based reconstruction method (right column, model). The r^2 values were from 0.849 to 0.979 (median: 0.949).

Figure 4 illustrates the comparison between measured (thin lines) and reconstructed central pressure waves (thick lines) in a representative case when the parameter values (from top $C \cdot R$, Z_c/R , and T_d) are set

Fig. 3. Summary of coefficient of determination (r^2) between measured and reconstructed central aortic pressure for each of the reconstruction methods (none, without reconstruction; TF, generalized transfer function method; model, model-based method).

at four times the average value obtained from the measured transfer function. Even with $C \cdot R$ and Z_c/R four times the average value, the model was still able to reconstruct central pressure reasonably well, though the scatter plot revealed slightly open hysteresis loops. This indicates that the sensitivity to $C \cdot R$ or Z_c/R to reproduce central pressure is low. In contrast, when T_d was increased four times, the reconstructed pressure differed significantly from the measured central pressure, indicating the high sensitivity to T_d to reproduce central pressure.

Figure 5 tabulates the changes in coefficient of determination by changing each model parameter (from left $C \cdot R$, Z_c/R , and T_d) one by one. The steeper slope

Fig. 4. Left: Comparison between measured (thin lines) and reconstructed central aortic pressure waves (thick lines) by the model-based method in a representative case when the parameter values (from top, $C \cdot R$, Z_c/R , T_d) are set at four times the average value obtained from the measured transfer function. Right: Scatter plots between measured and reconstructed central aortic pressure when each model parameter was set at four times (from top, $C \cdot R$, Z_c/R , T_d).

Fig. 5. Changes in coefficient of determination (r^2) between measured and reconstructed central aortic pressure by the changes in each of the model parameter values (from left, $C \cdot R$, Z_c/R , T_d). Unity value for the parameter indicates the average value obtained from the measured transfer function.

of this plot indicates that the model-based reconstruction of the central pressure is more sensitive to that parameter. It is clearly shown that the precise reconstruction is more sensitive to T_d than to $C \cdot R$ and Z_c/R .

DISCUSSION

Advantages of our method. We have developed a new method to reconstruct central aortic pressure from tonometric pressure measured at the radial artery. This method was capable of reproducing central pressure as faithfully as with the generalized transfer function method.

Our method has several advantages. First, although it is based on the method developed by Stergiopoulos *et*

al. [6], our method needs only radial arterial pressure and not radial arterial flow. As a result, we need the radial arterial pressure and the timing of the central pressure upstroke (for the determination of T_d) to reconstruct central pressure. Radial arterial flow was calculated from radial arterial pressure based on the vascular model. Second, the way to reconstruct central pressure is simple and straightforward. Unlike the generalized transfer function method [2–5], our method needs only the calculation of flow and pressure wave components in the time domain and does not need complicated calculation in the frequency domain involving Fourier transform, transfer function, and convolution integral. Because of this simplicity it is quite easily possible to implement this method in monitoring central pressure on-line. Although the time delay by the wave transmission is physically inevitable, reconstruction itself can be carried out almost in real time. Third, the approximation of the vascular system with a model provides the opportunity to individualize the transfer function between subjects and between clinical conditions. We will discuss this matter further in the next section. In summary, this method is especially useful in serial noninvasive monitoring of central pressure wave form. A combination of this method and the single-beat estimation method may enable noninvasive estimation of end-systolic elastance.

Sensitivity analysis. The model used in this study is only an approximation of the complicated vascular system with special emphasis on the pressure transmission between the aorta and the radial artery. However, because the model can reconstruct the central pressure precisely, the detailed form of the transfer function (especially in the high-frequency range) is not important in reproducing central pressure. We further demonstrated by the sensitivity analysis that the precision of the reconstruction was relatively preserved in the presence of rather drastic changes in $C \cdot R$ and Z_c/R values. Because the ranges we examined covered those seen in pathological and physiological conditions, it might be supposed that the changes in these parameters are not necessary to individualize the transfer function. In contrast, the large dependence of coefficient of determination on T_d indicated the need to change T_d according to subject and conditions. This is especially true when T_d was higher than the true value. In other words, the precision of the central pressure reconstruction deteriorates if T_d is overestimated. Thus T_d seems to be the key parameter to individualize the transfer function. Fortunately, because T_d can be estimated from carotid and radial artery pulse recordings, the method can individualize

the transfer function noninvasively. T_d can also be estimated with the use of second heart sound and diastolic notch of radial artery pulse recording. Although we used the average transmission delay in this study, it is possible to take pressure-dependence of delay into account because it can be estimated on a beat basis.

Similar study. Karamanoglu and Feneley [12] also studied the precision of the model-based transfer function in reconstructing central pressure, but from finger pressure. They have also shown that the model-based method was comparable in precision and that it was even better if the hemodynamics were altered. Although we did not study the precision in altered hemodynamics, the findings by Karamanoglu and Feneley [12] are basically consistent with our findings. In their study, however, model parameters could be obtained only invasively. The strategy to obtain model parameters noninvasively was not provided. We have shown that the key model parameter is obtainable noninvasively and that the average values of other parameters can give good precision.

Limitation. We hypothesized that the model-based method may be superior to the generalized transfer function method in reconstructing central pressure because the transfer function can be individualized. The results, however, indicated that the precision of reconstructed pressure was comparable. Individualization of the transfer function improves the precision in central pressure reconstruction, and approximation of the transfer function by the model would worsen it. We conjectured that the relative uniformity of the subjects in this study limited the benefit of this method. Whether better precision can be obtained by the model-based method in altered hemodynamics needs further study.

Our study also indicated that the accurate determination of T_d is vital in precisely reconstructing central aortic pressure. Future development of instruments to reconstruct central pressure needs accurate methods to determine T_d from noninvasive carotid and radial artery pulse recordings.

Central pressure reconstruction may be aggravated if $C \cdot R$ and Z_c/R values change beyond the range we examined in this study. We must determine the conditions under which these values can potentially exceed the range.

Conclusion. We conclude that our model-based method was capable of reconstructing central aortic pressure from radial arterial tonometric pressure as precisely as the generalized transfer function method. Furthermore, the model-based method provides an opportunity to individualize transfer function by changing the wave transmission delay parameter.

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