Two Arterial Effective Reflecting Sites May Appear as One to the Heart

Roberto Burattini, Grant G. Knowlen, and Kenneth B. Campbell

The relation between reflected waves and features of ascending aortic pressure waveforms and impedance patterns was investigated with a modified T-tube model of the systemic arterial circulation. Ascending aortic pressure and flow and descending aortic flow were measured in 10 dogs under basal conditions and under the effect of an agent (methoxamine) that caused vasoconstriction and an increase of mean aortic pressure. A broad range of aortic pressure amplitudes and features was obtained. These waveshapes were classified into four groups. Under basal conditions, cases for which a prominent diastolic fluctuation was present \((n=8)\) were grouped in A. Cases for which this fluctuation was absent \((n=2)\) were grouped in B. Groups C \((n=4)\) and D \((n=3)\) included cases that, under vasoconstricted conditions, did or did not display, respectively, a diastolic fluctuation in pressure. Arterial T-tube model parameters were estimated by simultaneously fitting the model to both ascending and descending aortic flows with aortic pressure as input. A good fit was obtained in any case considered. After parameter estimation, forward and reflected waves and impedance patterns at the entrance of head circulation (head and upper limbs) and body circulation (trunk and lower limbs) as well as their merger in the ascending aorta were determined. T-tube input impedance compared well with impedance data points obtained from the ratio of corresponding harmonics of ascending aortic pressure and flow. In some cases (group A), modulus and phase spectra displayed two distinct minima, in the range from 0 to 10 Hz. In some other circumstances, these minima were less distinct (groups B and C) and could even appear as one (group D). Whether one or two minima appeared in the ascending aortic impedance spectra at low frequency and whether a prominent diastolic fluctuation did or did not appear in aortic pressure, pressure and flow waveshapes proximal to the heart were explained by the presence of two effective reflecting sites in the systemic circulation. In group B, a diastolic fluctuation in pressure was absent despite the fact that head-end and body-end reflected waves were distinct. This happened because body-end reflected waves peaked corresponding to a minimum of the head-end reflected wave. In group D, a diastolic fluctuation in aortic pressure was absent because the body-end reflected wave moved into systole and superimposed on the head-end reflected wave. This superimposition was due to increased pulse wave velocity in the body transmission path as a result of decreased arterial distensibility. After comparing our T-tube model with a single-tube model representation of systemic circulation, we concluded that, in cases like those grouped in D, two effective reflecting sites appear as one to the heart. (Circulation Research 1991;68:85-99)

Changes in vascular impedance result in alterations in contour of the pressure wave in the ascending aorta. The majority of alteration is due to reflected waves, and it is generally agreed that such waves occur.\(^1\)\(^-\)\(^6\) Reflected waves are generated wherever there is a change in the vascular impedance of the path through which the blood flows or whenever the dimensions or elasticity of the bed assumes new values. Because the topology as well as the architecture of the arterial tree plays a key role in the way the reflective properties come about, multi-branched models were used to interpret these properties.\(^7\)\(^-\)\(^12\) Nevertheless, there is still some dispute as to whether most reflection arises and how much reflection of the pulse wave (pressure and flow) occurs.

Quality and quantity of reflection depend on the observation point. An important point of observation is the aortic root for the effect that reflections may have on left ventricular function. An essential aspect

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of arterial wave reflection, as seen from the aortic root, is whether reflected waves appear to arise from one or two functionally discrete reflecting sites. This issue is still unanswered.\textsuperscript{3,5,6,13} The presence of two arterial effective reflecting sites was first suggested and conceptually explained in terms of an asymmetrical T-tube model by McDonald in 1960 (see Reference 14). The short limb represents the circulation of head, neck, and upper limbs (head circulation), and the long limb represents the circulation of trunk and lower limbs (body circulation). This concept was taken up by O’Rourke and Taylor,\textsuperscript{15,16} O’Rourke,\textsuperscript{17-19} Avolio et al,\textsuperscript{20} and Nichols et al\textsuperscript{21} to explain arterial impedance patterns and pressure and flow wave shapes for a variety of animals and humans. In contrast to this theory, other studies of humans\textsuperscript{22-24} and dogs,\textsuperscript{25-28} based on the assumption that the arterial system behaves like a single-tube model, support the idea that one effective reflecting site may account for reflection phenomena. The differences in interpretation as to whether two or one effective reflecting sites are needed arise from two main facts. First, two rather than one minima may be observed in the range from 0 to 10 Hz of ascending aortic impedance spectra. Second, a prominent diastolic fluctuation may or may not appear in the ascending aortic pressure waveform. In the absence of this diastolic fluctuation, a single-tube model was able to describe experimental impedance and pressure–flow relations in the ascending aorta and to account for forward and reflected waves in this region.\textsuperscript{28} However, when this fluctuation was present, the single-tube model failed. In this circumstance, a modified version of the asymmetrical T-tube model, which possessed two reflecting sites of complex nature, was suitable.\textsuperscript{13,29,30}

A purpose of the present work was to address the issue of whether two instead of one effective reflecting sites account for 1) features of ascending aortic pressure waveform, even though a diastolic fluctuation is absent, and 2) impedance patterns, even though one minimum, instead of two minima, appears at low frequency. Another purpose was to identify possible circumstances for which two effective reflecting sites appear as one to the heart.

**Methods**

To obtain records of ascending aortic pressure in which diastolic fluctuations were and were not present, measurements were taken in dogs under basal and vasoconstricted conditions. To analyze the relation of wave reflection with both features of aortic pressure and impedance spectra, we used our modified T-tube model.\textsuperscript{29,30} Measurements of ascending aortic pressure and flow and descending aortic flow were used to estimate model parameters. The model with these estimated parameters was then used to determine impedance patterns and forward and reflected waves in the arteries proximal to the heart. These impedance patterns and reflected waves were related to the prominent features of aortic pressure.

To test whether one, instead of two, arterial effective reflecting sites can account for these features, a single-tube model\textsuperscript{27,28} was considered as a model competing with our T-tube model. When these two models were equivalent, that is, when they gave comparable estimates of global parameters and similar fits of experimental data, we concluded that two effective reflecting sites appeared as one.

**Experimental Preparation**

Ascending aortic pressure and flow and descending aortic flow were measured in 10 anesthetized, positive-pressure ventilated, open-chest dogs with an average weight of 26.8±2.0 kg. Anesthesia was induced with thiamylal sodium and maintained with isoflurane. The chest was opened with a midsternotomy, the pericardium was incised and sutured to the chest wall to form a cradle for the heart, and the root of the aorta was dissected clear of adherent fat and connective tissue. A flow probe was placed snugly around the aorta proximal to the brachiocephalic artery. Another flow probe was placed on the descending thoracic aorta just above the branching of the first intercostal arteries. In a first series of three dogs (1E to 3E) electromagnetic flow probes (Statham SP2202, Gould Instruments, Cleveland) were used. In a second series of seven dogs (1 to 7) ultrasonic flow probes (Transonic Systems Inc., Ithaca, N.Y.) were used. Pressure measurements were performed with a 6F catheter tip transducer (Millar, Houston), which was inserted through a puncture wound in the left ventricular apex and was advanced beyond the aortic valves to lie in the aorta in the immediate vicinity of the ascending aorta flow probe.

Measurements were taken in basal and vasoconstricted states. Vasoconstriction was produced by injection of methoxamine in 1-mg boluses at 1-minute intervals until mean arterial pressure rose to approximately 140 mm Hg.

Signals from the pressure transducers were amplified using Accudata 218 amplifiers (Honeywell Inc., Fort Washington, Pa.) with low-pass filters set at 100 Hz to avoid aliasing. All signals were displayed on a Model 7414A thermic recorder (Hewlett-Packard Co., Medical Electronic Division, Waltham, Mass.) and on various oscilloscopes. Digital data were collected on-line using a Hewlett-Packard 1000 computing system. All signals were amplified to use the full 10-V range and 12-bit resolution of the analog-to-digital converter. The analog-to-digital converter was multiplexed with an 18-μsec slew interval between channels. The sampling rate was 250 samples/sec. The respirator was shut off 15 seconds before collection and analog-to-digital conversion of 10 seconds of pressure and flow during steady states. Ten contiguous beats of pressure and flow were aligned according to the peak of the R wave of the ECG. With the aid of a computerized procedure, these beats were subsequently normalized to mean length (i.e., mean number of samples per cycle); then, ensemble averages were taken to reduce noise.\textsuperscript{31} Normalization was
necessary because the heart rate was nearly rather than absolutely constant; that is, the number of samples per beat differed by 3–4%. The resulting mean cycles of ascending aortic pressure and flow and descending aortic flow were submitted to the analysis procedure.

**Modified T-Tube Model**

To investigate the reflected waves in the ascending aorta, we described the arterial system by a modified T-tube model.\textsuperscript{29,30} The electrical analog of this model is shown in Figure 1. Model parameters are described in the legend. Total peripheral resistance (ratio of mean ascending aortic pressure, $P$, to cardiac output, $Q$) is

$$R_p = R_{p1} R_{p2} / (R_{p1} + R_{p2}) = P/Q$$  \hspace{1cm} (1)

where $R_{p1}$ and $R_{p2}$ are peripheral resistances of the head and body portions of the circulation, respectively. The characteristic impedance of each tube is a real constant since these tubes are taken to be loss free ($Z_{in} = \sqrt{1/c_i}$, $i=1,2$). The quantities $l_i$ and $c_i$ ($i=1,2$) are the individual tube iner- tance and compliance per unit length, respectively. The short and long tube lengths are $d_1$ and $d_2$, respectively. Because load compliances are $C_1$ and $C_2$, the sum $C_t = C_1 + C_2 + c_1 d_1 + c_2 d_2$ represents total compliance of the systemic arterial bed. According to our most recent version of the model,\textsuperscript{30} the load resistors, $R_{Li}$ ($i=1,2$), were chosen to adapt the load with the tube at high frequencies. Thus

$$R_{Li} = R_{pi} Z_{ci} / (R_{pi} - Z_{ci}) \quad i=1,2$$  \hspace{1cm} (2)

Wave reflection is therefore absent at high frequencies. The impedance as seen at the input of the parallel tubes (joining with ascending aorta) can be approximated by the parallel of $Z_{Li}$ and $Z_{ci}$\textsuperscript{29,30}.

In the frequency domain ($\omega$), the input impedance of individual tubes assumes the following expression\textsuperscript{28,32}:

$$Z_i(\omega) = Z_{ci} + j Z_{ci} \tan(\beta d_i) \quad i=1,2$$  \hspace{1cm} (3)

where $Z_{Li}$ is the load impedance and $\beta$ is the phase constant. The latter is related to the velocity of propagation of a sinusoidal wave in the head and body tubes (blood phase velocity, $c_{phi}$) by the following equation:

$$\beta_i = \omega / c_{phi} = j \omega \sqrt{(l/c_i)} \quad i=1,2$$  \hspace{1cm} (4)

The load impedances have the following expressions:

$$Z_{Li}(\omega) = R_{Li}(1 + j \omega R_{Li} C_i) / [1 + j \omega C_i (R_{pi} + R_{Li})] \quad i=1,2$$  \hspace{1cm} (5)

The impedance as seen from the entrance of the T-tube model (ascending aorta input impedance) results in

$$Z_a(\omega) = 1 / [1/Z_1(\omega) + 1/Z_2(\omega)]$$  \hspace{1cm} (6)
The number of free parameters that characterize our T-tube model equals eight: \( R_{p1}, R_{p2}, C_1, C_2, (c_1d_1), (c_2d_2), Z_{c1}, \) and \( Z_{c2} \). Because the equation

\[
(l_d)_i = Z_{c1}^2(c_1d_1) \quad i = 1, 2 \tag{7}
\]

holds for each tube, the parameters \((l_d)_i\) and \((l_d)_2\) can be computed from the estimates of \( Z_{c1} \) and \( Z_{c2} \) and \((c_1d_1)\) and \((c_2d_2)\).

The model as it is formulated requires a priori specification of the peripheral resistances in the two wave transmission paths.\(^{29,30}\) This requirement can be satisfied when flows in the ascending aorta and in the upper descending aorta are measured. If the ratio between descending aorta mean flow \((\bar{Q}_2)\) and cardiac output \((\bar{Q})\) is defined as \( K_{da} \), that is,

\[
K_{da} = \frac{\bar{Q}_2}{\bar{Q}} \tag{8}
\]

then from this ratio and knowledge of \( R_p \) (Equation 1), the resistances \( R_{p1} \) and \( R_{p2} \) can be calculated

\[
R_{p1} = R_p \left(1 - K_{da}\right) \tag{9}
\]

\[
R_{p2} = R_p / K_{da} \tag{10}
\]

Calculation of \( R_{p1} \) and \( R_{p2} \) reduces the number of free model parameters to six. The procedure for estimating the parameters is described below.

**Parameter Estimation Method**

To estimate all free parameters of the model, that is, \((c_1d_1), (c_2d_2), C_1, C_2, Z_{c1}, \) and \( Z_{c2}\), both the measurements of ascending and descending aortic flows were used together with measured aortic pressure.\(^{30}\) Parameter estimation was accomplished by minimizing (Powell's algorithm was used\(^{33}\)) a cost function formulated as follows:

\[
F_c = \sum \left[\frac{(Q_m(t_i) - Q(t_i))^2}{Q^2} + \frac{(Q_{dm}(t_i) - Q_d(t_i))^2}{Q_d^2}\right]/2 \tag{11}
\]

In this equation \( Q(t_i) \) and \( Q_m(t_i) \) are the sampled values of the measured and model-generated ascending aortic flows, respectively. \( Q \) is mean flow. \( Q_d(t_i) \) and \( Q_{dm}(t_i) \) are the sampled values of the measured and model-generated descending aortic flows, respectively. \( Q_d \) is mean flow. For practical reasons, \( Q_d(t_i) \) was measured 4-12 cm downstream from the input of body circulation. Therefore, this flow pulse is from input flow \( Q_d(t_i) \). The mean flow values \( Q \) and \( Q_d \) are equal since the transmission tube has been assumed to be loss free. A frequency domain approach was used to obtain \( Q_m(t_i)\) and \( Q_{dm}(t_i) \) from the model. The mean cycles of pressure and flow were interpolated to obtain 128 samples/beat. Harmonics of ascending aortic pressure, \( P(j\omega) \), were calculated by the fast Fourier transform.\(^{34}\) These harmonics were divided by the T-tube input impedance, \( Z_{c1}(j\omega) \), to calculate the corresponding harmonics of ascending aortic flow, \( Q_m(j\omega) \):

\[
Q_m(j\omega) = P(j\omega)/Z_{c1}(j\omega) \tag{12}
\]

Measurements of ascending aortic pressure and flow were essentially taken immediately upstream from the junction where division occurs between headward vessels and descending aorta. This pressure is the same as pressure at the entrance of head and body transmission tubes of the model, since these tubes are connected in parallel. If \( Z_{c1}(j\omega) \) is the input impedance of the body tube, the ratio \( P(j\omega)/Z_{c1}(j\omega) \) gives harmonics of flow at the entrance of this tube. From this ratio, flow harmonics, \( Q_{dm}(j\omega) \), at the tube location where the flow probe was placed, were calculated\(^{35,32}\):

\[
Q_{dm}(j\omega) = \frac{P(j\omega)}{Z_{c1}(j\omega)} \frac{1 - \Gamma_2(j\omega) \exp(j2\omega\Delta t)}{[1 - \Gamma_2(j\omega) \exp(j2\omega\Delta t)]} \tag{13}
\]

The fraction at the extreme right of Equation 13 shifts the flow harmonics forth by a time interval of \( \Delta t \). This time interval was calculated from the foot-to-foot delay between measured ascending, \( Q(t) \), and descending, \( Q_d(t) \), aortic flows. The reflection ratio, \( \Gamma_2(j\omega) \), at the input of body tube was calculated from \( Z_{c1}(j\omega) \) and \( Z_{c2} \) as follows:

\[
\Gamma_2(j\omega) = \frac{Z_{c2}(j\omega) - Z_{c1}}{[Z_{c2}(j\omega) + Z_{c1}]} \tag{14}
\]

When Equation 14 is substituted into Equation 13, the latter is then expressed in terms of the parameters of the lower body tube.

Ascending and descending aorta flow waveshapes in the time domain, that is, \( Q_m(t) \) and \( Q_{dm}(t) \), were obtained by applying the inverse Fourier transform to the harmonics obtained from Equations 12 and 13, respectively. These flows were fitted to the measured flows (Equation 11), and the six free model parameters mentioned above were estimated.

**Calculation of Reflected Waves**

Pressure and flow waves at the input of individual tubes can be dissected into their forward and reflected (or backward) traveling components. If time-dependent signals are written as Fourier series, it holds for each harmonic that

\[
P_{f}(j\omega) + P_{b}(j\omega) = P_{1}(j\omega) = P(j\omega) \tag{15}
\]

\[
Q_{f}(j\omega) + Q_{b}(j\omega) = Q_{1}(j\omega) \tag{16}
\]

where \( P \) is pressure and \( Q \) is flow. The subscripts \( f \) and \( b \) indicate forward and backward, respectively. According to Equations 15 and 16 above, pressure and flow waves at the input of individual tubes are the sum of forward and backward waves. Because of parallel connection of the tubes, the input pressure to each tube is the same and equals aortic pressure, \( P(j\omega) \). Input flow harmonics \( Q_{f1}(j\omega) \) and \( Q_{b1}(j\omega) \) differ from each other. Harmonics of measured ascending aortic flow are given by the sum
Harmonics of forward pressure can be obtained from ascending aortic pressure harmonics after estimation of model parameters, that is, after calculating tube input impedances $Z_{i}(j\omega)$. The following procedure is used. The relation between forward and backward transmitted harmonic waves is given by the reflection coefficient $\Gamma_{i}(j\omega)$:

$$Q_{b}(j\omega) = -\Gamma_{i}(j\omega)Q_{i}(j\omega) \quad i=1,2 \quad (19)$$

Substitution of Equation 18 into Equation 15 results in

$$P_{b}(j\omega) = P(j\omega)/[1+\Gamma_{i}(j\omega)] \quad i=1,2 \quad (20)$$

Because the following relation holds:

$$\Gamma_{i}(j\omega) = \frac{Z_{i}(j\omega)-Z_{ci}}{[Z_{i}(j\omega)+Z_{ci}]} \quad i=1,2 \quad (21)$$

by substituting Equation 21 into Equation 20, forward pressure harmonics in each tube can be calculated from $P(j\omega)$ and $Z_{i}(j\omega)$.

After forward pressure harmonics have been calculated and taking into account that the following relations hold$^{3,29,32}$:

$$P_{i}(j\omega)/Q_{i}(j\omega) = Z_{ci} \quad i=1,2 \quad (22)$$

$$P_{bi}(j\omega)/Q_{bi}(j\omega) = -Z_{ci} \quad i=1,2 \quad (23)$$

forward flow harmonics can be computed from Equation 22. Finally, backward pressure can be computed from Equation 18 and backward flow from Equation 23 or 19. Time-dependent signals are given by the inverse Fourier transform.

After setting ascending aorta characteristic impedance equal to the parallel combination of $Z_{ci}$ and $Z_{c2}$, we can infer forward ($P_{a}$, $Q_{a}$) and backward ($P_{b}$, $Q_{b}$) pressure and flow waves in the ascending aorta upstream of its junction with upper and lower body circulatory systems.

The reflection coefficient as a function of harmonic frequency at this point is

$$\Gamma(j\omega) = \frac{Z_{a}(j\omega)-Z_{c}}{[Z_{a}(j\omega)+Z_{c}]} \quad (24)$$

where $Z_{a}$ is given by Equation 6. By analogy with Equations 20, 22, 18, and 19, harmonics of forward and backward pressure and flow can be computed as $P_{f}(j\omega) = P(j\omega)/[1+\Gamma(j\omega)]$, $Q_{f}(j\omega) = P_{f}(j\omega)/Z_{c}$, $P_{b}(j\omega) = \Gamma(j\omega)P_{f}(j\omega)$, and $Q_{b}(j\omega) = -\Gamma(j\omega)Q_{f}(j\omega)$, respectively. Time-dependent signals are obtained from the inverse Fourier transform.

Ascending aortic pressure in the time domain, $P(t)$, can be obtained from either one or the other of the following relations:

$$P(t) = P + P_{f}(t) + P_{b}(t) \quad (25)$$
Pressure and Impedance Patterns

(\(R_p\)) by applying Equation 7. Total peripheral resistance (\(R_p\)) was computed from the ratio of mean ascending aortic pressure and flow. Total systemic arterial compliance (\(C_s\)) was obtained from the sum of \(C_f = c_d + C\).

After model parameters were determined, forward and reflected waves in the ascending aorta were computed by applying the procedure described above.\(^2\)

The single-tube model was compared with the T-tube model in terms of ability to reproduce ascending aortic pressure from flow. In those circumstances for which the two models were equally able to describe measured pressure waveshapes, we compared their estimates of global parameters, that is, \(Z_c\) and \(C_s\). Further, the reflected waves as predicted in the ascending aorta by these two models were superimposed for comparison.

Results
Pressure and Flow Waveshapes and Impedance Patterns

Dogs were characterized and grouped according to features of the pressure waveshape. Characteristic differences among dogs were observed in both systole and diastole. Features observed during diastole were used to separate animals into groups.

Under basal conditions, the systolic peak of the ascending aortic pressure occurred during early systole to mid-systole in eight dogs (see cases BA6 in Figure 3 and BA2E in Figure 4), while it occurred during late systole just before the incisura in two animals (BA3 in Figure 4). During the diastolic period, a prominent diastolic wave was seen in eight animals (BA6 in Figure 3 and BA3 in Figure 4), while this diastolic wave was not detectable in two animals (BA2E in Figure 4). Those that possessed a distinct diastolic wave were placed in group A; those that did not were placed in group B.

Group A and B dogs had distinctive impedance patterns. In group A animals, the impedance modulus and phase displayed two distinct minima in the range from 0 to 10 Hz (Figures 3a and 5b). In group B animals, these minima were less distinct (Figures 5c and 5d).

On the average, the vasoconstrictor agent caused an increase in mean aortic pressure and total peripheral resistance by 54% and 87%, respectively. Variations in the contour of ascending aortic pressure waveshapes were observed from basal to vasocon-
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FIGURE 4. Same kind of data as reported in Figure 3. These signals, from left to right, pertain to dogs 2E and 3, respectively, under basal (BA) conditions; and to dog 7 under vasoconstricted (VC) conditions. PAO, pressure in the ascending aorta; PB, backward pressure component; PF, forward pressure component.

Stratified states. An example of major changes is shown in Figure 3. Differences in pressure waveshapes among animals were also observed in the vasoconstricted state (compare VC6 in Figure 3 and VC7 in Figure 4). Peak systolic pressure occurred during late systole in all dogs. During diastole, a distinct diastolic

FIGURE 5. Input impedance spectra of head circulation (broken lines), body circulation (solid lines), and the whole systemic arterial tree (triangles) as predicted by our T-tube model. Circles are data points obtained from the ratio of ascending aortic pressure harmonics to the corresponding flow harmonics. Panels a and b pertain to case BA3, group A. Panels c and d pertain to case BA2E, group B.
wave was observed in four dogs (group C), while pressure fell in a smooth and almost exponential fashion in another three dogs (group D). Groups C and D also had distinctive impedance patterns (see below).

Differences among dogs in the flow waveforms were also observed, but they could not be characterized with a distinct set of features as could the pressure waveforms. A summary of hemodynamic variables averaged over each subgroup is presented in Table 1.

**Data Fit**

Our T-tube model fitted the ascending and descending aortic flow waveforms equally well in every dog under all conditions and among group A, B, C, and D dogs. Examples of how well the measured flow waveforms were fitted are given in Figure 6. The averages of estimated and calculated model parameters are given in Table 2. No remarkable differences between average tube and load parameters were detectable between basal-state groups A and B. One notable difference existed in estimated parameters between vasoconstricted-state groups C and D. The transmission time to the body-end reflection site, \( \tau_2 \), was less in group D (30.3 msec) than in group C (50.7 msec). Differences in estimated parameters between the basal-state groups and the vasoconstricted-state groups were seen in the magnitude and relative values of the terminal resistances (higher and relatively more alike in the vasoconstricted state than in the basal state) and in the transmission times in both body-end and head-end tubes (longer and more equal between groups A and B in the basal state and shorter and more different between groups C and D in the vasoconstricted state).

Our T-tube model with parameters estimated from fits to the time domain flow waveforms also reproduced well the input impedance spectra as calculated from frequency harmonics of ascending aortic pressure and flow waveforms. Examples are shown in Figure 5. It can be seen that the model accurately captured all important minima, maxima, and phase zero crossing features of these spectra. Therefore, interpretations of impedance features based on the T-tube model were warranted.

A summary of impedance features for the four groups (A, B, C, and D) is presented in Figure 7. From this figure it can be seen that the average features of impedance patterns in groups A and B are much the same as in the individual examples given in Figure 5. Similarly, the average features for groups C and D are representative of the individuals. Thus, in group C, two minima can be identified in the magnitude spectrum, while in group D only one broad, indistinct minimum can be identified.

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### Table 1: Hemodynamic Variables

<table>
<thead>
<tr>
<th></th>
<th>( P ) (mm Hg)</th>
<th>( Q ) (ml/sec)</th>
<th>( Q_2 ) (ml/sec)</th>
<th>( Q_2/Q )</th>
<th>( R_p ) (( g \cdot \text{cm}^{-1} \cdot \text{sec}^{-1} ))</th>
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<tbody>
<tr>
<td>A (n=8)</td>
<td>84.6±8.4</td>
<td>43.4±8.7</td>
<td>29.7±8.8</td>
<td>0.67±0.10</td>
<td>2.70±0.58</td>
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<tr>
<td>B (n=2)</td>
<td>99.5±9.4</td>
<td>65.3±19.3</td>
<td>47.0±14.0</td>
<td>0.72±0.00</td>
<td>2.16±0.45</td>
</tr>
<tr>
<td>C (n=4)</td>
<td>134±3</td>
<td>37.7±9.5</td>
<td>23.8±6.1</td>
<td>0.59±0.03</td>
<td>5.12±1.53</td>
</tr>
<tr>
<td>D (n=3)</td>
<td>136±9</td>
<td>41.8±6.2</td>
<td>24.8±4.5</td>
<td>0.63±0.02</td>
<td>4.48±1.01</td>
</tr>
</tbody>
</table>

Data are mean±SD for n cases. \( P \) and \( Q \) are ascending aortic mean pressure and mean flow, respectively. \( Q_2 \) is the descending aortic mean flow. Groups A and B include cases, under basal conditions, that did (A) or did not (B) display a prominent diastolic fluctuation in aortic pressure. Cases that did or did not display this fluctuation under vasoconstricted conditions are grouped in C and D.

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**Figure 6.** Examples of fit between experimental and model-predicted ascending (FAO) and descending (FDAO) aortic flows.
TABLE 2. Estimated and Calculated Parameters of the Modified T-Tube Model

<table>
<thead>
<tr>
<th>Parameter</th>
<th>A (n=8)</th>
<th>B (n=2)</th>
<th>C (n=4)</th>
<th>D (n=3)</th>
</tr>
</thead>
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<tr>
<td>$R_0$ ($10^9$ g·cm$^{-4}$·sec$^{-1}$)</td>
<td>9.10±3.27</td>
<td>7.70±1.57</td>
<td>13.8±3.7</td>
<td>10.9±2.2</td>
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<td>$R_p$ ($10^9$ g·cm$^{-4}$·sec$^{-1}$)</td>
<td>4.14±1.23</td>
<td>3.00±0.63</td>
<td>8.15±2.62</td>
<td>7.64±1.89</td>
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<td>$C_1$ ($10^8$ g·cm$^{-1}$·cm$^2$·sec$^{-1}$)</td>
<td>92.0±43.2</td>
<td>130±29</td>
<td>102±26</td>
<td>45.8±18.5</td>
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<tr>
<td>$C_2$ ($10^8$ g·cm$^{-1}$·cm$^2$·sec$^{-1}$)</td>
<td>161±52</td>
<td>143±12</td>
<td>100±16</td>
<td>87.7±20.8</td>
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<tr>
<td>$Z_{st}$ (g·cm$^{-4}$·sec$^{-1}$)</td>
<td>367±87</td>
<td>316±68</td>
<td>325±62</td>
<td>427±80</td>
</tr>
<tr>
<td>$Z_{st}$ (g·cm$^{-4}$·sec$^{-1}$)</td>
<td>514±143</td>
<td>391±18</td>
<td>508±162</td>
<td>561±130</td>
</tr>
<tr>
<td>$c_{id}$ ($10^8$ g·cm$^{-1}$·cm$^2$·sec$^{-2}$)</td>
<td>90.6±32.9</td>
<td>99.3±30.6</td>
<td>63.6±14.0</td>
<td>45.7±27.4</td>
</tr>
<tr>
<td>$c_{d2}$ ($10^8$ g·cm$^{-1}$·cm$^2$·sec$^{-2}$)</td>
<td>160±59</td>
<td>193±27</td>
<td>105±16</td>
<td>59.3±22.9</td>
</tr>
<tr>
<td>$l_{id}$ (g·cm$^{-3}$)</td>
<td>11.0±2.8</td>
<td>9.04±1.06</td>
<td>6.48±1.48</td>
<td>7.09±4.20</td>
</tr>
<tr>
<td>$l_{d2}$ (g·cm$^{-3}$)</td>
<td>38.4±12.8</td>
<td>29.3±1.4</td>
<td>26.9±12.5</td>
<td>16.3±1.9</td>
</tr>
<tr>
<td>$C_i$ ($10^8$ g·cm$^{-1}$·cm$^2$·sec$^{-2}$)</td>
<td>503±138</td>
<td>566±99</td>
<td>371±30</td>
<td>238±54</td>
</tr>
<tr>
<td>$Z_e$ (g·cm$^{-4}$·sec$^{-1}$)</td>
<td>211±48</td>
<td>173±24</td>
<td>196±45</td>
<td>240±46</td>
</tr>
<tr>
<td>$\tau_1$ (10$^{-3}$ seconds)</td>
<td>30.6±5.7</td>
<td>29.3±2.9</td>
<td>19.9±3.4</td>
<td>17.8±10.6</td>
</tr>
<tr>
<td>$\tau_2$ (10$^{-3}$ seconds)</td>
<td>74.8±11.9</td>
<td>75.2±7.2</td>
<td>50.7±7.3</td>
<td>30.3±5.3</td>
</tr>
</tbody>
</table>

Data are mean±SD over n cases. $Z_c$ is characteristic impedance as seen at the junction of head and body circulations. $r_i=Z_{st}(c_{idi})$ (i=1,2) is the wave transit time from the input to the effective reflecting sites of head and body circulations, respectively. See legend of Figure 1 for meaning of other parameters.

From average impedance patterns, the following parameters were estimated: 1) the frequency ($f_{sa}$) at the first zero crossing of head (i=1) and body (i=2) impedance phase angles, 2) the frequency ($f_{ma}$) at the first minimum of impedance moduli, and 3) the phase ($\theta_i$) of load reflection coefficients at $f_{ma}$. These estimates are reported in Table 3. No significant differences in values of $\theta_i(f_{sa})$ were detectable among groups A-D. Differences between the basal-state groups and the vasoconstricted-state groups were seen in the values of both $f_{ma}$ and $f_{sa}$ (lower and more equal between groups A and B in the basal state, and higher and more varied between groups C and D in the vasoconstricted state).

**Forward and Reflected Waves**

Forward and backward pressure waves at the entrance of head and body transmission paths were calculated from ascending aortic pressure after estimation of model parameters, as described in “Methods.” These waves are named head-end and body-end reflected waves, respectively. Their time course

**FIGURE 7.** Input impedance moduli of head circulation (broken lines), body circulation (solid lines), and the whole systemic arterial tree (triangles) as obtained from our T tube model using parameter estimates averaged over groups A (panel a), B (panel b), C (panel c), and D (panel d).
Figure 8. Pressure measured with head and waves merged reflected group from in observed aortic systolic peak thus contributing affected. In the ascending in the body effective reflecting site was practically superimposed on the head-end reflected wave (VC6 in Figure 3). Both these waves contributed to the late systolic peak of aortic pressure so that no diastolic fluctuation was observed.

Comparison With Single-Tube Model

In general, the single-tube model was not able to give as good a fit between experimental and model-predicted ascending aortic pressures as the T-tube model was. As shown in Figure 8, the single-tube model failed in fitting prominent fluctuations caused by two discrete reflected waves. Estimates of model parameters averaged over groups A–D are given in Table 4. The parameters of the single-tube and T-tube models that can be directly compared are characteristic impedance $Z_c$ and total compliance $C_t$. On the average, the single tube showed a tendency to overestimate $C_c$ and to underestimate $Z_c$ with respect to our T-tube model.

In cases with type B and type D pressures (no diastolic fluctuation), the single-tube model was able to fit pressure as well as the T-tube model did. Examples of these fits are shown in Figure 9. In basal conditions (type B), total compliance was significantly overestimated (about 91%) by the single tube with respect to the T tube, while characteristic impedance was underestimated by 18% (compare Table 4 with Table 2). Under vasoconstricted conditions (type D), total compliance was overestimated by only 9%, while characteristic impedance was underestimated by 10%.

Table 3. Parameters of Average T-Tube Model Impedances

<table>
<thead>
<tr>
<th></th>
<th>A</th>
<th>B</th>
<th>C</th>
<th>D</th>
</tr>
</thead>
<tbody>
<tr>
<td>$f_{a1}$ (Hz)</td>
<td>4.4</td>
<td>4.3</td>
<td>6.3</td>
<td>7.5</td>
</tr>
<tr>
<td>$f_{a2}$ (Hz)</td>
<td>1.8</td>
<td>2.1</td>
<td>2.7</td>
<td>3.9</td>
</tr>
<tr>
<td>$f_{d1}$ (Hz)</td>
<td>4.8</td>
<td>4.9</td>
<td>7.3</td>
<td>8.3</td>
</tr>
<tr>
<td>$f_{d2}$ (Hz)</td>
<td>2.0</td>
<td>2.2</td>
<td>3.1</td>
<td>4.6</td>
</tr>
<tr>
<td>$\theta(f_{a1})$ (rad)</td>
<td>-1.1</td>
<td>-1.2</td>
<td>-1.2</td>
<td>-1.1</td>
</tr>
<tr>
<td>$\theta(f_{a2})$ (rad)</td>
<td>-1.1</td>
<td>-1.0</td>
<td>-1.1</td>
<td>-1.2</td>
</tr>
<tr>
<td>$c_{pa1}/d_1$ (sec$^{-1}$)</td>
<td>30.5</td>
<td>31.6</td>
<td>48.6</td>
<td>51.5</td>
</tr>
<tr>
<td>$c_{pa2}/d_2$ (sec$^{-1}$)</td>
<td>12.5</td>
<td>13.6</td>
<td>19.1</td>
<td>30.4</td>
</tr>
</tbody>
</table>

Data pertain to impedance patterns obtained from our T-tube model using parameter estimates averaged over groups A, B, C, and D. $f_{a1}$ and $f_{a2}$ are frequencies at the first minimum of impedance modulus of head and body circulation, respectively. $f_{d1}$ and $f_{d2}$ are frequencies at the first zero crossing of impedance phase angle of head and body circulation, respectively. $\theta(f_{a1})$ and $\theta(f_{a2})$ are load reflection coefficients at $f_{a1}$ and $f_{a2}$, respectively. $c_{pa1}$ and $c_{pa2}$ are phase velocities in the head and body transmission paths, respectively.

is shown in Figures 3 and 4. In these figures are also shown the forward and reflected pressure waves inferred in the ascending aorta, upstream of its junction with head and body circulations, after setting ascending aorta characteristic impedance equal to the parallel combination of $Z_{c1}$ and $Z_{c2}$.

In dogs of group A, the peak of body-end reflected wave occurred in diastole, after the incisura of the ascending aortic pressure. This peak was related to a prominent diastolic fluctuation in this pressure (BA6 in Figure 3 and BA3 in Figure 4). In the same dogs, the head-end reflected wave peaked in midsystole, thus contributing to midsystolic peak of ascending aortic pressure (BA6 in Figure 3); in late systole, just before the incisura, thus determining a secondary systolic peak (BA3 in Figure 4); or coincidently with the incisura, so that peak systolic pressure was not significantly affected. The last circumstance was also observed in group B (BA2E in Figure 4). Differently from group A, in group B the head-end and body-end reflected waves merged in such a way that a prominent oscillation was not produced in the diastolic portion of aortic pressure.

During vasoconstriction, in group C the head-end reflected wave peaked in late systole, close to the incisura of aortic pressure, while the body-end reflected wave peaked in diastole, after the incisura. The former contributed to peak systolic pressure. The latter was related to a prominent diastolic fluctuation (VC7 in Figure 4). In group D, the wave reflected from the body effective reflecting site was practically superimposed on the head-end reflected wave (VC6 in Figure 3). Both these waves contributed to the late systolic peak of aortic pressure so that no diastolic fluctuation was observed.

![Figure 8](http://ahajournals.org)  
**Figure 8.** Pressure measured in the ascending aorta (full squares) is compared with pressure predicted by the single tube (broken line) and the T tube (solid line). Left panel pertains to dog 3 and right panel pertains to dog 6.
Table 4. Estimated and Calculated Parameters of Single-Tube Model

<table>
<thead>
<tr>
<th></th>
<th>A (n=8)</th>
<th>B (n=2)</th>
<th>C (n=4)</th>
<th>D (n=3)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Zc (g·cm⁻¹·sec⁻¹)</td>
<td>185±52</td>
<td>141±27</td>
<td>183±39</td>
<td>217±47</td>
</tr>
<tr>
<td>cd (10⁻⁸g⁻¹·cm¹·sec²)</td>
<td>112±54</td>
<td>123±5</td>
<td>115±34</td>
<td>56.5±28.3</td>
</tr>
<tr>
<td>C (10⁻⁶g⁻¹·cm¹·sec⁵)</td>
<td>633±330</td>
<td>960±128</td>
<td>353±56</td>
<td>203±33</td>
</tr>
<tr>
<td>C₂ (10⁻⁶g⁻¹·cm¹·sec²)</td>
<td>745±362</td>
<td>1,083±123</td>
<td>468±27</td>
<td>260±59</td>
</tr>
<tr>
<td>ld (g·cm⁻²)</td>
<td>3.60±2.24</td>
<td>2.58±1.05</td>
<td>3.67±1.05</td>
<td>2.25±0.44</td>
</tr>
<tr>
<td>Rₚ (g·cm⁻¹·sec⁻¹)</td>
<td>200±59</td>
<td>152±33</td>
<td>188±43</td>
<td>228±49</td>
</tr>
</tbody>
</table>

Data are mean±SD over n cases. See legend of Figure 2 for meaning of parameters. Values of total peripheral resistance, Rₚ, are given in Table 1.

Discussion

To obtain a variety of features of ascending aortic pressure, measurements were made for 10 dogs. The presence or absence of a prominent diastolic fluctuation in this pressure was an essential issue for this study. Because it has been shown previously that this fluctuation is often found in open-chest dogs under basal conditions,²⁻⁹⁻¹⁰ while little or no diastolic fluctuation may be seen under vasoconstricted or hypertensive conditions,¹¹⁻¹⁳⁻¹⁴⁻¹⁶⁻¹⁷⁻¹⁸⁻¹⁹⁻²⁰⁻²¹⁻²²⁻²³⁻²⁴⁻²⁵⁻²⁶ we performed our experiments under both basal conditions and under the effect of an agent that caused vasoconstriction and increase of mean aortic pressure.

Time Domain Analysis

In all cases for which a prominent diastolic fluctuation was evident in ascending aortic pressure, the peak value of the reflected wave from the body effective reflecting site occurred during the diastolic interval. In addition, the head-end reflected waves were characterized by the presence of a secondary peak. This peak merged with the body-end reflected wave to enhance the diastolic fluctuation in aortic pressure (see cases BA6 in Figure 3 and BA3 and VC7 in Figure 4). The presence of a secondary peak in the head-end reflected wave indicated the presence of multiple reflections within a heart cycle. A forward wave, traveling toward the tube load, is reflected back from this load to the junction of the T tube. At this junction, the backward wave is reflected toward the load where reflection takes place until equilibrium is established. Furthermore, when a wave reflected from the terminal load reaches the junction of the T tube, not only is this wave partially reflected back toward the load, but part of it is also reflected forth toward the heart and the other tube load. Therefore, the head-end reflected wave affects the forward wave in the body transmission path and vice versa. As a consequence, the forward waves in the head and body transmission paths show pronounced peaks corresponding to peaks in the body-end and head-end reflected waves, respectively (compare forward and reflected waves in Figures 3 and 4).

The absence of a prominent diastolic fluctuation in ascending aortic pressure had two different explanations in basal (group B) and vasoconstricted (group D) conditions. In group B (BA2E in Figure 4), a prominent secondary peak in the head-end reflected wave was absent. The peak of the body-end reflected wave, which occurred in diastole, superimposed on a minimum of the head-end reflected wave so that a prominent diastolic oscillation in ascending aortic pressure did not occur. In group D (VC6 in Figure 3), the body-end reflected wave moved toward systole and superimposed on the head-end reflected wave, thus contributing to the late systolic peak of aortic pressure wave. At the same time, the diastolic part of this wave fell in a smooth and almost exponential fashion. However, movement of body-end reflected wave into systole was not a general result under vasoconstriction. Differently from group D, in group C the body-end reflected wave was distinct from the head-end reflected wave so as to cause a diastolic fluctuation in pressure (VC7 in Figure 4). This discrepancy can be explained by a much lower wave transit time (τ₂, Table 2) in the body circulation of group D with respect to group C. Because the complexity of the load reflection coefficient may play a significant role in timing of traveling waves,²⁸ more detailed conclusions can be drawn after frequency domain analysis (see below). Simultaneous arrival of head-end and body-end reflected waves in systole caused higher pulse pressure (ΔP=42.1±4.4 mm Hg) in group D than in group C (ΔP=33.5±3.4 mm Hg), while mean aortic pressures were similar (Table 1). It is interesting to observe that the pulse pressure determined in group C was similar to the pulse pressure determined in groups A (ΔP=30.9±9.5 mm Hg) and B (ΔP=29.8±10.9 mm Hg), for which head-end and body-end reflected waves were also distinct.

Wave reflection explains differences in contour of flow with respect to pressure. Because the tubes are loss free, forward pressure and forward flow in each tube have the same shape and are in phase. Backward pressure and backward flow also have the same shape, but are 180° out of phase. Therefore, the sum of forward and backward waves yields flow wave shapes markedly different from pressure wave shapes. The same consideration holds for backward pressure and flow in the ascending aorta.³⁻⁴⁻²⁵⁻²⁶ In cases for which the head-end reflected wave did not give a significant contribution to systolic pressure, ascending aortic flow and pressure were of similar shape (BA2E in Figures 4 and 6). In contrast, when the
head-end reflected pressure wave determined a late systolic peak, the corresponding reflected flow wave (opposite in phase) provoked a sharp decrease of ascending aortic flow after its early systolic peak (VC7 in Figures 4 and 6).

**Frequency Domain Analysis**

A good approximation of ascending aortic impedance spectra (modulus and phase) was obtained by our T-tube model (Figure 5). Further, it was shown by Burattini (Figure 8 of Reference 13) that impedance patterns predicted by the model at the entrance of head and body circulations resembled the experimental patterns determined by O'Rourke and Taylor.16 Figures 5 and 7 show that fluctuations in ascending aortic impedance patterns resulted from a parallel combination of head and body input impedances. In most cases, these fluctuations displayed two minima in the range from 0 to 10 Hz (groups A, B, and C; Figures 7a, 7b, and 7c). The first minimum in the modulus of ascending aortic impedance corresponded to the first minimum in the modulus of body-end impedance, while the second minimum of the aortic impedance modulus corresponded to the first minimum of head-end impedance. This finding confirms that minima at low frequency of ascending aortic impedance modulus result from interaction of reflected waves arising from two effective reflecting sites located in head and body systemic circulation.4,16,19 In group D (Figure 7d), head and body impedances merged so one minimum was dominant in the aortic impedance modulus. In these cases, head-end and body-end reflected waves superimposed, thus simulating the effect of a single effective reflecting site.

Timing of reflected waves depends on the wave phase velocity \( c_{phi} \), on the effective length \( d_i \) of each transmission path, and since the load is complex, on phase \( \phi_i \) of load reflection coefficient.3,28 All these factors are involved in the equation, which according to Burattini and Di Carlo,28 allows calculation of \( d_i \):

\[
d_i = \frac{c_{phi}}{4f_{d_i}} (1 + \phi_i(f_{d_i})/\pi) \quad i=1,2
\]

where \( f_{d_i} \) is the frequency at the first zero crossing of the head \((i=1)\) and body \((i=2)\) impedance phase angles, and \( \phi_i(f_{d_i}) \) is the corresponding phase of the load reflection coefficient. Values reported in Table 3 indicate that \( f_{d_i} \) increased from basal to vasoconstricted conditions. Because there was no significant difference among values of \( \phi_i(f_{d_i}) \), and assuming that \( d_1 \) and \( d_2 \) do not change with physiological conditions, we can conclude that an increase in \( f_{d_i} \) indicates an increase in phase velocity. Within vasoconstricted conditions, group D (no diastolic fluctuation) showed an increase of \( f_{d_1} \) by 50% with respect to group C, while \( f_{d_2} \) increased by 14%. Therefore, a significant increase of phase velocity in the body-end transmission path explains superimposition of reflected waves in group D. Comparison of values of compliance in Table 2 indicates that greater pulse wave velocity in group D with respect to group C was determined by reduced distributed elasticity.

**Comparison With Single-Tube Model**

The present study confirmed our previous finding that the single-tube model is not able to account for a prominent diastolic oscillation in ascending aortic pressure since it suffers from the absence of a second transmission path.29 When two of these tubes were combined in parallel to constitute our T-tube model, aortic pressure-flow relations were satisfactorily described. Therefore, we can conclude that two, instead of one, arterial effective reflecting sites are seen by the left ventricle. However, a point of controversy was caused by the observation that, when a prominent diastolic oscillation in ascending aortic pressure was negligible, the single-tube model described aortic pressure-flow relations as well as the T-tube model did (Figure 9). This result suggests that two effective reflecting sites may appear as one to the heart. If this hypothesis holds, the formulations of the arterial system in terms of the single tube and T tube should be equivalent. In quantitative terms, model equivalence requires that, not only is a similar goodness of fit between experimental and model predicted pulse waves obtained from the two models, but also that their parameter estimates are comparable. For a correct comparison, we considered values of global parameters, \( Z_s \) and \( C_s \), estimated in cases for which the same goodness of fit was obtained from the single-tube and T-tube models, that is, groups B and D. The big discrepancy between single-tube and T-tube compliance found under basal conditions (compare \( C_s \) of group B in Tables 2 and 4) suggests that, under these circumstances, the two models are not equivalent representations of the systemic arterial tree. This conclusion is confirmed by the fact that the reflected waves predicted in the ascending aorta by these two models were also significantly different (see case BA2E in Figure 9). In contrast, under vasoconstricted conditions (group D), both the single-tube and T-tube models gave comparable estimates of \( C_s \) and \( Z_s \). Moreover, the reflected waves predicted by these models in the ascending aorta were practically coincident (see case VC6 in Figure 9). Only in these circumstances, the two models can be considered equivalent formulations of the systemic arterial tree and it can be concluded that two effective reflecting sites appear as one to the heart.

**Comparison With Aortic Pressure Waveshapes in Humans**

The aortic pressure wave rising to a late systolic peak and, then, falling smoothly during diastole with little or no diastolic fluctuation has been observed in elderly humans and in those with hypertension and arterial degeneration. These pressure waves are similar to those of dogs that were grouped in D in the present study and are represented by case VC6 in
Figure 9. Upper panels: Pressure measured in the ascending aorta (squares) is compared with pressure predicted by the single-tube (broken line) and T-tube (solid line) models. Lower panels: Comparison between backward pressure (PB) waves described by the single-tube (broken line) and the T-tube (continuous line) models in the ascending aorta. Left panels pertain to dog 2E under the basal condition. Right panels pertain to dog 6 under the vasoconstricted condition.

Figure 3. In contrast, pressure wave contour in children and young adults has a diastolic fluctuation as in most of our dogs under basal conditions.\textsuperscript{3,4,6,18} From results of this study and in accordance with O’Rourke,\textsuperscript{4,18} we can infer that progressive arterial stiffening with advancing years increases pulse wave velocity. As a consequence, the body-end reflected wave moves into systole and contributes to the late systolic peak. This conclusion is consistent with the fact that late systolic peak of aortic pressure was markedly augmented during the perturbation of the body-end reflected wave via bilateral compression of femoral arteries.\textsuperscript{22}

Murgo et al\textsuperscript{22} and Latham et al\textsuperscript{6,35} have drawn attention to inflection and secondary systolic fluctuation in pressure waves recorded in the ascending aorta of “normal” humans. This inflection has been interpreted to represent the foot of a wave reflected from terminal aorta. An early diastolic oscillation in aortic pressure, when present, was considered another part of this reflected wave. An aortic pressure pattern with a well-defined systolic inflection and secondary fluctuation similar to that designated as type A by Murgo et al is that plotted in our Figure 4, case BA3. In contrast to the explanation given by these authors, this figure shows that the late systolic peak is due to the head-end reflected wave, while the body-end reflected wave contributes to diastolic fluctuation. However, as the arterial pressure wave travels away from the ascending aorta down to the abdominal aorta, the longer the distance from ascending aorta, the more the reflected wave component moves into systole, thus enforcing the secondary systolic peak (see BA3 in Figure 4 and Figure 10). As a consequence, the systolic pressure pulse increases in amplitude from the ascending aorta to the effective reflecting site, while the systolic inflection point occurs earlier in systole. This finding explains experimental observations by Murgo et al\textsuperscript{22} and Latham et al\textsuperscript{35} along the aorta.

Limitations of Model

Our modified T-tube model is necessarily a reduced representation of the systemic arterial tree. Many aspects of arterial distribution and wave transmission have been disregarded for model simplification so that the number of parameters was kept low and could be estimated with a limited number of simultaneous measurements, that is, ascending aortic pressure and flow and descending aortic flow. The minimum set of measurements required was previously assessed after analysis of accuracy of model...
FIGURE 10. Upper panels: Pressure predicted by our T-tube model at half the distance to the body effective reflecting site (left panel) and at body effective reflecting site (right panel). Lower panels: Forward (continuous lines) and backward (broken lines) waves (without DC component) calculated at the same locations. Data pertain to dog 3 under the basal condition.

parameter estimates. The most significant aspects not represented in the model are wave attenuation, geometric and elastic tapering, nonlinear pressure–diameter relation, and vessel branching beyond main branching into head and body circulation at the level of the aortic arch. Therefore, head and body tubes were assumed to be uniform and loss free and were loaded with an effective low-pass filter complex load rather than with branching elements. Nevertheless, this model is a great improvement with respect to the original T-tube. Substitution of terminal resistors with complex terminal loads was essential to this improvement. Further, we showed previously that changes of the ratio $K_{3a}$ between $R_p$ and $R_{p2}$ did not affect significantly the estimates of the other model parameters. These findings suggest that geometric and elastic properties of proximal vessels are more important than peripheral resistances in accounting for wave transmission effects as seen from the ascending aorta. This conclusion is consistent with the fact that a significant discrete reflection was found by Latham et al in the abdominal aorta, where significant reduction in vessel size, wave speed, and wall elasticity occurred. However, to account for this discontinuity, a nonhomogeneous two-tube conceptual model, which located two effective reflecting sites in the body circulation, was supported by these authors. This model is comparable with the body-tube component of our T-tube model because both of them have a complex reflection coefficient at the main effective reflecting site. The main difference is the fact that, in the two-tube model, a narrower transmission tube, rather than a lumped complex load, was placed downstream to this site to account for wave reflection arising from peripheral resistances. Reflective properties of head circulation are disregarded in the two-tube model. In contrast, results of the present study indicate that these properties play a significant role.

Formulation of terminal loads of our T-tube model as well as evaluation of distances to head and body effective reflecting sites, and influence of nonuniform properties of transmission tubes, deserve further quantitative analysis. These properties may play different roles in different mammalian species for which the T-tube concept has been shown to be applicable.

References

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Key Words · arterial parameter estimation · arterial wave reflection · hypertension · single-tube arterial model · T-tube arterial model · vasconstriction