Aortic input impedance in infants and children

M. KEITH SHARP, GEORGE M. PANTALOS, LUANN MINICH, LLOYD Y. TANI, EDWIN C. MCGOUGH, AND JOHN A. HAWKINS

Biofluid Mechanics Laboratory, Department of Bioengineering and Department of Surgery and Department of Bioengineering, University of Utah, Salt Lake City 84112; and Department of Pediatrics, Division of Cardiology, and Department of Surgery, Division of Cardiothoracic Surgery, Primary Children's Medical Center, Salt Lake City, Utah 84113

Sharp, M. Keith, George M. Pantalos, Luann Minich, Lloyd Y. Tani, Edwin C. McGough, and John A. Hawkins. Aortic input impedance in infants and children. J Appl Physiol 88: 2227–2239, 2000.—Flow and pressure measurements were performed in the ascending aortas of six pediatric patients ranging in age from 1 to 4 yr and in weight from 7.2 to 16.4 kg. From these measurements, input impedance was calculated. It was found that total vascular resistance decreased with increasing patient weight and was approximately one to three times higher than those of adults. Conductance per unit weight was relatively constant but was approximately three times higher than for adults. Strong inertial character was observed in the impedance of four of the six patients. Among a three-element and two four-element lumped-parameter models, the model with characteristic aortic resistor (Rₐ) and inertia in series followed by parallel peripheral resistor (Rₚ) and compliance fitted the data best. Rₐ decreased with increasing patient weight and was one to three times higher than in adults, and Rₚ decreased with increasing patient weight and was 2 to 15 times higher. The Rₚ-to-Rₐ ratio differed significantly between infants and children vs. adults. The results suggested that Rₚ developed more rapidly with patient weight than did Rₐ. Compliance values increased with increasing patient weight and were 3-16 times lower than adult values.

flow; pressure; resistance; compliance

EARLY PRESSURE AND FLOW MEASUREMENTS and subsequent calculations of human adult aortic input impedance were made by Gabe et al. (3) and Patel et al. (11). Subsequent measurements were performed by Mills et al. (5), Nichols et al. (8), Murgo et al. (7), and O'Rourke and Avolio (9). Although considerable variability was found, in many individuals the modulus fell to a shallow minimum of 8–10% of the zero frequency (Z₀) found, in many individuals the modulus fell to a

and Avolio (9). Although considerable variability was expected. First, because of somewhat smaller aortic pulse pressure but much smaller cardiac stroke volume, it was expected that infant/child vascular systems would exhibit lower compliance. Compliance may be roughly estimated as stroke volume divided by pulse pressure. Lower compliance is possible in spite of the increased tissue elasticity of younger vessels (13), because compliance is a volumetric strain response, whereas elasticity is a linear strain response. Lower compliance would decrease the impedance phase, particularly at low frequency. Second, because average aortic pressures are somewhat smaller in infants and children, but average flow rates (cardiac outputs) are much lower, total vascular resistance must be increased. Because peripheral resistance (Rₚ) normally provides most of the total vascular resistance, this factor would raise the modulus not only at Z₀ but across all frequencies. Third, because of the shorter vessel lengths in infants and children, shorter wave reflection times and, therefore, modulus minima at higher frequencies were expected.

Several lumped-parameter models [with resistance (R), compliance (C), and inertance (L) elements] have been applied to the adult systemic vasculature (Fig. 1). The two-element model (2) (termed RC) provides the important characteristic of a drop in impedance modulus along with negative impedance phase with increasing frequency, although these characteristics are accompanied by an asymptotic modulus at high frequency of zero and an asymptotic phase at high frequency of −90°, both of which are nonphysiological. The three-element model (19) (termed RCR) provides a significant improvement over the two-element model in that its modulus has a nonzero high-frequency asymptote and
its phase has a high-frequency asymptote of zero. However, Burkhoff et al. (1) found that, whereas the RCR model provided a reasonable representation of afterload for predicting integrated measures of cardiac performance, such as stroke volume, stroke work, and average systolic and diastolic aortic pressures, it did not provide realistic aortic pressure and flow waveforms, and it significantly underestimated peak aortic flow. The four-element model (14) (termed RLRC) provides potential for improvement with a minimum modulus at intermediate frequency and increasing modulus for higher frequency (whereas the RCR model has a monotonically decreasing modulus) and positive phase at high frequency. An alternative four-element model (17) (termed RLRC2) also provides a minimum modulus at intermediate frequency but with a constant high-frequency asymptote compared with the increasing asymptote of the RLRC model. The RLRC2 model provides a transition from negative to positive phase at intermediate frequency but with a high-frequency asymptote of zero compared with the 90° asymptote of the RLRC model. Five-element models have also been proposed and compared with more simple models (14, 18).

This study provides what is thought to be the first report of simultaneous high-fidelity measurements of aortic pressure and flow and calculations of impedance in infants and children. The fits of the RCR, RLRC, and RLRC2 models to the results were evaluated.

**METHODS**

Six patients at the Primary Children's Medical Center (Salt Lake City, UT) were enrolled in this study between June 1996 and August 1997. Approval to conduct the investigation was obtained from the local Institutional Review Board, and written informed consent was obtained from the parents of the patients before enrollment in the study. All patients were admitted for repair of simple cardiac defects and had normal aortic anatomy, stable cardiac function, and no previous history of cardiac or thoracic surgery. The patients ranged in age from 0.8 to 4.0 yr (2.2 ± 1.4 yr, mean ± SD) and in weight from 7.2 to 16.4 kg (11.5 ± 3.9 kg). Five of the patients underwent surgery for closure of atrial septal defects, and one underwent surgery for pulmonary valvotomy and patch angioplasty of the right pulmonary artery. The ages and weights of the patients, along with average aortic pressures and flows, are presented in Table 1.

The patients, measurement procedures, and flow and pressure data are the same as those used by Pantalos et al. (10) in the investigation of intra-aortic balloon pump timing errors. Briefly, a transit time ultrasonic perivascular flow probe (10 or 14 mm, 100-Hz frequency response, Transonic Systems, Ithaca, NY) was positioned around the aorta, 1 cm downstream of the aortic valve after surgical exposure of the heart but before cannulation for cardiopulmonary bypass. A high-fidelity catheter tip pressure transducer (5-Fr MPC-500, 5-kHz frequency response, Millar Instruments, Houston, TX) was placed in the aortic root through a hole, and purse-string

**Table 1. Patient characteristics**

<table>
<thead>
<tr>
<th>Patient</th>
<th>Weight, kg</th>
<th>Age, yr</th>
<th>Heart Rate, beats/min</th>
<th>Mean Pressure, mmHg</th>
<th>Systolic Pressure, mmHg</th>
<th>Diastolic Pressure, mmHg</th>
<th>Mean Flow, l/min</th>
<th>Resistance, dyn·s·cm⁻²</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>8.6</td>
<td>1.1</td>
<td>152.4</td>
<td>55.17</td>
<td>62.07</td>
<td>47.37</td>
<td>1.42</td>
<td>3,113</td>
</tr>
<tr>
<td>2</td>
<td>8.4</td>
<td>1.5</td>
<td>124.2</td>
<td>48.99</td>
<td>54.05</td>
<td>42.87</td>
<td>1.09</td>
<td>3,593</td>
</tr>
<tr>
<td>3</td>
<td>7.2</td>
<td>0.8</td>
<td>160.2</td>
<td>73.17</td>
<td>90.41</td>
<td>56.43</td>
<td>1.55</td>
<td>3,771</td>
</tr>
<tr>
<td>4</td>
<td>15.0</td>
<td>3.8</td>
<td>141.6</td>
<td>72.15</td>
<td>79.11</td>
<td>67.87</td>
<td>3.06</td>
<td>1,844</td>
</tr>
<tr>
<td>5</td>
<td>13.2</td>
<td>2.0</td>
<td>147.0</td>
<td>48.79</td>
<td>52.88</td>
<td>46.18</td>
<td>1.67</td>
<td>2,336</td>
</tr>
<tr>
<td>6</td>
<td>18.4</td>
<td>4.0</td>
<td>105.0</td>
<td>66.43</td>
<td>75.10</td>
<td>56.35</td>
<td>3.47</td>
<td>1,530</td>
</tr>
</tbody>
</table>
suture, that was later used for the cardioplegia infusion line at the time cardiopulmonary bypass was initiated. The catheter was advanced down the ascending aortic lumen so that the sensor was at the level of the flow probe. Consequently, the pressure and flow measurements were made simultaneously at the same location in the ascending aorta. Flow and pressure waveforms were recorded on FM tape (MR-30 data tape recorder, 313-Hz frequency response; TEAC America, Montebello, CA).

Segments of the analog recordings were subsequently digitized (GW Instruments MacADIOS analog-to-digital board in a Macintosh 8100 computer running Superscope 2.17 data-acquisition software) over 5 s with 0.005-s sampling interval (10,000 samples). The beginnings of the waveforms were selected from digitized data by identifying the sharp rise in pressure between diastole and systole. The beginning and end of each pressure cycle were similarly identified. Ten cycles were used, except for patient 6, whose heartbeat frequency was low enough that only eight cycles of digitized data were available in the 5-s sampling window. A Fourier transform was calculated for each cycle of pressure and flow. Input impedance was then calculated for each cycle as the ratio of pressure and flow. The real and imaginary parts of the impedance values were averaged, and standard deviations were calculated. A threshold of quality for the resulting average impedance values was established. Harmonics that had standard deviations of either real or imaginary parts >20% of the zeroth harmonic modulus were excluded, along with all higher harmonics for the same patient data. The maximum number of harmonics was arbitrarily limited to 10. These impedance values were compared with the average impedance of five adults (8).

The RCR model was fit to the impedance of each patient and to the adult average. Two fitting procedures were evaluated. First, with the sum of characteristic aortic resistance (R\textsubscript{c}) and R\textsubscript{p} set equal to the measured modulus at Z\textsubscript{0}, R\textsubscript{c}, and C were found iteratively such that the normalized root mean square difference (NRMS) in measured and modeled impedance was minimized

\[
NRMS = \sqrt{\frac{\sum_{i=0}^{m} (\text{Re}(Z_i) - \text{Re}(A_i))^2 + (\text{Im}(Z_i) - \text{Im}(A_i))^2}{(m + 1)Z_0}}
\]

(1)

where Re and Im are the real and imaginary parts of the measured impedance (Z) and the modeled impedance (A), and m is the number of harmonics satisfying the accuracy criterion. Second, with the sum of R\textsubscript{c} and R\textsubscript{p} again set equal to Z\textsubscript{0}, R\textsubscript{c}, and C were found iteratively such that the first harmonic modulus and phase were fit exactly. The NRMS was also calculated for this second fitting method.

The four-element models were fit to the measured patient impedance values and to the adult average by procedures similar to the first method above. For the RLRC model, with the sum of R\textsubscript{c} and R\textsubscript{p} set equal to Z\textsubscript{0}, R\textsubscript{c}, C, and L were found iteratively such that NRMS was minimized. For the RLRC2 model, with R\textsubscript{p} set equal to Z\textsubscript{0}, R\textsubscript{c}, C, and L were found iteratively such that NRMS was minimized. For each fit, the sensitivity of NRMS to parameter values was tested, including, for the four-element models, variations in L and C with constant LC product (constant crossover frequency; see DISCUSSION).

In addition, arterial compliance was also calculated for the patient data by the area (4) and pulse pressure methods for comparison to the results of the curve fits described above. These calculations were based on the first complete cycle only. Mean systolic and mean diastolic pressures (Table 1) were also calculated from the digitized data for the first cycle only. General correlation coefficients (r) and 95% confidence intervals (CI\textsubscript{95}) were calculated for the correlations of results with both patient weight and age. ANOVA comparisons were performed for the results from pediatric patient vs. adults and for results from infants (patients 1 and 3) vs. the group of children (patients 2, 4, 5, and 6). (Although patient 1 was not technically an infant, being slightly older than 1 yr, the similarity of the response of patient 1 and true infant patient 3 promoted this grouping and description of the groups for comparison purposes.)

RESULTS

Example pressure and flow waveforms used for impedance calculations are shown in Fig. 2. The cycle periods were constant for each patient within ±1 sampling interval except for patient 5, for whom the variation was ±1.5 sampling intervals, and patient 6, for whom the variation was ±2.5 sampling intervals. Mean pressures and flows for each 10-cycle (8-cycle for patient 6) data set are given in Table 1. Although mean flow increased with patient weight and was reasonably

![Fig. 2. Example of aortic pressure (AOP; top) and aortic flow (AOQ; bottom) waveforms (patient 2). Ticks on abscissas indicate 1-s intervals.](image)

![Fig. 3. Mean R calculated as the ratio of mean pressure and mean flow. Correlation with patient weight is as follows: r = -0.988, 95% confidence interval (CI\textsubscript{95}) = -0.999 < r < -0.888.](image)
### Table 2. Fourier components of pressure, flow and impedance

<table>
<thead>
<tr>
<th>Harmonic</th>
<th>Patient 1</th>
<th></th>
<th>Patient 2</th>
<th></th>
<th>Patient 3</th>
<th></th>
<th>Patient 4</th>
<th></th>
<th>Patient 5</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pressure, mmHg</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Modulus</td>
<td>55.17</td>
<td>5.96</td>
<td>1.89</td>
<td>0.98</td>
<td>0.84</td>
<td>0.55</td>
<td>0.50</td>
<td>0.28</td>
<td>0.30</td>
</tr>
<tr>
<td>Phase, °</td>
<td>0.00</td>
<td>0.76</td>
<td>0.24</td>
<td>0.46</td>
<td>0.43</td>
<td>0.35</td>
<td>0.31</td>
<td>0.40</td>
<td>0.42</td>
</tr>
<tr>
<td>%SD 2</td>
<td>2.72</td>
<td>0.77</td>
<td>0.42</td>
<td>0.46</td>
<td>0.43</td>
<td>0.35</td>
<td>0.31</td>
<td>0.40</td>
<td>0.42</td>
</tr>
<tr>
<td>Flow, l/min</td>
<td>1.42</td>
<td>1.06</td>
<td>0.45</td>
<td>0.16</td>
<td>0.15</td>
<td>0.05</td>
<td>0.06</td>
<td>0.06</td>
<td>0.04</td>
</tr>
<tr>
<td>Modulus</td>
<td>5.60</td>
<td>0.66</td>
<td>1.13</td>
<td>3.50</td>
<td>4.57</td>
<td>14.47</td>
<td>7.99</td>
<td>8.34</td>
<td>7.50</td>
</tr>
<tr>
<td>Phase, °</td>
<td>6.00</td>
<td>13.82</td>
<td>87.19</td>
<td>25.63</td>
<td>24.76</td>
<td>65.97</td>
<td>58.02</td>
<td>58.05</td>
<td>79.87</td>
</tr>
<tr>
<td>%SD 2</td>
<td>6.50</td>
<td>27.56</td>
<td>14.03</td>
<td>16.24</td>
<td>34.98</td>
<td>25.55</td>
<td>26.41</td>
<td>22.52</td>
<td>12.42</td>
</tr>
<tr>
<td>Impedance, dyn·s·cm⁻¹</td>
<td>3,125.90</td>
<td>448.66</td>
<td>340.70</td>
<td>492.55</td>
<td>450.52</td>
<td>906.79</td>
<td>574.12</td>
<td>456.40</td>
<td>570.80</td>
</tr>
<tr>
<td>Modulus</td>
<td>3,594.30</td>
<td>434.53</td>
<td>312.10</td>
<td>328.41</td>
<td>462.83</td>
<td>589.03</td>
<td>852.30</td>
<td>475.30</td>
<td>768.55</td>
</tr>
<tr>
<td>Phase, °</td>
<td>2.08</td>
<td>0.53</td>
<td>0.65</td>
<td>1.47</td>
<td>2.27</td>
<td>7.14</td>
<td>5.49</td>
<td></td>
<td></td>
</tr>
<tr>
<td>%SD 2</td>
<td>1.89</td>
<td>0.82</td>
<td>1.96</td>
<td>2.15</td>
<td>6.54</td>
<td>5.85</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Pressure, mmHg</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Modulus</td>
<td>73.37</td>
<td>11.10</td>
<td>3.14</td>
<td>1.83</td>
<td>0.76</td>
<td>1.03</td>
<td>0.42</td>
<td>0.68</td>
<td>0.10</td>
</tr>
<tr>
<td>Phase, °</td>
<td>0.00</td>
<td>0.32</td>
<td>0.30</td>
<td>0.20</td>
<td>0.47</td>
<td>0.34</td>
<td>0.72</td>
<td>0.37</td>
<td>0.38</td>
</tr>
<tr>
<td>%SD 2</td>
<td>2.93</td>
<td>2.85</td>
<td>2.10</td>
<td>1.35</td>
<td>1.21</td>
<td>0.73</td>
<td>0.72</td>
<td>0.37</td>
<td>0.38</td>
</tr>
<tr>
<td>Flow, l/min</td>
<td>1.09</td>
<td>0.85</td>
<td>0.47</td>
<td>0.18</td>
<td>0.18</td>
<td>0.06</td>
<td>0.06</td>
<td>0.06</td>
<td>0.01</td>
</tr>
<tr>
<td>Modulus</td>
<td>6.70</td>
<td>12.10</td>
<td>2.70</td>
<td>2.85</td>
<td>5.31</td>
<td>3.28</td>
<td>1.23</td>
<td>1.88</td>
<td>2.40</td>
</tr>
<tr>
<td>Phase, °</td>
<td>0.00</td>
<td>0.12</td>
<td>0.31</td>
<td>0.21</td>
<td>0.06</td>
<td>0.11</td>
<td>0.02</td>
<td>0.02</td>
<td>0.05</td>
</tr>
<tr>
<td>%SD 2</td>
<td>1.54</td>
<td>1.12</td>
<td>0.31</td>
<td>0.21</td>
<td>0.06</td>
<td>0.11</td>
<td>0.02</td>
<td>0.02</td>
<td>0.05</td>
</tr>
<tr>
<td>Pressure, mmHg</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Modulus</td>
<td>72.15</td>
<td>4.98</td>
<td>1.98</td>
<td>1.15</td>
<td>1.27</td>
<td>0.34</td>
<td>0.89</td>
<td>0.11</td>
<td>0.50</td>
</tr>
<tr>
<td>Phase, °</td>
<td>0.00</td>
<td>0.38</td>
<td>0.24</td>
<td>0.16</td>
<td>0.18</td>
<td>0.15</td>
<td>0.28</td>
<td>0.22</td>
<td>0.19</td>
</tr>
<tr>
<td>%SD 2</td>
<td>1.89</td>
<td>2.48</td>
<td>1.16</td>
<td>0.47</td>
<td>0.36</td>
<td>0.14</td>
<td>0.27</td>
<td>0.04</td>
<td>0.11</td>
</tr>
<tr>
<td>Flow, l/min</td>
<td>3.06</td>
<td>4.94</td>
<td>3.13</td>
<td>1.99</td>
<td>1.57</td>
<td>1.59</td>
<td>2.11</td>
<td>1.89</td>
<td>1.09</td>
</tr>
<tr>
<td>Modulus</td>
<td>6.49</td>
<td>12.10</td>
<td>2.70</td>
<td>2.85</td>
<td>5.31</td>
<td>3.28</td>
<td>1.23</td>
<td>1.88</td>
<td>2.40</td>
</tr>
<tr>
<td>Phase, °</td>
<td>0.00</td>
<td>0.12</td>
<td>0.31</td>
<td>0.21</td>
<td>0.06</td>
<td>0.11</td>
<td>0.02</td>
<td>0.02</td>
<td>0.05</td>
</tr>
<tr>
<td>%SD 2</td>
<td>1.54</td>
<td>1.12</td>
<td>0.31</td>
<td>0.21</td>
<td>0.06</td>
<td>0.11</td>
<td>0.02</td>
<td>0.02</td>
<td>0.05</td>
</tr>
<tr>
<td>Pressure, mmHg</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Modulus</td>
<td>48.79</td>
<td>3.49</td>
<td>1.02</td>
<td>0.79</td>
<td>0.48</td>
<td>0.31</td>
<td>0.37</td>
<td>0.24</td>
<td>0.24</td>
</tr>
<tr>
<td>Phase, °</td>
<td>0.00</td>
<td>0.76</td>
<td>0.24</td>
<td>0.46</td>
<td>0.43</td>
<td>0.35</td>
<td>0.31</td>
<td>0.40</td>
<td>0.42</td>
</tr>
<tr>
<td>%SD 2</td>
<td>1.81</td>
<td>0.54</td>
<td>0.32</td>
<td>0.25</td>
<td>0.25</td>
<td>0.24</td>
<td>0.22</td>
<td>0.36</td>
<td>0.21</td>
</tr>
<tr>
<td>Flow, l/min</td>
<td>1.42</td>
<td>1.06</td>
<td>0.45</td>
<td>0.16</td>
<td>0.15</td>
<td>0.05</td>
<td>0.06</td>
<td>0.06</td>
<td>0.04</td>
</tr>
<tr>
<td>Modulus</td>
<td>2,340.50</td>
<td>216.41</td>
<td>146.44</td>
<td>270.51</td>
<td>228.19</td>
<td>331.55</td>
<td>295.82</td>
<td>510.55</td>
<td>277.85</td>
</tr>
<tr>
<td>Phase, °</td>
<td>0.00</td>
<td>0.12</td>
<td>0.31</td>
<td>0.21</td>
<td>0.06</td>
<td>0.11</td>
<td>0.02</td>
<td>0.02</td>
<td>0.05</td>
</tr>
<tr>
<td>%SD 2</td>
<td>4.36</td>
<td>0.74</td>
<td>0.88</td>
<td>1.98</td>
<td>1.73</td>
<td>7.25</td>
<td>4.65</td>
<td>12.83</td>
<td>4.02</td>
</tr>
</tbody>
</table>
correlated \((r = 0.886, \text{CI}_{95} = 0.265 < r < 0.988)\), mean pressure was not well correlated \((r = 0.214, \text{CI}_{95} = -0.723 < r < 0.874)\). Mean resistance calculated by the ratio of mean pressure and mean flow decreased with patient weight and was highly correlated as shown in Fig. 3.

Mean harmonics of pressure, flow, and impedance, along with standard deviations, are provided in Table 2. The impedance values listed are limited to the harmonics satisfying the accuracy criterion (that standard deviation of either real or imaginary parts of the impedance be no greater than 20% of the zeroth harmonic modulus). The accuracy criterion corresponded in all but one case with a modulus of flow \(<0.02\) l/min or a modulus of pressure \(<0.15\) mmHg. (The exception was patient 6, for whom the normalized standard deviations of the real and imaginary parts were 20.46 and 26.05, respectively, for a combination of a relatively low-pressure modulus of 0.21 mmHg and a relatively low-flow modulus of 0.028 l/min.) For patient 1, the number of harmonics was arbitrarily limited to 10, although the accuracy criterion was not exceeded. Sampling frequency may also limit the resolution of the Fourier transforms; however, in these measurements, the sampling frequency (200 Hz) greatly exceeded the frequency of the highest harmonic considered (25.4 Hz for the 10th harmonic for patient 1).

Normalized impedance modulus and phase for all patients are shown in Fig. 4. Two types of responses were found. Type A had low inerance with constant high-frequency asymptote of impedance modulus and a positive, but relatively small, impedance phase at high frequency. Type B had high inerance with impedance modulus that increased at high frequency and high-impedance phase at high frequency. The type A response, which was also facilitated by high \(R_c\), occurred in the two youngest patients (patients 1 and 3), and the type B response, with lower \(R_c\), occurred for the older patients (patients 2, 4, 5, and 6). A comparison of the fits of the lumped-parameter models, using an example of each type of response, is shown in Fig. 5.

The total vascular resistances \((Z_0)\) of the models were specified by the fitting procedures to be equal to the average impedance moduli at \(Z_0\) (Table 2) for each patient. The \(Z_0\) values, which resulted from the Fourier analysis, thus were nearly identical to the average resistances calculated from the average pressures and flows (Fig. 3). \(Z_0\) therefore, also decreased with patient weight and was highly correlated \((r = 0.988,\)
Although $R_c$ in the models showed a general decrease with increasing patient weight, the correlations were not strong (Fig. 6). $R_p$, however, decreased with patient weight and was well correlated (Fig. 7). $C$ increased with patient weight for all models (Fig. 8). $L$ decreased with patient weight for the RLRC2 model, but no clear trend was evident for the RLRC model (Fig. 9).
DISCUSSION

Considerable variation among patients in pressure and flow was evident in the waveforms (example in Fig. 2) and in their averages (Table 1). Because the patients were selected on the basis of absence of left ventricular and aortic abnormalities, the variability is not likely due to abnormal cardiac or vascular performance. That all patients had normal left ventricular performance appears to be supported by the pressures and flows being in an acceptable range and by the reasonable correlation of aortic flow with patient weight. The variability in pressure may be due in part to subject variability, including response to anesthesia. The anesthetic induction and maintenance agents used with these patients (fentanyl, pancuronium bromide, isoflurane, and nitrous oxide) are known to cause varying degrees of myocardial depression and arterial vasodilation, which affect the stroke volume and vascular resistance. The patient-to-patient variability in the amount of agent needed to achieve the desired anesthetic plane and the individual cardiovascular response to the agents, coupled with the normal range of cardiovascular regulation, may have contributed to the subject-to-subject variability found in this patient population.

The correspondence of high standard deviation (>20%) of impedance to low pressure and flow moduli approaching the accuracy of the respective measurements suggests that the quality of the impedance averages for increasing frequency was limited at this level by the accuracy of the pressure and flow measurements rather than by inherent variations in the input impedance. Note that considerable fluctuations in impedance (exceeding in some cases twice the value of the individual modulus) nonetheless exist in the harmonics that satisfy the accuracy criterion. These fluctuations may be representative of primary changes in vascular response (due, for instance, to varying vasoconstriction) or of changes in vascular response due to interacting secondary functions (for instance, respiration) but may also derive in part from inherent nonlinearities in vascular response.

For compliant vessels, greater pressure increases cross-sectional area for flow, resulting in lower impedance to flow. The influence of this effect was investigated by testing the correlation of $Z_0$ with zeroth harmonic pressure for each of the cycles. Decreases in $Z_0$ with increasing pressure were found in all patients, although correlations, other than for patient 1, were not strong ($r = -0.153, Cl_{95} = -0.714 < r < -0.528$; $r = -0.954$ and $Cl_{95} = -0.995 < r < -0.630$; $r = -0.975$ and $Cl_{95} = -0.997 < r < -0.785$; $r = -0.953$ and $Cl_{95} = -0.995 < r < -0.627$; $r = -0.988$ and $Cl_{95} = -0.999 < r < -0.890$).

Fig. 7. $R_p$ for each model. Correlations with patient weight are as follows: $RCR, r = -0.954$ and $Cl_{95} = -0.995 < r < -0.630$; $RCR2, r = -0.975$ and $Cl_{95} = -0.997 < r < -0.785$; $RLRC, r = -0.953$ and $Cl_{95} = -0.995 < r < -0.627$; $RLRC2, r = -0.988$ and $Cl_{95} = -0.999 < r < -0.890$.

Fig. 8. Arterial $C$ for each model. Correlations with patient weight are as follows: $RCR, r = 0.907$ and $Cl_{95} = 0.363 < r < 0.990$; $RLRC, r = 0.852$ and $Cl_{95} = 0.131 < r < 0.984$; $RLRC2, r = 0.952$ and $Cl_{95} = 0.618 < r < 0.995$. Statistics for $RCR$ model are not given because of inaccuracy of $C$ values. For reference, arterial $C$ calculated by area (Ref. 4) and pulse pressure methods are also shown: $RC$, area, incisura to end diastole; $RC2$, area, peak after incisura to end diastole; $RC3$, pulse pressure. Correlations with patient weight are as follows: $RC, r = 0.748$ and $Cl_{95} = -0.161 < r < 0.971$; $RC2, r = 0.910$ and $Cl_{95} = 0.377 < r < 0.990$; $RC3, r = 0.961$ and $Cl_{95} = 0.662 < r < 0.996$. 

Downloaded from http://jap.physiology.org/ by 10.220.33.3 on April 14, 2017
patient 3: \( r = -0.392 \), \( CI_{95} = -0.819 < r < 0.315 \); patient 4: \( r = -0.561 \), \( CI_{95} = -0.880 < r < 0.107 \); patient 5: \( r = -0.312 \), \( CI_{95} = -0.787 < r < 0.395 \); patient 6: \( r = -0.365 \), \( CI_{95} = -0.809 < r < 0.343 \). In the most highly correlated subject (patient 1, Fig. 10), a maximum zeroth harmonic pressure variation among cycles of \( \sim 10\% \) corresponded to a maximum \( Z_0 \) variation of \( \sim 20\% \). Correlations between higher harmonics of impedance and pressure were not evident. These small differences do not explain the much larger fluctuations observed among cycles in the higher harmonics of pressure, flow, and impedance but suggest that linearity of response should not be assumed in cases in which pressure varies more than in the present measurements (\( \sim 10\% \)).

The excellent correlation of mean resistance with patient weight (Fig. 3) appears to indicate that this measure of systemic vascular response was normal despite the effects of anesthesia and other influences and, furthermore, that patient weight was a good scaling parameter (correlation with patient age was slightly lower: \( r = -0.928 \), \( CI_{95} = -0.993 < r < -0.471 \), as it was for most results). The average resistance of the heaviest patient was only slightly higher than that of adults, despite the child’s weight being only about one-quarter that of adults. On the other hand, the resistance of the lightest patient was approximately three times that of adults. One might expect that conductance (inverse of resistance) per unit weight may be constant, and, indeed, this parameter varied only from \( 3.2 \times 10^{-5} \) to \( 4.0 \times 10^{-5} \) \( cm^5 \cdot dyn^{-1} \cdot s^{-1} \cdot kg^{-1} \) in the patients but was \( \sim 1.1 \times 10^{-5} \) \( cm^5 \cdot dyn^{-1} \cdot s^{-1} \cdot kg^{-1} \) in adults. Conductance per unit weight was only weakly correlated with weight among the patients \( (r = 0.222, CI_{95} = -0.719 < r < 0.876, \) and \( P = 0.493 \) between infants and children (ANOVA)) but was significantly different in infants and children compared with adults \( (P < 0.001) \).

Decreases in \( R_c \) (Fig. 6) and \( R_p \) (Fig. 7) with patient weight were also seen. Because of the wide range of \( R_c \) in the patients, the difference in \( R_c \) between infants and children vs. adults was not significant \( (RCR: P = 0.379, RLRC: P = 0.315, RLRC: P = 0.378, RLRC: P = 0.196) \), but the difference between infants and children was significant except for the RLRC2 model \( (RCR: P = 0.015, RLRC: P = 0.059, RLRC: P = 0.018, RLRC: P = 0.676) \). In the heaviest patient, \( R_p \) was only \( \sim 10\% \) higher than that of adults, whereas \( R_c \) was still approximately three times higher.

The ratio of \( R_c \) to total vascular resistance also dropped with increasing weight (except for the RLRC2 model, which exhibited an \( R_c \) value for best fit that was influenced by its parallel inductance). This trend is consistent with the obvious growth of the arteries from childhood to adulthood; however, because \( R_c \) represented a small fraction of the total vascular resistance, its development had little impact on \( Z_p \). The normalized \( R_c \) values \( (0.058–0.212) \) were higher than those for adults \( [-0.028–0.056 \) from data by Nichols et al. \( (8) \)], but, because of the wide range, the difference was not statistically significant among these groups except for the RLRC2 model \( (RCR: P = 0.323, RLRC: P = 0.222, RLRC: P = 0.321, RLRC: P = 0.083) \). The difference between infants and children, however, was significant \( (RCR: P = 0.006, RLRC: P = 0.072, RLRC: P = 0.006, RLRC: P < 0.090) \). The difference in the \( R_{ HC } \)-to-\( R_c \) ratio (Fig. 11) was significant for comparisons of both infants and children vs. adults \( (RCR: P = 0.049, RLRC: P = 0.028, RLRC: P = 0.046, RLRC: P < 0.001) \) and for infants vs. children \( (RCR: P = 0.018, RLRC: P = 0.093, RLRC: P = 0.017, RLRC: P = 0.022) \).

The reduction of \( R_p \) to a level near that of adults suggests that the peripheral circulation develops rapidly early in life, whereas the remaining difference between child and adult \( R_c \) suggests that development of the proximal arteries may follow the slower geometric growth of the body itself. The difference in the development rates is supported by the above comparisons of infant, child, and adult \( R_{ HC } \)-to-\( R_c \) ratio. The reasons for this difference in development rates remain
to be investigated. One attractive hypothesis is that $R_p$,
as well as other parameters, may decrease rapidly to
reduce hydraulic power requirements. However, in this
data set, power increased with patient weight ($r = 0.771$, CI95 = $-0.319 < r < 0.959$; RCR2, $r = 0.582$ and CI95 = $-0.436 < r < 0.946$; RLRC, $r = 0.642$ and CI95 = $-0.355 < r < 0.956$; RLRC2, $r = -0.425$ and CI95 = $-0.919 < r < 0.590$).

The decrease in the $R_c$-to-$Z_0$ ratio with increasing
weight is also evident in Fig. 4; however, other distin-
guishing features in the impedance plots also stand
out. With the exception of patients 1 and 3, the patient
data exhibited increasing modulus and phase with
increasing frequency at high frequency. The phase
increased to 60° or more for patients 2, 4, 5, and 6. This
behavior, which is characteristic of inertial flow, is
much stronger than for adults. Higher inertance, which
scales with $\rho/A$, where $\rho$ is blood density, $l$ is vessel
length, and $A$ is cross-sectional area, might be expected
in infants and/or children if $A$ scaled with $l^2$. Indeed, the
modeled inertance values were approximately an order
of magnitude larger than those for adults for the RLRC
model (Table 3), although the wide range of the infant/
child values made this difference statistically insignifi-
cant ($P = 0.340$). Although no trend with increasing
weight was evident for $L$ for the RLRC model, $L$ in the
RLRC2 model decreased with weight (Fig. 9).

On the other hand, patients 1 and 3 exhibited no
significant trend in modulus for high frequency, and
phase remained below 35° for patient 1 and lower than
15° for patient 3. Patient 3 was the youngest and
lightest subject, whereas patient 1 was the second
youngest and third lightest; however, the connection
between age or weight and the less inertial impedance
behavior of these patients is unclear.

The RLRC model produced the best fits of the imped-
ance data for all patients. Furthermore, this model also
produced the lowest values of the Akaike information
criterion (AIC) and the Schwarz criterion (SC; see Ref.
17), two indexes that account in different ways for the
expectation that models with more parameters should
provide better fits (Table 3). A survey of the curve fits
for each patient (examples in Fig. 5) reveals the follow-
ing characteristics.

First, the RCR model was severely limited in its
ability to match infant/child impedance because its
phase cannot be positive. The curve fits in this work
were based on errors between modeled and actual
impedance at all harmonics satisfying the accuracy
criterion, which included high-frequency harmonics.

Table 3. Lumped parameter element value results

<table>
<thead>
<tr>
<th></th>
<th>Patient 1</th>
<th>Patient 2</th>
<th>Patient 3</th>
<th>Patient 4</th>
<th>Patient 5</th>
<th>Patient 6</th>
<th>Adult</th>
</tr>
</thead>
<tbody>
<tr>
<td>$Z_0$ (specified)</td>
<td>3.126</td>
<td>3.594</td>
<td>3.813</td>
<td>1.889</td>
<td>2.341</td>
<td>1.539</td>
<td>1.300</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Curvefit results</th>
</tr>
</thead>
<tbody>
<tr>
<td>$R_c$, dyn·s·cm$^{-5}$</td>
</tr>
<tr>
<td>$C$, cm$^2$/dyn</td>
</tr>
<tr>
<td>AIC/SC</td>
</tr>
<tr>
<td>$R_c$, dyn·s·cm$^{-5}$</td>
</tr>
<tr>
<td>$C$, cm$^2$/dyn</td>
</tr>
<tr>
<td>AIC/SC</td>
</tr>
<tr>
<td>$R_c$, dyn·s·cm$^{-5}$</td>
</tr>
<tr>
<td>$C$, cm$^2$/dyn</td>
</tr>
<tr>
<td>AIC/SC</td>
</tr>
<tr>
<td>$R_c$, dyn·s·cm$^{-5}$</td>
</tr>
<tr>
<td>$C$, cm$^2$/dyn</td>
</tr>
<tr>
<td>AIC/SC</td>
</tr>
</tbody>
</table>

Series of letters represent resistance ($R$), compliance ($C$), and iner-tance ($L$) parameters employed in each model; 2 designates alternative
model using same parameters. For all subjects, RLRC model produced the lowest Akaike information criterion (AIC) and Schwartz criterion
(SC). For all models except RLRC2, resistance was specified as characteristic aortic resistance ($R_p$) + peripheral resistance ($R_p^*$) = zero
frequency ($Z_0$). For RLRC2, $R_p = Z_0$ was specified.
These curve fits produced unrealistic values of C because increasing compliance produced better (though still poor) fits of the higher harmonics of impedance as the modeled impedance phase approached zero. The best fits in several cases would actually be obtained with infinite C; however, the tabulated C values (Table 3) are those for which further increases in C produced insignificant improvements in NRMS. Better estimates of C were obtained with the RCR model by using fits based on the zeroth and first harmonics only (designated RCR2 in Fig. 14). Yet another compliance estimation method for the RCR model is based on a fit of the zeroth harmonic, an estimate of Rp from an average of the high-frequency harmonics of the impedance modulus and then a calculation of C based on an approximate fit of the first harmonic impedance modulus (18a). This method, however, depends on the first harmonic modulus being larger than the estimated Rp, which was not the case for the infants and children. The method resulted in imaginary values for C.

The area method of Liu et al. (4), another compliance estimation method based on the decay of pressure during diastole according to an RC model of cardiac afterload, produced greatly different C values depending on the interval of diastole chosen. C estimated by the area method is given by

$$C = \frac{\int_1^2 P \, dt}{Z_0(P_1 - P_2)} \quad (2)$$

where \(\int_1^2 P \, dt\) is the integrated pressure from time 1 to time 2, and \(P_1\) and \(P_2\) are the beginning and ending pressures. If the interval from the incisura to the end of diastole was used, the small pressure difference in the denominator of Eq. 2 produced large C values. If the interval from the peak in pressure after the incisura to end diastole was used, the much larger pressure difference produced much smaller C values. These two compliance values (for models RC and RC2 in Fig. 8) essentially bracketed the compliance resulting from the curve fits of the lumped-parameter models. C estimated by the pulse pressure method (for model RC3 in Eq. 8) was comparable to the higher area method results. C correlated strongly with patient weight (Fig. 8) and was significantly different in infants and children relative to adults (RCR2: \(P = 0.024\), RLRC: \(P < 0.001\), RLRC2: \(P < 0.001\), pressure wave forms were not available to calculate RC, RC2, and RC3 compliances, and statistics for the RCR model are not given because of the inaccuracy of the C values for this model). C in infants and children was approximately 10 times lower than that in adults. C was not significantly different between infants and children except for the RC3 model (RC: \(P = 0.383\), RC2: \(P = 0.203\), RC3: \(P = 0.053\), RCR2: \(P = 0.218\), RLRC: \(P = 0.506\), RLRC2: \(P = 0.188\)).

Second, for the patients with strong inertial character, the RLRC2 model produced impedance modulus curves with sharp minima, which adversely influenced the fits. In addition, its transition from negative phase to positive phase was abrupt and was followed by decreasing phase toward an asymptote of 0° (see Fig. 5B). This model did not appear to fit the character of these patients. For patients 1 and 3, the mismatch of the shape of the modeled impedance to the patient data was less dramatic but still noticeable.

The poor performance of the RLRC2 model can be explained by examining its impedance equation

$$Z = \frac{1}{R_c} + \frac{1}{j\omega L} + \frac{1}{(R_p - j\omega C)} \quad (3)$$

which can be nondimensionalized as

$$Z^* = \frac{n^2L^2/R_c^*}{1 + n^2L^2/R_c^*} + \frac{n\omega L}{1 + n^2\omega^2L^2/R_c^*} \quad (4)$$

where \(Z^* = Z/Z_0\), \(n = \omega/\omega_1\), where \(\omega_1\) is angular frequency and \(\omega_1\) is first harmonic angular frequency, \(L^* = \omega_1L/Z_0\), \(R^* = R/Z_0\), \(C^* = \omega_1R_cC\), and \(Z_0 = R_c\) for this circuit. The imaginary part of the impedance disappears (the phase crosses through zero) for small \(nL^*/R_c^*\) the ratio of aortic flow dissipation time to cycle period, indicating the relative impedance of the parallel L and Rc flow pathways) and large \(nC^*\) (\(C^*\) is the ratio of pressure dissipation time to cycle period, indicating the relative impedance of the parallel Rp and C flow pathways) when the inertance and compliance resonate for a normalized angular frequency of approximately \(n_X = (L^*C^*)^{-1/2}\). The crossover corresponds to the first imaginary term X1 in Eq. 4, canceling the second imaginary term X2. In the patients with inertial character, the normalized crossover frequency was \(~1.4–1.8\) which was close to the resonant frequency. At the crossover frequency, \(nL^*/R_c^*\) was small, allowing the first real term R1 to be small. In addition, \(nC^*\) was large, allowing the second real term R2 also to be small. The impedance, therefore, exhibited a precipitous drop near the crossover frequency (Fig. 12).

The physical interpretation of this modeled behavior is that the flow substantially bypassed the resistance of the proximal arteries and found a sufficiently large compliance downstream such that it offered little back pressure. For the patients with inertial behavior studied here, this behavior appears unrealistic. For patients 1 and 3, \(nC^*\) was large, but \(nL^*/R_c^*\) was not small at the crossover frequency, resulting in a sharp minimum in modulus.

For the adults represented by Nichols et al. (8), \(nC^*\) was large, but \(nL^*/R_c^*\) was not small at the crossover frequency; thus a shallow minimum results. In Eq. 4, both R2 and X2 drop sharply with increasing frequency. When the falling X2 matches the magnitude of the rising X1, these two terms cancel at the crossover frequency \(n_X\). Similarly, R1 rises to match the magnitude of the decreasing R2 at a normalized frequency of \(n_R = R_c^* n_X \times (L^*C^*)^{-1/2}\) for small \(nL^*/R_c^*\) and large \(nC^*\). When these frequencies are close to one another, which
occurred for the range of patient values of $R_c$ of 0.2–0.5, the potential for a sharp minimum in modulus exists. (Note in Fig. 12 that the $R_1$-$R_2$ match occurs near the $X_1$-$X_2$ crossover.)

This sharp minimum is prevented if the magnitude of $R_1$ rises above that of $X_1$ before the crossover frequency. This matching frequency, called the bypass frequency, is given by

$$n_b = \frac{R_c}{L}$$

and corresponds to the frequency at which the impedance of the inertance begins to exceed that of the $R_c$. Thus the ratio of the bypass and crossover frequencies controls the sharpness of the modulus minimum. If $n^* = n_b/n_x$ is greater than one, then a sharp minimum will occur. For patient 2, shown in Fig. 12, $n^* = 6.71$, the bypass frequency occurred beyond the maximum frequency on the graph, and a sharp minimum resulted. The RLRC2 model provided reasonable fits for patients 1 and 3 and for the adult, for which $n^* = 1.68$, 1.07, and 0.65, respectively, but excessively sharp minima for the other patients with larger $n^*$ ($n^* = 6.7, 7.8, 5.2$, and 7.8 for patients 2, 4, 5, and 6, respectively).

This sharp minimum, which depends on a delicate balance of inertial and compliant impedance, does not occur with the RLRC model, which produced the best quantitative fits and also provided qualitative shapes better matched to the patient data. Its advantageous features for the patient data included a more gradually and monotonically increasing phase with increasing frequency at high frequency and a softer minimum in modulus. The impedance of the RLRC model is given by

$$Z = R_c + j\omega L + \left(\frac{1}{R_p} + j\omega C\right)^{-1}$$

which may be nondimensionalized as

$$Z^* = R^*_c + \frac{R^*_p}{1 + n^2C^2} + j\left(nL^* - \frac{nC*R^*_p}{1 + n^2C^2}\right)$$

where the dimensionless parameters are the same as for the RLRC2 model, except $R^*_p = R_p/Z_0$ and $Z_0 = R_c + R_p$ for this circuit. For the RLRC model, the minimum modulus is defined by $R^*_c$ where the imaginary terms cancel and where the second real term in Eq. 6 is small. (Note in Fig. 12 that $R_1 = R^*_c$ provides a floor for the modulus.) In both the infant/child and adult data, the second real term was only a few percent of $R^*_c$ at the crossover frequency, which was again near $(LC)^{-1/2}$.

The physical interpretation is that the aortic compliance is large enough to accommodate the flow through the $R_c$, thereby decreasing impedance to near its minimum, before the frequency becomes high enough for inertance to influence the impedance. In this model, however, impedance is limited to a minimum of $R^*_c$ by the series resistance. Inertance serves only to increase impedance above the value of $R^*_c$ with increasing frequency. This concept is consistent with flow through relatively stiff conduits and branches, in which fluid inertia may grow to dominate viscous dissipation in the flow behavior, but viscous dissipation does not disappear and is never bypassed.

The RLRC model performed better than the RCR and RLRC2 models for all infant, child, and adult data in terms of lower NRMS, AIC, and SC (Table 3). The RLRC2 model produced lower NRMS values than the RCR model except for the adult data and produced lower AIC and SC except for the adult data and patient 3. Stergiopulos et al. (17) advocated the RLRC2 model over the RLRC and RCR models for fitting the response of dogs and adult humans; however, the present results are, to our knowledge, the first quantitative comparison of the fits of the two four-element models to human data. Yoshigi and Kell (20) compared the performance of 18 models with two to five elements (including RCR, RLRC, and RLRC2) in matching the impedance of chick embryos and found that the RLRC model and a three-element model resulting from the elimination of the proximal resistor from the RLRC model produced the best fits.

Westerhof et al. (18a) determined that, among many adult mammals, the ratio of $R_c$ to $Z_0$ ($R_c/Z_0$) was a
constant. In the six patients examined in this project, 
\( R_c/Z_0 \) dropped with increasing weight for the RCR, 
RCR2, and RLRC models but increased for the RLRC2 
model, although none of the correlations were strong. 
The value of \( R_c/Z_0 \) of \( \approx 0.1 \) for the heaviest subjects 
compared with the Westerhof et al. constant value of 
0.055 and the value of 0.042 obtained from the Nichols 
et al. (8) data suggests that this ratio may change 
during the development of humans. Such development 
appears to include a rapid reduction in \( R_p \) and a slow 
reduction in \( R_c \), as discussed above.

Westerhof et al. (18a) also found that the ratio of 
pressure dissipation time (\( R_pC/\) to heart cycle period \( T \) 
(\( R_pC/T \)) was a constant, hypothesizing that a lower 
limit of this ratio was necessary from a cardiovascular 
function perspective to maintain diastolic pressure 
high enough to provide adequate coronary flow. \( C^* = \omega R_pC \) (the same as \( R_pC/T \) except for a factor of \( 2\pi \)) 
increased for all models, although correlations were not 
strong except for the RCR2 and RLRC2 models (Fig. 
13). The strong increase in \( C \) dominated the decrease in 
\( R_c \), as discussed above.

Scaling arguments were used to compare wave reflec-
tion effects in the pediatric patients to those in adults. 
Wave speed (\( W \)) may be estimated by

\[
W = \sqrt{\frac{V_a}{\rho C}} 
\]

where \( V_a \) is arterial volume. If \( \rho \) is assumed to be
constant with patient weight and \( V_a \) is assumed to scale directly with patient weight, then the 2–16 times smaller compliance in infants and children might combine with the 4–10 times smaller volume to produce little difference in wave speed. Indeed, when compliance per unit weight was calculated from the pediatric patients and compared with values for the average adult, no trend or strong correlation with weight was found for the RLRC model. With the assumption that artery length scales with the cube root of weight, wave reflection times estimated from the wave speeds and artery lengths were 1.6–2.2 times shorter in the infants and children than in the average adult. However, the heart cycle periods in the patients were also 1.5–2.3 times shorter. Thus no significant differences in wave reflection effects between infants and children vs. adults were found. These estimates appear to support the speculation that wave reflection time is regulated to minimize cardiac work by avoiding reflection of high-pressure pulses during the same systolic period (6).

In summary, these results provide the first calculations of aortic input impedance in pediatric patients. For four of the six patients, the results exhibited strong inertial character compared with adults, with increasing modulus and large positive phase with increasing frequency at high frequency. The RLRC lumped-parameter model reproduced the character of the impedance data best. For this model, compliance and \( R_p/R_c \) differed greatly from those of adults. Furthermore, \( R_c \) and \( R_p/R_c \) varied considerably between infants and children, and \( R_p \) and \( Z_0 \) were well correlated with patient weight. These comparisons demonstrate that the overall character of the circulation is different in infants and children compared with adults and suggest that significant changes are in progress during development from infant to child.

We thank Transonic Systems (Ithaca, NY) for providing the ultrasonic flow probes and meter for this project.

Address for reprint requests and other correspondence: M. K. Sharp, Biofluid Mechanics Laboratory, Dept. of Bioengineering, Univ. of Utah, 50 S. Central Campus Dr., Rm. 2480, Salt Lake City, UT 84112 (E-mail: m.k.sharp@m.cc.utah.edu).

Received 15 March 1999; accepted in final form 4 February 2000.

REFERENCES

18a. Westerhof N, Elzinga G, and Sipkema P. The measure-