Computation of aortic flow from pressure in humans using a nonlinear, three-element model

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Netherlands Organization for Applied Scientific Research, TNO Biomedical Instrumentation, Academic Medical Centre, 1105 AZ Amsterdam; Department of Pulmonary Diseases, Erasmus University, 3000 DR Rotterdam, and Department of Anesthesiology, Academic Hospital, University of Limburg, 6202 AZ Maastricht, The Netherlands

WESSELING, K.H., J.R.C. JANSEN, J.J. SETTELS, AND J.J. SCHREUDER. Computation of aortic flow from pressure in humans using a nonlinear, three element model. J. Appl. Physiol. 74(5): 2566-2573, 1993.—We computed aortic flow pulsations from arterial pressure by simulating a nonlinear, time-varying three-element model of aortic input impedance. The model elements represent aortic characteristic impedance, arterial compliance, and systemic vascular resistance. Parameter values for the first two elements were computed from a published, age-dependent, aortic pressure-area relationship (G. J. Langewouters et al. J. Biomech. 17: 425-435, 1984). Peripheral resistance was predicted from mean pressure and model mean flow. Model flow pulsations from aortic pressure showed the visual aspects of an aortic flow curve. For evaluation we compared model mean flow from radial arterial pressure with thermodilution cardiac output estimations, 76 times, in eight open heart surgical patients. The pooled mean difference was +7%, the SD 22%. After using one comparison per patient to calibrate the model, however, we followed quantitative changes in cardiac output that occurred either during changes in the state of the patient or subsequent to vasoactive drugs. The mean deviation from thermodilution cardiac output was +2%, the SD 8%. Given these small errors the method could monitor cardiac output continuously.

CARDIAC OUTPUT is an important hemodynamic variable. It is studied in physiological experiments often in relation with changes in pressure. It is monitored in surgical and critical care patients. There is a trend in physiological research, as well as in the operating room and intensive care unit, to require essentially continuous cardiac output monitoring.

Pressure pulse analysis has been used for beat-to-beat cardiac output monitoring (9, 15). Traditionally, such methods computed cardiac stroke output from certain characteristics of an arterial pressure pulse, based on a variety of simple models of the arterial system. Characteristics used included systolic area, heart rate, and mean and diastolic pressure. In these methods, aortic properties were assumed constant under varying distending pressures. They were not constant, however, and models were not sufficiently detailed. Consequently, pressure pulse analysis often proved to be of insufficient precision (6).

More recently, several studies have shown the suitability of a three-element model of arterial input impedance to describe the relationship between aortic pressure and flow (2, 16). Once model parameters are found, flow can be computed from measured pressure by simulating the model (2). This flow then provides a continuous measure of cardiac output. Integrated over one heartbeat it provides stroke volume; integrated over one minute it provides cardiac minute volume.

We investigated whether this model could monitor cardiac output continuously. We were fortunate in that a precise thermodilution technique is now available (4), and we anticipated that this improved technique would permit a more reliable evaluation.

METHODS
Computing Flow From Pressure
During the development we made three decisions. 1) We would compute flow by simulating the response of a three-element model of arterial input impedance to arterial pressure (Fig. 1). To do this, the parameter values of the model elements must be known. Two of the model parameters, characteristic impedance and arterial compliance, can be derived from an aortic pressure-area relationship. 2) We would use the arctangent model of aortic mechanics of Langewouters et al. (7) (Fig. 2). The use of this arctangent relationship causes the model elements to be nonlinearly dependent on pressure. The third element, total systemic peripheral resistance, is a variable. Its value is not known but is an outcome of the model.
FIG. 1. Diagram of 3-element model used in this study to compute flow. Z₀, characteristic impedance of proximal aorta; Cₘ, windkessel compliance of arterial system; Rₚ, total systemic peripheral resistance. Z₀ and Cₘ have nonlinear, pressure-dependent properties, indicated by stylized S symbol. Rₚ varies with time, as symbolized by arrow. Q(t), blood flow as function of time; P(t), arterial pressure waveform; Pₘ(t), windkessel pressure.

Simulation. 3) We would use the current peripheral resistance value as a best estimate of that parameter in the simulation of the next heat.

The model. To compute aortic flow from pressure we base the pressure-flow relationship on the traditional (1, 20) but adequate (16) model shown in Fig. 1. This model has three elements representing the three major properties of the aorta and arterial system: aortic characteristic impedance (Z₀), a dynamic property of the aorta that impedes pulsatile outflow from the ventricle; windkessel or buffer compliance (Cₘ), the ability of the aorta and arterial system to elastically store the cardiac stroke output from the left ventricle; and peripheral resistance (Rₚ), the Poiseuille resistance of all vascular beds together. The value of total peripheral resistance in the model is the sum of Z₀ and Rₚ. The special symbols in Fig. 1 indicate the nonlinear, pressure-dependent properties of Z₀ and Cₘ and the time varying property of Rₚ.

During systole, flow is into the model; during diastole this inflow is dissipated in the periphery. Given input pressure, aortic systolic inflow is principally determined by the time constant Z₀Cₘ. If this time constant is too short, computed stroke volume will be too small. The peripheral resistance element value is not a major determinant of systolic inflow. Diastolic outflow from the windkessel into the peripheral resistance and windkessel pressure decay are principally determined by the time constant RₚCₘ. If input pressure decay equals model windkessel pressure decay, the diastolic inflow is zero. Windkessel compliance is a common factor in both time constants.

Model parameter values. To carry out the computations numerically, proper values for the model parameters Z₀, Cₘ, and Rₚ must be found. We do have precise and detailed results on the viscoelasticity of the human aorta in the form of pressure-area relations (7), and Z₀ and Cₘ can be computed from it.

The aortic characteristic impedance Z₀ can be expressed as

\[ Z₀ = \sqrt{\rho/(AC')} \]  

where \( \rho \) is the density of blood, \( A \) is the cross-sectional area of the aorta, and \( C' \) is the aortic compliance per unit length. C' is the derivative of the pressure-area relationship with respect to pressure (P)

\[ C' = \frac{dA}{dP} \]  

Windkessel compliance, \( Cₘ \), represents the lumped compliance of the entire arterial system. We assume its value to be equal to the compliance of one unit length of thoracic aorta times an aortic effective length, \( l \)

\[ Cₘ = lC' \]  

For an adult patient we assumed for \( l \) a value of 80 cm (19). More strictly, \( l \) could depend on patient height and weight.

FIG. 2. A: arctangent pressure-area relationship. It fits measured pressure-area curves of human aorta well (7) and is characterized by 3 parameters: maximal area (\( A_{max} \) ), pressure at inflection point (\( P_i \) ), and width parameter (\( P_f \)). Measurements of the 3 parameters are plotted in B-D, together with their regressions on age (A), as listed in Table 1. Open circles, female (F) aortas, filled circles, male (M) aortas. Parameters \( P_f \) and \( P_i \) show only small scatter around regression, but scatter for \( A_{max} \) is substantial. [Adapted from data in Langewouters et al. (7).]
TABLE 1. Summary of regression equations for arctangent aortic model parameters vs. patient age and gender (7)

<table>
<thead>
<tr>
<th>Parameter</th>
<th>F</th>
<th>M</th>
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<tbody>
<tr>
<td>$A_{\text{max}}$, cm$^2$</td>
<td>4.12</td>
<td>5.62</td>
</tr>
<tr>
<td>$P_0$, mmHg</td>
<td>72 - 0.89A</td>
<td>76 - 0.89A</td>
</tr>
<tr>
<td>$P_1$, mmHg</td>
<td>57 - 0.44A</td>
<td>57 - 0.44A</td>
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F, female; M, male; $A_{\text{max}}$, maximal area; $P_0$, inflection pressure; $P_1$, width parameter; A, patient age (in yr). We used these data to obtain proper pressure-area relation for a patient, given gender and age.

We define $R_o$ as the ratio of average pressure to average flow. Its value changes only slowly (15) compared with a heartbeat interval. We therefore use its current computed value to simulate the flow of the next beat. For the first beat, at the start of the simulation, a reasonable initial value is assumed. We take the ratio of 100 mmHg mean pressure and 3 l/min cardiac output. From true mean pressure and computed mean flow the next approximation is computed, and so on. $R_o$ converges from the initial value to the correct value in a few heartbeats.

Pressure-area relationship. $Z_o$ and $C_w$ have been presented in terms of the aortic pressure-area relation and its derivative. We must now define that relation. According to Langewouters et al. (7), the thoracic aortic cross-sectional area can be described as a function of pressure by an arctangent with three parameters (see Fig. 2A)

$$A(P) = A_{\text{max}} \left[ 0.5 + \frac{1}{\pi} \arctan \left( \frac{P - P_0}{P_1} \right) \right] \quad (4)$$

$A_{\text{max}}$ is the maximal cross-sectional area at very high pressure. Parameter $P_0$ defines the position of the inflection point on the pressure axis (40 mmHg in Fig. 2); $P_1$ defines the width between the points at one-half and three-quarter amplitude (50 mmHg in Fig. 2). Parameters $P_0$ and $P_1$ compare with the mean and the SD of a probability function.

The compliance per unit length is the derivative of $A(P)$ with respect to $P$ (Eq. 2) and depends on $P$

$$C(P) = \frac{A_{\text{max}}/\pi P_1}{1 + \left( \frac{P - P_0}{P_1} \right)^2} \quad (5)$$

Given correct values for the three arctangent parameters $A_{\text{max}}, P_0$, and $P_1$, the model parameters $Z_o$ and $C_w$ can be computed for any pressure $P$.

Arctangent parameters. The three arctangent parameters have been determined for 45 human thoracic aortas in vitro (7) and are presented in graphic form in Fig. 2 (B–D). It appears that $P_0$ and $P_1$ regress tightly on patient age (A), given patient gender. $A_{\text{max}}$ does not regress on age, at least not over the observed age range. Its average value is indicated separately for females and males. However, an individual's value can deviate as much as 40% from the group average. The statistical information is in Table 1. Thus given patient gender and age, we can compute $A_{\text{max}}, P_0$, and $P_1$ using the data in Table 1 to find a proper pressure-area relationship for a subject, but with uncertainty in the maximal area. To improve accuracy in monitoring an individual patient, a proper value for maximal area must be obtained in another way. If no other information is available, the group average value is used as a best estimate. The error involved from this is as follows.

Importance of maximal area. Calculations show that computed model flow using the group average maximal area deviates by the same percentage amount from true flow as the group average maximal area deviates from a patient's true area. Given the 20% scatter SD in the maximal area, model flow cannot be computed with an accuracy better than this percentage. An accurate measurement of aortic diameter can yield a patient individual value for maximal area, improving accuracy. An indicator dilution estimate of cardiac output can also be used, as was proposed by Warner (17). The initial population average value of $A_{\text{max}}$ is then multiplied by $k$, an individual patient's ratio of thermodilution to model flow cardiac output.

Model flow computation. Model flow, finally, is computed by simulating the behavior of the model under the applied arterial pressure pulsation. The values for windkessel compliance and characteristic impedance, as a function of instantaneous pressure, are inserted in the model. Because these values depend on pressure as was shown above, the model behavior is nonlinear. Simulation is done digitally, and model computations are repeated for each new pressure sample taken. Left ventricular stroke volume is computed by integrating model flow during systole; cardiac output is computed by multiplying stroke volume with instantaneous heart rate. The systolic duration and the heartbeat interval are derived from the pressure waveform. We call this approach the model flow method.

Patients

We evaluated the model flow method using arterial pressure and thermodilution cardiac output recorded in the operating room. After obtaining informed consent, we studied eight male patients who suffered from multiple coronary vessel disease but had intact aortic valves. All patients were monitored in 1989, and their signals were digitized. Except for two patients whose data were accidentally destroyed, all patients studied in that year were analyzed. The patients underwent elective coronary artery bypass graft surgery. Lorazepam (5 mg) was administered as premedication 2 h before surgery. A peripheral venous catheter, a 20-gauge radial artery cannula with a Medisize continuous flush device, and a 7 Fr Swan-Ganz pulmonary artery catheter were inserted under local anesthesia. Before induction of anesthesia a series of baseline hemodynamic measurements was performed under spontaneous breathing conditions. Anesthesia was induced with an initial dose of 7.5 µg/kg of sufentanil and was maintained with a continuous infusion of sufentanil at a rate of 3.75 µg·kg$^{-1}$·h$^{-1}$. Pancuronium bromide (0.1 mg/kg) was given for muscle relaxation. The patients were ventilated at 10 cycles/min with an oxygen-air mixture having an inspired oxygen fraction of 50%.
Experimental Techniques

In all patients, we recorded pressure in the radial artery using a standard Hewlett-Packard model 78334A arterial pressure channel with a 15-Hz built-in cutoff frequency. In two patients, we could additionally record ascending aortic pressure using a Millar microtip pressure transducer, but only during series e (see Measurement Protocol), just before going on bypass. A Baxter COM-2 device computed thermodilution cardiac output. A Siemens model 900B servoventilator was used for mechanical ventilation. It also provided the time pulses to trigger injections of 5 ml of room temperature 5% glucose with a computer-controlled power injector. An interface box passed all monitor and control lines, providing electrical isolation. This interface comprises three buttons to flush, start, and, if necessary, break the automatic thermodilution injections. The interface was connected to an IBM-compatible personal computer of AT level controlling the hardware system. Because aortic pressure was only incidentally available in these patients but radial pressure was available throughout the operation, the latter was used as input to the model. An aortic flow pulsation signal for comparison with the computed model flow pulsation was also not available in these patients. Instead, average model flow was compared with thermodilution estimates of average cardiac output.

To improve the accuracy of the thermodilution estimates, we used the technique of phase-controlled injections, equally spread over the ventilatory cycle (4). Using bolus injections, we estimated cardiac output four times, each in a different phase of the ventilatory cycle, and averaged the results. Each injection was delayed from the start of a ventilatory period over either 0, 25, 50, or 75% of the duration of the period. We waited ~30 s between injections. We averaged model flow cardiac output over two ventilatory cycles beginning at each thermodilution injection. Each series of four provided one comparison between model flow and thermodilution cardiac output.

Measurement Protocol

At various moments during the operation, measurements were done after major changes in patient state. Surgery was suspended, and hemodynamics were usually stable during the measurement periods. We did series s during spontaneous breathing, before induction of anesthesia. Series a–k were taken with the patient main- tained on the ventilator. The code letters a–k correspond to the following 11 instants: a, 3 min after induction; b, immediately after sternotomy; c–e, at 15, 10, and 5 min before bypass, respectively; f–i, at 3, 8, 13, and 18 min after bypass, respectively; j, 3 min after sternal fixation; and k, after end of surgery. Because of surgical and anesthetic events and depending on the duration of the operation, not every series was always performed in all patients. Seven to 12 series of four thermodilution estimations were thus carried out. In one patient an extra series was done. The series a thermodilution cardiac output was always obtained and was used to calibrate model flow cardiac output. During series b, c, and e before bypass, and during series h, j, and k after bypass, sodium nitroprusside was infused in doses of between 0.5 and 1.5 μg·kg⁻¹·min⁻¹. This drug has a strong vasodilating effect. Other vasoactive drugs were administered as needed during the operation.

We performed an off-line sensitivity analysis of the model flow method by repeating the model flow computations on the digitized data base. Patient age and pressure zero level were artificially varied. The first estimate was computed with unmodified parameters. Computations were then repeated with patient age entered incorrectly at +10 and −10 yr off true age. Age appears in the values for the arctangent parameters P₀ and P₁. The computations were repeated two more times with the correct age but with a pressure offset of +10 and −10 mmHg. Insensitivity to the absolute level of pressure is important because the pressure transducer membrane is not always precisely at heart level. The mean difference of model flow with simultaneous thermodilution cardiac output and its SD and range were used to judge sensitivity.

Our study could have produced good results accidentally because circumstances were right or patients were stable. We decided, therefore, to use the earlier, so-called “corrected” characteristic impedance (cz) method (19) on the same data. This method has been evaluated before (5), and its behavior is known. It computes beat-to-beat stroke volume, Vозд, by integrating the area under the systolic portion of the arterial pressure pulse, dividing the area by the cz method’s characteristic impedance calibration factor, Z₁

\[
V₂ = \frac{1}{Z₁} \int_{T_e} P(t) - P_d dt
\]

A correction is then applied that accounts for changes in mean pressure (Pₘ) and heart rate (f). This results in corrected stroke volume estimates, V₁

\[
V₁ = \frac{1}{Z₁} \int_{T_e} P(t) - P_d dt + 0.66 + 0.005f - 0.01A(0.014Pₘ - 0.8)V₂
\]

where P₀ is arterial end-diastolic pressure, Tₑ is the ejection period, and A is the age of the patient. The method is calibrated for each individual patient similarly to the model flow method, but adjusting Z₁.

RESULTS

Figure 3 presents an example of measured aortic and radial arterial pressure and computed model flow pulsations from each pressure signal. Compared with aortic pressure, radial pressure appears delayed and distorted. Model flow computed from aortic pressure shows the typical characteristics of an aortic waveform: a steep upstroke, a slow downstroke, a steep end-systolic downstroke terminated in a sharp dicrotic wave, and a flat near-zero diastolic flow. The flow wave computed from the distorted radial pressure is also distorted. In particular, peak flow is exaggerated.

Table 2 lists patient age, height, and model flow calibration factor k, pressure, pulse rate, and thermodilution cardiac output ranges. k indicates by which factor the patient aortic cross-sectional area is different from the population average. For our eight patients, k averages 107 ± 22%. This is not significantly different from 100% (t test, 2-sided, at P = 0.05). The 22% SD is not signifi-
AORTIC FLOW COMPUTED FROM PRESSURE

**FIG. 3.** A: simultaneous aortic (solid line) and radial pressures (dashed line). B: simulated model flow from aortic (solid line) and radial pressures (dashed line). Model flow computed from aortic pressure shows typical characteristics of an aortic flow pulsation: steep upslope at beginning of systole, a gradual down slope terminated in a steep final phase, and sharp dicrotic notch at end systole, followed by a period of almost zero flow in diastole. Model flow computed from radial pressure is delayed and distorted and shows typical characteristics less convincingly.

<table>
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<th>TABLE 2. Summary of patient information</th>
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<td>15</td>
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| k, model flow calibration factor (dimensionless); Pₐ, mean pressure; f, heart rate; Qₐ₉₅, quadruple thermodilution cardiac output. Deviations of k values from 1 are indication of accuracy of model flow method before calibration. Averaged over group of 8 patients, k is 1.07 ± 0.24 (SD). All patients were male.

Significantly different (F test, at P = 0.05) from the 17% SD found earlier in the Aₘₐₓ value for male patients (7). Pressure, heart rate, and cardiac output varied approximately over a 5:8 range during the operations.

The results of the 76 comparisons with thermodilution cardiac output are summarized in Table 3. We used determination a in each patient to compute the calibration factor k. The remaining 68 comparisons were taken into the postcalibration error statistics. Mean error ranged from -0.1 to 0.4 l/min. Its average of 0.1 l/min (2%) is not significantly different from zero. The SD ranged from 6 to 11% with an average of 8%. By comparison, the offset of the CZ method ranged from -0.6 to 0.8 l/min with a pooled average of 0.3 l/min. Scatter for the CZ method ranged from 7 to 21% with a pooled average of 12%.

A scatter diagram of model flow vs. thermodilution cardiac output is shown in Fig. 4. It presents the individual data taken in the eight patients summarized in Table 3. We obtained from 7 to 13 quadruple measurements in a patient. After using measurement a for calibration, at least 6 and at most 12 measurements per patient remained, representing major changes in patient state (see Measurement Protocol). Figure 5 is the errorgram of the model flow method showing the difference between the model flow and thermodilution estimates vs. their paired means. All differences except one are within ±1 l/min. Except for five outliers all differences stay within -0.50 and +0.65 l/min. The distribution of the 68 errors is also shown.

Pooled results of the sensitivity analysis for age and pressure are given in Table 4. Overall, errors do not differ greatly under the applied changes in the age parameter and in the offset pressure. Judging from the error minima and maxima per condition, neither do they differ greatly in individual patients.

**DISCUSSION**

The model flow method uses a three-element model of aortic input impedance to compute a flow pulsation from an arterial pressure pulsation (2). A new feature in this model is that two parameter values are computed (pre-
AORTIC FLOW COMPUTED FROM PRESSURE

Adequacy of Three-Element Model

The three-element model was proposed in 1930 by Broemser and Ranke (1, 20). Burkhoff et al. (2) state that the three-element model provides a reasonable representation of left ventricular afterload for predicting stroke volume but significantly underestimates peak aortic flow in dogs. McDonald and Nichols (10), using a pressure gradient method, similarly found that peak flow is more difficult to compute. We could not do a peak flow comparison because pulsatile flow was not measured in these patients. For most applications, however, possibly incorrect peak flows would have little consequence as long as beat-to-beat stroke volume and cardiac output are correct.

Sensitivity Analysis

The arctangent parameters \( P_0 \) and \( P_1 \) regress tightly on age (see Fig. 2) but not without scatter. This scatter can be interpreted as some patients having apparently older aortas for their age and some having younger aortas for their age. The possible effect this has on model flow-computed cardiac output was studied with our sensitivity analysis. We found that varying the patient age \( \pm 10 \text{ yr} \) hardly affected the error figures. We conclude that any effect of difference between calendar age and physiological age on aortic mechanical properties is small and does not affect significantly the precision of the model flow method.

During surgery it is often difficult to monitor the height of the arterial pressure transducer with respect to heart level. Awake subjects may change their position, anesthetized subjects may be tilted for surgical reasons, or the table may be changed in height. This may cause errors in the pressure level. We assumed that this effect would be less than \( \pm 10 \text{ mmHg} \) and computed the subsequent model flow error. Table 4 shows that there is little effect. Relative position of the pressure transducer, therefore, the aortic compliance did not depend on degree of sclerosis for pressures in the physiological range.

Apparently, the increased aortic wall stiffness with increasing degree of sclerosis is effectively compensated by the larger diameter. For the patient group studied we may expect arteriosclerosis to be present and thus may find aortic areas larger than average. This we did find (Table 3, \( k = 1.07 \)) but could not prove statistically (SD = 0.24).

TABLE 4. Summary of sensitivity analysis

<table>
<thead>
<tr>
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<th>Error Offset, l/min</th>
<th>Error Scatter, l/min</th>
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<tbody>
<tr>
<td></td>
<td>Mean</td>
<td>SD</td>
</tr>
<tr>
<td>Correct</td>
<td>0.09</td>
<td>0.22</td>
</tr>
<tr>
<td>A – 10 yr</td>
<td>0.17</td>
<td>0.16</td>
</tr>
<tr>
<td>A + 10 yr</td>
<td>0.07</td>
<td>0.29</td>
</tr>
<tr>
<td>P – 10 mmHg</td>
<td>0.03</td>
<td>0.15</td>
</tr>
<tr>
<td>P + 10 mmHg</td>
<td>0.10</td>
<td>0.23</td>
</tr>
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</table>

Situations: correct, pooled errors of Table 3 given for comparison; A ± 10 yr, all patient ages were entered 10 yr too young or too old; P ± 10 mmHg, pressure transducer simulated 10 mmHg (13 cm) too high or too low with respect to heart level.
With certainty we can state, therefore, that the upper methods are based on different physical principles. Errors of the thermodilution and model flow methods remain. They are statistically independent errors because model flow mean error, after calibration with thermodilution technique in patients. We assume, however, this has not been demonstrated for the phase-controlled techniques. Averaging over a large number of estimations, represents the differences between two imprecise techniques. Averaging several randomly injected estimates reduces this error. Averaging a series of phase-controlled estimations, however, reduces the error much more. In a previous study (4), averaging four phase-controlled injections reduced the errors from 14 to 3%. Thus our thermodilution errors are probably limited to no more than 5% SD.

Error Analysis

The postcalibration error mean and SD (Table 3) thus represent the differences between two imprecise techniques. Averaging over a large number of estimations, thermodilution probably has a mean error near zero (3). This has not been demonstrated for the phase-controlled injection technique in patients. We assume, however, that any mean error is small enough to be ignored. The model flow mean error, after calibration with thermodilution, is also near zero (Table 3). Thus only the scatter errors of the thermodilution and model flow methods remain. They are statistically independent errors because the methods are based on different physical principles. With certainty we can state, therefore, that the upper limit of scatter error for each technique must be below the 8% observed in this study. For the phase-controlled thermodilution technique we estimated an error SD of maximally 5%. Thus part of the 8% error in the comparison is due to errors in the reference method. We subtracted this error as explained elsewhere (18). The resulting error of the model flow method alone becomes 6%; that of the CZ method, 11%

Comparison With CZ Method

Pressure pulse analysis has been used to obtain beat-to-beat cardiac output monitoring (9). Under stable hemodynamic circumstances these techniques performed quite well, but when blood pressure, heart rate, peripheral resistance, or vascular tone changed, deviations occurred. An earlier method, the CZ method, tried to correct for this aberrant behavior (Eqs. 6 and 7), achieving improved precision under varying hemodynamic conditions and drug regimens (5, 13, 14, 19). Recently published errors for the CZ method are near the ones reported here in Table 3; older results indicate larger errors. This may be due to a less precise reference cardiac output method (18).

Precision of Thermodilution Estimates

The thermodilution technique is usually associated with large errors. Jansen et al. (4) have shown that, under conditions of mechanical ventilation, these errors are due in part to flow modulation and follow a systematic pattern. Averaging several randomly injected estimates reduces this error. Averaging a series of phase-controlled estimations, however, reduces the error much more. In a previous study (4), averaging four phase-controlled injections reduced the errors from 14 to 3%. Thus our thermodilution errors are probably limited to no more than 5% SD.

Conclusion

The model flow method extracts precise cardiac output information from arterial pressure, a signal that is often already recorded for other reasons. The nonlinear three-element model, representing the three major characteristics of arterial input impedance, allows a precise computation of stroke volume and cardiac output. The arctangent aortic pressure-area relation is a useful approximation of aortic nonlinearity. Radial artery pressure can be used to compute beat-to-beat values, although aortic pressure provides a visually better model flow waveform. For highest accuracy and precision a calibration of the model parameters is required.

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REFERENCES

8. LANGEWOUTERS, G. J., K. H. WESSELING, AND W. J. A. GOED-


