Evaluation of the Pulse-Contour Method of Determining Stroke Volume in Man

By Edwin L. Alderman, M.D., Angelo Branzi, M.D., William Sanders, M.A., Byron W. Brown, Ph.D., and Donald C. Harrison, M.D.

SUMMARY
The pulse-contour method for determining stroke volume has been employed as a continuous rapid method of monitoring the cardiovascular status of patients. Twenty-one patients with ischemic heart disease and 21 patients with mitral valve disease were subjected to a variety of hemodynamic interventions. These included exercise, isoproterenol infusion, and practolol, morphine or pentazocine administration. The pulse-contour estimations, using three different formulae derived by Warner, Kouchoukos, and Herd, were compared with indicator-dilution outputs. The pressure-pulse-contour computations were performed by a small on-line cardiac catheterization laboratory computer. A comparison of the results of the two methods for determining stroke volume yielded correlation coefficients ranging from 0.59 to 0.84. The better performing Warner formula yielded a coefficient of variation of about 20%. The type of hemodynamic interventions employed did not significantly affect the results using the pulse-contour method. This method yielded nearly as reliable results with the catheter positioned in the axillary artery as in the central aorta. Good results were obtained using the pulse-contour method in patients with atrial fibrillation as compared to those with sinus rhythm.

Although the correlation of the pulse-contour and indicator-dilution stroke volumes is high, the coefficient of variation is such that small changes in stroke volume cannot be accurately assessed by the pulse-contour method. However, the simplicity and rapidity of this method compared to determination of cardiac output by Fick or indicator-dilution methods makes it a potentially useful adjunct for monitoring critically ill patients.

Additional Indexing Words:
Indicator-dilution cardiac output  Cardiac output  Cardiac catheterization
Computer monitoring  Arterial-pressure pulse contour

Analysis of the arterial pulse contour for determining stroke volume is employed widely as a rapid method of monitoring cardiovascular function in critically ill patients. On-line computer capability in coronary and intensive care units makes continuous observation of stroke volume by this method possible. Although the Fick and indicator-dilution methods are also used to determine cardiac output, they are cumbersome to perform repetitively in critically ill patients.

The technic of estimating the stroke volume from the arterial pulse contour has its theoretic origins in Windkessel theory.1 The application of Windkessel theory is based on the assumption that the central aorta and

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proximal large arteries can be viewed as a chamber which fills only during systole and drains during both systole and diastole. The systolic filling of this chamber is, in a complex way, related to the pressure rise during systole. By means of various applications of Windkessel theory, several investigators have attempted to determine stroke volume solely on the basis of pressure-waveform and pressure-transmission characteristics.\textsuperscript{2} Warner\textsuperscript{3} related the intraaortic pressure rise to an initial Fick or indicator-dilution determination of the stroke volume. Subsequent determinations of the pressure rise could then be related to changes in stroke volume. Analogous formulae by Kouchoukos\textsuperscript{5} and Herd\textsuperscript{8} have been developed and utilized clinically.

An evaluation of the accuracy and reproducibility of the pulse-contour method was carried out in a variety of patients undergoing routine cardiac catheterization studies. The results of the Warner, Kouchoukos, and Herd formulae were compared against indicator-dilution cardiac outputs. Computations were performed by a computer system comparable to those used in special patient care units.

Methods

Patient Selection

Forty-two patients with heart disease were studied during cardiac catheterization. Twenty-one of these patients (19 males, two females) had coronary artery disease and were in sinus rhythm. The average age of this group of patients was 48 years. The other twenty-one patients, 10 of whom were in atrial fibrillation, had mitral valve disease. Two thirds of these patients had mitral stenosis and one third had mitral regurgitation of mild to moderate extent. Patients with severe mitral regurgitation or evidence of aortic valve disease were excluded. The latter group was not studied because of the possibility of aortic pressure-contour distortion.

Procedure

Stroke volume was measured by indicator-dilution methods and by the pulse-contour method under a variety of hemodynamic conditions. A no. 7 or 8 French catheter was positioned in the main pulmonary artery. A 90-cm polyethylene catheter with an internal diameter of 0.114 cm (0.045 in) and an external diameter of 0.157 cm (0.062 in) was advanced retrograde into the central aorta from the brachial artery. A 105-cm catheter was used when catheterization was performed from the femoral artery. The aortic catheter was connected by stiff 152.4-cm (60-in) Teflon tubing with an internal diameter of 0.185 cm (0.073 in) and an external diameter of 0.356 cm (0.140 in) to a Statham P23Db pressure transducer via Luer-lok connections. A Hewlett-Packard no. 8805A amplifier was used which has a 3dB rolloff at 200 Hz. Comparison of the central aortic pressure obtained through the fluid catheter system with the pressure recorded from a no. 5 French Honeywell type MCP-055 solid-state catheter revealed no significant waveform distortion. The first derivative of the aortic pressure curve obtained from both catheter systems was very similar and was well below those values normally obtained from the left ventricle.

A standard sequence, outlined in table 1, was followed for the comparison of pulse-contour and indicator-dilution outputs during each hemodynamic condition. At the start of the sequence the central aortic catheter was connected to the pressure transducer. The pressure signal was keyed into the catheterization laboratory computer which calculated the stroke volume using the pulse-contour method. The aortic catheter was then connected to a Waters cuvette dye densitometer; the output was also processed by the catheterization laboratory computer.\textsuperscript{7} Indocyanine, 3.75 mg in 15 cc of diluent, was injected in 1 sec into the pulmonary artery and sampled from the aortic catheter using a Harvard pump, withdrawing blood at 20 cc/min. The aortic catheter was then reconnected to the pressure transducer and another determination of the stroke volume using the pulse-contour method was performed. Thus, for each hemodynamic condition a stroke volume was determined initially by the pulse-contour method, then by the indicator-dilution method, and again by the pulse-contour method.

This basic sequence, outlined in table 1, was modified in several ways. During the initial resting control condition two separate indicator-dilution outputs were performed. The first pulse-contour determination of stroke volume was paired with the first indicator-dilution cardiac output and the second pulse-contour determination was paired with the second indicator-dilution cardiac output in order to obtain two distinct estimations of the pulse-contour formula constants $K_1$ and $K_2$. The constant obtained from the initial pulse-contour determination in the resting condition ($K_1$) was used to calculate estimated stroke volumes ($SV_1$) from the initial pulse-contour determinations during subsequent hemodynamic conditions. Likewise, the constant obtained from the second pulse-contour determination in the
resting condition \((K_2)\) was used to calculate estimated stroke volumes \((SV_2)\) from the second pulse-contour determinations during subsequent hemodynamic conditions. The two pressure determinations and the two indicator-dilution outputs required less than 8 min to perform.

The basic sequence outlined in table 1 was modified in nine patients with ischemic heart disease who had arterial catheterization from the brachial rather than from the femoral artery. The initial pulse-contour determination and initial indicator-dilution cardiac output were performed with the arterial catheter in the central aorta. The catheter was then partially withdrawn and positioned under fluoroscopic control in the axillary artery. A second indicator-dilution cardiac output and a second pulse-contour determination were then performed. This modified sequence was carried out at rest and during all subsequent hemodynamic conditions in these patients in order to evaluate the effects of a peripheral catheter location on the pulse-contour method.

The hemodynamic condition of each patient was altered from the resting supine position in the following ways:

**Exercise.** Pulse-contour and indicator-dilution determinations of stroke volume were made with the supine patient performing leg exercise on a pedal ergometer. The load was adjusted individually so that steady-state exercise could be achieved and maintained. Sixteen of 21 patients with coronary heart disease and all 21 patients with mitral valvular disease were studied during steady-state exercise.

**Isoproterenol.** Isoproterenol, 1–3 \(\mu g/min\), was administered to eight patients with coronary heart disease until a stable tachycardia, 40% in excess of the resting heart rate, was obtained.

**Other drugs.** Ten patients with coronary artery disease received one of three intravenous drugs. Practolol, 0.6 mg/kg i.v., was given to four patients; morphine sulphate, 8 mg i.v., was given to three patients; and pentazocine, 48 mg i.v., was given to three patients. Pulse-contour and indicator-dilution cardiac output determinations were made 20 min following infusion of the above drugs.

**Exercise following Drug Administration.** Several patients had additional determinations of stroke volume by the pulse-contour and indicator-dilution methods while exercising following drug administration. Although these determinations were not numerically sufficient to comprise a separate group, they provided useful data on the reliability of the pulse-contour method within an individual patient.

**Computer Methods**

All computations were performed by a Hewlett-Packard 2115A digital computer which was programmed for on-line pressure-pulse analysis and other hemodynamic computations.

| Table 1 |
| Standard Procedure Used to Obtain Data for Comparison of the Pulse-Contour and Indicator-Dilution Methods |

<table>
<thead>
<tr>
<th>Sequence of measurements during resting condition</th>
<th>Determine pulse-contour constants</th>
<th>Sequence of measurements during hemodynamic intervention</th>
<th>Calculate stroke volumes</th>
</tr>
</thead>
<tbody>
<tr>
<td>1 Pulse contour</td>
<td>(\rightarrow K_1)</td>
<td>1 Pulse contour</td>
<td>(\rightarrow) Estimated (SV_1)</td>
</tr>
<tr>
<td>2 Indicator dilution</td>
<td>(\rightarrow K_2)</td>
<td>2 Indicator dilution</td>
<td>True (SV)</td>
</tr>
<tr>
<td>3 Indicator dilution</td>
<td>(\rightarrow K_2)</td>
<td>3 Pulse contour</td>
<td>(\rightarrow) Estimated (SV_2)</td>
</tr>
</tbody>
</table>

Abbreviations: \(SV\) = stroke volume; \(K\) = pulse-contour formula constant.

**Table 2**

Comparison of Manual and Computer Calculation of Pulse-Contour Formulae \((N = 17)\)

<table>
<thead>
<tr>
<th>SV/K ratio</th>
<th>Formula</th>
<th>Warner</th>
<th>Roushoukos</th>
<th>Herd</th>
</tr>
</thead>
<tbody>
<tr>
<td>Manual (mean calculation)</td>
<td>1047</td>
<td>1716</td>
<td>207</td>
<td></td>
</tr>
<tr>
<td>Computer (mean calculation)</td>
<td>1054</td>
<td>1723</td>
<td>207</td>
<td></td>
</tr>
<tr>
<td>SD of the differences</td>
<td>62</td>
<td>138</td>
<td>14</td>
<td></td>
</tr>
<tr>
<td>Correlation coefficient</td>
<td>0.93</td>
<td>0.96</td>
<td>0.94</td>
<td></td>
</tr>
</tbody>
</table>

Abbreviations: SD = standard deviation.
Indicator-Dilution Determination of Stroke Volume

The electrical output of the Waters densitometer was digitized at a rate of 10 samples/sec and processed by the cardiac catheterization laboratory computer. The integrated area of the indicator-dilution curve was calculated using the Hamilton formula. The indocyanine dosage was entered manually into the computer from a keyboard. Prior studies in our laboratory have shown that the standard deviation of the difference of computer-calculated outputs from manually calculated outputs is less than 6% of the mean. A comparison of computer-calculated outputs to simultaneous Fick cardiac outputs showed that the SD of the differences is approximately 11% of the mean. Duplicate indicator-dilution cardiac outputs performed under the resting condition in this study were reproducible to within 2% of each other.

Pulse-Contour Determination of Stroke Volume (SV)

Every arterial pressure sampled by the computer for pulse-contour measurement was analyzed using three formulae. The detailed derivation of these formulae is given in references cited below.

Warner Formula:

\[ SV = K \times \sqrt{P_{md}} \times (1 + Sa/Da) \]

where \( P_{md} \) = mean pressure difference between the last 80 msec of systole and the last 80 msec of diastole; \( Sa \) = integrated pressure above 20 mm Hg from a point 80 msec before the onset of systole to a point 80 msec before the onset of diastole; and \( Da \) = integrated pressure above 20 mm Hg from a point 80 msec before the onset of diastole to a point 80 msec before the onset of systole.

Kouchoukos Formula:

\[ SV = K \times Psa \times (1 + Ts/Td) \]

where \( Psa \) = integrated area of the systolic pressure curve from the onset of systole to the end of systole above the diastolic pressure; \( Ts \) = duration of systole in msec; and \( Td \) = duration of diastole in msec.

Herd Formula:

\[ SV = K \times (M-D) \]

where \( M \) = mean pressure throughout both systole and diastole; and \( D \) = diastolic pressure.

All three formulae depend upon the detection of the diastolic pressure, which is readily obtainable as the minimum pressure. The Warner and Kouchoukos formulae depend upon detection of the dicrotic notch to establish the end of systole and the onset of diastole. Detection of the dicrotic notch was obtained by utilizing a pattern-recognition algorithm currently employed in intensive care unit monitoring systems.

All three formulae for pulse-contour analysis are of the format: stroke volume = K x pressure-pulse computation. With the terms transposed this becomes \( SV/K \) = pressure-pulse computation. For each arterial-pulse contour analyzed the computer immediately calculated the heart rate and the pressure-pulse computation, i.e. the \( SV/K \) ratio for each of the three formulae. The arterial pressure was sampled by the computer for nine beats and digitized at a rate of 100 samples/sec. The nine beats were ranked according to peak systolic pressure. The \( SV/K \) ratios of the middle three of nine beats, based on peak systolic pressure, were averaged to generate the final \( SV/K \) ratio for each formula. Thus, for each hemodynamic condition two separate arterial pressures were analyzed. For each pressure analysis a heart rate, an \( SV/K \) (Warner), an \( SV/K \) (Kouchoukos), and an \( SV/K \) (Herd) were

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generated. From the SV/K ratio, the K value for each formula was calculated during the resting control condition. After each subsequent hemodynamic intervention the estimated stroke volume was calculated from the SV/K ratio.

A comparison was made of the SV/K ratios calculated by the computer with the ratios calculated by manual planimetry on identical pressures recorded at a paper speed of 50 or 100 mm/sec. Seventeen arterial pressures were compared in five patients covering the various hemodynamic and catheter conditions employed in this study. The SV/K ratios generated by the computer had a high correlation with the manually calculated SV/K ratios for each of the three pulse-contour formulae (table 2). The sd of the difference between the computer and manual calculations given as a percentage of the manual data mean was 5.9% for the Warner formula, 8.1% for the Kouchoukos formula, and 6.7% for the Herd formula.

**Statistical Methods**

Comparison of the stroke volumes as determined by pulse-contour and indicator-dilution methods was performed by fitting straight lines by least squares, using the estimate based on the pulse-contour value as the dependent variable and the indicator-dilution value as the true or independent variable. For each regression line the slope, intercept, correlation coefficient, and standard error of the estimate of true value from pulse-contour estimate were calculated. The indicator-dilution stroke volumes and the estimates of stroke volume, obtained using the three

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**Figure 2**

The estimated stroke volume obtained using the Kouchoukos pulse-contour formula (ordinate) is graphed against the "true" stroke volume determined using the indicator-dilution method (abscissa) under the same conditions as in figure 1. The correlation coefficient is 0.59 and the best-fitting regression (solid line) is \( y = 1.04x + 15.4 \). The dashed lines indicate the 95% limits for an individual determination.

**Figure 3**

The estimated stroke volume obtained using the Herd pulse-contour formula (ordinate) is graphed against the "true" stroke volume determined using the indicator-dilution method (abscissa) under the same conditions as in figure 1. The correlation coefficient is 0.79 and the best-fitting regression (solid line) is \( y = 0.94x + 11.1 \). The dashed lines indicate the 95% limits for an individual determination.

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pulse-contour formulae, were compared with regard to the following considerations: (1) Quantitative precision. The ability of pulse-contour methods to estimate quantitatively the true stroke volume and to detect directional changes in stroke volume. (2) Reproducibility. The consistency of the stroke volume estimation when determined under the same situation at different times. (3) Hemodynamic intervention. The effect of exercise, isoproterenol, and other drug interventions on the performance of the pulse-contour method. (4) Patient group. The influence of disease (i.e. ischemic heart disease vs mitral disease) and cardiac rhythm (i.e. sinus rhythm vs atrial fibrillation) on the pulse-contour method. (5) Catheter position. The effect of catheter position in central aorta or axillary artery on the performance of the pulse-contour method.

Results

Precision

An overall comparison of the results of the pulse-contour and indicator-dilution methods for determining stroke volume, regardless of hemodynamic condition, cardiac rhythm, disease state, or catheter position, yielded correlation coefficients of 0.84 for the Warner formula, 0.59 for the Kouchoukos formula, and 0.79 for the Herd formula. Figures 1–3 show the data for the three methods; table 3 contains additional summary statistics for comparing the three methods. From the figures it can be seen that all three formulae tended to overestimate the indicator-dilution stroke volume. The extent of this bias was on the order of 10–15 cc/beat. The standard error of the estimation of the true (indicator-dilution) stroke volume was ±24% using the Warner formula, ±29% using the Herd formula, and ±52% using the Kouchoukos formula.

Directional changes in stroke volume were inaccurately determined by the pulse-contour method in about one quarter of the estimations (table 4). All three formulae were about equal in their ability to detect these directional changes. Declines in stroke volume were less well detected by all three formulae than were increases in stroke volume. The explanation for this is that hemodynamic interventions which diminished cardiac output produced

<table>
<thead>
<tr>
<th>Catheter site</th>
<th>Warner formula</th>
<th>Kouchoukos formula</th>
<th>Herd formula</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ao + Ax</td>
<td>0.35</td>
<td>0.05</td>
<td>9</td>
</tr>
<tr>
<td>Ao + Ax</td>
<td>0.54</td>
<td>0.04</td>
<td>14</td>
</tr>
<tr>
<td>Ao + Ax</td>
<td>0.83</td>
<td>0.08</td>
<td>22</td>
</tr>
</tbody>
</table>

Table 3: Comparison of Pulse-Contour and Indicator-Dilution Methods for Determining Stroke Volume

Abbreviations: Ao = aorta; Exer = exercise; Isoprot = isoproterenol; N = number of observations; Au = axillary artery; r = correlation coefficient; b = slope; c = intercept; cv = coefficient of variation on the true x scale.

- Includes eight observations under other hemodynamic conditions.

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Table 4

Comparison of Pulse-Contour Method with the Indicator-Dilution Method for Determination of Changes in Stroke Volume*

<table>
<thead>
<tr>
<th>Indicator-dilution method</th>
<th>Pulse-contour method</th>
</tr>
</thead>
<tbody>
<tr>
<td>Change in SV</td>
<td>No. &quot;correct&quot; determinations</td>
</tr>
<tr>
<td>Direction</td>
<td>Warner</td>
</tr>
<tr>
<td>Increase</td>
<td>39</td>
</tr>
<tr>
<td>Decrease</td>
<td>32</td>
</tr>
</tbody>
</table>

*Data obtained from patients with coronary heart disease.

very small decreases in stroke volume. The error for a single pulse-contour estimation of stroke volume is such that the greater the change in the true stroke volume, the more likely is the directional change to be accurately detected. In general, a directional change of at least 15-20% in stroke volume was required for reliable pulse-contour detection.

Reproducibility

Measurement of the stroke volume by the pulse-contour method was very reproducible while the patient remained in the same hemodynamic condition. Figure 4A illustrates a comparison of the first and second estimations of stroke volume during the exercise condition using the Warner formula. The correlation coefficient was 0.94. The regression line closely follows the line of identity. The Kouchookus and Herd formulae had similar degrees of reproducibility. The reproducibility of the pulse-contour computation during the resting condition was evaluated by examining the two separate SV/K ratios calculated by

![Figure 4](image)

(A) Two pulse-contour estimations of the stroke volume during the same exercise condition (Warner formula) are graphed against each other (SV₁ and SV₂). These stroke volume estimations were separately computed from pressures sampled 5-8 min apart. The best-fitting line \((y = 0.93 x + 4.3)\) obtained from these data is very close to the line of identity. (B) Two separate computer calculations of the Warner pressure pulse-contour integration (SV/K) during the resting control period are graphed against each other (SV/K-1 and SV/K-2). These SV/K computations were made separately from pressures sampled 5-8 min apart. The best-fitting line \((y = 0.77 x + 236)\) is close to the line of identity over the range of values encountered in this study (700-1400).

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the computer during the resting control condition. The correlation coefficient for these two determinations was 0.86 using the Warner formula. The regression line is shown in figure 4B.

The indicator-dilution stroke volumes of most patients ranged from 50 to 80 cc; however, the overall range was 17–166 cc. There was a tendency for the stroke volume in a given patient to change less than 25% of the resting control value irrespective of the hemodynamic intervention used. Interventions such as exercise and isoproterenol, although increasing the cardiac output by as much as 50–100%, only increased the stroke volume by 10–25%. Therefore, stroke volumes following hemodynamic challenge tended to remain relatively close to the control stroke volume.

It is possible that these small stroke volume shifts resulting from hemodynamic intervention may produce only minor random changes in the pulse-pressure contour. Since this method relies on an initial independent cardiac output determination to establish the relationship between the pressure-pulse contour and the stroke volume, even if the pulse contour were randomly affected by a hemodynamic intervention, there would still be some correlation of the estimated and true stroke volume across patients because of the relative constancy of the stroke volume within a given patient. In order to evaluate whether the pulse-contour method was successful because of its dependence upon a control stroke volume determination, the data were analyzed separately for each individual patient who was studied under two or more hemodynamic conditions. For each patient the stroke volumes determined using the pulse-contour method were plotted against the stroke volumes using the indicator-dilution method, and a regression line was calculated. If one assumed that the pulse-contour method contributed nothing to the estimation of stroke volume, then the slopes of these individual patient regression lines would distribute randomly around an average slope of zero. The actual average slope, obtained from the regression lines for the eleven subjects who were evaluated under two or more hemodynamic conditions, was 0.86, which is significantly different from zero at the 0.1% level.

Hemodynamic Condition

The effects of exercise, isoproterenol, pentazocine, morphine, or practolol on the heart rate and stroke volume of the patients studied are summarized in table 5. Exercise and isoproterenol administration increased the heart rate by 28–40%. The stroke volume increased to a lesser extent following exercise or isoproterenol with the increases ranging from 2 to 16%. In nine of ten patients who received either practolol, morphine, or pentazocine, the stroke volume was decreased below the resting value.

The better performing Warner and Herd formulae were not significantly affected by the type of hemodynamic condition during which they were used (table 3). The stroke volume, as determined by the Warner formula during isoproterenol administration, had a 0.95 correlation with the stroke volume determined by the indicator-dilution method. The same formula achieved a 0.88 correlation during exercise, and a 0.83 correlation during the administration of other drugs. Figure 5 illustrates the regression lines determined from a comparison of the pulse-contour stroke volumes and the indicator-dilution stroke volumes measured during different hemodynamic conditions. The Warner formula demonstrated the narrowest range of slopes and intercepts.

Patient Group

The results obtained with all three formulae were not affected substantially by the patient’s disease or cardiac rhythm (table 3). The stroke volume as determined by the Warner formula had a 0.85 correlation with the stroke volume determined by the indicator-dilution methods in patients with mitral disease who were in sinus rhythm. An identical comparison, with the exception that the patients with mitral-valvular disease were in atrial fibrillation, achieved a correlation coefficient of 0.83. The Warner formula had a similar level of
correlation in patients with coronary artery disease who were in sinus rhythm. The coefficient of variation which ranged from 12 to 20% in patients with coronary artery disease was 27–37% in patients with mitral valvular disease, using the Warner formula.

Catheter Position
Catheter positioning in the axillary artery had a relatively minor influence on the ability of the pulse-contour method to estimate stroke volume. Figure 6 shows the stroke volumes estimated using the pulse-contour method against the stroke volumes obtained using the indicator-dilution method for both the axillary and central-aorta catheter positions. The regression lines for both sets of data indicate that at low stroke volumes the axillary catheter position tends to overestimate the stroke volume. Conversely, at high stroke volumes the axillary catheter position tends to underestimate the stroke volume in comparison to the aortic catheter position.

Discussion
The Warner, Kouchoukos, and Herd formulae have been extensively evaluated in animal studies. Estimation of the stroke volume using the Warner formula had a 0.98 correlation with indicator-dilution methods following infusion of angiotensin and acetylcholine.4 The stroke volume estimation using the Kouchoukos formula was compared with beat-by-beat electromagnetic flow measurements in dogs.5 An array of hemodynamic interventions was employed and an overall correlation of 0.928 was obtained. Stroke volume estimation using the Herd formula had a 0.93 correlation with indicator-dilution cardiac outputs following hemorrhage and vagal stimulation in dogs.6

All three formulae have also been evaluated in human subjects. The Warner formula was evaluated following exercise, 70° tilt, and g-suit plus tilt. Other than the 70° tilt intervention, the estimated stroke volume was only 6–7% different from Fick or indicator-dilution outputs.3 In a later study4 using a modified Warner formula, comparisons of stroke volume estimations were made with the indicator-dilution method following exercise and atrial pacing. Agreement between the two methods was as good as repeat indicator-dilution cardiac outputs. Stroke volumes obtained using the Kouchoukos formula were compared with outputs obtained using indicator-dilution methods in 31 patients in the first 24 hours postcardiac surgery.10 The correlation was 0.95, and the standard error of the estimate was ±16%. The Herd formula was evaluated in 15 patients under varied situations using the radial artery pressure.11 A 0.959 correlation with indicator-dilution methods was obtained.

In the study reported here, the Warner formula seems to be the best available formula for estimating stroke volume from the pressure contour. It yields a coefficient of variation of about 20%, which is the smallest variation of the three formulae evaluated. The original Warner formula has been modified by taking the square root of the term $P_{mol}$.4

Table 5

<table>
<thead>
<tr>
<th>Pt group</th>
<th>Data</th>
<th>Hemodynamic condition</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>Rest</td>
</tr>
<tr>
<td>Coronary</td>
<td>HR</td>
<td>70.5</td>
</tr>
<tr>
<td></td>
<td>SV</td>
<td>69.4</td>
</tr>
<tr>
<td>Mitral-sinus rhythm</td>
<td>HR</td>
<td>74.4</td>
</tr>
<tr>
<td></td>
<td>SV</td>
<td>60.5</td>
</tr>
<tr>
<td>Mitral-atrial fibrillation</td>
<td>HR</td>
<td>83.3</td>
</tr>
<tr>
<td></td>
<td>SV</td>
<td>45.4</td>
</tr>
</tbody>
</table>

Abbreviations: HR = heart rate; SV = stroke volume (indicator-dilution method).

*Pentazocine, practolol, morphine.
Compared to the modified formula, which was used in this study, was not significantly influenced by the type of hemodynamic intervention or by the cardiac rhythm. The pulse-contour measurement of stroke volume was quite consistent in closely spaced duplicate determinations. Over
Comparisons of the estimated stroke volume by the pulse-contour method (ordinate) against the "true" stroke volume determined by the indicator-dilution method (abscissa) are graphed for both the central aorta and axillary artery catheter positions. Only data from those patients who had both aortic and axillary measurements are shown. (A) Data obtained using the Warner formula. The regression line for the aortic catheter position is \( y = 1.12x - 0.8 \) and for the axillary catheter position, \( y = 0.96x + 10.8 \). (B) Data obtained using the Kouchoukos formula. The regression line for the aortic catheter position is \( y = 2.05x + 62.7 \) and for the axillary catheter position, \( y = 1.10x + 2.3 \). (C) Data obtained using the Herd formula. The regression line for the aortic catheter position is \( y = 1.15x - 3.4 \) and for the axillary catheter position, \( y = 0.88x + 12.4 \).
longer periods of time progressively greater deviation from the true value might occur, necessitating recalibration using a repeat indicator-dilution determination of cardiac output.

The Kouchoukos formula was the least successful in estimating the true stroke volume. The Herd formula was developed as a relatively uncomplicated calculation which could be handled by inexpensive analog electronics. In this study it performed nearly as well as the Warner formula.

Studies in dogs, using the Kouchoukos formula, had suggested that the administration of catecholamine substantially reduced the accuracy of the pulse-contour method. This was attributed to altered compliance of the central arterial vessels in response to the drug. The data from human subjects in our study did not suggest that the pulse-contour estimation was adversely affected by isoproterenol. It is probable that there is substantially greater beta-receptor activity in the central arterial vessels of dogs than in adult human subjects. Large species and age differences in beta-receptor activity have been demonstrated in helically cut strips of thoracic and abdominal aorta from rats, rabbits, guinea pigs, and cats.

The axillary artery catheter position did not significantly alter the ability of the pulse-contour method to estimate stroke volume. The axillary artery position was evaluated because arterial catheters cannot always be passed retrograde into the central aorta without fluoroscopy. Hamilton and Remington obtained good results estimating the stroke volume from central aorta pressure data, but had little success using brachial or femoral artery pressure data. In our study the axillary artery catheter position yielded sufficiently good results to suggest that the catheter location was central enough to minimize pulse-contour distortion.

The pulse-contour method calculates the stroke volume on a beat-by-beat basis. One might therefore expect difficulties in atrial fibrillation comparing the pulse-contour stroke volume estimation against an indicator-dilution cardiac output. The method employed in this study averaged the stroke volume estimate from the three most representative beats of a nine-beat sample. The better performing Warner and Herd formulae had the same correlation in atrial fibrillation as in sinus rhythm; however, the coefficients of variation were greater in atrial fibrillation. The only hemodynamic intervention employed in these patients was exercise, thus limiting conclusions concerning the overall applicability of this method to patients in atrial fibrillation.

Although the pulse-contour method for determining stroke volume achieved a high degree of correlation with the indicator-dilution method, the coefficient of variation is quite substantial. Estimates of stroke volume using the pulse-contour method, which are as much as one third above or below the indicator-dilution stroke volume, will occur at least once in 20 times. As a result, the pulse-contour method cannot be expected to measure accurately small changes in stroke volume. Conversely, large changes in stroke volume, such as those seen early in the recovery of patients from cardiac surgery, may be detected quite satisfactorily. In a comparison of the pulse-contour method and the indicator-dilution method in such patients, Kouchoukos reported a 0.95 correlation with a standard error of ±16% in the first 24 hours postoperation; however, the correlation fell to 0.70 with a standard error of ±31% in the second 24 hours.

The detection of changes in the stroke volume of critically ill patients is important in assessing a patient’s prognosis and response to therapy. Determination of the stroke volume by Fick or indicator-dilution methods is too cumbersome to perform repetitively, and for this reason the pulse-contour method offers an attractive, although very approximate, alternative. It might be expected that there would be less variation in multiple pulse-contour measurements in an individual patient than the considerable variation noted in this study using multiple measurements made in different patients. This is suggested by the reproducibility of the pulse-contour computation.
obtained during the resting and exercise conditions, although at narrowly spaced intervals. The availability of continuous-flush attachments for arterial catheter-transducer systems and of small inexpensive digital computers allows for continuous monitoring of the cardiac output using this method. For these reasons, the pulse-contour method is readily applicable to the coronary care and intensive care situation. Although this study did not directly evaluate such patients, the data obtained would suggest consideration of this method for these units.

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Evaluation of the Pulse-Contour Method of Determining Stroke Volume in Man

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