Absolute Determination of Cardiac Output in Intra-aortic Balloon Pumped Patients Using the Radial Arterial Pressure Trace

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SUMMARY We describe a new method for the absolute determination of cardiac output in intra-aortic balloon pumped (IABP) patients. The method uses the known pumping volume of the IABP balloon and the radial arterial pressure trace, which is commonly used to monitor IABP patients, to determine the cardiac output. Two pressure excursions denoted by P2-P0 and P-P0, characterizing IABP balloon deflation, and ventricular ejection, respectively, are extracted from the radial trace. The cardiac output (CO) is then determined by the simple relation: CO = (BV) x (P2-P0)/(P-P0) x HR where HR is the heart rate, and the value for pumped balloon volume (BV) is corrected for the effect of the pressure in the patient's aorta. Comparison with dye dilution and thermal dilution procedures as carried out on a routine basis in a clinical setting produced a good correlation (r = .928). When fit to a straight line through zero output, the data yielded a constant of proportionality of 0.973 between the above formula and the clinical procedures. The procedure does not disturb the patient in any way, and enables continuous monitoring of cardiac output. This has been implemented using a real-time, miniaturized computer and allows much more information to be obtained than in usual single measurements.

THE RADIAL ARTERIAL PRESSURE CONTOUR which is commonly used to monitor intra-aortic balloon pumped (IABP) patients consists of several pulse segments representing discrete events in the cardiac cycle: ventricle ejection of blood into the aorta, balloon inflation accompanied by runoff from the aorta into the arterial tree, and balloon deflation, with the inflation transient separated from the balloon deflation by a short, relatively flat "plateau" region (fig. 1). The signal thus contains well defined cardiac and balloon events. Since one knows, and can control the pumping volume of the intra-aortic balloon, analysis of the radial pressure contour provides an opportunity to determine in absolute terms the stroke volume of the heart. This concept was enunciated by Arthur Kantrowitz (unpublished) in 1969. In this paper we present such an analysis, and describe a simple method for determining stroke volume and cardiac output on a continuous, beat by beat basis in IABP patients. The distortion of the arterial waveform as the pulse propagates from the aorta to the radial artery does not affect the features of the waveform relevant to the analysis, and the pressure is measured using the same radial cannula commonly used in IABP patients. The method described here gives the cardiac output (CO) directly in absolute terms and thus differs from the various empirical formulas which have been used in the past to determine CO within a constant of proportionality from the central arterial pressure contour.

The basic idea of the method is that two pressure excursions denoted by P2-P0 and P3-P0 can be extracted from the radial artery trace, one characteristic of balloon deflation within the descending thoracic aorta, the other of the change in aortic size due to ventricular ejection into the aortic root. As illustrated in figure 1, the pressure change P2-P0 extends from the end of the plateau-like region of the pulse (just before the point F in fig. 1) to the pressure minimum at the end of diastole (point G in fig. 1). The pressure change P3-P0 represents the excursion from the minimum at A to the systolic maximum at B. The cardiac output is then given by: CO = BV x (P2-P0)/(P3-P0) x HR

where HR is the heart rate, and BV the balloon pumping volume. The definition of P0, P1, and P2 for several other types of radial traces with varying plateau characteristics is shown in figure 2. We have found that the very simple for-
FIGURE 1. Patient IABP radial arterial pressure tracing, showing discrete events in IABP-cardiac cycle. Segment ABC corresponds to ventricular ejection, followed by balloon inflation and rapid run-off of blood from the aorta into the arterial tree-segment CDE. In the "plateau" region EF, the balloon is still fully inflated with the slow run-off of blood from the aorta. At F the balloon is deflated, with complete deflation occurring at G just prior to the beginning of the next systole. The radial pressure excursions P₁-P₀ and P₂-P₀ characterizing balloon deflation and ventricle ejection, respectively, are as indicated.

Figure 2. Additional examples of IABP patient tracings showing definition of pressures P₀, P₁, P₂ in cases of varying "plateau" region characteristics.

Figure 3. Patient radial trace showing the effect of late deflation. Overlap of the deflation and ventricle ejection pulses results in much reduced size of the balloon deflation pulse.
The key factor in our method is that the use of a calibrating pressure change P1-P0, corresponding to a known balloon deflation volume within the aorta, allows the heart pulse to be characterized very simply by a pressure difference, P1-P0.

Methods

Radial arterial traces from IABP patients in the Myocardial Infarction Research Unit (MIRU) and the Surgical Intensive Care Unit (SICU) of the Massachusetts General Hospital were analyzed. The measurements were carried out without disturbing the patient in any way, and the same radial arterial cannula which is routinely used for patient monitoring was used. Some of the analyses were carried out by manual computation using Equation (1) above, and some using a miniaturized, on-line digital computer which was pre-programmed to carry out the analysis.

Avco IABP systems were used. The volume meters of the systems were recalibrated at 30 cc at a back pressure of 75 mm of mercury. (For normal IABP use, a precise volume meter calibration is not essential, and is not usually performed. To apply the formula presented here, such a calibration should be undertaken.) With this calibration a setting of 40 cc corresponded to a pumped volume of approximately 39 cc, while a setting of 20 cc corresponded to approximately 21 cc, at the same back pressure, 75 mm. To insure a well defined filling volume for a given final pressure, the IABP systems were operated so as to effect complete emptying of the balloon to atmospheric pressure during the deflation part of the cycle ("automatic" mode on the Avco pump). When in use the actual pumped volume of the intra-aortic balloon was determined using the instantaneous aortic pressure just before deflation which is obtained from the balloon console pressure. For operation of the system without balloon stretching, the usual mode of operation, the balloon console pressure is in equilibrium with the aortic pressure just before deflation at the point P in figure 4. The other portions of the console balloon pressure curve are discussed in the figure caption. The dependence of balloon volume on aortic pressure will depend upon the type of console used, especially the "dead" volume of the system (the volume of connecting tubes within the console). For Avco IABP systems, each additional 10 mm of aortic back pressure corresponds to a decrease in pumped volume of approximately 1½ cc. Thus with a system whose volume meter has been calibrated to 30 cc at a back pressure of 75 mm, an IABP patient having a 30 cc intra-aortic balloon and an aortic pressure just prior to deflation of 95 mm (at point P in fig. 4) would have a true pumped volume of 27 cc and for this patient, a value for BV of 27 cc would be used in the above formula (1). For a patient with an aortic pressure of 65 mm, a volume of 31½ cc would be used, etc. (The Avco balloons are a few cc oversized so that 31½ cc would not, in general, cause balloon stretching.)

Cardiac outputs were determined using the radial arterial trace, and balloon console pressure trace as described, and compared with measurements obtained using standard clinical methods. The clinical measurements were carried out as a routine part of therapy in either the MIRU or SICU units. For some patients, clinical cardiac outputs were determined using a thermal dilution system while for others an indicator dye dilution method was used. In the thermal dilution measurements 10 cc of 0°C dextrose solution was injected via a flow-directed Swan-Ganz catheter (Edwards Laboratories) into the pulmonary artery. Cardiac output was determined from the resulting temperature curve using an Edwards cardiac output computer. In the dye dilution measurements, indocyanine green dye was injected via a Swan-Ganz catheter into the pulmonary artery, with arterial blood withdrawn from an in-dwelling radial cannula. The dye was detected with a Gilford densitometer and the dye curve analyzed using the Stewart-Hamilton formula which was implemented either manually or using a Lexington Instruments cardiac output computer.

The thermal dilution comparison measurements were carried out simultaneously with the IABP pressure curve analyses, with a given determination representing the average of 2 to 6 such comparisons, all carried out within approximately a 5 to 15 minute period. The dye dilution comparisons (generally 2 measurements) were carried out approximately 10 minutes before or after the arterial pressure analyses, since the radial arterial trace was not available during the dye dilution measurements.

Results

Comparison of the Method With Thermal Dilution and Dye Dilution Methods

A comparison of the present method with clinical methods is shown in figure 5. The IABP pressure trace measurements derived using formula (1) are presented as derived, with no constant factor added. The data represent 44 output comparisons in 40 patients, with the recorded outputs ranging from 1.65 to 8.2 L/min. The mean outputs were 4.84 L/min using formula (1) and 4.93 L/min using thermal and dye dilution.

Satisfactory computations were performed in approximately 95% of the IABP patients. In the remaining 5%, no plateau region could be defined while still maintaining proper timing of inflation and deflation of the balloon. In these traces, the inflation transient was not distinct from the deflation pulse, so that the pressure excursion P1-P0 could not be defined.

For the 95% which were analyzed, the relatively simple formula (1) showed surprisingly good agreement with the clinical measurements. A straight line fit to the data through zero output (both methods should yield zero at zero output) of the form: CO(1) = K x CO(TD/DYE) gives a "best fit"
value of K of 0.973, while the value of K, which minimizes the percentage error for each measurement, is 0.993. Other fits to the data in which an offset is allowed at zero output are given in table 1. The over-all correlation coefficient for the data is \( r = 0.928 \). The standard deviation of the percentage difference between the two methods, a measure of the percentage error to be expected in a given determination of cardiac output, is approximately 12%.

**Continuous Monitoring of Cardiac Output**

The method permits a continuous, beat by beat analysis, without interfering with the patient. Figure 6A shows a continuous recording of CO over approximately a 1 hour period using the method as implemented on a special purpose computer which was pre-programmed to perform the calculation of Equation (1). The pressure analysis was carried out at approximately 10 sec intervals using 5 sec of patient pulse data, with each point plotted as received in "real-time" and representing the updated average of the prior four measurements. Thermal dilution measurements were performed at the times indicated. In figure 6B a second continuous recording of CO over a short period of time is shown. The patient who had been lying tranquilly was sufficiently disturbed by the preparations for the thermal dilution CO measurement that his cardiac output transiently increased followed by a period of higher CO than that existing before the patient had been aroused. In both examples

**Figure 5.** Comparison of cardiac outputs determined using the radial trace analysis algorithm (ordinate) with those determined by means of indicator dye or thermal dilution (abscissa) in 40 IABP patients. Lines indicating ±20% deviation from perfect correspondence are shown. The filled circles represent the average over simultaneous algorithm/thermal dilution comparisons while the open circles represent algorithm/dye dilution comparisons.

**Figure 6.** Upper trace) Continuous cardiac output recording (40 sec moving average) over approximate 1 hour period. Arrows indicate times when thermal dilution (TD) injections were made. (Average over TD's CO = 6.6 L/min; average over corresponding computer readings, CO = 6.5 L/min). Lower trace) Recorded large variation in cardiac output near the time of thermal dilution measurements (no moving average). Bringing the thermal dilution apparatus to the patient's bedside and arousing the patient produced a transient increase in output followed by a higher output than that which had existed prior to the measurement. (Thermal dilution measurements at arrows averaged 6.2 L/min, while the average over corresponding computer readings yielded 6.3 L/min.)
shown, the cardiac output fluctuated substantially over relatively short time periods, and the continuous display of CO can thus be used to yield information not available in conventional single measurements.

Acknowledgment

The author gratefully acknowledges useful discussion with Dr. Arthur Kantrowitz and Dr. Harry Petschek of Avco Everett Research Laboratory, Inc., and with Dr. Mortimer J. Buckley, Dr. Hermann K. Gold, Dr. Robert C. Leinbach and Dr. Eldred D. Mundth of Massachusetts General Hospital and Harvard Medical School. He would also like to thank Mr. Alfred Magro of Avco for assistance in data collection, and Mr. Armando Federico also of Avco who collaborated in developing the special purpose computer used in part of this study. The clinical comparisons were made possible by the assistance and cooperation of Mr. John Hinckley and Mr. John Drake at the MIRU and Dr. Myron B. Laver and his staff at the SICU at the Massachusetts General Hospital.

References


| Table 1. Regression Line Fits to the Cardiac Output Data |
|---------------------------------|---------------------------------|----------------|
| Description                     | Equation of fit                 | SEE            |
| 1. Line through zero output     |                                |                |
| a) Fit minimizes absolute error | CO (1) = 0.973 × CO (TD/Dye)   | 0.56 L/min     |
| b) Fit minimizes percentage of error | CO (1)/CO (TD/Dye) = 0.993 | 0.12 (12%) |
| 2. Allowing offset at zero output | | |
| a) CO (TD/Dye) dependent variable | CO (1) = 0.87 × CO (TD/Dye) + 0.55 | 0.53 L/min |
| b) CO (1) dependent variable    | CO (TD/Dye) = 0.99 × CO (1) + 0.14 | 0.57 L/min |

Abbreviations: CO (1) = formula (1) determination of cardiac output; CO (TD/Dye) = thermal or dye dilution determination of cardiac output; SEE = standard error of the estimate.
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doi: 10.1161/01.CIR.53.3.417

Circulation is published by the American Heart Association, 7272 Greenville Avenue, Dallas, TX 75231
Copyright © 1976 American Heart Association, Inc. All rights reserved.
Print ISSN: 0009-7322. Online ISSN: 1524-4539

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