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 \( P_r - P_o \) and \( P_o - P_s \), representing IABP balloon deflation, and ventricular ejection, respectively, are extracted from the radial trace. The cardiac output \( (CO) \) is then determined by the simple relation:

\[
CO = \frac{BV \times (P_r - P_o) / (P_o - P_s)}{HR}
\]

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 yield 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 This concept was enunciated by Arthur Kantrowitz (unpublished) in 1969.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 aortic pressure contour.

The basic idea of the method is that two pressure excursions denoted by \( P_r - P_o \) and \( P_o - P_s \) 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 \( P_r - P_o \) 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 \( P_o - P_s \) represents the excursion from the minimum at A to the systolic maximum at B. The cardiac output is then given by:

\[
CO = \frac{BV \times (P_r - P_o) / (P_o - P_s)}{HR}
\]

where HR is the heart rate, and BV the balloon pumping volume. The definition of \( P_o \), \( P_r \), and \( P_s \) for several other types of radial traces with varying plateau characteristics is shown in figure 2. We have found that the very simple for-
mula (1) gives good agreement with conventional clinical methods of measurement, as described below. Several features of the above formula should be noted:

1) The pressure excursion \( P_2-P_0 \), representing balloon deflation, occurs at a point in time where there is very little effect due to runoff of blood from the aorta into the arterial tree, and no appreciable overlap of other portions of the signal. Little effect due to reflections is seen; if the timing of the deflation is altered slightly, making the width of the end-diastolic minimum wider or narrower, there is very little change in \( P_1 \), and in the ratio \( P_2-P_0/P_1-P_0 \) used in the above formula. If substantial reflections were present they would show up upon widening the diastolic minimum.

2) Most of the energy delivered by the left ventricle goes into stretching the aorta as it is inflated. The increased pressure which is developed is related to the change in aortic volume by a nonlinear pressure-volume relation. The pressure excursions \( (P_2-P_0) \) and \( (P_1-P_0) \) occur over nearly the same range in magnitude so that errors due to nonlinearities in the aortic compliance are minimized. (At relatively large ratios of \( P_2-P_0/P_1-P_0 \), the effect of nonlinearity of the aortic compliance may have to be taken into account. For the cardiac disease patients whose radial traces we have analyzed, most of the pressure ratios have been below 2 for which very little effect of the nonlinearity has been observed.)

3) The distortion of the pressure pulse in propagating along the arterial tree to the radial artery seems to leave the ratio \( P_2-P_0/P_1-P_0 \) intact. Most of the pulse distortion occurs in the smaller vessels and the displacement of the balloon relative to the ventricle within the aorta has little effect on the relative magnitude of the two radial pulses used in the analysis.

4) The balloon pumping volume \( (BV) \) used in the above formula is not necessarily the nominal balloon volume. The true inflation or deflation volume will in general depend on the magnitude of the "back pressure" of the patient's aorta against which the balloon must fill. The aortic pressure at the time just prior to deflation is readily obtained from the balloon console pressure which is normally displayed in an IABP system as discussed below.

5) The above formula assumes that balloon timing has been correctly set. In particular, the formula cannot be used if balloon deflation has been set so late as to permit incomplete balloon deflation at the time of the opening of the aortic valve at the beginning of systole of the subsequent beat. This situation is undesirable from a physiological point of view, and in the case of cardiac output determination, the balloon deflation and ventricle ejection pulses overlap, and the above formula will not give correct readings. Late deflation's effect on the radial arterial trace is shown in figure 3.

6) The use of a pressure difference \( P_2-P_0 \) to characterize the patient pulse tacitly includes the pulse duration \( \tau \) in the determination of stroke volume. Since, in an exponentially decaying system, as in a windkessel model for aortic drainage, a difference in pressure over a time \( \tau \) which is much less than the decay time constant \( \tau_0 \) of the system is approximately \( P_2-P_0 \approx \frac{P_1}{\tau_0} \). Thus the above formula (1) has some similarity in form to some of the semi-empirical windkessel formulas used to model the central aortic pressure

\[
\text{(1) } \frac{P_2-P_0}{P_1-P_0} = \frac{1}{1 + \frac{\tau}{\tau_0}} 
\]

\[
\text{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 \( P_1-P_0 \) and \( P_2-P_0 \) characterizing balloon deflation and ventricle ejection, respectively, are as indicated.}
\]

\[
\text{Figure 2. Additional examples of IABP patient tracings showing definition of pressures } P_0, P_1, P_2 \text{ in cases of varying "plateau" region characteristics.}
\]

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

**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 \( P_r - P_o \) 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: \( CO(1) = K \times CO(TD/DYE) \) 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
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 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.

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doi: 10.1161/01.CIR.53.3.417

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