Minimally invasive cardiac output monitors

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Revalidation FOR ANAESTHETISTS
RCA Revalidation matrix
Matrix reference 2A04

Key points

Minimally invasive cardiac output monitors have varying degrees of 'invasiveness' with some being totally non-invasive and others only marginally less invasive than a pulmonary artery catheter (PAC).

All minimally invasive cardiac output monitors have their own sources of potential error.

Users of minimally invasive cardiac output monitors should be aware of their potential sources of error and clinical limitations.

Most currently available minimally invasive cardiac output monitors compare well with the PAC in haemodynamically stable patients.

Information regarding the impact of minimally invasive cardiac output monitors on clinical outcome is scarce.

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Monitoring cardiac output is a common practice in anaesthesia and critical care. It is used as a marker of oxygen delivery to tissues and can identify patients at high risk of significant morbidity, mortality or both. It is also used in guiding treatment, primarily for fluid resuscitation and the use of vasoactive and inotropic drugs. First introduced in 1970, the Swan–Ganz catheter has become an important tool for calculating cardiac output. However, the publication of a number of studies showing little or no change in clinical outcome has resulted in their use declining. Despite this, the Swan–Ganz or similarly designed thermodilution catheters are still considered the gold standard for cardiac output measurement, against which all new monitors are measured.

Although widely available and easy to use once inserted, some degree of skill is required to position the pulmonary artery catheter (PAC) accurately. This may not be available at all institutions. Also, PACs have a number of disadvantages, principally arising from their 'invasive' nature. Complications of infection, pulmonary artery rupture, arrhythmias on insertion, thrombosis, and embolism have all been reported.

The search for a new, less invasive method of measuring cardiac output has led to the introduction of many new devices into clinical practice. The term 'minimally invasive cardiac output monitors' collectively describes all devices that calculate cardiac output without requiring insertion of a PAC. Each of these devices, however, utilizes different techniques to determine cardiac output. As such, each have their own sources of potential error and degree of 'invasiveness' (Table 1). In this article, we aim to review the principles behind the calculation of cardiac output and potential sources of error for devices used in clinical practice. In addition, we review some of the clinical situations in which the accuracy of these monitors may be limited.

Pulse contour analysis

Erlanger and Hooker first described the theory for pulse contour analysis in 1904. They suggested that cardiac output was proportional to arterial pulse pressure.[1] Pulse contour devices available today utilize the same principle and relate the contour of the arterial pressure waveform to stroke volume and systemic vascular resistance. An algorithm is used to determine the cardiac output and produce a continuous read out (Fig. 1).

Each device can be set to display a range of user-determined physiological variables, including cardiac output, cardiac index, and heart rate. All these devices additionally provide information on stroke volume variation (SVV) with respiration as an indicator of fluid responsiveness.

SVV is the difference between maximum and minimum stroke volumes over the respiratory cycle and is caused by changes in preload with alterations in intra-thoracic pressure. SVV can be used as an indicator of fluid responsiveness.[1] In general, patients with an SVV of $<10\%$ are unlikely to be fluid responsive, whereas those with an SVV of 15% or greater are likely to benefit from fluid resuscitation.

There are a number of devices currently in clinical use that use the pulse waveform for continuous cardiac output calculation; of which, three are described here. These three devices all use different methods of calculating cardiac output. The PiCCO system (Pulsion Medical Systems, Munich, Germany) uses a thermistor-tipped arterial line in a proximal artery to measure the aortic trace waveform morphology. An algorithm is used to determine the cardiac output by integrating the area under the curve of the arterial pressure *vs* time trace. A central venous catheter (CVC) is used to calibrate the system using a transpulmonary thermodilution technique described below.[2]

The FloTrac/Vigileo system (Edwards Lifesciences, Irvine, CA, USA) also utilizes a blood flow sensor, attached to a standard

doi:10.1093/bjaceaccp/mkr044 Advance Access publication 3 October, 2011

Table 1 Comparison and limitations of minimally invasive cardiac output measurement techniques (PiCCO system, LiDCO monitor, FloTrac/Vigileo system, USCOM device). PAC, pulmonary artery catheter; CVC, central venous catheter; E'_{CO_2}, end-tidal carbon dioxide

Device	Invasiveness	Equipment required	Limitations
PAC	+++	Central venous access Thermodilution capable PAC	Not continuous in standard form
Pulse contour analysis PiCCO	++	Thermistor-tipped arterial line in a central vessel CVC	Requires intermittent recalibration and a central arterial catheter Less accurate with significant aortic regurgitation, arrhythmia, or intra-aortic balloon pump
Pulse contour analysis LiDCO	++	Arterial line ±CVC	Requires intermittent recalibration
Pulse contour analysis FloTrac/ Vigileo	+	Arterial line	Less accurate for absolute measurement than calibrated pulse wave devices Less accurate with significant aortic regurgitation, arrhythmia or intra-aortic balloon pump
Oesophageal Doppler	+	Transoesophageal Doppler probe	Poorly tolerated unless tracheal tube present Relies on assumed proportion of blood flow through the descending aorta
USCOM	−	Transthoracic Doppler probe	Uses nomogram for valve area estimation Not accurate with significant valve stenosis
Gas re-breathing	−	Rebreathing circuit Dedicated in-line E'_{CO_2} and pulse oximetry probe	Requires tracheal intubation and stable tidal volumes during measurements
Transpulmonary thermodilution	++	Thermistor-tipped arterial line CVC	Less accurate with pulmonary congestion and in the presence of shunting
Lithium dilution	+/++	Arterial line ±CVC	Inaccurate with intercurrent lithium use Contraindication to lithium preclude use Contraindicated in patients <40 kg and first trimester of pregnancy
Thoracic bioimpedance	−	Cutaneous electrodes only	Accuracy with haemodynamic instability not well tested Limited usefulness in awake patients

arterial catheter. Cardiac output is calculated every 20 s using a recently upgraded algorithm. Multiplication of arterial pulsatility [standard deviation (SD) of the pressure wave over 20 s] and a constant (K), derived from the patient's specific vascular compliance, results in stroke volume, which is then multiplied by heart rate to calculate cardiac output. The specific vascular compliance is updated every minute and is based on age, height, gender, and weight and waveform characteristics. Unlike other pulse contour devices, the FloTrac/Vigileo does not require external calibration.[3]

Strictly speaking, the LiDCO monitor (LiDCO, Cambridge, UK) uses pulse power analysis rather than pulse contour analysis. It uses an algorithm based on the law of conservation of mass for continuous cardiac output calculation. The pulse power, rather than pulse contour, is measured. By using the assumption that net power has a linear relationship with net flow, an algorithm is used to calculate cardiac output. A standard arterial line only is required. The LiDCO is calibrated using lithium dilution (see below). This can be done centrally or peripherally.[4]

Cardiac output monitoring utilizing pulse contour analysis is one of the most extensively studied of all the minimally invasive monitoring systems. In general, they all show good agreement with cardiac output measurements made using a PAC. Despite this, the user should be aware of a number of sources of potential error, which may be more pronounced in some clinical settings.

All pulse contour analysis monitors rely on an optimal arterial signal. Over- or under-damped traces may lead to inaccurate cardiac output measurement. Arrhythmias, aortic regurgitation, and the use of an intra-aortic balloon pump all affect the pulse contour and have been shown to affect accuracy.[2] Changes in systemic vascular resistance may also lead to inaccuracies in cardiac output measurement.

The accuracy of these devices has been researched in a number of different settings. The most consistently used method of assessing their accuracy has been to measure their mean bias in comparison with another method of cardiac output measurement using the Bland–Altman method.

The PiCCO device has been extensively studied and compared with pulmonary artery thermodilution-derived cardiac output measurements. In 1999, Goedje and colleagues showed that pulse wave analysis-derived cardiac output measurements with the PiCCO device correlated well with those from the PAC in post-cardiac surgical patients. These patients had cardiac outputs ranging from 3.0 to 11.8 litre min^{-1} and systemic vascular resistances ranging from 252 to 2434 dyn s cm^{-5}. They showed a mean bias of 0.07 litre min^{-1} (2 SD 1.4 litre min^{-1}). This strong correlation remained even when significant variations in haemodynamics and vascular tone were present.[5]

The performance of the FloTrac/Vigileo system has also been researched in a range of clinical situations. To date, the data regarding the accuracy of the Vigileo system in comparison with both PACs and PiCCO devices have been conflicting. Despite software upgrades, a recent study of 21 critically ill patients in an intensive care unit showed underestimation of cardiac output by more than 2 litre min^{-1} in 41% of measurements.[6] However, the advantage of this device being less invasive than others may outweigh this reduction in reliability and it has been suggested that the FloTrac/Vigileo may be more useful for measuring trends than absolute values.

The LiDCO pulse pressure device correlates well with the PAC thermodilution technique. It has been studied in a range of settings,

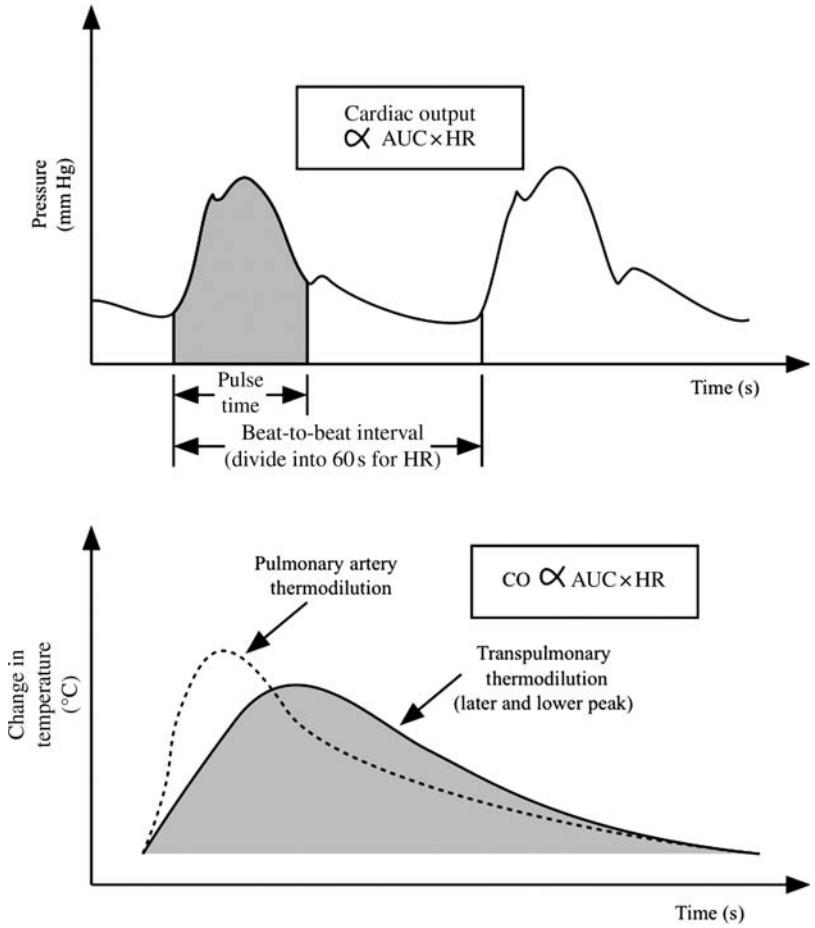

Fig 1 The PiCCO monitor (Pulsion Medical Systems) uses the area under the curve (AUC) of the pressure–time curve to calculate cardiac output. It is calibrated using the transpulmonary thermodilution method (HR, heart rate; CO, cardiac output).

including in patients with hyperdynamic circulations. Costa and colleagues[7] studied 23 patients in intensive care after liver transplantation and showed good agreement with the PAC thermodilution technique in patients with cardiac outputs of both <8 and >8 litre min^{-1}. In this study, the mean bias between cardiac outputs measure by pulse pressure analysis with the LiDCO device and intermittent PAC thermodilution measurements was 0.29 litre min^{-1} (2 SD 2.17 litre min^{-1}).[7]

Aortic Doppler

Cardiac output can be estimated using Doppler ultrasound to determine the flow of blood through the aorta. Most devices in current practice use a Doppler probe inserted into the oesophagus to measure descending aortic flow (Fig. 2). The USCOM device (Ultrasonic Cardiac Output Monitors, Sydney, Australia) is truly non-invasive and uses a probe placed suprasternally to measure flow through the aorta or on the left chest to measure transpulmonary flow.[8]

The volume of blood passing through the aortic valve over a given cardiac cycle is the stroke volume. Multiplying the stroke volume by the heart rate gives the cardiac output. Doppler ultrasound is used to measure the stroke volume and once an optimal flow profile has been obtained, the blood flow velocity is determined from the shift in frequency of red blood cells. This is done by the ultrasound processor using the Doppler equation:

$$V = (f_d \times c)/(2 \times f_0 \times \cos\theta)$$

where V is the velocity of blood, f_d the Doppler shift in frequency, c the speed of ultrasound in tissue (1540 m s^{-1}), f_0 the initial ultrasound frequency, and θ the angle of ultrasound beam in relation to the blood flow.

The velocity–time integral (VTI) is calculated from the area under the velocity–time curve and used as the stroke distance (Fig. 2). An estimate of aortic cross-sectional area (CSA) is taken either from a nomogram (height, weight, and age) or utilizing M-mode ultrasound. Cardiac output is then calculated using the equation: CO=CSA×VTI×HR. Finally, a correction factor must be used as the measurement utilizes the descending aorta, as only ~70% of cardiac output passes through this vessel.

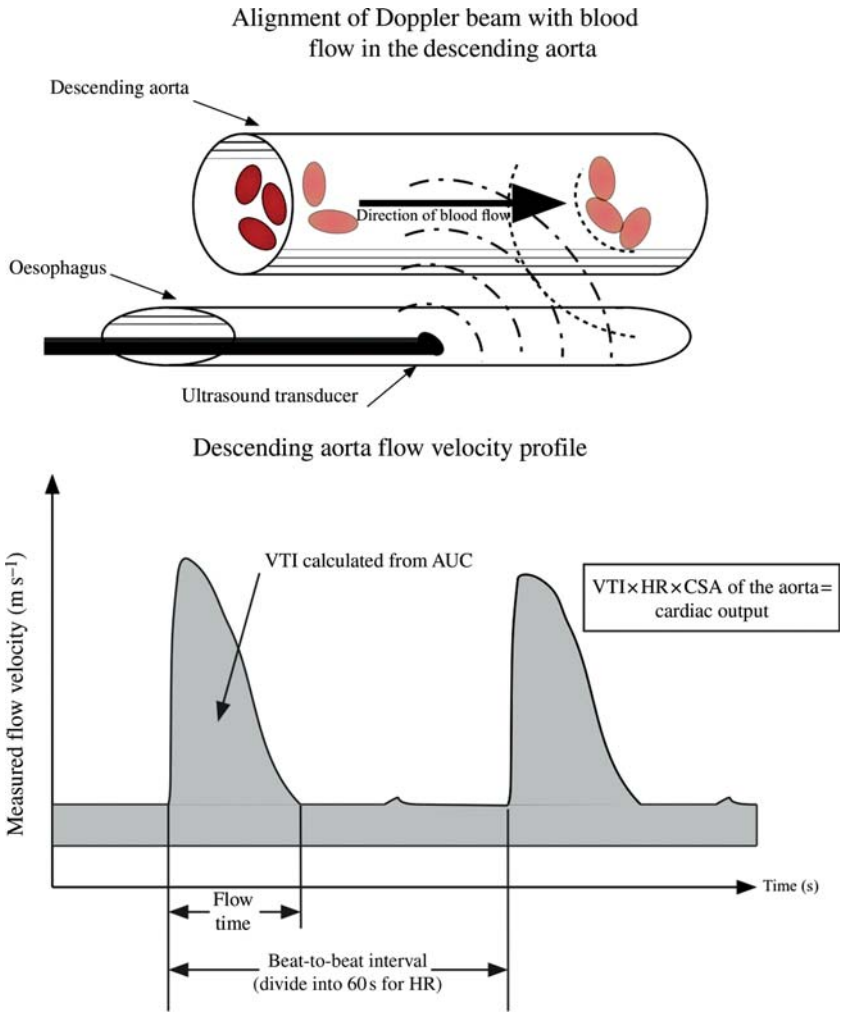

Fig 2 The oesophageal aortic Doppler probe is inserted into the oesophagus and manipulated to achieve the optimal velocity–time curve. The VTI is calculated from the area under the curve and then the cardiac output (CO) is calculated from the product of the VTI, heart rate (HR), and CSA of the aorta.

Estimation of CSA may be an important source of error for this method of cardiac output measurement. The use of a nomogram may introduce measurement error, especially as CSA will change with change in vascular tone and volume status.[1] The probe position is critical in reducing measurement error for both blood flow measurement and aortic cross-sectional measurement. Even small misalignments of the ultrasound beam with blood flow will lead to underestimation of flow when using the Doppler equation. Finally, the above equations assume laminar flow and any turbulent flow in the aorta will reduce measurement accuracy.

Cardiac output measurement by oesophageal Doppler has been extensively studied in a wide range of patients. A meta-analysis of 21 studies of critically ill patients in intensive care and operating departments showed a mean bias of 0.19 litre min^{-1} (range −0.69 to 2.00 litre min^{-1}) and an 86% clinical agreement between oesophageal Doppler and PAC thermodilution methods when measuring changes in cardiac output. However, the clinical agreement for absolute cardiac output measurements was only 52%.[9] The USCOM device has shown variable results, especially in low and high cardiac output states.[8]

Gas rebreathing

The partial gas rebreathing monitor utilizes the indirect Fick equation to determine cardiac output. A rebreathing apparatus is attached to the patient's tracheal tube and serial measurements are taken every 3 min. At steady state, the amount of CO_2 entering the lungs via the pulmonary artery is proportional to the cardiac output and equals the amount exiting the lungs via expiration and pulmonary veins. During 30 s of rebreathing, the amount entering does not change, but the amount eliminated by expiration decreases and the E'_{CO_2} increases in proportion to the cardiac output.[1,2]

Unlike pulse contour analysis monitors, this device is unable to give any additional haemodynamic values such as SVV. Tracheal

intubation is required as are fixed ventilator settings, which limits the number of patients in which this method can be used. It is also inaccurate with assisted spontaneous breathing patients. Partial rebreathing has been shown to be more accurate in less critically ill patients with normal alveolar gas exchange when compared with PAC thermodilution.[10] Severe chest trauma, significant intrapulmonary shunt, low minute ventilation, and high cardiac output may all reduce accuracy.[1] For these reasons, partial gas rebreathing is limited in its clinical applicability.

Transpulmonary thermodilution

Transpulmonary thermodilution is used to calibrate the pulse contour PiCCO monitor. It uses the same principles of thermodilution as PACs. The cold injectate is introduced into the superior vena cava via a CVC. An arterial line with a thermistor is placed in a major artery (femoral, axillary, or brachial) and the change in temperature of the blood is measured after the injectate has traversed the right heart, pulmonary circulation, and left heart. The change in temperature over time curve begins later and has a lower peak compared with the curve from a PAC (Fig. 1). The thermodilution equation is used to calculate the cardiac output. Other variables can also be measured, including global end-diastolic volume as a measure of preload and extra-vascular lung water as a measure of pulmonary oedema.[11]

A number of potential sources of error have been identified. Thermodilution measured via a PAC measures right heart cardiac output, whereas transpulmonary thermodilution measures left heart cardiac output. In the majority of patients, these are equal; however, the presence of an intra-cardiac or intra-pulmonary shunt will lead to cardiac output measurements which differ from the 'gold standard'. Indicator loss into the lungs, especially in patients with pulmonary oedema, has been suggested as a reason for poor correlation in some studies; however, it has been estimated that 96–97% of the indicator reaching the pulmonary artery is recovered in the aorta.[11] Indicator recirculation occurs when an increased amount of the cold injectate leaves the blood and enters the tissues, for example, in pulmonary oedema, and later re-enters the blood. Indicator recirculation will cause an abnormal prolongation of thermodilution curve and lead to underestimation of cardiac output.[11]

In general, transpulmonary thermodilution has shown good correlation with PAC thermodilution.[11] Goedje and colleagues[5] showed a very low mean bias of -0.29 litre min^{-1} (2 SD 1.31 litre min^{-1}) in their group of haemodynamically variable patients with a thermistor-tipped arterial catheter (PiCCO) inserted in the femoral artery.

Lithium dilution

Lithium dilution is an alternative indicator dilution method for measuring cardiac output. It is used to calibrate the LiDCO pulse contour device. This method uses 0.5–2 ml (maximum cumulative dose 20 ml) boluses of lithium chloride (0.15 mmol ml^{-1}) as the indicator. The lithium is injected via a central or peripheral venous line and measured via aspiration of blood through an arterial catheter, with an attached disposable electrode selective for lithium, at a constant rate of 4 ml min^{-1}. A correction factor is applied for serum sodium levels to determine baseline voltage. The change in voltage is electronically converted to plasma lithium concentration and the resulting lithium concentration *vs* time curve is used to calculate plasma flow. Plasma flow is converted to blood flow by dividing by 1 − PCV (packed cell volume),[3,11]

Lithium dilution shows good correlation with PAC thermodilution in normal and in hyperdynamic conditions, provided that there is no indicator loss and constant blood flow.[3,7] Mean bias between lithium dilution using a LiDCO device and thermodilution using a PAC has been shown to be 0.11 litre min^{-1} (2 SD 1.94 litre min^{-1}).[7] Reduced accuracy may be seen in patients who are on long-term lithium treatment and in the presence of non-depolarizing neuromuscular blocking agents.[1,11] The use of lithium dilution monitors is contraindicated in patients weighing <40 kg and those in the first trimester of pregnancy. Finally, concerns have been raised regarding the repeated drawing and discarding of blood (3–4 ml) over time in critically ill patients.[10]

Transthoracic electrical bioimpedance

One of the least invasive methods of measuring cardiac output, transthoracic electrical bioimpedance (TEB), measures the electrical resistance of the thorax to a high frequency, very low magnitude current. Six electrodes are placed on the patient and the resistance to current flowing from the outermost to innermost electrodes is measured. The bioimpedance is indirectly proportional to the content of thoracic fluid. Tissue fluid volume, pulmonary and venous blood, and the aortic blood volume all contribute to the TEB measurement. Changes in cardiac output will change the amount of aortic blood and will be reflected in a change in TEB.[1]

Stroke volume is calculated using the formula SV = VEPT × VET × EPCI, where VEPT is the volume of electrically participating tissue (calculated using gender, height, and weight), VET the ventricular ejection time taken from the R–R interval, and EPCI the ejection phase contractility index which is indirectly proportional to TEB.[1]

A number of problems have been identified with TEB in clinical practice. Interference by electrocautery (diathermy) may limit its use intraoperatively. The system is also very sensitive to movement and thus is unlikely to be of benefit in awake patients in critical care. Arrhythmias may lead to inaccuracy due to an irregular R–R interval. Initial studies in older devices showed inconsistent results in critically ill patients; however, newer second-generation devices show improved accuracy. They have been studied in cardiac surgical patients and show good correlation intraoperatively with a mean bias of -0.28 litre min^{-1}. Results were less promising in the immediate postoperative period, potentially due to the use of steel wires.[12] Despite this, further evidence is still needed to determine the accuracy of TEB devices in haemodynamically unstable patients.[1]

Summary

The ability to accurately measure cardiac output remains an integral part of diagnosing and managing critically ill patients. There are many minimally invasive devices currently available on the market designed to reduce the risks associated with the use of PAC. These devices have variable degrees of 'invasiveness', with some being only marginally less invasive than PACs.

Pulse contour devices with or without calibration are perhaps the most extensively studied and are widespread in use. In general, they show good correlation with PACs thermodilution-derived cardiac output measurements when their limitations are taken into account. Aortic Doppler ultrasound and gas rebreathing techniques have also been extensively studied and shown to give accurate results, but are limited to use in patients with tracheal tubes in place. Owing to their limitations, thoracic electrical bioimpedance monitors, while truly non-invasive, are not yet in widespread clinical use. Users should be aware of the limitations of each of these devices when interpreting measurements.

While there has been much research looking at the accuracy of these monitors in comparison with PACs, there is currently very little information in the literature regarding the effectiveness of these monitors in improving clinical outcomes. Until further work in this area has been conducted, it remains with individual clinicians and institutions to weigh up the risk–benefit ratio based on their patient mix, clinical skills, and resources.

Declaration of interest

None declared.

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Please see multiple choice questions 5–8.