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Physical principles that enable cardiac output monitoring

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Introduction

The primary homeostatic role of the cardiovascular system is to ensure that all the metabolic demands of the tissues are met. This is achieved by the delivery of oxygen and nutrients to the tissues with simultaneous removal of waste products of metabolism and cellular respiration. The delivery of oxygen to the tissues (DO_2) is dependent on cardiac output (CO) and arterial oxygen content (CaO_2) and can be calculated with the following equation:[1]

$$DO_2 = CaO_2 \times CO$$

CaO_2 in turn is dependent on the haemoglobin concentration (Hb), haemoglobin oxygen saturation (SpO_2), and partial pressure of oxygen (PO_2) in arterial blood and can be calculated with the following equation:[1]

$$CaO_2 = (1.39 \times Hb \times SpO_2 / 100) + (0.003 \times PO_2)$$

Continuous non-invasive monitoring of SpO_2 can be achieved with photoplethysmography using the modified Beer-Lambert's law, while PO_2 and Hb can be measured by blood gas analysis, all of which is beyond the scope of this review.[2] However, in the absence of anaemia and hypoxaemia, DO_2 is dependent on CO. CO is defined as the volume of blood ejected from the right or left ventricle per unit of time and is therefore the product of HR and stroke volume (SV):[3]

$$CO = HR \times SV$$

CO monitoring has become commonplace in the management of critically ill patients, those undergoing anaesthesia for major surgical procedures, and patients with limited cardiopulmonary reserve.[4] Numerous methods have been used to measure CO, all of which derive a value from an inference of SV or flow, both of which cannot be measured directly. For this reason, each of the CO monitoring techniques is coupled with potential sources of error and limitations to their use. Despite this, the clinical necessity for an ideal CO monitor is strongly asserted by a wealth of literature suggesting that therapies should be directed at improving DO_2 with goal-directed interventions, whilst liberal fluid therapies may be harmful.[5]

This review aims to explain the physical principles that enable each of the CO monitoring techniques. An appreciation of the physics behind these techniques is essential to understanding their strengths and limitations in clinical application.

The ideal CO monitor[6]

The ideal CO monitor should be:

- non-invasive or minimally invasive without the need for calibration;
- quick and easy to set up;
- safe to use;
- operator independent with minimal specialised training required;
- appropriate for use in a wide variety of clinical settings;
- appropriate for use in adult and paediatric patients;
- appropriate for use in intubated and non-intubated patients; and
- provide real-time, continuous, accurate and reproducible data that are reliable under different physiological conditions.

No single CO monitor is presently able to satisfy all the clinical criteria above.

Types of CO monitors[7]

Each CO monitoring technique can be classified as one of the following:

- Invasive techniques: These techniques make use of a central line and arterial line or a pulmonary artery flotation catheter (PAFC)/Swan-Ganz catheter. Invasive techniques include the Fick principle, dye dilution, lithium dilution, thermodilution, pressure waveform analysis, and pulse pressure contour analysis.
- Minimally invasive techniques: These techniques make use of transoesophageal ultrasound and the Doppler effect.
- Non-invasive techniques: These techniques include bioimpedance, bioreactance, and tonometry, which make use of the application of electrodes or cuffs to the surface of the body.

A variety of physical principles have been exploited in the calculation of CO using the above techniques. These principles will be elaborated on below.

Invasive techniques

Fick principle

The Fick principle, developed by Adolf Eugen Fick in the late 1800s, asserts that the total amount of a substance that is taken up or excreted by an organ is equivalent to the product of the blood flow through that organ and the difference in the arterial and venous concentrations of the substance in the blood supplying and leaving that organ respectively.[8] If one considers the lungs as the organ of interest, oxygen can be considered as a "substance" that is completely added, from the alveoli to the blood as it moves through the pulmonary circulation. Hence, if the pulmonary arterial oxygen content (CvO_2) as well as the systemic arterial oxygen content of blood (CaO_2) is known, the pulmonary blood flow (PBF) can be calculated provided the amount of oxygen taken up by the blood through the alveoli (VO_2) is known.

$$PBF = VO_2 / (CaO_2 - CvO_2)$$

Since the entire right ventricular CO passes through the pulmonary circulation, it can be assumed that the right ventricular CO is equal to the PBF. The pulmonary and systemic circulations are arranged in series. Consequently, it is assumed that the left ventricular CO is equivalent to the right ventricular CO and therefore, systemic CO can be calculated.

This method of CO monitoring requires the patient to breathe 100% oxygen from a closed system incorporating a carbon dioxide absorber and a spirometer that contains a known amount of oxygen over a period of one minute. Oxygen readily diffuses across the alveolar-capillary barrier along a concentration gradient. The rate of diffusion (V_{gas}) has been described by Fick's law[9] and is directly proportional to the partial pressure gradient ($P_1 - P_2$) across the barrier and the surface area available for diffusion (A), while it is inversely proportional to the thickness of the alveolar-capillary membrane (T).[9]

$$V_{gas} = D_{Constant} \times (A / T) \times (P_1 - P_2)$$

After one minute, the amount of oxygen drawn through the spirometer is measured as the VO_2. Further placement of a PAFC is required so that CvO_2 can be calculated from mixed venous blood gas analysis while CaO_2 is calculated from peripheral arterial blood gas analysis.

Dye dilution technique

This method of CO calculation requires the injection of a marker or dye (usually indocyanine green) into the blood.[2] The dye is injected into a central vein or the pulmonary artery through a central venous catheter or PAFC. The dye should mix completely with the blood, circulate, and remain in the intravascular space while being minimally metabolised. CO can be calculated if the change in concentration of the dye, at a point downstream of the initial injection point (usually a peripheral arterial line) can be measured and plotted against time.[2] An optical dye detector is usually utilised at the arterial line.

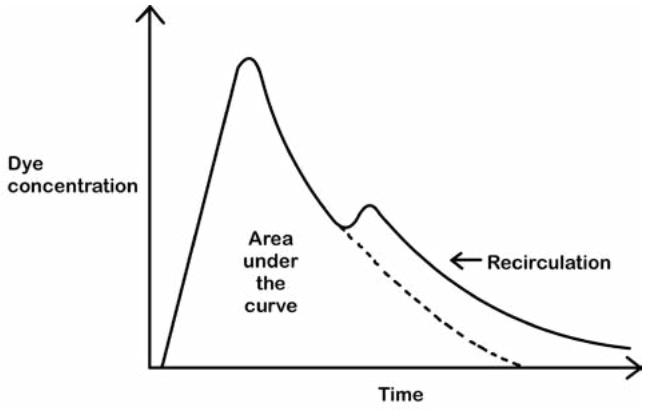

Figure 1: A graph depicting a washout curve

The resulting graph of dye concentration plotted against time is termed a washout curve with an initial peak concentration reached rapidly after dye injection with a subsequent exponential decrement in dye concentration as it is diluted and "washed out" (Figure 1).[2] The dye travels at a range of velocities from the point of injection towards the detector due to laminar flow.[2] This results in a slightly delayed, rounded peak to the plasma concentration at the detection site after injection.[2] Finally, the concentration of dye decays exponentially and predictably. However, some dye recirculates and will result in a second peak due to fast-moving dye. The rate at which dye is "washed out" is directly proportional to the flow rate (CO) and the concentration of dye injected.[2]

The Stewart-Hamilton equation can be used to calculate the CO based on the following mathematic principles:

- The volume of a dye or marker is equal to the mass of the dye divided by the concentration.
- If concentration is measured over time, the rate of change of volume (flow) can be calculated.
- Hence, CO (flow) can be calculated by dividing the mass (amount) of dye injected by the mean concentration of dye over time (area under the exponential decay curve) using the following formula:[2]
  - CO = amount of dye injected / area under the graph

Lithium chloride dilution technique

Calculating CO using lithium chloride as a marker makes use of the same dilution technique described in the dye dilution method above. Lithium chloride is an ion that is not normally present in plasma, so only a small dose of this marker is necessary to produce a reliable washout curve.[2] Secondly, the small dose of the marker reduces the risk of toxicity or an adverse reaction.[2] Lithium chloride is not a dye and a different method of concentration sampling is necessary at the arterial line. Following a bolus of lithium chloride injected into a central vein, the resulting lithium concentration-time curve is recorded by drawing a blood sample past an electrochemical lithium-ion detector attached to the patient's peripheral arterial line.[2] CO can be calculated from the known dose of lithium chloride injected and the area under the graph generated using the Stewart-Hamilton equation.[2]

Thermodilution technique

Dye and lithium chloride dilution techniques are confounded by the recirculation of the marker that needs to be excluded by the logarithmic transformation of concentration measurements and extrapolation of the exponential decay curve to the baseline of the graph. This confounder is avoided by making use of the thermodilution method of CO calculation. Traditionally, a PAFC is placed through the right internal jugular vein and advanced through the right atrium, tricuspid valve, right ventricle, and finally floated through the right ventricular outflow tract to settle in the pulmonary artery.[2] A measurement of 10–15 ml cold saline at 0 °C is injected into the proximal port, which is situated in the right ventricle. The change in blood temperature is measured at the distal port situated in the pulmonary artery and plotted over time, generating a thermal washout curve.[2]

Akin to both dye and lithium chloride dilution techniques, a modification of the Stewart-Hamilton equation is used to calculate CO if the volume of cold saline ($Vol_{injectate}$), temperature of the cold injectate ($T_{injectate}$), temperature of the blood (T_{blood}), average temperature measured by the thermistor (T_{mean}), and the times at which the temperature drop is first detected (T_1) and returns to normal (T_2), is known.[2]

$$CO = [Vol_{injectate} \times (T_{blood} - T_{injectate})] / [T_{mean} \times (T_2 - T_1)]$$

The temperature change is measured by two thermistors, one at the site of proximal injection and one downstream of the injection site. A thermistor is a temperature-sensitive resistor.[2] It is constructed from a semiconductor material with a resistance inversely proportional to temperature (negative temperature coefficient) and is known as a negative thermal conductivity thermistor.[2] They are small, cheap and exhibit excellent accuracy and rapid response times.[2] When connected to a Wheatstone bridge circuit, the resistance generated can be measured accurately. Furthermore, by passing a constant current (I) through the electrical circuit connected to the thermistor, a temperature change will result in a change in resistance (R) in the circuit, which in turn will result in a change in potential difference (V) across the circuit.[2] This change in potential difference allows electronic detection of the temperature change at the distal port on the PAFC.

The use of a PAFC can be avoided by the injection of cold injectate into a central venous catheter whilst measuring the resultant temperature change at a peripheral arterial line. This method is known as trans-pulmonary thermodilution and the resulting washout curve is less pronounced due to the circulation of injectate through both the right and left heart chambers as well as the pulmonary circulation before reaching the thermistor. The use of a PAFC to measure CO is still considered the gold standard, but its use has declined over the years. Whilst the PAC-Man study found no evidence of benefit or harm in its use in critically ill patients, its invasive nature, required technical skill and risk of complication have led to the use of less invasive CO monitors.[10]

Pressure waveform analysis and pulse pressure contour analysis

Pressure waveform analysis and pulse pressure contour analysis can be used to calculate CO from measurements obtained from a peripheral arterial catheter, based on theory first described by Erlanger and Hooker in 1904.[11] They theorised that CO was proportional to arterial pulse pressure.[12] Arterial pulse pressure is different from arterial blood flow. A German physiologist, Otto Frank, described the Windkessel phenomenon that explains the relationship between arterial pulse pressure and flow.[13] During systole, the contraction of the left ventricular musculature generates a pressure gradient between the left ventricle and the aorta. This pressure gradient (ΔP) drives the flow of blood (Q), which is proportional to the pressure exerted and the radius of the aorta (r) whilst being inversely proportional to the viscosity (η) of blood and the length (L) of the vasculature (Hagen-Poiseuille law).[14]

$$Q = (\Delta P \times \pi r^4) / (8 \times \eta \times L)$$

Some of the kinetic energy of the blood ejected into the aorta is converted into potential energy that is stored in the stretched elastic tissue of the vessel walls. During diastole, the aortic valve closes allowing the arterial pressure to fall, and the stored potential energy is converted back to kinetic energy during the elastic recoil of the vessel walls. This results in augmented diastolic blood flow and continual flow during diastole.

Pulse pressure contour analysis devices relate the shape of an arterial pressure wave to SV and systemic vascular resistance.[15] The steeper the upstroke of the arterial pressure waveform during systole, the greater the pressure and likely contractility, provided the aortic valve is not diseased.[15] The steeper the downstroke of the arterial waveform during diastole, the less elastic recoil and hence the lower the systemic vascular resistance.[15] The area under the pressure-time curve during the systolic part of the cardiac cycle is the mean arterial pressure during systole and is proportional to SV (Figure 2). While devices differ by making use of unique proprietary algorithms to calculate CO, they all aim to calculate SV.[15] SV is then multiplied by HR to determine CO.

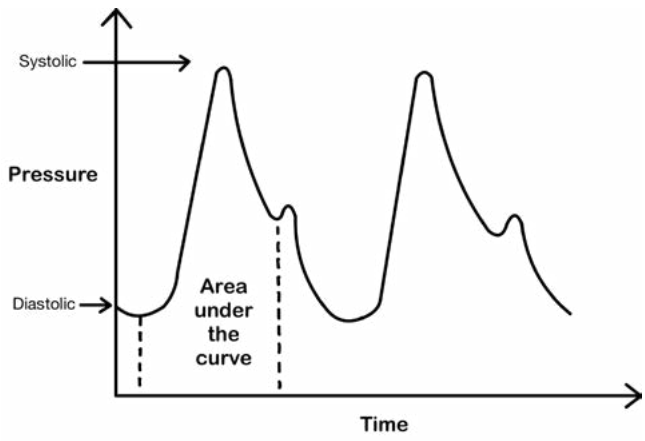

Figure 2: A graph depicting an arterial pulse pressure waveform

The use of arterial waveform analysis in the calculation of CO is made possible by signal transduction. A transducer is an electronic device that converts one form of energy to another.[2] In this instance, an arterial cannula is connected to a pressure transducer via a fluid-filled catheter. The fluid-filled catheter is pressurised, relatively inelastic and heparinised to ensure that it does not clot at the tip. This allows arterial pressure waves to be transmitted and reflected through the cannula to the transducer. The pressure transducer consists of four thin wafers of semiconductor material connected to form a Wheatstone bridge resistor network, bound to a membrane.[2] Small movements of the heparin-saline solution in the catheter are induced by the arterial pressure wave. This causes the pressure transducer diaphragm to oscillate. The oscillation causes a change in resistance and subsequent change in voltage generated across the Wheatstone bridge resistor network whilst a constant current is applied (Ohm's law).[2] This change in potential difference is displayed electronically as a change in gauge pressure relative to atmospheric pressure.[2] The pressure reading is continuously plotted against time to generate a pressure-time curve. An optimal arterial signal is required for the pulse pressure contour analysis calculation of CO.

Minimally invasive techniques

Ultrasound, aortic Doppler and echocardiography

Ultrasound and the Doppler principle can be used to calculate CO without the need for intravenous cannulation. This modality makes use of a lubricated ultrasound probe, placed either orally and advanced into the oesophagus in a minimally invasive approach, or suprasternal in a non-invasive approach.[7] The ultrasound probe contains a ceramic piezoelectric crystal that emits sound waves into the tissues at a frequency > 20 kHz when an alternating current is applied to it.[2] This causes the crystal to expand and contract repetitively, producing vibrations and emitting ultrasonic waves.[2] The high-frequency waves penetrate tissue while a variable number of waves are reflected at interfaces between tissues of differing densities.[2] Some waves travel further before being reflected and consequently return to the probe later. The kinetic energy of the reflected waves is converted back into an electrical signal when they return to the probe.[2] This electric signal is displayed on a monitor to provide a graphical representation of tissue at differing distances from the probe. If tissue is moving towards or away from the ultrasound probe, the reflected waves undergo a frequency shift before returning to the probe.[2] The change in frequency is termed the Doppler effect and can be used to calculate the velocity (Vel) of the moving tissue if the initial ultrasound frequency (F_o), Doppler shifted frequency (F_d), speed of ultrasound in tissue (C = 1 540 $m.s^{-1}$), and the angle of the ultrasound beam relative to the tissue (θ) is known, using the Doppler equation:[2]

$$Vel = (F_d \times C) / (2 \times F_o \times Cos\ \theta)$$

To calculate CO with the use of ultrasound, one must exploit its utility in measuring distances and velocities. The volume of a cylinder can be calculated if its length and cross-sectional area are known. Similarly, if the radius of the descending aorta is known (determined from nomograms based on the patient's height, weight and age, or by measurement on ultrasonography), and the velocity of red blood cells in the descending aorta is known (determined from the velocity-time integral = area under the velocity-time curve generated using the Doppler method above), one can calculate the volume of a cylinder of blood that moves through the aorta in one cardiac cycle (SV).[16] This, in turn, can be multiplied by HR to determine CO. Lastly, a correction factor needs to be applied to estimate the total CO since approximately 70% of the CO passes through the descending aorta.[16]

Non-invasive techniques

Bioimpedance

Thoracic electrical bioimpedance is one of the least invasive methods of calculating CO. This modality aims to estimate CO by exploiting Ohm's law. Ohm's law describes the relationship between potential difference (V), current flow (I), and resistance (R) in an electrical circuit as follows:[2]

$$V = I \times R$$

Impendence (Z) is the measure of resistance to electrical current when an alternating current is applied to the circuit.[7] Thus, bioimpedance is the measure of resistance offered by organic tissue to the conduction of an alternating current. Exploiting these principles enables the calculation of CO when a high frequency (alternating), low amplitude current is applied across the thorax. The thorax contains tissues with differing ionic concentrations and hence offers various impedances to current flow. Vascular-rich organs and blood vessels have high ionic content and are classified as low-impedance tissues (150 Ω/cm), whilst less vascular tissues are classified as high-impedance tissues (1 274 Ω/cm).[7] During systole, the aorta distends and the relative proportion of low-impedance tissue increases. By applying a constant alternating current to the thorax via electrodes placed on either side of the neck and the diaphragm, a change in impedance during the cardiac cycle can be inferred by a change in the potential difference between the electrodes.[7] SV is then calculated using the following formula:[12]

$$SV = VEPT \times VET \times EPCI$$

Where VEPT is the thoracic volume, VET is the ventricular ejection time determined from the R-R interval on the electrocardiogram, and EPCI is the ejection phase contractility index resulting from the product of bioimpedance and total fluid conductivity.[12] SV, in turn, is multiplied by HR to calculate CO.

Bioreactance

Bioreactance is an improvement on bioimpedance modalities because it reduces error due to movement artefacts, variance in patient body habitus, and inconsistencies in electrode positioning.[7] It relies solely on changes in phase shifts of alternating currents and voltages.[7]

Conclusion

Numerous CO modalities exist, each exploiting physical principles to aid the clinician in the perioperative management of patients. Their use as an adjunct to traditional clinical patient assessment may aid in the reduction of morbidity and mortality. At present, no ideal CO monitor exists, each has their strengths and limitations. An understanding of the physical principles will ensure appropriate interpretation of the data represented. This in turn will provide additional value in guiding appropriate therapeutic strategies.

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References

1. Dunn J-OC, Mythen MG, Grocott MP. Physiology of oxygen transport. BJA Educ. 2016;16(10):341-8. https://doi.org/10.1093/bjaed/mkw012.
2. Middleton B, Phillips J, Thomas R, Stacey S. Physics in anaesthesia. Oxfordshire: Scion Publishing, Ltd; 2012.
3. Kam P, Power I. Principles of physiology for the anaesthetist. 2nd ed. CRC Press; 2008.
4. Pinsky MR. Hemodynamic evaluation and monitoring in the ICU. Chest. 2007;132(6):2020-9. https://doi.org/10.1378/chest.07-0073.
5. Lees N, Hamilton M, Rhodes A. Clinical review: goal-directed therapy in high risk surgical patients. Crit Care. 2009;13(5):231. https://doi.org/10.1186/cc8039.
6. Geerts BF, Aarts LP, Jansen JR. Methods in pharmacology: measurement of cardiac output. Br J Clin Pharmacol. 2011;71(3):316-30. https://doi.org/10.1111/j.1365-2125.2010.03798.x.
7. Betteridge N, Armstrong F. Cardiac output monitoring. Anaesth Intensive Care Med. 2022;23(2):101-10. https://doi.org/10.1016/j.mpaic.2021.10.018.
8. Fick A. On the measurement of blood mass in the heart ventricles. Sitzber Physik Med Ges Würzburg. 1870;36:16-28. German.
9. Fick A. V. on liquid diffusion. Lond Edinb Dublin Philos Mag J Sci. 1855;10(63):30-9. https://doi.org/10.1080/14786445508641925.
10. Harvey S, Harrison DA, Singer M, et al. Assessment of the clinical effectiveness of pulmonary artery catheters in management of patients in intensive care (PAC-Man): a randomised controlled trial. Lancet. 2005;366(9484):472-7. https://doi.org/10.1016/S0140-6736(05)67061-4.
11. Erlanger J. An experimental study of blood-pressure and of pulse-pressure in man. Bull Johns Hopkins Hosp. 1904;12:145-378.
12. Funk DJ, Moretti EW, Gan TJ. Minimally invasive cardiac output monitoring in the perioperative setting. Anesth Analg. 2009;108(3):887-97. https://doi.org/10.1213/ane.0b013e31818ffd99.
13. Frank O. The basic shape of the arterial pulse. First treatise: mathematical analysis. J Mol Cell Cardiol. 1990;22(3):255-77. https://doi.org/10.1016/0022-2828(90)91460-O.
14. Pfitzner J. Poiseuille and his law. Anaesthesia. 1976;31(2):273-5. https://doi.org/10.1111/j.1365-2044.1976.tb11804.x.
15. Saugel B, Kouz K, Scheeren TWL, et al. Cardiac output estimation using pulse wave analysis-physiology, algorithms, and technologies: a narrative review. Br J Anaesth. 2021;126(1):67-76. https://doi.org/10.1016/j.bja.2020.09.049.
16. Drummond KE, Murphy E. Minimally invasive cardiac output monitors. BJA Educ. 2011;12(1):5-10. https://doi.org/10.1093/bjaceaccp/mkr044.