Determination of cardiac output

Cardiac output is **the amount of blood that the ventricle pumps per unit time (minute)**. This is actually the **flow of blood through the riverbed**. Except in exceptional situations (short-term unsteady state) *both chambers* always pump practically the *same volume of* blood. It is also true that the *same amount of blood* flows (in the absence of a pathological short circuit) through both pulmonary *and systemic circulations*.

A slight exception to this rule is the so-called **physiological short circuit:**

- The portion of blood from the systemic circulation that nourishes the distal bronchi returns to the heart through the pulmonary veins, along with the oxygenated blood flowing from the alveoli. In fact, it bypasses the pulmonary circulation.

More specifically, we should say that the amount of blood flowing through the left heart and the systemic circulation is about 1-3% higher than the amount flowing through the right heart and the pulmonary circulation.

The resting value of cardiac output is about **4-8 l.min^{-1}**. A simple equation applies to cardiac output:

$$SV = TO \times SF \qquad (1)$$

SV - cardiac output, TO - heart rate, SF - heart rate. If we substitute the average physiological values, we get (after converting *ml* to *l*) eg.:

4,9 *l/min* = 70 *ml/beat* x 70 *bpm*

If necessary, cardiac output can increase up to five times compared to the resting value, mainly due to an increase in heart rate (it can rise up to 180 - 220 beats / min) but initially also an increase in heart rate. Resting pulse volume depends on the phase of the respiratory cycle, as changes in intrathoracic pressure during respiration affect venous return within the chest. Of course, the variability of heart rate also causes some variability in cardiac output - for some methods, it is recommended that cardiac output be measured at the end of the expiration period (or at another fixed point in the respiratory cycle).

Cardiac output depends significantly on body size. However, it depends (like many other physiological parameters) on the size of the body surface area rather than increasing in direct proportion to body weight. In other words, in obese people of a given height, weight increases faster than body surface area and cardiac output. In order to get rid of the dependence on the size of the body surface and to be able to evaluate the pure working ability of the heart, we introduce a **cardiac index** (cardiac index − CI) by the relation:

$$CI = \frac{SV}{BS} \qquad (2)$$

where BS is the body surface area, calculated after substituting the height and weight of the individual using one of the body surface area calculation formulas. Nowadays, the cardiac index is usually calculated automatically using software-integrated relationships evaluating measurement results, such as the thermodilution method. **Cardiac index decreases in heart failure and failure** increases in conditions associated with hyperkinetic circulation (eg fever, anemia, arteriovenous shunts, avitaminosis B1, etc.).

In the case of heart failure, another parameter we evaluate is the ejection fraction.

To date, there is no method for determining cardiac output and cardiac index that is generally considered the gold standard. The available methods are either inaccurate or invasive or expensive (or combine several disadvantages).

The determination of cardiac output is performed by the following methods:

- Doppler echocardiography.
- Dilution methods and especially thermodilution method.
- Fick's principle and traditional method.
- Non-invasive modification of Fick's principle.
- Other.

Doppler echocardiography

We use relation (1) to calculate cardiac output by Doppler echocardiography. We are based on knowledge of heart rate and heart rate calculation. We determine heart rate by adding up all the volume that has flowed through the heart valve during one heartbeat. The volume flowed (dV) at each instant is equal to the duration of the instant (dt) times the blood velocity (v) times the cross-sectional area of the valve (S). The total heart rate is then the sum of the volumes flown during all moments of one heart revolution. Expressed by the formula:

$$dV = S \times v \times dt = \frac{\pi d^2}{4} v\, dt \qquad (3)$$

TO = sum of all *dV*s during one pulse

where d is the diameter of the valve (determined echocardiographically) and v is the blood velocity at a given moment determined on the basis of the Doppler effect by Doppler echocardiography. The Doppler effect is manifested by the fact that the frequency of ultrasound increases when reflected from the blood cells flowing towards the probe and, conversely, the frequency decreases when reflected from the blood cells flowing towards the probe. More detailed information about the Doppler effect can be found here.

For mathematicians: This statement means that the heart rate is the time integral of the flow of the aortic valve during systole and the mitral valve during diastole.

Doppler echocardiography is non-invasive, inexpensive, and relatively accurate. However, it requires an experienced echocardiographer.

Dilution method and the thermodilution method

In the classical method (according to Stewart-Hamilton), a known amount of indicator (dye, easily determinable substance, radioisotope) is injected into the venous system. Then the time course of its concentration at one place of the arterial bed is determined. It is not necessary to know the distance between the two places. The principle of the method is (simply) based on the following fact: The higher the cardiac output, the higher the blood velocity in the riverbed and the faster the indicator flows through the place of measurement of its concentration. The indicator must be well tolerated or degraded by the body and must not be toxic. An example of a contrast agent is, for example, a lithium salt solution (e.g. LiCl, lithium chloride); the concentration of lithium ions Li + is then measured.

For Mathematicians: Cardiac output is calculated based on the following equation:

$$SV = \frac{n_{ind}}{\int_0^\infty c_{ind}\, dt} \qquad (4)$$

where SV is the cardiac output, n_{ind} is the total amount of injected indicator and c_{ind} is the concentration of the indicator at the measuring point (eg in mol / l) at a certain point in time of length dt. Instead of substance amount and substance concentration, we could alternatively use weight and mass concentration (in kg / l) in the formula. Integration from zero to infinity in practice means that we measure concentration from the moment the indicator is applied until the first time! flows through the measurement site (therefore its concentration drops back to zero for the first time). The above formula can be understood as follows: If there was no dilution between the application of the substance and the measurement of its concentration "downstream", then (as already mentioned) it would be the case that the higher the cardiac output, the faster the dye flows around places of measurement. Therefore, in such a case, the value of the integral will be smaller (the sum of dtis smaller) and therefore we also get a higher cardiac output (divided by the integral, which has decreased). If the indicator is mixed and diluted in the longitudinal direction between application and measurement (which will definitely happen), then the result will not be affected. During dilution, the concentration (c_{ind}) decreases , but at the same time the time dt during which the dye flows through the measuring point is extended - the value of the integral does not change. It is also possible to dilute the indicator "in width" by the fact that, for example, the overall cross-section of the riverbed is higher at the measuring point than at the application site. Think about why even in this case the value of the integral (or one integration step) does not change. Hint: As the cross section increases, the flow rate decreases.

Thermodilution method

A free modification of the classical method is the **thermodilution method** of SV measurement using a **Swan-Ganz catheter**. Here, the indicator applied is heat (more precisely cold), as a bolus (for example 20 ml) of saline at 0 ° C is administered. The temperature is measured by a sensor located at the end of the catheter. This sensor is located at a known distance downstream (for example 18 cm) from the application site.

Unlike the classical method, it is necessary to know exactly the distance of the application site from the measurement site. Unlike other indicators, heat (cold) "runs" freely out of the vascular bed (the area around the vessel also cools down), and the equation valid for the classical dilution method (5) cannot therefore be used without any reservations.

The measurement principle is as follows: If the cardiac output is higher, then the cold will flow to the measurement site faster and will be less diluted (there will be a sharper drop in temperature and a subsequent sharper rise). Conversely, if the cardiac output is low, then it will take longer for the cold to reach the measurement site, and after this time the cold will be more diluted (there will be a slight decrease in temperature and the temperature will gradually return to normal). Calibration (finding out which temperature curve corresponds to a given cardiac output) of the method was performed by simultaneous measurements using other methods. Thus, a simultaneous measurement was used with another method (which is more difficult to perform / invasive, and therefore the thermodilution method is chosen instead), which allows to accurately determine the given cardiac output, while the corresponding values obtained by the simultaneous execution of the thermodilution method were assigned to the determined values, thus creating so-called thermodilution curves, which allow to assign specific values of cardiac output to the values determined by the thermodilution method. The thermodilution method of measurement can be

considered as **invasive** *to the extent that it requires the presence of a* ***Swan-Ganz catheter*** in the right heart compartment and the lung. However, it is not in itself an indication for catheterization and is used primarily where a catheter has been introduced for other reasons, primarily for measuring pressure. The accuracy of the method is not ideal, so several measurements (cold bolus application) are used in succession and the result is averaged.

Fick's principle and Fick's classical method

Note: In this section, blood that flows into the lungs through the pulmonary arteries will be referred to as venous. Blood that flows into the pulmonary veins and then into the systemic arteries will be referred to as arterial.

Fick's principle is a simple application of the law of conservation of matter. The amount of oxygen that flows into the lungs in the venous blood plus the amount that is consumed from the breathed air must equal the amount of O_2 that flows out in the arterial blood. In other words **The difference between the (substance) amount of oxygen that flows out of the lungs in the arterial blood in one minute and the amount that flows in the venous blood is equal to the amount of O_2 that is consumed from the breathed air in one minute** . Expressed by the equation:

$$Q_{O_2Art} - Q_{O_2Ven} = Q_{O_2Air} \qquad (5)$$

where Q_{O2Art} is the amount of oxygen discharged in the arterial blood in (mol / min), Q_{O2Ven} represents the inflow in the venous blood and Q_{O2Air} air supply.

The inflow and outflow of O2 through the blood can be expressed as the product of cardiac output and the substance oxygen concentration in the venous or arterial blood (in mol / l):

$$Q_{O_2Ven} = SV \times c_{O_2Ven} \qquad (6)$$

$$Q_{O_2Art} = SV \times c_{O_2Art} \qquad (7)$$

Substituting into relation (5) we get:

$$SV \times (c_{O_2Art} - c_{O_2Ven}) = Q_{O_2Air} \qquad (8)$$

So cardiac output can be determined as:

$$SV = \frac{Q_{O_2Air}}{c_{O_2Art} - c_{O_2Ven}} \qquad (9)$$

In this derivation, we used inflows and outflows of the amount of oxygen (in mol / min). Alternatively, we could use oxygen mass flows (in g / min). Gaseous oxygen volume flows (at constant pressure and temperature) are also sometimes used. Although the volume of oxygen exhaled from the breathed gas mixture is easily expressed in ml / min, the inflow and outflow of oxygen in the blood must be expressed somewhat in a break in ml O_2 / min. By this expression is meant what would be the volume of oxygen that flowed in and out of the blood if this oxygen were in the gaseous state (at constant pressure and temperature).

Practical implementation of the classical method: The amount of oxygen consumed from the breathed mixture in one minute (Q_{O2Air}) can be measured by the patient breathing into a closed bag, where the exhaled CO_2 is absorbed . Oxygen consumption from this gaseous mixture results in a decrease in volume, which is converted to the corresponding exhaled amount of O_2. The oxygen concentration in the mixed venous blood flowing into the lungs is determined by measuring the pO_2 in the mixed venous blood - therefore **catheterization of the central veins with a Swan-Ganz catheter is necessary**. The concentration of O_2 in arterial blood flowing from the lungs is determined **by taking arterial blood** and measuring pO_2. Although this classical method is relatively accurate, it is rarely used for its invasiveness.

Modern noninvasive Fick modification methods

Fick's principle (law of conservation of matter) can also be applied to substances other than oxygen. For example, for CO_2 the amount of venous blood flowing minus the amount of arterial blood flowing out is equal to the amount exhaled by the air. Or$Q_{CO_2Ven} - Q_{CO_2Art} = Q_{CO_2Air} \qquad (10)$

By adjustment we get a relation for cardiac output SV based on CO_2 measurement:

$$SV = \frac{Q_{CO_2Air}}{c_{CO_2Ven} - c_{CO_2Art}} \qquad (11)$$

Non-invasive measurement of arterial blood gases: To calculate the arterial blood gas concentration, we need to measure arterial pO_2 or arterial pCO_2. The so-called **“end –tidal“ pO_2 and end-tidal pCO_2** offer an interesting way to measure these pressures non-invasively. We measure pO_2, resp. pCO_2 of exhaled air at the end of deep exhalation. It is assumed that this air already comes all from the alveoli and therefore has the same pO_2

and pCO_2 like blood flowing from these alveoli. This procedure avoids the need for arterial blood collection. Unfortunately, the method fails in the presence of poorly ventilated areas of the lungs (atelectasis and obstruction), which can, of course, in extreme cases progress to pathological pulmonary shunts (eg ARDS).

Non-invasive measurement of venous blood gases: A method called **"rebreathing"** can be used, especially for measuring venous pCO_2. The method consists of rebreathing repeatedly into the gas bag until the pCO_2, stabilizes in the bag , but for a maximum of 45 seconds. This steady state arises when the pCO_2 in the venous blood and the pCO_2 in the sac equal. By measuring pCO_2 in the bag, we also measure pCO_2 in venous blood. This procedure avoids the need for annoying central vein catheterization.

Use of gases other than O_2 and CO_2: Fick's principle has been modified to use more exotic gases than oxygen and carbon dioxide. The measurement is performed by the patient starting to breathe a mixture that contains the substance. The partial pressure of this substance in the arterial blood is then measured. The advantage of this method is that when using a gas that does not normally occur in the air, the venous inflow of this substance before the start of the measurement is zero. Equation (5) is thus reduced to:

$$Q_{exot\ Art} = Q_{exot\ Air} \qquad (12)$$

and cardiac output after adjustment is calculated as:

$$SV = \frac{Q_{exot\ Air}}{c_{exot\ Art}} \qquad (13)$$

where $c_{exot\ Art}$ is the (substance) concentration of exotic gas in arterial blood (in mol / l), $Q_{exot\ Air}$ rate of uptake of this gas from the breathed mixture (in mol / min). This method also circumvents the need for central vein catheterization.

Summary: Non-invasive or low-invasive methods of measuring cardiac output based on the use of the Fick principle may become an accurate and inexpensive method of measuring cardiac output in the future. Methods based on inhalation of gases other than O_2 and CO_2 may be used. The possible use of oxygen and carbon dioxide has so far encountered problems with the accuracy of the conversion of partial pressures to concentrations, where it also depends on the effect of pH, the interaction of both gases in binding to hemoglobin, etc. Lung inhomogeneity can also cause problems.

Other methods

Magnetic resonance: The resonant properties of protons in the nucleus change depending on the velocity. Magnetic resonance imaging can thus be used as an *accurate* way to measure aortic flow. The method is *expensive*, it is used only experimentally.

Mathematical analysis of the pulse wave: The shape and amplitude of the pulse wave (the way the pressure in the artery changes from systolic to diastolic and back) depends on the cardiac output. The pulse wave is measured using either a conventional inflatable cuff or a sensor that sticks to the skin at the site of the artery. By mathematical analysis of this wave, the value of cardiac output can be obtained. The problem is that the shape of the pulse wave also depends very much on the properties of the arteries. For example, in the elderly, where the elasticity of the aorta and its elastic effect are lost, systolic blood pressure typically increases, but diastolic blood pressure remains normal. The method might be useful after calibrating a person (for the properties of their arteries) using another method for continuous monitoring of cardiac output.

Chest impedance (electrical resistance) measurement: Chest electrical resistance can be measured using several chest electrodes. Resistance changes during the heart revolution due to changes in the volume of blood in the heart and can therefore be used in the calculation of heart rate and subsequent cardiac output. The method is cheap and non-invasive, but unfortunately inaccurate.