Principles of pressure transducer function and sources of error in clinical use

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Abstract
The invasive measurement of physiological pressures is a common requirement in anaesthesia and intensive care medicine. From arterial blood pressure to intracranial pressure, these calculated variables give a swift graphical and numerical representation of a patient's current physiological status. This allows us to respond rapidly to conditions outside our preferred parameters and to carefully titrate treatments to target effects. These systems are, however, not infallible. An understanding of the principles of their function will promote appropriate use and an ability to recognize and react to sources of error. This article aims to furnish the reader with this level of understanding in order to inform their academic and clinical practice.

Keywords Calibration; damping; drift; energy; force; harmonics; natural frequency; pressure; strain gauge; Wheatstone bridge

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Learning objectives

After reading this article, you should be able to:
- define pressure and force in physical terms
- list commonly measured pressures that may require a transducer in anaesthetic practice
- draw a model of the components of an invasive arterial pressure measurement system and describe the ideal properties of each component
- explain the principles of the strain gauge and illustrate a Wheatstone bridge circuit
- outline the concepts of natural frequency, harmonics and Fourier analysis
- discuss the common sources of error in monitoring devices, how these are classified and what can be done to minimize their impact

Pressure, force and area

Pressure (P) is the force (F) exerted on an object by something in contact with it. It is defined as force per unit area (A):

$$P = F/A$$

Force is the physical ability to cause a change in the velocity of an object with mass (m) – i.e. acceleration (a). This is given by Newton's second law of motion: $F = ma$. Force is measured in Newtons (N). Area is measured in square metres (m^2); therefore pressure can be measured in N/m^2. The SI unit of pressure is the Pascal (Pa), which is equal to one N/m^2. Numerous other units of pressure are seen in anaesthetic practice and often relate to the original methods of pressure measurement:
- Liquid manometry – pressure is given as the height of the column of a given liquid and is equal to the product of column height, liquid density and gravitational force. This gives rise to cmH_2O, mmHg. Seen in mercury sphygmomanometers.
- Aneroid gauge – literally 'without air', a method of measurement reliant on the deforming properties of gas pressure on a system, e.g. Bourdon gauge – a coiled tube which is flattened out at higher pressures, manipulating a needle on a measurement dial. Seen on gas cylinders. Absolute pressure is the sum of the gauge pressure and atmospheric pressure.[1]

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Pressure measurement in anaesthesia

It is common (and often mandated) to measure certain biological pressures during anaesthesia and critical care admissions. Whilst non-invasive techniques, e.g. for the measurement of blood pressure, are sufficient in many cases, it is sometimes necessary to invasively monitor physiological pressure variables, commonly including:
- airway pressures
- non-invasive blood pressure
- invasive arterial pressure
- central venous pressure
- pulmonary arterial pressure
- intracranial pressure
- intra-abdominal pressure
- limb compartment pressure
- endotracheal tube cuff pressure
- uterine pressure/contraction as tocodynamometer component of cardiotocography.

Invasive arterial pressure measurement serves as a good example to illustrate the function of pressure transducers and the sources of error in their clinical use.[1,2]

Blood pressure measurement

Blood pressure is the force exerted by blood on the walls of the vessels which contain it. It has two components:
- static component consisting of the volume, mass and density of the blood
- dynamic component from the pulsatile nature of blood flow.

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Blood pressure is often formulaically represented as the product of cardiac output and systemic vascular resistance. It is the complex interplay between the contractility of the ventricles and the resistance to flow brought about by blood viscosity and vessel wall calibre. Furthermore, systolic, diastolic and mean arterial pressure elements can all be determined to varying degrees of accuracy by invasive and non-invasive techniques. It is no surprise, then, that the waveform produced during invasive arterial pressure monitoring is complex in itself and can be employed to derive much useful information about a patient's cardiovascular status.[1]

Transducers

A transducer is a device which converts energy from one form into another. Examples include:

- microphones (sound to electrical energy)
- radio antennae (radio waves to electrical energy)
- audio speakers (electrical energy to sound)
- solar panels (light energy to electrical energy).

The electromanometer measures invasive pressures by use of a pressure transducer. This converts mechanical energy to electrical energy, allowing an electrical signal to be transmitted to a processing unit before being displayed numerically or graphically for interpretation by the user. The method of converting mechanical to electrical energy is via a strain gauge. This gauge can sit at the tip of a catheter situated in a vessel or space whose pressure is to be measured. For example, the intracranial pressure monitors ('bolts') employed in neurosurgery consist of a strain gauge tipped catheter attached to a display box. Once this catheter is inserted, it is not possible to recalibrate the system as the gauge is intracranial. Errors in its function therefore require complete revision. It is possible to use a similar arrangement for intravascular devices, but more commonly the transducers are situated externally allowing easy access. This can be demonstrated when we examine the structure of an invasive arterial pressure set.

The main components shown in Figure 1 are:

- A: fluid bag under pressure of 300 mmHg, usually saline or heparinized saline
- B: pressure transducer with strain gauge
- C: flush mechanism to allow manual flush of catheter (constant 3–4 ml/hour fluid flushes the catheter to prevent thrombus formation and blockage)
- D: three-way tap to allow access for aspiration of blood from catheter and manual flush with syringe
- E: fluid filled line connecting cannula to transducer
- F: arterial cannula
- G: wire transmitting transducer signal to processing unit
- H: display module to numerically and graphically represent the processed signal.

As will be explored later, errors can occur anywhere along this system.[1–3]

Strain gauge and Wheatstone bridge

The continuous column of fluid in this system means that the arterial pressure waveform can be transmitted to the external transducer relatively unchanged. The fluid column reaches a flexible diaphragm which interfaces with the transducer unit. This diaphragm is displaced by oscillations in the fluid column caused by changes in arterial pressure. The adjacent strain gauge is, in turn, distorted by the movement of the diaphragm. Distortion of the resistor wires of the gauge causes a change in their resistance.

Resistance is a measure of the difficulty of passing an electrical current through an object. According to Ohm's Law, the potential difference (or voltage (V)) across a conductor is proportional to the current (I) passing through it:

$$V \propto I$$

The constant of proportionality used to produce the familiar equation is resistance (R)

$$V = IR$$

Therefore:

$$R = V/I$$

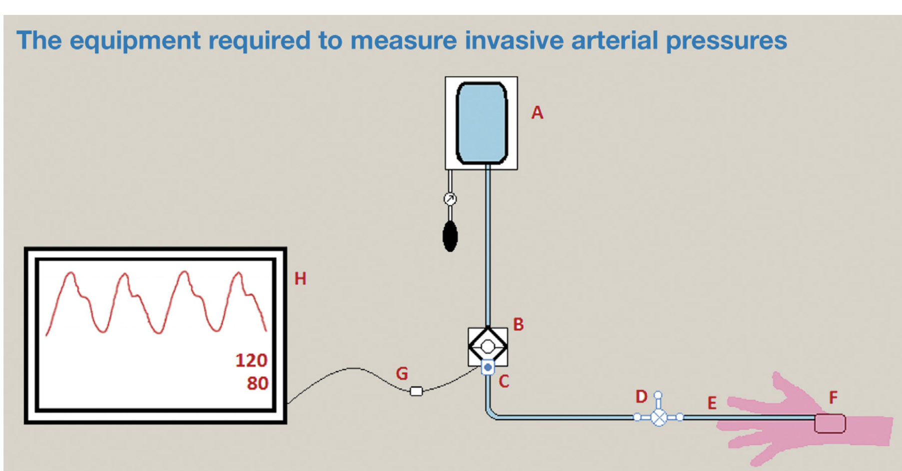

Figure 1

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Resistance is related to the material of the object, but also to its physical size. As the length (L) of a wire increases, resistance also increases — they are directly proportional:

$$R \propto L$$

As the cross-sectional area (A) increases, however, the resistance decreases — they are inversely proportional:

$$R \propto 1/A$$

Combining these two conditions gives:

$$R \propto L/A$$

Resistivity is the electrical resistive property of a material regardless of its dimensions. It is also used as the constant of proportionality (ρ) for the above resistance formula, producing the equation:

$$R = \rho L/A$$

As can be seen from these conditions, a change in the length and cross-sectional area of the resistor will cause a change in the resistance. In the strain gauge, movement of the diaphragm causes a lengthening or shortening (and respective narrowing or widening) of the resistor.

This change in resistance can be measured due to the unique arrangement of the Wheatstone bridge circuit seen in Figure 2.

As two resistors in the circuit are of known and constant resistance (R_1 and R_2) and the third is manually variable (R_3), the unknown fourth (R_4) can be calculated by a null-deflection technique. The galvanometer in the circuit will experience null-deflection when both sides of the parallel circuit are of equal resistance. Therefore, by manipulating the manually variable resistor to a point of null-deflection on the galvanometer, the unknown resistor value can be calculated. Use of microprocessors makes this system fully automated with a rapid response time. This transduced signal is transmitted for processing and display.

Oscillation of the fluid column following a change in pressure within the system (e.g. by ventricular contraction) will, however, not be a singular event. Reflection of the oscillatory wave between the arterial wall and the diaphragm can lead to resonance within the system. Each system will have a given frequency at which it will continue to oscillate — its natural frequency. Just as a well-pitched note has the ability to cause resonation in a glass to the point of it shattering, an input at the natural frequency of the arterial line system can amplify the signal by up to 40%, causing inaccuracy. An accurate and reliable device must be able to overcome these difficulties.[1–3]

Natural frequency, harmonics and frequency response

In order to overcome the issues of resonance, the natural frequency of the system can be manipulated to lie outside of the range to be measured. The natural frequency is calculated as:

$$f_n = r/2\ \sqrt{S/\pi\rho l}$$

(where f_n: natural frequency, r: radius, S: stiffness, ρ: density, l: length).

So as to avoid frequencies encountered during measurement, the natural frequency of the system should be ten times the fundamental frequency (1st harmonic) or higher. As can be seen above, this can be achieved by use of stiff, short and wide tubing with low density fluid within.

The fundamental frequency, or 1st harmonic, relates to the lowest frequency wave in the complex blood pressure waveform. This corresponds with the heart rate — 60–120/minute or 1–2 Hz (Hertz is the SI unit of frequency and is equal to one

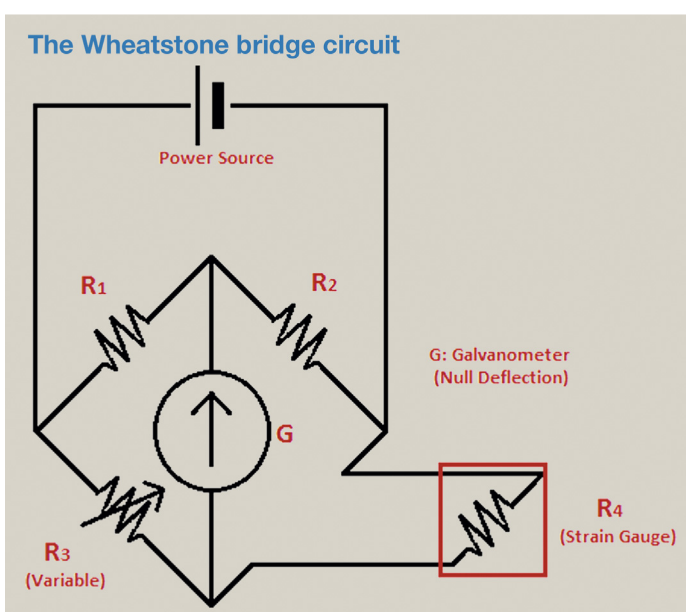

Figure 2

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cycle per second, therefore 1 Hz = 60 cycles/minute). The familiar arterial pressure waveform is actually made up of many individual sine waves of increasing frequency and decreasing amplitude – i.e. 2 Hz (2nd harmonic), 3 Hz (3rd harmonic), and so on. The method of analysing and dividing the signal into these constituent waves is Fourier analysis. Each harmonic is superimposed onto the first harmonic to produce an increasingly accurate representation of the true pressure waveform. Therefore, the more sine waves the system can compute, the better the accuracy. The resulting representation comprises the first ten or so harmonics – any harmonics higher than this are insignificant due to their low amplitude.

The ability of the system to deal with and produce an output for all frequencies it encounters is termed the frequency response. A frequency response of 30 Hz would mean the system can manage the first ten harmonics of a waveform with a fundamental frequency of 3 Hz – i.e. a heart rate of 3/second or 180/minute; 30–40 Hz is considered a reasonable frequency response for an arterial pressure transducer.

Another method of dealing with issues of resonance and noise (irregular fluctuations in a signal which are not a part of it and can mask it) is to optimize the degree of damping in the system.[1,2]

Damping

Whilst oscillation is the tendency of a system to move either side of a baseline, damping is the tendency to diminish this oscillation. Damping occurs due to a dissipation of the energy within the system. Underdamping gives a rapid, exaggerated and prolonged period of oscillation after a step change; the resultant signal is not useful due to overshoot of the extreme values – pressure changes are overestimated leading to falsely high systolic and falsely low diastolic readings. Overdamping, however, results in the pressure changes being underestimated (falsely low systolic, falsely high diastolic) and an increasingly extended time taken for the trace to approach the true value, a phenomenon called phase lag. A situation where there is no overshoot of the trace would appear to be ideal, but there is still delay in the approach to the true value (critical damping). The compromise is to allow a small degree of overshoot in order to achieve a true value promptly. This situation is known as optimal damping.

The level of damping present in a system is represented by the damping coefficient (*D*). This value increases as the degree of oscillation decreases. In the circumstances described above, *D* assumes the following values:

- $D = 0$ No damping: oscillations continue indefinitely (at natural frequency)
- $D < 0.64$ Underdamping: overshoot, prolonged oscillation
- $D = 0.64$ Optimal damping: ideal compromise between overshoot and rapidity
- $D = 1$ Critical damping: no overshoot, but phase lag
- $D > 1$ Overdamping: no oscillation, underestimation of true values

The presence of gas bubbles in the line increase damping due to compressibility. Blood clots, excessive connectors and 3-way taps also cause damping. Decrease in diameter of the line will increase damping:[3]

$$D \propto 1/r^3$$

Errors

Very few measurement systems could be said to be entirely accurate and reliable and there are many potential sources of error, increasingly so over time. Regarding the arterial pressure transducer model, the initial setup can be a source of error in itself. For example, if the transducer is set at the wrong vertical level, the output measurement will not represent the true value accurately. It must be set at the level of the right atrium. Placing the transducer too low will cause an overestimate of around 7.4 mmHg for every 10 cm discrepancy. A transducer set too high will under-read by the same degree. In specific situations, the transducer may be set elsewhere, e.g. at the level of the tragus (as with intracranial pressure monitoring) as an estimation of the level of the Circle of Willis if there are concerns surrounding cerebral perfusion pressure.

Furthermore, accuracy will fall as the placement of the arterial cannula becomes more peripheral due to the narrowing and heightening of the waveform. The more peripheral the cannula, the higher the systolic pressure will read. For example, a line in the dorsalis pedis artery will give a higher systolic output than one in the brachial artery.

More generally, there are some concepts in clinical measurement which are important to mention and understand.[1,3]

Linearity, drift, hysteresis and calibration

Linearity is a description of the proportionality of the displayed to the true values in a measurement system (accuracy is subtly different as it is the absolute difference in these values). The degree of difference is quoted as the non-linearity figure. Transducers often display a correct value at the extremes of measurement but may differ in the middle-range. If the maximum difference between a true and a displayed value was 5 mmHg, this would be expressed as ±2.5 mmHg and the non-linearity would be given by expressing this as a percentage of the maximum output value (perhaps 250 mmHg). For example:

$$\text{Non-linearity} = \pm 2.5\ \text{mmHg}/250\ \text{mmHg} = \pm 1\%$$

The degree of non-linearity in an arterial pressure transducer system is probably a little lower than this. It is also possible to electronically account for such conditions during signal processing.

Drift describes the degree of change in a measured value over time when the actual value remains constant. It can take one of two forms. Offset drift is where the displayed value has the same degree of error over all measurement values e.g. if the transducer constantly read 10 mmHg over the true pressure. This issue can be addressed with a one-point calibration. This is often achieved by ‘zero-ing’ the arterial line – calibrating to atmospheric pressure. When the difference in measured and true pressures is not constant across all values (e.g. 10 mmHg over at 100 mmHg but 20 mmHg over at 150 mmHg), the error is termed gradient drift. Because the error is not constant, one-point calibration is insufficient for correction; two-point calibration must be used.

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Hysteresis is a phenomenon where the degree of error is dependent on whether the true value is increasing or decreasing, e.g. airway pressures during inspiration or expiration. The level of error caused by hysteresis is usually insignificant in anaesthetic monitoring.[1–3] ◆

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