Arterial Transducers and Damping

Dr. Gareth Davies
Department of Anaesthesia
Paarl Hospital

Introduction

A transducer is a device that converts one form of energy to another form of energy. When measuring intra-arterial blood pressure, the arterial pulse pressure mechanical waveform is transmitted via a column of fluid in the arterial catheter and tubing to a pressure transducer, where it is converted into an electrical signal (hydraulic coupling). This signal is then processed, amplified and converted into a visual display by a microprocessor.

The apparatus required for measuring intra-arterial blood pressure includes the following components:
- Intra-arterial catheter
- Fluid filled tubing
- Pressure Transducer
- Infusion/flushing system
- Signal processor, amplifier and display

Basic Principles

A wave is a disturbance that travels through a medium, transferring energy, but not matter. The simplest wave form is the sine wave.

![Fig 1: The sine wave]

The following key terminology can be used to describe a sine wave:
- **Amplitude**: the maximal displacement from zero
- **Frequency**: the number of cycles per second – expressed as Hertz (Hz)
- **Wavelength**: the distance between two points on the wave that have the same value
- **Phase**: the displacement of one wave in comparison to another

Sine waves are fundamentally important when considering arterial transducers, as any waveform may be represented by combining together sine waves of different frequencies, amplitudes and phases. Thus the complex arterial pulse pressure wave can be broken down into a number of different sine waves (Fourier analysis).

This wave consists of a fundamental wave (the pulse rate) and a series of other harmonic waves. These smaller waves have frequencies that are multiples of the fundamental wave (e.g. 2 Hz, 3 Hz, etc.).

A microprocessor performs the function of breaking down the complex waveform into the fundamental wave and at least 10 or more harmonics of higher frequency, to give an accurate representation of the original waveform.
The figures below demonstrate this analysis with two sine waves:

Fig 2: Two sine waves with differing frequency, amplitude and phase

Fig 3: The sum of the two sine waves

Natural frequency and resonance

All materials have a frequency at which they oscillate freely – this is called their natural frequency. When a force with a similar frequency to the natural frequency is applied to the system it will begin to oscillate at its maximum amplitude. This phenomenon is known as resonance.

This can be demonstrated when pushing someone on a playground swing, where the swing acts as a pendulum. Pushing the swing at its natural frequency makes the swing go higher and higher (maximum amplitude), while trying to push the swing faster or slower produces smaller arcs. This occurs because the energy the swing absorbs is maximised when the pushes match the swing’s natural oscillations.

If the natural frequency of an intra-arterial blood pressure measuring system lies close to any of the frequencies of the sine wave components of the arterial pulse pressure waveform, then the system will resonate, producing excessive amplification and signal distortion (amplitude distortion).

The figure below demonstrates that at lower frequencies there is minimal amplitude distortion, however at higher frequencies, where the transduced signal approaches the resonant frequency of the system, significant distortion occurs.

Fig 4: Amplitude distortion at increasing frequency

It is thus imperative that all the components of an intra-arterial monitoring system have a very high natural frequency – at least 8 – 10 times higher than the fundamental frequency of the arterial waveform (the pulse rate). So for the system to remain accurate for heart rates up to 180bpm, the natural frequency must be at least $\frac{180\text{bpm} \times 10}{60\text{sec}} = 30$ Hz.
For a fluid filled system the resonant frequency ($f_0$) can be calculated with the following equation:

$$f_0 = \frac{1}{2\pi} \sqrt{\frac{\pi r^2 \Delta P}{p \ell \Delta V}}$$

$r$ = the radius of the tubing  
$p$ = the density of the fluid  
$\ell$ = the length of the tubing  
$\Delta P$ = the elastance of the system  
$\Delta V$ = the elastance of the system

Thus, to obtain the highest possible natural frequency in an intra-arterial blood pressure monitoring system, we require shorter, wider, stiffer cannulae and tubing, with lower density fluid.

**Damping**

In practice the monitoring systems used do not possess a high natural frequency, and damping is introduced to decrease or eliminate amplitude distortion. Damping reduces the energy in the system (by creating friction in the fluid pathway) and thus reduces the amplitude of the oscillations. Some degree of damping is required in all systems, but if excessive (overdamping) or insufficient (underdamping), can be a major source of error.

Overdamping can be caused by:
- three way taps  
- bubbles and clots  
- vasospasm  
- narrow, long or compliant tubing  
- kinks in the cannula or tubing

**Overdamping** will result in an under-reading of systolic blood pressure and an over-reading of diastolic blood pressure. The response time of the system is also increased.

In an underdamped system pressure waves overshoot, with excessively high systolic blood pressures and low diastolic blood pressures.

**Critical damping** is the amount of damping required to prevent any overshoot. The damping coefficient in a critically damped system is 1. However, this results in a system that is relatively slow to respond. A damping coefficient of 0.64 (termed **optimal damping**) provides the best compromise between responsiveness and distortion.

![Fig 5: Damping co-efficient (D) – demonstrating the response to a step change in pressure](image)

With a damping co-efficient of 0.64, amplitude is accurately replicated up to 2/3 of the natural frequency (within 2%), and only displays a maximal amplitude distortion of 6% at natural frequency.

In the clinical setting the damping co-efficient of a system can be calculated by using the “fast flush” test (see Fig 6 below). The monitoring system is flushed with high-pressure saline via a flush system, creating oscillating waves, resonating at the natural frequency of the system. The trace obtained is then printed and the amplitude of two successive waves measured, allowing the calculation of the
amplitude ratio \(^{\frac{A_2}{A_1}}\). The amplitude ratio is converted to the corresponding damping co-efficient by using available conversion tables. Generally, a lower ratio corresponds to a higher damping co-efficient and a higher ratio represents a lower damping co-efficient.

![Amplitude Waveform Diagram](image)

**Fig 6: “Fast-flush” test**

The fast-flush test is also used to determine the natural frequency of the system. The distance between the peaks of 2 successive waves (one cycle length) is measured and applied to the following formula:

\[
\text{natural frequency} = \frac{\text{paper speed in mm/sec}}{\text{length of one cycle in mm}}
\]

Therefore, in summary: damping is introduced into monitoring systems with low natural frequencies to decrease or eliminate amplitude distortion due to resonance.

**Phase shift**

As discussed before, the complex arterial pulse pressure waveform can be deconstructed into the fundamental wave with at least 10 additional harmonics. When these waves are summated by a microprocessor, not only their amplitude, but also their phase relationship will affect the displayed waveform. When no damping is present in a system, harmonic waves equal to the natural frequency of the system will be delayed by 90°, whilst those with a very low frequency will demonstrate almost no delay. Waves between these two extremes will demonstrate a variable amount of *phase lag*. 
To correct this phase lag, all harmonics need to be delayed in a linear proportion to their frequency. Harmonics with a frequency of 1f (the fundamental frequency) will need to be delayed by 90°, 2f by 180°, 4f by 360°, etc. This produces a delay in the displayed result, but allows the phase relationship of all waves to be preserved.

Fortuitously, optimal damping (damping co-efficient = 0.64) provides precisely this proportional delay and allows the accurate summation of all harmonics.

**Transducers**

In the intra-arterial blood pressure measuring system the arterial pulse pressure is transmitted to a flexible diaphragm by a column of fluid – displacing the diaphragm. The commonest method of measuring this displacement is with a strain gauge. Strain gauges utilise the principle that the electrical resistance of wire or silicone increases with stretch. The flexible diaphragm is attached to wire or silicone strain gauges and then incorporated into a Wheatstone bridge circuit (see Fig 8 below). With movement of the diaphragm the gauges are stretched or compressed, altering their resistance.

**Fig 8: The Wheatstone bridge circuit.** When $R_2/R_1 = R_x/R_3$, the circuit is balanced and no deflection is registered on $V_G$

These circuits are constructed with 3 resistors of known resistance ($R_1$, $R_2$ and $R_3$) and one of unknown resistance ($R_x$). The resistance of $R_3$ is variable and can be adjusted to match the resistance in $R_x$, until the galvanometer ($V_G$) produces a zero deflection (voltage between B and D is zero). When pressure is applied to the diaphragm in the measuring system, the two sides of the bridge become unbalanced and current flows. The resistance in $R_2$ is then altered to maintain a zero balance between B and D, and the magnitude of this change in resistance is proportional to the pressure applied.
Modern Wheatstone bridge circuits have strain gauges in all four positions. The diaphragm is attached in such a way that when pressure is applied to it, the gauges on one side are compressed (reducing their resistance) and those on the other side are stretched (increasing their resistance). The bridge becomes unbalanced and the potential difference generated is proportional to the pressure applied. Using four strain gauges has 2 major benefits: (i) the system is four times more sensitive than a single strain gauge system and (ii) as a change in temperature can affect strain gauges, any temperature change in this system affects all of the strain gauges equally (in a single strain gauge setup temperature change can skew readings).

**Zeroing and levelling**

Zeroing

Atmospheric pressure must be discounted from the pressure measurement, for the pressure transducer to read accurately. This is achieved by exposing the transducer to atmospheric pressure and then calibrating the pressure reading to zero. As transducers are prone to baseline drift, this should be performed several times a day.

Levelling

The pressure transducer must be set at the level of the patient's heart, to measure blood pressure correctly. By convention this is set at the level of the right atrium, at the 4th intercostal space in the mid-axillary line. If the transducer is placed above or below this level, the hydrostatic pressure exerted by the column of fluid (in this case blood) is being measured in addition to the blood pressure. For a 10cm error in levelling, the measured blood pressure will be over- or under-read by 7.4 mm Hg (over-read if the transducer is too low, under-read if too high).

**Summary**

The measurement of invasive blood pressure is one of the most valuable clinical tools in the management of critically ill patients. The underlying physical principles may be complex, but are essential to master – allowing the accurate interpretation of displayed values, and the identification of possible sources of error (resonance, damping, phase-shift, etc.)

**References**

1. Dyer R. The characteristics of the ideal catheter-transducer system for blood pressure measurement. *UCT FCA Part I ARC 2014*
3. Jones A, Pratt O. Physical principles of intra-arterial blood pressure measurement. *Anaesthesia Tutorial of the Week* 137 8 June 2009
4. Joubert, IA. Invasive blood pressure measurement and damping. *UCT FCA Part I ARC 2006*
6. Morgan Jr GE, Mikhail MS, Murray MJ. Patient monitors. *Clinical Anaesthesiology*
8. Van West C. Transducers and damping. *UCT FCA Part I ARC 2010*