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Textbook of Hemodynamic Monitoring & Therapy in the Critically Ill

Editors
Atul Prabhakar Kulkarni
Amol Trimbakrao Kothekar
Jigeeshu Vasistha Divatia

2nd Edition

JAYPEE

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CHAPTER

1

Physical Principles and Sources of Error in Invasive Blood Pressure Measurement in the Critically Ill

Atul Prabhakar Kulkarni

"The discovery of the blood pressure was more important than the discovery of the blood"

—Johannes Muller[1]

This is how Stephan Hale described his first attempt at the measurement of blood pressure in a horse.[1]

"In December I caused a mare to be tied down alive on her back; she was 14 hands high, and about 15 years of age; had a fistula of her withers, was neither very lean nor yet lusty; having laid open the left crural artery about three inches from her belly, I inserted into it a brass pipe whose bore was one-sixth of an inch in diameter. I fixed a glass tube of nearly the same diameter, which was 9 feet in length: Then untying the ligature of the artery, the blood rose in the tube 8 feet 3 inches perpendicular above the level of the left ventricle of the heart; when it was at its full height, it would rise and fall at and after each pulse 2, 3, or 4 inches."

HISTORY[1,2]

1733: First ever-arterial cannulation—Rev Stephen Hales.

1828: Poiseuille cannulation of small arteries, use of a mercury manometer for the measurement of arterial blood pressure. Also Ca^{++} for anticoagulation.

1856: Arterial cannulation of an artery in an amputee —Faivre.

1855: Vierordt suggested that arterial pressure can be measured by measuring the pressure required to stop the pulsations in the artery.

1881: The first sphygmomanometer was invented—Samuel Siegfried Karl Ritter von Basch.

1896: Improvement in the sphygmomanometer by Scipione Riva-Rocci.

1947: Direct measurement of the arterial pressure using strain gauge—Lambert and Wood.

1949: Continuous recording of arterial blood pressure in brachial artery—Peterson.

1961: Catheter over needle cannulation of artery—Barr.

INTRODUCTION

One of the most common parameters used to assess hemodynamic status of the critically ill patients is blood pressure. "Blood pressure may be the vital sign, we measure the most and understand the least." Blood pressure measurement techniques have evolved over last 3 centuries to improve its accuracy. There has been a technological evolution for increasing the accuracy in monitoring patient's blood pressure. This has involved measurement of manual blood pressure using a mercury sphygmomanometer to invasive arterial blood pressure (ABP) measurement. It has further evolved to include the oscillometry, applanation tonometry, and the volume clamp methods. Sir Otto Frank published his work on the analysis of arterial pressure pulse in 1899, which led to dynamic response monitoring.[3,4] Invasive blood pressure measurement has since become the gold standard for monitoring blood pressure. Since there are multiple components in the monitoring system and a human interface, multiple sources of errors exist. In this chapter, we describe these sources of error and ways to avoid these errors, to obtain accurate measurements.

CONVENTIONS TO REMEMBER WHILE MEASURING CARDIOVASCULAR PRESSURES

- All pressures are expressed as millimeters of mercury, except central venous pressure (CVP), which may be expressed as cmH_2O. 1 mm Hg = 1.36 cmH_2O.

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Ideally, however, even CVP measurement should be done using a transducer because:

- Use of an open tubing, and improper handling of the tube, can cause infections.
- The frequency response of this system is slow, which is liable to cause inaccuracies in the measurement.

- All cardiovascular pressures are referenced to the heart, in what is known as the phlebostatic axis or plane **(Fig. 1)**. "Phlebostatic axis" was established by Winsor and Burch; as the external landmark for the right atrium and pulmonary catheter tip. It is located by identifying the patient's fourth intercostal space, and drawing an imaginary line that extends down the lateral aspect of the thorax.[5] The midaxillary line lies in the middle of anterior and posterior chest borders. Phlebostatic point is the point where these two lines converge. A line or plane passing through the phlebostatic point is termed as phlebostatic axis, or plane. There is an argument that the phlebostatic axis may not be located at exactly the same point in all patients. However, it does not matter in routine clinical practice, if the clinician marks the point on the chest with a marker pen while taking the first reading and levels the transducer at this point, and takes all subsequent readings with the transducer leveled consistently there, then it is accurate enough for clinical use. It is important to remember that it is the trend of change in the (monitored) variable, and not each single value observed, that is important for monitoring and therapy. The air-fluid zero point must be "leveled" or "referenced" horizontal to the precise point where the end of the pressure-monitoring catheter is located inside a vessel. This is necessary because pressure is being measured through a fluid-filled catheter and a certain length of connecting tubing. The weight of fluid within the tubing exerts "hydrostatic pressure" and is proportional to the height of the fluid column. Every 1 cm air-fluid interface difference from this leveling point will lead to an increase or decrease of 0.74 mm Hg in the measured pressure. To cite an example, if the transducer level is off by 10 cm, there will be a difference of 7.4 mm Hg in the measured pressure in the either direction. While this may not be significant in the measurement of large pressures (arterial pressure), it can make the therapy inappropriate when measuring pressures, which are of lower magnitude [such as CVP and pulmonary artery occlusion pressure (PAOP)].

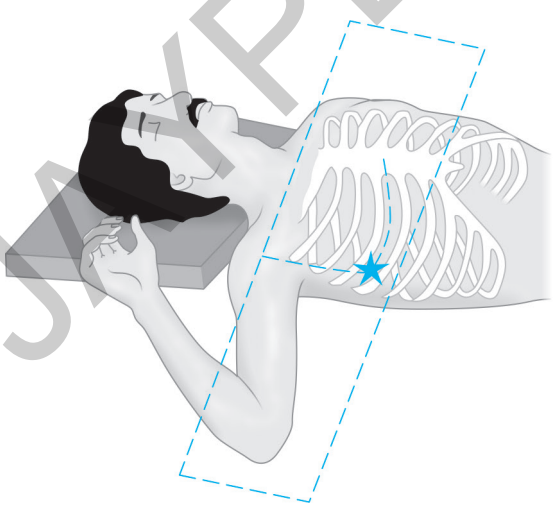


Fig. 1: Phlebostatic point, axis, and plane.

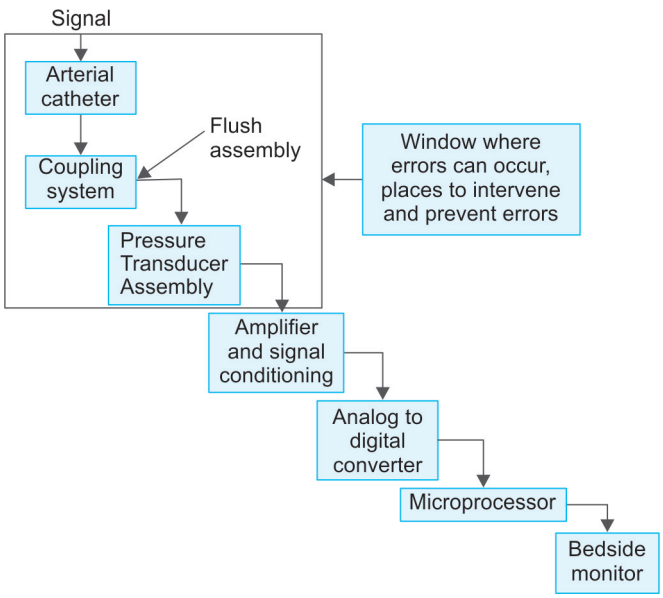
- All pressures, because the transducer is zeroed to atmosphere before measurement, are pressures above the ambient atmospheric pressure. During zeroing the transducer, the clinician is establishing the fact that the monitor has to treat the ambient pressure (760 mm Hg) as zero, and therefore all pressures measured by us at the bedside are 760 + measured pressure in mm Hg.

COMPONENTS OF THE MONITORING SYSTEM USED FOR INVASIVE PRESSURE MEASUREMENTS (FLOWCHART 1)

- An intravascular catheter to access/cannulate the patient's blood vessels.
- Connecting tubing (which is noncompliant) and stopcocks—to connect the intravascular catheter (in the patient's blood vessel) to the monitoring system.
- A pressure transducer to convert the mechanical impulse of a pressure wave into an electrical signal through movement of a displaceable sensing diaphragm. Modern transducers use a silicon wafer as the diaphragm, which is a stiff, low compliance pressure sensing device, capable of bending and creating a small volume change in response to applied pressure. The transducers function on the principle of the Wheatstone bridge (see later in the following test).
- A continuous flush device fills the pressure tubing with normal saline (in pediatric patients Ringer's lactate should be used, unless contraindicated) and helps prevent blood from clotting in the catheter by continuously flushing fluid through the system at a rate of 1–3 mL/hour.

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Flowchart 1: Components of the monitoring system used for invasive pressure measurements, showing window for sources of errors and their rectification.



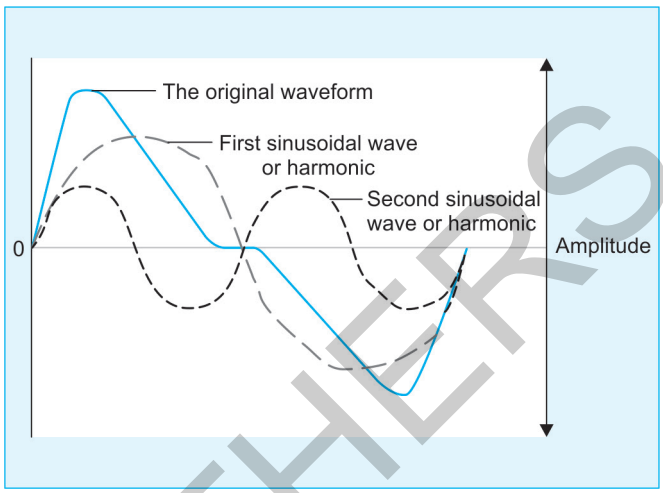


Fig. 2: Fourier series (multiple sinusoids or wavelets) forms the final complex waveform (see text for detailed explanation).

- An amplifier increases the low voltage signal from the pressure transducer (usually measured in microvolts) to a signal that can be displayed on an oscilloscope or other display device. Most amplifiers include electronic filters to filter out unwanted physiologic "noise."
- A monitor to display waveforms along with digital fields (to display numerical data).
- A processor or microcomputer, to calculate hemodynamic parameters from the measured variables.
- A recorder and a printer, for printing the patient data for the hospital records. Modern monitors are now equipped with interface, which allows direct storage of data in the electronic medical record (EMR).

ARTERIAL PRESSURE WAVEFORM: A FOURIER SERIES[6]

Fourier, a 19th century French mathematician, showed that complex repetitive waveforms could be analyzed as being the sum of a series of much simpler waveforms called sinusoids **(Fig. 2)**. A complex waveform thus can be broken down into a series of sinusoids in the same phase (same direction and passing through the zero amplitude) but of decreasing amplitudes (and having increasing frequency). Of these sinusoids, the first 6–10 sinusoids contribute the maximum information about the final waveform. The first sinusoid in this series, with the lowest frequency, is called fundamental frequency (or the first harmonic). For cardiovascular pressures, this equates to the heart rate. The second harmonic is a sinusoidal waveform that has twice the frequency and the third harmonic three times the frequency of the first harmonic, and so on. Arterial waveform, which is a complex waveform, can have as many as 30 harmonics. Therefore, reproducing the first 10 harmonics can produce a close approximation of the final arterial waveform.

If the heart rate of patient is 60 beats/minute, then it has the frequency of 1 Hz, this is called the fundamental frequency. Since the arterial waveform can have as many as 30 harmonics, the monitoring recording system would require a frequency response that is undistorted up to 30 Hz to accurately reproduce the original waveform. The natural frequency of the arterial pulse (brachial and radial) ranges between 16 and 24 Hz. Thus, it follows that for faithfully reproduction of the original arterial waveform, the monitoring system should therefore have the natural frequency greater than this.

SITES OF CANNULATION, ARTERIAL CATHETER, AND END-HOLE ARTIFACT

The discussion about catheter types, techniques of insertion are beyond the scope of this write-up. Most clinicians perform arterial cannulation using the "catheter over needle" technique described by Barr in 1961. Nowadays commercial kits with Seldinger technique of different sizes are also available for cannulating arteries at various anatomical sites.

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Problems of weakly palpable pulse can be overcome by using transillumination (infants) or miniature Doppler probes in adults. Difficulties in threading catheters can be overcome by using appropriately sized catheters, a catheter over guidewire (Seldinger technique) or a liquid stylet (after obtaining free-flow flushing with saline while threading the catheter), etc. A meta-analysis showed that first attempt success rate was higher when ultrasound guidance was used to cannulate the radial artery.[5] A recent meta-analysis suggested that though ultrasound may improve first success rate, it did not change the overall success rates, while cautioning that this conclusion must be viewed with caution due to significant heterogeneity in the trials.[6] However, use of ultrasound during arterial cannulation has become a routine practice now.

It is important to use a catheter of appropriate size (i.e., largest possible diameter but smallest length, see the formula for natural frequency in the following text). This is important because the arterial catheter tip points toward the heart. The blood flowing toward the catheter has kinetic energy. When the tip of catheter meets the blood having kinetic energy, some of this energy is converted into pressure, leading to a false increase (by 2–10 mm Hg) in the measured systolic blood pressure (SBP). This augmentation of systolic pressure due to addition of converted kinetic energy is known as the end-pressure product or the end-hole artifact.[7]

CONNECTING THE ARTERIAL CATHETER TO THE TRANSDUCER: THE COUPLING SYSTEM[8]

The typical catheter-transducer system used in the intensive care unit (ICU) setting is considered to be an "underdamped, second-order dynamic system." A second-order system is characterized by three mechanical parameters:

1. *Elasticity:* The stiffness of the system.
2. *Mass:* The weight of the material moving through the system.
3. *Friction:* The resistance encountered as the mass oscillates.

In the ICU, "elasticity" reflects the flexibility of the pressure transducer diaphragm and the compliance of the pressure tubing. "Mass" reflects the fluid or blood traveling through the catheter and tubing. "Friction" refers to both the resistance of the fluid, as it moves with each pulsatile change in blood pressure, as well as the viscosity of the fluid itself.

Each system has a characteristic natural frequency (the number of "oscillations" per unit time, natural frequency being the frequency at which it will oscillate freely, and in the absence of sustained stimulus). The arterial pressure waveform is propagated through the systemic circulation at a given frequency. The resulting sine wave, occurring at the rate of the patient's pulse, is called the first harmonic or fundamental frequency (see in the earlier text). If the natural frequency of the monitoring system is the same as the arterial pulse-pressure wave, or the pulse pressure wave reflects off a large catheter or obstruction in the system, then, a wave reflection or "harmonic" is generated which may be a sum of two waves, falsely augmenting the systolic blood pressure and reducing the diastolic blood pressure, though the mean arterial pressure will remain the same **(Fig. 3)**. The natural frequency at which a catheter-extension tubing system tends to oscillate can be calculated using the following formula:

$$f_n = \frac{1}{2\pi}\sqrt{\frac{\pi D^2}{4\rho L}\cdot\frac{\Delta V}{\Delta P}}$$

where, volume/change in pressure = compliance; D: diameter, L: length of the tubing, and ρ is density of the solution.

It is evident from the formula above that longer the length of the system, lower the natural frequency of the system, and the more likely it is to resonate with the pulse-pressure waves. In order to keep the natural frequency of the monitoring system higher than the frequencies in the pulse pressure wave (normally likely to be up to 20–30 Hz),

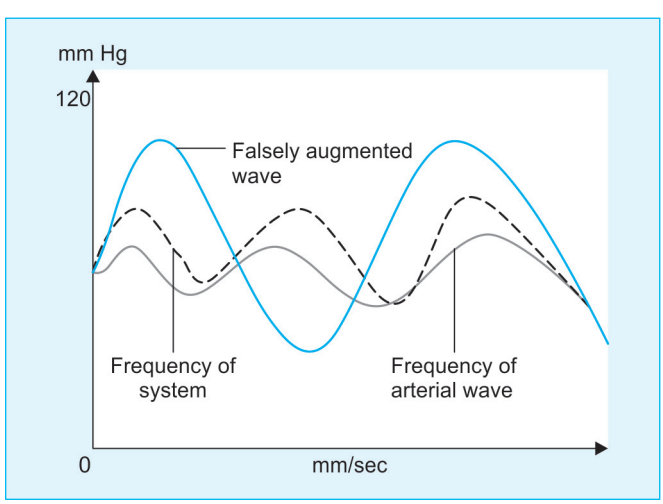


Fig. 3: Augmented systolic pressure due to summation of natural frequency of the monitoring system and of the waveform.

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it is desirable to maximize the diameter of the catheter and tubing. The length of the catheter and the tubing should also be kept to the minimum (in the ideal world, the catheter would be directly connected to the transducer). This is because, if the length is increased to 6–8 feet, the resultant overshoot in SBP can vary from 2 to 31.3%. Small air bubbles in the tubing can cause hyperresonance due to summation, while large air bubbles in the tubing cause damping of the displayed waveform.

RESONANCE AND DAMPING IN THE MONITORING SYSTEM AND THEIR EFFECT ON THE WAVEFORM[9]

To prevent resonance (ringing) in the system, some degree of damping is essential in the system. There are frictional forces within the monitoring system or "damping factors" which include size and interval of the pressure wave, which affect the time it takes for the wave to dissipate and disappear. Without some extent of damping within the system (an "underdamped" system), the pressure waves would tend to reverberate, i.e., resonate (as discussed earlier) within the catheter and tubing, overestimation of systolic blood pressure. The damping coefficient can be calculated by the formula:

$$\beta = 1/r^3$$

A narrow, peaked waveform is characteristic of an underdamped pressure waveform **(Fig. 4 and Table 1)**. This waveform tends to overestimate SBP while underestimating diastolic blood pressure (DBP). At the opposite end of the spectrum is the overdamped system, which underestimates the SBP while overestimating DBP, this is seen at the bedside as a widened slurred tracing, a common phenomenon in the ICU **(Fig. 5 and Table 2)**.

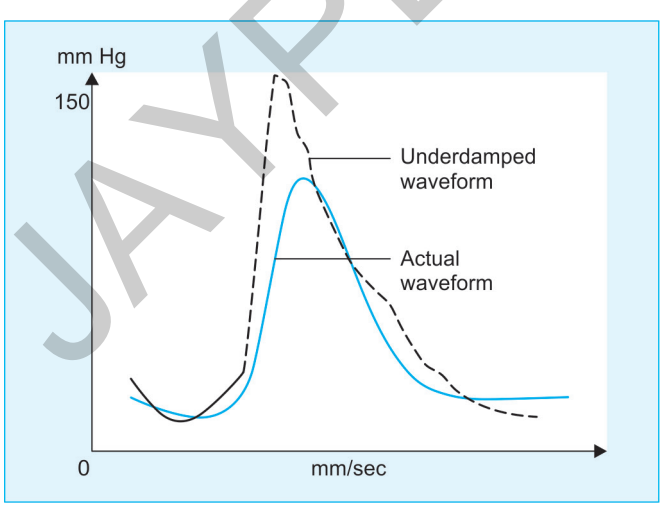


Fig. 4: Underdamped waveform.

TABLE 1: Characteristics and causes of underdamped waveforms.

Nature of waveform	*Effect on measurements*	*Causes*
Narrow, peaked tracing	Overestimates SBP	Long tubing increased vascular resistance
	Underestimates DBP	
	MAP remains unchanged	

(DBP: diastolic blood pressure; MAP: mean atrial pressure; SBP: systolic blood pressure)

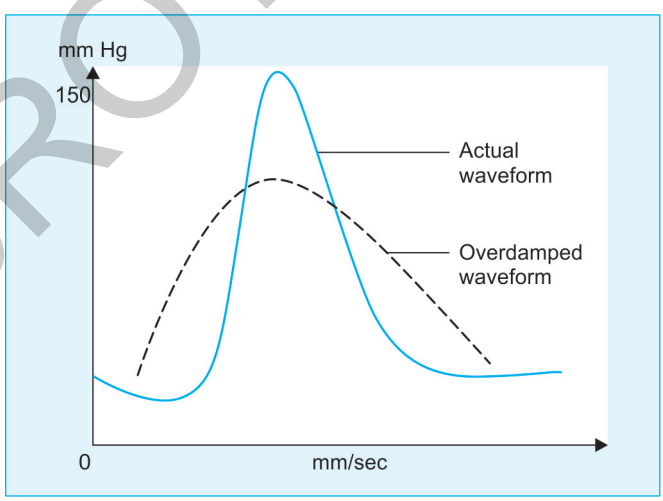


Fig. 5: Overdamped waveform.

TABLE 2: Characteristics and causes of overdamped waveforms.

Nature of waveform	*Effect on measurements*	*Causes*
Widened and slurred pressure tracing	Underestimates SBP	• Air bubbles • Overly compliant tubing • Blood clots/fibrin • Catheter kinks • Stopcocks/injection ports • No fluid in flush bag/low flush bag pressure
	Overestimates DBP	
	MAP remains unchanged	

(DBP: diastolic blood pressure; MAP: mean arterial pressure; SBP: systolic blood pressure)

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Although damping seems to be a solely a theoretical issue, underdamped and overdamped waveforms are encountered on a daily basis in the ICU. The ability to recognize when these potential sources of error, or "dynamic response artifacts," are present is essential to being able to effectively analyze and apply hemodynamic measurements in the care of the critically ill and avoid potentially detrimental therapy.

THE WHEATSTONE BRIDGE PRINCIPLE (HOW THE TRANSDUCER FUNCTIONS)[8]

In 1833, Samuel Hunter Christie invented the Wheatstone bridge, which was improved and popularized by Sir Charles Wheatstone in 1843. The Wheatstone bridge is the *interconnection of four resistances* forming a bridge. The four resistances in circuit are referred to as arms of the bridge. The bridge is used for finding the value of an unknown resistance connected with two known resistors, one variable resistor and a galvanometer. To find the value of unknown resistance the deflection on galvanometer is made zero by adjusting the variable resistor. This point is known as balance point of Wheatstone bridge. The Wheatstone bridge can also be used for the measurement of small changes of resistance **(Fig. 6)**.

The pressure transducers contain four strain gauges which are designed so that the resistances of two of the strain gauges at opposite sides of the bridge increase while the resistances of the other two decrease. This results in a larger potential change at the galvanometer connections. The potential is then amplified before display. Modern pressure transducers utilize the same principle, except that they employ the semiconductor technology instead of wires. Silicon crystals also change resistance in linear proportion when pressure is applied. Using integrated circuit technology, a Wheatstone bridge is etched into the undersurface of the silicon pressure-sensing diaphragm. An epoxy coat is given over the entire structure to produce bonded strain gauge. As pressure is applied to the strain gauge, the wires increase in length and decrease in diameter, increasing the resistance to flow of current through the wires of the Wheatstone bridge. This change in the wire's electrical resistance causes a voltage change that can be quantified to reflect the amount of pressure that changed the wires length and diameter. The electrical signal can be amplified and measured; when calibrated, it is proportional to the pressure change. Thus, changes in resistance and current are measured, then electronically converted and displayed as systolic, diastolic, and mean arterial pressures. The preamplifier provides energizing current to the circuit and receives the output current from the transducer. It also contains the controls for calibration to preset pressure (by altering the resistance so that the output is equal to preset pressure) and zero it relative to atmosphere (by changing resistance so that the net output, i.e., output current to preamplifier is zero). After passing through the preamplifier, the electrical analog of the pulse-pressure wave passes through the amplifier section of the monitor, where the signal is boosted up to drive the readout device with sufficient voltage that is proportionate to the pressure change.

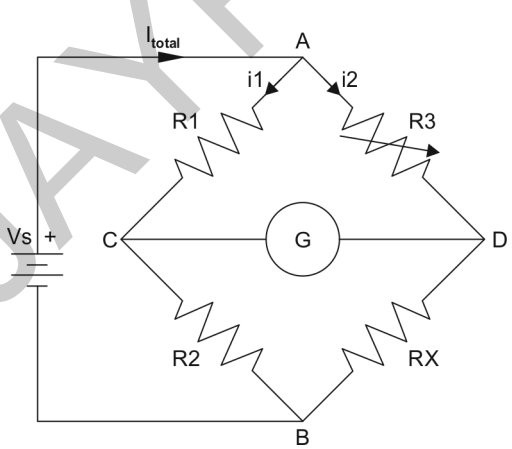


Fig. 6: Wheatstone bridge circuit diagram.

The clinician will have little control over actual electronic components. The system can be calibrated by connecting it to a manometer and raising the pressure to sequentially to 50 mm Hg, 100 mm Hg, and 200 mm Hg. We have a little more control than this, over the next components of the pressure monitoring system: the catheter in the artery and the fluid-filled tubing, the level of the transducer relative to heart, and we can perform dynamic response testing (DRT) for the system that we are using.

DYNAMIC RESPONSE TESTING OF A HEMODYNAMIC MONITORING SYSTEM (FAST FLUSH OR SQUARE WAVE TEST)[7,10]

For accurate capture of basic data, the clinician must ensure that the system is performing optimally. For this, the dynamic response characteristics of the system must be tested. Only when system accuracy, also termed fidelity, has been confirmed, the analog waveform can be accepted as an accurate reflection of a patient's status. The dynamic

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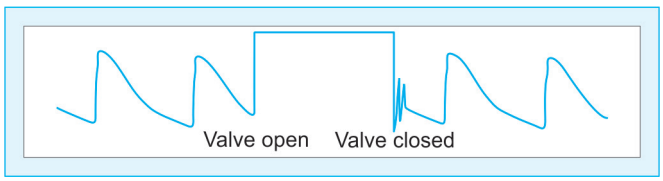


Fig. 7: Dynamic response testing using the square wave or fast flush test.

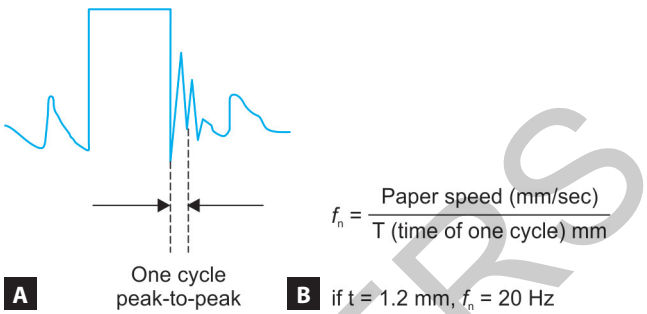
response of a hemodynamic monitoring system is defined by its natural frequency and the damping coefficient. Though performing DRT seems complicated at first, it can be performed quite easily using the fast flush test or square wave test.

Fast flush test (Square wave test): Open the valve of the continuous flush device to perform a "fast flush test;" this increases the flow through the catheter-tubing system is transiently to 30 mL/hour from the usual 1–3 mL/hour. This generates an acute rise in pressure within the system such that a square wave is generated on the bedside monitor. With closure of the valve, sinusoidal pressure waves (oscillations or wavelets) of progressively decreasing amplitude are generated. A system with appropriate dynamic response characteristics will return to the baseline pressure waveform within 1–2 oscillations. For actual calculation a strip chart recorder connected to the monitor is required. The rule of the thumb is, if there are more than two wavelets, the system is underdamped and if there are none, it is overdamped. After performing the fast flush test, record and print the resulting square waveform and subsequent oscillations on calibrated strip chart paper **(Fig. 7)**.

When and how Often the Dynamic Response Testing Should be Performed?[7,10]

The DRT should be performed as follows:

- After opening the system for the first time.
- When the accuracy of measurement is doubtful or when patient's clinical condition does not match the monitored parameters.
- When we suspect an overdamped or underdamped waveform.
- Before implementing interventions or making changes to ongoing interventions.
- Physiological changes place higher demands on the monitoring system.
- At least every 8–12 hours or at the time of each change of shift.



Figs. 8A and B: Calculating the natural frequency.

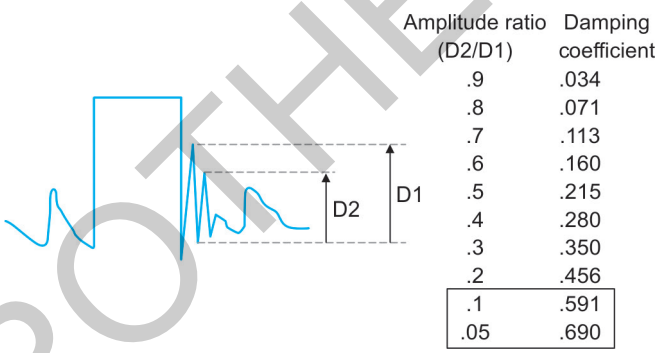


Fig. 9: Calculating damping coefficient and amplitude ratio.

- *Determining natural frequency (f_n):*[10]
  - Measure the distance [period (t) of one cycle] in millimeters between two oscillations (one small box on the calibrated strip chart paper equals 1 mm) **(Fig. 8A)**.
  - Calculate f_n by using the formula f_n = paper speed (mm/s)/t of one cycle (mm) (The standard paper speed is 25 mm/s) **(Fig. 8B)**.
- *Determining the amplitude ratio and damping coefficient:*[10]
  - Measure the amplitude (D) in millimeters of two successive oscillations (D1 and D2) **(Fig. 9)**.
  - *Calculate the amplitude ratio:* Divide D2 by D1 (D2/D1).
- *Determining the dynamic response characteristics of a monitoring system:*[10]
  - Plot fn along the x-axis.
  - Plot the amplitude ratio along the y-axis on the right.
  - Find the intersection of the two lines **(Fig. 10)**. Where the two lines intersect on the f_n-versus-amplitude ratio graph, determines if the system is able to correctly reproduce the hemodynamic waveform (adequate, optimal) or if the system is not functioning at a desirable level (overdamped, underdamped, and unacceptable).

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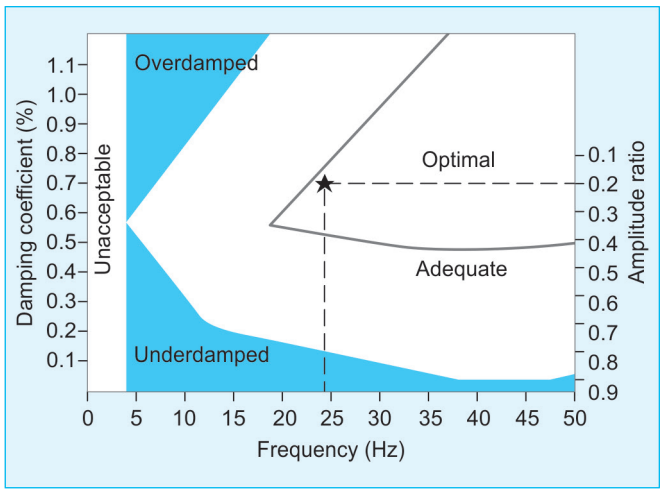


Fig. 10: Plotting the dynamic response characteristics of a monitoring system.[10]

CONCLUSION

It is important for every clinician to understand the physical principles behind pressure transduction in clinical medicine and understand where errors are likely to occur due to equipment related problems. Once optimal functioning of the monitoring is ensured, it is necessary that zeroing at the correct reference point, periodic calibration of the monitoring systems, and maintenance of the monitoring system such as removing blood from the catheter/tubing, ensuring that the flush bag is not empty, or ensuring adequate pressure in the flush bag will allow accurate acquisition of hemodynamic data.

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