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Fluid-filled hemodynamic monitoring systems

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Intravascular pressures are commonly measured in acute and critical illness. Intravascular or “blood” pressure is the physical pressure that blood exerts on the vessel wall. This pressure is important because the difference in intravascular pressure at any two points in the vascular network is the driving force for blood circulation. The pressure at the root of the aorta is much greater than the pressure in the vena cava so that blood flows from the arterial to the venous side, delivering oxygen and other nutrients to cells along its path. Intravascular pressures commonly measured in small animals are peripheral arterial blood pressure (ABP) and central venous pressure (CVP). Pulmonary arterial pressure (PAP) and pulmonary arterial occlusion pressure (PAOP; also called pulmonary capillary wedge pressure or “wedge pressure”) are also directly measured; however, these measurements are less commonly performed in the veterinary patient.

Peripheral arterial blood pressure may be directly evaluated by measuring intraluminal pressure following placement of an intravascular catheter measurement system, or it can be indirectly measured by surface-applied cuff pressure and flow-detection methods such as Doppler ultrasound or oscillometry. Central venous pressure can only be measured directly by measuring pressure transmitted from the central vein into a catheter-associated measuring system. This chapter deals specifically with the technical aspects of the direct measurement of intravascular pressure using fluid-filled monitoring systems.

Basic pressure principles

Pressure is defined as a force per unit area:

$$P = F/A \quad (8.1)$$

where P is pressure, F is applied force, and A is the cross-sectional area over which force is applied. To interpret measured pressure, it must be compared with a known pressure standard. For instance, one can only appreciate the effect of pressure applied to the surface of a rubber band if one knows the pressure the rubber band exerts back on the operator—whether or not the band will distort (stretch) depends on the difference between these pressures. In medicine, the accepted standard pressure against which physiologic pressures are compared is barometric pressure (P_B), or the pressure in the earth’s atmosphere, which is approximately 760 mm Hg (1031 cm H₂O) at sea level. Clinically, we often are more interested in knowing the intravascular pressure at a given site compared with P_B (i.e., “What is the ABP as measured in the femoral artery of this cat?”) than in knowing the difference in pressure between two different sites.

To understand the physiologic determinants and importance of intravascular pressure, one must also understand Ohm’s law of hydrodynamics, which is expressed as follows:

$$\Delta P = Q \times R \quad (8.2)$$

where ΔP is the pressure difference between an upstream and a downstream measurement site, Q is the flow of blood between those sites, and R is the resistance between those sites. The ΔP may be referred to as the **driving pressure**. Blood flow (Q) is the volume of blood movement per unit time and is directly correlated with ΔP . Resistance to flow, R , is the force opposing forward fluid movement. In very basic terms, the higher the driving pressure (the pressure differential between the arterial and venous sides of a blood vessel), the more likely blood will flow through that blood vessel; the higher the resistance across the blood vessel, the less likely blood will flow through that vessel. An important concept herein is that blood flow through a blood vessel or to an organ is not guaranteed by a “normal” driving pressure if high resistance is present; conversely, blood flow through an organ may be adequate at low driving pressure if the resistance through that blood vessel or organ is also low.

Determinants of intravascular pressure

Determinants of systemic arterial pressure

Systemic ABP is the product of cardiac output (CO) and systemic vascular resistance (SVR) such that

$$ABP = CO \times SVR \quad (8.3)$$

This equation is derived from Ohm’s law, where CO is blood flow (Q), SVR is vascular resistance (R), and ABP is driving pressure (ΔP).

Systemic ABP is determined by cardiac factors such as heart rate and contractility, by blood volume, and by systemic (also called peripheral) vascular tone. Blood pressure is under tight minute-to-minute and long-term control by an integrated neural, endocrine, and paracrine system. Full discussion of these physiologic influences is outside the scope of this chapter, and details can be found elsewhere.^{1,2}

Determinants of central venous pressure

Central venous pressure is frequently used as a surrogate for right atrial (RA) pressure under conditions of no flow obstruction into the RA and normal tricuspid valve function. CVP is very close to right ventricular end diastolic pressure; this value represents the filling pressure of the right side of the heart. Factors determining central venous (RA) pressure include effective blood volume at the site of measurement, pleural pressure (which influences transmural pressure across the great veins and right heart), venous vascular resistance, and right heart “function.” Changes in CVP are commonly used to estimate blood volume in critically ill veterinary patients.

Despite its widespread use for this purpose, there is evidence from human literature that there is no relationship between fluid responsiveness and either CVP or a change in CVP.³

Determinants of pulmonary arterial pressure

Pulmonary arterial pressure (PAP) is analogous to ABP, but is measured in the pulmonary circulation. Pulmonary arterial pressure is the product of CO and pulmonary vascular resistance (PVR):

$$PAP = CO \times PVR \quad (8.4)$$

Pulmonary vascular resistance is affected by multiple factors such as pleural pressure variation (in disease or secondary to the respiratory cycle) and pulmonary venous pressures. A thorough review of these factors is available elsewhere.⁴

Indications for direct monitoring

Direct intravascular pressure measurement is an integral part of the critically ill patient’s monitoring plan. Central venous pressure must always be directly measured. Indications for CVP monitoring are discussed elsewhere (see Chapter 11, Central Venous Pressure Monitoring). Pulmonary arterial pressure is commonly estimated in small animals via Doppler echocardiographic evaluation using modified Bernoulli equation rather than measured directly. However, any time precise, continuous, or frequent measurements of PAP are desirable, it must be measured directly with a pulmonary arterial catheter (see Chapter 12, Cardiac Output Monitoring).

Systemic ABP can be measured either by a direct method using a peripheral arterial catheter system or by an indirect method using surface-applied pressure from an inflatable cuff partnered with flow-detection methodology (sphygmomanometry). The most common methods of indirect ABP measurement in dogs and cats are Doppler ultrasonic or oscillometric techniques, which are discussed in Chapter 10, Noninvasive Arterial Blood Pressure Monitoring.

Most human intensive care references consider direct pressure monitoring the “gold standard” against which indirect methods are compared.^{5–8} Direct pressure measurement is generally more accurate because detection and measurement of arterial pressure occurs directly in the vessel lumen. Cuff measurement techniques depend on blood *flow* to provide pressure measurement. Cuff methods are particularly poor in patients with low blood pressure secondary to myocardial failure or in cases where significant alteration in vascular resistance occurs (shock).^{5,7} In these cases, there can be large differences between values provided by indirect methods versus

Box 8.1 Indications for continuous direct arterial blood pressure measurement^{7,8}

- Hypotensive states with actual or potential tissue malperfusion
- Significant peripheral vasoconstriction
- Severely hypertensive states
- Vasodilator therapy
- Intraoperative and postoperative monitoring of critically ill patients

direct measurement. The Association for the Advancement of Medical Instrumentation (AAMI) has set performance standards for indirect ABP monitors in humans.⁹ Multiple veterinary studies have found that commonly used indirect methods fail to meet these standards in dogs and cats, though study methods did not match those of the AAMI.^{10–13} Therefore, continuous invasive pressure monitoring is indicated in many critically ill patients, and particularly those in shock states (see Box 8.1).

Despite the advantages noted with direct intravascular pressure measurement, this technique is more technically challenging and measurement errors are noted even when equipment is properly calibrated, leveled, and zeroed. The potential for inaccuracy is high for directly measured systolic and diastolic ABP. Directly measured mean arterial blood pressure (MAP) is more consistent and is a useful value to monitor, as it is the mean (continuous) pressure the organs “see.” The inherent pitfalls of even the gold standard of pressure monitoring underscore the importance of evaluating the whole patient—reported values that do not fit the whole clinical picture should be considered suspect, and the monitoring system investigated for sources of error. System inaccuracy and sources of error will be addressed specifically later in the chapter.

Both central venous and arterial pressure monitoring require large venous and arterial catheterization, respectively, and therefore carry the risks of these techniques. Complications, troubleshooting, and contraindications for central venous and arterial catheterization are discussed in Chapters 4 and 5, respectively.

Types of direct intravascular pressure monitoring systems

Direct pressure monitoring systems in veterinary practice are usually (normal saline) **monitoring systems**

consisting of fluid-filled tubing that connects a vascular catheter at or near the site of interest to a measuring device. Pressure waves move from the area of interest within the body (such as the cranial vena cava for CVP measurement), through the catheter and fluid-filled tubing either to a water manometer or to a pressure-transducer–processor–display system. Fluid must completely fill the system because fluid is relatively noncompressible and therefore transmits pressure waves well; air is too compressible to accurately transmit pressure waves. The fluid column between the site of interest and the measurement system must be unobstructed for the measuring system to provide accurate information.

The **water manometer** is the simpler of the two common measurement systems. The intravascular catheter is attached to a fluid-filled system of tubing and a manometer (see Fig. 8.1). There is a continuous fluid column between the catheter tip within the patient’s body and the manometer. The pressure at the catheter tip supports a column of fluid within the manometer; the pressure is then reported as the height of the fluid within the column. Therefore, when used properly, this technique allows for direct measurement of the pressure at the catheter tip. Measurements are manually performed intermittently. Water manometers are calibrated in centimeters of water (cm H₂O) and are generally only used for CVP measurement in veterinary practice. Peripheral ABP is too high to allow for practical measurement with a water manometer (ABP is higher than the pressure produced by the standard fluid column, so arterial blood would shoot out the top of the water



Figure 8.1 A water manometer being used to measure central venous pressure (CVP) in a ferret. Note the bottom of the manometer, at 0 cm in height, is being leveled with the ferret’s right atrium, establishing the right atrium as the zero reference point to which the CVP will be compared.

manometer). Water manometers generally report a single value, which is considered the mean intravascular pressure in the cavity (vessel, cardiac chamber) of interest at the reference height (more regarding reference height, leveling, and zeroing below).

For direct arterial blood pressure measurement, a fluid-filled catheter system attached to a pressure-transducer–processor–display system. This system is commonly used for systemic blood pressure because it is designed to accommodate the higher pressures generated in arterial blood vessels and permits continuous monitoring. There is a continuous fluid column between the catheter tip within the patient's body and a pressure transducer, which changes (transduces) the physical pressure wave into an electronic signal (see Fig. 9.1). The electronic signal is transmitted via a cable to a processor where the signal is amplified and processed into a real-time display of graphic waveform and numeric pressure values. Digital monitors generally display pressure in millimeters of mercury (mm Hg), and report systolic, mean, and diastolic pressures. Transducer–processor–display systems are often one component of patient monitors designed to simultaneously report multiple physiologic parameters (such as temperature, respiratory rate, electrocardiogram, multiple pressures, pulse oximetry; see Fig. 8.2). Blood pressure waveforms are displayed on the monitor screen in addition to the numerical pressure values that are displayed.

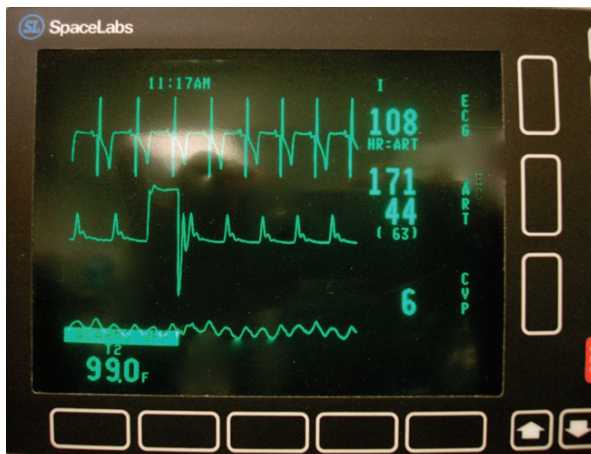


Figure 8.2 The screen of a multiparameter monitor displaying a simultaneous lead I ECG, heart rate, arterial blood pressure waveform, arterial blood pressure values, central venous pressure waveform, mean central venous pressure, and rectal temperature.

Measurement system components

Both the manometer system and the pressure-transducer–processor–display system begin with an intravascular catheter and specialized fluid-filled tubing. Beyond the tubing, the manometer system consists of a three-way stopcock, a water manometer, and a fluid reservoir (source). Beyond the tubing, the electronic system consists of at least one stopcock, a pressure transducer, a pressurized fluid reservoir, a flush device, cable connecting the transducer to a processor, and the processor–display, which is generally a single unit.

For the pressure (and waveform, with an electronic system) to be reported faithfully, each of the system components must meet certain physical and technical criteria. Even if each of the system components meets the required criteria, combining the components alters the physical properties such that the system as a whole may not provide accurate information. Therefore, the operator should be able to do the following: appropriately set up the system; recognize waveform patterns in electronic systems consistent with system malfunction; and test the electronic system for fidelity.

Intravascular catheter

The catheter's gauge, length, insertion site, orientation, and proximity to the vessel wall all affect reported pressures. Frictional resistance to fluid movement within the system increases as the catheter's inner diameter decreases or the catheter lengthens. Therefore, the ideal catheter for pressure-wave transmission would be short with a large bore. However, because the catheter itself increases resistance within the vessel at the insertion site and thus alters pressure, the catheter should ideally occupy no more than 10% of the vessel lumen.⁵ The reality for dogs and cats is somewhere in the middle: in general, the catheter selected should be one that reasonably fits within the vessel while minimizing the likelihood of vascular occlusion.

The catheter insertion site should be chosen for cleanliness, ease of placement, and maintenance. If a "stiff" catheter is used, it should remain straight along its entire path to minimize occlusion. Catheters for CVP monitoring are most commonly inserted into the external jugular vein in both dogs and cats and threaded into the cranial vena cava. Alternately, a long, flexible catheter may be placed into a saphenous vein and threaded cranially into the thoracic vena cava. Because saphenous-inserted thoracic caval catheters yield similar values as those placed through the external jugular vein,¹⁴ this is common practice in dogs and cats. Venous catheter insertion is discussed in detail in Chapter 4, Catheterization of the Venous Compartment.

Common insertion sites for arterial catheters in dogs are the perforating metatarsal artery (commonly called the dorsal pedal artery) and the radial, coccygeal, femoral, and auricular arteries. In cats, the femoral artery is most commonly used; the perforating metatarsal artery can also be used in larger cats for short durations (hours). Information regarding arterial catheter insertion can be found in Chapter 5, Arterial Puncture and Catheterization. Mean pressure is lower in distal portions of the arterial tree compared with the aorta because pressure-wave energy is lost as heat generated by frictional flow resistance along the vessel length. The pressure difference along the arterial tree is not considered a significant factor when selecting measurement sites.

Catheter tip orientation in relation to direction of blood flow affects the measured pressure value in both arterial and venous systems. A catheter tip facing into the blood flow (upstream) will measure a slightly higher pressure value than the actual intravascular pressure; a catheter tip facing away from the flow of blood (downstream) will measure a pressure value slightly lower than actual pressure. These differences are due to alteration in kinetic and potential energy at the catheter tip; these differences are relatively small in the vascular catheters commonly placed in veterinary practice.

Noncompliant tubing

The tubing that connects the intravenous catheter and pressure transducer must be made of specialized material to prevent pressure-wave energy from being absorbed (or “dampened”) by the tubing wall. This tubing is called rigid, noncompliant, or high-pressure tubing. The tubing must be completely filled with fluid without any air bubbles present. Use of standard fluid extension tubing between the vascular catheter and the pressure transducer will lead to significant error in the pressure and waveform measurement. Noncompliant tubing is less important for CVP measurement as the pressure values are significantly lower than systemic arterial pressure.

Water manometer

Water manometers are used for intermittent measurement of CVP, which is detailed in Chapter 11, Central Venous Pressure Monitoring. The water manometer is a plastic tube that is marked in centimeters along its length and has a “zero,” or reference, mark near the bottom of the column height (sometimes the bottom is the zero mark, as in Fig. 8.1). The zero mark is used as a reference site for aligning the manometer at the level of the right atrium prior to CVP measurements (see

Zeroing the Transducer, below). If a commercially produced water manometer is unavailable, one can create a water manometer from standard fluid extension tubing hand-marked with centimeters indicated along its length. The base of the water manometer fits into a three-way stopcock with the noncompliant-tubing-catheter system on the second port and the fluid reservoir on the third.

Although it is called a “water” manometer, the tubing column is filled with a biologically compatible crystalloid fluid to perform the measurement. The fluid reservoir for the water manometer is usually a 20-mL syringe filled with isotonic crystalloid. Full details regarding assembly and use of the water manometer pressure measurement system are available in Chapter 11, Central Venous Pressure Monitoring, specifically in Protocol 11.1, Intermittent Central Venous Pressure Measurement.

Pressure transducer

When an electronic pressure monitoring system is used, the noncompliant tubing is attached to a pressure transducer (see Fig. 9.1). The transducer has a pressure-sensitive membrane that distorts in response to pressure changes that are created when pressure waves strike it. The transducer converts the membrane distortion into an electronic signal using an integral electronic circuit that functions by the “Wheatstone bridge” principle.⁵ The generated electrical signal is transmitted to the processor–display unit by a shielded electrical cable. Most pressure transducers used in an integral veterinary medicine are classified as “disposable,” meaning that they are intended for single use in humans. Disposable transducers are sterile in their packaging and electronically precalibrated; some models come with high-pressure tubing attached.

Pressure transducers have a female adapter for connection to the fluid reservoir. The fluid reservoir is a bag of sterile isotonic crystalloid and administration set attached to the transducer. The bag of isotonic crystalloid is held under constant pressure such that a small volume of fluid continuously flushes through the system, preventing blood from flowing back and contaminating the monitoring system. The infused volume is generally 2–4 mL/hour (read manufacturer’s specifications for more precise value). Most transducers have a pigtail or lever protruding from the housing that activates an integrated “flush valve” that allows rapid infusion (“fast flush”) of fluid from the pressurized fluid bag through the system to clean the transducer “head.”

Handled with care, “disposable” transducers may be cleaned and reused following ethylene oxide



Figure 8.3 Part of a direct arterial pressure monitoring system in a dog. The dog's perforating metatarsal (dorsal pedal) arterial catheter is attached to a T-port, which is connected to noncompliant, fluid-filled tubing by a three-way stopcock. This stopcock is optional to the system and is in place here for serial arterial blood sampling. The noncompliant tubing is attached to the pressure transducer, which is secured to a yellow sandbag with medical tape; the sandbag provides a stable base for the transducer and helps raise the transducer's zero point (top of the transducer's stopcock) to the level of the patient's right atrium. The red pigtail on the transducer can be used to provide rapid system flush from the pressurized fluid bag (not pictured; its tubing leads from the pigtail to the fluid bag out the top of the frame). The grey cable runs from the transducer to the electronic processor and display monitor (not pictured).

resterilization. Resterilization can lead to damage of the pressure-sensitive membrane, which can lead to inaccurate pressure reporting. When a resterilized transducer is used, the operator should always perform a calibration prior to performing clinical measurements (see below for information about calibration).

Assembling the electronic direct pressure measurement system

A full list of supplies and step-by-step instructions required to assemble the electronic direct pressure measuring system are available in Protocol 8.1. Figure 8.3 shows a portion of a direct arterial pressure monitoring system in use.

Technical aspects of the electronic direct pressure measurement system

Direct pressure monitoring is relatively complex and far from foolproof. There are many technical points that must be carefully followed for accurate measurement. It

is extremely important that the operator understand the technical principles that affect its fidelity, how to properly configure and test the system for fidelity, and how to recognize and correct for system error.

Zeroing the transducer

To interpret measured pressure, it must be compared with a known pressure standard. The standard pressure against which direct intravascular pressure measurements are compared is barometric pressure (P_B), which is approximately 760 mm Hg (1031 cm H_2O) at sea level.

The electronic pressure measurement system is calibrated to standard pressure, or “zeroed,” by opening the transducer's stopcock port (see Fig. 9.1) to the atmosphere and depressing the “zero” button on the electronic pressure monitor (see Protocol 8.2 for instructions on zeroing a transducer). Zeroing the transducer to atmospheric pressure makes discussion of physiologic pressures easier (i.e., “The patient's MAP is 98 mm Hg” rather than “The patient's MAP is $760 + 98 = 858$ mm Hg”). Transducer zeroing should be done at initial system setup, any time system components are removed or replaced, or if any problems occur with system readings. If a transducer system fails to zero, the transducer, the monitor, or the connecting cable may be faulty (most often it is the transducer, especially when using a resterilized disposable transducer). These items should be sequentially changed until the faulty component is identified and replaced.

Water manometers are always open to the atmosphere and are thus inherently “zeroed” to atmospheric pressure. Therefore, no zeroing actions need to be performed on a water manometer, but its top must always remain open to the air or readings will be inaccurate.

Calibrating the system

After the system is zeroed to the atmosphere, it must be calibrated prior to use. To calibrate a measuring instrument is to compare and align its readings with known standards so that the measuring instrument can provide accurate readings. Calibrating a water manometer device simply involves comparing the markings on the manometer against those on a centimeter ruler. Such calibration is unnecessary on commercially available manometers, since they come premarked, but is required if regular fluid extension tubing is used to contain the fluid column.

Calibration is required for electronic systems each time the system is assembled and again any time problems arise. The transducer must be calibrated to confirm that when standard pressure is applied to the transducer,

Protocol 8.1 Assembling the electronic direct pressure measurement system for CVP or ABP monitoring**Items Required**

- Indwelling vascular catheter at the appropriate site
- ≤ 2 lengths of high-pressure tubing with locking fittings—keep as short as feasible
- Stopcocks with locking fittings—2 or fewer
- 1–2 sterile infusion plugs with locking fittings
- Sterile pressure transducer with integrated fast flush device
- Sandbag and medical tape, or other stabilizing device to fix the transducer
- Bag of isotonic crystalloid flush solution with standard administration set attached; standard extension set(s) as needed
 - Most commonly the flush solution is heparinized—ask clinician for any special considerations.
- Pressure bag of appropriate size for fluid bag used
- Monitor, with its power cord and transducer-to-monitor cable
- Power source

Procedure

1. Collect necessary supplies.
2. Plug the monitor in, turn it on, and attach the transducer-to-monitor cable to the monitor. Configure the monitor per manufacturer's instructions to display pressure from the socket into which you have inserted the transducer-to-monitor cable.
3. Perform hand hygiene and don clean examination gloves.
4. Aseptically prepare the patient's catheter port onto which the monitoring system will be attached. Flush the patient's catheter gently to ensure patency.
5. Heparinize the flush solution unless the clinician instructs otherwise. Standard dilution is 1 unit heparin per milliliter of flush. Mark the fluid bag to indicate its additives. Remove all air from the fluid bag by inserting a 22-gauge needle into the medication port and withdrawing gas until none remains. This minimizes the risk of air embolism from the pressurized system.
6. Attach a standard fluid administration set, with the roller clamp closed, to the heparinized fluid bag and squeeze the drip chamber until it is approximately half-full. Prime the fluid line with flush using gravity flow, then reclamp the fluid line. Handle fluid line aseptically such that the end remains sterile.
7. Insert the fluid bag into a pressure bag and inflate to approximately 300 mm Hg. Hang the pressurized bag and fluid line near the patient's area.
8. Remove the pressure transducer (see Fig. 9.1) from its packaging and manipulate all moving parts to ensure they move as intended. Handle aseptically such that all ports remain sterile. Inspect for any evidence of damage, particularly if the unit has been resterilized.
9. Attach the primed flush fluid line to the transducer. If necessary, attach one length of high-pressure tubing to the transducer on the patient side. If necessary, add more high-pressure tubing to reach the patient, to a maximum of two lengths total. Place a sterile, locking injection port to the transducer's zeroing (air) port.
10. The vented cap on a new transducer's stopcock should be thrown away, as its vents may allow contaminants to enter the system.
11. If blood sampling from the catheter is desired, place a stopcock between the high-pressure tubing and the catheter or catheter's T-port. The unused port should have a sterile, locking injection port attached.
12. Use the fast flush device on the transducer to prime the entire system until fluid drips from the end of the high-pressure tubing. Flush slowly to avoid air bubble formation in the tubing.
13. Attach the monitoring system to the patient's vascular catheter or the catheter's T-port.
14. Tighten all connections and engage all fitting locks to prevent inadvertent disconnection and subsequent blood loss.
15. Plug the monitor cable into the transducer and fix the transducer to its stabilizing device (i.e., sandbag with tape) such that the transducer's zeroing (air) port is at the appropriate height (at the level of the right atrium).
16. Zero and level the system. (See the sections Zeroing the Transducer and Leveling the Transducer, for more information.)
17. Turn all stopcocks in the system open to the patient.
18. Perform a fast flush test and make any necessary adjustments to the system to optimize the system's dynamic response. (See text for more information about dynamic response.)

Protocol 8.2 Zeroing a pressure transducer**Items Required**

- An assembled electronic direct pressure measurement system, attached to its monitor via the transducer cable

Procedure

1. Plug in and turn on the monitor. Configure the monitor such that the screen displays an option to zero the transducer (see manufacturer's instructions for specific steps).
2. Perform hand hygiene and don clean examination gloves.
3. Turn the transducer's stopcock "off" to the patient's side of the system so there is a continuous fluid column between the stopcock's injection cap and the transducer's pressure diaphragm.
4. Remove the injection cap from the stopcock, maintaining system asepsis.
5. Depress the "zero" button on the monitor.
6. The pressure tracing should be a flat line at the zero mark on the display.
7. If the pressure reads a number other than 0, or if the pressure tracing is not a flat line at the zero mark on the graph, the transducer may be faulty, in which case it must be replaced with another flushed, sterile transducer. If this does not remedy the issue, the transducer cable or monitor itself may be faulty and may need to be replaced.
8. Once the monitor displays "0" with the transducer's fluid interface open to the atmosphere via the stopcock, aseptically replace the injection cap and turn the stopcock "off" to the injection cap, opening the fluid column between the transducer and the patient end of the system.

the correct pressure is reported. Calibration should be conducted with the transducer attached to the monitor, as instructed by the monitor's manufacturer. Alternately, a water manometer filled with fluid to a set height may be attached to the transducer's stopcock and the display's reading compared with the known applied pressure from the fluid column, remembering that 1.36 cm H₂O exerts the same pressure as 1 mm Hg. Calibration problems can be due to transducer, monitor, or connecting-cable problems.

Leveling the transducer

A transducer is both zeroed, to eliminate the influence of atmospheric pressure from vascular pressure readings, and **leveled**, to eliminate the influence of *gravity*. For both CVP and ABP, the "level" reference point is the right atrium (RA); thus the RA is called the **zero reference point**. To level the measuring system, the transducer should be placed at the height of the RA and zeroed at that point by opening the three-way valve to air. This step should be performed *prior to every measurement* (see Protocol 8.3). An external anatomic landmark that correlates well with the RA reference point is the sternum of the cat or dog lying in lateral recumbency. As stated in Chapter 11, Central Venous Pressure Monitoring, in a sternally recumbent animal the RA lies at a point roughly 40% the height of a vertical line that extends from the sternum to the top of the dorsal spinous process just caudal to the shoulder.

Once the transducer is zeroed at RA height, it must remain at RA height for every measurement. The reason for this requirement may best be explained by example. In a patient undergoing CVP monitoring, if the transducer falls below RA level by 10 cm (approximately 4 inches), a 10-cm blood fluid column is exerting pressure on the transducer in addition to the actual CVP. Though this additional pressure is not a change in CVP, the transducer will "see" both the actual CVP and the additional 10-cm blood column and will report a value equal to the CVP plus 10 cm H₂O. The opposite is true if the transducer ends up higher than its zero reference point—the pressure reported will be falsely low in such cases (see Fig. 8.4). For peripheral ABP monitoring, such changes are less likely to create clinical confusion because as a proportion of the pressure of interest, the error is smaller. For instance, while a difference of 10 cm H₂O (7.4 mm Hg) in CVP can greatly influence a clinician's decision-making process, that same 7.4 mm Hg difference in MAP is less likely to cause an error in clinical judgment. This example underscores the importance of evaluating the whole clinical picture before making decisions, and the importance of proper transducer leveling at the zero reference point (the RA) prior to *every measurement*, particularly when monitoring CVP.

Dynamic response of the system

Intravascular pressures are pulsatile in nature. Reflection of pressure waves through the vessel creates multiple

Protocol 8.3 Leveling a pressure transducer at the zero reference point (the right atrium)**Items Required**

- An electronic direct pressure measurement system, attached to a powered-on monitor and to the patient
- Carpenter's level
- Piece of string

Procedure

1. Configure the monitor such that the screen displays an option to zero the transducer (see manufacturer's instructions for specific steps).
2. Perform hand hygiene and don clean examination gloves.
3. Turn the transducer's stopcock "off" to the patient's side of the system so there is a continuous fluid column between the stopcock's injection cap and the transducer's pressure diaphragm.
4. Remove the injection cap from the stopcock, maintaining system asepsis.
5. Ensure the transducer's fluid-air interface is at the height of the right atrium. Assess this level using the string and carpenter's level for accuracy—this will enhance accuracy and repeatability from one operator to the next for repeated measures.
 - a. With the patient in lateral recumbency, RA height is approximately the height of the sternum.
 - b. For a dog or cat in sternal recumbency, the RA is approximately 40% the distance from the sternum to the dorsal spinous process just caudal to the scapula.
6. Secure the monitor at this height if performing continuous monitoring and the patient is immobile.
7. Depress the "zero" button on the monitor.
8. The pressure tracing should be a flat line at the zero mark on the display.
9. If the pressure reads a number other than 0, or the pressure tracing is not a flat line at the zero mark on the graph, the transducer may be faulty, in which case it must be replaced with another flushed, sterile transducer. If this does not remedy the issue, the transducer cable or monitor itself may be faulty and may need to be replaced.
10. Once the monitor displays "0" with the transducer's fluid-air interface open at RA height, aseptically replace the injection cap and turn the stopcock "off" to the injection cap, opening the fluid column between the transducer and the patient end of the system.
11. Allow the system to equilibrate and record the measured value.

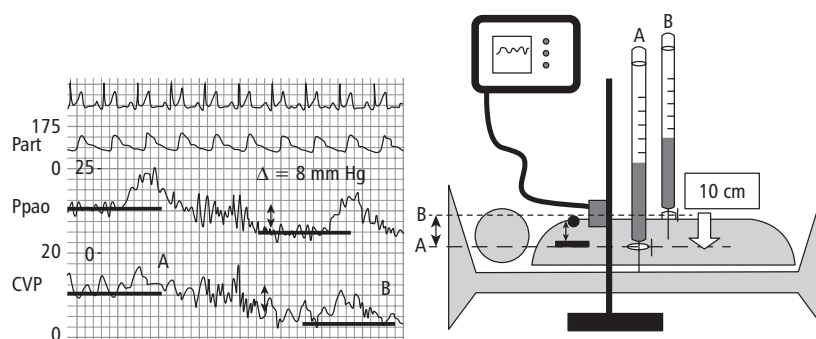


Figure 8.4 Illustration of what happens when the patient's right atrium (RA; level shown by dashed line A) falls below the transducer's zero reference point (at the level of dotted line B) as the bed is lowered: (left) change in recorded pressures (P_{art} , peripheral arterial; P_{pao} , pulmonary arterial occlusion [wedge]; CVP, central venous pressure); (right) water manometers (tubes A and B) that demonstrate how lowering the RA relative to the transducer lowers the measured pressures. In this example, the bed was lowered by 10 cm, which means that the patient's RA (the proper level) is 10 cm lower than the current zero reference point (B), which translates to a pressure drop of 10 cm H_2O , or approximately 8 mm Hg. Figure reprinted and legend adapted from Magder S, Invasive intravascular hemodynamic monitoring: technical issues, *Critical Care Clinics*, Vol. 23, pp. 401–414, 2007, with permission from Elsevier.

oscillating waves of different amplitude and frequency (Fourier series) that summate to create the observed waveform. An intravascular pressure monitoring system must have the physical properties required to measure pressures within the expected range and must be able to respond adequately to physiologic pressure pulsations. The ideal system would report the pressure waves of interest and no others. The ability of a system to accurately display the shape and amplitude of the pulse pressure waveform is determined by the system's **dynamic response**, also called the system's frequency response. The system's dynamic response is determined by its physical properties, specifically its mass, elasticity, and friction.¹⁵ Dynamic response is discussed in terms of natural frequency and damping coefficient, both of which are measurable and have significant impact on a waveform's appearance.

Effect of a system's natural frequency

When stimulated, every structure naturally vibrates at a characteristic frequency, which is expressed in cycles per second or hertz (Hz). This frequency is called the structure's **natural frequency**, fundamental frequency, or resonant frequency. Adding components together (as in connecting a catheter, noncompliant tubing, and transducer) alters a system's natural frequency. It is important that the natural frequency of a fluid-filled monitoring system not coincide with the frequency of physiologic pressure waves, because frequency overlap causes summation (from the patient and the system) and results in exaggerated waveforms and numerical values. Exaggerated results are due to *too low a system natural frequency*, and such exaggeration leads to what is often called overshoot, **ringing**, or resonance of the waveform.¹⁵ Ringing causes pointy, spiked waveforms, falsely high systolic pressure readings, and falsely low diastolic pressure readings.

A fluid-filled monitoring system will have optimal responsiveness if its natural frequency is as high as possible. Though individual materials made for intravascular pressure monitoring are designed with this principle in mind, once a catheter–tubing–transducer system is assembled, the natural frequency drops to minimal requirements for humans.⁵ Because dogs and cats generally have pulse rates that exceed humans', their pulse pressure waveforms have a higher frequency than people's, and thus almost certainly overlap the natural frequency of most measurement systems. This overlap means that without correction through damping (see below), our patients' pulse pressure waveforms will almost always ring, systolic pressures will be falsely high, and diastolic pressures will be falsely low. One way to

maximize a system's natural frequency is to keep the system as simple as possible, for instance with as short a tubing length and as few components as possible.

Effect of a system's damping coefficient

Damping is loss of the pulse pressure energy between the catheter tip and the transducer. Damping is due to frictional resistance along the system's length, absorption of energy by the tubing and other materials, and air bubbles, which are more compressible than fluid. The more damped a system is, the more quickly it returns to "zero" after an applied stimulus due to this energy loss.

Damping in a pressure system is measured and expressed as the **damping coefficient**; the higher the coefficient, the more significant the damping. An **overdamped** pressure waveform has slurred upstrokes and downstrokes, loss of detail, and a generally flattened appearance; overdamped systems cause a falsely narrowed pulse pressure with falsely low systolic and falsely high diastolic pressure readings (see Fig. 9.7). Conversely, **underdamped** waveforms contain nonphysiologic points and spikes, extra waves, and appear exaggerated—they are overshoot or have excessive ringing, as discussed previously regarding natural frequency. Underdamping causes falsely high systolic and falsely low diastolic pressure readings. A system with an infinitely high natural frequency does not require damping to produce accurate results (see Fig. 8.5), but because available systems have natural frequency overlap as noted previously, they usually require some operator-implemented damping.

Determining the system's dynamic response

Many catheter–tubing–transducer systems have weak natural frequencies to faithfully reproduce physiologic pressure data; thus, these systems are inherently underdamped. Within a certain range, adjusting the system's damping can help produce more accurate values and waveforms (Fig. 8.5). To know whether a system has adequate dynamic response, its natural frequency and damping coefficient should be determined and plotted onto a graph such as Figure 8.5. These properties are measured by performing a fast flush or "**square wave**" test on the system. When the transducer's fast flush device is activated, the transducer–tubing–catheter system is exposed to the high pressure in the flush fluid bag (300 mm Hg). To perform the test, the fast flush device should be opened briefly (<1 second) and released quickly several times to produce multiple square waves for analysis. The high pressure should appear on the recorder as a square waveform as shown in Figure 8.6. A normal square wave has a nearly 90° upstroke, a flat

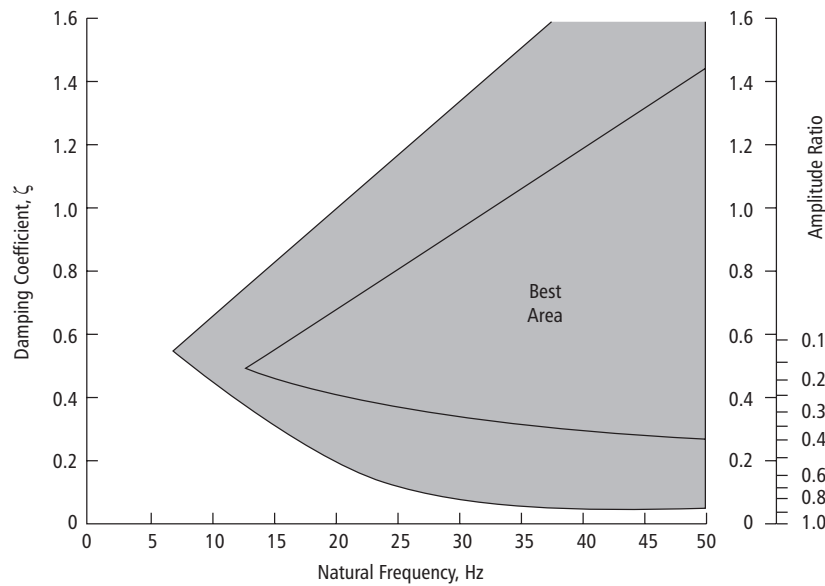


Figure 8.5 Use of natural frequency and damping coefficient to determine if the catheter–tubing–transducer system is producing accurate pressure waveforms and values. Plot the system’s frequency and damping coefficient; the intersection falls in the gray area for an adequately responsive system. Overdamped readings will fall above the gray area and underdamped readings will fall below the gray area. This figure and its legend were altered and used here with permission from *Hemodynamic Waveform Analysis*; Ahrens RS, Taylor LA; Technical considerations in obtaining hemodynamic waveform values, pp. 209–258, copyright Saunders 1992.

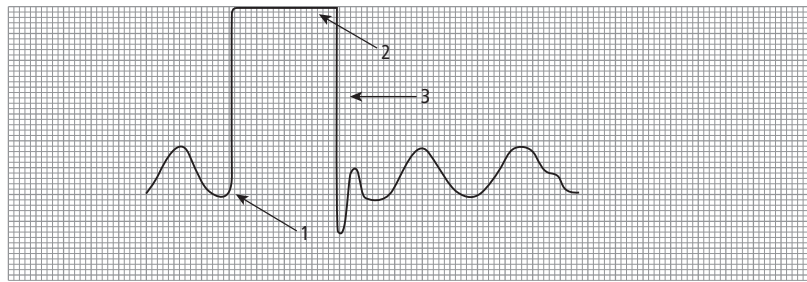


Figure 8.6 Performing a square wave test. The fast flush device is activated at point 1. Squaring of the waveform occurs as the transducer is exposed to 300 mm Hg pressure from the flush fluid bag and has a flat plateau (2) with right angles on both sides. A rapid downstroke occurs as the fast flush device is released (3). This test should be performed several times and the square waves analyzed to determine the system’s natural frequency and damping coefficient. These values can then be plugged into the graph in Figure 8.5 to determine whether the system has adequate dynamic response. This figure used with permission from *Hemodynamic Waveform Analysis*; Ahrens RS, Taylor LA; Technical considerations in obtaining hemodynamic waveform values, pp. 209–258, copyright Saunders 1992.

plateau at 300 mm Hg, and a rapid downstroke as the fast flush device is released. The top is squared, as the test’s name implies. Protocol 8.4 describes how to determine a system’s natural frequency and damping coefficient, and thus the system’s dynamic response, using waveforms from square wave tests.

A square wave test should be performed when the system is initially assembled and any time there are questions about the fidelity of measurement results.

Optimizing dynamic response by altering a system’s natural frequency and damping coefficient

There are many steps one can take to optimize a system’s dynamic response by either increasing the natural frequency or altering the damping coefficient. Underdamping is far more common in dogs and cats than overdamping. An example from a ringing, underdamped

Protocol 8.4 Determining the dynamic response of a fluid-filled monitoring system (Figs. 8.7–8.9)**Items Required**

- An electronic direct pressure measurement system, attached to a powered-on monitor (with printer) and to the patient
- Calculator
- Straightedge (i.e., ruler)
- Pen and paper

Procedure

1. Ensure the pressure bag containing the flush fluid is pressurized to 300 mm Hg.
2. Perform hand hygiene and don clean examination gloves.
3. Perform a square wave test during diastole by activating the transducer's fast flush device briefly (for <1 second) and quickly releasing the device. A square wave should appear on the monitor—print this waveform with 1–2 seconds of strip on either side. Repeat this process 3–5 more times and collect the square waves for analysis.
4. Evaluate the square waves to determine the system's natural frequency. The natural frequency is the frequency with which the waveform oscillates after the fast flush device is released. For example: The paper speed in Figure 8.7 is 25 mm/sec. Note the number of blocks between oscillation peaks—in this case just over 2 blocks between oscillations. Divide the paper speed by the number of blocks to determine the system's natural frequency: $(25 \text{ mm/sec}) \div (2 \text{ mm}) = 12.5 \text{ cycles/sec} = 12.5 \text{ Hz}$, which is a minimally acceptable natural frequency for measurements in humans.
5. Determine the frequency on all of the square waves collected, and average the values to reach the mean natural frequency. Use this mean natural frequency in the dynamic response graphic plot (see step 8).
6. Evaluate the square wave to determine the system's damping coefficient. This is done by comparing the length of two successive oscillations and dividing the second (smaller) oscillation by the first (see Fig. 8.8). The number generated is called the amplitude ratio. For example: The paper speed in Figure 8.8 is 25 mm/sec. Measure the length of two successive oscillations (i.e., from peak to valley, and from that same valley to the next peak). Divide the smaller oscillation by the larger to determine the amplitude ratio. Here, $6 \text{ mm} \div 28 \text{ mm} = 0.21$, which is the amplitude ratio.
7. Determine the amplitude ratio on all of the square waves collected and average the values to reach the mean amplitude ratio. Use the graph in Figure 8.9 to determine the damping coefficient from the amplitude ratio.
8. Use a straightedge to plot the mean natural frequency against the damping coefficient on the graph in see fig 8.5 above. If the intersecting point falls into the gray area, the system is adequately responsive. Overdamped readings will fall above the gray area and underdamped readings will fall below the gray area.

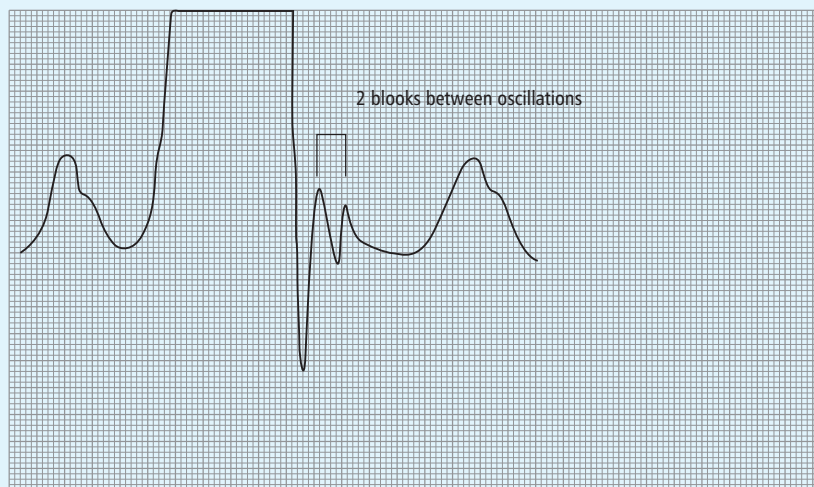


Figure 8.7 Step 4, determine system's natural frequency. Paper speed 25 mm/sec. First, note the number of blocks between oscillation peaks—in this case just over 2 blocks between oscillations. Then divide the paper speed by the number of blocks to determine the system's natural frequency: $(25 \text{ mm/sec}) \div (2 \text{ mm}) = 12.5 \text{ cycles/sec} = 12.5 \text{ Hz}$, which is a minimally acceptable natural frequency for measurements in humans. Figure altered and used with permission from *Hemodynamic Waveform Analysis*; Ahrens RS, Taylor LA; Technical considerations in obtaining hemodynamic waveform values, pp. 209–258, copyright Saunders 1992.

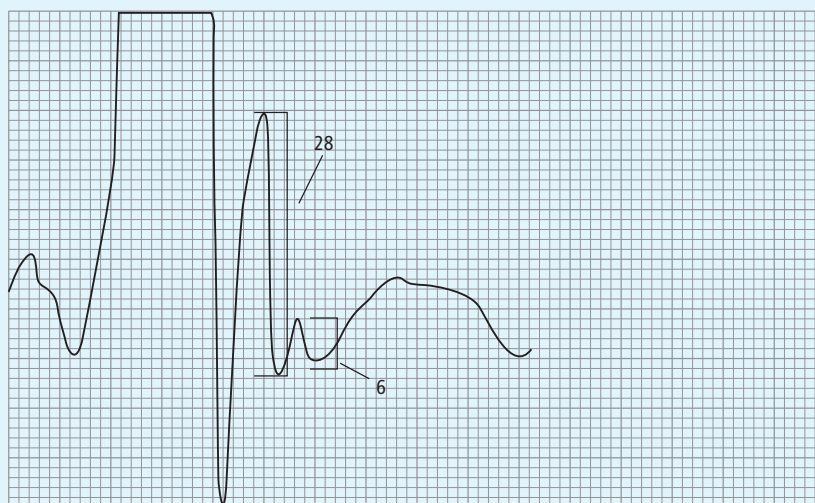


Figure 8.8 Step 6, calculating the amplitude ratio to determine damping coefficient. Paper speed 25 mm/sec. First measure the length of two successive oscillations (i.e., from peak to valley, and from that same valley to the next peak). Then divide the smaller oscillation by the larger to determine the amplitude ratio. Here, $6\text{ mm} \div 28\text{ mm} = 0.21$, which is the amplitude ratio. Figure altered and used with permission from *Hemodynamic Waveform Analysis*; Ahrens RS, Taylor LA; Technical considerations in obtaining hemodynamic waveform values, pp. 209–258, copyright Saunders 1992.

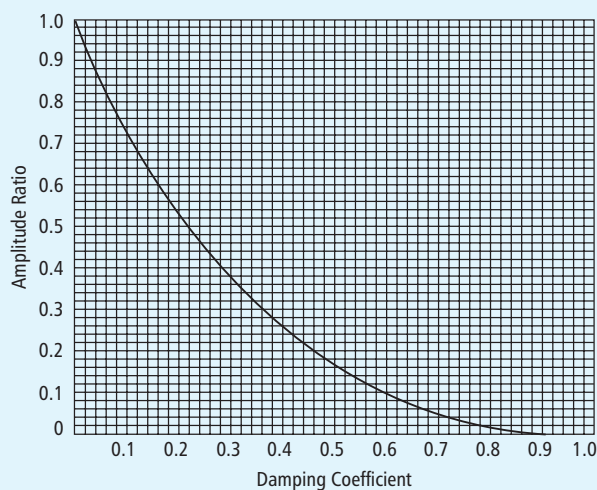


Figure 8.9 Step 7, Use this graph to determine the damping coefficient from the amplitude ratio. Figure altered and used with permission from *Hemodynamic Waveform Analysis*; Ahrens RS, Taylor LA; Technical considerations in obtaining hemodynamic waveform values, pp. 209–258, copyright Saunders 1992.

system and one from an overdamped system are shown in the square wave tests in Figure 8.10 Further discussion of the effects of underdamping and overdamping on ABP waveforms and values is found in Chapter 9, Direct Systemic Arterial Blood Pressure Monitoring. Measures

to take that may help optimize the system's dynamic response are listed in Box 8.2.

Though it is important for accurate pressure readings and interpretation, it is unclear that the systems in clinical use today commonly have adequate dynamic

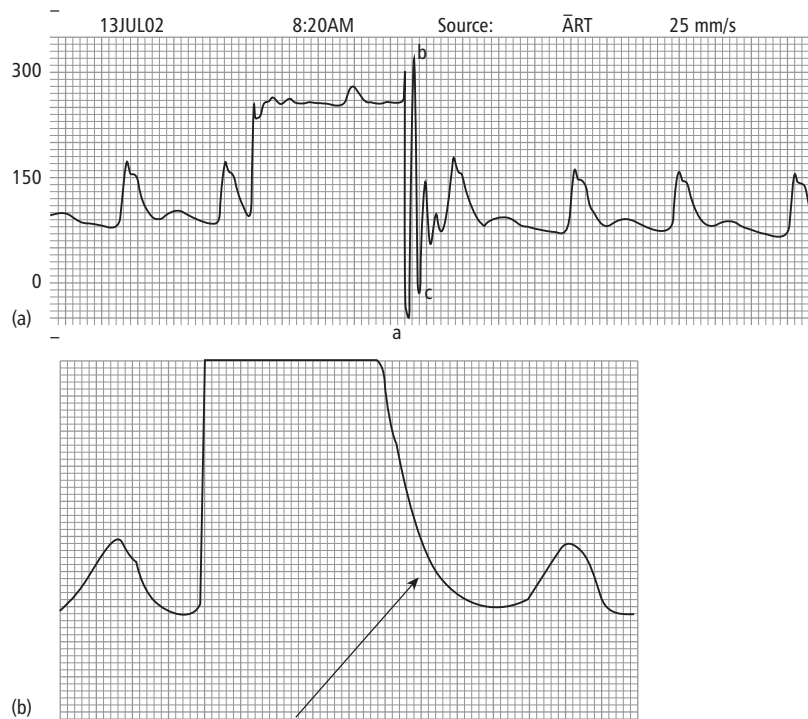


Figure 8.10 Square wave test waveforms: (a) Example of a square wave generated from an underdamped system. Note the spiky, pointy nature of the arterial pressure waveform and the subjectively apparent excessive ringing of the square wave test compared with the normal example in Figure 8.6. To confirm underdamping, calculate the amplitude ratio: measure vertically from point a to point b, 37 mm; then from point b to point c, 34 mm; then divide the smaller length by the larger, $34 \text{ mm} \div 37 \text{ mm} = 0.92$. An amplitude ratio of 0.92 corresponds to a damping coefficient of less than 0.1, which is extremely low and corroborates the impression of an underdamped waveform. (b) Example of a square wave generated from an overdamped system. Note the slurred arterial pressure waveform with no apparent dicrotic notch, and the slurred downstroke of the square wave test with lack of oscillations at baseline (arrow). It is very difficult to determine damping coefficient in the absence of any measurable oscillations, but the damping coefficient here is high, probably greater than 0.6.¹⁶ Part (b) is altered and used here with permission from *Hemodynamic Waveform Analysis*; Ahrens RS, Taylor LA; Technical considerations in obtaining hemodynamic waveform values, pp. 209–258, copyright Saunders 1992.

Box 8.2 Measures to help optimize a system's dynamic response

The corrections for overdamped and underdamped systems are similar, so the following actions can be tried in the case of either problem.^{16,17}

- Simplify the system as much as possible to increase its natural frequency:
 - Remove as many lengths of tubing between the catheter and the patient as possible such that the tubing does not exceed 3–4 feet in length.
 - Remove any unnecessary stopcocks.
 - Consider removing any T-ports or other connections.
- Ensure only noncompliant tubing is present between the catheter and the transducer, particularly for arterial pressure monitoring.
- Check for and remove any visible clots or air bubbles.
- Check tubing for kinks or occlusions.
- Gently aspirate and flush the catheter to assess for occlusion.
- If the catheter is <18-gauge (or 7 Fr), compliant, or long, consider replacement with a shorter, larger-bore, stiffer catheter.
- If the waveform is underdamped, consider insertion of a damping device into the system.

response.¹⁵ It is important to remember both this source of inaccuracy and that *pressure waveforms and pressure values are altered by simple changes the operator makes to the system* such as alterations in system components and damping. Thus, the gold-standard invasive pressure monitoring system is also inherently flawed by inevitable operator manipulations.

Summary

Fluid-filled monitoring systems provide important information in patients with cardiovascular instability. Though they are more complicated than noninvasive measurement techniques, they can provide continuous monitoring. As with any technique, one becomes more proficient with practice and use. A solid understanding of the principles behind fluid-filled monitoring systems allows the clinician to make the most of these systems' capabilities and avoid common technical errors. It is important to remember that there is a potential for technical error if equipment is incorrectly configured, inaccurately zeroed, uncalibrated, or poorly leveled. It is also important to remember that actions we take to optimize the dynamic response of a system (altering natural frequency and damping) can change the reported pressure results and mislead the clinician.

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