Review Articles

Hemodynamic Monitoring: From Catheter to Display

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Introduction

Hemodynamic variables are almost always monitored in the modern intensive care unit, and although the techniques are widely used, problems frequently occur that are neither understood nor detected by the clinical staff. Several excellent survey articles and texts have been written especially for blood pressure monitoring [1–5]. This monograph was prepared for practicing physicians and nurses to give them a better understanding of the theory of hemodynamic monitoring. Clinically applicable and easy to follow suggestions are given to minimize measurement errors.

Components of Pressure Monitoring Systems

The components of a direct blood pressure monitoring system for critically ill patients are shown in figure 1 and described below. Those numbered 1 through 7 in the figure and listed below are known as the 'plumbing system' and must always be sterile because there is direct contact with the patient's circulatory system. Usually these components are disposable or single use items and are often discarded after 24–48 h to minimize the risks of patient infection. Components 8 through 11 in figure 1 are used for processing and displaying pressure waveforms and derived hemodynamic parameters.

(1) Catheter. Arterial and pulmonary artery catheters provide access to the patient's blood vessels to (a) monitor intravascular pressure and (b) provide a site for blood gas analysis and other tests. These catheters are typically placed by percutaneous method, either by the Seldinger over the needle technique [6], or by introducing the catheter through a needle. It is desirable to have the arterial catheter tip advanced to a central location – i.e., thoracic aorta or subclavian artery – since measurements at peripheral
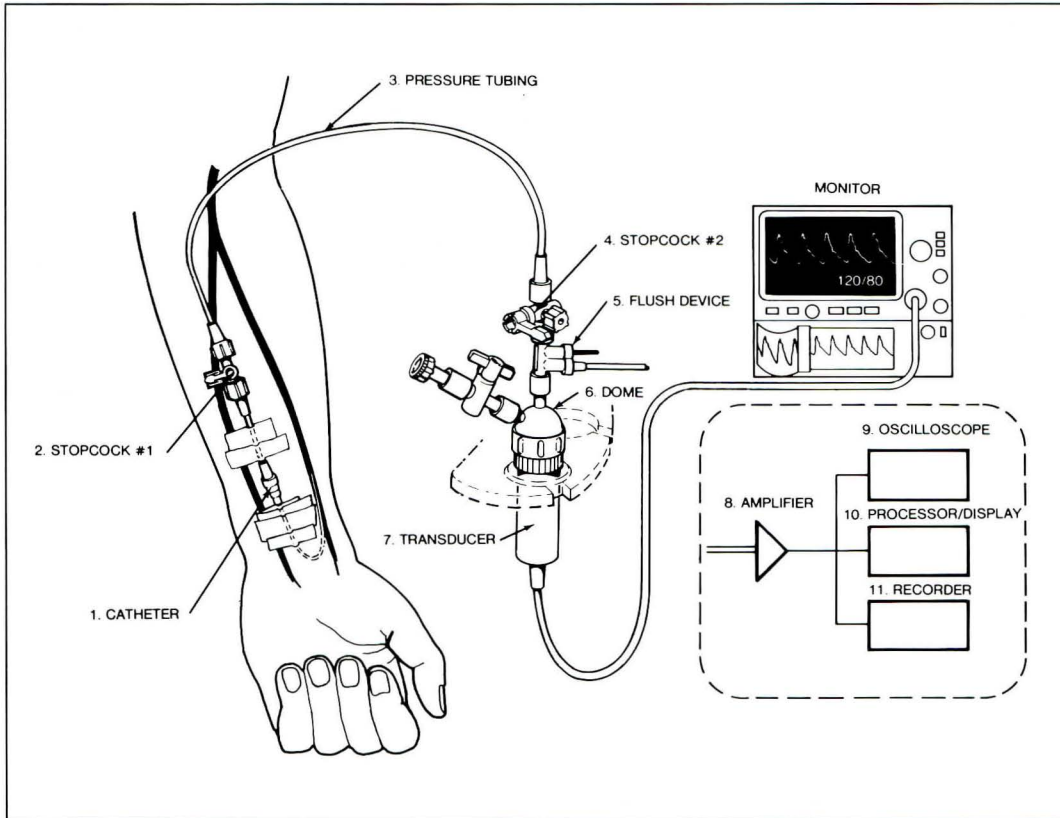
Fig. 1. Illustration of the eleven components used in a direct-pressure monitoring system. The components and problems associated with them are independent of whether the catheter is in an artery (radial, brachial or femoral) or in the pulmonary artery. The components are: (1) the catheter system, (2) stopcock No. 1 for sample withdrawal, (3) pressure extension tubing used to attach the catheter to the transducer, (4) stopcock No. 2 located near the transducer, (5) continuous-flush device, (6) transducer dome – usually the disposable type with a membrane, (7) pressure transducer, (8) amplifier system, (9) oscilloscope, and (10) digital processing and display module. [From ref. 4, with permission.]

sites may not accurately reflect pressures measured at a central arterial location. Central arterial pressure provides the driving force for blood flow to the vital organs – the heart, the kidney and the brain [4, 7–9].

(2) Sampling stopcock: Stopcock No. 1 (component 2 in fig. 1) is used as a site for withdrawal of blood for analysis. This stop-cock is either connected directly to an arterial or pulmonary artery catheter or may be isolated by 6–12 in of pressure tubing. Most stopcocks used in clinical monitoring are disposable, have Luer-Lock fittings to prevent disconnection, and are made of of clear plastic so air bubbles can be readily seen and easily removed.

Stopcocks with compliant seals are unsatisfactory for use in direct blood pressure monitoring systems because they act as shock absorbers, smoothing out the pressure waveform so that systolic pressure is underestimated and diastolic pressure is overestimated (fig. 7c).

When filling the plumbing system with fluid, precautions must be taken to be sure all central switching cavities of the stopcock are filled and entrapped air bubbles are removed. Because they are especially vulnerable sources of patient contamination, especially during blood sampling and the zeroing process, stopcocks must be handled with extreme care: ports not in active use should be covered with sterile caps and personnel should never touch open ports [10, 11].

(3) Pressure tubing. Catheter and stopcock are normally attached to the flush device and transducer by pressure tubing that should be noncompliant, i.e., rigid or semirigid pressure tubing, and as short in length as possible. Long lengths of tubing must be avoided, even though transducers are frequently mounted at the bedside with 6–8 ft of tubing attached to them. These long lengths of tubing have been shown to cause serious distortions because they decrease dynamic response [12, 13].

(4) Transducer stopcock No. 2. This stopcock is usually put in place to allow disconnection of the flush device and transducer from the patient when the patient is moved or when initially filling the system with fluid.

(5) Continuous flush device. This device is used not only to initially fill the pressure monitoring system with fluid, but also to prevent clotting in the catheter by continuously flushing fluid through the system at a rate of 1–3 ml/h. More than 10 different models of flush devices are available, most of which prevent catheter clotting by continuous flush. However, only a few permit adequate dynamic testing with a fast flush because their fast-flush valves close too slowly. Some have large volume displacements and seriously degrade the plumbing system's dynamic response performance [12–15]. Methodology presented below should make it possible for the clinical staff to evaluate system dynamics as a guide to the selection of appropriate flush devices.

(6) Transducer dome. Disposable plastic diaphragm transducer domes (see fig. 1) isolate the nonsterile transducer from the patient so the transducer can be reused rapidly without resterilization. However, if the dome is not properly positioned on the transducer, the pressure waveform is not well coupled to the pressure transducer with consequent distortion of the waveform. Most disposable diaphragm domes are more easily coupled by adding a few drops of sterile saline onto the transducer diaphragm while pressurizing the dome [16, 17].

(7) Pressure transducer. These are available in a variety of sizes and shapes. Conventional pressure transducers are usually resistive devices (e.g., a Wheatstone Bridge) that convert the movement of the sensing diaphragm to an electrical signal. The diaphragm is moved by the pressure pulses that go through the catheter tubing system. These movements are converted to an electrical signal of 5.0 μV/V of excitation/mm Hg of pressure applied. For a typical patient with a pressure of 120/80 mm Hg, the output from a transducer with a 5.0 V excitation is 0.003 V (3 mV) while the pulse pressure is only 1 mV. Small signals from the transducer are amplified and processed by the bedside monitor.

Table I. Factors to be considered when selecting a pressure transducer

Ruggedness	withstand repeated clinical use and abuse
Stability	zero and sensitivity are stable with time and temperature
Interchangeability	do they operate with all of your monitors?
Easy to set up	are they easy to fill and use? consider cost, inconvenience, and problems of filling, coupling and calibration
Cost	also consider setup time, repair, cleaning, sterilization, and cost of disposables

Table II. Desirable specifications for pressure amplifier systems

Bandwidth	DC to 50 Hz
Zero stability	± 1 mm Hg (with time and temperature)
Sensitivity stability	$\pm 1\%$ (with time and temperature)

Standards developed by manufacturers have greatly simplified the clinical user's efforts for selection of transducers which are compatible with specific monitors [18]. Several disposable pressure transducers are now available. They are smaller, have better technical qualities (i.e., greater stability and accuracy, lower and more stable zero offset) and can also better withstand the rigors of clinical use such as bumping and dropping than currently available reusable transducers [19]. They also cost less than $15 and their price is expected to decrease as they are more widely used. Accordingly, reusable transducers will become obsolete (table I).

(8) Amplifier system. The output voltage required to drive an oscilloscope or strip recorder is provided by an amplifier system between the transducer and the display [20]. Transducer excitation may be either direct current (DC) or alternating current (AC). The usual voltage used is 4–8 V RMS (root mean square).

Amplifiers typically increase the amplitude of the pressure signal 250–1,000 times. The amplifier system may include low pass filters that filter out unwanted high frequency signals. Pressure amplifier frequency response should be 'flat' from 0 to 50 Hz to avoid waveform distortion. Processing and recording systems should accurately reproduce 6–10 harmonics of the pressure waveform [1, 14, 15]. A heart rate of 180 beats/min requires a bandwidth of at least 30 Hz. As pulse pressure is always superimposed on a steady mean pressure, a DC response is needed. Specifications for amplifier systems are given in table II.

(9) Oscilloscope. Pressure waveforms are best visualized on a calibrated oscilloscope to ascertain the dynamic response of the pressure monitoring system. Although all pressure waveforms are not displayed continuously and simultaneously, the ability to display each pressure waveform on an oscilloscope makes it possible to detect respiratory variations, artifacts, and distortions that enter the pressure signal.

(10) Digital processing and display. Although digital displays have serious limitations, they provide a simple method for presenting quantitative data from the pressure waveform [20, 21]. Therefore, this type of display is incorporated into most modern

pressure-monitoring equipment. Systolic, diastolic, as well as mean pressure and heart rate are computed from the arterial pressure waveform.

(11) Strip chart recorders. These are commonly, but not universally included as a part of direct pressure-monitoring systems. Strip chart recorders are invaluable for documenting dynamic response characteristics, respiratory variations in pulmonary artery pressures, and aberrant rhythms and pressure waveforms [22–27].

Static Calibration

Zeroing and calibrating the transducer are the two most important steps in setting up the direct pressure-monitoring system. Yet, they are the steps in which mistakes are made most often.

Zeroing the Transducer

The accuracy of blood pressure readings depends on establishing an accurate reference point from which all subsequent measurements are made. The patient's midaxillary line (right heart level) is the reference point most commonly used. The zeroing process is used to compensate for offset caused by hydrostatic pressure differences, offset in the pressure transducer, amplifier, oscilloscope, recorder and digital displays.

Zeroing is accomplished by opening an appropriate stopcock to atmosphere and aligning the resulting fluid-air interface with the midaxillary reference point (see fig. 2a, b). Two methods of system zeroing are in common use.

(1) The transducer-amplifier display system can be zeroed by opening the transducer to atmosphere through stopcock No. 2 (see fig. 2a). The fluid-air interface point of stopcock No. 2 is then moved vertically to be level with the midaxillary line (fig. 2a).

(2) Stopcock No. 1 can be opened to expose the transducer and tubing system to atmospheric pressure. When this is accomplished the fluid-air interface point of stopcock No. 1 should be placed at the midaxillary line (see fig. 2).

One must be sure to zero the transducer to the fluid-air contact point and not to the transducer diaphragm. Improper zeroing such as this is one of the most common errors in blood pressure measurement, a consequence of which can be major therapeutic decision errors. In addition, any change in the transducer's location – either by moving it up or down, or rotating it on its axis – after its zero position has been established, alters the zero. Each centimeter the zero location is deviated from the midaxillary reference point[1] results in an error of 0.74 mm Hg. However, it is possible to fix the transducer at the midaxillary line by mounting it on the patient's upper arm (fig. 3a, b). This location not only optimizes the system's dynamic response characteristics by keeping the tubing length very short, but the transducer will remain at or near the midaxillary line while the patient is lying in bed.

An alternative method for zeroing the transducer is to mount it on a pole, placing

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[1]   A word of caution: when using a plastic membrane dome that couples to the transducer, the diaphragm must be positioned slightly below the patient's midaxillary line or negative pressures will not be measured correctly. The plastic membrane pulls away from transducer's metal diaphragm if a vacuum greater than −20 mm Hg (−20 mm Hg = 27 cm H_2O) is applied. Consequently, the transducer cannot sense the correct pressure signal.
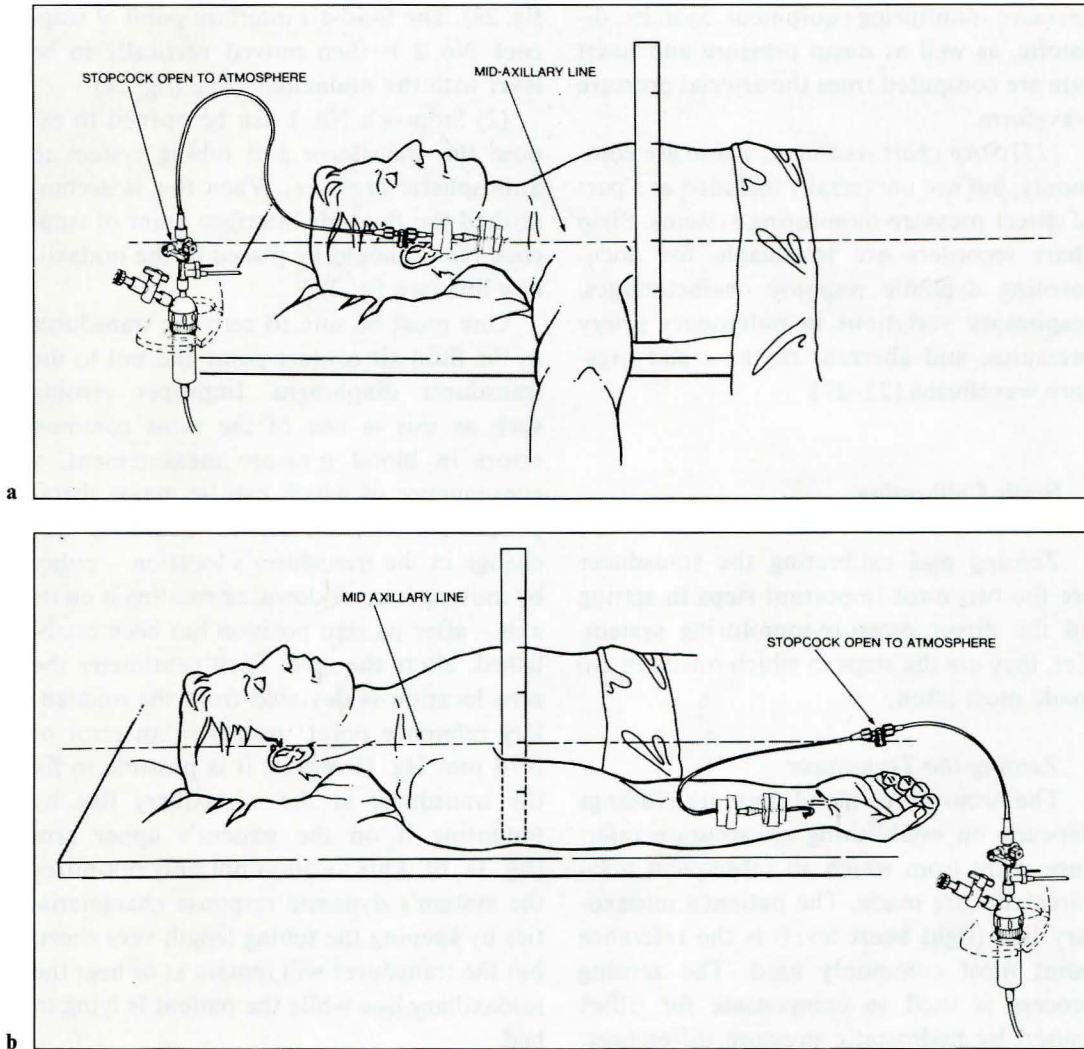
Fig. 2. Two methods of zeroing a pressure transducer. Note, the place at which the water-air interface occurs should always be at the midaxillary line when zeroing. **a** Placing the stopcock near the transducer at the midaxillary line. **b** Placing the stopcock nearer the catheter at the midaxillary line. Note: Size of transducer and plumbing components were enlarged for illustration purposes. [From ref. 4, with permission.]

the stopcock (see fig. 2b) at the end of the interconnecting tubing at the midaxillary line. The amplifier display system is then adjusted to 'zero' to compensate for amplifier offsets or display imperfections.

Two electronic zeroing methods might be used depending on the type of amplifier-monitor system used. The first is a simple push button method in which any offset in the plumbing system, transducer or ampli-
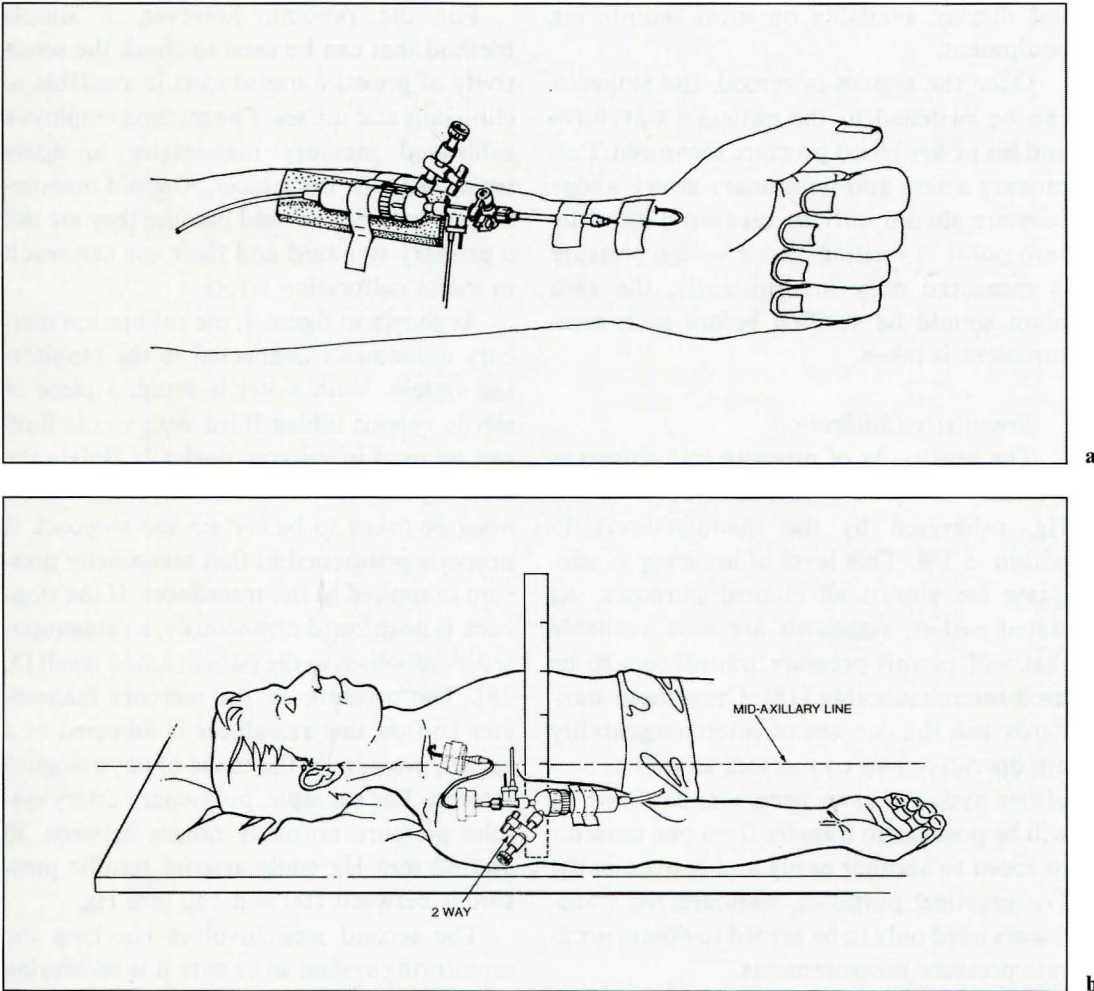
Fig. 3. Transducer mounted on the patient to minimize length of interconnecting tubing needed and thus optimize dynamic response characteristics and also minimize 'zero' shift problems due to changing hydrostatic pressure. (Change of position of transducer relative to the midaxillary line.) **a** Arterial pressure transducer mounted near the radial insertion site. **b** Transducer mounted on the upper arm for attachment to a pulmonary artery catheter (Swan-Ganz). Note: Size of transducer and plumbing components were enlarged for illustration purposes.

fier is compensated for electronically. The second method entails a simple potentiometer screw driver adjustment to set the zero point. If multiple display devices are used (e.g., oscilloscope, strip recorder, tape recorder, etc.), special care is required to be sure all elements are properly zeroed. This often requires that each device be adjusted to zero separately. In many cases, the zero point can be validated by reviewing the dig-

ital display available on most monitoring equipment.

Once the system is zeroed, the stopcock can be switched to the patient's waveform and his or her blood pressure measured. Pulmonary artery and pulmonary artery wedge pressure should only be measured after the zero point is verified. Since wedge pressure is measured only intermittently, the zero point should be verified before each measurement is taken.

Sensitivity Calibration

The sensitivity of pressure transducers is typically 5.0 μV/V excitation applied/mm Hg, calibrated by the manufacturers to within ±1%. This level of accuracy is adequate for almost all clinical purposes. As stated earlier, standards are now available that will permit pressure transducers to be used interchangeably [18]. Once these standards and the concept of interchangeability are operative and connectors as well as amplifier systems have been standardized, it will be possible to transfer from one monitor or room to another easily and conveniently. For practical purposes, standardized transducers need only to be zeroed to obtain accurate pressure measurements.

Sensitivity adjustments are built into some commercially available amplifier systems to compensate for imperfections in the transducer and amplifier systems. These adjustments will no longer be needed when standardized transducers become available. Meanwhile, the best possible pressure monitoring system is one incorporating a fixed-sensitivity transducer. Once these transducers are manufactured as disposable items and high-quality stable monitor amplifiers are developed, the pressure monitoring process will be greatly simplified.

For the present, however, a simple method that can be used to check the sensitivity of pressure transducers is available to clinicians and nurses. The method employs a calibrated mercury manometer to apply pressure to the transducer. Aneroid manometers should not be used because they are not a primary standard and their use can result in major calibration errors.

As shown in figure 4, the calibration mercury manometer connected to the monitoring system. With a sterile setup, a piece of sterile venous tubing filled with sterile fluid can be used to microbiologically isolate the mercury manometer. However, *extreme care must be taken* to be certain the stopcock is properly positioned so that manometer pressure is applied to the transducer. If the stopcock is positioned erroneously, a catastrophic air embolism to the patient could result [5, 28]. The pressure on the mercury manometer and on the transducer is adjusted to a known pressure in the range of physiological interest. For example, pulmonary artery systolic pressure normally ranges between 30 and 60 mm Hg while arterial systolic pressure is between 100 and 150 mm Hg.

The second step involves checking the monitoring system to be sure it is measuring pressure accurately. If the system is not calibrated within ±2 mm Hg, it should then be rezeroed and the calibration rechecked. If it is still unacceptable, the sensitivity of the amplifier system and that of the recording or display devices (e.g., oscilloscope, strip recorder, tape recorder, or digital display) connected to the pressure system should be checked and adjusted.

Unfortunately, most pressure calibration tests incorporated into commercially available monitors are only internal checks. They do not consider variations in transducer sen-
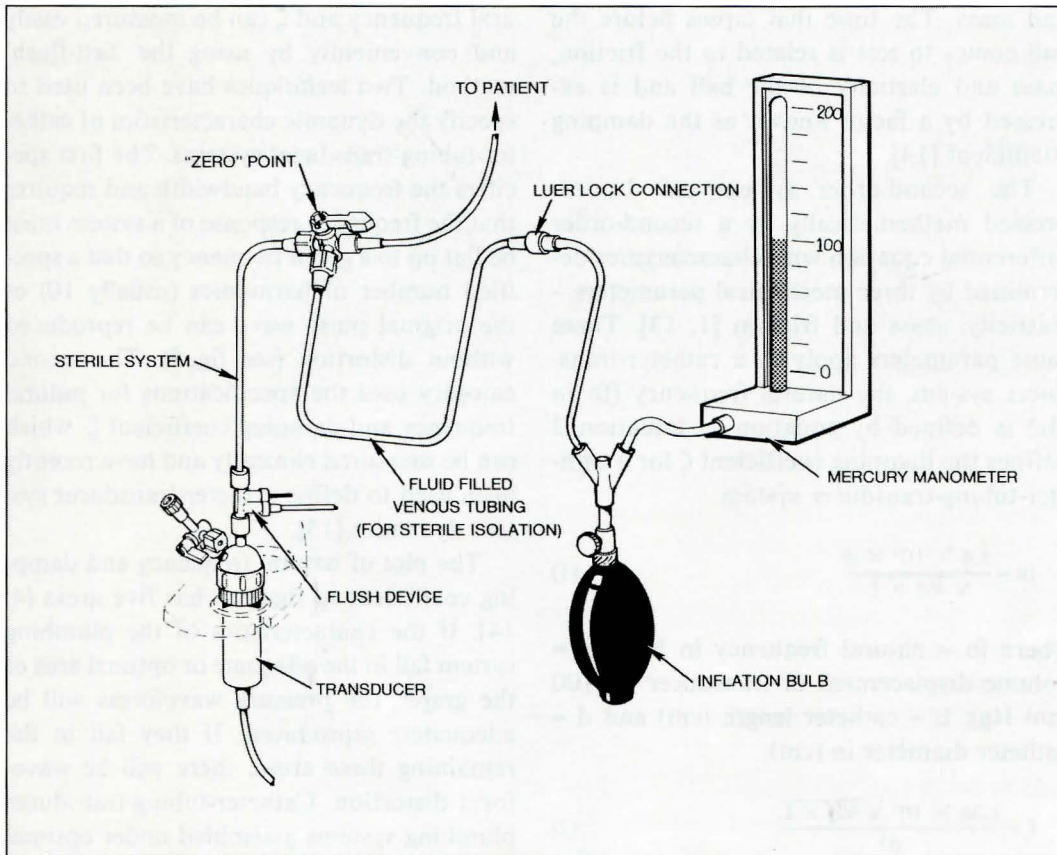
Fig. 4. Illustration of a mercury manometer sensitivity and calibration checking method which can be used in the clinical setting. The sterile fluid-filled venous tubing provides isolation from the 'contaminated' tubing of the mercury manometer. Care should be taken to assure that no air gets into the catheter entering the patient. A mercury manometer should be used to assure accuracy and the pressure in the manometer should be elevated to a pressure near the expected pressure of the patient (arterial 100–150 mm Hg, pulmonary artery 20–50 mm Hg).

sitivity. Therefore, it is important to apply a known pressure to the transducer and go through the calibration procedure. When fixed-sensitivity standard transducers become available, the sensitivity-calibration step will no longer be necessary.

The catheter-tubing-transducer systems which are used in intensive care settings typically behave as underdamped second-order dynamic systems. This is analogous to a bouncing tennis ball [14]. Dropped on a hard flat surface, the ball initially bounces several times, the height of each successive bounce gradually diminishing until the ball comes to rest. Each bounce has a characteristic frequency determined by the ball's elasticity

and mass. The time that lapses before the ball comes to rest is related to the friction, mass and elasticity of the ball and is expressed by a factor known as the damping coefficient [14].

The second-order system can be expressed mathematically by a second-order differential equation with characteristics determined by three mechanical parameters – elasticity, mass and friction [1, 13]. These same parameters apply to a catheter-transducer system, the natural frequency (fn in Hz) is defined by equation 1. Equation 2 defines the damping coefficient ζ for a catheter-tubing-transducer system.

$$fn = \frac{1.4 \times 10^3 \times d}{\sqrt{Vd \times L}} \tag{1}$$

where fn = natural frequency in Hz; Vd = volume displacement of transducer (μl/100 mm Hg); L = catheter length (cm) and d = catheter diameter in (cm).

$$\zeta = \frac{1.36 \times 10^3 \sqrt{Vd \times L}}{d^3} \tag{2}$$

where ζ = damping coefficient; d = diameter (cm); L = catheter length (cm), and VD = volume displacement of the transducer in μl/100 mm Hg (for the best transducers about 0.01 μl/100 mm Hg).

If the damping coefficient ζ is less than 1.0, a system is said to be underdamped – i.e., it will oscillate (see fig. 7a, d for example). If ζ is 1.0, the system is critically damped, and, if it is greater than 1.0, the system is overdamped – i.e., it will not oscillate (see fig. 7c for example).

For an underdamped second-order system, the natural frequency and the damping coefficient define the system's dynamic characteristics. In the clinical setting the nat-ural frequency and ζ can be measured easily and conveniently by using the 'fast-flush' method. Two techniques have been used to specify the dynamic characteristics of catheter-tubing-transducer systems. The first specifies the frequency bandwidth and requires that the frequency response of a system must be flat up to a given frequency so that a specified number of harmonics (usually 10) of the original pulse wave can be reproduced without distortion (see fig. 5). The second category uses the specifications for natural frequency and damping coefficient ζ, which can be measured clinically and have recently been used to define catheter-transducer system dynamics [13].

The plot of natural frequency and damping coefficient in figure 6 has five areas [4, 14]. If the characteristics of the plumbing system fall in the adequate or optimal area of the graph, the pressure waveforms will be adequately reproduced. If they fall in the remaining three areas, there will be waveform distortion. Catheter-tubing-transducer plumbing systems assembled under optimal conditions are underdamped; a few fall into the unacceptable area [4, 14]. As will be noted from figure 6, if the system's natural frequency is less than 7.5 Hz, i.e. unacceptable, the pressure waveform will be distorted no matter what the damping coefficient is. However, if the natural frequency can be increased to 24 Hz, the value of the damping coefficient can range from about 0.15 to 1.1 and still not distort the waveform. Therefore, to optimize the dynamic response of any pressure-monitoring system, its natural frequency should be as high as possible. Equation 1 shows that this is best accomplished in a system with short tubing, catheters which have big internal diameters, and transducers and other components with
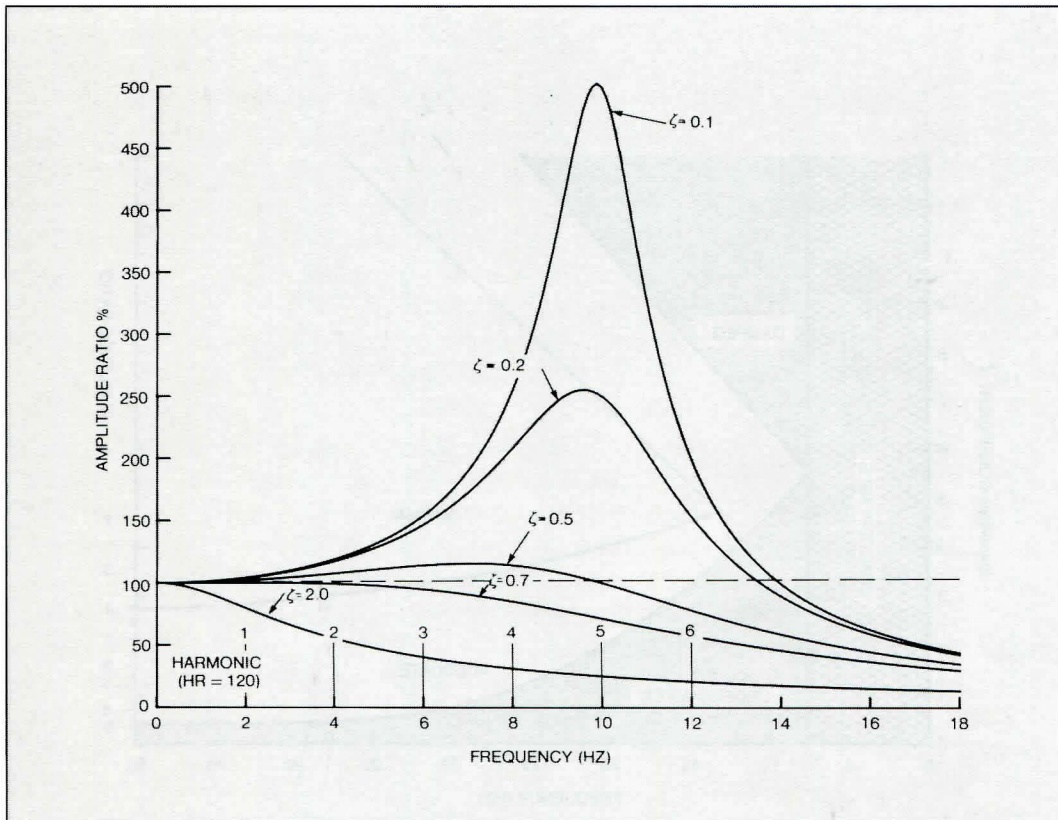
Fig. 5. Family of frequency versus amplitude ratio plots for 5 different damping coefficients ζ. Natural frequency of the plot shown is 10 Hz. A damping coefficient of 0.1 occurs when a system is very underdamped. A damping coefficient of 2.0 is for a system which is overdamped. The dashed line shows the frequency versus amplitude characteristics which would occur if the system had a 'flat' frequency response. Along the frequency axis are plotted the harmonics of the pressure wave if the heart rate (HR) were 120 beats/min (2/s). Note that by the 5th harmonic (10 Hz), if the damping coefficient were 0.1, the true signal would be amplified 5 times. If the damping coefficient were 2.0, there would be an attenuation to about one fourth of the amplitude. In both cases there would be gross waveform distortion because neither situation reflects a 'high fidelity' system's dynamic response. Fidelity of the system can be improved by (1) increasing the natural frequency or (2) adjusting the damping coefficient to be in the range of 0.5–0.7. [From ref. 4, with permission.]

small volume displacements. The damped area on the graph in figure 6 is more commonly denoted overdamped. Waveforms recorded with a system whose response is in the damped area will result in the underestimation of systolic and overestimation of diastolic pressure. The pulse pressure is flattened and the fast features in the waveform such as the dicrotic notch are lost (see fig. 7c, 8d).
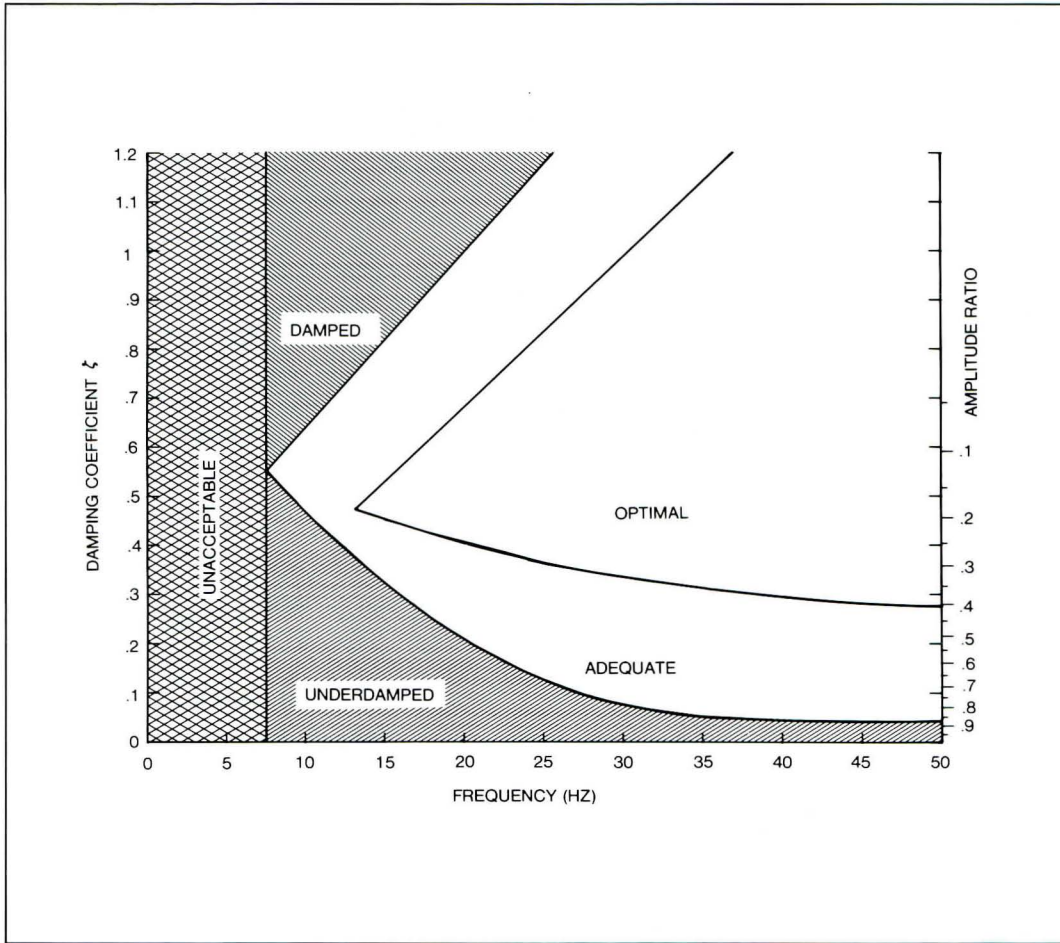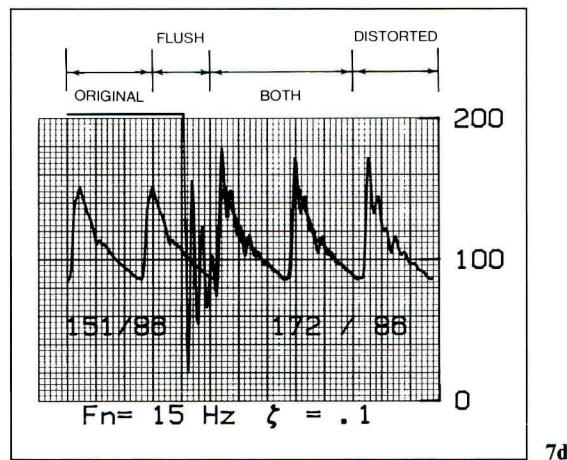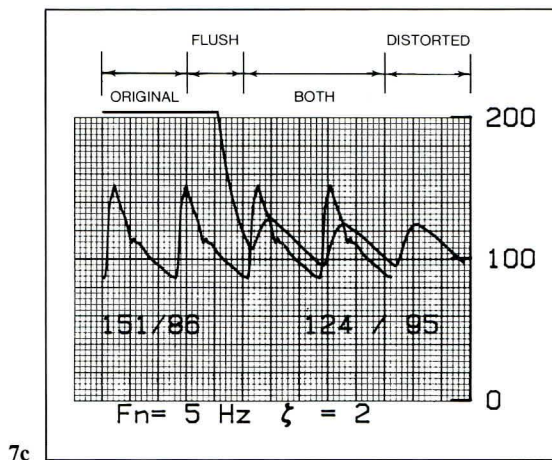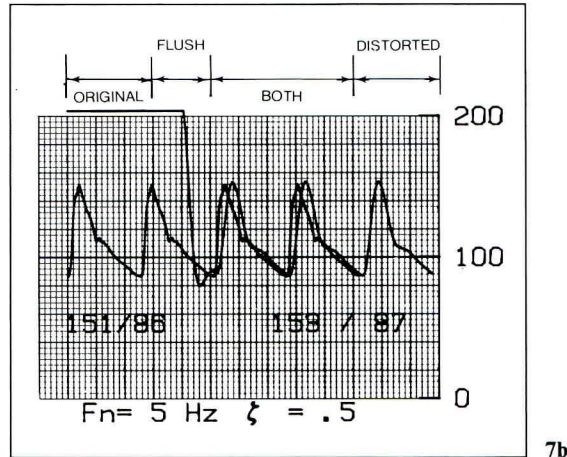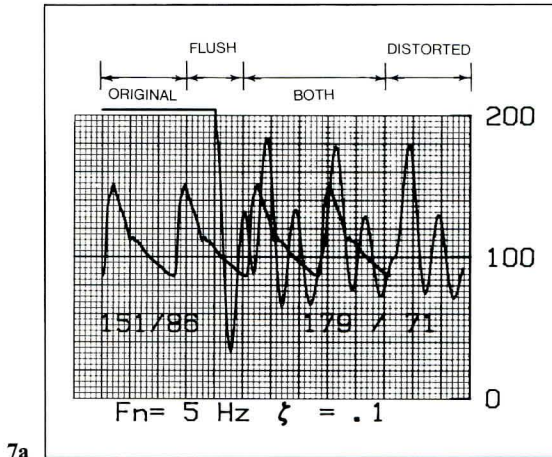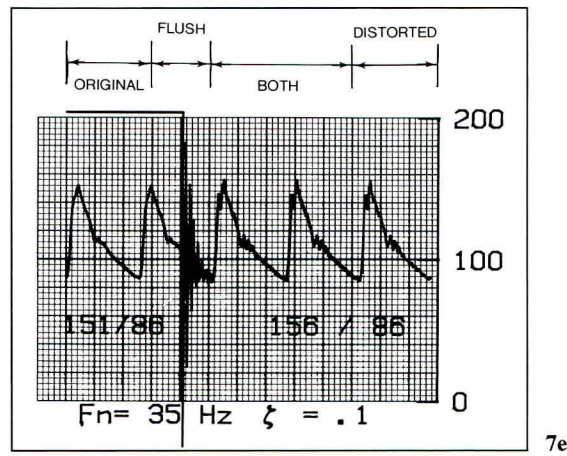
Fig. 6. Natural frequency versus damping coefficient ζ plot which illustrates the five areas into which catheter-tubing-transducer systems fall. Systems which are in the optimal area will reproduce even the most demanding (fast heart rate and rapid systolic upstroke) arterial or pulmonary artery waveforms without distortion. Systems which are in the adequate area will reproduce most 'typical' patient waveforms with little or no distortion. All other areas will cause serious and clinically important waveform distortion. Note scale on the right is used to estimate the damping coefficient from the amplitude ratio determined during fast flushing (see fig. 9 for example). [From ref. 4, with permission.]

Fig. 7. Illustration of patient waveforms and how they are distorted by different catheter-tubing-transducer plumbing systems (paper speed 25 mm/s). The initial waveform (before the flush) is the original patient waveform. Then the next two pulses show the original patient waveform and a fast-flush with a specific catheter-transducer system response superimposed (distorted). The dynamic characteristics, natural frequency (fn) and damping coefficient ζ are noted. The final pulse shows only the distorted waveform. The systolic/diastolic pressures in millimeters Hg are shown (original waveform on the left and distorted waveform on the right). The same patient's arterial waveform is demonstrated passing through five different plumbing systems. **a** An underdamped (ζ = 0.10) low natural frequency (fn = 5 Hz) system which falls in the unacceptable area of figure 6. Note the 28 mm Hg overestimation of systolic pressure and the 15 mm Hg underestimation of diastolic pressure.

7a — FLUSH / DISTORTED / ORIGINAL / BOTH — 200 / 100 / 0 — 151/86   179 / 71 — Fn= 5 Hz ζ = .1

7b — FLUSH / DISTORTED / ORIGINAL / BOTH — 200 / 100 / 0 — 151/86   159 / 87 — Fn= 5 Hz ζ = .5

7c — FLUSH / DISTORTED / ORIGINAL / BOTH — 200 / 100 / 0 — 151/86   124 / 95 — Fn= 5 Hz ζ = 2

7d — FLUSH / DISTORTED / ORIGINAL / BOTH — 200 / 100 / 0 — 151/86   172 / 86 — Fn= 15 Hz ζ = .1

7e — FLUSH / DISTORTED / ORIGINAL / BOTH — 200 / 100 / 0 — 151/86   156 / 86 — Fn= 35 Hz ζ = .1

b A near optimal damping (ζ = 0.5) but low natural frequency (fn = 5 Hz). Still unacceptable; although the systolic pressure is only slightly elevated, the shape of the waveform is dramatically altered. **c** An over-damped system (ζ = 2.0 with fn = 5 Hz) with unacceptable characteristics. Note the distortion of the waveform and the errors in systolic (–27 mm Hg) and diastolic (+9 mm Hg) pressures. **d** An underdamped (ζ = 0.10) system (same damping as waveform **a**) but with increased natural frequency (fn = 15 Hz). The quality of the waveform is improved over that of **a**, but it is still distorted. **e** An underdamped (ζ = 0.10) system with identical damping coefficient as **a** and **d**, but with further increase in natural frequency (fn = 35Hz). Note now that there is very little waveform distortion.
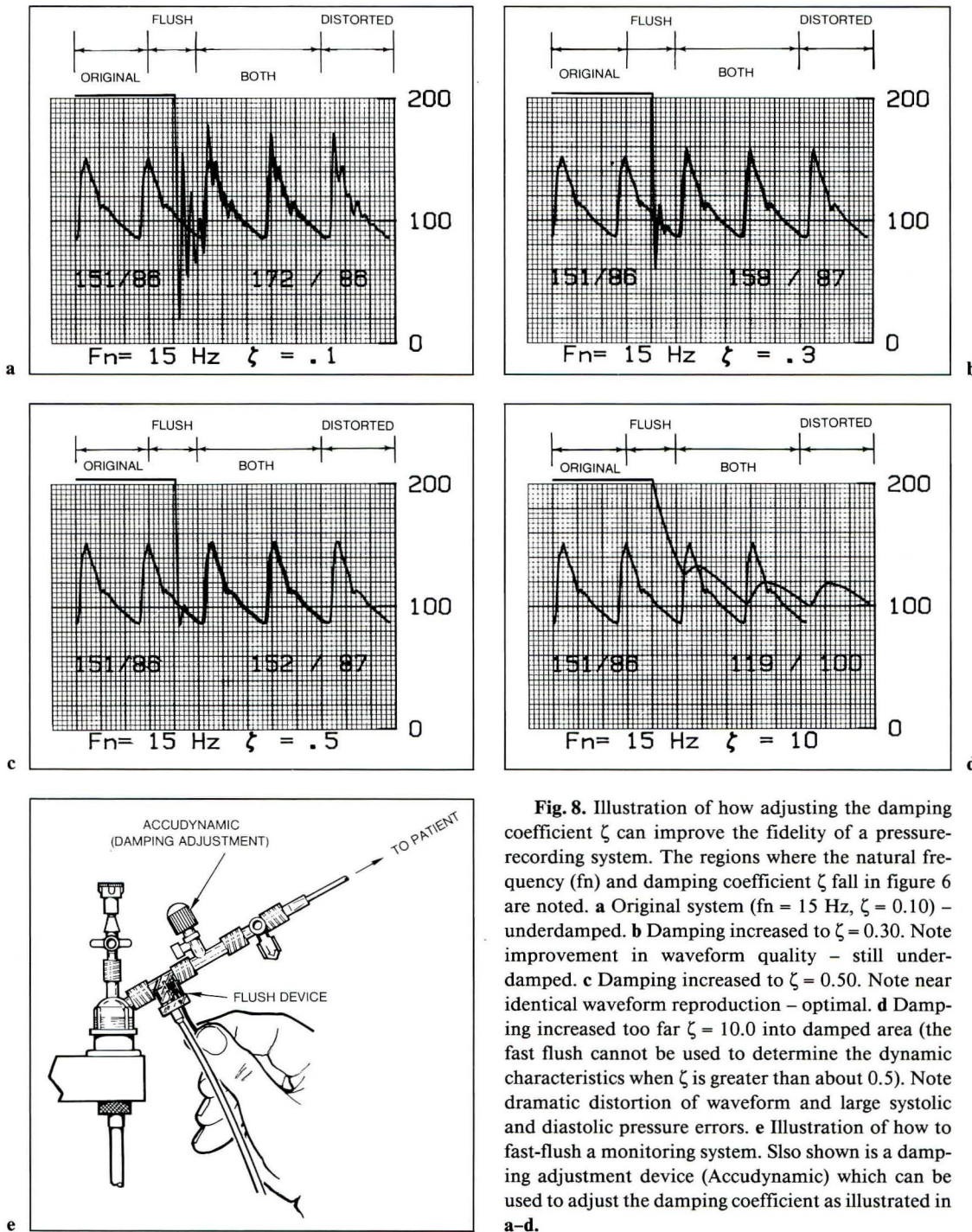
Fig. 8. Illustration of how adjusting the damping coefficient ζ can improve the fidelity of a pressure-recording system. The regions where the natural frequency (fn) and damping coefficient ζ fall in figure 6 are noted. **a** Original system (fn = 15 Hz, ζ = 0.10) – underdamped. **b** Damping increased to ζ = 0.30. Note improvement in waveform quality – still underdamped. **c** Damping increased to ζ = 0.50. Note near identical waveform reproduction – optimal. **d** Damping increased too far ζ = 10.0 into damped area (the fast flush cannot be used to determine the dynamic characteristics when ζ is greater than about 0.5). Note dramatic distortion of waveform and large systolic and diastolic pressure errors. **e** Illustration of how to fast-flush a monitoring system. Slso shown is a damping adjustment device (Accudynamic) which can be used to adjust the damping coefficient as illustrated in **a–d.**
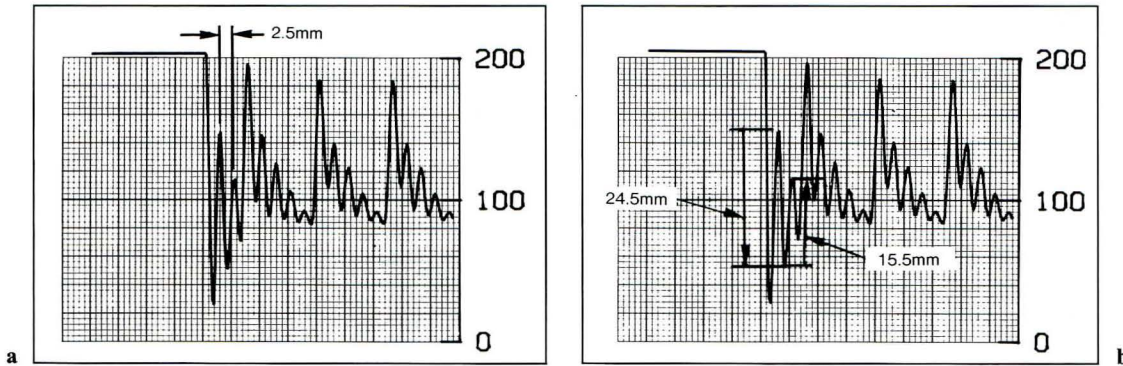
Fig. 9. Demonstration of how to determine natural frequency (fn) and damping coefficient ζ from a strip chart recording. **a** Determination of natural frequency. Paper speed 25 mm/s. The time of one cycle is 2.5 mm, therefore fn = 10 Hz. fn = paper speed (mm/s)/time (mm) = 25/2.5 = 10 Hz. **b** Determination of damping coefficient. Amplitude ratio = B/A = 15.5/24.5 = 0.6. Then from figure 6 (right hand scale) ζ = 0.14.

The importance of the interaction of natural frequency and damping coefficient are demonstrated in figure 7a, d, e, where natural frequency is varied from 5–35 Hz but the damping coefficient remains fixed at 0.1. Figure 7a–c shows the effects of a changing damping coefficient while holding the natural frequency fixed. From figures 7 and 8, it is evident that a system with a high natural frequency permits a broader latitude of damping coefficient than do systems with lower natural frequencies. Consequently, every effort must be made to maximize natural frequency if systolic and diastolic pressures are to be measured accurately.

Equations 1 and 2 can be used to estimate natural frequency and damping coefficient. However, the dynamic response characteristics of identical catheter-tubing-transducer plumbing system setups may be different because setup procedures vary. The primary reason for these setup differences is air bubbles trapped somewhere in the system. The

bubbles can become trapped when the system is being filled with fluid or they may form when air from the continuous flush fluid comes out of solution [29]. Therefore, in the clinical setting, it is mandatory to test the adequacy of each pressure monitoring system's dynamic response. This can be done easily using the fast-flush technique.

The fast flush is produced by opening the valve of the continuous flush device (for example, by pulling and quickly releasing the pigtail on the Intraflo). The rapid closure generates a square wave from which the natural frequency and damping coefficient of the plumbing system can be measured. Fast-flush testing should be performed at regular intervals, at least once each 8 h and following manipulation of the plumbing system, such as after drawing blood samples.

Natural frequency is estimated by measuring the width of one full oscillation on a strip chart recorder (see fig. 9a), following a fast flush and determined by dividing the

paper speed by the measured width. To determine the damping coefficient, any two successive peak amplitudes are measured and an amplitude ratio obtained by dividing the measured height of the lower peak by that of the amplitude of the larger peak (see fig. 9b) to get the ratio. This ratio is then converted either graphically or numerically (equation 3) to the damping coefficient [4, 14]. The graphical solution for estimating the damping coefficient is performed using the scale on the right side of figure 6.

$$\zeta = \frac{-\ln\,(\text{ratio})}{\{\pi^2 + [\ln\,(\text{ratio})]^2\}^{0.5}} \qquad (3)$$

where ζ = damping coefficient.

In the clinical setting, 2 or 3 fast flushes should be performed so that at least one occurs during the diastolic runoff phase of the pressure waveform to prevent reading errors related to the patient's pressure pulse. Once the natural frequency and the damping coefficient ζ have been determined, these data can be plotted on the graph in figure 6 to ascertain the adequacy of dynamic response. Once physicians, nurses, or technologists become familiar with the fast-flush dynamic-response testing technique, they should be able to estimate the adequacy of dynamic-response characteristics by studying the fast-flush waveform on the bedside oscilloscope.

Some bedside monitors and recorders may compromise the fast-flush technique, especially monitors with built-in low-pass filters. These filters roll off frequencies above 8 or 12 Hz. Unfortunately, they were built in to compensate for underdamped plumbing systems but they prevent 'optimizing' the system's natural frequency. These filters should be expanded to at least 50 Hz or eliminated completely. Strip chart recorders, especially thermal paper recorders, may have difficulty responding to the oscillations of systems with natural frequencies above 20 Hz. Ink-writing recorders respond much better to higher frequencies than thermal recorders and, therefore, are recommended.

Several factors that lead to poor dynamic responses are (a) air bubbles in the system, usually caused by a poor initial plumbing system setup, (b) pressure tubing that is too long, too compliant, or the diameter of which is too small, and (c) pressure transducers that are too compliant. Although air bubbles are frequently, and sometimes intentionally, added to pressure monitoring systems to damp the waveform, they exert a detrimental effect because they not only increase the damping coefficient, they also decrease the natural frequency (equations 1, 2 and fig. 6). The best way to enhance the system's dynamics is to improve its natural frequency (equation 1).

Techniques that increase the natural frequency of the monitoring system include:

(1) Eliminate air bubbles. Use transparent tubing in the fluid pathways so that air bubbles can be seen and removed. They are not detected easily at interconnecting points or between the diaphragm dome and the metal transducer diaphragm. Therefore, plastic diaphragm domes should either not be used or properly attached.

(2) Using simple systems. Keep the system simple, using the fewest number of components possible.

(3) Eliminate compliant elements. Use only high-quality, low-compliance pressure tubing, keep its length as short as possible. Use low-compliance catheters and transducers as well as tightly sealing, low-compliance

stopcocks. Eliminate other compliant elements such as medication injection sites.

After all of the above steps have been taken, the system's dynamic response may still be underdamped (see fig. 6). This is the case for many plumbing configurations, especially those that include a pulmonary artery catheter (Swan-Ganz). Corrective devices may then be used. The Accudynamic [14] and the CorrecTorr [30] (see fig. 8e) are disposable correction devices that are easy to use and allow the damping coefficient to be adjusted without degrading the system's natural frequency, much like the increase in damping seen in figure 8a–d.

The dynamic response of the pressure monitoring system must be optimum to measure systolic and diastolic pressure accurately. However, if only mean pressure measurements are required, dynamic response characteristics are of less importance. But, since physicians are increasingly using derived parameters, all pressures must be measured accurately. If the system's dynamic response is inadequate, whether underdamped or overdamped, there will be errors in systolic pressure. Those recorded from an underdamped system will be overestimated, whereas systolic pressure from an overdamped system will be underestimated. Diastolic pressures are also affected but are much more tolerant of dynamic response inadequacies (see fig. 7 and 8).

The use of invasive pressure-monitoring systems entails iatrogenic risks to the patient and their use is justified only if accurate and reliable data are obtained. Merely looking at the waveform will not provide the information required to determine the adequacy of the system's dynamic response. However, the fast-flush technique will verify the response of the system.

Some simple rules that should help to optimize the dynamic response are: (1) maximize the monitoring system's natural frequency; (2) measure the system's natural frequency using the fast-flush test; for a paper speed of 25 mm/s one oscillation should be completed in less than 2 mm (2 mm = 12.5 Hz), and (3) damping coefficient should range between 0.2 and 0.5, which means that the ratio of successive peaks should be between 0.53 and 0.16.

Signal Amplification, Processing and Display

Once the pressure signal has been transmitted to the transducer, the processing and display modulor, bedside monitor operates on that signal. Most monitors not only display the heart rate and systolic, diastolic and mean pressure, but they also display the processed waveform on an oscilloscope, and provide an analog output for a recorder or for transmission to a central display [20]. Figure 10 provides a block diagram of a typical bedside monitor processing system and indicates how each of the parameters is derived.

Systolic pressure, which is detected by a 'peak' detector is not the same for each heart beat. Since this variability would cause the digital display to appear unstable, most manufacturers filter or smooth the derived systolic pressure. Thus the digital value shown on the display is an average of a finite number of preceding beats.

Most systems employ simple low-pass filters with a cutoff frequency of about 0.2 Hz or less (time constant = 0.80 s). Figure 11 indicates how a patient's pressure waveform
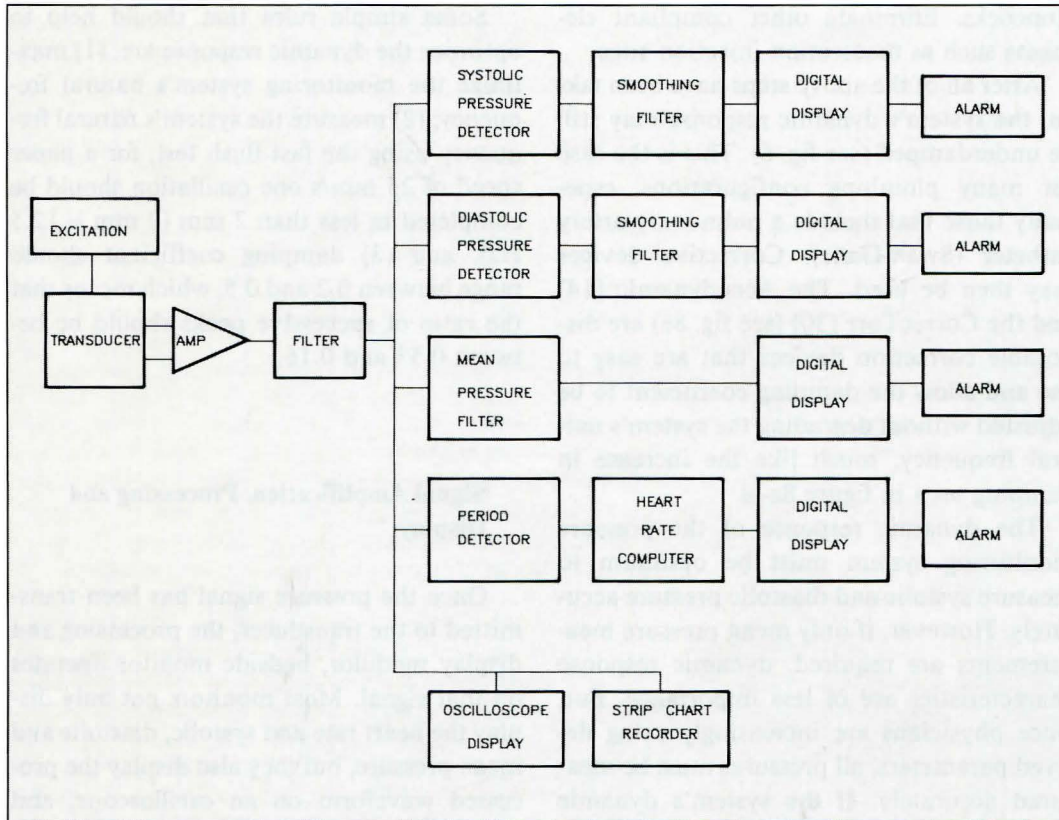
Fig. 10. Block diagram of a bedside blood pressure-monitoring system showing the different elements used. The transducer is excited by an electrical voltage. As a result of the excitation and the pressure applied to the transducer a small electrical signal proportional to the pressure is generated. The amplifier increases the amplitude of the resultant pressure signal. In most systems the pressure waveform is processed through a low-pass filter to eliminate unwanted high frequency signal components. The resulting pressure signal is then sent to 4 processing modules (systolic, diastolic, mean pressure, and heart rate). The same signal is usually made available for display on an oscilloscope and strip chart recorder. In the processing modules the various parameters are determined and presented on a digital display.

is processed to derive systolic and diastolic pressures. The original analog waveform (fig. 11a) is processed to derive beat to beat systolic and diastolic pressures (fig. 11b). Then these pressures are filtered (smoothed) to provide more stable readings (fig. 11c). Whereas the filtering has the advantage of stabilizing the digital display, it has two major disadvantages:

(1) The systolic peak detector often detects artifacts as well as real signals. Accordingly, if there are overshoots coupled into the pressure waveform due to poor dynamics of an underdamped system or because of

catheter whip, these artifacts are sensed and the systolic pressure is falsely increased.

(2) Respiratory variations in the pressure signals can be large, especially for pressures measured in the right side of the heart (Swan-Ganz catheter). Unfortunately, smoothing and filtering distorts the real signals and precludes measurement of these pressures at end expiration when transmural pressure approaches zero [22–27, 31, 32].

Mean pressure is computed by averaging the instantaneous pressure within the time interval of each heart cycle [32]. The averaging function is accomplished easily and effectively by a low pass filter which in most systems, has a cutoff frequency of about 0.1 Hz (time constant = 1.59 s). The mean pressure signal determined from low-pass filtering is generally free from overshoot, damping, and catheter whip artifacts. However, because the signal is filtered, it is difficult to remove cardiac variations and retain respiratory variations. This applies to a majority of systems which employ low pass filters that average over an entire respiratory cycle (see fig. 12).

Mean pressure is sometimes estimated using the following equation:

Mean pressure =

diastolic pressure + (systolic – diastolic)/3   (4)

This equation provides a surprisingly good estimate of arterial pressure for many patient waveforms (see fig. 13a). However, if there are dynamic response defects or if there are artifacts on the waveform, the systolic or diastolic pressure will be distorted. Consequently, the estimate of mean pressure using equation 4 will also be in error.

Mean systolic pressure (i.e., mean pressure during systole) is an increasingly important measurement because it is the pressure against which the heart pumps. Some clinicians use equation 5 to estimate mean systolic pressure.
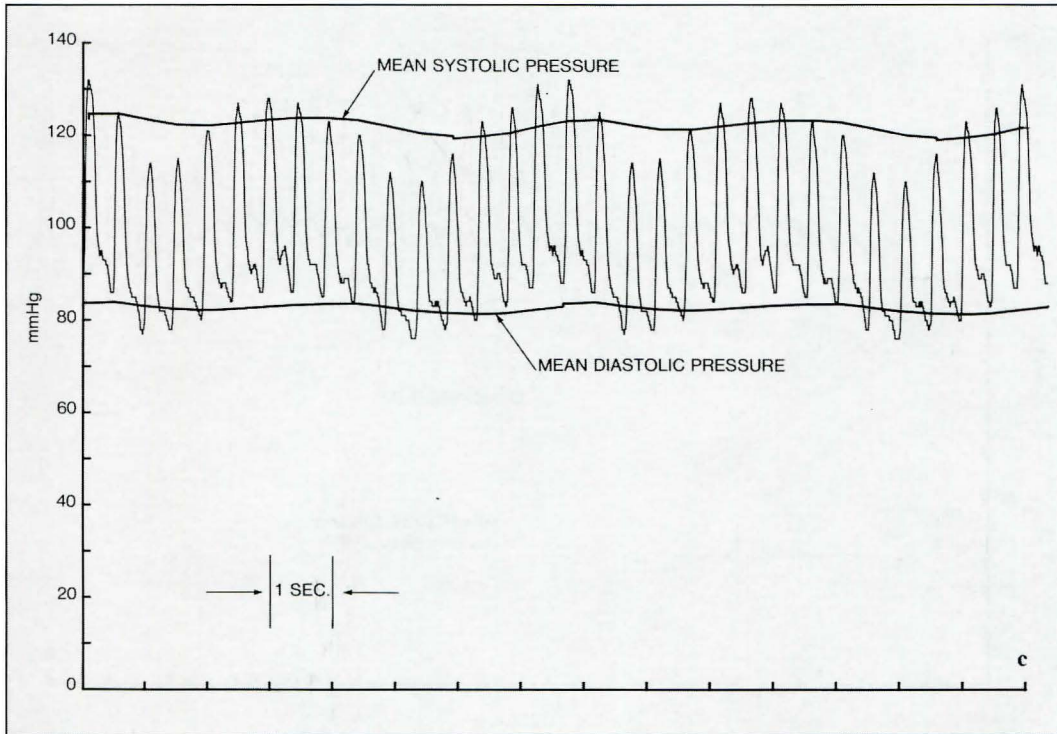
Mean systolic pressure =

diastolic pressure + 2*(systolic – diastolic)/3   (5)

For arterial pressure, this estimate is also good (see fig. 13b).

Many monitors detect heart rate from the arterial pressure waveform. By detecting the upstroke of the systolic ejection phase of the arterial pressure waveform, heart period (P) and thus heart rate (HR = 1/P) can be determined. However, heart rate is not a steady value but varies with respiration and other physiologic parameters. Therefore manufacturers of most monitors smooth out the heart rate so that the digital display is stable and easy to read. Heart rate determined from the pressure signal has two major advantages over that determined from the electrocardiogram (ECG): (1) the arterial pressure signal usually has fewer artifacts than does the ECG so the heart rate and, therefore, rate alarms are more reliable, and (2) heart rate determined from the pressure signal is indicative of the mechanical action of the heart; thus, it is a better indicator of blood flow since arrhythmias, such as premature ventricular depolarizations determined from the ECG, may be totally ineffective mechanically (see fig. 13).

Standard techniques have not been established for measuring systolic and diastolic pressures during arrhythmias. Some of the problems encountered are shown in figure 14. A monitor may not display accurate heart rate or systolic and diastolic pressures. Until standard techniques are established, the most meaningful digitally displayed value is the mean arterial pressure.
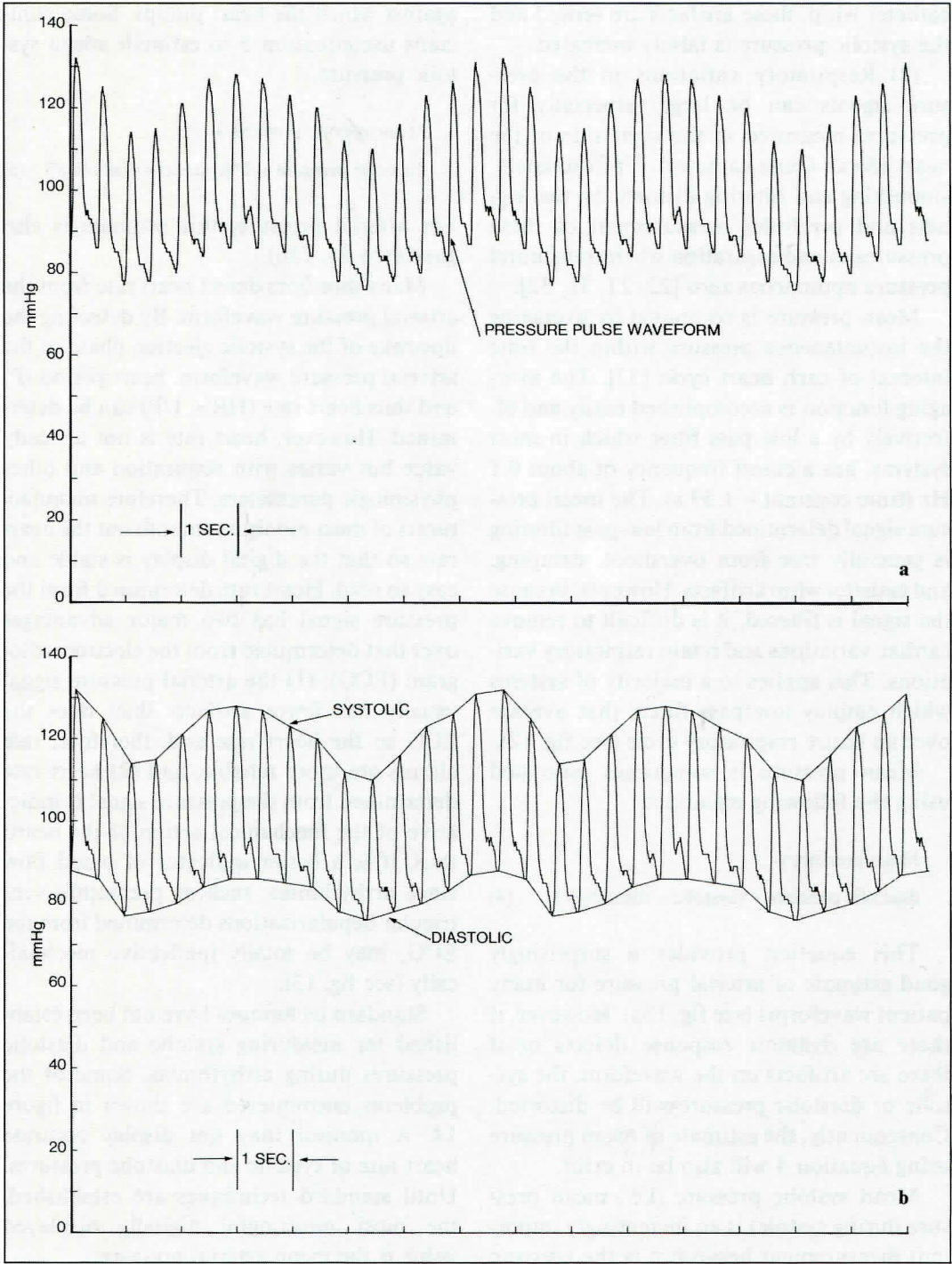
Fig. 11. The processing of systolic and diastolic pressure by the bedside monitor in a patient with pulsus paradoxus resulting from spontaneous breathing. **a** Original analog pulse pressure waveform. **b** Peak and valley detection – note how the systolic and diastolic pressure change with the systolic pressure varying from 132 to as low as 110 mm Hg. The diastolic pressure varies from 88 to 76 mm Hg. **c** Filtered systolic and diastolic pressures – note how they are smoothed out.

Accurate Measurement of Pulmonary Artery Pressure

Ever since it was introduced, the balloon-tipped flow-directed pulmonary artery catheter (Swan-Ganz) has been widely used in intensive care units. The ease with which it is usually inserted may lead one to conclude that measurement of pulmonary artery (P_A) and wedge pressure (P_W), are easily and reliably accomplished [25, 33, 34]. This conclusion is not consistent with data reported by the originators of the technique.

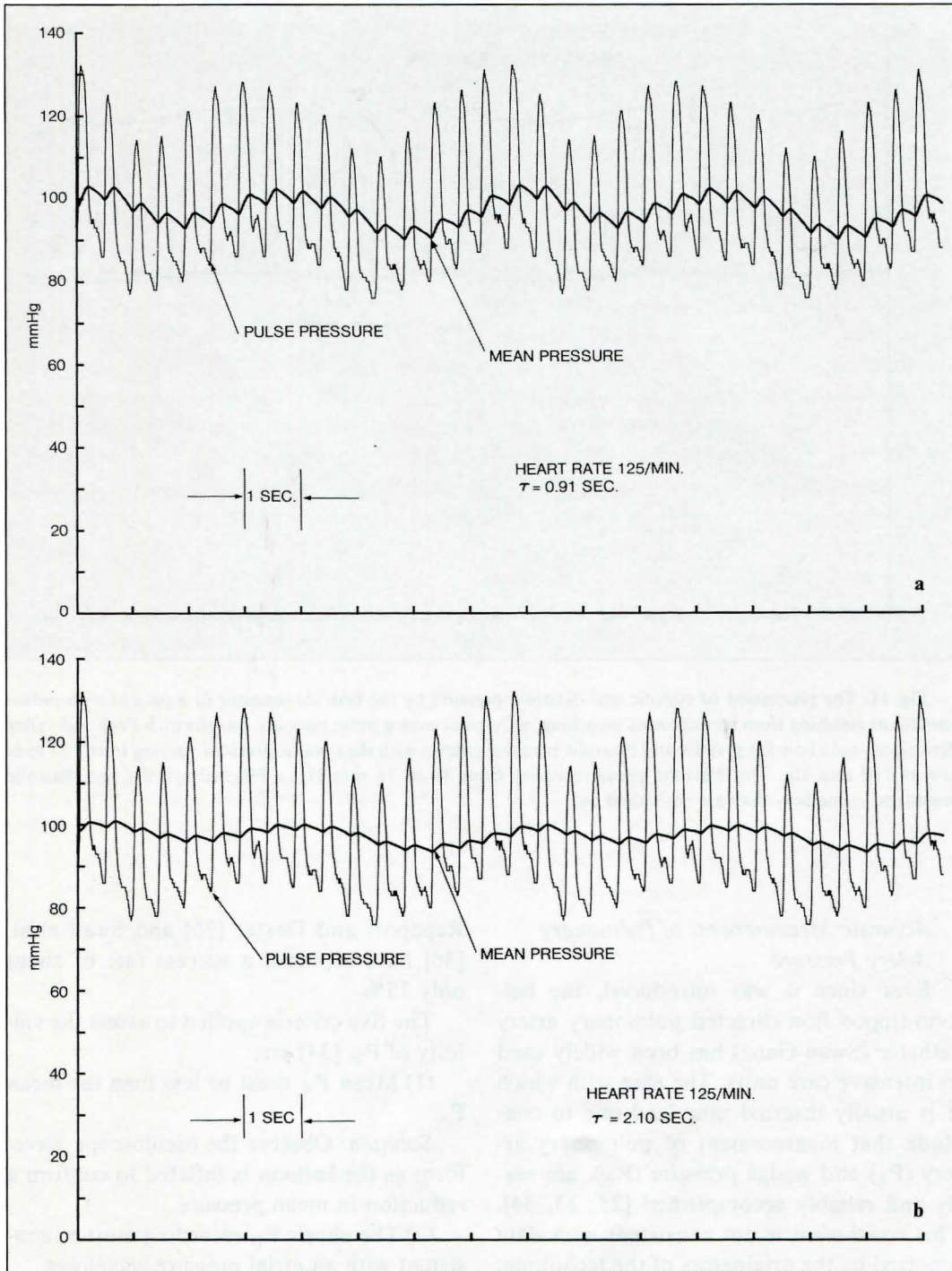Rapaport and Dexter [35] and Swan et al. [36] have reported a success rate of about only 75%.

The five criteria applied to assess the validity of P_W [34] are:

(1) Mean P_W must be less than the mean P_A.

Solution: Observe the oscilloscope waveform as the balloon is inflated to confirm a reduction in mean pressure.

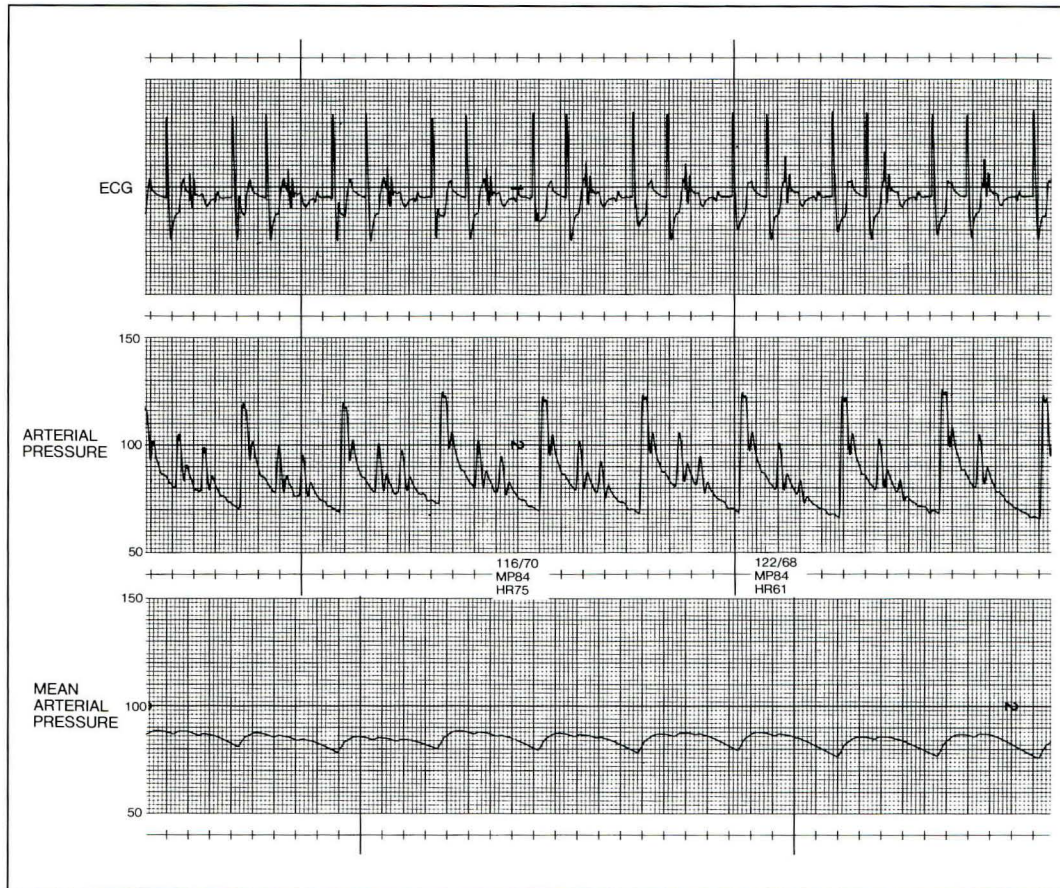(2) The phasic P_W recording must be consistent with an atrial pressure waveform.

PULSE PRESSURE

MEAN PRESSURE

HEART RATE 125/MIN.
τ = 0.91 SEC.

1 SEC.

a

PULSE PRESSURE

MEAN PRESSURE

HEART RATE 125/MIN.
τ = 2.10 SEC.

1 SEC.

b

12

Fig. 12. Illustration of the effects of filters used to determine mean pressure. Note the problems of determining the real mean pressure because of pressure variability due to the patient's breathing pattern (respiratory variation). **a** Fast filter (time constant = 0.91 s –single pole filter bandwidth 0.17 Hz). Change in mean pressure approximately 12 mm Hg. **b** Slow filter (time constant = 2.10 s –single pole filter bandwidth 0.076 Hz). Change in mean pressure approximately 7 mm Hg.
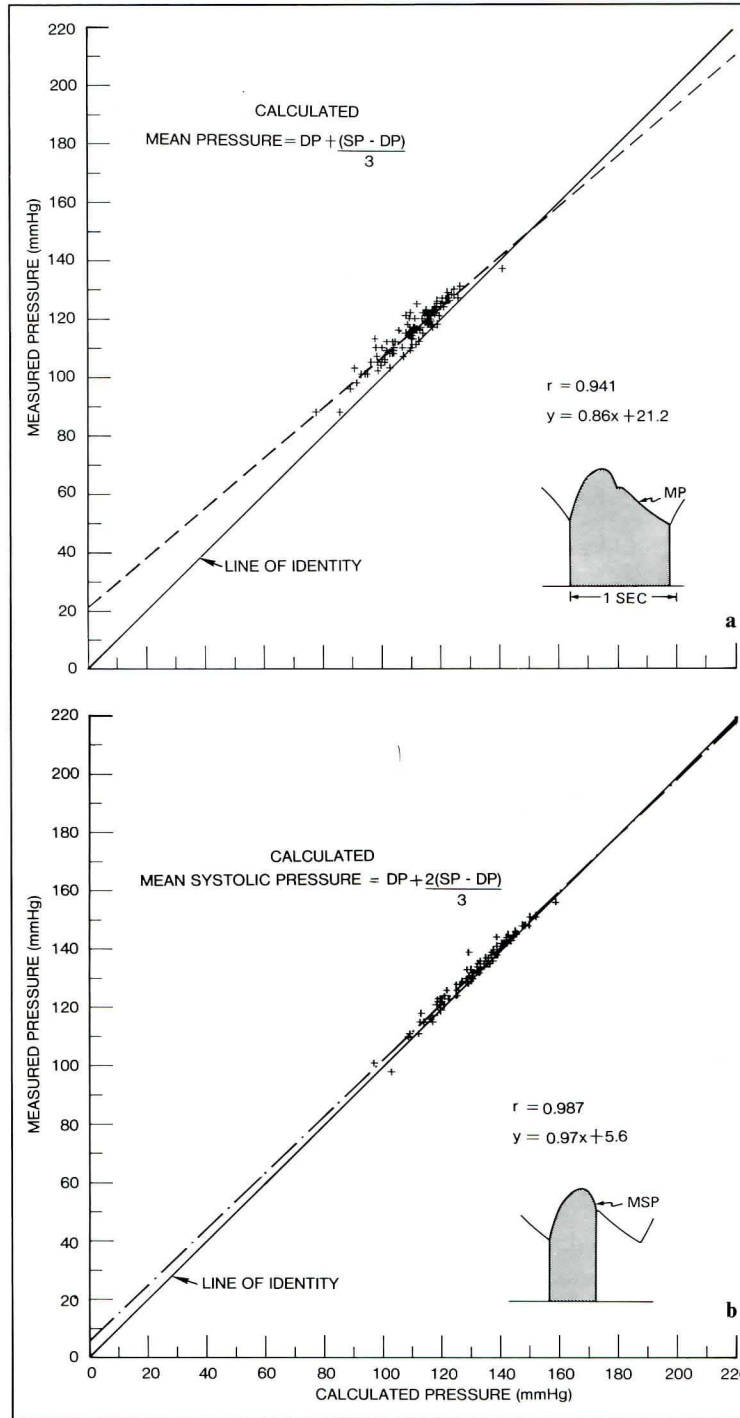
Fig. 13. Illustration of how arterial mean pressure (MP) (**a**) and arterial mean systolic pressure (MSP) (**b**) are determined and how well the usual calculations compare. Shown graphically for mean pressure and in equation form for both (upper left). The correlation is surprisingly good for both as can be seen from the correlation coefficient and the cross-plot of calculated versus measured (line through the origin is the line of identity; line which intersects at 21 mm Hg is the best-fit line for panel a).

CALCULATED

$$\text{MEAN PRESSURE} = DP + \frac{(SP - DP)}{3}$$

r = 0.941

y = 0.86x + 21.2

MEASURED PRESSURE (mmHg)

LINE OF IDENTITY

MP

1 SEC

a

CALCULATED

$$\text{MEAN SYSTOLIC PRESSURE} = DP + \frac{2(SP - DP)}{3}$$

r = 0.987

y = 0.97x + 5.6

MEASURED PRESSURE (mmHg)

LINE OF IDENTITY

MSP

CALCULATED PRESSURE (mmHg)

b

13

ECG

ARTERIAL PRESSURE

150
100
50
116/70
MP84
HR75
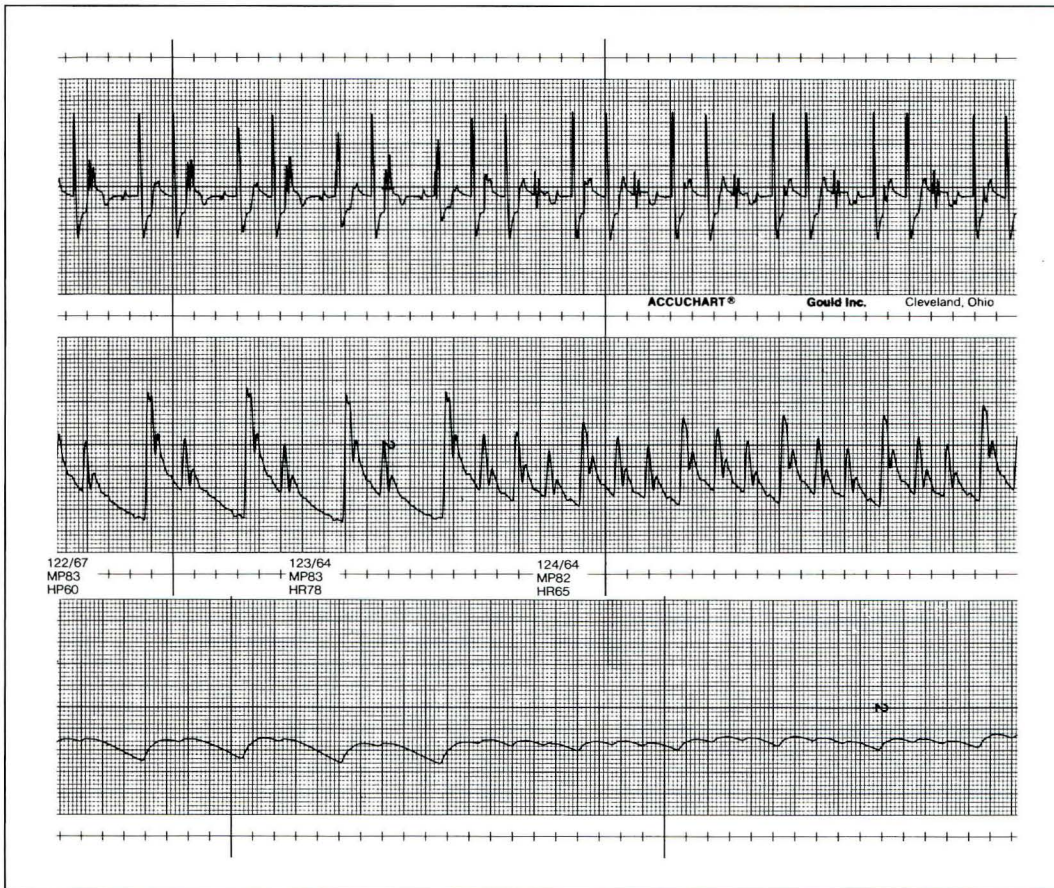122/68
MP84
HR61

MEAN ARTERIAL PRESSURE

150
100
50

14a

Solution: Observe the waveform on an oscilloscope as the balloon is inflated to make certain that the waveshape changes from its pulsatile shape of P_A pressure to the smoother 'atrial' shape. Care should be exercised to be certain that the balloon is not overtly inflated, in which case a falsely elevated P_W pressure is displayed.

(3) Free flow should be present when the catheter is in the wedge position so that the tip is in free communication with fluid column to the left atrium.

Solution: Fast flushing the catheter when it is in the wedge position, is the most simple and reliable method of demonstrating free flow [25, 33, 34]. A quick (1.0 s or less) fast flush with the continuous flush device should elicit an 'adquate' dynamic response. If it does not, there is a high probability that the P_W measurement will be inaccurate [33, 34]. Although some physicians may be concerned lest fast flush in the wedge causes injury, a 1.0-second flush delivers only about 1.0 ml of fluid. This tech-

ACCUCHART®   Gould Inc.   Cleveland, Ohio

122/67   123/64   124/64
MP83   MP83   MP82
HP60   HR78   HR65

14b

Fig. 14. Illustration of the effect of arrhythmia on the measurement of arterial pressure. **a, b** Continuous tracings of a patient with an arrhythmia are shown. The three strips show the ECG at the top, the arterial pressure waveform in the middle and the mean arterial pressure at the bottom (time constant 0.91 s – see fig. 12 for explanation). Also noted at approxi- mately 6-second intervals are the systolic/diastolic, mean pressure and heart rate (from ECG tracing) determined by a bedside monitor. Note the monitor has difficulty detecting heart rate and measuring the systolic and diastolic pressures. The most meaningful measure from all the 'digital' values is the mean arterial pressure.

nique has been used routinely in our hospital for several years without adverse consequence.

(4) A palpable 'give' or jerk should occur as the catheter is pulled back from the wedge position.

Solution: This test is inappropriate with the balloon-tipped pulmonary artery catheter since it should never be pushed into a 'wedge' position.

(5) Highly oxygenated blood must be aspirated from the wedge position. Rapaport
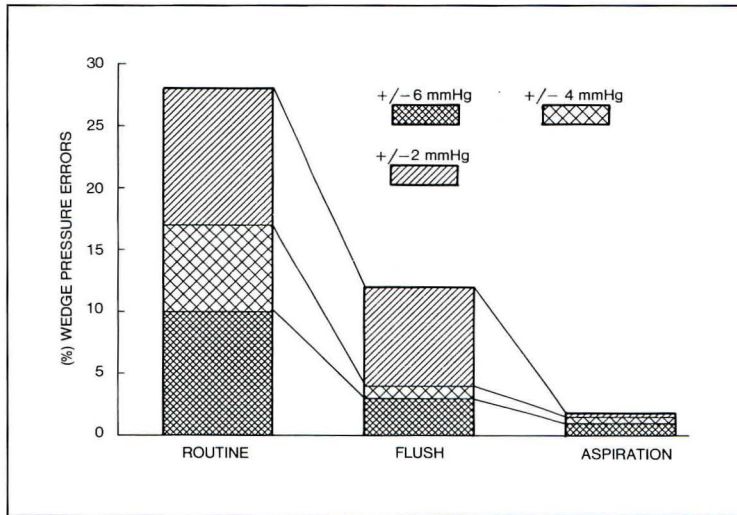
Fig. 15. P_w errors and how they can be improved by application of waveform analysis and dynamic response testing. In the usual clinical setting (routine) there are 28% of P_w measurements which will be in error by $\pm$ 2 mm Hg, 17% with errors of $\pm$ 4 mm Hg, and 10% with errors of $\pm$ 6 mm Hg. By identifiying and correcting technical problems using waveform analysis and dynamic response testing (flush) (fast-flush test) the error rate can be dramati-cally reduced, see second bar. For example the number of errors of $\pm$ 4 mm Hg (considered to be clinically important errors) can be reduced from 17 to 4%. Further verification of a P_w is made by aspiration of capillary blood, which leads to further reduction in error (see third bar). The additive confirmation given by aspiration of wedge blood is not warranted with each measurement of P_w in the clinical care setting.

and Dexter [35] considered this to be the most important criterion.

Solution: Although this procedure is possible in a clinical situation, it is not recommended for routine use [25, 33, 34]. When 'wedge' samples are withdrawn with the balloon-tipped catheter, a sufficient volume must be withdrawn to clear both the catheter and pulmonary-artery dead space. The sample should therefore be withdrawn at a slow and steady rate.

The frequency with which errors occur in P_W measurements and how they can be reduced by applying these criteria are summarized in figure 15. Usually, an error of $\pm$ 2 mm Hg is found in 28% of the P_W measurements whereas an error of $\pm$ 4 mm Hg is found 17% of the time. By identifying the technical problems by the fast-flush technique, the rate of error can be reduced dramatically [33, 34].

In summary, pulmonary artery parameters can be measured accurately if the following steps are adhered to:

(1) Zero the monitor accurately

(2) Make strip chart recordings of all P_A pressures for a time period covering at least 3 respiratory cycles. Do not use digital displays

(3) Perform and record dynamic response testing (fast-flush) for each position (i.e., wedge and P_A). If

the response is not adequate, resolve the adequacy of the plumbing system before proceeding

(4) Obtain phasic (i.e., systolic and diastolic) as well as the mean pressures from the oscilloscope or strip chart recording at end expiration, when the transmural pressure is nearest zero

(5) Measure the pressures accurately and record them to document the patient's status and for future clinical decision-making

Cardiac Output Determination

Cardiac output is commonly measured in the intensive care to assist in the management of critically ill patients [37–40]. This parameter can be measured conveniently and rapidly by the thermodilution technique. Made practical by Swan et al. [36] and Ganz et al. [37] through their pulmonary artery catheter, this technique merely requires the injection of cold solution as an indicator. Neither the injection of a foreign substance such as green dye, nor the insertion of an arterial catheter is required.

Cardiac output can also be measured by the indocyanine green dye dilution procedure by applying the Stewart-Hamilton technique. Using the dye dilution technique, it is possible to determine cardiac output by dividing the area under the dilution curve by the quantity of indicator which was injected (equation 5). Principles applicable to the indocyanine green dye dilution technique are the same as those of the 'cold' solution.

$$CO = \frac{quantity}{area} =$$

$$\frac{VI \times (TB - TI) \times SI \times CI \times CT \times 60}{\int BT(t)dt \times SB \times CB} \quad (6)$$

where CO = Cardiac output; VI = volume of injectate (l), TB = initial blood temperature

($^\circ$C); TI = injectate temperature ($^\circ$C); CI = specific gravity – injectate; CB = specific gravity – blood; SI = specific heat – injectate; SB = specific heat – blood

$$\text{for 5\% dextrose } \frac{SI \times CI}{SB \times CB} = 1.08;$$

CT = correction factor for loss of thermal signal in the cathether and $\int BT(t)dt$ = the area under the time versus temperature thermodilution curve.

The thermodilution method for determining cardiac output poses several problems in that many of the basic assumptions required by the Stewart-Hamilton technique are not satisfied. Firstly, the exact amount of thermal indicator injected cannot be quantitated precisely. Secondly, indicator is lost at various stages and this loss of indicator (heat loss) leads to errors.

A block diagram of the thermodilution measuring system with typical thermodilution curves and time of injection indicated (fig. 16) shows the transit time for the cooled blood moving from the injection site in the right atrium, to the pulmonary artery measurement site. The cold (i.e., room temperature or iced physiological solution) is thoroughly mixed with blood in the right ventricle. To calculate the area under the curve, a baseline temperature must be established before the injection. In turn, the end point is usually determined by extrapolating to this baseline temperature. Various algorithms are used to determine the endpoint and the area under the curve. The two methods in common use either fit the downslope of the curve with an exponential curve to the baseline or truncate the curve and add an estimated area after the truncation point. Both methods can lead to errors.
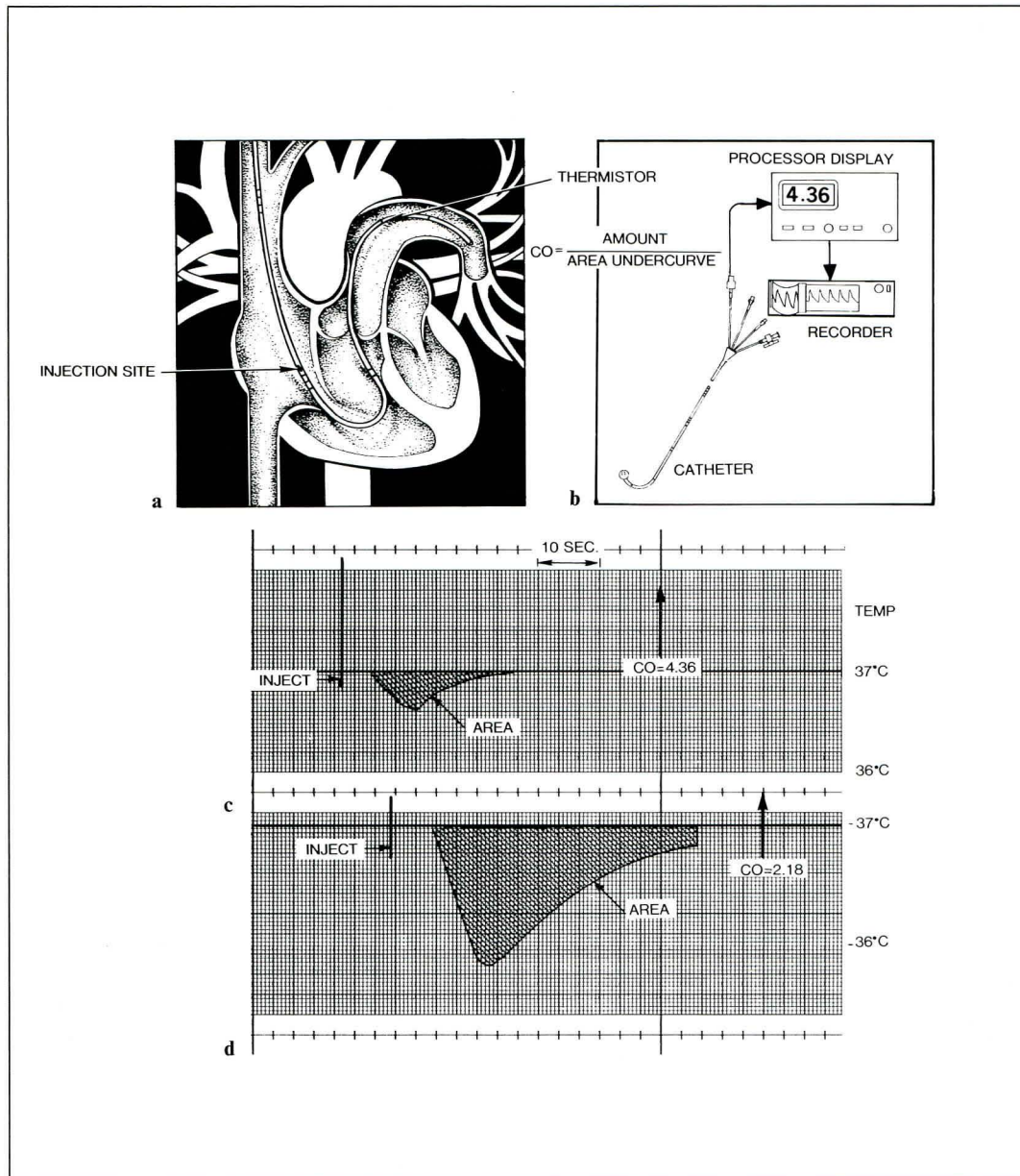
Fig. 16. Schematic diagram of the thermodilution measurement of cardiac output. Note that a recorder of some type should always be used to verify the quality of the thermodilution curve. **a** Thermodilution catheter placement in the pulmonary artery. Note location of injection site and thermistor. **b** Configuration of thermodilution catheter connected to a cardiac output processor and recorder. **c** Typical temperature-time plot sensed by the thermistor near the catheter tip. Cardiac output (CO) is 4.36 liter/min. **d** Temperature-time plot for low cardiac output (2.18 liter/min). Note the larger area and broader dispersion because of the lower flow.

To ensure that the results of the thermodilution cardiac output measurement are trustworthy, it is recommended that the thermodilution curves be visualized either on the monitor screen or on a strip recorder. Recent studies have shown that synchronizing the injections with the respiratory cycle improves reproducibility of the technique [40]. Measurement and physiological variations in the thermodilution determinations require that at lease three reproducible curves are obtained. Averaging the three curves gives a more representative assessment of cardiac output.

Until recently, the injection of ice-cold infusate has been recommended because it gives the largest signal and, as a consequence, the largest signal to noise ratio. A variety of techniques have been used to ensure that saline injections are ice-cold and to minimize cross-contamination between injections. Input temperature may be measured with a second thermistor at the injection port of the Swan-Ganz catheter and powered injection syringe has also been proposed [41]. However, room temperature injectate has shown better reproducibility. This simpler and less expensive injection configuration is therefore preferred [42].

A varying amount of thermal indicator is retained in the catheter after each injection because of the 0.75 ml dead space of the catheter and because of the variable length of catheter which is exposed to room temperature while the remaining length is at body temperature. For this reason, the first cardiac output measurement should be discarded and at least three good cardiac output measurements within ± 5 to $\pm 10\%$ should be obtained. The injection of 10-ml samples is recommended for adults. The following general procedures are recommended:

(1) Be sure the catheter tip is correctly placed, i.e., there should be a good P_A pressure waveform, as well as a good dynamic response as determined by the fast-flush technique

(2) Inject 10-ml volumes into the patient either at peak-inspiration or at end-expiration

(3) Review the adequacy of the curve shape, i.e., no double humps, smooth downslope

(4) After discarding the first injection, repeat the procedure 3 more times at approximately 1-min intervals, at either peak-inspiration or at end-exhalation

(5) Record and review curves for the injection and obtain results within $\pm 10\%$ of each other. if these results are not obtained, the problems should be resolved and the series repeated

(6) Validate cardiac output results with other physiological measures such as arteriovenous differences in oxygen content

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