MONITORING, AMBULATORY. See AMBULATORY MONITORING.

MONITORING, FETAL. See FETAL MONITORING.

MONITORING, HEMODYNAMIC

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INTRODUCTION

The word monitor has a variety of meanings, depending on the context. A monitor can be any device for checking on or regulating the performance of a machine, aircraft, or a patient. A patient monitor is usually thought of as something that watches, warns, or cautions if there is a life-threatening event. A more rigorous definition of patient monitoring is Repeated or continuous observations or

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measurements of the patient, his or her physiological status, and the functions of life support equipment for the purpose of guiding management decisions, including when to make therapeutic interventions and assessment of those interventions (1). As a result, a monitor should not only alert physicians and nurses to potentially life-threatening events, but perhaps should also control devices that maintain life. The primary emphasis of this section deals with hemodynamic monitoring of the critically ill patient who is in an intensive care unit (ICU), but many of the principles apply to all hospitalized patients.

Hemodynamic monitoring relates to monitoring of the blood pressure and blood flow in the cardiovascular system. The cardiovascular system consists of the heart, lungs, and blood vessels, and has a most important function in maintaining life in complex animals, such as humans. Oxygen and fuel must be transported from their source to the individual cells that consume them. The resulting waste products of metabolism must then be disposed of. Thus, the heart and blood vessels transport nutrients to the body and remove the waste products. Clearly, if this system does not function properly, the organism could be compromised. As a consequence clinically applicable methods have been developed to assess the function of the cardiovascular system. Hemodynamic monitoring is one part of this complex monitoring strategy. Typical parameters measured when performing hemodynamic monitoring are heart rate and rhythm, measured through analysis of the electrocardiogram (ECG), blood pressure measurements in various locations in the cardiovascular system, and estimates of blood flow usually using cardiac output as a measure.

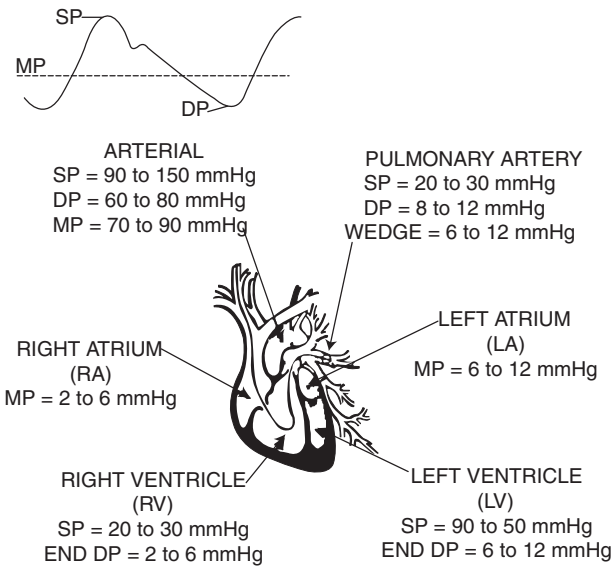

Figure 1. Outline drawing of the heart with its chambers and typical pressures (expressed in mmHg) for each area of the heart. Note the pressures are systolic (SP), diastolic (DP), and mean (MP), as shown on the arterial pressure waveform.

THEORY

Hemodynamic monitoring permits minute-to-minute surveillance of the cardiovascular system and provides physiologic data to assist in diagnosis as well as to guide therapy (2–5). The cardiovascular system consists of the heart, lungs, and blood vessels that supply blood to the body and return blood from the peripheral tissue.

It is beyond the scope of this section to describe the detailed anatomy of the cardiovascular system. However, to understand the principles of hemodynamic monitoring knowledge of the functional aspects of the cardiovascular system is essential.

HEART

The heart is made up of four chambers: the right atrium and the right ventricle and the left atrium and the left ventricle (see Fig. 1). The right atrium accepts blood from the systemic circulation (head, arms, and legs) via the superior and inferior vena cava. On atrial contraction the tricuspid valve between the right atrium and right ventricle opens and blood flows into the right ventricle. On ventricular contraction the right ventricle pumps blood through the pulmonic valve into the pulmonary artery and to the lungs where oxygen is added and carbon dioxide is removed. Blood flows from the lungs to the pulmonary veins and then into the left atrium. On atrial contraction the blood flows into the left ventricle through the mitral valve. On ventricular contraction the left ventricle pumps blood through the aortic valve to the systemic circulation (aorta and the systemic vasculature).

The basic contractile element of the heart is the sarcomere, which is composed of myofilaments, contractile proteins that interdigitate and slide along one another during contraction. Shortening of the sarcomere is the functional unit of heart contraction. Physiologic and pharmacologic agents can change the contractile characteristics of the sarcomeres. Rate and contractility of the heart are controlled by sympathetic and parasympathetic innervation, as well as circulating catecholamines.

Control of Heart Performance

Mechanisms regulating cardiac (heart) output involve not only factors controlling performance of the heart as a pump, but also factors affecting the systemic vascular system and its resistance. Typically, the heart can increase its output to a level of almost five times its resting value. There are two methods by which the heart regulates its cardiac output in response to stress, injury, or disease: by changing heart rate and stroke volume.

Heart Rate Control

Heart rate can be changed rapidly and is thus one of the most effective ways for the heart to change its cardiac output. For a healthy person, an increase in heart rate can more than double the cardiac output when the heart rate increases to near 180 beats $\cdot$ min^{-1}. However, if a patient with heart disease increases their heart rate to $>$120 beats $\cdot$ min^{-1} they may have deleterious responses because of the increased demand for oxygen by the heart muscle. Blood flow in the heart muscle occurs primarily

during diastole (the relaxation phase of heart contraction). Increasing heart rate decreases the time for cardiac circulation during diastole. In normal subjects, decreasing the heart rate to ~ 50 beats $\cdot$ min^{-1} may not decrease cardiac output because there is increased diastolic filling time that increases stroke volume.

Stroke Volume Changes

The stroke volume of an intact ventricle is influenced by (*1*) ventricular end-diastolic volume (called preload), (*2*) ventricular afterload, and (*3*) contractility.

Preload. Preload is the term used to define the end-diastolic stress in the wall of the ventricle. For example, zero preload would result in the ventricle ejecting no blood. However, with increased preload, ventricular ejection generally increases linearly until the capacity of the pump (heart) is exceeded. Since the end-diastolic volume so profoundly influences the myocardial fiber length it has a great influence on the myocardial performance. The Frank–Starling law describes this principle and is illustrated graphically in Fig. 2. The most accessible measure of right ventricular preload is the right atrial pressure. Left atrial pressure is used to estimate left ventricular preload. Since the left ventricle does most of the work of the heart, it is usually the first part of the heart muscle to fail. Consequently, the measurement or estimation of the left atrial pressure is important in assessing a patient's hemodynamic status.

Afterload. Afterload is a measure of the impedance (resistance) against which the right or left ventricles must eject blood. Resistance (R) is calculated by measuring blood flow and pressure and then using Ohm's law {Eq. 1}.

$$R = \frac{\text{mean blood pressure}}{\text{cardiac output}} \quad (1)$$

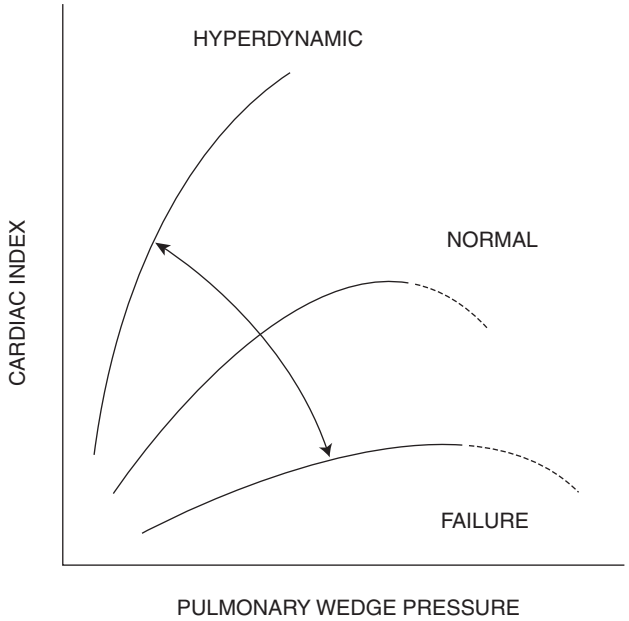

Figure 2. Frank–Starling curve of the heart showing the ventricular performance (cardiac index) plotted against the end-diastolic volume typically estimated by using pulmonary artery wedge pressure. Note to the right of these curves, there is a pulmonary wedge pressure above which the heart in ineffective in producing increased flow.

Systemic Circulation

Blood flow to the periphery of the body is controlled by local autoregulation and by the autonomic nervous system. Local autoregulation of blood flow helps tissue meet its oxygen requirements. For example, with decreased blood flow, metabolic byproducts increase, causing local vasodilatation that tends to increase blood flow. There are baroreceptors, similar to blood pressure transducers, located in the aortic arch and the carotid sinus which sense blood pressure. Via the baroreceptor reflex mechanism, the body regulates the blood pressure. In addition, chemoreceptors in the carotid sinus and other locations regulate respiration by responding to changes in CO_2 and O_2.

Pulmonary Circulation

The pulmonary arterial vessels differ markedly from systemic arterial vessels; they have thinner walls, less muscle, and have a resistance to blood flow about one-sixth that of the systemic circulation.

Contractility. Contractility is a measure of how a healthy heart performs. A healthy heart pumps vigorously and nearly empties its ventricles with each beat and is said to have excellent contractility. On the other hand, a compromised heart may not be able to empty effectively.

HEMODYNAMIC MONITORING

Bedside hemodynamic monitoring makes use of data gathering procedures that were formerly only done in diagnostic cardiac catheterization laboratories. Understanding the relationship between the pressure and blood flow in the cardiovascular system is the primary reason for performing hemodynamic monitoring. The cardiovascular system responds to many and varied stimuli and can be affected by physical conditioning, drugs, disease, blood loss, injury, and myocardial insult such as a heart attack. Because of the complexity of factors controlling the body, it is necessary to make hemodynamic measurements on the system to understand disease processes and provide optimum therapy to the patient.

Electrocardiogram

Electrocardiogram (ECG) monitoring is used to determine heart rate and detect arrhythmias, pacemaker function, and myocardial ischemia (lack of blood flow to the heart muscle). To permit optimum ECG monitoring the signal quality must be excellent (6). Since the ECG electrical signal from the heart is only 0.5–2.0 mV at the skin's surface, it is best measured by properly preparing the skin and optimally placing the electrodes. Skin can be properly prepared by removing oil from the surface and abrading the skin to remove the dry dead outer layer (stratum

granulosum). In 90% of patients, proper skin preparation reduces electrode resistance from as high as 200 to as low as 10 kΩ. Good electrode placement allows the electrodes to receive the maximum ECG signal with minimum noise. By placing the electrodes over bony prominence, such as the sternum or clavicles, muscle artifact (EMG) can be reduced. Motion artifact caused by movement of electrodes can be minimized not only by proper skin preparation, but also by taping a strain-relieving loop in the lead wires to prevent movement artifact. Shielded wire on the ECG leads helps minimizes pickup of alternating current (ac) electrical fields from 60 Hz power sources, electrosurgical units, and other sources like radio transmitters. The two leads that connect the patient form a loop through which magnetic fields pass and can induce unwanted voltages. Pickup from magnetic fields can be minimized by decreasing the loop area, by keeping the lead wires close together (usually twisted pairs), and by avoiding draping the lead wires over motors, lights, or other electrically powered instruments.

Electrocardiogram Arrhythmia Monitoring

Early in the development of monitoring techniques, the application of computer technology to detect patterns of the electrocardiogram caught the attention of those who sought to improve care of the critically ill. The computer appeared to be a logical candidate for relieving the nursing and medical staff of the tedious chore of continuously visually monitoring a multichannel oscilloscope.

Arrhythmia monitoring is one of the most sophisticated of the bedside monitor's tasks. People-based arrhythmia monitoring is expensive and unreliable, and those who do it find the task to be tedious and stressful. Today virtually every bedside monitor has rhythm monitoring built in. These monitors use computers and a variety of algorithms to detect and classify ECG rhythm abnormalities. Classifying these rhythm abnormalities is important to hemodynamic monitoring since irregular rhythms can cause dramatic inefficiencies in how the heart works as a pump. For example, Fig. 3 shows three strip recordings of the ECG and the corresponding pressure waveform from three different arrhythmias (ventricular tachycardia, couplet, which is two beats of abnormal electrical origin, and bigeminy where every other beat is from abnormal electrical origin). Note that several of the abnormal beats are hardly effective at creating any change in the arterial pressure. Those same beats deliver small stroke volumes to the patient's systemic circulation. As a consequence, one cannot assume that the cardiac output remains constant or increases just because the heart rate increases.

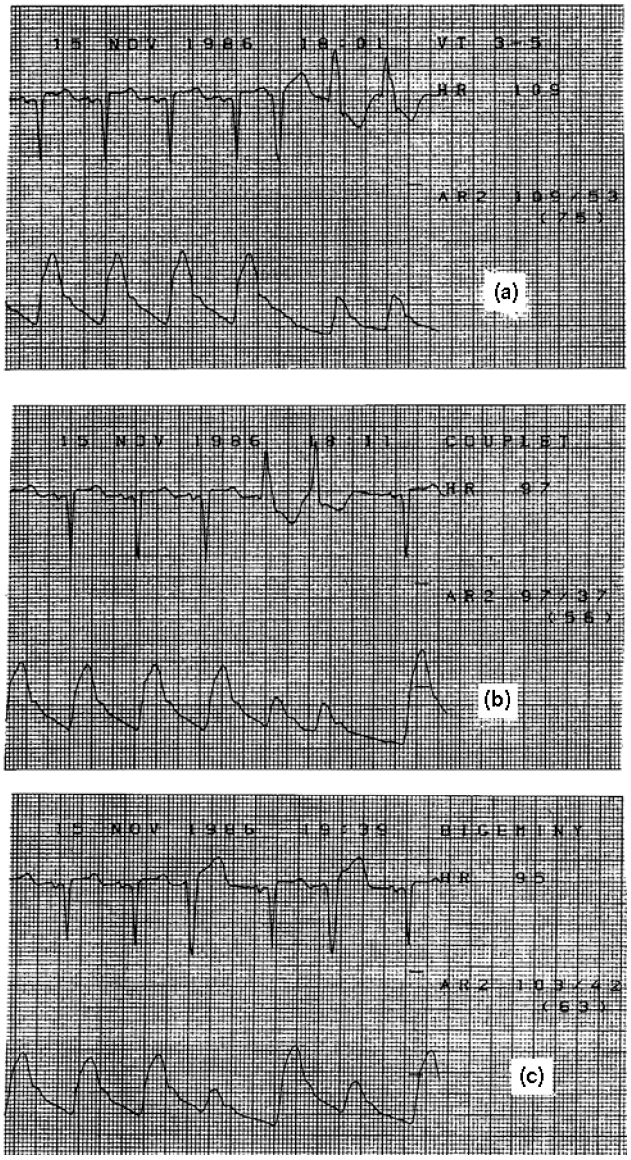

Figure 3. Electrocardiogram and arterial pressure waveforms with three different abnormal rhythms. (a) Ventricular tachycardia (VT), which occurs during the last two beats of the strip. (b) Couplets where two successive beats have an abnormal electrical origin. (c) Bigeminy where every other beat is from an abnormal electrical origin. Pressures are expressed in millimeters of mercury. For example, the patient in (a) has a systolic arterial pressure of 109 mmHg, a diastolic pressure of 53 mmHg, and a mean arterial pressure of 75 mmHg.

MEASUREMENTS

Blood Pressure Monitoring

Arterial blood pressure can be measured by both direct and indirect methods. However, central venous pressure (CVP), pulmonary artery (PA), and pulmonary capillary wedge pressure (PCWP) used to estimate left atrial pressure, at present, can only be measured by direct invasive methods.

Arterial Blood Pressure: Indirect Measurement; Using a Cuff. Recently, the American Heart Association has updated its recommendations on accurate "indirect" measures of blood pressure (7). The update reports that the auscultatory technique with trained observer and mercury manometer continues to be the method of choice for measurement of a patient's blood pressure in a physician's office. The report also suggests that hybrid devices that use electronic transducers rather than mercury have promise. The report indicates that oscillometric devices can also be used, but only after careful validation.

Unfortunately, the indirect measurement of arterial pressure has serious limitations for patients in shock usually signaled by low blood pressure. Also, since virtually all reliable indirect pressure measurement techniques require cuff inflation, such measurements can only be made intermittently.

Direct Blood Pressure Measurements. The direct measurement of blood pressure allows for continuous and accurate assessment of blood pressures. Direct and continuous pressure monitoring allows detection of dangerous hemodynamic events and provides the information necessary to initiate and regulate patient therapy to prevent catastrophic events. However, monitoring of pressures provides valuable information only when it is obtained in a technically satisfactory manner.

To accomplish direct blood pressure measurements, it is necessary to insert a catheter directly into the cardiovascular system (8). This invasive technique has risks that must be weighted against the benefits that can be obtained. These risks include, infection, blood loss, insertion site damage and other factors (9,10). For many patients who are in shock or who have cardiac disease, the benefits far outweigh the risks. Formal methods for assessing these risks have been published by the Coalition for Critical Care Excellence (11).

Blood pressure can be measured on both the pulmonary (right heart) and systemic (left heart) sides of the circulatory system. Measurements of both pulmonary and systemic parameters yield different and important cardiovascular status. The CVP reflects the patient's circulating blood volume or venous tone, and right atrial and ventricular pressures (right ventricular preload). To measure the right atrial pressure accurately a catheter must be placed in a major vein within the chest or directly in the right atrium. The CVP values fluctuate about atmospheric pressure. The level of the right heart is usually taken as the zero reference point from which all other blood pressures are measured. The CVP gives an indication of only the function of the right heart, and not left heart's performance.

To measure the left atrial pressure, it is necessary to place a catheter tip through the atrial septum from the right atrium (usually done only with fluoroscopic control in the cardiac catheterization laboratory) or estimating it by placing a pulmonary artery (Swan–Ganz) catheter in the wedged position by inflating its balloon near the catheter tip.

EQUIPMENT

Components of Direct Pressure Monitoring Systems

The components of a direct blood pressure monitoring system for critically ill patients are shown in Fig. 4 (6,8). The components numbered 1–7 in the figure are known as

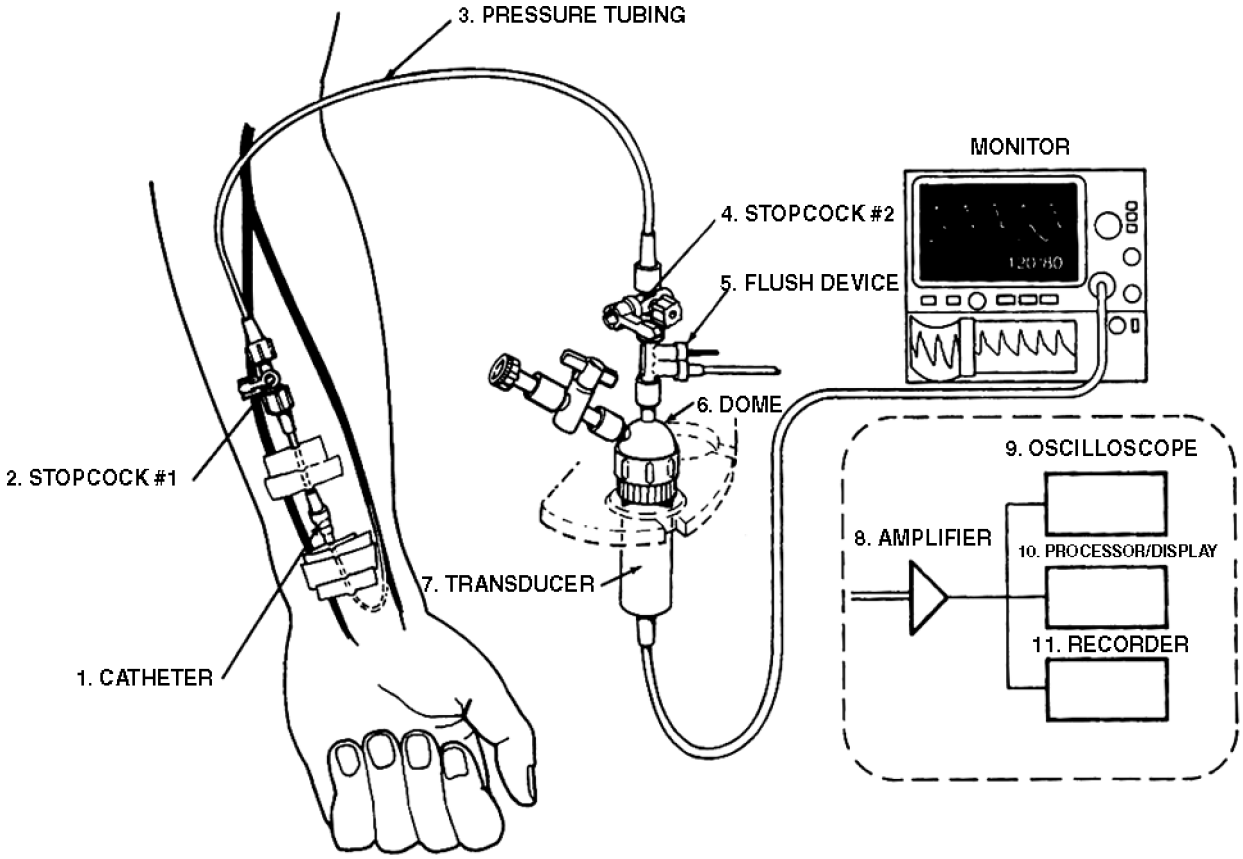

Figure 4. The 10 components used to monitor direct blood pressure. The monitoring components are nearly independent of whether the catheter is in an artery (radial, brachial, or femoral) or in the pulmonary artery. Size of transducer and plumbing components are enlarged for illustration purposes. [Reproduced from Ref. 6, with permission.]

the "plumbing" system and must always be sterile because the fluid contained therein comes in direct contact with the patient's blood. Today virtually all of these components are disposable or single-use items to minimize patient infection. Components 8–11 in Fig. 4 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 samples for blood to allow determination of blood gas and other laboratory testing parameters. These catheters are typically placed by the percutaneous method, either by the Seldinger "over-the-needle" technique or by introducing the catheter through a needle (8).
2. Sampling stopcock. Stopcock 1 is used as a site for withdrawing blood for analysis. When filling the catheter-tubing-transducer system with fluid, precautions must be taken to be sure all central switching cavities of the stopcock are filled and that entrapped air bubbles are removed. Because stopcocks are especially vulnerable sources of patient contamination, these devices must be handled with extreme care; ports not in active use should be covered with sterile caps and medical personnel should never touch open ports of the stopcocks.
3. Pressure tubing. The catheter and stopcock are normally attached to a continuous flush device and transducer by noncompliant pressure tubing. To optimize the dynamic response of the catheter-tubing-transducer system, long lengths of tubing must be avoided.
4. Stopcock 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 when initially filling the pressure monitoring system, but also to help prevent blood from clotting in the catheter. These devices provide a "continuously flush" of fluid at a rate of from 1 to 3 $mL \cdot h^{-1}$.

6,7. Transducer dome and Pressure transducer. Today virtually all transducers used for monitoring are highly reliable, standardized, disposable devices (12,13).

8. Amplifier system. The output voltage required to drive an oscilloscope or strip-chart recorder is provided by an amplifier system inserted between the transducer and display. Pressure transducer excitation is provided either from a direct current (dc) or ac source at a voltage of 4–8 V revolutions per second (rms). Most amplifier systems 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 pressure waveform distortion.
9. Oscilloscope. Pressure waveforms are best visualized on a calibrated oscilloscope or other similar display panel.
10. Digital processing and display. Digital displays provide a simple method for presenting quantitative data from the pressure waveform. They are found on most modern pressure monitoring equipment. Systolic, diastolic, and mean pressure are typically derived from the pressure waveforms.
11. Strip-chart recorders. Frequently, strip-chart recorders are used to document dynamic response characteristics, respiratory variations in pulmonary artery pressures, and aberrant rhythms and pressure waveforms.

STATIC CALIBRATION

Zeroing and calibrating the transducer are two important steps in setting up the direct pressure-monitoring system.

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 (14). 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 the atmosphere and aligning the resulting fluid-air interface with the midaxillary reference point.

Once the system is zeroed the stopcock can be switched to allow the patient's waveform to be displayed. Pulmonary artery and pulmonary artery wedge pressure are especially susceptible to improper zeroing and should be measured only after the zero point has been verified.

Sensitivity Calibration

The sensitivity of most pressure transducers is fixed at 5.0 $\mu V \cdot V^{-1}$ of excitation applied per 1 mmHg (0.13 kpa) and calibrated by the manufacturers to within $\pm 1\%$. This degree of accuracy is adequate for clinical purposes. As a consequence standardized transducers need only to be zeroed to obtain accurate pressure measurements (12,13).

CHECKING DYNAMIC RESPONSE

In the critical care setting, where most hemodynamic monitoring is carried out, the catheter-tubing-transducer systems used can usually be characterized as an underdamped second-order dynamic system analogous, for example, to a bouncing tennis ball. A second-order dynamic system can be expressed mathematically by a second-order differential equation with characteristics determined by three mechanical parameters: elasticity, mass, and friction. These same parameters apply to a catheter-tubing-transducer system where the natural frequency (f_n in Hz) and damping coefficient determine the dynamic characteristics for a catheter-tubing-transducer system. For an underdamped second-order system f_n and define the system's

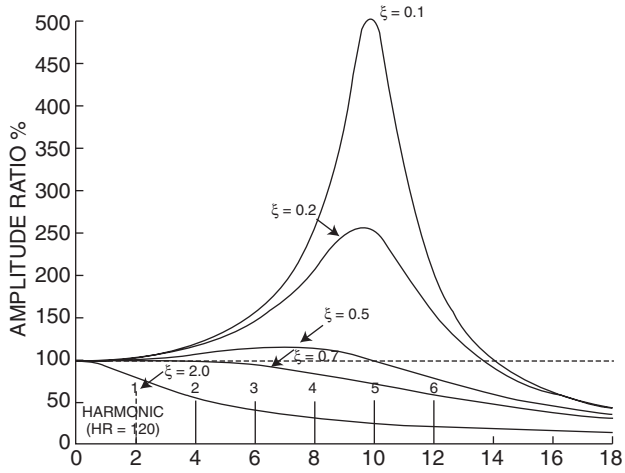

Figure 5. Family of frequency versus amplitude ratio plots for five different damping coefficients ζ and natural frequencies f_n of the plot shown is 10 Hz. When $\zeta = 0.1$, the system is very underdamped, and when $\zeta = 2.0$, it is overdamped. The dashed line shows the frequency versus amplitude characteristic that 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 were 120 beats $\cdot$ min^{-1} (2 beats $\cdot$ s^{-1}). Note that by the fifth harmonic (10 Hz) if $\zeta = 0.1$ the true signal would be amplified five times. If $\zeta = 2.0$ there would be an attenuation to about one-fourth of the amplitude. In both cases there would be a gross waveform distortion because neither situation reflects a high fidelity system dynamic response. Fidelity of the system can be improved by increasing the f_n or adjusting ζ to be in the range of 0.5–0.7. [Reproduced from Ref. 6, with permission.]

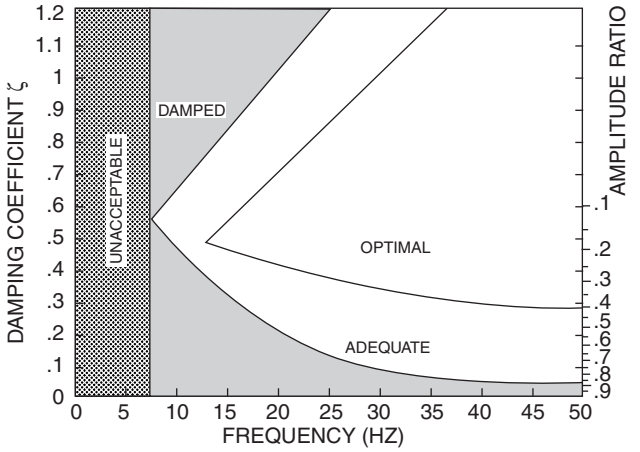

Figure 6. Plot of f_n versus ζ illustrating the five areas into which catheter-tubing-transducer systems fall. Systems in the optimal area will reproduce even the most demanding (fast heart rate and rapid systolic upstroke) arterial or pulmonary artery waveforms without distortion. Systems in the adequate area will reproduce most typical patient waveforms with little or no distortion. All other areas will cause serious and clinically important waveform distortion. Note the scale on the right can be used to estimate ζ from the amplitude ratio determined during fast flush testing (11). See Fig. 8 for an example of waveforms. [Reproduced from Ref. 6, with permission.]

dynamic characteristics. In the clinical setting f_n and can be measured easily and conveniently by using the "fast-flush" method.

Dynamic response characteristics of catheter-tubing-transducer systems have been defined by two interrelated techniques. The first technique specifies the frequency bandwidth and requires that the system frequency response 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 (Fig. 5).

The second technique specifies f_n and The plot of f_n and in Fig. 6 has five areas (6,15). If the characteristics of the catheter-tubing-transducer "plumbing" system fall in the adequate or optimal area of the graph, the pressure waveforms will be adequately reproduced. If the characteristics fall into one of the remaining three areas, there will be pressure waveform distortion. Most catheter-tubing-transducer systems assembled under optimal conditions are underdamped, but a few fall into the unacceptable areas of the chart. Methods for optimizing the catheter-tubing-transducer system components have been outlined (15–20). In the clinical setting, there are dramatic differences between each patient setup; therefore it is mandatory to verify the adequacy of each pressure-monitoring system by testing them. The testing can be done easily using the fast-flush technique.

A fast flush is produced by opening the valve of the continuous flush device, for example, by pulling and quickly releasing the pigtail valve on a continuous flush device. The rapid valve closure generates a near square wave pressure signal from which f_n and of the catheter-tubing-transducer system can be measured.

Once the fast-flush test has been executed two or three times, the dynamic response characteristics (f_n and) can quickly and easily be determined. Natural frequency f_n can be estimated by measuring the period of each full oscillation on a strip-chart recorder following a fast flush (Fig. 7a) and calculating the frequency from the period. Damping coefficient can be determined by using the amplitudes of any two successive peak-to-peak values measured after a fast flush. The amplitude ratio is calculated by dividing the measured height of the smaller peak-to-peak value by that of the amplitude of the larger peak-to-peak value (Fig. 7b). The amplitude ration can then be converted to a damping coefficient by using the scale in the right side of Fig. 6.

Once f_n and have been determined, these data can be plotted on the graph of Fig. 6 to ascertain the adequacy of dynamic response. Some bedside monitors and recorders may compromise the clinical user's ability to use the fast-flush technique because the monitors have built-in low-pass filters. These filters should be expanded to at least 50 Hz or be eliminated.

Several factors lead to poor dynamic responses: (*1*) air bubbles in the system usually caused by a poor initial catheter-tubing-transducer system setup, (*2*) pressure tubing that is too long, too compliant, or a diameter that is too small, and (*3*) pressure transducers that are too compliant. The best way to enhance the system's dynamics is to improve f_n.

Invasive pressure monitoring systems have patient risks, such as a source of infection and air embolism. In addition, great care is required by clinical users to optimize

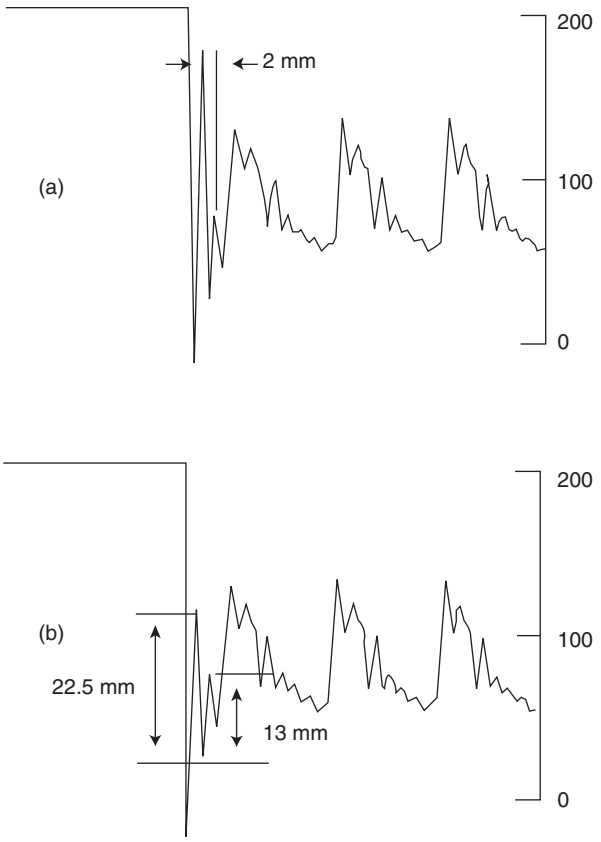

Figure 7. Measuring dynamic response parameters from the fast-flush waveform, (a) The natural frequency f_n can be determined by using a strip-chart recording to measure the period of one full oscillation, as shown. In this example, one full cycle is 2 mm and at a paper speed of 25 mm $\cdot$ s^{-1} this results in $f_n = 12.5$ Hz $=$ 25 mm $\cdot$ s^{-1}/2 mm. (b) Determining the damping coefficient ζ required measuring two successive peak-to-peak values of the resulting oscillations. The amplitude ratio of the two successive peaks is taken, giving a value of $0.58 = 13/22.5$. With use of the amplitude ratio and the scale on the right side of Fig. 6, the damping coefficient $\zeta = 0.17$. Plotting the natural frequency and damping coefficient on Fig. 6 shows that this system is underdamped.

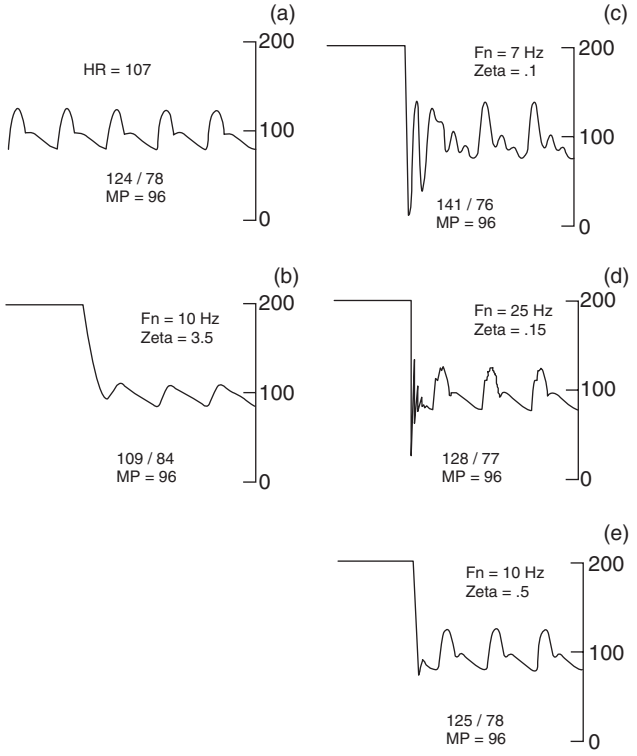

Figure 8. Arterial pressure waveforms were obtained from the same patient. Shown are Systolic/Diastolic and Mean Pressure (*MP*). In panel the (a) Patient's actual arterial pressure waveform as if recorded with a catheter-tipped transducer is shown, (b) shown the same patient's arterial waveform recorded with an overdamped system ($\zeta = 3.5$). Note the fast-flush signal (upper left) returns slowly to the patient waveform. Systolic pressure is underestimated, diastolic pressure is overestimated, and *MP* is unchanged, (c) An underdamped condition ($\zeta = 0.1$) with low $f_n = 7$ Hz. After the fast flush, the pressure signal oscillates rapidly (rings). Systolic pressure is overestimated, diastolic is slightly underestimated, and MP is correct, (d) shows an underdamped condition ($\zeta = 0.15$), but with high $f_n = 25$ Hz. The pressure waveform is slightly distorted and systolic, diastolic, and mean pressures are close to the actual pressures, (e) shown an ideally damped pressure monitoring system ($\zeta = 0.5$). The undershoot after the fast flush is small and the original patient waveform is adequately reproduced. [Reproduced from Ref. 6, with permission.]

dynamic response and proper zeroing to provide accurate and reliable data. Merely looking at pressure waveforms will not provide the information required to determine the adequacy of the system's dynamic response (see Fig. 8). Fast-flush testing to determine these parameters is essential.

SIGNAL AMPLIFICATION, PROCESSING, AND DISPLAY

Once the pressure signal has been transmitted to the transducer, the 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.

Placement of the Pulmonary Artery Catheter

The balloon-tipped, flow-directed, pulmonary artery catheter (Swan–Ganz) came into widespread use in 1970 (21). The follow-up development by Ganz of a practical thermal dilution attachment to the pulmonary artery catheter permitted convenient and easy measurement of cardiac output (22). Since these early developments with the Swan–Ganz catheter, the pulmonary artery catheter has been fitted with optical fibers which allow measurement of mixed venous oxygen saturation (23).

The pulmonary artery catheter is inserted into the right side of the circulation using the percutaneous technique typically using entry from either the internal jugular or the subclavian vein. The catheter is floated into the pulmonary artery without use of fluoroscopy, using the hemodynamic pressure waveforms as a guide (Fig. 9).

Accurate Measurement of Pulmonary Artery Pressure. 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

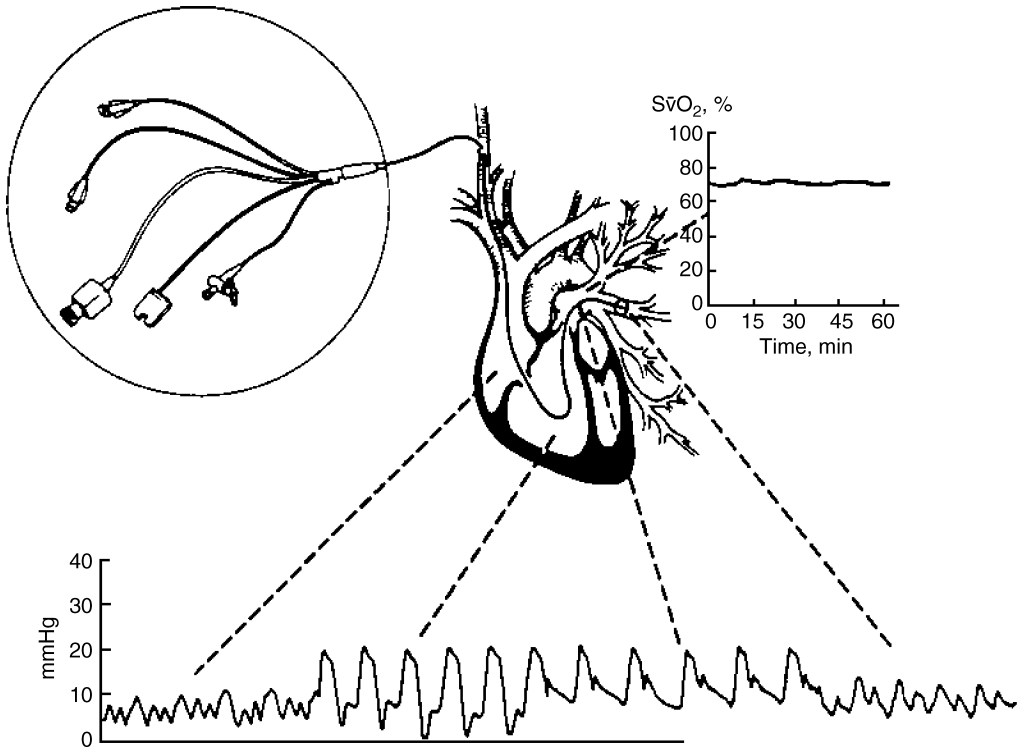

Figure 9. Composite illustration showing normal pressure waveforms obtained as a fiber optic balloon flotation pulmonary artery catheter (Swan-Ganz) is advanced from the right atrium to the pulmonary artery wedge position. [From Daily and Tilkian in Reading List (1986), with permission.]

usually inserted may lead one to conclude that the measurements of pulmonary artery and wedge pressure (PCWP) are easily and reliably measured. However, such is not the case.

Pulmonary artery pressures can be measured accurately only if the following steps are taken (24–27):

1. The monitor is properly zeroed.
2. Strip-chart recordings of all PA pressures for a time period covering at least three respiratory cycles are obtained. Using only the monitor's digital displays is insufficient.
3. Dynamic response testing (fast flush) should be obtained when the catheter is in each position (i.e., wedge and PA). If the dynamic response is not adequate, the problems with the catheter-tubing-transducer system must be resolved before accurate pressures can be measured.
4. Pressures (i.e., systolic, diastolic, and mean pressures) should be assessed from a monitor's display or a strip-chart recording. The pressure measures should be made at the end expiration when the transmural pressure is nearest zero.

CARDIAC OUTPUT DETERMINATION

Cardiac output is the volume of blood ejected by the heart every minute. Cardiac output is a helpful measurement since it can be used to evaluate the overall cardiac status of the critically ill patient, as well as help make the diagnosis of cardiovascular disease. Ideally a cardiac output measurement system would be continuous, automatic, minimally invasive, accurate, fast, inexpensive, and easy to use clinically. The most common method used to measure cardiac output in critically ill patients is still the indicator dilution method. The pulmonary artery catheter (Swan–Ganz) introduced in the 1970s revolutionized the ease with which cardiac output could be measured.

The thermal dilution method requires injection of cold physiological solution, usually normal saline, into the superior vena cava or right atrium. Cardiac output is determined by measuring the area under the time–temperature curve measured in the pulmonary artery that results from the injection of the cold solution.

The thermal dilution method for determining cardiac output relies on several assumptions that are not always correct. First, the exact amount of thermal indicator injected cannot be quantitated precisely. Second, indicator is lost at various stages and this loss of indicator (heat loss) leads to errors.

A block diagram of the thermal dilution measuring system with typical thermal dilution curves and time of injection indicated are shown in Fig. 10. Figure 10c and d show the transit time for the cooled blood moving from the injection site in the right atrium to the pulmonary artery measurement site. Calculation of cardiac output requires measuring the area under the curve. Consequently, a baseline temperature must be established before the injection. In turn, the end point is usually determined by extrapolating to the baseline temperature.

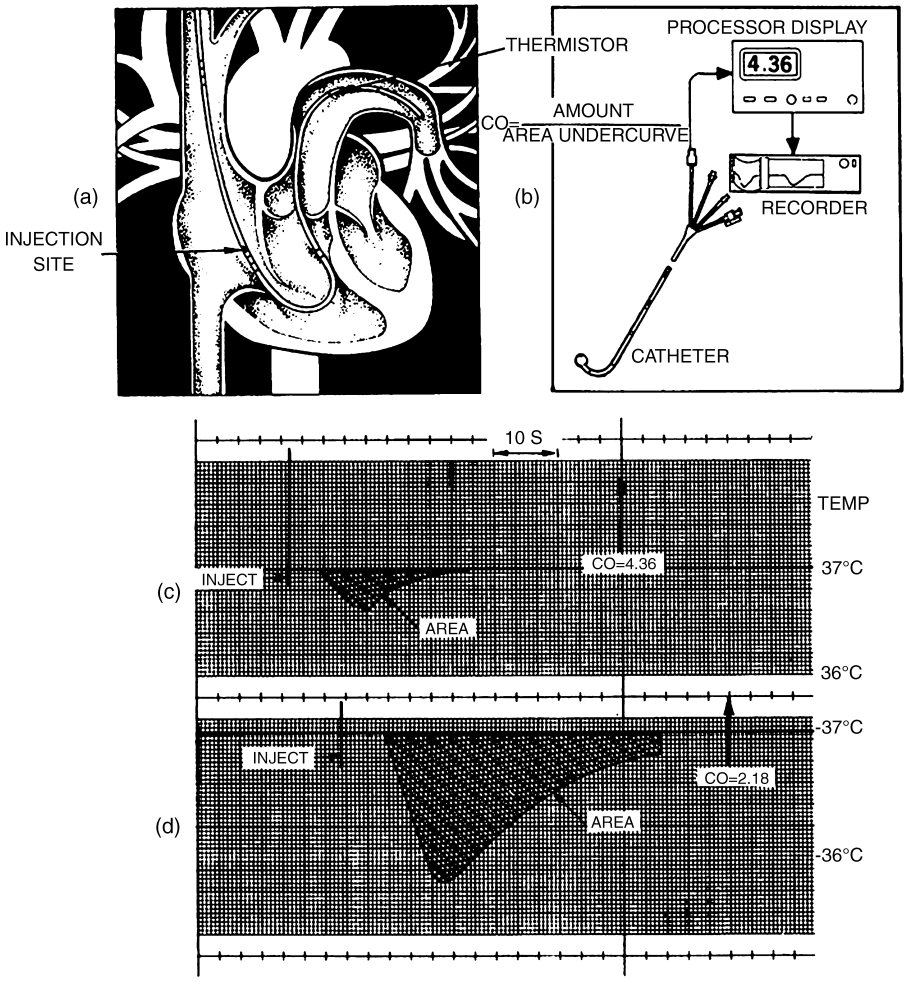

Figure 10. Schematic diagram of the thermal dilution measurement of cardiac output. A recorder of some type should always be used to verify the quality of the thermal dilution curve, (a) shows the thermal dilution catheter placed into the pulmonary artery. Note the location of injection site and thermistor, (b) shows the connection of the thermal dilution catheter connected to a cardiac output processor and recorder, (c) shows a typical temperature-time plot sensed by the thermistor near the catheter tip after an iced saline injection. The cardiac output determined in this case was 4.36 $L \cdot min^{-1}$. (d) Shows a similar temperature–time plot for a patient with low cardiac output (2.18 $L \cdot min^{-1}$). Note the larger area and broader dispersion of the waveform caused by the lower flow. [Reproduced from Ref. 9, with permission.]

To ensure that accurate thermal dilution cardiac output results are obtained, it is recommended that the thermal dilution curves be presented on a monitoring screen or on a strip-chart recorder. Studies have shown that synchronizing the injections with the respiratory cycle improves the technique's reproducibility (28). Since there is considerable variability in cardiac output between measures, at least three reproducible curves are usually obtained. Averaging the findings from these three curves gives a more representative assessment of cardiac output.

In recent times, the complexities of using the Swan–Ganz pulmonary artery catheter have result in controversies. Some clinicians feel that such systems should only be used only when needed and then only sparingly while others have a differing viewpoint (29,30). Still others have questioned the ability of making accurate central venous and pulmonary artery occlusion pressures and whether it matters (26,27). Many of those issues will be resolved in the future when there might be better methods for measuring hemodynamic parameters. Until that time, physicians and nurses caring for critically ill patients who require hemodynamic monitoring should be aware of several how to oriented publications (31–34).

Alarming Based on Hemodynamic Parameters. Clinical hemodynamic monitoring is now several decades old. What started from a simple beginning has since seen many dramatic changes in both the development of new medical devices and skills of the clinicians to use those devices. However, it is my feeling that we are not yet at optimum hemodynamic monitoring. Some recent publications on the topic are illustrative. Sander and colleagues in Germany have recently looked at categories of patients with elevated heart rates who are at higher risk of cardiac complications (35). Their work resulted in an editorial comment Vital are vital signs (36). Additional recent work at Vanderbilt University indicates that volatility in vital signs is a good predictor of death in trauma patients (37). Finally, the problem of false alarms continues to be a huge problem with current bedside monitors. As part of a Master of Science thesis in Medical Informatics at the University of Utah, an investigator found that only about one-third of the standard alarms for patients in a variety of ICU care were true alarms. Thus about two-thirds of the alarms are false. However, if the alarming system used heart rates determined from both the ECG and the Arterial Blood Pressure, the number of alarms decreased by $\sim$ 50% and the false alarm rate was only $\sim$ 25% (38). Having smarter and better hemodynamic monitoring with better and smarter alarming systems will be crucial for to future monitoring systems.

COMPUTERIZED DECISION SUPPORT

Much has been learned about hemodynamic measurements and how to use the data to calculate derived patient parameters. These parameters can then be used to determine patient status and augment patient therapy (39–42).

Using hemodynamic data available from bedside monitors and combining that data, in a structured and coded electronic patient record allows for optimal computerized decision support (42–44). Morris and his colleagues have stated the value of computerized decision support well (45). Only adequately explicit protocols contain enough detail to lead different clinicians to the same decision when faced with the same clinical scenario. Guidelines of care provide only general guidance to patient care and require clinicians considerable latitude in which care decision should be made. Computerized protocols, on the other hand, can be patient-specific and evidence based (46). Using computerized decision-support tools variation in clinical practice can be reduced and favorable effects on improve patient outcomes can be accomplished (45,46).

FUTURE

There are still needs for improvement in hemodynamic monitoring. Being able to make the measurements continuously, less invasively, and more reliably are areas where progress is needed. Clearly, using computer aided decision-support technology to help reduce false alarms and to guide clinicians in making better patient diagnosis and more timely and more optimal and effective treatment decision offer ample opportunity for future research and progress.

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See also BLOOD PRESSURE MEASUREMENT; CARDIAC OUTPUT, FICK TECHNIQUE FOR; CARDIAC OUTPUT, INDICATOR DILUTION MEASUREMENT; CARDIAC OUTPUT, THERMODILUTION MEASUREMENT; ELECTROCARDIOGRAPHY, COMPUTERS IN; HEMODYNAMICS.