use a common piezoelectric transducer, so a *gate* is required to turn off the signal from the transmitter during reception. A one-stage gate is not sufficient to isolate the large transmitter signals from the very small received signals. Therefore, two gates in series are used to turn off the transmitter.

The optimal transmitted signal is a pulse-modulated sine-wave carrier. Although it is easy to generate this burst electrically, it is difficult to transduce this electric burst to a similar acoustic burst. The crystal transducer has a high Q (narrow bandwidth) and therefore rings at its resonant frequency long after the electric signal stops. Therefore, the transducer is modified to achieve a lower Q (wider bandwidth) by adding mass to the back [Figure 8.9(d)] or to the front [Figure 8.9(e)]. The Q is not lowered to a desirable value of about 2 to 5, because this would greatly decrease both the efficiency of the transmission and the sensitivity of the reception. The Q is generally 5 to 15, so some ringing still exists.

When we generate a short sine-wave burst, we no longer have a single frequency. Rather, the pulse train of the repetition rate is multiplied by the carrier in time, producing carrier sidebands in the frequency domain. This spectrum excites the transducer, producing a field that is more complex than that for continuous-wave excitation. This causes spectral spreading of the received signal.

LASER DOPPLER BLOOD FLOWMETER

In a laser Doppler blood flowmeter, a 5 mW He–Ne laser beams 632.8 nm light through fiber optics into the skin (Khaodhiar and Veves, 2006). Moving red blood cells in the skin frequency shift the light and cause spectral broadening. Reflected light is carried by fiber optics to a photodiode. Filtering, weighting, squaring, and dividing are necessary for signal processing. Capillary blood flow has been studied in the skin and many other organs.

8.5 THERMAL-CONVECTION VELOCITY SENSORS

PRINCIPLE

The thermodilution methods described in Sections 8.1 and 8.2 depend on the mixing of the heat indicator into the entire flow stream. In contrast, thermal velocity sensors depend on convective cooling of a heated sensor and are therefore sensitive only to local velocity.

Figure 8.13(a) shows a simple probe. The thermistor R_u is heated to a temperature difference ΔT above blood temperature by the power W dissipated by current passing through R_u. Experimental observations (Grahn *et al.*, 1969) show that these quantities are related to the blood velocity u by

$$\frac{W}{\Delta T} = a + b \log u \tag{8.20}$$

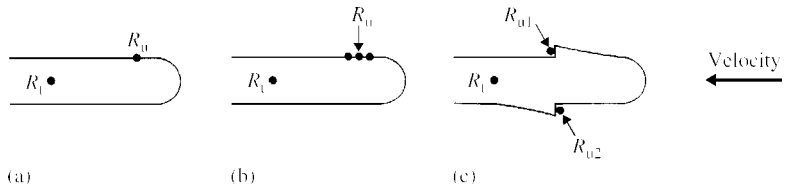


Figure 8.13 Thermal velocity probes (a) Velocity-sensitive thermistor R_u is exposed to the velocity stream. Temperature-compensating thermistor R_t is placed within the probe. (b) Thermistors placed down- and upstream from R_u are heated or not heated by R_u, thus indicating velocity direction. (c) Thermistors exposed to and shielded from flow can also indicate velocity direction.

where a and b are constants. Thus the method is nonlinear, with a large sensitivity at low velocities and a small sensitivity at high velocities.

PROBES

Catheter-tip probes are designed with two types of sensors (Cobbold, 1974). The first type uses the thermistors shown in Figure 8.13 and provides a high sensitivity and reasonable resistance values. Because the thermistor shown in Figure 8.13(a) is cooled equally for both directions of velocity, the output of the instrument is a full-wave-rectified replica of the true velocity. To overcome this limitation, the probe shown in Figure 8.13(b) has two additional thermistors located a few tenths of a millimeter downstream and upstream from R_u. Depending on the direction of velocity, one or the other is heated by the heat carried through the blood from the thermistor R_u. These two additional thermistors are placed in a bridge that is balanced for zero velocity. A comparator detects the bridge unbalance and switches the output from positive to negative. The probe shown in Figure 8.13(c) uses two velocity sensors arranged so that one is exposed to the fluid velocity while the other is shielded from the fluid velocity.

The second type of sensor uses a glass bead with a thin strip of platinum deposited on its surface. The platinum may be painted on and then fired in a furnace, or it may be *sputtered* (deposited by electric discharge in a vacuum). A disadvantage of platinum-film sensors is their low resistance (a few ohms) and low sensitivity.

A real question arises about what is actually being measured. When a catheter is inserted into a blood vessel, the sensor may be centered and thus measure maximal velocity, or it may be against the wall of the vessel and thus measure a low velocity. One way of ensuring that the sensor is not against the wall is to rotate the catheter, searching for the maximal output. Catheters are also sensitive to radial velocity of blood, as well as to radial vibrations of the catheter (catheter whip). Thus, in addition to any errors due to measuring velocity, errors in trying to estimate flow can arise from lack of knowledge about location of the sensor. Either type of probe (if it is made sufficiently small) can be placed at the end of a hypodermic needle and inserted perpendicular to the vessel for measuring velocity profiles.

CIRCUIT

A *constant-current* sensor circuit cannot be used for two reasons. First, the time constant of the sensor embedded in the probe is a few tenths of a second—much too long to achieve the desired frequency response of 0 to 25 Hz. Second, to achieve a reasonable sensitivity at high velocities, the sensor current must be so high that when the flow stops, lack of convection cooling increases the sensor temperature more than 5 °C above the blood temperature and fibrin coats the sensor.

The *constant-temperature* sensor circuit shown in Figure 8.14 overcomes both of these problems. The circuit is initially unbalanced by adjusting R_1. The unbalance is amplified by the high-gain op amp, and its output is fed back to power the resistance bridge. Operation of the circuit is as follows: Assume that thermistor R_u is 5 °C higher than blood temperature because of self-heating. If the velocity increases, R_u cools and its resistance increases. A more positive voltage enters the noninverting op-amp terminal, so v_b increases. This increases bridge power and R_u heats up, thus counteracting the original cooling. The system uses high-gain negative feedback to keep the bridge always in balance. Thus R_u remains nearly constant, and therefore its temperature remains nearly constant. The high-gain negative feedback divides the sensor time constant by a factor equal to the loop gain, so frequency response is greatly improved. In effect, if the sensor becomes slightly cooled, the op amp can provide a large quantity of power to rapidly heat it back to the desired temperature.

The circuit operates satisfactorily with only one sensor, R_u, provided that the blood temperature is constant. Should the blood temperature vary, a temperature–compensating thermistor R_t is added to keep the bridge in balance. So that its rise in temperature is very small, R_t must have a much lower resistance–temperature coefficient than R_u, to ensure that R_t is a sensor of temperature and not of velocity. The thermal resistance of R_t can be lowered by making it large in size, by using a heat sink, or by placing it within the probe so that the effective cooling area is much larger. Another solution is to increase

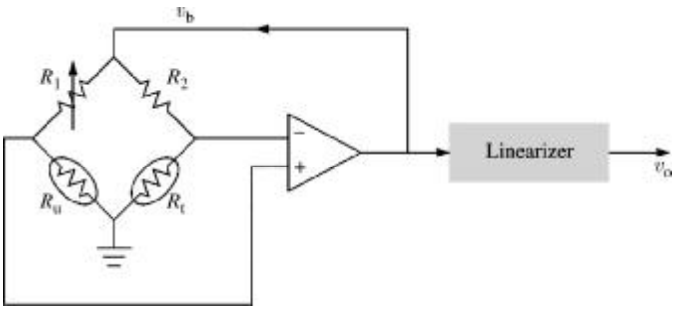


Figure 8.14 Thermal velocity meter circuit A velocity increase cools R_u, the velocity-measuring thermistor. This increases voltage to the noninverting op-amp input, which increases bridge voltage v_b and heats R_u. R_t provides temperature compensation.

the resistance values for R_2 and R_t so that their power dissipation is much lower.

A linearizer is required to solve (8.20). We may square v_b to obtain W and then use an antilog converter to obtain v_o. For the directional probe shown in Figure 8.17(b), a unity-gain inverting amplifier and switch may be used to yield the direction of flow.

Calibration can be accomplished by using a sinusoidal-flow pump or a cylindrical pan of liquid rotating on a turntable.

The main use of thermal-velocity sensors is to measure the velocity of blood and to compile velocity profiles in studies of animals, although such sensors have also been regularly used to measure velocity and acceleration of blood at the aortic root in human patients undergoing diagnostic catheterization. The same principle has also been applied to the measurement of the flow of air in lungs and ventilators by installing a heated platinum wire in a breathing tube.

8.6 CHAMBER PLETHYSMOGRAPHY

Plethysmographs measure changes in volume. The only accurate way to measure changes in volume of blood in the extremities noninvasively is to use a chamber plethysmograph. By timing these volume changes, we can measure flow by computing $F = dV/dt$. A cuff is used to prevent venous blood from leaving the limb—hence the name *venous-occlusion plethysmography* (Seagar *et al.*, 1984).

EQUIPMENT

Figure 8.15 shows the equipment used in a venous-occlusion plethysmograph. The chamber has a rigid cylindrical outer container and is placed around the leg.

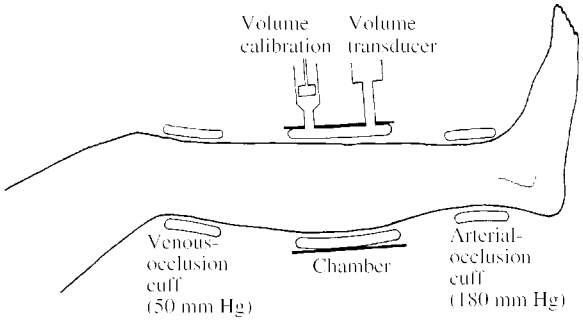


Figure 8.15 In chamber plethysmography, the venous-occlusion cuff is inflated to 50 mm Hg (6.7 kPa), stopping venous return. Arterial flow causes an increase in volume of the leg segment, which the chamber measures. The text explains the purpose of the arterial-occlusion cuff.

7

BLOOD PRESSURE AND SOUND

Robert A. Peura

Determining an individual's blood pressure is a standard clinical measurement, whether taken in a physician's office or in the hospital during a specialized surgical procedure. Blood-pressure values in the various chambers of the heart and in the peripheral vascular system help the physician determine the functional integrity of the cardiovascular system. A number of direct (invasive) and indirect (noninvasive) techniques are being used to measure blood pressure in the human. The accuracy of each should be established, as well as its suitability for a particular clinical situation.

Fluctuations in pressure recorded over the frequency range of hearing are called *sounds*. The sources of heart sounds are the vibrations set up by the accelerations and decelerations of blood.

The function of the blood circulation is to transport oxygen and other nutrients to the tissues of the body and to carry metabolic waste products away from the cells. In Section 4.6 we pointed out that the heart serves as a four-chambered pump for the circulatory system. This is illustrated in Figure 4.12. The heart is divided into two pumping systems, the right side of the heart and the left side of the heart. The pulmonary circulation and the systemic circulation separate these two pumps and their associated valves. Each pump has a filling chamber, the atrium, which helps to fill the ventricle, the stronger pump. Figure 7.15 is a diagram that shows how the electric and mechanical events are related during the cardiac cycle. The four heart sounds are also indicated in this diagram.

Figure 7.1 is a schematic diagram of the circulatory system. The left ventricle ejects blood through the aortic valve into the aorta, and the blood is then distributed through the branching network of arteries, arterioles, and capillaries. The resistance to blood flow is regulated by the arterioles, which are under local, neural, and endocrine control. The exchange of the nutrient material takes place at the capillary level. The blood then returns to the right side of the heart via the venous system. Blood fills the right atrium, the filling chamber of the right heart, and flows through the tricuspid valve into the right ventricle. The blood is pumped from the right ventricle into the pulmonary artery through the pulmonary valve. It next flows through the pulmonary arteries, arterioles, capillaries, and veins to the left atrium. At the pulmonary capillaries, O_2 diffuses from the lung alveoli to the blood,

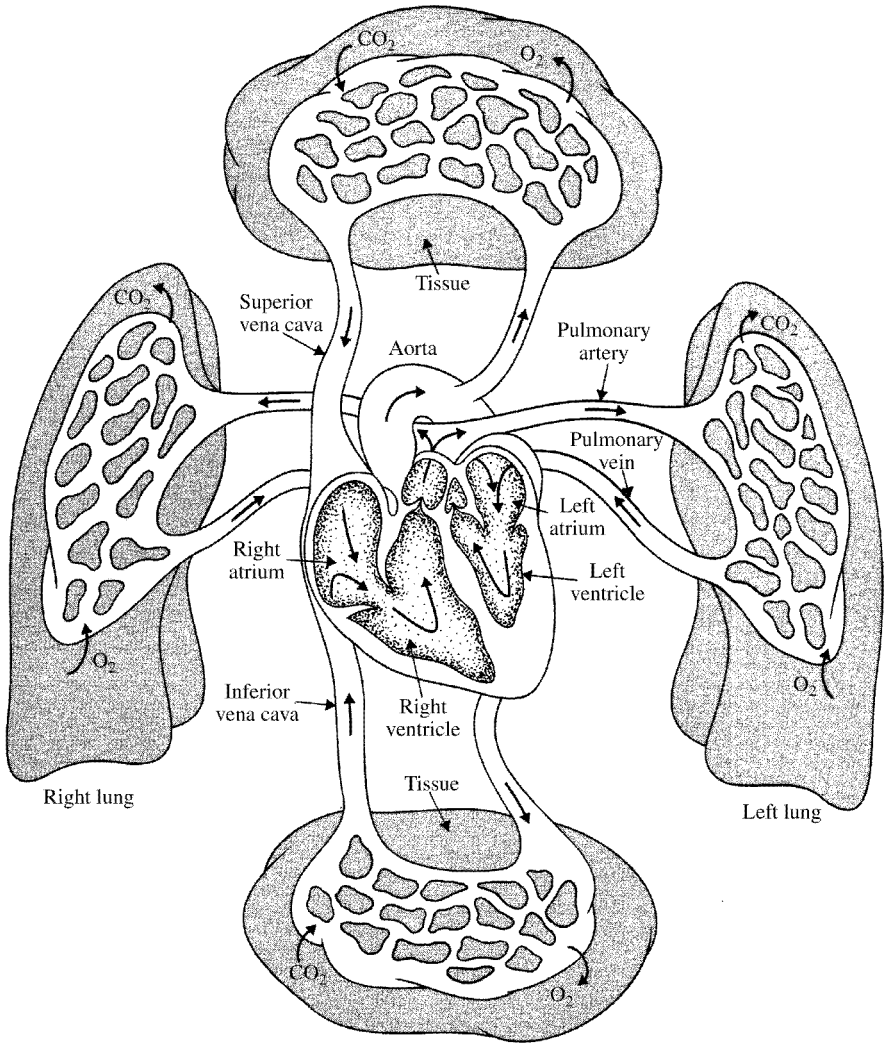


Figure 7.1 The left ventricle ejects blood into the systemic circulatory system. The right ventricle ejects blood into the pulmonary circulatory system.

and CO_2 diffuses from the blood to the alveoli. The blood flows from the left atrium, the filling chamber of the left heart, through the mitral valve into the left ventricle. When the left ventricle contracts in response to the electric stimulation of the myocardium (discussed in detail in Section 4.6), blood is pumped through the aortic valve into the aorta.

The pressures generated by the right and left sides of the heart differ somewhat in shape and in amplitude (see Figure 7.2). As we noted in Section 4.6, cardiac contraction is caused by electric stimulation of the cardiac muscle. An electric impulse is generated by specialized cells located in the sino-atrial node of the right atrium. This electric impulse quickly spreads over both atria.

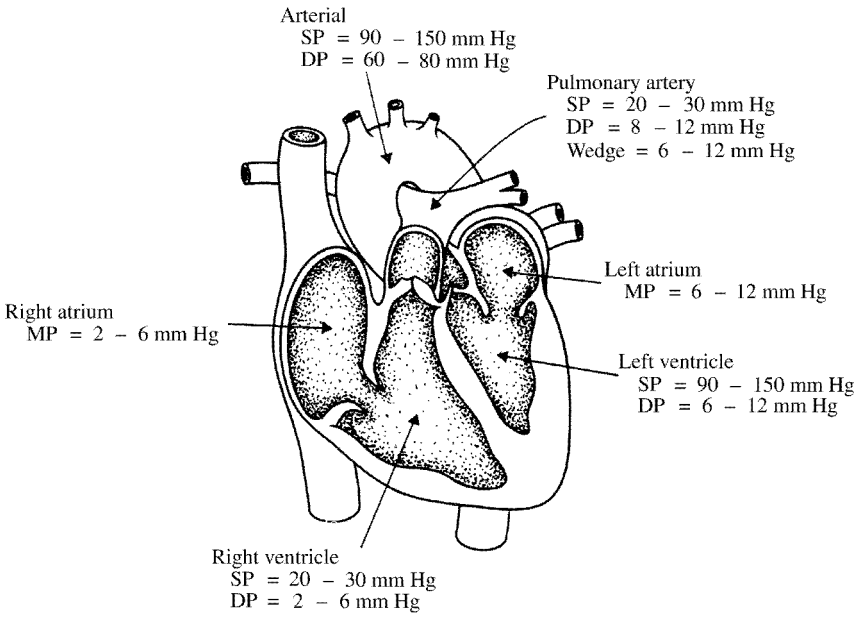


Figure 7.2 Typical values of circulatory pressures SP is the systolic pressure, DP is the diastolic pressure, and MP is the mean pressure. The wedge pressure is defined in Section 7.13.

At the junction of the atria and ventricles, the electric impulse is conducted after a short delay at the atrioventricular node. Conduction quickly spreads over the interior of both ventricles by means of a specialized conduction system, the His bundle, and the Purkinje system. Conduction then propagates throughout both ventricles. This impulse causes mechanical contraction of both ventricles. Mechanical contraction of the ventricular muscle generates ventricular pressures that force blood through the pulmonary and aortic valves into the pulmonary circulation and the systemic circulation, causing pressures in each. Section 7.9 describes the correlation of the four heart sounds with the electric and mechanical events of the cardiac cycle. Briefly, the heart sounds are associated with the movement of blood during the cardiac cycle. Murmurs are vibrations caused by the turbulence in the blood moving rapidly through the heart.

7.1 DIRECT MEASUREMENTS

Blood-pressure sensor systems can be divided into two general categories according to the location of the sensor element. The most common clinical method for directly measuring pressure is to couple the vascular pressure to an external sensor element via a liquid-filled catheter. In the second general

category, the liquid coupling is eliminated by incorporating the sensor into the tip of a catheter that is placed in the vascular system. This device is known as an *intravascular pressure sensor*.

A number of different kinds of sensor elements may be used; they include strain gage, linear-variable differential transformer, variable inductance, variable capacitance, optoelectronic, piezoelectric, and semiconductor devices (Isik, 2006). Cobbold (1974) compares the significant electric and mechanical properties of commercial pressure sensors. This section describes the principles of operation of an extravascular and an intravascular system. For a description of other sensors, see Chapter 2.

EXTRAVASCULAR SENSORS

The extravascular sensor system is made up of a catheter connected to a three-way stopcock and then to the pressure sensor (Figure 7.3). The catheter–sensor system, which is filled with a saline–heparin solution, must be flushed with the solution every few minutes to prevent blood from clotting at the tip.

The physician inserts the catheter either by means of a surgical cut-down, which exposes the artery or vein, or by means of percutaneous insertion, which

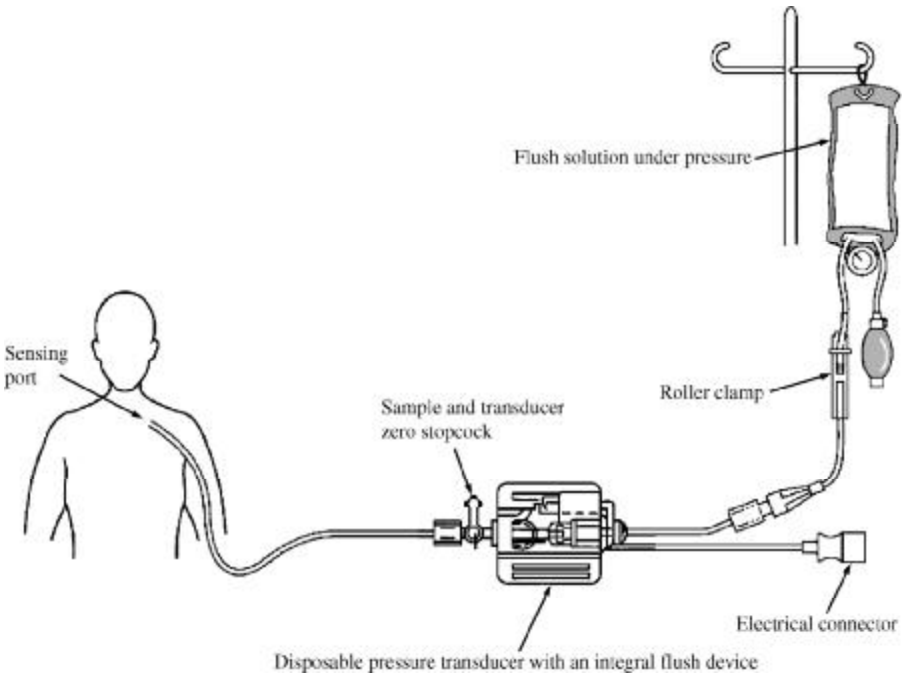


Figure 7.3 Extravascular pressure-sensor system A catheter couples a flush solution (heparinized saline) through a disposable pressure sensor with an integral flush device to the sensing port. The three-way stopcock is used to take blood samples and zero the pressure sensor.

involves the use of a special needle or guide-wire technique. Blood pressure is transmitted via the catheter liquid column to the sensor and, finally, to the diaphragm, which is deflected. Figure 2.2(a) shows an early pressure sensor, in which the displacement of the diaphragm is transmitted to a system composed of a moving armature and an unbonded strain gage. Figure 2.5 shows a modern disposable blood-pressure sensor.

INTRAVASCULAR SENSORS

Catheter-tip sensors have the advantage that the hydraulic connection via the catheter, between the source of pressure and the sensor element, is eliminated. The frequency response of the catheter–sensor system is limited by the hydraulic properties of the system. Detection of pressures at the tip of the catheter without the use of a liquid-coupling system can thus enable the physician to obtain a high frequency response and eliminate the time delay encountered when the pressure pulse is transmitted in a catheter–sensor system.

A number of basic types of sensors are being used commercially for the detection of pressure in the catheter tip. These include various types of strain-gage systems bonded onto a flexible diaphragm at the catheter tip. Gages of this type are available in the F 5 catheter [1.67 mm outer diameter (OD)] size. In the French scale (F), used to denote the diameter of catheters, each unit is approximately equal to 0.33 mm. Smaller-sized catheters may become available as the technology improves and the problems of temperature and electric drift, fragility, and nondestructive sterilization are solved more satisfactorily. A disadvantage of the catheter-tip pressure sensor is that it is more expensive than others and may break after only a few uses, further increasing its cost per use.

The fiber-optic intravascular pressure sensor can be made in sizes comparable to those described above, but at a lower cost. The fiber-optic device measures the displacement of the diaphragm optically by the varying reflection of light from the back of the deflecting diaphragm. (Recall that Section 2.14 detailed the principles of transmission of light along a fiber bundle.) These devices are inherently safer electrically, but unfortunately they lack a convenient way to measure relative pressure without an additional lumen either connected to a second pressure sensor or vented to the atmosphere.

A fiber-optic microtip sensor for *in vivo* measurements inside the human body is shown in Figure 7.4 (a) in which one leg of a bifurcated fiber bundle is connected to a light-emitting diode (LED) source and the other to a photodetector (Hansen, 1983). The pressure-sensor tip consists of a thin metal membrane mounted at the common end of the mixed fiber bundle. External pressure causes membrane deflection, varying the coupling between the LED source and the photodetector. Figure 7.4(b) shows the output signal versus membrane deflection. Optical fibers have the property of emitting and accepting light within a cone defined by the acceptance angle θ_A, which is equal to the fiber numerical aperture, N_A (Section 2.14). The coupling between LED source and detector is a function of the overlap of the two acceptance angles

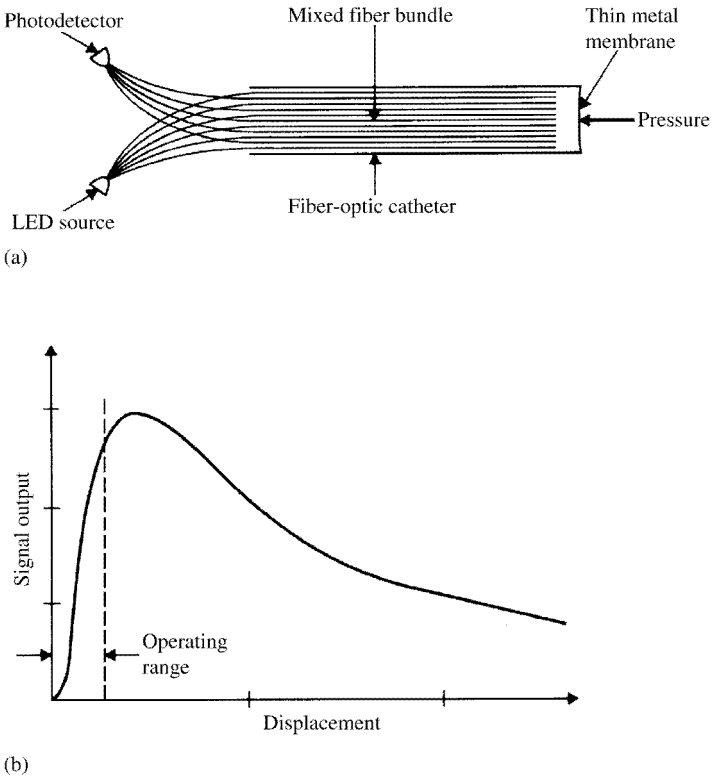


Figure 7.4 (a) Schematic diagram of an intravascular fiber-optic pressure sensor. Pressure causes deflection in a thin metal membrane that modulates the coupling between the source and detector fibers. (b) Characteristic curve for the fiber-optic pressure sensor.

on the pressure-sensor membrane. The operating portion of the curve is the left slope region where the characteristic is steepest.

Roos and Carroll (1985) describe a fiber-optic pressure sensor for use in magnetic resonance imaging (MRI) fields in which a plastic shutter assembly modulates the light transversing a channel between source and detector. Neuman (2006) described a fiber-optic pressure sensor for intracranial pressure measurements in the newborn. Figure 7.5 shows a schematic of the device, which is applied to the anterior fontanel. Pressure is applied with the sensor such that the curvature of the skin surface is flattened. When this applanation occurs, equal pressure exists on both sides of the membrane, which consists of soft tissue between the scalp surface and the dura. Monitoring of the probe pressure determines the dura pressure. Pressure bends the membrane, which moves a reflector. This varies the amount of light coupling between the source and detector fibers.

Air pressure from a pneumatic servo system controls the air pressure within the pressure sensor, which is adjusted such that diaphragm—and thus

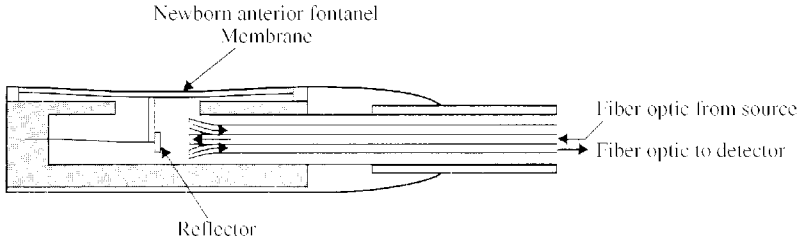


Figure 7.5 Fiber-optic pressure sensor for intracranial pressure measurements in the newborn. The sensor membrane is placed in contact with the anterior fontanel of the newborn.

the fontanel tissue—is flat, indicating that the sensor air pressure and the fontanel or intracranial pressure are equal.

Similar pressure-unloading techniques are used in thin, compliant sensors that measure interface pressures between the skin and support structures such as seat cushions (Webster, 1991).

Silicon fusion bonding is used to fabricate micro silicon pressure-sensor chips (Howe *et al*., 1990). A wedge-shaped cavity is etched in a silicon wafer to form a diaphragm. Piezoresistive strain gages are implanted and a metal connection is made in order to create a sensor for a catheter tip micropressure sensor.

DISPOSABLE PRESSURE SENSORS

Traditionally, physiological pressure sensors have been reusable devices, but most modern hospitals have adopted inexpensive, disposable pressure sensors in order to lower the risk of patient cross-contamination and reduce the amount of handling of pressure sensors by hospital personnel. Because reusable pressure sensors are subject to the abuses of reprocessing and repeated user handling, they tend to be less reliable than disposable sensors.

By micromachining silicon, a pressure diaphragm is etched and piezoresistive strain gages are diffused into the diaphragm for measuring its displacement. This process results in a small, integrated, sensitive, and relatively inexpensive pressure sensor. This silicon chip is incorporated into a disposable pressure-monitoring tubing system. The disposable pressure sensor system also contains a thick-film resistor network that is laser trimmed to remove offset voltages and set the same sensitivity for similar disposable sensors. In addition, a thick-film thermistor network is usually incorporated for temperature compensation. The resistance of the bridge elements is usually high in order to reduce self-heating, which may cause erroneous results. This results in high output impedance for the device. Thus, a high-input impedance monitor must be used with disposable pressure sensors.

Pressure sensors can monitor blood pressure in postsurgical patients as part of a closed-loop feedback system. Such a system injects controlled amounts of the drug nitroprusside to stabilize the blood pressure (Yu, 2006).

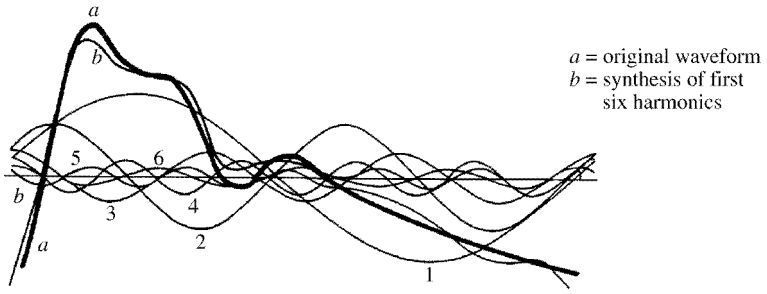
7.2 HARMONIC ANALYSIS OF BLOOD-PRESSURE WAVEFORMS

The basic sine-wave components of any complex time-varying periodic waveform can be dissected into an infinite sum of properly weighted sine and cosine functions of the proper frequency that, when added, reproduce the original complex waveform. It has been shown that researchers can apply techniques of Fourier analysis when they want to characterize the oscillatory components of the circulatory and respiratory systems, because two basic postulates for Fourier analysis—periodicity and linearity—are usually satisfied (Attinger *et al.*, 1970).

Cardiovascular physiologists and some clinicians have been employing Fourier-analysis techniques in the quantification of pressure and flow since this method was established in the 1950s. Early Fourier analysis used bandpass filters. More recent analysts have used computer techniques to obviate the need for special hardware. The advantage of the technique is that it allows for a quantitative representation of a physiological waveform; thus it is quite easy to compare corresponding harmonic components of pulses.

O'Rourke (1971) points out that the physician who turns to a standard medical textbook for assistance in interpreting the arterial pulse is likely to be confused, misled, and disappointed. He further indicates that in recent years, analysis of the frequency components of the pulse appears to have yielded more information on arterial properties than any other approach. He proposes that the arterial pulse be represented in terms of its frequency components.

The blood-pressure pulse can be divided into its fundamental component (of the same frequency as the blood-pressure wave) and its significant harmonics. Figure 7.6 shows the first six harmonic components of the blood-pressure wave



Harmonic	Amplitude (%)
1	100
2	63.2
3	29.6
4	22.2
5	14.8
6	11.8

Figure 7.6 The first six harmonics of the blood-pressure waveform The table gives relative values for amplitudes. (From T. A. Hansen, "pressure measurement in the human organism." *Acta Physiologica Scandinavica*, 1949, 19, Suppl. 68, 1–227. Used with permission.)

and the resultant sum. When we compare the original waveform and the waveform reconstructed from the Fourier components, we find that they agree quite well, indicating that the first six harmonics give a fairly good reproduction. Note that the amplitude of the sixth harmonic is approximately 12% of the fundamental. We can achieve more faithful reproduction of the original waveform by adding higher harmonic components.

7.3 DYNAMIC PROPERTIES OF PRESSURE-MEASUREMENT SYSTEMS

An understanding of the dynamic properties of a pressure-measurement system is important if we wish to preserve the dynamic accuracy of the measured pressure. Errors in measurement of dynamic pressure can have serious consequences in the clinical situation. For instance, an underdamped system can lead to overestimation of pressure gradients across stenotic (narrowed) heart valves. The liquid-filled catheter sensor is a hydraulic system that can be represented by either distributed- or lumped-parameter models. Distributed-parameter models are described in the literature (Fry, 1960), which gives an accurate description of the dynamic behavior of the catheter-sensor system. However, distributed-parameter models are not normally employed, because the single-degree-of-freedom (lumped-parameter) model is easier to work with, and the accuracy of the results obtained by using these models is acceptable for the clinical situation.

ANALOGOUS ELECTRIC SYSTEMS

The modeling approach taken here develops a lumped-parameter model for the catheter and sensor separately and shows how, with appropriate approximations, it reduces to the lumped-parameter model for a second-order system. Figure 7.7 shows the physical model of a catheter–sensor system. An increase in pressure at the input of the catheter causes a flow of liquid to the right from the catheter tip, through the catheter, and into the sensor. This liquid shift causes a deflection of the sensor diaphragm, which is sensed by an electromechanical system. The subsequent electric signal is then amplified.

A liquid catheter has inertial, frictional, and elastic properties represented by inertance, resistance, and compliance, respectively. Similarly, the sensor has these same properties, in addition to the compliance of the diaphragm. Figure 7.7(b) shows an electric analog of the pressure-measuring system, wherein the analogous elements for hydraulic inertance, resistance, and compliance are electric inductance, resistance, and capacitance, respectively.

The analogous circuit in Figure 7.7(b) can be simplified to that shown in Figure 7.8(a). The compliance of the sensor diaphragm is much larger than that of the liquid-filled catheter or sensor cavity, provided that the saline solution is bubble-free and the catheter material is relatively noncompliant. The resistance and inertance of the liquid in the sensor can be neglected compared to

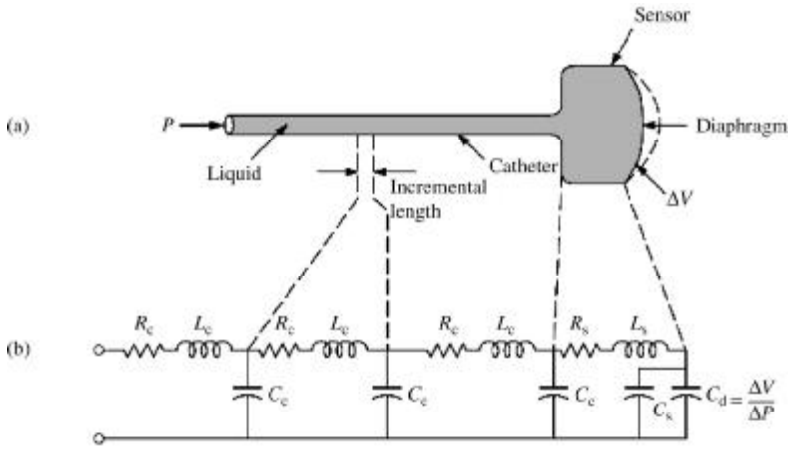


Figure 7.7 (a) Physical model of a catheter–sensor system. (b) Analogous electric system for this catheter–sensor system. Each segment of the catheter has its own resistance R_c, inertance L_c, and compliance C_c. In addition, the sensor has resistance R_s, inertance, L_s, and compliance C_s. The compliance of the diaphragm is C_d.

those of the liquid in the catheter. Let us now derive equations relating the resistance and inductance to the properties of the system.

The liquid resistance R_c of the catheter is due to friction between shearing molecules flowing through the catheter. It can be represented by the equation

$$R_c = \frac{\Delta P}{F} \left(\text{Pa}\cdot\text{s/m}^3\right) \tag{7.1}$$

or

$$R_c = \frac{\Delta P}{\overline{u}A}$$

where

p = pressure difference across the segment in Pa (pascal = N/m^2)
F = flow rate, m^3/s
$\overline{u}$ = average velocity, m/s
A = cross-section area, m^2

Poiseuille's equation enables us to calculate R_c when we are given the values of catheter length L, in meters; radius r, in meters; and liquid viscosity η, in pascal-seconds. The equation applies for laminar or Poiseuille flow. It is

$$R_c = \frac{8\eta L}{\pi r^4} \tag{7.2}$$

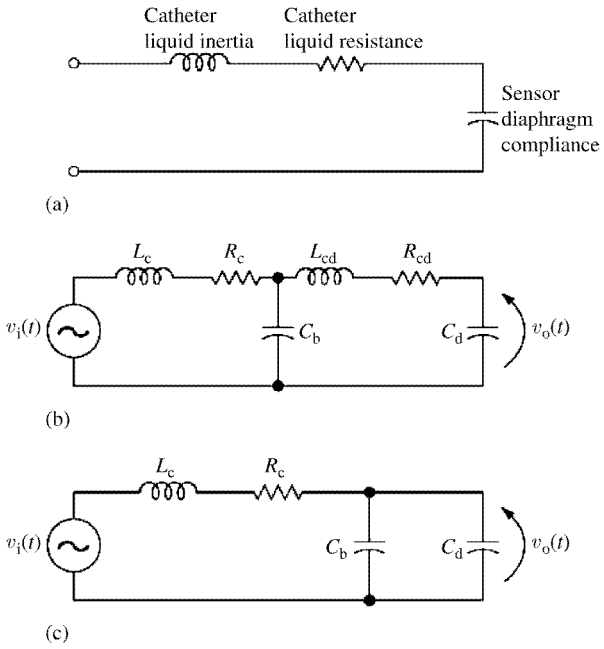


Figure 7.8 (a) Simplified analogous circuit. Compliance of the sensor diaphragm is larger than compliance of catheter or sensor cavity for a bubble-free, noncompliant catheter. The resistance and inertance of the catheter are larger than those of the sensor, because the catheter has longer length and smaller diameter. (b) Analogous circuit for catheter–sensor system with a bubble in the catheter. Catheter properties proximal to the bubble are inertance L_c and resistance R_c. Catheter properties distal to the bubble are L_{cd} and R_{tf}. Compliance of the diaphragm is C_d; compliance of the bubble is C_b. (c) Simplified analogous circuit for catheter–sensor system with a bubble in the catheter, assuming that L_{cd} and R_{cd} are negligible with respect to R_c and L_c.

The liquid inertance L_c of the catheter is due primarily to the mass of the liquid. It can be represented by the equation

$$L_c = \frac{\Delta P}{dF/dt}\left(\text{Pa}\cdot\text{s}^2/\text{m}^3\right) \tag{7.3}$$

or

$$L_c = \frac{\Delta P}{aA}$$

where a = acceleration, m/s^2.

This equation reduces further to

$$L_c = \frac{m}{A^2}$$

or

$$L_c = \frac{\rho L}{\pi r^2} \tag{7.4}$$

where m = mass of liquid (kg) and ρ = density of liquid (kg/m^3).

Equations (7.2) and (7.4) show that we can neglect the resistive and inertial components of the sensor with respect to those of the liquid catheter. The reason for this is that the liquid-filled catheter is longer than the cavity of the sensor and of smaller diameter. Geddes (1970) develops a more refined model of fluid inertance—one based on kinetic-energy considerations—in which he considers that the effective mass is four-thirds times that of the fluid in the catheter.

The compliance C_d of the sensor diaphragm is given by the equation

$$C_d = \frac{\Delta V}{\Delta P} = \frac{1}{E_d}$$

where E_d is the volume modulus of elasticity of the sensor diaphragm.

We can find the relationship between the input voltage v_i, analogous to applied pressure, and the output voltage v_o, analogous to pressure at the diaphragm, by using Kirchhoff's voltage law. Thus,

$$v_i(t) = \frac{L_c C_d d^2 v_o(t)}{dt^2} + \frac{R_c C_d d v_o(t)}{dt} + v_o(t) \tag{7.5}$$

Using the general form of a second-order system equation derived in Section 1.10, we can show that the natural undamped frequency ω_n is $1/(L_c C_d)^{1/2}$ and that the damping ratio ζ is $(R_c/2)(C_d/L_c)^{1/2}$. For the hydraulic system under study, by substituting (7.2) and (7.4) into the expressions for ω_n and ζ, we can show that

$$f_n = \frac{r}{2}\left(\frac{1}{\pi\rho L}\frac{\Delta P}{\Delta V}\right)^{1/2} \tag{7.6}$$

and

$$\zeta = \frac{4\eta}{r^3}\left(\frac{L(\Delta V/\Delta P)}{\pi\rho}\right)^{1/2} \tag{7.7}$$

Table 7.1 lists a number of useful relationships and pertinent constants.

We can study the transient response and the frequency response of the catheter–sensor system by means of the analogous electric circuit. In addition, we can study the effects of changes in the hydraulic system by adding appropriate elements to the circuit. For example, an air bubble in the liquid makes the system more compliant. Thus its effect on the system is the same as

Table 7.1 Mechanical Characteristics of Fluids

Parameter	Substance	Temperature	Value
η	Water	20 °C	0.001 Pa·s
η	Water	37 °C	0.0007 Pa·s
η	Air	20 °C	0.000018 Pa·s
ρ	Air	20 °C	1.21 kg/m^3
$\Delta V/\Delta P$	Water	20 °C	0.53×10^{-15}m^5/N per ml volume
η	Blood	All	$\cong 4 \times \eta$ for water

that caused by connecting an additional capacitor in parallel to that representing the diaphragm compliance. Example 7.1 illustrates how the analogous circuit is used.

EXAMPLE 7.1 A 5 mm-long air bubble has formed in the rigid-walled catheter connected to a Statham P23Dd sensor. The catheter is 1 m long, 6 French diameter, and filled with water at 20 °C. (The isothermal compression of air $\Delta V/\Delta P$ is 1 ml/cm of water pressure per liter of volume.) Plot the frequency-response curve of the system with and without the bubble. (Internal radius of the catheter is 0.46 mm; volume modulus of elasticity of the diaphragm is 0.49×10^{15} N/m^5.)

ANSWER The analogous circuit for the hydraulic system with and without the bubble is shown in Figure 7.8(b) and (c). We can calculate the values of the natural frequency f_n and the damping ratio ζ without the bubble by using (7.6) and (7.7). That is,

$$f_n = \frac{r}{2}\left(\frac{1}{\pi L}\frac{\Delta P}{\rho \Delta V}\right)^{1/2}$$

$$= \frac{0.046 \times 10^{-2}}{2}\left(\frac{1}{\pi(1)}\frac{0.49 \times 10^{15}}{1 \times 10^{3}}\right)^{1/2} = 91\,\text{Hz}$$

$$\zeta = \frac{4\eta}{r^3}\left(\frac{L}{\pi\rho}\frac{\Delta V}{\Delta P}\right)^{1/2}$$

$$= \frac{4(0.001)}{(0.046 \times 10^{-2})^3}\left(\frac{1}{\pi}\frac{1}{(1 \times 10^{3})(0.49 \times 10^{15})}\right)^{1/2} = 0.033$$

The frequency response for the catheter–sensor system is shown in Figure 7.9.

The next step is to calculate the new values of ζ and f_n, for the case in which a bubble is present. Because the two capacitors are in parallel, the total capacitance for the circuit is equal to the sum of these two. That is,

$$C_t = C_d + C_b \tag{7.8}$$

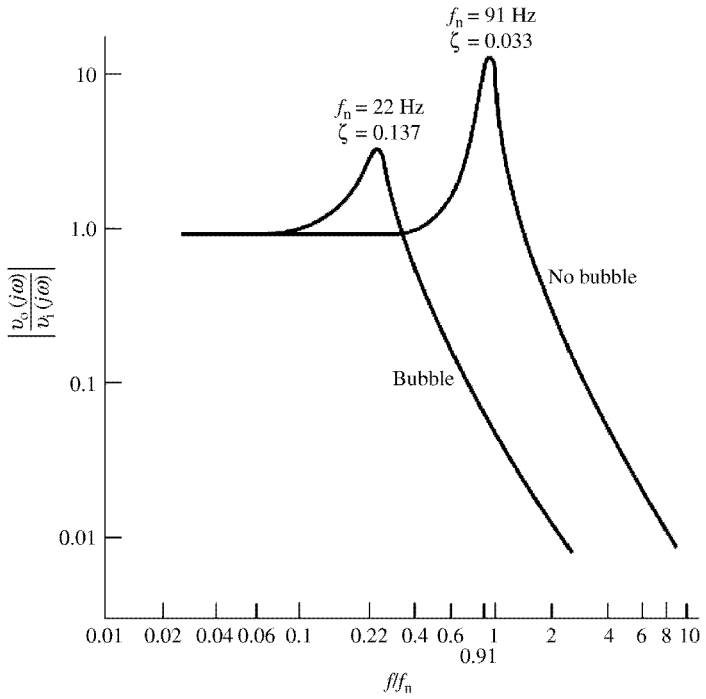


Figure 7.9 Frequency-response curves for catheter–sensor system with and without bubbles. Natural frequency decreases from 91 to 22 Hz and damping ratio increases from 0.033 to 0.137 with the bubble present.

or

$$C_t = \frac{\Delta V}{\Delta P_d} + \frac{\Delta V}{\Delta P_b}$$

The value of $\Delta V/\Delta P_d = 1/E_d = 2.04 \times 10^{-15}\ m^5/N$. The volume of the bubble is

$$\pi r^2 l = 3.33 \times 10^{-9}\ m^3 = 3.33 \times 10^{-6}\ \text{liter}$$

One centimeter of water pressure is 98.5 N/m^2. Thus

$$\Delta V/\Delta P_b = \frac{3.33 \times 10^{-9} 1 \times (1 \times 10^{-3} m^3/1)}{98.5\ N/m^2} = 3.38 \times 10^{-14}\ m^5/N$$

Consequently, $C_t = 3.38 \times 10^{-14}\ m^5/N$. We can find the new values for f_n and ζ by referring to (7.6) and (7.7) and assuming that the only parameter that changes is the value of $\Delta V/\Delta P$. Thus

$$f_{n,\text{bubble}} = f_{n,\text{no bubble}} \left(\frac{\Delta P \Delta V_{\text{total}}}{\Delta P/\Delta V_{\text{no bubble}}} \right)^{1/2}$$

or

$$f_{\text{n, bubble}} = 92\left(\frac{2.04 \times 10^{-15}}{3.38 \times 10^{-14}}\right)^{1/2} = 22\,\text{Hz}$$

and

$$\zeta_{\text{bubble}} = \zeta_{\text{no bubble}}\left(\frac{\Delta V/\Delta P_{\text{total}}}{\Delta V/\Delta P_{\text{no bubble}}}\right)^{1/2} = 0.137$$

The frequency response for the system with the bubble present is shown in Figure 7.9. Note that the bubble lowers f_n and increases ζ. This lowering of f_n may cause distortion problems with the higher harmonics of the blood-pressure waveform.

EXAMPLE 7.2 By changing only the radius of the catheter, redesign the (no-bubble) catheter of Figure 7.9 to achieve the damping ratio $\zeta = 1$. Calculate the resulting natural frequency f_n.

ANSWER From Eq. (7.7), $r^3/r_0^3 = \zeta/\zeta_0$.

$$r^3 = (1.0/0.033)(0.46\,\text{mm})^3 = 0.0032 \quad r = 0.147\,\text{mm}$$

From Eq. (7.6), $f_n/f_{n0} = r/r_0$,

$$f_n = 91(0.147)/(0.5) = 29\,\text{Hz}$$

EXAMPLE 7.3 Water is not a perfect liquid because it has a finite volume modulus of elasticity. Therefore, a theoretical upper limit of high-frequency response exists for a water-filled sensor–catheter system. Find the maximal f_n for a sensor that has a 0.50 ml liquid chamber and that is connected to the pressure source by means of a #20 ($r = 0.29\,\text{mm}$) 50 mm-long steel needle.

ANSWER A steel needle is used for the catheter, so the volume modulus of elasticity of the catheter is assumed to be zero. We do not consider the volume modulus of elasticity of the diaphragm because we want the theoretical upper limit of frequency response. Thus $\Delta V/\Delta P$ for this example is that of water, $0.53 \times 10^{-15}\,\text{m}^5/\text{N}$ per milliliter volume. The total volume of the water is equal to the volume of the liquid chamber of the sensor plus the volume of the cylindrical needle.

$$V_t = 0.5 + \pi(0.029)^2(5)\,\text{ml} = 0.513\,\text{ml}$$

$$\left(\frac{\Delta V}{\Delta P}\right)_{\text{water}} = (0.53 \times 10^{-15})(0.513) = 0.272 \times 10^{-15}\,\text{m}^5/\text{N}$$

$$f_n = \frac{0.029 \times 10^{-2}}{2}\left(\frac{1}{\pi(0.05)(1000)(0.27 \times 10^{-15})}\right)^{1/2} = 700\,\text{Hz}$$

7.4 MEASUREMENT OF SYSTEM RESPONSE

The response characteristics of a catheter–sensor system can be determined by two methods. The simplest and most straightforward technique involves measuring the transient step response for the system. A potentially more accurate method—but a more complicated one because it requires special equipment—involves measuring the frequency response of the system.

TRANSIENT STEP RESPONSE

The basis of the transient-response method is to apply a sudden step input to the pressure catheter and record the resultant damped oscillations of the system. This is also called the *pop* technique, for reasons that will become evident in the following discussion. The transient response can be found by the method shown in Figure 7.10.

The catheter, or needle, is sealed in a tube by a screw adaptor that compresses a rubber washer against the insert. Flushing of the system is accomplished by passing excess liquid out of the three-way stopcock. The test is performed by securing a rubber membrane over the tube by means of an O-ring. Surgical-glove material is an excellent choice for the membrane. The technician pressurizes the system by squeezing the sphygmomanometer bulb, punctures the balloon with a burning match or hot soldering iron, and observes the response. The response should be observed on a recorder running at a speed that makes it possible to distinguish the individual oscillations. However, if the frequency bandwidth of the recorder is inadequate, the technician can use a storage oscilloscope or data acquisition system.

Figure 7.11 shows an example of the transient response. In this case the response represents a second-order system. The technician can measure the

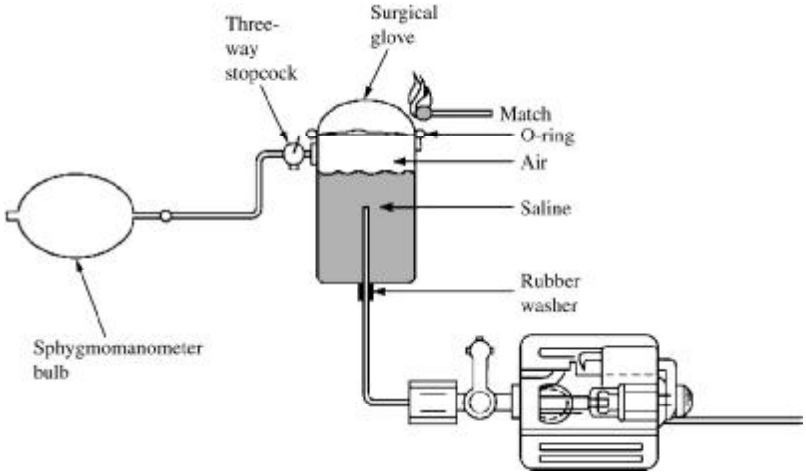


Figure 7.10 Transient-response technique for testing a pressure-sensor–catheter-sensor system.

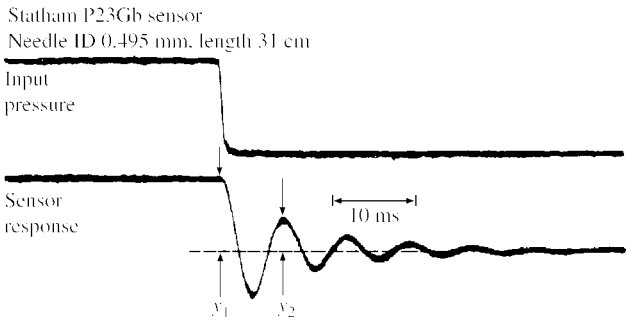


Figure 7.11 Pressure-sensor transient response Negative-step input pressure is recorded on the top channel; the bottom channel is sensor response for a Statham P23Gb sensor connected to a 31 cm needle (0.495 mm ID). (From I. T. Gabe, ''Pressure measurement in experimental physiology,'' In D. H. Bergel (ed.), I, Vol. I, New York: Academic Press, 1972.)

amplitude ratio of successive positive peaks and determine the logarithmic decrement Λ. Equation (1.38) yields the damping ratio ζ. The observer can measure T, the time between successive positive peaks, and determine the undamped natural frequency from $\omega_n = 2\pi/\left[T(1-\zeta^2)^{1/2}\right]$.

SINUSOIDAL FREQUENCY RESPONSE

As we noted before, the sinusoidal frequency-response method is more complex because it requires more specialized equipment. Figure 7.12 is a

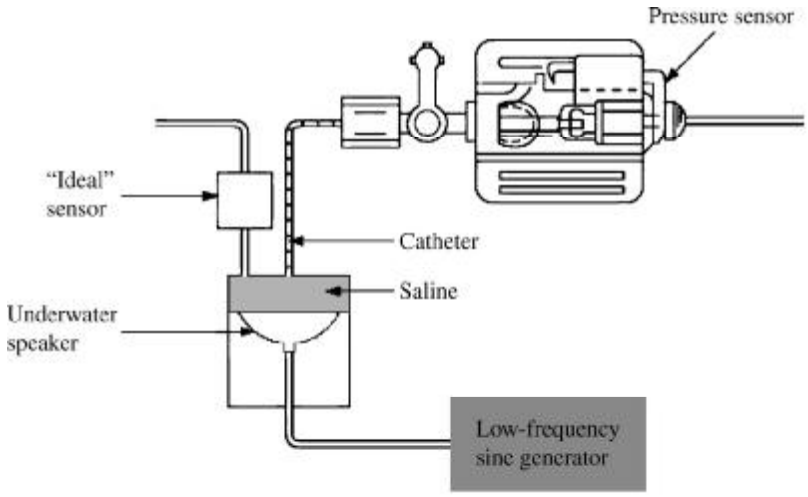


Figure 7.12 A sinusoidal pressure-generator test system A low-frequency sine generator drives an underwater-speaker system that is coupled to the catheter of the pressure sensor under test. An ''ideal'' pressure sensor, with a frequency response from 0 to 100 Hz, is connected directly to the test chamber housing and monitors input pressure.

schematic diagram of a sinusoidal pressure-generator test system. A pump produces sinusoidal pressures that are normally monitored at the pressure source by a pressure sensor with known characteristics. This is used because the amplitudes of the source-pressure waveforms are not normally constant for all frequencies. The source pressure is coupled to the catheter sensor under test by means of bubble-free saline. The air is removed by boiling the liquid.

We can find an accurate model for the catheter-sensor system by determining the amplitude and phase of the output as a function of frequency without the constraint of the second-order system model required in the transient-response case. In some cases, resonance at more than one frequency may be present.

7.5 EFFECTS OF SYSTEM PARAMETERS ON RESPONSE

We have shown in our discussion of the model of the catheter–sensor system that the values of the damping ratio ζ and natural frequency ω_n are functions of the various system parameters. This section reports on experimental verification of these theoretical derivations (Shapiro and Krovetz, 1970). By using step transient-response and sinusoidal pressure-generation techniques similar to those described above, these investigators determined the effects on the performance of the catheter–sensor system of de-aerating water and of using various catheter materials and connectors. They found that even minute air bubbles, which increase the compliance of the catheter manometer system, drastically decreased the damped natural frequency $\omega_d = 2\pi/T$. For a PE-190 catheter of lengths 10 to 100 cm, the damped natural frequency decreased by approximately 50% to 60% for an unboiled-water case compared to a boiled-water case. Length of the catheter was shown to be inversely related to the damped natural frequency for Teflon and polyethylene catheters for the diameters tested (0.58 to 2.69 mm). The theoretical linear relationship between the damped natural frequency and $1/(\text{catheter length})^{1/2}$ seemed to hold within experimental errors. A linear relationship was found to exist between the inner diameter of the catheter and the damped natural frequency for both polyethylene and Teflon catheters, as predicted by the model equations.

In comparing the effect of catheter material on frequency response, Shapiro and Krovetz (1970) found that because Teflon is slightly stiffer than polyethylene, it has a slightly higher frequency response at any given length. As expected, the increased compliance of silicone rubber tubing caused a marked decrease in frequency response. The authors concluded that silicone rubber is a poor material for determining parameters other than mean pressure.

They examined the effect of connectors on the system response by inserting—in series with the catheter—various connecting needles that added little to the overall length of the system. They found that the damped natural frequency was linearly related to the needle bore for needles of the same length. The connector serves as a simple series hydraulic damper that

decreases the frequency response. They suggested that the fewest possible number of connectors be used and that all connectors be tight fitting and have a water seal. In further tests, they found that coils and bends in the catheter cause changes in the resonant frequency. However, the magnitude of these changes was insignificant compared with changes caused by factors that affected compliance.

7.6 BANDWIDTH REQUIREMENTS FOR MEASURING BLOOD PRESSURE

When we know the representative harmonic components of the blood-pressure waveform—or, for that matter, any periodic waveform—we can specify the bandwidth requirements for the instrumentation system. As with all biomedical measurements, bandwidth requirements are a function of the investigation.

For example, if the mean blood pressure is the only parameter of interest, it is of little value to try to achieve a wide bandwidth system. It is generally accepted that harmonics of the blood-pressure waveform higher than the tenth may be ignored. As an example, the bandwidth requirements for a heart rate of 120 bpm (or 2 Hz) would be 20 Hz.

For a perfect reproduction of the original waveform, there should be no distortion in the amplitude or phase characteristics. The waveshape can be preserved, however, even if the phase characteristics are not ideal. This is the case if the relative amplitudes of the frequency components are preserved but their phases are displaced in proportion to their frequency. Then the synthesized waveform gives the original waveshape, except that it is delayed in time, depending on the phase shift.

Measurements of the derivative of the pressure signal increase the bandwidth requirements, because the differentiation of a sinusoidal harmonic increases the amplitude of that component by a factor proportional to its frequency. As with the original blood-pressure waveform, a Fourier analysis of the derivative signal can estimate the bandwidth requirements for the derivative of the blood pressure. The amplitude-versus-frequency characteristics of any catheter–manometer system used for the measurement of ventricular pressures that are subsequently differentiated must remain flat to within 5%, up to the 20th harmonic (Gersh *et al.*, 1971).

7.7 TYPICAL PRESSURE-WAVEFORM DISTORTION

Accurate measurements of blood pressure are important in both clinical and physiological research. This section gives examples of typical types of distortion of blood-pressure waveform that are due to an inadequate frequency response of the catheter-sensor system.

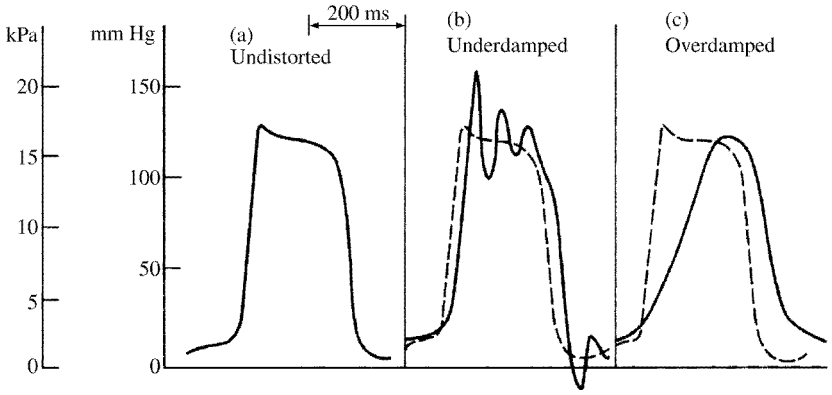


Figure 7.13 Pressure-waveform distortion (a) Recording of an undistorted left-ventricular pressure waveform via a pressure sensor with bandwidth dc to 100 Hz. (b) Underdamped response, where peak value is increased. A time delay is also evident in this recording. (c) Overdamped response that shows a significant time delay and an attenuated amplitude response.

There may be serious consequences when an underdamped system leads to overestimation of the pressure gradients across a stenotic (narrowed) heart valve. Figure 7.13 shows examples of distortion of pressure waveform. The actual blood-pressure waveform [Figure 7.13(a)] was recorded with a high-quality pressure sensor with a bandwidth from dc to 100 Hz. Note that in the underdamped case, the amplitude of the higher-frequency components of the pressure wave are amplified, whereas for the overdamped case these higher-frequency components are attenuated. The actual peak pressure [Figure 7.13(a)] is approximately 130 mm Hg (17.3 kPa). The underdamped response [Figure 7.13(b)] has a peak pressure of about 165 mm Hg (22 kPa), which may lead to a serious clinical error if this peak pressure is used to assess the severity of aortic-valve stenosis. The minimal pressure is in error, too; it is $-15\,\text{mm Hg}(-2\,\text{kPa})$ and the actual value is 5 mm Hg (0.7 kPa). There is also a time delay of approximately 30 ms in the underdamped case.

The overdamped case [Figure 7.13(c)] shows a significant time delay of approximately 150 ms and an attenuated amplitude of 120 mm Hg (16 kPa); the actual value is 130 mm Hg (17.3 kPa). This type of response can occur in the presence of a large air bubble or a blood clot at the tip of the catheter.

An underdamped catheter–sensor system can be transformed to an overdamped system by pinching the catheter. This procedure increases the damping ratio ζ and has little effect on the natural frequency. (See Problem 7.6.)

Another example of distortion in blood-pressure measurements is known as *catheter whip*. Figure 7.14 shows these low-frequency oscillations that appear in the blood-pressure recording. This may occur when an aortic ventricular catheter, in a region of high pulsatile flow, is bent and whipped about by the accelerating blood. This type of distortion can be minimized by

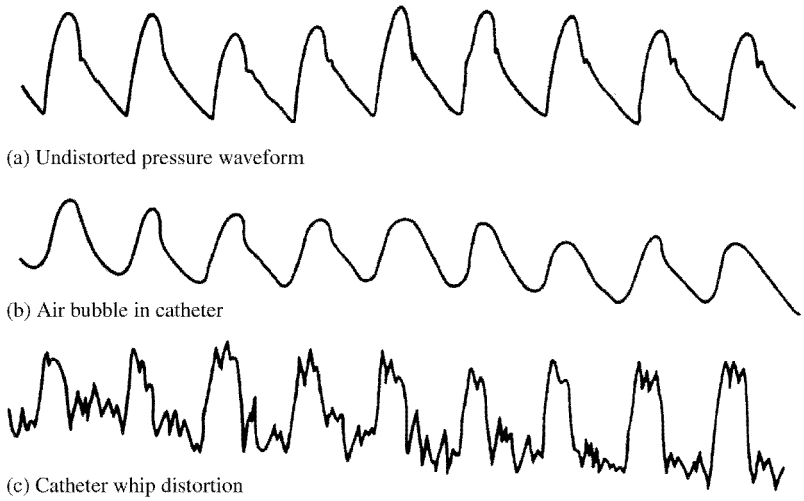


Figure 7.14 Distortion during the recording of arterial pressure The bottom trace is the response when the pressure catheter is bent and whipped by accelerating blood in regions of high pulsatile flow.

the use of stiff catheters or by careful placement of catheters in regions of low flow velocity.

7.8 SYSTEMS FOR MEASURING VENOUS PRESSURE

Measurements of venous pressure are an important aid to the physician for determining the function of the capillary bed and the right side of the heart. The pressure in the small veins is lower than the capillary pressure and reflects the value of the capillary pressure. The intrathoracic venous pressure determines the diastolic filling pressure of the right ventricle (Rushmer, 1970). The central venous pressure is measured in a central vein or in the right atrium. It fluctuates above and below atmospheric pressure as the subject breathes, whereas the extrathoracic venous pressure is 2 to 5 cm H_2O (0.2 to 0.5 kPa) above atmospheric. The reference level for venous pressure is at the right atrium.

Central venous pressure is an important indicator of myocardial performance. It is normally monitored on surgical and medical patients to assess proper therapy in cases of heart dysfunction, shock, hypovolemic or hypervolemic states, or circulatory failure. It is used as a guide to determine the amount of liquid a patient should receive.

Physicians usually measure steady state or mean venous pressure by making a percutaneous venous puncture with a large-bore needle, inserting a catheter through the needle into the vein, and advancing it to the desired position. The needle is then removed. A plastic tube is attached to the

intravenous catheter by means of a stopcock, which enables clinicians to administer drugs or fluids as necessary. Continuous dynamic measurements of venous pressure can be made by connecting to the venous catheter a high-sensitivity pressure sensor with a lower dynamic range than that necessary for arterial measurements.

Problems in maintaining a steady baseline occur when the patient changes position. Errors may arise in the measurements if the catheter is misplaced or if it becomes blocked by a clot or is impacted against a vein wall. It is normal practice to accept venous-pressure values only when respiratory swings are evident. Normal central venous pressures range widely from 0 to 12 cm H_2O (0 to 1.2 kPa), with a mean pressure of 5 cm H_2O (0.5 kPa).

Esophageal manometry uses a similar low-pressure catheter system (Velanovich, 2006). A hydraulic capillary infusion system infuses 0.6 ml/min to prevent sealing of the catheter orifice in the esophagus.

7.9 HEART SOUNDS

The auscultation of the heart gives the clinician valuable information about the functional integrity of the heart. More information becomes available when clinicians compare the temporal relationships between the heart sounds and the mechanical and electric events of the cardiac cycle. This latter approach is known as *phonocardiography*.

There is a wide diversity of opinion concerning the theories that attempt to explain the origin of heart sounds and murmurs. More than 40 different mechanisms have been proposed to explain the first heart sound. A basic definition shows the difference between heart sounds and murmurs (Rushmer, 1970). Heart sounds are vibrations or sounds due to the acceleration or deceleration of blood, whereas murmurs are vibrations or sounds due to blood turbulence.

MECHANISM AND ORIGIN

Figure 7.15 shows how the four heart sounds are related to the electric and mechanical events of the cardiac cycle. The first heart sound is associated with the movement of blood during ventricular systole (Rushmer, 1970). As the ventricles contract, blood shifts toward the atria, closing the atrioventricular valves with a consequential oscillation of blood. The first heart sound further originates from oscillations of blood between the descending root of the aorta and ventricle and from vibrations due to blood turbulence at the aortic and pulmonary valves. Splitting of the first heart sound is defined as an asynchronous closure of the tricuspid and mitral valves. The second heart sound is a low-frequency vibration associated with the deceleration and reversal of flow in the aorta and pulmonary artery and with the closure of the semilunar valves (the valves situated between the ventricles and the aorta or the pulmonary trunk).

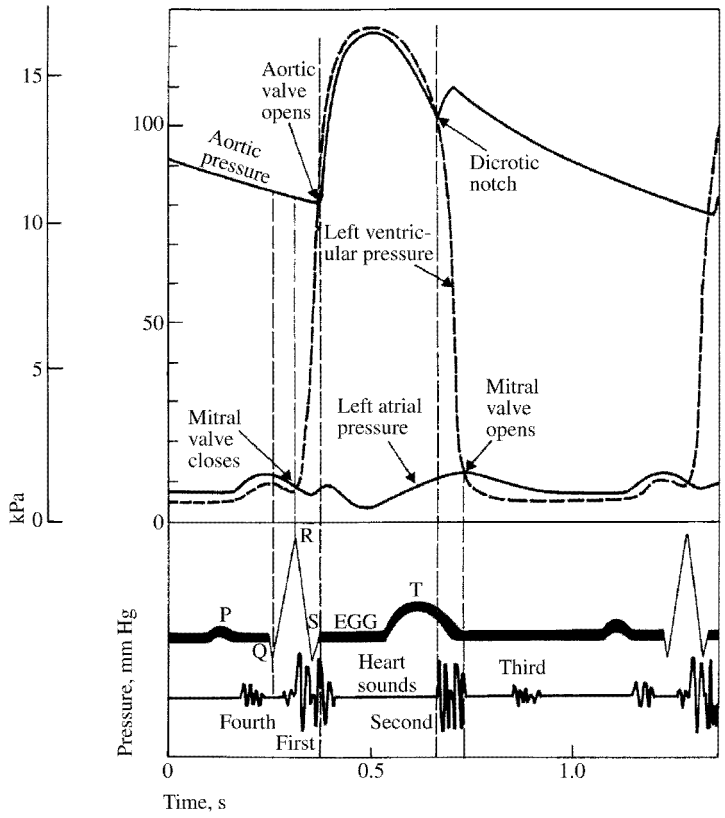


Figure 7.15 Correlation of the four heart sounds with electric and mechanical events of the cardiac cycle.

This second heart sound is coincident with the completion of the T wave of the ECG.

The third heart sound is attributed to the sudden termination of the rapid-filling phase of the ventricles from the atria and the associated vibration of the ventricular muscle walls, which are relaxed. This low-amplitude, low-frequency vibration is audible in children and in some adults.

The fourth or atrial heart sound—which is not audible but can be recorded by the phonocardiogram—occurs when the atria contract and propel blood into the ventricles.

The sources of most murmurs, developed by turbulence in rapidly moving blood, are known. Murmurs during the early systolic phase are common in children, and they are normally heard in nearly all adults after exercise. Abnormal murmurs may be caused by stenoses and insufficiencies (leaks) at the aortic, pulmonary, and mitral valves. They are detected by noting the time of their occurrence in the cardiac cycle and their location at the time of measurement.

AUSCULTATION TECHNIQUES

Heart sounds travel through the body from the heart and major blood vessels to the body surface. Because of the acoustical properties of the transmission path, sound waves are attenuated and not reflected. The largest attenuation of the wavelike motion occurs in the most compressible tissues, such as the lungs and fat layers.

There are optimal recording sites for the various heart sounds, sites at which the intensity of sound is the highest because the sound is being transmitted through solid tissues or through a minimal thickness of inflated lung. There are four basic chest locations at which the intensity of sound from the four valves is maximized (Figure 7.16).

Heart sounds and murmurs have extremely small amplitudes, with frequencies from 0.1 to 2000 Hz. Two difficulties may result. At the low end of the spectrum (below about 20 Hz), the amplitude of heart sounds is below the threshold of audibility. The high-frequency end is normally quite perceptible to the human ear, because this is the region of maximal sensitivity. However, if a phonocardiogram is desired, the recording device must be carefully selected for high frequency-response characteristics. That is, a light-beam, ink-jet, or digital-array recorder would be adequate, whereas a standard pen strip-chart recorder would not.

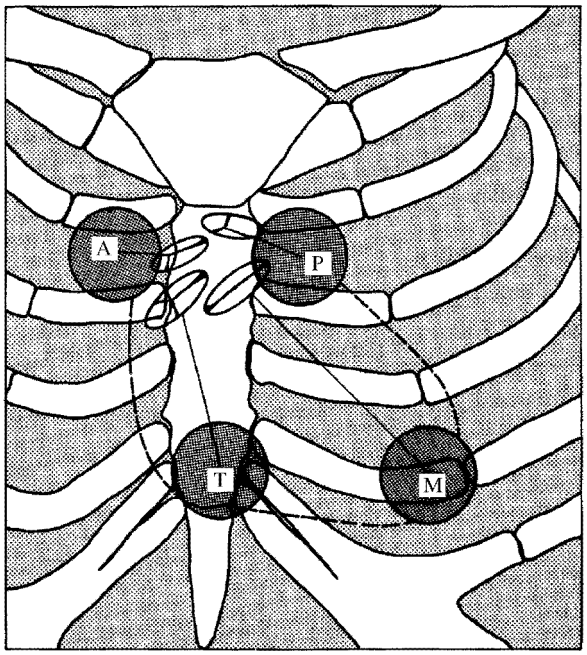


Figure 7.16 Auscultatory areas on the chest A, aortic; P, pulmonary; T, tricuspid; and M, mitral areas. (From A. C. Burton, *Physiology and Biophysics of the Circulation*, 2nd ed. Copyright © 1972 by Year Book Medical Publishers, Inc., Chicago. Used by permission.)

Because heart sounds and murmurs are of low amplitude, extraneous noises must be minimized in the vicinity of the patient. It is standard procedure to record the phonocardiogram for nonbedridden patients in a specially designed, acoustically quiet room. Artifacts from movements of the patient appear as baseline wandering.

STETHOSCOPES

Stethoscopes are used to transmit heart sounds from the chest wall to the human ear. Some variability in interpretation of the sounds stems from the user's auditory acuity and training. Moreover, the technique used to apply the stethoscope can greatly affect the sounds perceived.

Ertel *et al.* (1966a; 1966b) have investigated the acoustics of stethoscope transmission and the acoustical interactions of human ears with stethoscopes. They found that stethoscope acoustics reflected the acoustics of the human ear. Younger individuals revealed slightly better responses to a stethoscope than their elders. The mechanical stethoscope amplifies sound because of a standing-wave phenomenon that occurs at quarter wavelengths of the sound. Figure 7.17 is a typical frequency-response curve for a stethoscope; it shows that the mechanical stethoscope has an uneven frequency response, with many resonance peaks.

These investigators emphasized that the critical area of the performance of a stethoscope (the clinically significant sounds near the listener's threshold of hearing) may be totally lost if the stethoscope attenuates them as little 3 dB.

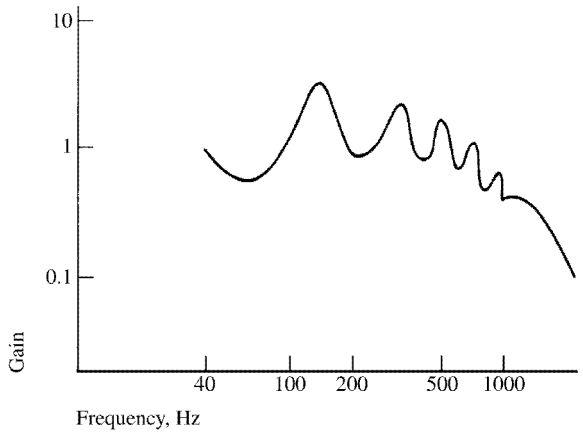


Figure 7.17 The typical frequency-response curve for a stethoscope can be found by applying a known audio-frequency signal to the bell of a stethoscope by means of a headphone-coupler arrangement. The audio output of the stethoscope earpiece was monitored by means of a coupler microphone system. (From Ertel *et al.* (1966); by permission of American Heart Association.)

A physician may miss, with one instrument, sounds that can be heard with another.

When the stethoscope chest piece is firmly applied, low frequencies are attenuated more than high frequencies. The stethoscope housing is in the shape of a bell. It makes contact with the skin, which serves as the diaphragm at the bell rim. The diaphragm becomes taut with pressure, thereby causing an attenuation of low frequencies.

Loose-fitting earpieces cause additional problems, because the leak that develops reduces the coupling between the chest wall and the ear, with a consequent decrease in the listener's perception of heart sounds and murmurs.

Stethoscopes are also useful for listening to the sounds caused by airflow obstruction or lung collapse (Loudon and Murphy, 2006).

Engineers have proposed many types of electronic stethoscopes. These devices have selectable frequency-response characteristics ranging from the ''ideal'' flat-response case and selected bandpasses to typical mechanical-stethoscope responses. Physicians, however, have not generally accepted these electronic stethoscopes, mainly because they are unfamiliar with the sounds heard with them. Their size, portability, convenience, and resemblance to the mechanical stethoscope are other important considerations.

7.10 PHONOCARDIOGRAPHY

A phonocardiogram is a recording of the heart sounds and murmurs (Vermariën, 2006). It eliminates the subjective interpretation of these sounds and also makes possible an evaluation of the heart sounds and murmurs with respect to the electric and mechanical events in the cardiac cycle. In the clinical evaluation of a patient, a number of other heart-related variables may be recorded simultaneously with the phonocardiogram. These include the ECG, carotid arterial pulse, jugular venous pulse, and apexcardiogram. The indirect carotid, jugular, and apexcardiogram pulses are recorded by using a microphone system with a frequency response from 0.1 to 100 Hz. The cardiologist evaluates the results of a phonocardiograph on the basis of changes in waveshape and in a number of timing parameters.

7.11 CARDIAC CATHETERIZATION

The cardiac-catheterization procedure is a combination of several techniques that are used to assess hemodynamic function and cardiovascular structure. Cardiac catheterization is performed in virtually all patients in whom heart surgery is contemplated. This procedure yields information that may be crucial in defining the timing, risks, and anticipated benefit for a given patient (Grossman, 1974). Catheterization procedures are performed in specialized

laboratories outfitted with x-ray equipment for visualizing heart structures and the position of various pressure catheters. In addition, measurements are made of cardiac output, blood and respiratory gases, blood-oxygen saturation, and metabolic products. The injection of radiopaque dyes into the ventricles or aorta makes it possible for the clinician to assess ventricular or aortic function. In a similar fashion, injection of radiopaque dyes into the coronary arteries makes possible a clinical evaluation of coronary-artery disease. In the following paragraphs, we shall discuss a number of specific procedures carried out in a catheter laboratory.

Clinicians can measure pressures in all four chambers of the heart and in the great vessels by positioning catheters, during fluoroscopy, in such a way that they can recognize the characteristic pressure waveforms. They measure pressures across the four valves to determine the valves' pressure gradients.

An example of a patient with aortic stenosis will help illustrate the procedure. Figure 7.18(a) shows the pressures of the stenotic patient before the operation: Note the pressures in the left ventricle and in the aorta and the systolic pressure gradient. Figure 7.18(b) reflects the situation after the operation: Note the marked decrease in the pressure gradient brought about by the insertion of a ball-valve aortic prosthesis. These pressures may be measured by using a two-lumen catheter positioned such that the valve is located between the two catheter openings. The clinician can find the various time indices that describe the injection and filling periods of the heart directly from the recordings of blood pressure in the heart.

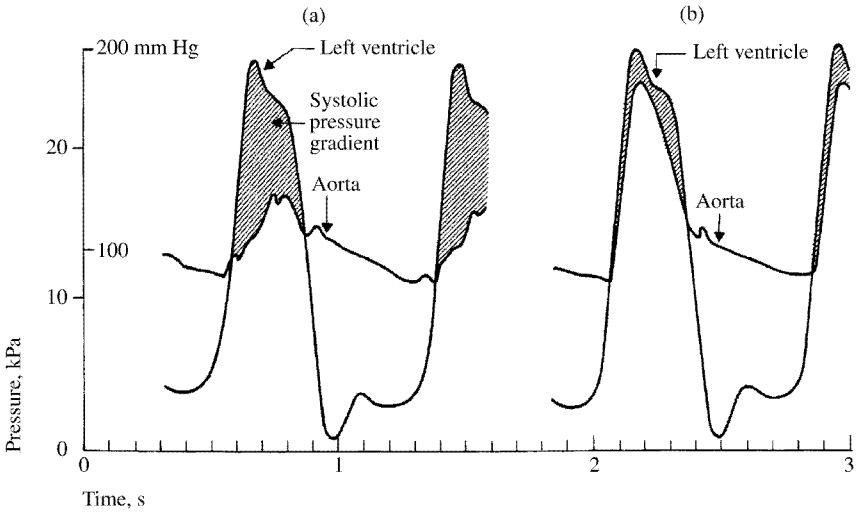


Figure 7.18 (a) Systolic pressure gradient (left ventricular aortic pressure) across a stenotic aortic valve. (b) Marked decrease in systolic pressure gradient with insertion of an aortic ball valve.

Clinicians can also use balloon-tipped, flow-directed catheters without fluoroscopy (Ganz and Swan, 1974). An inflated balloon at the catheter tip is carried by the bloodstream from the intrathoracic veins through the right atrium, ventricle, and pulmonary artery and into a small pulmonary artery—where it is wedged, blocking the local flow. The wedge pressure in this pulmonary artery reflects the mean pressure in the left atrium, because a column of stagnant blood on the right side of the heart joins the free-flowing blood beyond the capillary bed.

This catheter is also commonly used to measure cardiac output using the principle of thermodilution. Cardiac output is valuable for assessing the pumping function of the heart and can also be measured using dye dilution, the Fick method, and impedance cardiography (Sections 8.1, 8.2 and 8.7).

Blood samples can be drawn from within the various heart chambers and vessels where the catheter tip is positioned. These blood samples are important in determining the presence of shunts between the heart chambers or great vessels. For example, a shunt from the left to the right side of the heart is indicated by a higher-than-normal O_2 content in the blood in the right heart in the vicinity of the shunt. The O_2 content is normally determined by an oximeter (Section 10.3). Cardiac blood samples are also used to assess such metabolic end products as lactate, pyruvate, CO_2, and such injected substances as radioactive materials and colored dyes.

Angiographic visualization is an essential tool used to evaluate cardiac structure. Radiopaque dye is injected rapidly into a cardiac chamber or blood vessel, and the hemodynamics are viewed and recorded on x-ray film, movie film, or videotape. (In Section 12.6 we will discuss the principles of radiography and fluoroscopy.) Specially designed catheters and power injectors are used in order that a bolus of contrast material can be delivered rapidly into the appropriate vessel or heart chamber. Standard angiographic techniques are employed, where indicated, in the evaluation of the left and right ventricles (ventriculography), the coronary arteries (coronary arteriography), the pulmonary artery (pulmonary angiography) and the aorta (aortography). During heart catheterization, ectopic beats and/or cardiac fibrillation frequently occur. These are usually caused by a mechanical stimulus from the catheter or from a jet of contrast material. For this reason, clinicians must have a functional defibrillator (Section 13.2) readily available in the catheterization laboratory.

Until recently, the common procedure for directly viewing the coronary arteries was through angiography. An intravascular coronary ultrasound (IVUS), a tiny ultrasound ''camera'' on the tip of a coronary catheter, can now be threaded into the coronary arteries to provide a cross-sectional view from the inside out. These IVUS pictures allow the clinician to determine the location and characteristics of the plaque.

The percutaneous translumenal coronary angioplasty (PTCA) catheter is used to enlarge the lumen of stenotic coronary arteries, thereby improving distal flow and relieving symptoms of ischemia and signs of myocardial hypoperfusion. After initial coronary angiography is performed and the

coronary lesions are adequately visualized, a guiding catheter is introduced and passed around the aortic arch. The PTCA catheter is then placed over the guide wire and connected to a manifold (for pressure recording and injections) and to the inflation device. The guide wire is generally advanced into the coronary artery, across from and distal to the lesion to be dilated. A balloon catheter is advanced over the wire and placed across the stenosis. The pressure gradient across the stenosis is measured by using the pressure lumens on the PTCA catheter. This measurement is done to determine the severity of the stenosis. The balloon is repeatedly inflated—usually for 30 to 60 s each time—until the stenosis is fully expanded. Test injections are performed to determine whether the coronary artery flow has been improved. (Clinicians observe the distal runoff by using a radiopaque dye.)

A successful PTCA is an alternative to coronary by-pass surgery for a large proportion of patients with coronary artery disease. It avoids the morbidity associated with thoracotomy, cardiopulmonary by-pass, and general anesthesia. In addition, the hospital stay is much shorter. The patient can be discharged about a day after the procedure. Restenosis following coronary angioplasty has been shown to recur in 15% to 35% of cases. In order to help prevent restenosis a stent is commonly inserted. The stent, an expandable metal mesh tube, acts as a scaffold at the site of the blockage. It pushes against the intima of the artery to keep it open. Stenting has helped to reduce the stenosis reclosure rate. However, restenosis still occurs within a year of angioplasty or even longer because of cell regrowth. The use of drug-eluding stents, which release a drug over time directly to the artery intima most likely to reblock have been shown to significantly reduce the restenosis rate to low single digits.

Areas of a valve orifice can be calculated from basic fluid-mechanics equations (Herman *et al.*, 1974). Physicians can assess valvular stenosis by measuring the pressure gradient across the valve of interest and the flow through it.

Bernoulli's equation for frictionless flow (Burton, 1972) is

$$P_t = P + \rho g h + \frac{\rho u^2}{2} \tag{7.9}$$

where

P_t = fluid total pressure
P = local fluid static pressure
ρ = fluid density
g = acceleration of gravity (Appendix A.1)
h = height above reference level
u = fluid velocity

We first assume frictionless flow for the model shown in Figure 7.19 and equate total pressures at locations 1 and 2. We assume that the difference in

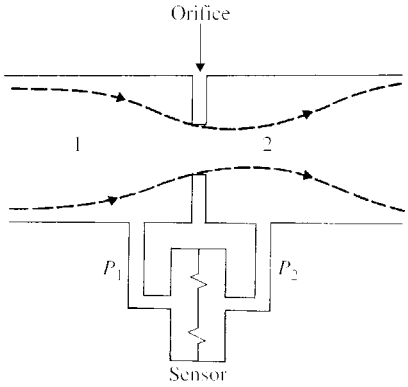


Figure 7.19 Model for deriving equation for heart-valve orifice area P_1 and P_2 are upstream and downstream static pressures. Velocity u is calculated for minimal flow area A at location 2.

heights is negligible and that the velocity at location 1 is negligible compared with u, the velocity at location 2. Then (7.9) reduces to

$$P_1 - P_2 = \frac{\rho u^2}{2} \tag{7.10}$$

from which

$$u = \left(\frac{2(P_1 - P_2)}{\rho}\right)^{1/2}$$

At location 2, the flow $F = Au$, where A is the area. Hence

$$A = \frac{F}{u} = F\left(\frac{\rho}{2(P_1 - P_2)}\right)^{1/2} \tag{7.11}$$

In practice, there are losses due to friction, and the minimal flow area is smaller than the orifice area. Hence (7.10) becomes

$$A = \frac{F}{c_d}\left(\frac{\rho}{2(P_1 - P_2)}\right)^{1/2} \tag{7.12}$$

where c_d is a discharge coefficient. It has been found empirically that for semilunar valves, septal defects, and patent ductus, $c_d = 0.85$, whereas for mitral valves, $c_d = 0.6$ (Yellin *et al.*, 1975).

EXAMPLE 7.4 Calculate the approximate area of the aortic valve for the patient with the aortic and left-ventricular pressures shown in Figure 7.18(a). The patient's cardiac output was measured by thermodilution as 6400 ml/min and the heart rate as 78 bpm. Blood density is 1060 kg/m^3.

ANSWER From Figure 7.18(a), the ejection period is 0.31 s, and the average pressure drop is 7.33 kPa. During the ejection period, the flow (in SI units) is

$$\begin{aligned} F &= (6.4 \times 10^{-3}\ \text{m}^3/\text{min})(1/78\ \text{min/beat})(1/0.31\ \text{beats/s}) \\ &= 264 \times 10^{-6}\ \text{m}^3/\text{s} \end{aligned}$$

From (12) we have

$$\begin{aligned} A &= \frac{264 \times 10^{-6}}{0.85}\left(\frac{1060}{2(7330)}\right)^{1/2} \\ &= 83 \times 10^{-6}\ \text{m}^2 = 83\ \text{mm}^2 \end{aligned}$$

7.12 EFFECTS OF POTENTIAL AND KINETIC ENERGY ON PRESSURE MEASUREMENTS

In certain situations, the effects of potential- and kinetic-energy terms in the measurement of blood pressure may yield inaccurate results.

Bernoulli's equation (7.9) shows that the total pressure of a fluid remains constant in the absence of dissipative effects. The static pressure P of the fluid is the desired pressure; it is measured in a blood vessel when the potential- and kinetic-energy terms are zero.

We first examine the effect of the potential-energy term on the static pressure of the fluid. When measurements of blood pressure are taken with the patient in a supine (on-the-back) position and with the sensor so placed that it is at heart level, no corrections need be made for the potential-energy term. However, when the patient is sitting or standing, the long columns of blood in the arterial and venous pressure systems contribute a hydrostatic pressure, ρgh.

For a patient in the erect position, the arterial and venous pressure both increase to approximately 85 mm Hg (11.3 kPa) at the ankle. When the arm is held above the head, the pressure in the wrist becomes about 40 mm Hg (5.3 kPa). The sensor diaphragm should be placed at the same level as the pressure source. If this is not possible, the difference in height must be accounted for. For each 1.3 cm increase in height of the source, 1.0 mm Hg (133 Pa) must be added to the sensor reading.

The kinetic-energy term $\rho u^2/2$ becomes important when the velocity of blood flow is high. When a blood-pressure catheter is inserted into a blood vessel or into the heart, two types of pressures can be determined—side (static) and end (total) pressures. ''Side pressure'' implies that the end of the catheter has openings at right angles to the flow. In this case, the pressure reading is accurate because the kinetic-energy term is minimal. However, if the catheter pressure port is in line with the flow stream, then the kinetic energy of the fluid

Table 7.2 Relative Importance of the Kinetic-Energy Term in Different Parts of the Circulation

Vessel	Vel (cm/s)	KE (mm Hg)	Systolic (mm Hg)	Systolic (kPa)	% KE of Total
Aorta (systolic)					
At rest	100	4	120	(16)	3
Cardiac output at 3 × rest	300	36	180	(24)	17
Brachial artery					
At rest	30	0.35	110	(14.7)	0.3
Cardiac output at 3 × rest	90	4	120	(16)	3
Venae cavae					
At rest	30	0.35	2	(0.3)	12
Cardiac output at 3 × rest	90	3.2	3	(0.4)	52
Pulmonary artery					
At rest	90	3	20	(2.7)	13
Cardiac output at 3 × rest	270	27	25	(3.3)	52

SOURCE: From A. C. Burton, *Physiology and Biophysics of the Circulation*. Copyright 1972 by Year Book Medical Publishers, Inc., Chicago. Used by permission.

at that point is transformed into pressure. If the catheter pressure port faces upstream, the recorded pressure is the side pressure plus the additional kinetic-energy term $\rho u^2/2$. On the other hand, if the catheter pressure port faces downstream, the value is approximately $\rho u^2/2$ less than the side pressure. When the catheter is not positioned correctly, artifacts may develop in the pressure reading.

The data given in Table 7.2 demonstrate the relative importance of the kinetic-energy term in different parts of the circulation (Burton, 1972). As Table 7.2 shows, there are situations in the aorta, venae cavae, and pulmonary artery in which the kinetic-energy term is a substantial part of the total pressure. For the laminar-flow case, this error decreases as the catheter pressure port is moved from the center of the vessel to the vessel wall, where the average velocity of flow is less. The kinetic-energy term could also be important in a disease situation in which an artery becomes narrowed.

EXAMPLE 7.5 Determine whether the kinetic-energy term is significant for measurements of pressure in the human descending aorta. Assume that the peak velocity of flow in the center of the aorta is approximately 1.5 m/s and that the density of the blood ρ is 1060 kg/m^3.

ANSWER The kinetic energy (K.E.) term is $(1/2)\rho V^2$

$$\text{K.E.} = \frac{1}{2}\left(1050\,\text{kg/m}^3\right)(1.5\,\text{m/s})^2 = 1181\,\text{Pa}$$

or in terms of mm Hg,

$$\text{K.E.} = 1181/133 = 9\,\text{mm Hg}$$

The orientation of the pressure catheter can significantly affect the measured aortic pressure. Under laminar flow conditions, this error may be decreased by moving the tip to wall where the average flow velocity is less.

7.13 INDIRECT MEASUREMENTS OF BLOOD PRESSURE

Indirect measurement of blood pressure is an attempt to measure intra-arterial pressures noninvasively. The most standard manual techniques employ either the palpation or the auditory detection of the pulse distal to an occlusive cuff. Figure 7.20 shows a typical system for indirect measurement of blood pressure. It employs a sphygmomanometer consisting of an inflatable cuff for occlusion of the blood vessel, a rubber bulb for inflation of the cuff, and either a mercury or an aneroid manometer for detection of pressure.

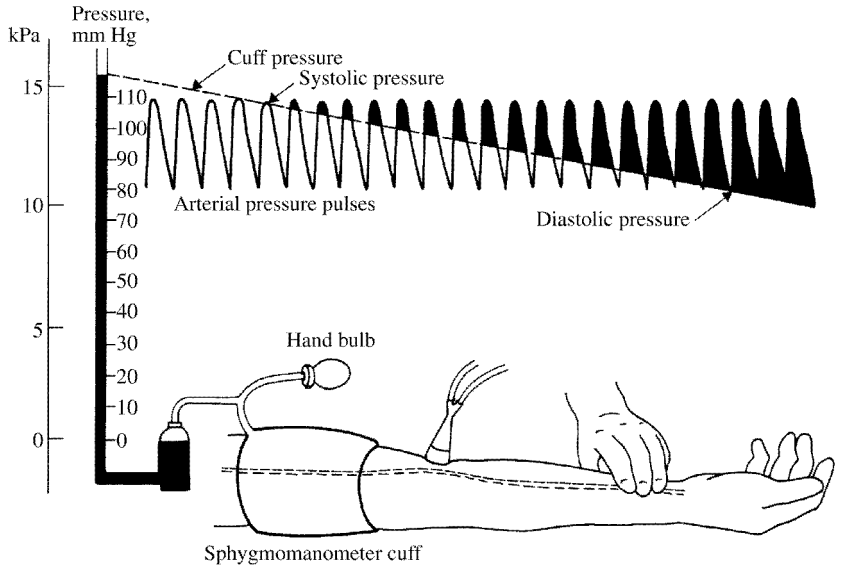


Figure 7.20 Typical indirect blood-pressure measurement system The sphygmomanometer cuff is inflated by a hand bulb to pressure above the systolic level. Pressure is then slowly released, and blood flow under the cuff is monitored by a microphone or stethoscope placed over a downstream artery. The first Korotkoff sound detected indicated systolic pressure, whereas the transition from muffling to silence brackets diastolic pressure. (From R.F. Rushmer, *Cardiovascular Dynamics*, 3rd ed., 1970. Philadelphia: W.B. Saunders Co. Used with permission.)

Blood pressure is measured in the following way. The occlusive cuff is inflated until the pressure is above systolic pressure and then is slowly bled off (2 to 3 mm Hg/s) (0.3 to 0.4 kPa/s). When the systolic peaks are higher than the occlusive pressure, the blood spurts under the cuff and causes a palpable pulse in the wrist (Riva–Rocci method). Audible sounds (Korotkoff sounds) generated by the flow of blood and vibrations of the vessel under the cuff are heard through a stethoscope. The manometer pressure at the first detection of the pulse indicates the systolic pressure. As the pressure in the cuff is decreased, the audible Korotkoff sounds pass through five phases (Geddes, 1970). The period of transition from muffling (phase IV) to silence (phase V) brackets the diastolic pressure.

In employing the palpation and auscultatory techniques, you should take several measurements, because normal respiration and vasomotor waves modulate the normal blood-pressure levels. These techniques also suffer from the disadvantage of failing to give accurate pressures for infants and hypotensive patients.

Using an occlusive cuff of the correct size is important if the clinician is to obtain accurate results. The pressure applied to the artery wall is assumed to be equal to that of the external cuff. However, the cuff pressure is transmitted via interposed tissue. With a cuff of sufficient width and length, the cuff pressure is evenly transmitted to the underlying artery. It is generally accepted that the width of the cuff should be about 0.40 times the circumference of the extremity. However, no general agreement appears to exist about the length of the pneumatic cuff (Geddes, 1970). If a short cuff is used, it is important that it be positioned over the artery of interest. A longer cuff reduces the problem of misalignment. The cuff should be placed at heart level to avoid hydrostatic effects.

The auscultatory technique is simple and requires a minimum of equipment. However, it cannot be used in a noisy environment, whereas the palpation technique can. The hearing acuity of the user must be good for low frequencies from 20 to 300 Hz, the bandwidth required for these measurements. Bellville and Weaver (1969) have determined the energy distribution of the Korotkoff sounds for normal patients and for patients in shock. When there is a fall in blood pressure, the sound spectrum shifts to lower frequencies. The failure of the auscultation technique for hypotensive patients may be due to low sensitivity of the human ear to these low-frequency vibrations (Geddes, 1970).

There is a common misconception that normal human blood pressure is 120/80, meaning that the systolic value is 120 mm Hg (16 kPa) and that the diastolic value is 80 mm Hg (10.7 kPa). This is not the case. A careful study (by Master *et al*., 1952) showed that the age and sex of an individual determine the ''normal value'' of blood pressure.

A number of techniques have been proposed to measure automatically and indirectly the systolic and diastolic blood pressure in humans (Cobbold, 1974). The basic technique involves an automatic sphygmomanometer that inflates and deflates an occlusive cuff at a predetermined rate. A sensitive

detector is used to measure the distal pulse or cuff pressure. A number of kinds of detectors have been employed, including ultrasonic, piezoelectric, photoelectric, electroacoustic, thermometric, electrocardiographic, rheographic, and tissue-impedance devices (Greatorex, 1971; Visser and Muntinga, 1990). Three of the commonly used automatic techniques are described in the following paragraphs.

The first technique employs an automated auscultatory device wherein a microphone replaces the stethoscope. The cycle of events that takes place begins with a rapid (20 to 30 mm Hg/s) (2.7 to 4 kPa/s) inflation of the occlusive cuff to a preset pressure about 30 mm Hg higher than the suspected systolic level. The flow of blood beneath the cuff is stopped by the collapse of the vessel. Cuff pressure is then reduced slowly (2 to 3 mm Hg/s) (0.3 to 0.4 kPa/s). The first Korotkoff sound is detected by the microphone, at which time the level of the cuff pressure is stored. The muffling and silent period of the Korotkoff sounds is detected, and the value of the diastolic pressure is also stored. After a few minutes, the instrument displays the systolic and diastolic pressures and recycles the operation. Design considerations for various types of automatic indirect methods of measurement of blood pressure can be found in the literature (Greatorex, 1971).

The ultrasonic determination of blood pressure employs a transcutaneous Doppler sensor that detects the motion of the blood-vessel walls in various states of occlusion. Figure 7.21 shows the placement of the compression cuff over two small transmitting and receiving ultrasound crystals (8 MHz) on the arm (Stegall *et al*., 1968). The Doppler ultrasonic transmitted signal is focused on the vessel wall and the blood. The reflected signal (shifted in frequency) is detected by the receiving crystal and decoded (Section 8.4). The difference in frequency, in the range of 40 to 500 Hz, between the transmitted and received signals is proportional to the velocity of the wall motion and the blood velocity. As the cuff pressure is increased above diastolic but below systolic, the vessel opens and closes with each heartbeat, because the pressure in the artery oscillates above and below the applied external pressure in the cuff. The opening and closing of the vessel are detected by the ultrasonic system.

As the applied pressure is further increased, the time between the opening and closing decreases until they coincide. The reading at this point is the systolic pressure. Conversely, when the pressure in the cuff is reduced, the time between opening and closing increases until the closing signal from one pulse coincides with the opening signal from the next. The reading at this point is the diastolic pressure, which prevails when the vessel is open for the complete pulse.

The advantages of the ultrasonic technique are that it can be used with infants and hypotensive individuals and in high-noise environments. A disadvantage is that movements of the subject's body cause changes in the ultrasonic path between the sensor and the blood vessel. Complete reconstruction of the arterial-pulse waveform is also possible via the ultrasonic method. A timing pulse from the ECG signal is used as a reference. The clinician uses the pressure in the cuff when the artery opens versus the time

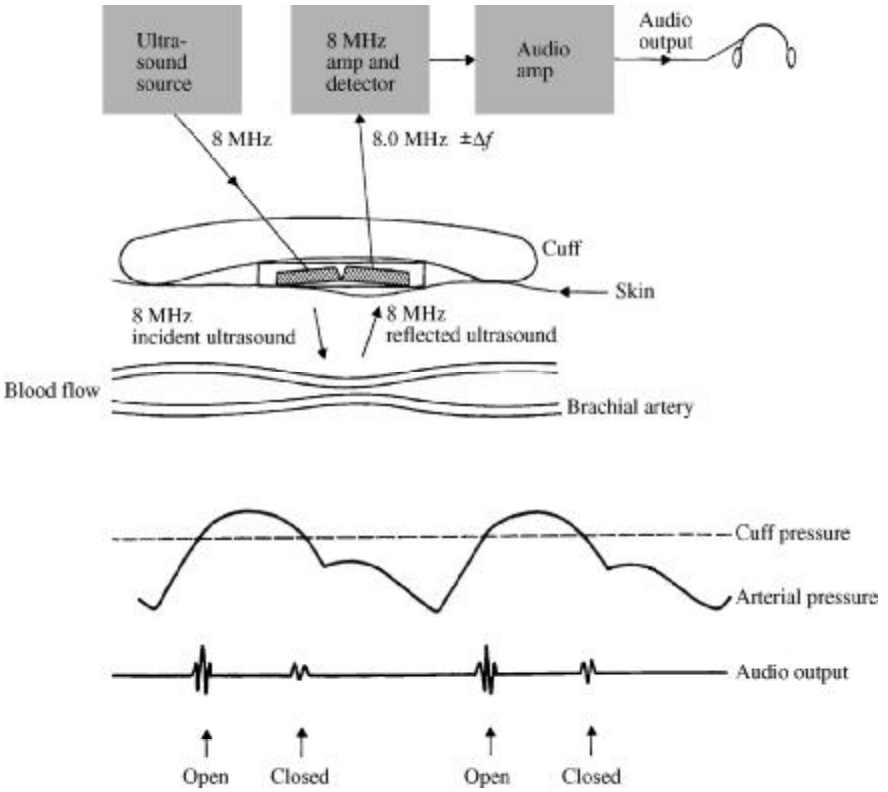


Figure 7.21 Ultrasonic determination of blood pressure A compression cuff is placed over the transmitting (8 MHz) and receiving ($8\,\text{MHz} \pm \Delta f$) crystals. The opening and closing of the blood vessel are detected as the applied cuff pressure is varied. (From H. F. Stegall, M. B. Kardon, and W. T. Kemmerer, ''Indirect measurement of arterial blood pressure by Doppler ultrasonic sphygmomanometry.'' *J. Appl. Physiol.*, 1968, 25, 793–798. Used with permission.)

from the ECG R wave to plot the rising portion of the arterial pulse. Conversely, the clinician uses the cuff pressure when the artery closes versus the time from the ECG R wave to plot the falling portion of the arterial pulse.

The oscillometric method, a noninvasive blood pressure technique, measures the amplitude of oscillations that appear in the cuff pressure signal which are created by expansion of the arterial wall each time blood is forced through the artery. The uniqueness of the oscillometric method, a blood-pressure cuff technique, is that specific characteristics of the compression cuff's entrained air volume are used to identify and sense blood-pressure values. The cuff-pressure signal increases in strength in the systolic pressure region, reaching a maximum when the cuff pressure is equal to mean arterial pressure. As the cuff pressure drops below this point, the signal strength decreases proportionally to the cuff air pressure bled rate. There is no clear transition in cuff-pressure oscillations

to identify diastolic pressure since arterial wall expansion continues to happen below diastolic pressure while blood is forced through the artery (Geddes, 1984). Thus, oscillometric monitors employ proprietary algorithms to estimate the diastolic pressure.

Ramsey (1991) has indicated that, using the oscillometric method, the mean arterial pressure is the single blood-pressure parameter, which is the most robust measurement, as compared with systolic and diastolic pressure, because it is measured when the oscillations of cuff pressure reach the greatest amplitude. This property usually allows mean arterial pressure to be measured reliably even in case of hypotension with vasoconstriction and diminished pulse pressure.

When the cuff pressure is raised quickly to pressures higher than systolic pressure it is observed that the radial pulse disappears. Cuff pressures above systolic cause the underlying artery to be completely occluded. However, at suprasystolic cuff pressures, small amplitude pressure oscillations occur in the cuff pressure due to artery pulsations under the upper edge of the cuff, which are communicated to the cuff through the adjacent tissues. With slow cuff-pressure reductions, when the cuff pressure is just below systolic pressure, blood spurts through the artery and the cuff-pressure oscillations become larger. Figure 7.22 illustrates the ideal case in which the cuff pressure is monitored by a pressure sensor connected to a strip chart recorder. A pressure slightly above systolic pressure is detected by determining the shift from small-amplitude oscillations at cuff pressure slightly above systolic pressure and when the cuff pressure begins to increase amplitude (point 1). As the cuff

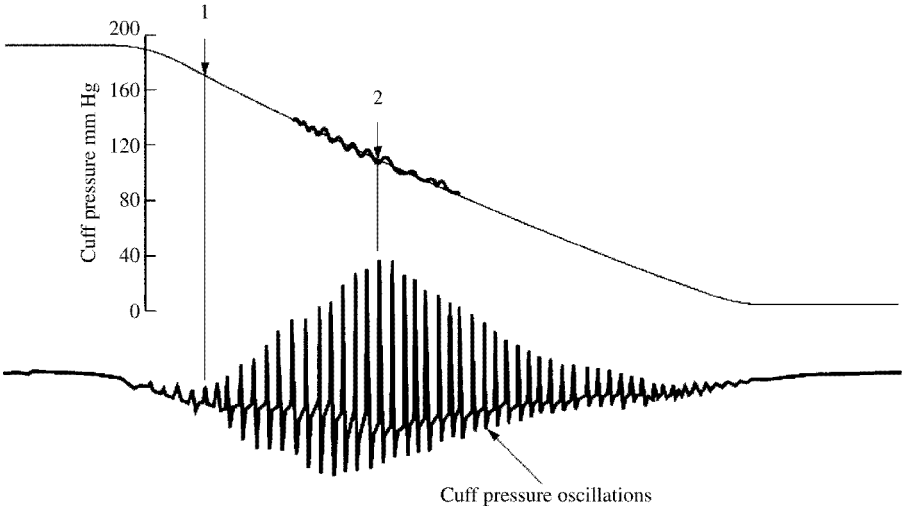


Figure 7.22 The oscillometric method A compression cuff is inflated above systolic pressure and slowly deflated. Systolic pressure is detected (point 1) where there is a transition from small amplitude oscillations (above systolic pressure) to increasing cuff-pressure amplitude. The cuff-pressure oscillations increase to a maximum (point 2) at the mean arterial pressure.

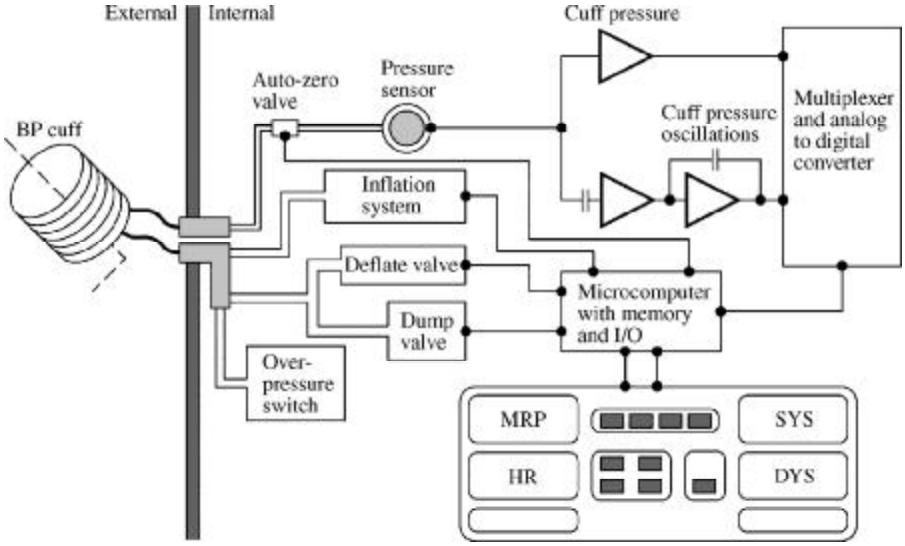


Figure 7.23 Block diagram of the major components and subsystems of an oscillometric blood-pressure monitoring device, based on the Dinamap unit, I/O = input/output; MAP = mean arterial pressure; HR = heart rate; SYS = systolic pressure; DYS = diastolic pressure. [From Ramsey M III. Blood Pressure monitoring: automated oscillometric devices, *J. Clin. Monit.* 1991, 7, 56–67]

continues to deflate, the amplitude of the oscillations increases reaching a maximum, and then decreases as the cuff pressure is decreased to zero. Point 2 in Figure 7.22 is the maximum cuff-pressure oscillation, which is essentially true mean arterial pressure. Since there is no apparent transition in the oscillation amplitude as cuff pressure passes diastolic pressure, algorithmic methods are used to predict diastolic pressure.

The system description begins with the blood-pressure cuff, which compresses a limb and its vasculature by the encircling inflatable compression cuff pressures (Ramsey, 1991). The cuff is connected to a pneumatic system (see Figure 7.23). A solid-state pressure sensor senses cuff pressure, and the electric signal proportional to pressure is processed in two different circuits. One circuit amplifies and corrects the zero offset of the cuff-pressure signal before the analog-to-digital digitization. The other circuit high-pass filters and amplifies the cuff-pressure signal. Cuff pressure is controlled by a microcomputer that activates the cuff inflation and deflation systems during the measurement cycle.

7.14 TONOMETRY

The basic principle of tonometry is that, when a pressurized vessel is partly collapsed by an external object, the circumferential stresses in the vessel wall are removed and the internal and external pressures are equal. This approach

has been used quite successfully to measure intra-ocular pressure and has been used with limited success to determine intraluminal arterial pressure.

The force-balance technique can be used to measure intra-ocular pressure. Based on the Imbert–Fick law, the technique enables the clinician to find intra-ocular pressure by dividing the applanation force by the area of applanation. Goldmann (1957) developed an applanation tonometer, which is the currently accepted clinical standard. With this technique, the investigator measures the force required to flatten a specific optically determined area. Mackay and Marg (1960) developed a sensor probe that is applied to the corneal surface; the cornea is flattened as the probe is advanced. The intra-ocular pressure is detected by a force sensor in the center of an annular ring, which unloads the bending forces of the cornea from the sensor.

Forbes *et al.* (1974) developed an applanation tonometer that measures intra-ocular pressure without touching the eye. An air pulse of linearly increasing force deforms and flattens the central area of the cornea, and it does so within a few milliseconds. The instrument consists of three major components. The first is a pneumatic system that delivers an air pulse the force of which increases linearly with time. As the air pulse decays, it causes a progressive reduction of the convexity of the cornea and, finally, a return to its original shape.

The second component, the system that monitors the applanation, determines the occurrence of applanation with microsecond resolution by continuously monitoring the status of the curvature of the cornea. Figure 7.24 (a) and (b) shows the systems of optical transmission and detection and the light rays reflected from an undisturbed and an applanated cornea, respectively.

Two obliquely oriented tubes are used to detect applanation. Transmitter tube T directs a collimated beam of light at the corneal vertex; a telecentric receiver R observes the same area. The light reflected from the cornea passes through the aperture A and is sensed by the detector D. In the case of the undisturbed cornea, the detector receives little or no light. As the cornea's

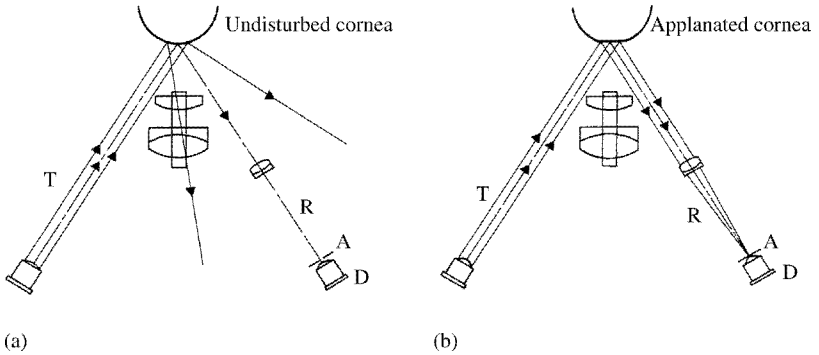


Figure 7.24 Monitoring system for noncontact applanation tonometer [From M. Forbes, G. Pico, Jr., and B. Grolman, "A Noncontact Applanation Tonometer, Description and Clinical Evaluation," *J. Arch. Ophthalmology*, 1975, 91, 134-140. Copyright© 1975, American Medical Association. Used with permission.]

convexity is progressively reduced to the flattened condition, the amount of light detected is increased. When the cornea is applanated, it acts like a plano-mirror with a resulting maximal detected signal. When the cornea becomes concave, a sharp reduction in light detection occurs. The current source for the pneumatic solenoid is immediately shut off when applanation is detected in order to minimize further air-pulse force impinging on the cornea. A direct linear relationship has been found between the intra-ocular pressure and the time interval to applanation.

The principles of operation of the arterial tonometry are very similar to those for the ocular tonometry, discussed above. The arterial tonometer measures dynamic arterial blood pressure, i.e., it furnishes continuous measurements of arterial pressure throughout the total heart cycle (Eckerle, 2006). The instrument sensor is placed over a superficial artery that is supported from below by bone. The radial artery at the wrist is a convenient site for arterial tonometer measurements. The arterial tonometer suffers from relatively high cost when compared to a conventional sphygmomanometer. One significant advantage of the arterial tonometer is its ability to make noninvasive, non-painful, continuous measurements for long periods of time.

Figure 7.25 shows an arterial tonometer model that depicts system operation in which the arterial blood pressure, P, from a superficial artery and the

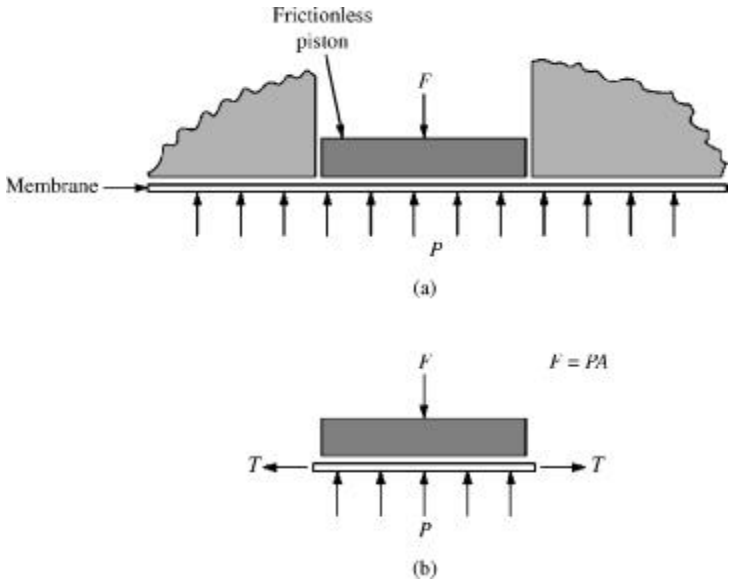


Figure 7.25 Idealized model for an arterial tonometer (a) A flattened portion of an arterial wall (membrane). P is the blood pressure in a superficial artery, and F is the force measured by a tonometer transducer. (b) A free-body diagram for the idealized model of (a) in which T is the membrane tensile force perpendicular to both F and P. [From Eckerle, J. D., "Tonometry, arterial." In J. G. Webster (ed.), *Encyclopedia of Medical Devices and Instrumentation*, 2nd ed. New York: Wiley, 2006, Vol. 6, pp. 402–410.]

force, F, is measured by a tonometer sensor. The artery wall is represented by a flat ideal membrane, M. A free-body diagram is used to describe the force balances. The ideal membrane only transmits a tensile force, T, without any bending moment. Vertical force balance shows that the tension vector, T, is perpendicular to the pressure vector. Thus, the force, F, is in quadrature to and independent of T and only depends on the blood pressure and the area of the frictionless piston, A. Thus, measurement of the force, F, permits direct measurement of the intra-arterial pressure.

Eckerle (2006) indicates that several conditions must be met by the tonometer sensor and an appropriate superficial artery for proper system operation:

1. A bone provides support for the artery, opposite to the applied force.
2. The hold-down force flattens the artery wall at the measurement site without occluding the artery.
3. Compared to artery diameter, the skin thickness over the artery is insignificant.
4. The artery wall has the properties of an ideal membrane.
5. The arterial rider, positioned over the flattened area of the artery, is smaller than the artery.
6. The force transducer spring constant K_T is larger than the effective spring constant of the artery.

When all these conditions hold, it has been shown on a theoretical basis that the electrical output signal of the force sensor is directly proportional to the intra-arterial blood pressure (Pressman and Newgard, 1963). However, a major practical problem with the above approach, using a single arterial tonometer, is that the arterial rider must be precisely located over the superficial artery. A solution to this problem is the use of an arterial tonometer with multiple element sensors. Figure 7.26 shows a linear array of force sensors and arterial riders positioned such that at least one element of the array is centered over the artery. A computer algorithm is used to automatically select from the multiple sensors, the sensor element which is positioned over the artery. One approach is to use two pressure distribution characteristics in the vicinity of the artery in which an element-selection algorithm searches for a (spatial) local minimum in diastolic pressure in a region near the maximum pulse amplitude (Eckerle, 2006). The sensor with these characteristics is assumed to be centered over the artery, and the blood pressure from this sensor is measured with this element.

In addition to positioning the sensor over the artery, the degree of arterial flattening is another important factor for accurate tonometric pressure measurements. The hold-down force F_1, (in Figure 7.26), which causes arterial flattening, is a function of the interaction of anatomical factors. The hold-down force for each subject must be determined before tonometric readings can be taken. The hold-down force is gradually increased (or decreased) while recording the tonometer sensor output.

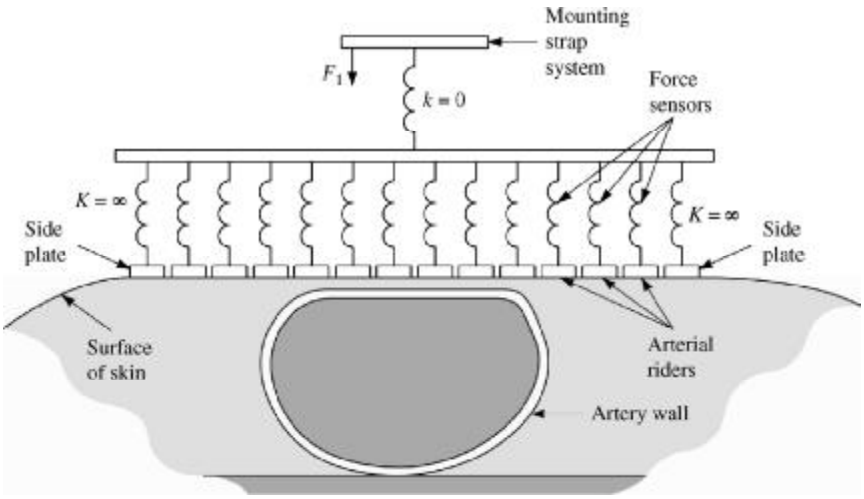


Figure 7.26 Multiple-element arterial tonometer The multiple element linear array of force sensors and arterial riders are used to position the system such that some element of the array is centered over the artery. [From Eckerle, J. D., "Tonometry, arterial," in J. G. Webster (ed.), *Encyclopedia of Medical Devices and Instrumentation.* 2nd ed. New York: Wiley, 2006, vol. 6, pp. 402–410.]

Multiple-element tonometer sensors have been manufactured from a monolithic silicon substrate using anisotropic etching to define pressure-sensing diaphragms (10 μm thick in the silicon). Piezoresistive strain gages in the diaphragms are fabricated using integrated-circuit (IC) processing techniques. The strain gage's resistance is used to determine the pressure exerted on each sensor element.

Note that the radial artery is not the only measurement site at which a tonometer may be applied. Other possible sites for tonometric measurements include the brachial artery at the inner elbow (the antecubital fossa), the temporal artery in front of the ear, and the dorsalis pedis artery on the upper foot (Eckerle, 2006). Arterial tonometers have not been commercially successful because of inaccuracy caused by wrist movement, tendons overlying arteries, etc.

Gizdulich and Wesseling (1990) measure the arterial pressure in the finger continuously and indirectly by using the Peñás method. They apply counter-pressure just sufficient to hold the arteries under a pressure cuff at their unstressed diameter at zero transmural pressure monitored by an infrared plethysmograph. Because the method occludes the veins, continuous use for more than 20 min causes discomfort and swelling. Thus, the pressure must be relieved periodically.

EXAMPLE 7.6 Design a noninvasive (no breaks in the skin) system for measuring the velocity of propagation of a blood-pressure wave from the aortic valve in the heart to the radial artery on the wrist. Name and describe the sensors required, their placement, the expected waveforms, and indicate the times required to measure velocity of propagation.

ANSWER Place a piezoelectric pressure sensor on the neck to measure carotid pulse. Place a similar sensor on the wrist to measure radial pulse. Use a digital storage oscilloscope to display both waveforms and measure the difference in timing of pulse upstroke t for the two waveforms. Measure distance from the aortic root to neck d_n. Measure distance from aortic root to wrist d_w.

$$\text{Velocity} = \text{distance/time} = (d_w - d_n)/t$$

PROBLEMS

7.1 Compare the transient-step and sinusoidal-frequency methods for determining the response characteristics of a catheter-sensor system.

7.2 Find (a) the damping ratio, (b) the undamped frequency, and (c) the frequency-response curve of the pressure sensor for which the transient response to a step change in pressure is shown in Figure 7.11.

7.3 Find the frequency-response curve of the sensor in Problem 7.2, given that its chamber is filled with the whole blood at body temperature (37 °C). The original data in Problem 7.2 were obtained with water at 20 °C.

7.4 What happens to the frequency response of a P23Dd sensor, 6 F, 1 m, water-filled catheter system (at 20 °C) when a tiny pinhole leak occurs at the junction of the catheter and sensor? The leak allows a 0.40 ml/min flow for a pressure head of 100 mm Hg (13.3 kPa). Plot frequency-response curves for the system with and without the leak. (An intentional leak is often desirable to permit constant flushing of the catheter and thus inhibit the formation of clots.)

7.5 A low-pass filter is added to the catheter of a pressure-sensor–catheter-sensor system by pinching the catheter. The system consists of a Statham P23Dd sensor and a 1 m, 6 F, polyethylene catheter. The pinch effectively reduces the diameter of the catheter to 25% of its original diameter.

a. How long must the pinch be for the system's damping factor to be equal to 0.7?

b. Sketch the frequency response for the system with and without the pinch.

c. Sketch the time response for the system with and without the pinch when it is excited by a 100 mm-Hg step input.

d. Discuss how faithfully the two systems will reproduce the blood-pressure waveform for humans, dogs, and shrews with heart-rate variations of 1 to 3.3 beats/s, 1.5 to 5 beats/s, and 12 to 22 beats/s, respectively.

7.6 A heart murmur has a frequency of 300 Hz. Give the block diagram and sketch waveforms for the special instrumentation that enables us to show the occurrence of this murmur on a 0 to 80 Hz pen recorder.

7.7 Name the two basic causes of abnormal heart murmurs. For each type, give an example and show on a sketch when it occurs relative to systole and diastole.