Lecture Four

Blood pressure and sound

Introduction

Determining an individual’s blood pressure is a standard clinical measurement. 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 pressures generated by the right and left sides of the heart differ somewhat in shape and in amplitude (see Figure 1). 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.

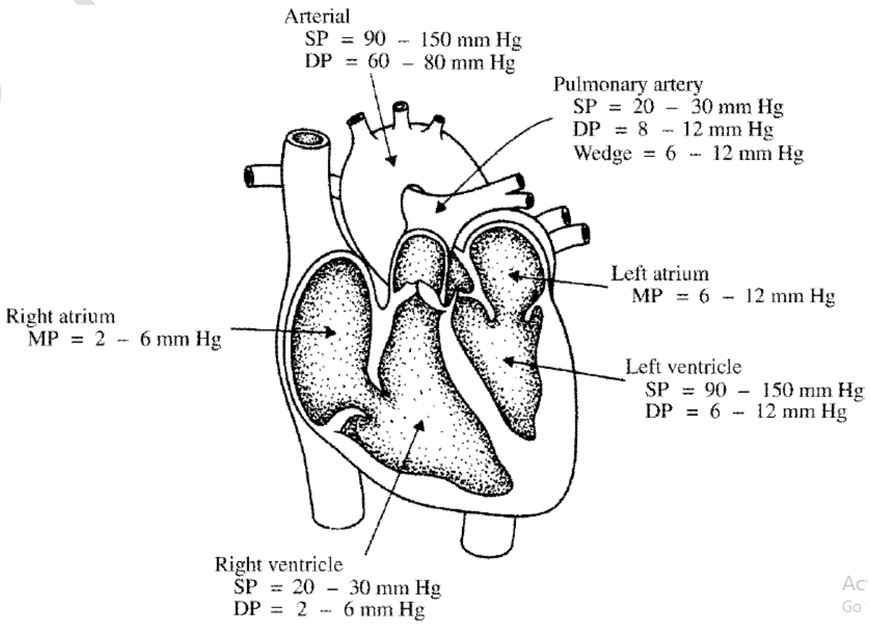


Figure 1 Typical values of circulatory pressures SP is the systolic pressure, DP is the diastolic pressure, and MP is the mean pressure.

The various types of pressure are demonstrated below. The fundamental force plays a crucial role in numerous physiological processes. The formula, $P = F/A$, represents the calculation of pressure, where "P" stands for pressure, "F" for force, and "A" for area. The different types of pressure are:

- Arterial blood pressure (ABP): The pressure exerted by blood within arteries and used in critical care units (Intensive Care Unit (ICU) and Coronary Care Unit (CCU)).
- Central venous pressure (CVP): The pressure within the large veins near the heart.
- Pulmonary artery pressure: The pressure within the pulmonary artery, which carries blood from the heart to the lungs.
- Spinal fluid pressure: The pressure of the cerebrospinal fluid, which surrounds the brain and spinal cord.
- Intraventricular brain pressure: The pressure within the ventricles of the brain, which are fluid-filled cavities.
- Intraocular pressure: The pressure within the eyeball.

Understanding these different pressure types is crucial in various medical fields for diagnosing and treating various conditions. Typical haemodynamic pressure values in the basic circulatory system are as follows:

Arterial system 30–300 mmHg
Venous system 5–15 mmHg
Pulmonary system 6–25 mmHg

The most frequently monitored pressures, which have clinical usefulness in medium and long-term patient monitoring, are:

1. The arterial pressure and 2. The venous pressure.

There are two basic methods for measuring blood pressure:

1. Direct and, 2. Indirect.

Direct measurement

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.

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 2). 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.

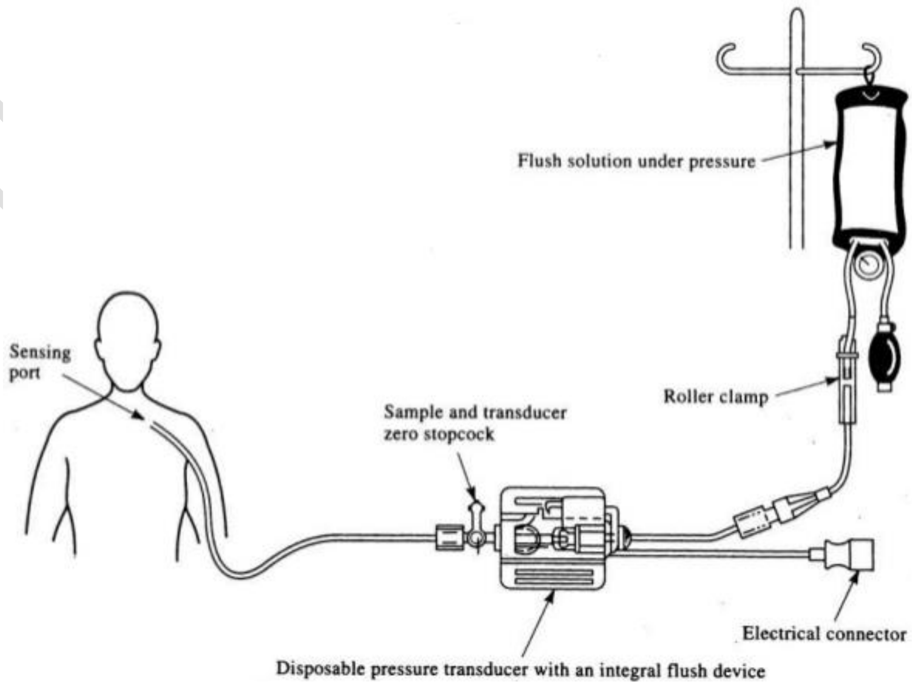


Figure 2 Extravascular pressure-sensor system A catheter couples a flush solution (heparinized saline) through a disposable pressure sensor with an integral flush device to the sensing port. The three-way stopcock is used to take blood samples and zero the pressure sensor.

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 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 3 shows a modern disposable blood-pressure sensor.

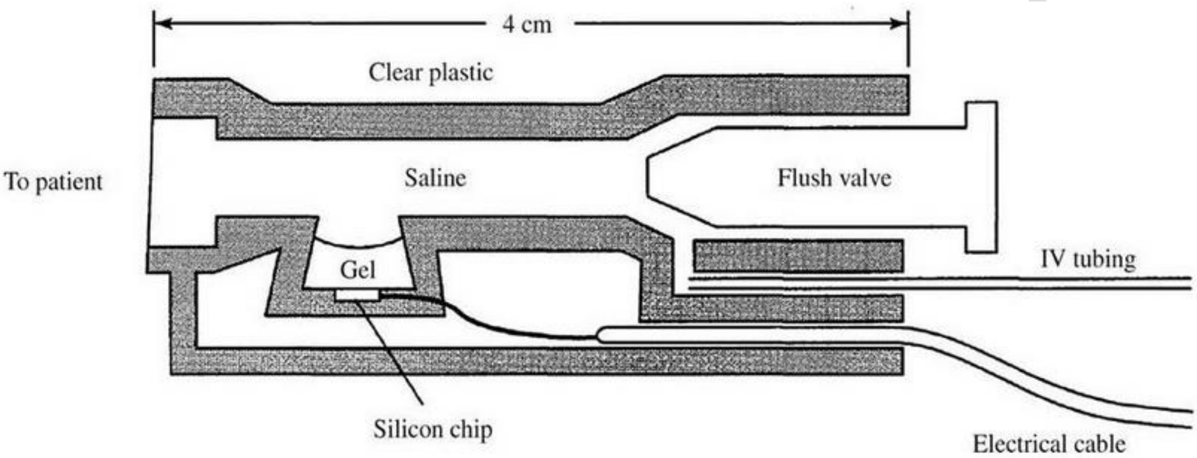


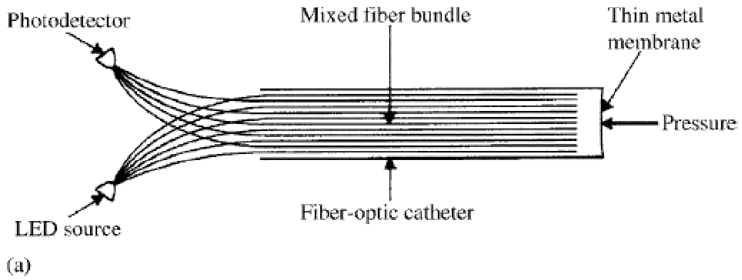
Figure 3 isolation in *a* disposable blood-pressure sensor. Disposable blood-pressure sensors are made of clear plastic so air bubbles are easily seen. Saline flows from an intravenous (IV) bag through the clear IV tubing and the sensor to the patient. This flushes blood out of the tip of the indwelling catheter to prevent clotting. A lever can open or close the flush valve. The silicon chip has a silicon diaphragm with a four-resistor Wheatstone bridge diffused into it. Its electrical connections are protected from the saline by a compliant silicone elastomer gel, which also provides electrical isolation. This prevents electric shock from the sensor to the patient and prevents destructive currents during defibrillation from the patient to the silicon chip.

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 basic type of sensor includes strain gage system bonded onto a flexible diaphragm at the catheter tip. Gages of this type are available in the F5 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. 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.

A fiber-optic microtip sensor for *in vivo* measurements inside the human body is shown in Figure 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. 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 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. The coupling between LED source and detector is a function of the overlap of the two acceptance angles on the pressure-sensor membrane. The operating portion of the curve is the left slope region where the characteristic is steepest.



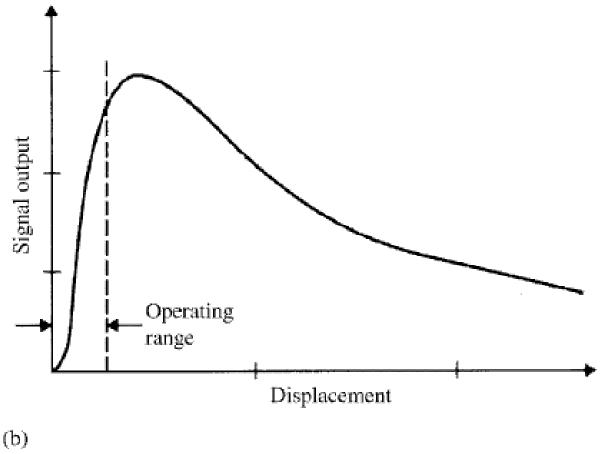


Figure 4 (a) Schematic diagram of an intravascular fiber-optic pressure sensor. Pressure causes deflection in a thin metal membrane that modulates the coupling between the source and detector fibers, (b) Characteristic curve for the fiber-optic pressure sensor.

Figure 5 describes a fiber-optic pressure sensor for intracranial pressure measurements in the newborn, 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 the fontanel tissue-is flat, indicating that the sensor air pressure and the fontanel or intracranial pressure are equal.

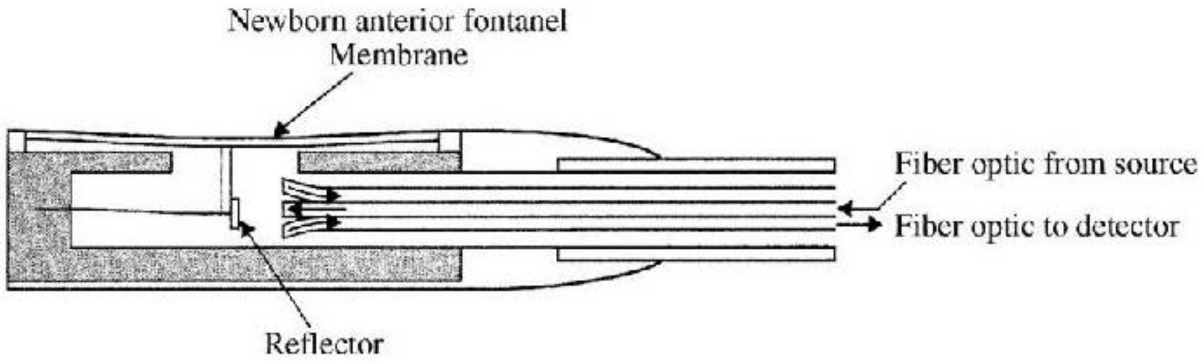


Figure 5 Fiber-optic pressure sensor for intracranial pressure measurements in the newborn. The sensor membrane is placed in contact with the anterior fontanel of the newborn.

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. *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 (is the medical condition where there is too much fluid in the blood. Fluid volume excess in the intravascular compartment occurs due to an increase in total body sodium content and a consequent increase in extracellular body water), or circulatory failure.

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 standard practice to accept venous pressure values only when respiratory swings are evident. Normal central venous pressures range widely from 0 to 12cm H_2O (Oto 1.2kPa), with a mean pressure of 5 cmH_2O (0.5kPa).

Esophageal manometry uses a similar low-pressure catheter system (Velanovich, 2OO6). A hydraulic capillary infusion system infuses O.6 ml/ min to prevent sealing of the catheter orifice in the oesophagus.

Heart sounds

Introduction

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 origin of heart sounds and murmurs. 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 1 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. 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).

The 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.

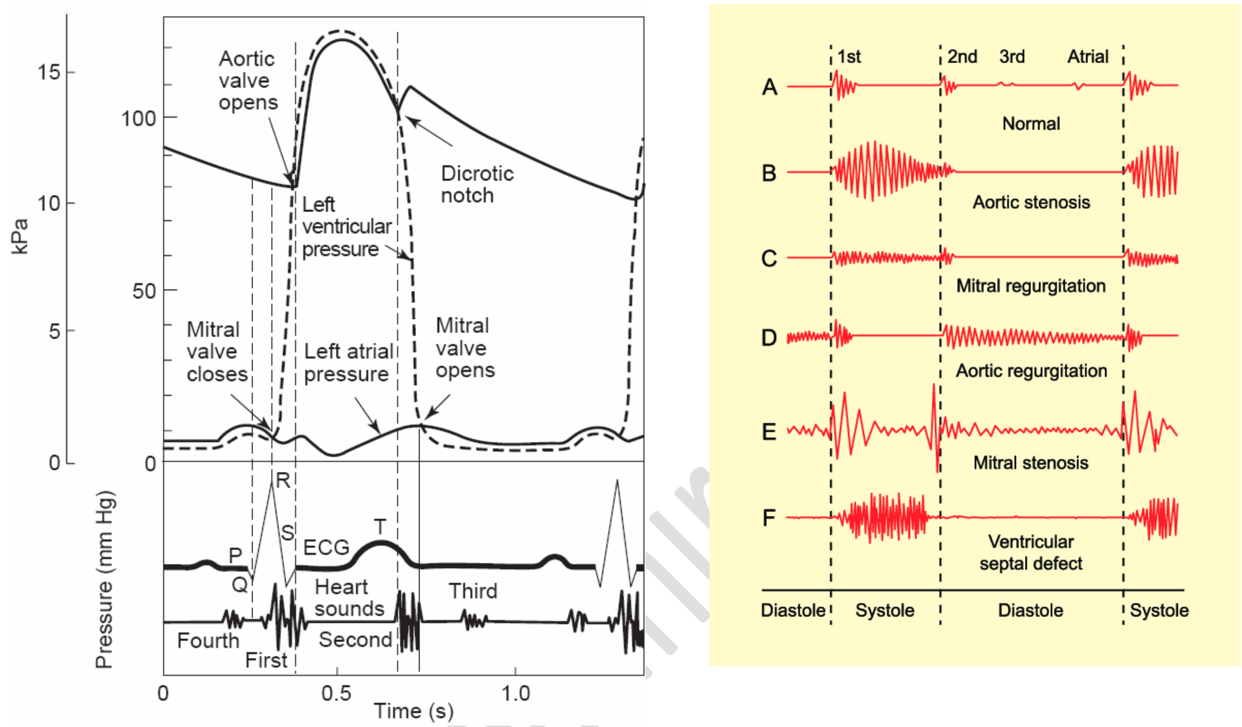


Figure 1 Correlation of the four heart sounds with electric and mechanical events of the cardiac cycle.

AUSCULTATION TECHNIQUES

Heart sounds travel through the body from the heart and major blood vessels to the body surface. The sound waves are attenuated and the largest attenuation of the wavelike motion occurs in the most compressible tissues, such as the lungs and fat layers.

There are four basic chest locations at which the intensity of sound from the four valves is maximized (Figure 2). In these sites, the intensity of sound is the highest because the sound is being transmitted through solid tissues or through a minimal thickness of inflated lung.

Heart sounds and murmurs have extremely small amplitudes, with frequencies from 0.1 to 2000Hz. 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.

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 non-bed ridden patients in a specially designed, acoustically quiet room. Artifacts from movements of the patient appear as baseline wandering.

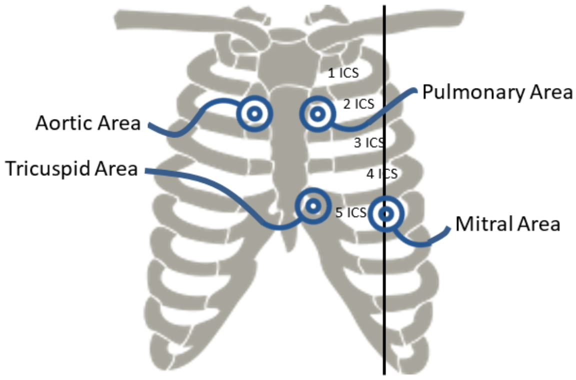


Fig. 2
Areas on the Precordium for Auscultation of Heart.

Figure 2 Auscultatory areas on the chest A, aortic; P, pulmonary; T, tricuspid; and M, mitral areas.

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. The stethoscope acoustics reflected the acoustics of the human ear. Younger individuals revealed slightly better responses to a stethoscope than their elders.

Figure 3 is a typical frequency-response curve for a stethoscope; it shows that the mechanical stethoscope has an uneven frequency response, with many resonance peaks. 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 3dB.

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. When the stethoscope chest piece is firmly applied, the diaphragm becomes taut with pressure, thereby causing an attenuation of low frequencies. A physician may miss, with one instrument, sounds that can be heard with another. In addition, 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.

Engineers have proposed many types of electronic stethoscopes. These devices have selectable frequency-response characteristics ranging from the "ideal" flat-response case and selected band-passes 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.

Stethoscopes are also useful for listening to the sounds caused by airflow obstruction or lung collapse.

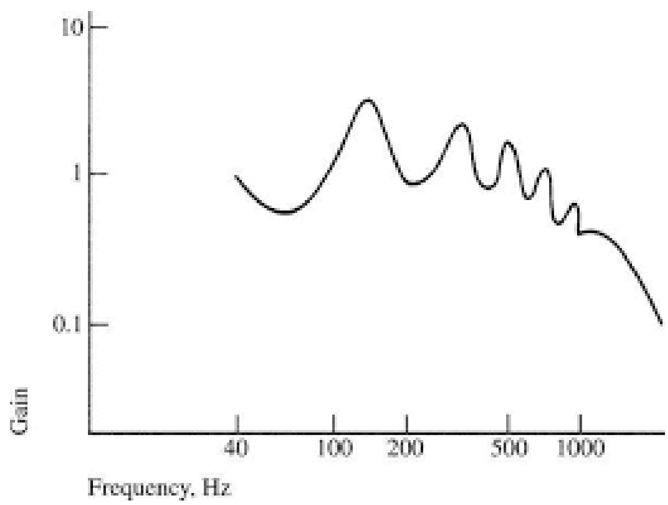


Figure 3 The typical frequency-response curve for a stethoscope

INDIRECT MEASUREMENTS OF BLOOD PRESSURE

This method measures the 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 4 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.

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. 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. The period of transition from muffling (phase IV) to silence (phase V) brackets the diastolic pressure.

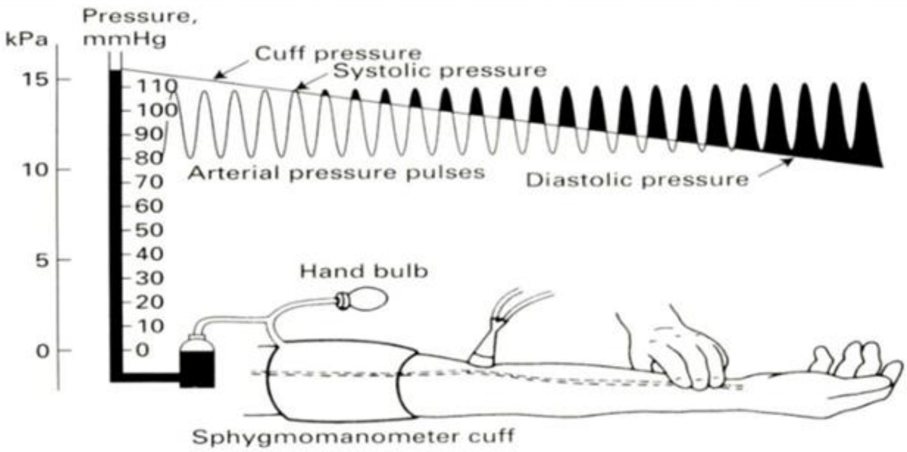


Figure 4 The sphygmomanometer cuff is inflated by a hand bulb to pressure above the systolic level. Pressure is then slowly released, and blood flow under the cuff is monitored by a microphone or stethoscope placed over a downstream artery. The first Korotkoff sound detected indicated systolic pressure, whereas the transition from muffling to silence brackets diastolic pressure.

Since the normal respiration and vasomotor waves modulate the normal blood-pressure levels, several measurements should be taken when employing the palpation and auscultatory techniques. The disadvantage of this technique is failing to give accurate pressures for infants and hypotensive patients.

To obtain accurate results, the clinician should use the correct size of the occlusive cuff. The pressure applied to the artery wall is assumed to be equal to that of the external cuff. Generally, the width of the cuff should be about 0.40 times the circumference of the extremity. The cuff should be positioned over the artery of interest, and 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. The failure of the auscultation technique for hypotensive patients may be due to low sensitivity of the human ear to these low-frequency vibrations.

A number of techniques have been proposed to measure *automatically and indirectly* the systolic and diastolic blood pressure in humans. 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. 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 mmHg/s) (2.7 to 4 kPa/s) inflation of the occlusive cuff to a preset pressure about 30mmHg 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 3mmHg/s) (0.3 to 0.4kPa/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.

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 5 shows the placement of the compression cuff over two small transmitting and receiving ultrasound crystals (8 MHz) on the arm. 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. 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 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.

Ultrasonic Based Blood Pressure Measurement...

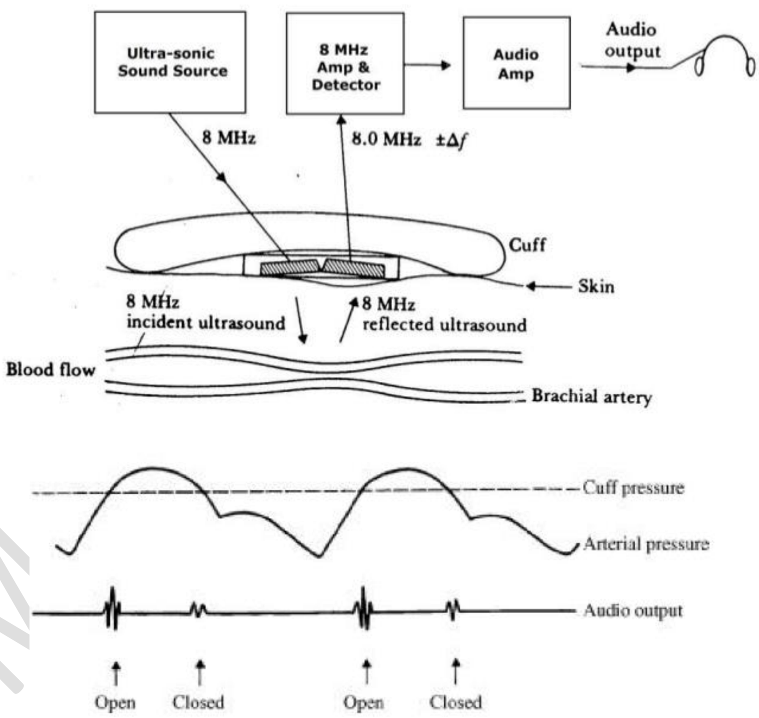


Figure 5 The Ultrasonic determination of blood pressure A compression cuff is placed over the transmitting (8 MHz) and receiving (8 MHz ±Δ*f*) crystals. The opening and closing of the blood vessel are detected as the applied cuff pressure is varied.

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 <u>cuff pressure is equal to mean arterial pressure</u>. 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. Thus, oscillometric monitors employ proprietary algorithms to estimate the diastolic pressure.

Figure 6 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 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 6 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.

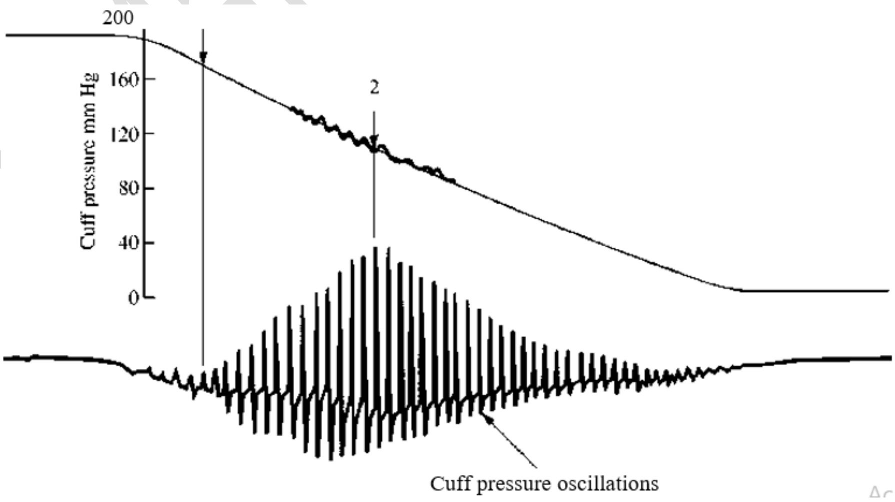


Figure 6 The oscillometric method A compression cuff is inflated above systolic pressure and slowly deflated. Systolic pressure is detected (point 1) where there is a transition from small amplitude oscillations (above systolic pressure) to increasing cuff-pressure amplitude. The cuff-pressure oscillations increase to a maximum (point 2) at the mean arterial pressure.

The system description begins with the blood-pressure cuff, which compresses a limb and its vasculature by the encircling inflatable compression cuff pressures. The cuff is connected to a pneumatic system (see Figure 7). 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.

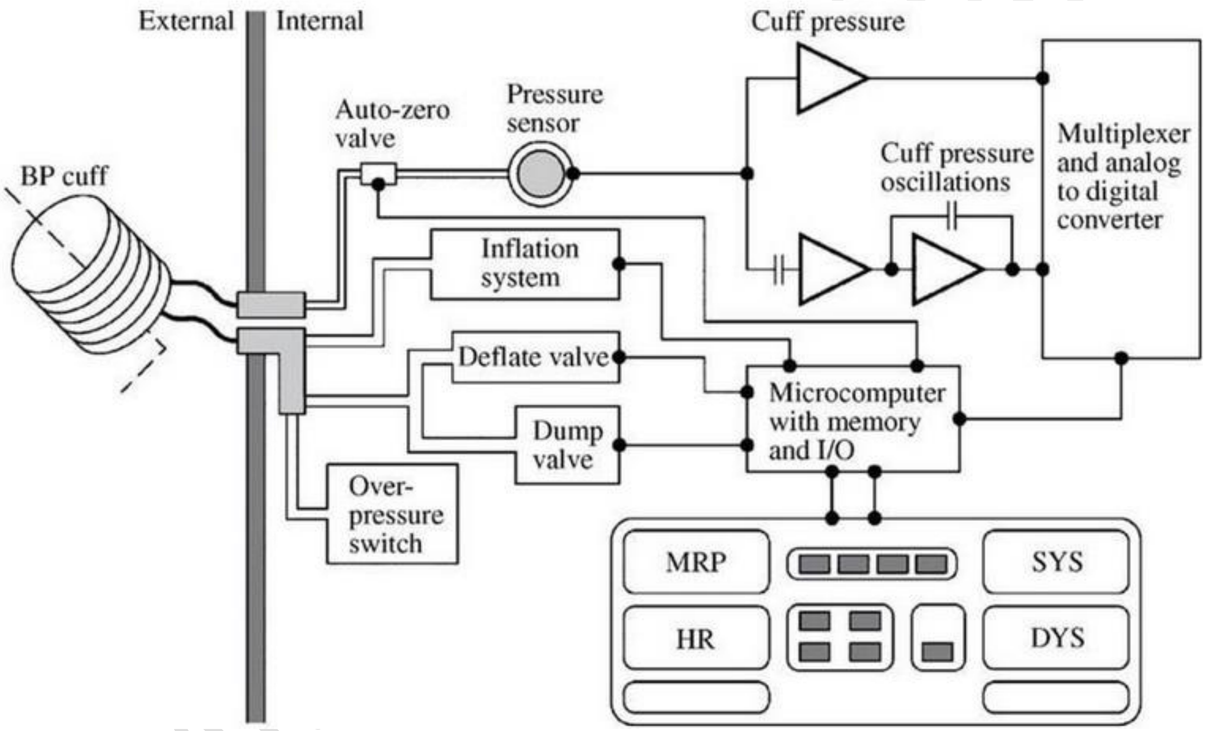


Figure 7 Block diagram of the major components and subsystems of an oscillometric blood-pressure monitoring device, based on the Dinamap unit, I/O = input/output; MAP = mean arterial pressure; HR = heart rate; SYS = systolic pressure; DYS = diastolic pressure.