Sahand University of Technology

Lecture 5
BLOOD PRESSURE AND SOUND

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Summer 2016
Determining an individual's **blood pressure** is a standard clinical measurement.

A number of **direct (invasive)** and **indirect (noninvasive)** techniques are being used to measure blood pressure in the human.

**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.

**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.

The transportation is made possible by a “pressurized vessel” system (the arteries, veins, arterioles, venules and capillaries).

Figure 7.1  The left ventricle ejects blood into the systemic circulatory system. The right ventricle ejects blood into the pulmonary circulatory system.
Figure 7.2  Typical values of circulatory pressures SP is the systolic pressure, DP the diastolic pressure, and MP the mean pressure.
Blood Pressure and Resistance

Pulse pressure (PP) = SP - DP
Mean pressure (MP) = DP + PP/3
• Blood pressure is an important measure of the state of the heart and blood vessels.

• It is defined as the force the blood exerts against the blood vessel walls.

• The pumping action of the heart generates a blood flow through the vessels.

• When this flow is met by resistance from vessel walls, blood pressure results.
Blood Pressure Measurement

Stephen Hales, Year 1733
Blood pressure measurement (2 main approaches)

1-Direct measurement
Truly measure the pressure inside the vessel.
Catheter

2-Indirect measurement
Using sound or vibration on the skin surface to monitor (guess) the SBP (systolic pressure) and DBP (diastolic pressure).
e.g. Auscultatory method, Oscillometric method

The actual pressure sensor can be:
- Strain gage
- Variable inductance
- Variable capacitance
- Optoelectronic
- Piezoelectric, etc...
Direct method

Direct measurement = Invasive measurement

• A vessel is punctured and a catheter (a flexible tube) is guided in

• The most common sites are brachial and radial arteries but also other sites can be used e.g. (Femoral artery)

• Direct blood-pressure sensor systems can be divided into two general categories extravascular and intravascular sensor systems

• This method is precise but it is also a complex procedure involving many risks.

• Used only when essential to determine the blood pressure continuously and accurately in dynamic circumstances
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 sensor is located behind the catheter and the vascular pressure is transmitted via this liquid-filled catheter.

**Advantage**: Good Stability

**Disadvantage**: The frequency response of the system is limited by the hydraulic properties, in particular the low-pass filter effect of the tubing system.
Direct method

- Catheter connected to a pressure sensor through 3-way stopcock

- System is filled with saline-heparin solution (anticoagulant agent), must be flushed every few minutes

- Catheter inserted through surgical cut down or percutaneous insertion

- BP info is transmitted via the catheter fluid to the sensor diaphragm

- (why we use water, but not air?)

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• A thin flexible metal diaphragm is stretched across the opening of the transducer top.

• The diaphragm is connected to an inductive bridge (or resistive Wheatstone bridge) strain gauge which flexes the strain gauge an amount proportional to the applied pressure.

• A clear plastic dome, filled with fluid sits atop the diaphragm and provides the hydraulic coupling / connection to the catheter. Electrical connector typically houses the bridge circuit.
Blood Pressure: Strain gage/ Wheatstone Bridge

If $R_1$ and $R_3$ increase and $R_2$ and $R_4$ decrease by $\Delta R$, then

$$\Delta V_0 = 0 \iff \frac{R_1}{R_2} = \frac{R_4}{R_3}$$

$$\Delta v_0 = \frac{\Delta R}{R_0} v_i$$
Lumped-parameter model for the catheter and sensor

- The liquid filled catheter-sensor is a hydraulic system that can be best modeled by distributed parameters.
- 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.

- Analogous elements for hydraulic inertance, resistance, and compliance are electric inductance, resistance, and capacitance, respectively.
The liquid resistance $R_c$ of the catheter is due to friction between shearing molecules flowing through the catheter.

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

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

$p = \text{pressure difference across the segment in Pa (pascal = N/m}^2\text{)}$

$F = \text{flow rate, m}^3/\text{s}$

$\bar{u} = \text{average velocity, m/s}$

$A = \text{cross-section area, m}^2$

The equation applies for laminar or Poiseuille flow. It is

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

Catheter length L (m), radius r(m) and liquid viscosity q (Pa.S).
The liquid inertance LC of the catheter is due primarily to the mass of the liquid. It can be represented by the equation

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

where $a =$ acceleration, m/s$^2$

$$L_c = \frac{\Delta P}{aA} \quad \Rightarrow \quad L_c = \frac{m}{A^2} \quad \Rightarrow \quad L_c = \frac{\rho L}{\pi r^2}$$

where $m =$ mass of liquid (kg) and $\rho =$ density of liquid (kg/m$^3$)

These two 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.
The compliance $Cd$ of the sensor diaphragm is given by the equation:

$$Cd = \frac{\Delta V}{\Delta P} = \frac{1}{Ed}$$

where $Ed$ 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,

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

Using the general form of a second-order system equation, we can show that the natural undamped frequency is and the damping ratio is

$$\omega_n = \frac{1}{\sqrt{(R_c/2)(C_d/L_c)^{1/2}}}$$

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

$$f_n = \frac{r}{2} \left( \frac{1}{\pi \rho L} \frac{\Delta P}{\Delta V} \right)^{1/2}$$
A number of useful relationships and pertinent constant:

Table 7.1 Mechanical Characteristics of Fluids

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Substance</th>
<th>Temperature</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\eta$</td>
<td>Water</td>
<td>20 °C</td>
<td>0.001 Pa·s</td>
</tr>
<tr>
<td>$\eta$</td>
<td>Water</td>
<td>37 °C</td>
<td>0.0007 Pa·s</td>
</tr>
<tr>
<td>$\eta$</td>
<td>Air</td>
<td>20 °C</td>
<td>0.000018 Pa·s</td>
</tr>
<tr>
<td>$\rho$</td>
<td>Air</td>
<td>20 °C</td>
<td>1.21 kg/m³</td>
</tr>
<tr>
<td>$\Delta V/\Delta P$</td>
<td>Water</td>
<td>20 °C</td>
<td>$0.53 \times 10^{-15} \text{m}^5/\text{N per ml volume}$</td>
</tr>
<tr>
<td>$\eta$</td>
<td>Blood</td>
<td>All</td>
<td>$\approx 4 \times \eta$ for water</td>
</tr>
</tbody>
</table>

We can study the transient response and the frequency response of the catheter-sensor system by means of the analogous electric circuit.
Read EXAMPLE 7.1:

The frequency response for the catheter-sensor system (Statham P23Dd sensor)

Figure 7.10 Frequency-response curves for catheter-sensor system with and without bubbles. Natural frequency decreases from 91 Hz to 22 Hz and damping ratio increases from 0.033 to 0.137 with the bubble present.
(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 $R_{cd}$. 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 $L_c$.

*Read relations and equations.*
Using an intravascular system eliminates the entire plumbing system, by making the measurement at the site!
1- Eliminates the time delay introduced by the tubing system
2- Allows high fidelity measurement of the high frequency components of the BP signal

Typical sensors used:
1- Various types of strain gauges bonded onto a flexible diaphragm at the catheter tip
2- Fiber optic systems where the displacement measurement of the diaphragm is made optically
Strain Gauges

Metal foil strain gauge

Temperature compensated strain-gauge

Thin-film aluminum strain gauges

Semiconductor strain gauge
Gages of this type are available in the F 5 catheter (1.67 mm OD) size. In the French scale (F), used to denote the diameter of catheters, each unit is approximately equal to 0.33 mm.

**Advantage:**
- Small size

**Disadvantages:**
- Problems of temperature
- Electric drift
- Fragility
- Nondestructive sterilization
- More expensive
- Break after
- High cost per use.
Intravascular fiber-optic pressure sensor

The fiber-optic intravascular pressure sensor can be made in sizes comparable to those described above, but at a lower cost.

(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.

Because he left slope region where the characteristic is steepest.
Intravascular fiber-optic pressure sensor

Sensing interferometer path length difference $\delta = 2L_0$

Cavity length $L_0$ of the Fabry-Perot sensing interferometer

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The curvature of the skin surface is flattened when equal pressure exists on both sides of the membrane,

Monitoring of the probe pressure determines the dura pressure.

The membrane position is determined by a reflector that is attached to the membrane and 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.

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.
Fiber optic pressure sensor

Advantages:
• These devices are inherently safer electrically.
• The frequency response is not limited by the hydraulic properties of the system.
• No time delay.
• The fiber-optic intravascular pressure sensor can be made in sizes comparable to catheter-tip sensors, but at a lower cost.

Disadvantages:
• 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.
Disposable pressure sensors

• Disposable sensors decrease the risk of patient cross-contamination and reduce the amount of handling by hospital personnel.

• Reusable pressure sensors are subject to the abuses of reprocessing and repeated user handling, they tend to be less reliable than disposable sensors.

**Cheaper and more reliable than reusable pressure sensors**
Disposable pressure sensors

Micromachining silicon:
A pressure diaphragm is etched and photo-resistors are diffused into the diaphragm for measuring its displacement.

This process results in
• a small,
• Integrated,
• Sensitive,
• and relatively inexpensive pressure sensor.
• Incorporated into a disposable pressure-monitoring tubing system.
• Contains a thick-film resistor network that is laser-trimmed to remove offset voltages and set the same sensitivity for similar disposable sensors.
• thick-film thermistor network is usually incorporated for temperature compensation

Application:
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.
Harmonic analysis of blood-pressure waveforms

The basic sine-wave components of any complex time-varying periodic wave-form can be dissected into an infinite sum of properly weighted sine and cosine functions of the proper frequency.

The blood-pressure pulse can be divided into its fundamental component (of the same frequency as the blood-pressure wave) and its significant harmonics.

Compare the original wave-form and the waveform reconstructed from the Fourier components.

The first six harmonics of the blood-pressure waveform. The table gives relative values for amplitudes.
The response characteristics of a catheter-sensor system can be determined by two methods:

1- Measuring the transient step response for the system:
   The simplest and most straightforward technique

2- Measuring the frequency response of the system:
   A potentially more accurate method but a more complicated one because it requires special equipment-involves.
The transient-response method

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.
The transient-response

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).

\[
\Lambda = \ln \frac{y_n}{y_{n+1}}
\]

\[
\varepsilon = \frac{\Lambda}{\sqrt{4\pi^2 + \Lambda^2}}
\]

\[
\omega_n = \frac{2\pi}{\left[T(1 - \varepsilon^2)^{1/2}\right]}
\]
The sinusoidal frequency-response method is more complex because it requires more specialized equipment.

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.
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.
The values of the damping ratio $\zeta$ and natural frequency $\omega_n$ are functions of the various system parameters.

• Minute air bubbles, which increase the compliance of the catheter manometer system, drastically decreased the damped natural frequency $\omega_d = 2\pi/T$.

Exp.: 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 are inversely related to the damped natural frequency for Teflon and polyethylene catheters for the diameters tested (0.58 to 2.69 mm).

• The linear relationship between the damped natural frequency and $1/(\text{catheter length})^{1/2}$ experimental errors.

• A linear relationship 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:
- Because Teflon is slightly stiffer than polyethylene, it has a slightly higher frequency response at any given length.
- 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.

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. The fewest possible number of connectors, tight-fitting of connectors and a water seal are suggested.

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.
Bandwidth requirements are a function of the investigation.

e.g.: 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.

e.g.: The bandwidth requirements for a heart rate of 120 beats/min (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 wave-shape can be preserved, however, even if the phase characteristics are not ideal, the waveform is delayed in time, depending on the phase shift.

• Measurements of the derivative of the pressure signal increase bandwidth requirements, because the differentiation of a sinusoidal harmonic increases the amplitude of that component by a factor proportional to frequency.
**Typical 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. (The amplitude of the higher-frequency components are amplified)

(c) Overdamped response that shows a significant time delay and an attenuated amplitude response. (higher-frequency components are attenuate)
Typical pressure-waveform distortion

• The underdamped 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 mm Hg (-2 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 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 $\zeta$ and has little effect on the natural frequency. (See Problem 7.6.)
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 the use of **stiff catheters** or by **careful placement** of **catheters** in regions of **low flow velocity**.
• 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 cmH2O (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 hypevolemic states, or
- circulatory failure.

It is used as a guide to determine the amount of liquid a patient should receive.

*Read the measurement method*

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 cmH2O (0 to 1.2 kPa), with a mean pressure of 5 cmH2O (0.5 kPa).

• Esophageal manometry uses a similar low-pressure catheter system.
• This test will tell your doctor if your esophagus is able to move food to your stomach normally.
The cardiac-catheterization procedure is 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 crucial in defining the timing, risks, and anticipated benefit for a given patient.

Catheterization procedures are performed in specialized laboratories outfitted with x-ray (fluoroscopy) equipment for visualizing heart structures and the position of various pressure catheter (injection of radiopaque dye).

In addition, measurements are made of cardiac output, blood and respiratory gases, blood-oxygen saturation, and metabolic product.

Clinicians can also use balloon-tipped, flow-directed catheters without fluoroscopy.
Applications:
- Measure pressures across the four valves to determine the valves pressure gradients.
- Measure cardiac output using the principle of thermodilution (using dye dilution, the Fick method, and impedance cardiography).
- Blood samples such as $O_2$ and $CO_2$ (shunts detection).
- Catheterization for angiographic visualization
- Enlarge the lumen of stenotic coronary arteries
- Assess valvular stenosis

During heart catheterization (because of mechanical stimulus, ectopic beats and/or cardiac fibrillation) frequently occur. For this reason, clinicians must have a functional defibrillator.
(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.
Areas of a valve orifice can be calculated from basic fluid-mechanics equations.

Bernoulli's equation for frictionless flow:

\[ P_t = P + \rho gh + \frac{\rho u^2}{2} \]

- \( P_t \): fluid total pressure
- \( P \): local fluid static pressure
- \( \rho \): fluid density
- \( g \): acceleration of gravity
- \( h \): height above reference level
- \( u \): fluid velocity

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.
Calculation of areas of a valve orifice

Assume:
- Frictionless flow for the model,
- Equate total pressures at locations 1 and 2.
- The difference in heights is negligible
- The velocity at location 1 is negligible compared with \( u \), the velocity at location 2.

Then:

\[
P_1 - P_2 = \frac{\rho u^2}{2}
\]

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

At location 2, the flow \( F = Au \), where \( A \) is the area. Hence

\[
A = \frac{F}{u} = F \left( \frac{\rho}{2(P_1 - P_2)} \right)^{1/2}
\]
In practice, there are losses due to friction, and the minimal flow area is smaller than the orifice area. Hence $A$ becomes:

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

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).

Read EXAMPLE 7.3.
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.

- Based on Bernoulli's equation, 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.

$$P_t = P + \rho gh + \frac{\rho u^2}{2}$$
The effect of the potential-energy

We first examine the effect of the potential-energy term on the static pressure of the fluid:

**No corrections** need be made for the potential-energy Term while
- The patient in a supine (on-the-back)
- The sensor placed at heart level

Other Positions:
- However, when the patient is sitting or standing, the long columns of blood in the arterial and venous pressure systems contribute a hydrostatic pressure, $\rho gh$.
- 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 mmHg (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 1.31-cm increase in height of the source, 1.0 mm Hg (133 Pa) must be added to the sensor reading.
Convenient reference point

Phlebostatic axis

4th intercostal space

Outermost anterior chest

Midchest

Outermost posterior chest

Reference point
The kinetic-energy term $\frac{\rho u^2}{2}$ 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:
1- side (static) pressure
2- end (total) pressure

"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.

- If the catheter pressure port faces upstream, the recorded pressure is the side pressure plus the additional kinetic-energy term $\frac{\rho u^2}{2}$.
- If the catheter pressure port faces downstream, the value is approximately less than the side pressure.
- When the catheter is not positioned correctly, artifacts may develop in the pressure reading.
The effect of the kinetic-energy

<table>
<thead>
<tr>
<th>Vessel</th>
<th>Vel (cm/s)</th>
<th>KE (mm Hg)</th>
<th>Systolic (mm Hg)</th>
<th>% KE of Total</th>
</tr>
</thead>
<tbody>
<tr>
<td>Aorta (systolic)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>At rest</td>
<td>100</td>
<td>4</td>
<td>120</td>
<td>(16)</td>
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<tr>
<td>Cardiac output at 3 × rest</td>
<td>300</td>
<td>36</td>
<td>180</td>
<td>(24)</td>
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<td>Brachial artery</td>
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<tr>
<td>At rest</td>
<td>30</td>
<td>0.35</td>
<td>110</td>
<td>(14.7)</td>
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<tr>
<td>Cardiac output at 3 × rest</td>
<td>90</td>
<td>4</td>
<td>120</td>
<td>(16)</td>
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<tr>
<td>Venae cavae</td>
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<td></td>
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</tr>
<tr>
<td>At rest</td>
<td>30</td>
<td>0.35</td>
<td>2</td>
<td>(0.3)</td>
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<tr>
<td>Cardiac output at 3 × rest</td>
<td>90</td>
<td>3.2</td>
<td>3</td>
<td>(0.4)</td>
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<td>Pulmonary artery</td>
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<td></td>
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<tr>
<td>At rest</td>
<td>90</td>
<td>3</td>
<td>20</td>
<td>(2.7)</td>
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<tr>
<td>Cardiac output at 3 × rest</td>
<td>270</td>
<td>27</td>
<td>25</td>
<td>(3.3)</td>
</tr>
</tbody>
</table>

The effect of the kinetic-energy

- As 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.
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.
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 (2-3 mm Hg/s), 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.
Advantages:
- The auscultatory technique is simple and requires a minimum of equipment.

Disadvantages:
- The auscultatory cannot be used in a noisy environment (whereas the palpation technique).
- In employing the palpation and auscultatory techniques, you should take several measurements, 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.
- The observations differ from observer to another.
- The observations do not always correspond with intra-arterial Pressure.
- A mechanical error might be introduced into the system e.g. Mercury leakage, air leakage, obstruction in the cuff etc.
Hints:

- Hearing acuity of the user must be good for low frequencies from 20 to 300 Hz.

- Using an **occlusive cuff** of the correct size.

- The pressure applied to the artery wall is assumed **equal** to that of the external cuff. The cuff pressure is **transmitted** across interposed tissue.

- The failure of the auscultation technique for hypotensive patients may be due to low sensitivity of the human ear to these low-frequency vibrations.
Automatic indirect blood-pressure measurement

Three of the commonly used automatic indirect blood-pressure measurement:

Method 1:

- Using microphone instead of stethoscope.
- A rapid (20-30 mm Hg/s) (2.7-4 kPa/s) inflation about 30 mm Hg higher,
- The flow of blood beneath the cuff is stopped by the collapse of the
- Reduce cuff pressure slowly (2-3 mm Hg/s)
- The first Korotkoff sound is detected by the microphone.
- The muffling and silent period of the Korotkoff sounds is detected
- The instrument displays the systolic and diastolic
Method 2:

- The ultrasonic determination,
- A transcutaneous Doppler sensor that detects the motion of the blood-vessel walls,
- The reflected signal (shifted in frequency) is detected by the receiving crystal,
- 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.

A compression cuff is placed over the transmitting (8 MHz) and receiving (8 MHz ± $\Delta f$) crystals. The opening and closing of the blood vessel are detected as the applied cuff pressure is varied.
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.
Advantages:

- It can be used with infants and hypotensive individuals.
- It can be used in high-noise environments.
- Complete reconstruction of the arterial-pulse waveform is also possible via the ultrasonic method.

Disadvantage

That movements of the subject's body cause changes in the ultrasonic path between the sensor and the blood vessel.
Oscillometric pressure measurement

- A non-invasive blood pressure technique,
- Measures the amplitude of oscillations

- These oscillations appear in the cuff-pressure signal which are created by expansion of the arterial wall each time blood is forced through the artery.

- 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 bleed 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.

- 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 (Ramsey 1991).

- This property usually allows mean arterial pressure to be measured reliably even in case of hypotension with vasoconstriction and diminished pulse pressure.
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 cuff is connected to a pneumatic system

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.
• The basic principle: 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 (Goldmann (1957) applanation tonometer, clinical standard)

• With this technique, the investigator measures the force required to flatten a specific optically determined area.
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.
• The second and third component, the system that continuously monitors the applanation and the status of the curvature of the cornea, determines the occurrence of applanation with microsecond resolution.

Two obliquely oriented tubes are used to detect applanation.

1. Transmitter tube T directs a collimated beam of light at the corneal vertex.
2. A telecentric receiver R observes the same area.

Forbes et al. (1974) developed an applanation tonometer without touching the eye. An air pulse of linearly increasing force deforms and flattens the central area of the cornea (within a few ms.)
In the case of the **undisturbed cornea**, the detector receives little or no light.

As the cornea's convexity is reduced to the **flattened** condition, the **amount** of **light** detected is increased.

When the cornea is applanated, it acts like a piano-mirror with a resulting maximal detected signal.

- 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.
Monitoring system for noncontact applanation tonometer
The principles of operation of the arterial tonometry are very similar to those for the ocular tonometry, discussed above.

- **Continuous measurements** of arterial pressure throughout the total heart cycle.
- The instrument sensor is placed over a superficial artery that is **supported from below by bone** (The radial artery at the wrist).
Disadvantage:
• The arterial tonometer suffers from relatively high cost when compared to a conventional sphygmomanometer.

Advantage:
• One significant advantage of the arterial tonometer is its ability to make noninvasive, nonpainful, continuous measurements for long periods of time.
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 $P$. 
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.
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.

Also:
The hold-down force $F_1$ 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.
Arterial tonometry (multiple element sensors)

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.
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.
Schematic illustrations and photographs of a thin conformable piezoelectric pressure sensor.
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**.

*Heart sounds* are **vibrations** or **sounds** due to the **acceleration** or **deceleration** of **blood**, whereas *murmurs* are **vibrations** or **sounds** due to **blood turbulence**.
Heart sounds: Mechanism and Origin

Correlation of the four heart sounds with 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.

  **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).

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 Murmurs

The sources of murmurs (in French Souffle), developed by turbulence in rapidly moving blood, are known.

Common murmurs:
Murmurs during the early systolic phase are common in children, and they are normally heard in nearly all adults after exercise.

Abnormal murmurs:
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 wave-like motion occurs in the **most compressible tissues**, such as the **lungs and fat**.

There are **optimal** recording **sites** for the various heart sounds.

**Auscultatory areas on the chest**
A, aortic; P, pulmonary; T, tricuspid; and M, mitral areas.
Heart sounds and murmurs have extremely small amplitudes, with frequencies from 0.1 to 2000 Hz.

Two difficulties may result:

1- At the low end of the spectrum (below about 20 Hz), the amplitude of heart sounds is below the threshold of audibility.

2- 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 would not.
• 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.
**Stethoscopes**

Stethoscopes are used to transmit heart sounds from the chest wall to the human ear.

The mechanical stethoscope amplifies sound because of a standing wave phenomenon that occurs at quarter wavelengths of the sound.

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.

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

When the stethoscope chest piece is firmly applied chest piece, 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
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.
A **phonocardiogram** is a recording of the heart sounds and murmurs.
Phonocardiography

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.