Chapter 7 Blood Pressure and Sound

- Cardiovascular system: Fig. 4.12, Fig. 7.15, Fig. 7.1, Fig. 7.2
- Blood circulation: to transport oxygen and other nutrients to the tissues and carry metabolic waste products away from cells
- Blood pressure measurement: to determine the functional integrity of the cardiovascular system
- Sound: fluctuations in pressure recorded over audio frequency, vibration due to acceleration and deceleration of blood

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

- Sensors: strain gage, LVDT, variable inductance, variable capacitance, optoelectric, piezoelectric, semiconductor

*Extravascular Sensors*
- Catheter-sensor system: liquid-filled catheter (saline-heparin), three-way stopcock, pressure sensor (usually Si diaphragm with piezoresistive sensor), flush solution (Fig. 7.3, Fig. 2.2, Fig. 14.15)
- Catheter insertion: surgical cut-down or percutaneous insertion

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

- Catheter-tip sensor: high-frequency response without time delay
- Strain gage with F5 catheter (F: French scale, 1 F ≈ 0.33 mm OD): expensive
- Fiber-optic intravascular pressure sensor: optical measurement of diaphragm deflection, electrical safe, and usually cheaper
- Fiber-optic microtip sensor: Fig. 7.4, requires extra efforts for relative pressure measurement
- Fiber-optic intracranial pressure sensor: Fig. 7.5, pneumatic servo system controls the air pressure within the sensor until the diaphragm is flat (null-balanced mode)
- Catheter-tip micropressure sensor (micro silicone pressure sensor chips): Si diaphragm and piezoelectric strain gage

**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.
Disposable Pressure Sensors

- Lower risk of patient contamination
- No repeated user handling by hospital personnel
- Micromachined Si diaphragm with piezoresistive strain gage
- Laser trimmed thick-film resistor network: offset voltage, sensitivity, and temperature compensation
- High resistance bridge to reduce self heating: an amplifier with high input impedance is required
7.2 Harmonic Analysis of Blood Pressure Waveforms

- Fourier analysis: Fig. 7.6, quantitative representation of a physiologic waveform, fundamental frequency, significant harmonics
- Frequency representation of arterial pulse waveform

![Image](http://ejwoo.com)

**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.)

7.3 Dynamic Properties of Pressure Measurement Systems

- Understanding of the dynamic properties of pressure measurement system is requires for the correct interpretation of the measured pressure waveforms.
- Distributed parameter model: Fig. 7.7
- Lumped parameter model: Fig. 7.8
Analogous Electric Systems

- Model simplification from Fig. 7.7 to Fig. 7.8
- Electric and fluid mechanics analogy
- Variables

<table>
<thead>
<tr>
<th>Electric System</th>
<th>Fluid Mechanics</th>
</tr>
</thead>
<tbody>
<tr>
<td>Voltage or potential difference, $V$ [V]</td>
<td>Pressure difference, $P$ [Pa] or [mmHg]</td>
</tr>
<tr>
<td>Current, $I$ [A]</td>
<td>Flow, $F$ [m$^3$/s] or [L/min]</td>
</tr>
<tr>
<td>Charge, $Q$ [C]</td>
<td>Volume, $V$ [m$^3$] or [L]</td>
</tr>
</tbody>
</table>

- Physical properties

<table>
<thead>
<tr>
<th>Electric System</th>
<th>Fluid Mechanics</th>
</tr>
</thead>
<tbody>
<tr>
<td>Resistance, $R = \frac{V}{I}$ [$\Omega$]</td>
<td>Liquid resistance, $R = \frac{\Delta P}{F}$ [Pa·s / m$^3$]</td>
</tr>
<tr>
<td>Inertance, $L = \frac{V}{dI/dt}$ [H] or [V·s / A]</td>
<td>Inertance, $L = \frac{\Delta P}{dF/dt} = \frac{\Delta P}{aA}$ [Pa·s$^2$ / m$^3$]</td>
</tr>
<tr>
<td>Capacitance, $C = \frac{Q}{V} = \frac{[Idt]}{V}$ [F] or [C / V]</td>
<td>Compliance, $C = \frac{\Delta V}{\Delta P} = \frac{1}{E_d}$ [m$^5$ / N]</td>
</tr>
</tbody>
</table>
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 inerance 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 inerance \( L_c \) 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 \( L_c \).

- Liquid resistance: due to friction or liquid viscosity, \( \eta \) [Pa·s]
  \[
  R = \frac{8\eta L}{\pi r^4} \text{[Pa·s/m}^3] \quad \text{(from Poiseuille equation for laminar or Poiseuille flow)}
  \]

- Liquid inertance: due mainly to the mass, \( m \) [kg] or density, \( \rho \) [kg/m3]
  \[
  L = \frac{m}{A^2} = \frac{\rho L}{\pi r^2} \text{[Pa·s}^2/m^3] \]

- Compliance: due to elasticity or Young's modulus

- From the equivalent electric circuit model,
  \[
  v_i(t) = L_c C_d \frac{d^2 v_o(t)}{dt^2} + R_c C_d \frac{dv_o(t)}{dt} + v_o(t)
  \]
  \[
  f_n = \frac{\omega_n}{2\pi} = \frac{1}{\sqrt{L_c C_d}} = \frac{1}{2} \sqrt{\frac{1}{\pi \rho L} \frac{\Delta P}{\Delta V}}
  \]
  \[
  \zeta = 4\eta \sqrt{\frac{L(\Delta V/\Delta P)}{\pi \rho}}
  \]

See Example 7.1, Fig. 7.9
7.4 Measurement of System Response

**Transient Step Response**

- Simple
- Pop technique: Fig. 7.10 and Fig. 7.11

- From Eq. (1.38), $\omega_n = \frac{2\pi}{T\sqrt{1-\zeta^2}}$

**Sinusoidal Frequency Response**

- More accurate and complicated
- Fig. 7.12

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**Figure 7.9** 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.
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., Cardiovascular Fluid Dynamics, vol I, New York: Academic Press, 1972.)
7.5 Effects of System Parameters on Response

- **Response:** $\omega_n$ and $\zeta$
- **Air bubble** $\Rightarrow C_c \uparrow \Rightarrow \omega_d = \frac{2\pi}{T}$: for PE-190 with $L=50 \sim 100$ cm, unboiled water reduces the damped natural frequency, $\omega_d$ by 50 ~ 60 % compared to boiled water
- **$L \uparrow \Rightarrow \omega_d \downarrow$**: for Teflon and polyethylene with diameters of 0.58 ~ 2.69 mm, $\omega_d \propto \frac{1}{\sqrt{L}}$
- **For Teflon and polyethylene catheter,** $\omega_d \propto (inner \ diameter)$
- **Teflon is slightly stiffer than polyethylene** $\Rightarrow$ Teflon has slightly higher frequency response
- **Si rubber tubing (high compliance)** $\Rightarrow$ marked decrease in frequency response: do not use rubber connections except for the measurement of mean pressure
- **Minimize any connectors and all connectors must be tight-fitting with a water-seal**
- **Catheter coiling and bending** $\Rightarrow$ insignificant change in frequency response

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**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.
7.6 Bandwidth Requirements for Measuring Blood Pressure

- Measure up to 10th harmonics of blood pressure: for example, if HR is 120 bpm (2 Hz), BW must be at least 20 Hz.
Flat amplitude response with linear phase shift ⇒ distortionless measurement with time delay
If the differentiation of BP waveform is needed, BW must be up to 20th harmonics at least.

7.7 Typical Pressure Waveform Distortion

Fig. 7.13: underdamped and overdamped
Fig. 7.14: air bubble and catheter whip

7.8 Systems for Measuring Venous Pressure

Venous pressure: for determining the function of capillary bed and the right side of the heart
- BP in the small vein ≈ the capillary pressure
- Intrathoracic venous pressure ≈ diastolic filling pressure of the right ventricle
- BP in a central vein or the right atrium ≈ central venous pressure
- Extrathoracic venous pressure ≈ 2 ~ 5 cmH₂O or 0.2 ~ 0.5 kPa
- Reference level for venous pressure is the right atrium
Central venous pressure is an important indicator of myocardial performance.
- Frequently monitored for proper therapy in case of heart dysfunction, shock, hypovolemic or hypervolemic states, or circulatory failure
- A guide to determine the amount of liquid a patient should receive
- Normal central venous pressure: 0 ~ 12 cmH₂O (0 ~ 1.2 kPa) with a mean value of 5 cmH₂O (0.5 kPa)
- It fluctuates above and below atmospheric pressure due to breathing

Procedure
- Percutaneous venous puncture using a large-bore needle
- Insert/advance an intravenous (IV) catheter through the needle into the vein
- Remove the needle
- Attach a plastic tube to IV catheter using a stopcock for drugs or fluid delivery
- Connect IV catheter to pressure sensor with higher sensitivity and lower dynamic range
• Confirm respiratory swings in the waveform

☑ Problems and errors
• Baseline wandering due to the patient movement
• Catheter misplacement
• Blood clot or air bubble
• Impact against a vein wall

☑ Esophageal manometry is similar procedure as the measurement of central venous pressure. It uses a hydraulic capillary infusion of 0.6 mL/min to prevent sealing of the catheter orifice in the esophagus.

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

7.9 Heart Sounds

☑ Auscultation of the heart: clinical information about the functional integrity of the heart
☑ Phonocardiography: temporal relationship between the heart sounds and the mechanical and electrical events of the cardiac cycle, Fig. 7.15
Heart sounds: vibrations or sounds due to the acceleration or deceleration of blood
Murmurs: vibrations or sounds due to blood turbulence

**Mechanism and Origin**

- The first heart sound: blood movement during ventricular systole (blood shift toward the atria) and blood oscillation due to the closure of the AV valves, asynchronous closure of the tricuspid and mitral valves split the first heart sound
- The second heart sound: low frequency vibration associated with the deceleration and reversal of flow in the aorta and pulmonary artery, closure of the semilunar valves, coincident with the completion of the T-wave
- The third heart sound: low frequency and low amplitude, in most children and some adults, sudden termination of the rapid filling phase of the ventricles from the atria and the associated vibration of the relaxed ventricular muscle walls
- The fourth heart sound (atrial heart sound): not audible but recordable, the atrial contraction and propulsion of blood into the ventricles
- Murmurs: turbulence in rapidly moving blood
  - In early systole phase: in most children and adults after exercise
  - Abnormal murmurs: stenosis and leakage of aortic, pulmonary, and mitral valves, note the timing during a cardiac cycle and the location of measurement

![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.)](image)
Auscultation Techniques

- Sound transmission: heart $\Rightarrow$ major blood vessel $\Rightarrow$ body surface, with the largest attenuation in compressible tissues such as the lung and fat, no reflection
- Optimal recording sites: maximal intensity locations, Fig. 7.16 (optimal sites for sounds from four heart valves)
- Sounds and murmurs: very small and $0.1 \sim 2000$ Hz
  - Low frequency ($< 20$ Hz): extremely small and not audible
  - High frequency: audible but hard to record
- Phonocardiography recording: acoustically quiet room, patient movement $\Rightarrow$ baseline wandering

Stethoscopes

- Transmit heart sounds from the chest wall to the human ear, Fig. 7.17
- Variability in interpretation: user's auditory acuity, training, technique of applying the stethoscope
- Mechanical filtering of sounds: contact pressure, bell shape, earpiece fitting, air leakage
- Electronic stethoscope: size, portability, convenience, shape, sound familiarity

![Frequency-Response Curve](http://ejwoo.com)

**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 P. Y. Ertel, M. Lawrence, R. K. Brown, and A. M. Stern, *Stethoscope Acoustics I, "The Doctor and his Stethoscope."* Circulation 34, 1996; by permission of American Heart Association.)
7.10 Phonocardiography

- Phonocardiogram: recording of the heart sounds and murmurs
- Simultaneous recording: ECG, carotid arterial pulse, jugular venous pulse, apex cardiogram
- Carotid arterial pulse, jugular venous pulse, and apex cardiogram: use microphone with 0.1 ~ 100 Hz response

7.11 Cardiac Catheterization

- Cardiac catheterization: hemodynamic function and cardiovascular structure
- For all patients prior to heart surgery
- Cardiac cath lab equipments: X-ray fluoroscopy and angiography (ventriculography, coronary arteriography, pulmonary angiography, aortography), ECG, BP, respiration, cardiac output, blood and respiratory gases, blood-oxygen saturation, metabolic products, defibrillator
- Catheters: two-lumen catheter, Swan-Ganz catheter, dye injection catheters, PTCA catheter
- Valve orifice are estimation: Fig. 7.19

![Figure 7.18](http://ejwoo.com) (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.
7.12 Effects of Potential and Kinetic Energy on Pressure Measurements

- Bernoulli's equation: $P_t = P + \rho gh + \frac{\rho u^2}{2}$ (Eq. 7.8) with $P_t =$ total pressure, $P =$ static pressure (desired BP), $\rho =$ fluid density, $u =$ fluid velocity
- Potential energy term ($\rho gh$)
- Kinetic energy term ($\frac{\rho u^2}{2}$)

7.13 Indirect Measurements of Blood Pressure

- Noninvasive blood pressure (NIBP): intraarterial pressure, Fig. 7.20, cuff deflation of 2 ~ 3 mmHg/s or 0.3 ~ 0.4 kPa/s
- Palpation technique: Rica-Rocci method
- Auscultatory technique: Korotkoff sounds
- Cuff size: cuff width $\approx 40\%$ of the circumference of the extremity
- Cuff location: heart level
- Automatic NIBP
  - Microphone
  - Ultrasound
  - Oscillometric method

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.
Figure 7.20 Typical indirect blood-pressure measurement system The sphygmomanometer cuff is inflated by a hand bulb to pressures 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 indicates 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.)

Figure 7.21 Ultrasonic determination of blood pressure A compression cuff is placed over the transmitting (8 MHz) and receiving (8 MHz ± Δf) crystals. The opening and closing of the blood vessel are detected as the applied cuff pressure is varied. (From H. F. Stegall, M. B. Karedon, and W. T. Kemmerer, "Indirect Measurement of Arterial Blood Pressure by Doppler Ultrasonic Sphygmomanometry," *J. Appl. Physiol.*, 1968,25,793-798. Used with permission.)
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.
7.14 Tonometry

- Force balance technique
- Applanation tonometer: intraocular pressure measurement
- Arterial tonometer

Figure 7.24 Monitoring system for noncontact applanation tonometer  (From M. Forbes, G. Pico, Jr., and B. Frolman, “A Noncontact Applanation Tonometer, Description and clinical Evaluation,” *J. Arch. Ophthalmology*, 1975, 91, 134-140. Copyright © 1975, American Medical Association. Used with permission.)
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. New York: Wiley, 1988, pp.2270-2276.

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. New York: Wiley, 1988, pp. 2270-2276.