EE 4BD4 Lecture 26

Plethysmography

Definition

- Measure volume change over time
- Flow = dvolume/dt

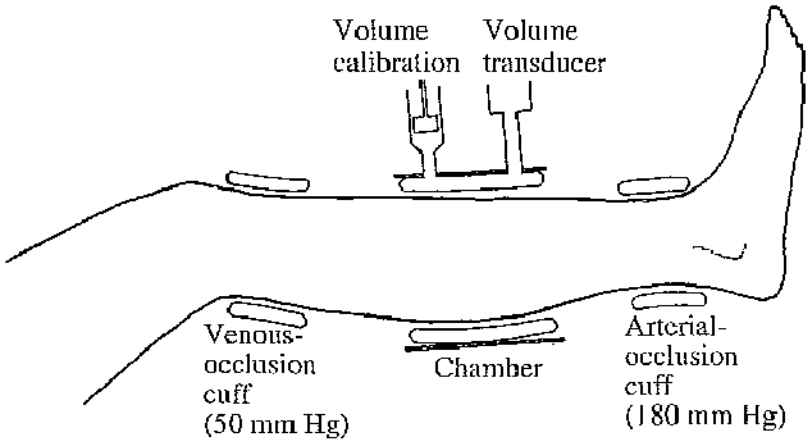


Figure 8.15 In chamber plethysmography, the venous-occlusion cuff is inflated to 50 mm Hg (6.7 kPa), stopping venous return. Arterial flow causes an increase in volume of the leg segment, which the chamber measures. The text explains the purpose of the arterial-occlusion cuff.

Sequence

- Occlude venous return (raise venous cuff pressure to 50 mm Hg
- Allow limb segment to fill from arteries
- Volume of leg increase can be measure by recording increase in air pressure in chamber (air is compressible)
- If cuff completely encloses limb segment inflow is adequately measured
- If cuff only partially encloses limb, arterial pressure cuff used to ensure volume change only due to blood entering limb segment
- When limb pressure exceeds 50 mm Hg change in volume levels off

Results

- After plateau reached venous cuff pressure released
- Slow return indicates venous thrombosis
- Modern plethysmography systems use electrical impedance measurement

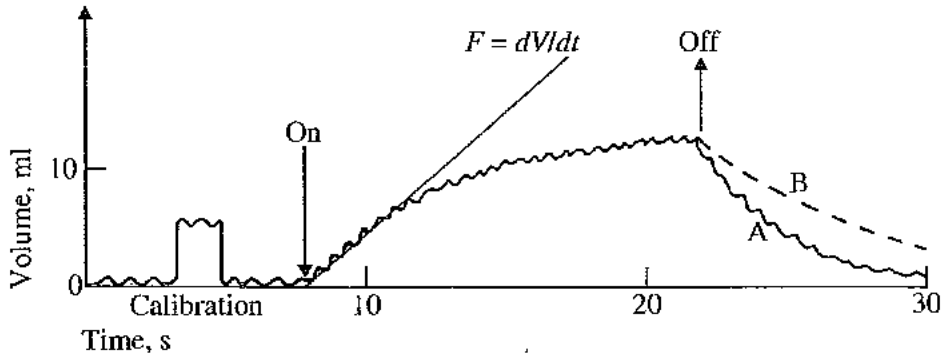


Figure 8.16 After venous-occlusion cuff pressure is turned on, the initial volume-versus-time slope is caused by arterial inflow. After the cuff is released, segment volume rapidly returns to normal (A). If a venous thrombosis blocks the vein, return to normal is slower (B).

Electrical-Impedance Plethysmography

- Nyboer (1970) developed first equations but Swanson (1976) simplified the approach

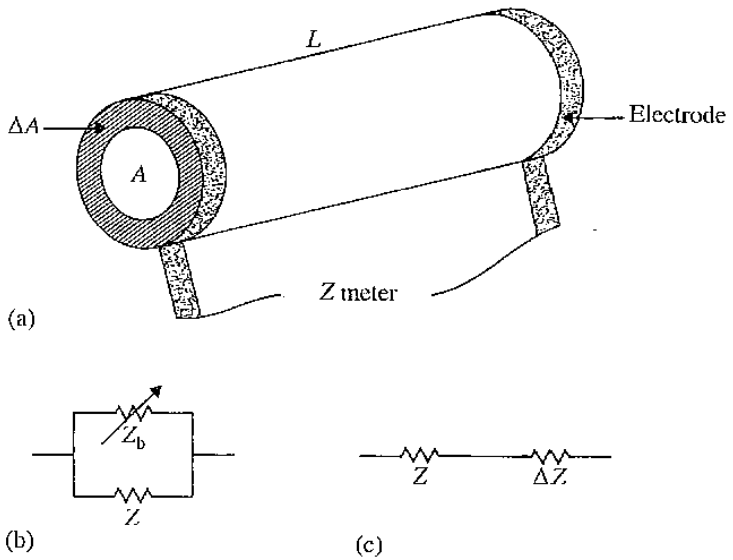


Figure 8.17 (a) A model for impedance plethysmography. A cylindrical limb has length L and cross-sectional area A. With each pressure pulse, A increases by the shaded area ΔA. (b) This causes impedance of the blood, Z_b, to be added in parallel to Z. (c) Usually ΔZ is measured instead of Z_b.

Assumptions and Derivation

- Expansion of arteries is uniform (probably valid in healthy vessels)
- Resistivity of blood ρ_b does not change (in fact decreases with velocity due to alignment of cells with the stream, real at dc but small reactive component at higher frequencies)
- Lines of current are parallel with arteries
- Additional blood flow causes parallel impedance Z_b

$$Z_b = \frac{\rho_b L}{\Delta A}$$

$$\Delta V = L\,\Delta A = \frac{\rho_b L^2}{Z_b}$$

Derivation (cont'd)

- We are really measuring the change ΔZ

$$\Delta Z = \frac{Z Z_b}{Z + Z_b} - Z = \frac{-Z^2}{Z + Z_b}$$

$$Z \ll Z_b, \qquad \frac{1}{Z_b} \cong \frac{-\Delta Z}{Z^2}$$

$$\Delta V = \frac{-\rho_b L^2 \Delta Z}{Z^2}$$

Measurement Considerations

- Need currents > 1 mA so that SNR is maximized
- To avoid electro-stimulation use frequencies > 20 kHz
- Skin electrode impedances decrease with frequency, i.e. two orders of magnitude as frequency goes from 100 to 100 kΩ
- Higher frequencies (> 100 kHz) cause a stray capacitance and tissue impedance becomes complex

Two or Four Electrodes?

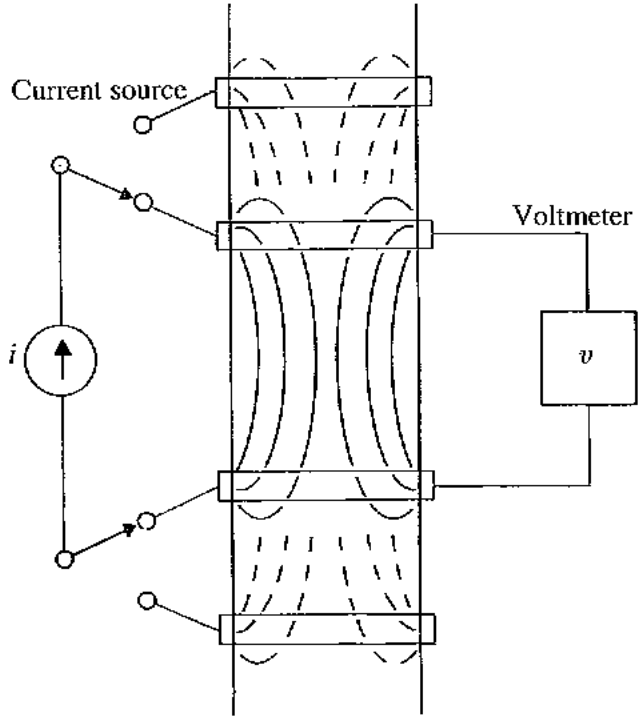


Figure 8.18 In two-electrode impedance plethysmography, switches are in the position shown, resulting in a high current density (solid lines) under voltage-sensing electrodes. In four-electrode impedance plethysmography, switches are thrown to the other position, resulting in a more uniform current density (dashed lines) under voltage-sensing electrodes.

Two or Four Electrodes? (cont'd)

- Current density highest near electrodes so Z at electrodes has highest weight
- Skin electrode impedance changes with heart beat
- Need uniform current flow
- Electrode/electrolyte impedance can cause ΔV at each electrode

Four Electrode Design

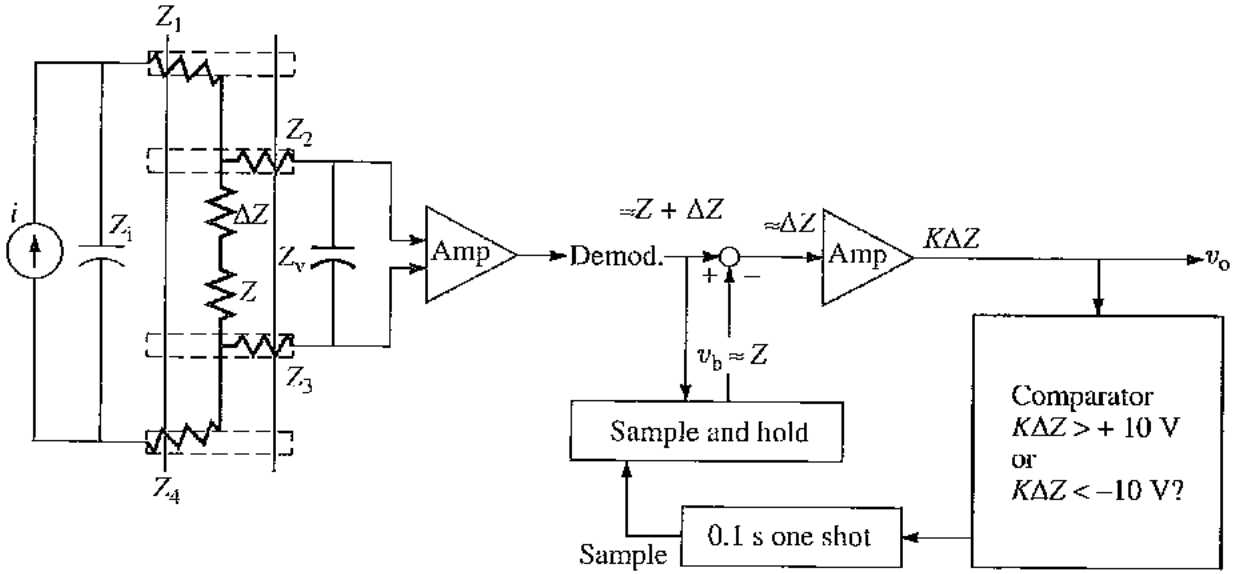
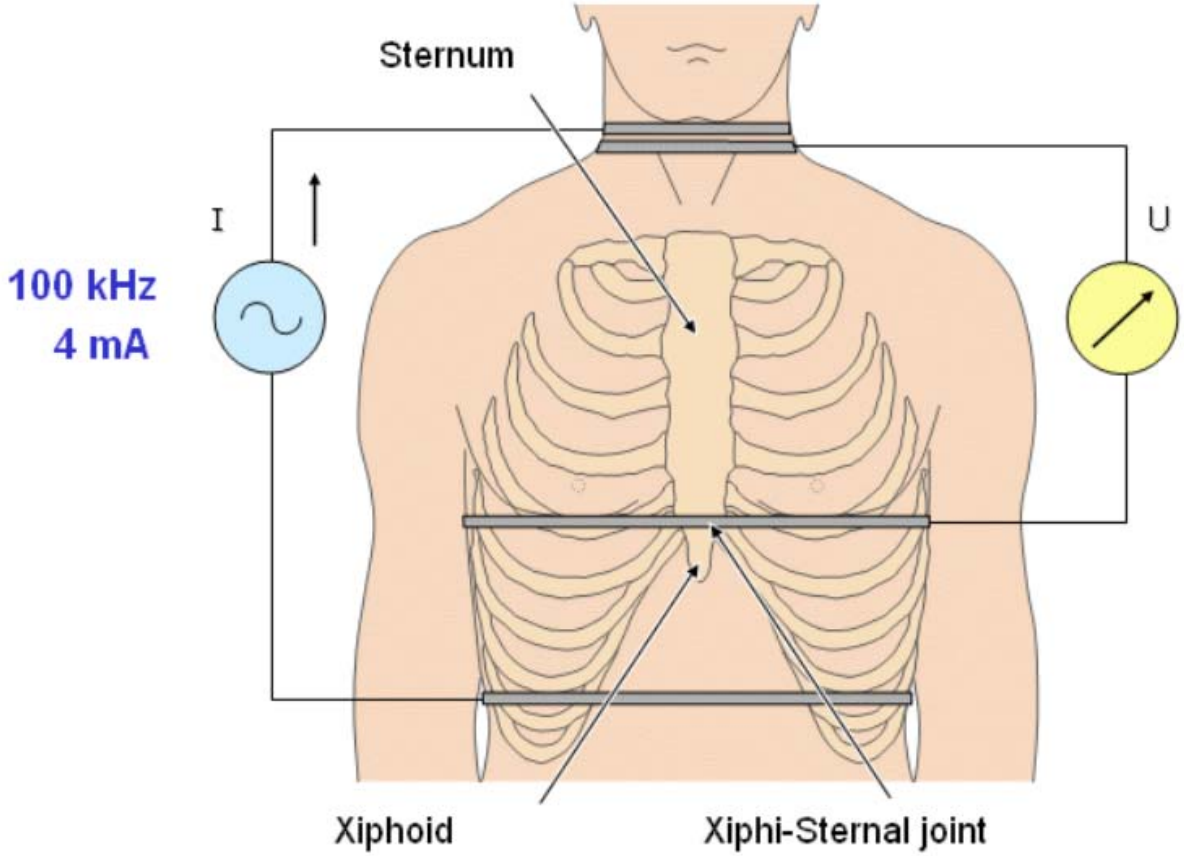


Figure 8.19 In four-electrode impedance plethysmography, current is injected through two outer electrodes, and voltage is sensed between two inner electrodes. Amplification and demodulation yield $Z + \Delta Z$. Normally, a balancing voltage v_b is applied to produce the desired ΔZ. In the automatic-reset system, when saturation of v_o occurs, the comparator commands the sample and hold to sample $Z + \Delta Z$ and hold it as v_b. This resets the input to the final amplifier and v_o to zero. Further changes in ΔZ cause changes in v_o without saturation.

Instrumentation Specifications

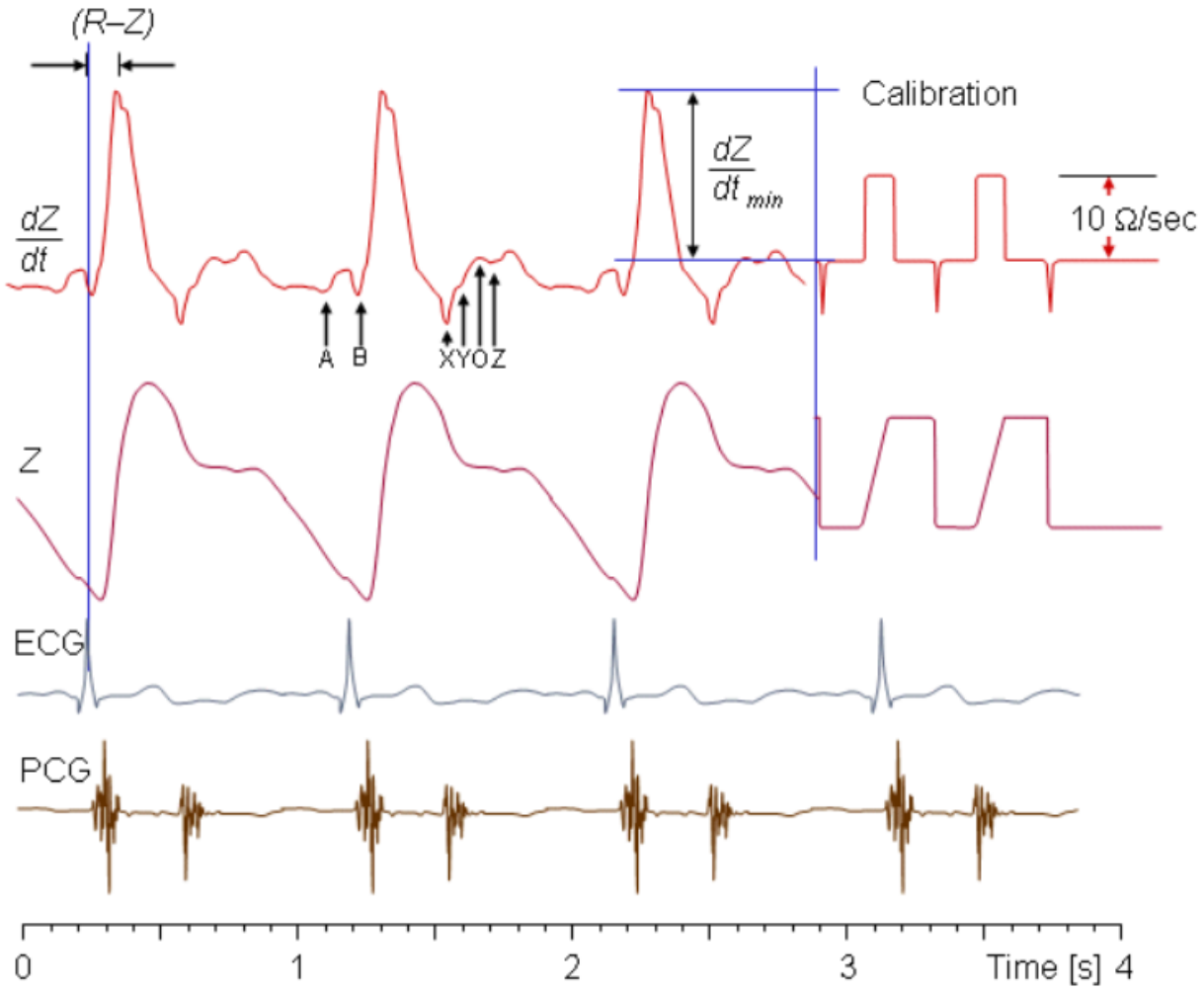
- I is constant current source
- Z_i results from stray and cable capacitance (at 100 kHz, 15 pf causes a stray impedance about 100 kΩ)
- In reality changes in Z_1, Z_4 and ΔZ cause current to divide between Z_i and Z
- ΔZ is small as are changes in Z_1, Z_4 and careful design can make Z_i large (Z_i and Z also 90^o out of phase)
- Z_{amp} is assumed ∞ but Z_v results from cable, stray and amplifier input impedance
- Good design should keep Z_3 and Z_4 constant and Z_v high
- Amplifier gain is modest since measured voltage is relatively high (e.g. 4 mA * 40 Ω = .16 V)
- Demodulator required (typical AM demodulator, rectifier plus low pass filter)

Impedance Cardiography (Instrumentation)



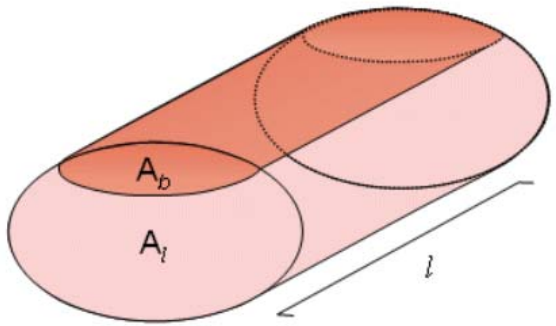
Impedance Calculations

(*Z* is inverted, up indicates lower impedance)



Determining Change in Thoracic Blood Volume

- A_b, A_t, Z_b and Z_t are cross-sectional area of blood volume and thoracic tissue volume and impedance of blood and tissue respectively



$$Z = \frac{Z_b Z_t}{Z_b + Z_t}$$

where Z = longitudinal impedance of the model

Change in Blood Volume (cont'd)

- Impedance change of thorax w.r.t. change in blood volume

$$dZ = \frac{Z^2}{Z_b^2} dZ_b$$

- Impedance of blood volume

$$Z_b = \frac{\rho_b l}{A_b}$$

- Relationship between changes in blood volume and impedance

$$dv_b = d(lA_b) = -\frac{\rho_b l^2}{Z_b^2} dZ_b \qquad dv_b = -\frac{\rho_b l^2}{Z^2} dZ$$

Determining Stroke Volume

- Change in thoracic impedance $\frac{\Delta Z}{\Delta t} = f'(Z)$
- Setting Δt = ejection time t_e $\Delta Z = f'(Z) \cdot t_e$

$$SV = \rho_b \frac{l^2}{Z^2} \left|\frac{dZ}{dt}\right|_{min} \cdot t_e$$

where SV = stroke volume [ml]

ρ_b = resistivity of the blood [Ω·cm]

l = mean distance between the inner electrodes [cm]

Z = mean impedance of the thorax [Ω]

$\left|\frac{dZ}{dt}\right|_{min}$ = absolute value of the maximum deviation of the first derivative signal during systole [Ω/s]

t_e = ejection time [s]

Calculating Parameters from Impedance Signals