Module -5 : Flow Measurement

Requirements for measurement ranges – blood flow in a single vessel, tissue blood flow, and respiratory gas flow. Electromagnetic flowmeters – principle, methods of magnetic field excitation, perivascular probes, intravascular probes. Ultrasonic blood flowmeters– propagation of ultrasound in the tissue, ultrasonic Doppler flowmeters, blood flow measurement through Doppler imaging. Indicator dilution method – principle and working, thermodilution method, Fick method, thermistor velocity probe, impedance cardiography.

By
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Requirements for Measurement Ranges :

Blood Flow in a Single Vessel:

Blood flow rate is the rate at which volume crosses a surface and the unit is (mL/min)

- Blood flow rate, as well as mean flow velocity in a blood vessel, can be roughly estimated by the size of the blood vessel, because the vessel size can vary adaptively with the blood flow rate.

- In an actual arterial system, a correlation exists between blood flow rate or velocity and arterial diameter.

- Blood flow rate is roughly proportional to the third power of vessel diameter, while mean flow velocity is roughly proportional to the diameter.

- The ratio of flow rates in a larger artery of about 2 cm in diameter to that in a capillary of 6 μm is more than 109 and to that of flow velocity is about 2000. No one measurement method is available which is applicable to the whole range of flow rates or velocity. Thus, different methods should be used for different ranges of flow or velocity.

Blood Flow in a Single Vessel:

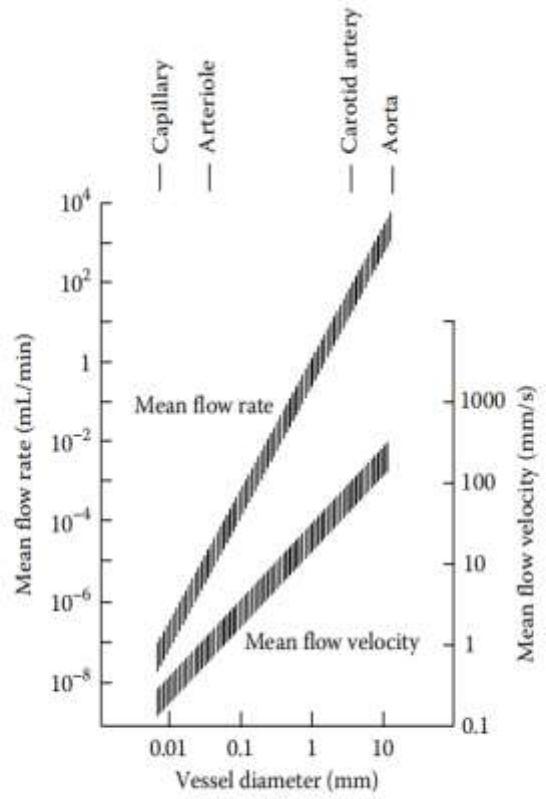


FIGURE 3.1 Rough estimate of mean blood flow rates and mean velocities in the arteries of different sizes.

- In a blood vessel, or in any conduit having a flowing fluid, flow velocity is not always uniform over a cross section but has a velocity distribution.

- If a conduit is assumed as a long straight tube having a circular cross section, and flow is assumed as steady and laminar, a parabolic velocity profile is seen.

Blood Flow in a Single Vessel:

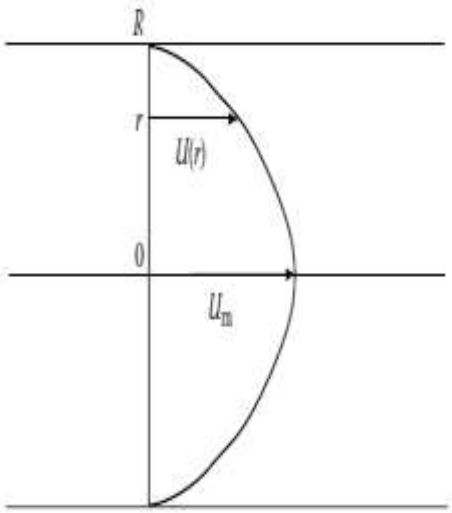


FIGURE 3.2 Parabolic velocity profile, which is realized in steady laminar flow in a long circular conduit.

- The backward velocity component is commonly observed even in small arteries, and the discrimination of flow direction is usually required in instantaneous velocity measurements. Backward flow occurs at the end of the systole and is substantial in patients with aortic valvular insufficiency.

- The velocity, $U(r)$, at a point at distance r from the center of the tube is expressed as:

$$U(r) = U_m\left(1 - \frac{r^2}{R^2}\right)$$

Where

R is the internal radius of the tube

Um is maximum velocity

- Thus, flow rate, Q, can be expressed as:

$$Q = \int_0^R U(r)2\pi r dr = \frac{1}{2}\pi R^2 U_m$$

- The flow rate Q divided by the cross sectional area πr2 is mean velocity, and is Um/2, i.e., the mean velocity is just a half the maximum velocity when the velocity profile is parabolic.
- Where blood flow is not steady but pulsatile, the velocity profile differs from the parabolic one.

Tissue Blood Flow:

- Tissue blood flow differs significantly for different tissues and physiological conditions. Figure 3.4 shows normal ranges of tissue blood flow rates for different organs in man (Folkow and Neil 1971).

FIGURE 3.3 Schematic of a typical velocity profile in the artery in one cardiac cycle.

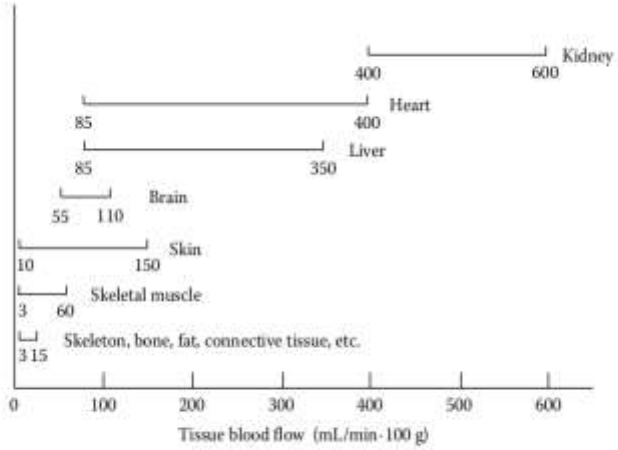


FIGURE 3.4 Rounded figure of tissue blood flow in different organs. Each range roughly indicates variation from the rest to the maximal vasodilatation.

Tissue blood flow is usually represented as the volume flow rate per unit mass of the tissue. If a region of a tissue is perfused uniformly, the tissue blood flow is estimated by the total blood flow into this region.

- Even if flow is not uniformly distributed, the average tissue blood flow can be defined as a rough estimate of the circulatory condition.

- For example, the average tissue blood flow in a segment of a limb can be regarded as an index of the peripheral circulation.

- The local tissue blood flow in an organ is also an important quantity for clinical diagnosis and physiological studies. If the local tissue blood flow can be measured at every part of an organ, the complete flow distribution will be obtained.ation.

Respiratory Gas Flow:

- The ventilation of the lungs can be assessed by studying a gas volume and its variations in the lung. Inspiratory and expiratory gas volumes or gas flow measurements provide data which characterizes the ventilation of the lung while actual gas volumes in the lung cannot be simply determined, because some amount of gas always remains in the lung even at maximal expiratory effort.
- The gas flow in airways is almost equal to the time derivative of the gas volume in the lungs as long as the temperature, pressure, and water vapor content of the ventilating gas is unchanged. Thus, flow measurement in the airways can be substituted by an instantaneous lung volume measurement.
- In clinical spirometry, measurement ranges of flow should cover peak flow rate at a maximal expiratory effort. According to the standard presented by the American Thoracic Association, the required ranges for flow and volume measurements are 0–12 L/s and 0–7 L.
- In respiratory measurement, gas composition may change significantly. Oxygen and carbon dioxide contents in expired air vary depending on the gas transfer rate in the lung. Oxygen content is increased when pure oxygen is added to the inspired air.
- Tracer gases such as helium and argon are sometimes used in the pulmonary function tests. In anesthetic monitors, flow measurements will be required for the air containing an anesthetic gas.
- A difference in the relative composition of gases may affect flow measurements due to the difference in the physical properties of these gases.
- Although a flow measurement method, which is unaffected by gas composition is desirable, existing methods are more or less affected by the physical properties of these gases, and thus

Electromagnetic flow meters: Principle:

The electromagnetic flowmeter is based on a principle that when a fluid containing electric charges flows in a magnetic field, an electromotive force is generated. If a particle having a charge *q* moves with a velocity ***U*** in a magnetic field of magnetic flux density ***B***, then a force ***F*** will be exerted on the particle, which is expressed in vector form as

$$\boldsymbol{F} = q(\boldsymbol{U} \times \boldsymbol{B}) \qquad (3.7)$$

In an electrolyte solution, such as blood flowing across a magnetic field, ions of positive and negative charges will move in opposite directions, and consequently, an electric field E will be generated, so that F is balanced with the electric force qE, i.e.,

$$q\boldsymbol{E} + q(\boldsymbol{U} \times \boldsymbol{B}) = 0 \qquad (3.8)$$

Thus, if two electrodes are placed along this electric field, a potential difference

$$V = \boldsymbol{S} \cdot \boldsymbol{E} = -\boldsymbol{S} \cdot (\boldsymbol{U} \times \boldsymbol{B}) \qquad (3.9)$$

will appear between this electrode pair, where S is a vector corresponding to a segment connecting the locations of two electrodes. These relations are shown in Figure 3.5. If U and B are perpendicular, the electromotive force V can be obtained as

$$V = dUB \qquad (3.10)$$

where ,

d is the distance between two electrodes, U = |U| and B = |B|.

Electromagnetic flow meters: Principle:

In blood vessels, the velocity is not uniform. Nevertheless, it can be shown that Equation 3.10 is still valid by taking U as the mean velocity as long as the velocity profile is axisymmetric about the longitudial axis of the vessel. This is an advantage of the electromagnetic flowmeter. Using mean velocity, the flow rate Q is expressed as

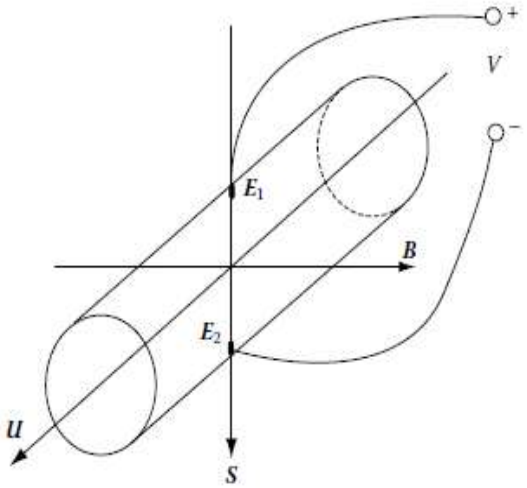


FIGURE 3.5 Relationship between flow velocity U, magnetic flux density B, developed electric field E, and electromotive force V.

$$Q = \frac{\pi d^2 U}{4} = \frac{\pi d V}{4B}$$

As long as the flow is axisymmetric. In SI units, Q is in m3/s, B in Tesla, d in meter, and V in volts are used.

Methods of magnetic field excitation

In earlier studies of electromagnetic blood flow metres static magnetic fields were used however due to a large electrode polarization potential superimposed on the blood flow induced signal it is difficult to obtain table records even if non polarized electrodes are used

To Eliminate the effects Polarization potential and to simplify the probe design the alternating magnetic field was introduced

Using AC field excitation a blood flow induced signal is obtained as an AC potential and can be separated from the DC polarisation potential

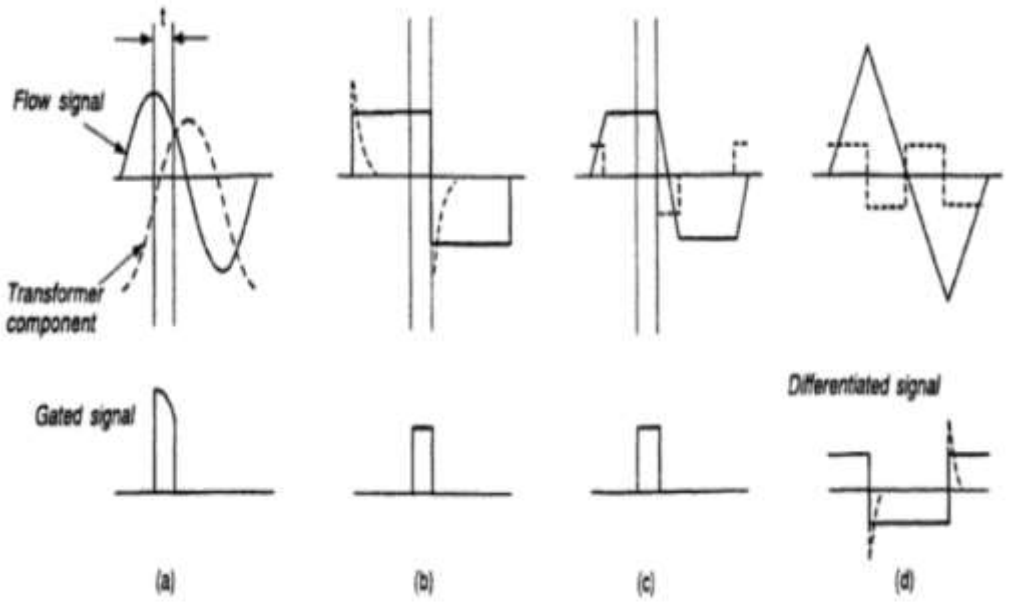


Figure 3.9 Flow signals and transformer components with gated or differentiated signals corresponding to sine wave (**a**), square wave (**b**), trapezoidal wave (**c**), and sawtooth wave (**d**) excitations.

Methods of magnetic field excitation Continued

An alternating magnetic field also induces a large AC potential directly in a loop circuit composed by lead wires and electrolytes between two electrodes. This potential is called the Transformer component.

To overcome all these issues and to detect only the blood flow component selectively Different techniques have been proposed as shown in the figure

In order to minimise the effect of the transformer component the excitation frequency should be chosen carefully.

In commercial instruments and excitation frequency of 100 to thousand hertz has been used. While the excitation frequency is selective in some flow metres

Perivascular probes

1. Below figure shows the different configurations of perivascular electromagnetic flowmeter probes

2. In the coreless probe, the excitation field strength is proportional to the excitation current and number of turns of the coil

3. By using an iron core , the filed strength can be increased about twice and temperature increase can be reduced significantly.

4. The EMF is proportional to the vascular diameter and the blood velocity.

5. To increase the signal amplitude the magnetic flux density should be increased thus iron core is preferred.

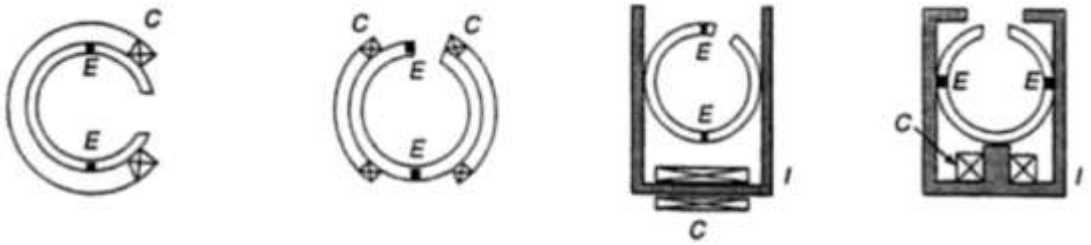


Figure 3.10 Different configurations of coreless and cored probes; *C*: coil, *E*: electrode, *I*: iron core.

Perivascular probes continued

6.leakage currents and capacitive leakage from exciting coils causes serious noise if insulation is insufficient. To eliminate these noise electrostatic shielding is performed

7. Perivascular probes for vessels from 35mm down to 0.5 mm in diameter are available commercially.

8. There should be a good contact between electrode and the vessel hence the inner diameter of the probe must fit the outer diameter of the vessel.

9. Fig shows a typical configuration of perivascular probes. The blood vessel is passed through the slot and allowed to expand into the lumen so both the surface will have a good contact

10. For chronic disease treatment these probes are implanted inside the animal body with a transcutaneous connector to which signal and excitation cables are connected

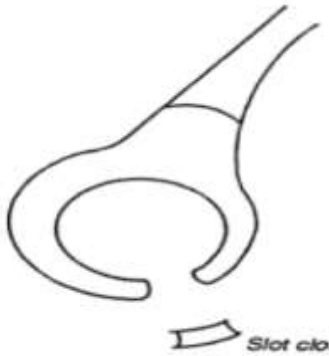


Figure 3.11 A typical perivascular probe with the slot closure.

Perivascular probes continued

The removable electromagnetic flow probe had a pair of coils with pliable Core and 2 Gold electrodes. All the components enclosed in a Slender silastic moulding of uniform cross sectional area. The probe was positioned around the ascending aorta as shown in the figure

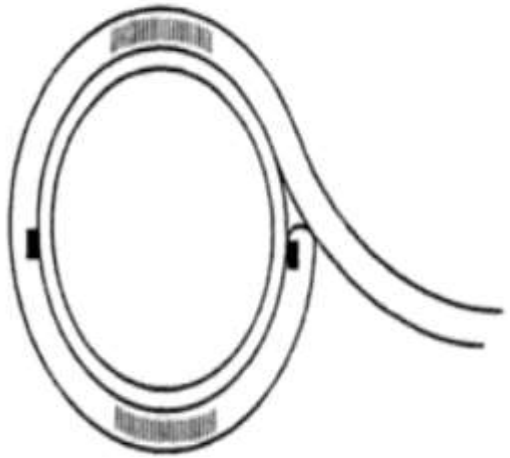


Figure 3.12 A removal electromagnetic flow probe positioned around the aorta. A pair of coils with pliable cores and electrodes was enclosed in a flexible silastic molding. (From Williams, B. T. et al., *Rev. Surg.*, 26, 227, 1969. With permission.)

Intravascular probes

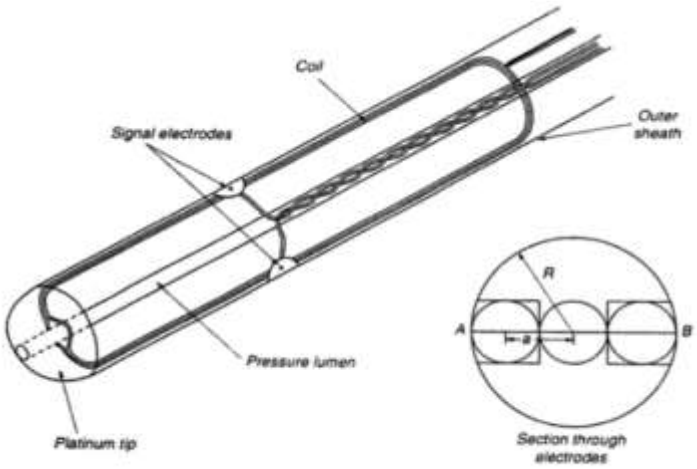


Figure 3.13 An electromagnetic velocity probe using a coreless coil. (From Mills, C. J. and Shillingford, J. P., *Cardiovasc. Res.*, 1, 263, 1967. With permission.)

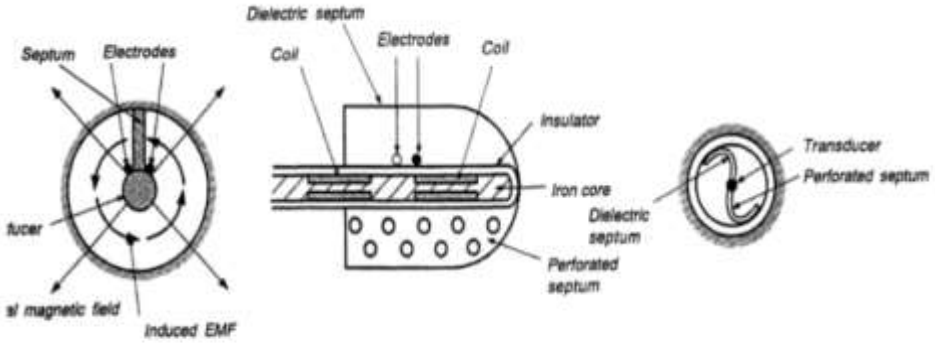


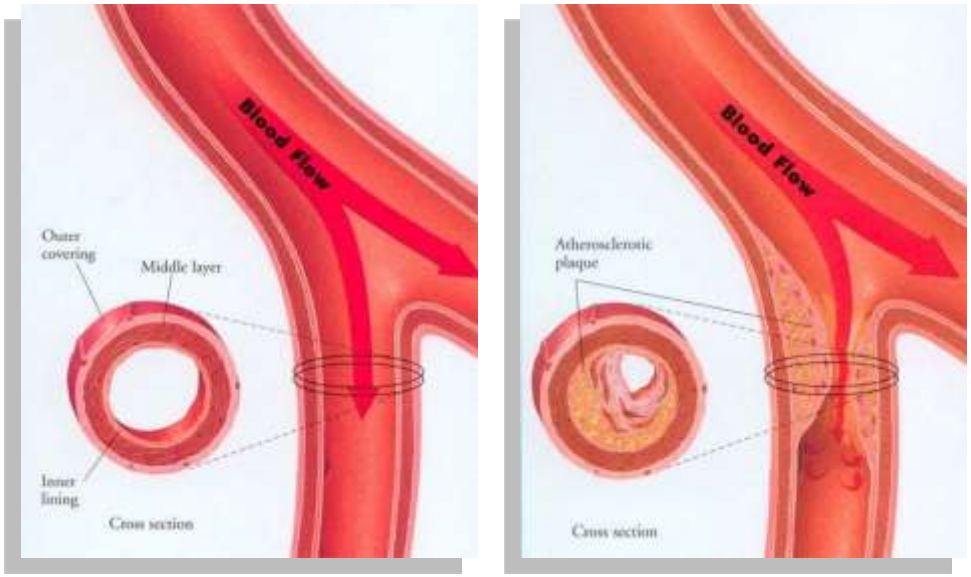
Figure 3.15 A radial field intravascular velocity probe. (From Kolin, A., *IEEE Trans. Biomed. Eng.*, BME-16, 220, 1969. With permission.)

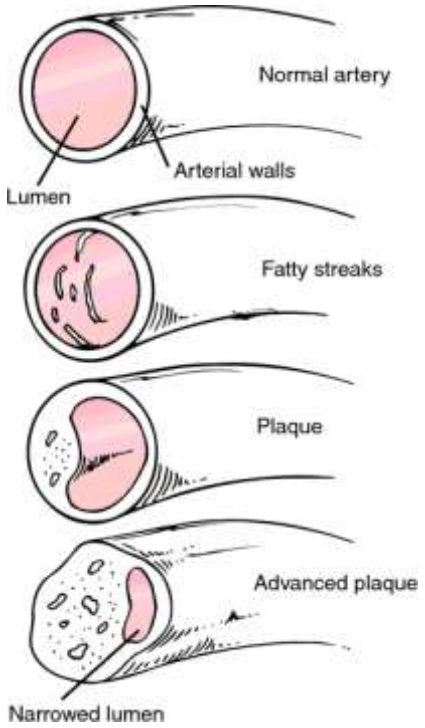
1. Catheter type electromagnetic velocity probes have been developed and used clinically for monitoring blood velocity e in the large arteries and veins

2.The electromagnetic velocity Pro using core less coil was designed by Mills. The design of a modified probe is shown in the figure

3. A catheter 3 mm in diameter contains a coil of 30 turns with signal electrodes on its surface

4. Clinical trials were made on the dogs





Heart	• Coronary artery disaese(CAD) • Anging(Chest pain which become worse with physical actvity) • Arrythmia(irregular heart beats) • Heart attack
Brain	• Carotid artery disease(numbness, fatigue, dizzness, and headaches) • Stroke
Arms &Legs	• Peripherial arterial disease(numbness and serious infections) • Intermittent claudication(leg pain on walking)

Normal blood flow velocity 0,5 m/s – 1 m/s (Systolic, large vessel)

ULTRASONIC FLOWMETERS

Basics

•Ultrasonic waves are sound waves above human hearing (>20 kHz)

• Typical frequencies are 20 kHz - 20 MHz.

•Sound waves are longitudinal pressure waves caused by vibrations in a medium

•Several types of ultrasonic sensors are available- the most common are dynamic or piezoelectric sensors

•A typical dynamic sensor is a thin, low mass diaphragm, stretched over passive electromagnet.

• Such diaphragms operates at frequencies up to 100 kHz

Propagation of Ultrasound in the Tissue

Ultrasound propagates in the soft tissue with a velocity of about 1500m/s.
The sound velocity c, frequency f, and wavelength λ are related as

$$c\, f = \lambda$$

sound at a point in a medium is characterized by the sound pressure and the sound particle velocity (or medium velocity).

If a sound wave propagates in one direction, the amplitude of sound pressure P is proportional to that of sound particle velocity U, so that

$$P = \rho c U$$

where ρ isthe density of the medium.

The ratio of the amplitude of sound pressure to particle velocity
is called a characteristic impedance (or a characteristic acoustic impedance) of the medium, that is

$$Z = P/U = \rho c$$

Propagation of Ultrasound in the Tissue..... Continued.....

At the boundary between two media, having different characteristic impedances, the sound wave will be partially reflected. If a sound wave with the sound pressure amplitude pi impinges perpendicularly on a boundary between two media having characteristic impedances Z1 and Z2, then the

sound pressure of the reflected wave is given as

$$p_r = \left| \frac{Z_1 - Z_2}{Z_1 + Z_2} \right| p_i$$

Sound energy (or sound power) I is expressed as

$$I = \frac{1}{2} pu = \frac{1}{2} \frac{p^2}{Z}$$

For an incident sound wave with sound energy Ii, the sound energy of the reflected wave Ir, is obtained as

$$I_r = \left(\frac{Z_1 - Z_2}{Z_1 + Z_2} \right)^2 I_i$$

Propagation of Ultrasound in the Tissue..... Continued.....

Thus the sound energy of transmitted sound It, is given as

$$I_t = I_i - I_r = \frac{4Z_1Z_2}{(Z_1 + Z_2)^2} I_i$$

When ultrasound propagates in tissue, it is attenuated by absorption, reflection, and scattering.
If a plane wave propagates in the x direction in an absorbing medium, and if it has a sound pressure p(0) at x = 0, then sound pressure at x is represented as

$$p(x) = p(0)\, e^{-\alpha x}$$

where α is the absorption coefficient. It varies almost in proportion to the sound frequency. Because sound energy is proportional to a square of sound pressure, sound energy at x, I(x) is expressed as

$$I(x) = I(0)\, e^{-2\alpha x}$$

Ultrasonic Doppler Flowmeters

When a flowing fluid contains particles by which ultrasound is scattered, the velocity of particles can be determined by the Doppler shift of the scattered ultrasound.

The scattering in which particle size is less than the wavelength of the sound is called Rayleigh scattering, and in that case, the intensity of the scattered sound wave is proportional to the fourth power of the sound frequency.

Thus, the scattered signal will be significantly stronger at higher sound frequencies. The sound intensity, however, decreases exponentially with increasing propagation distance

If the ratio of the absorption coefficient, α, and the frequency, f, is assumed to be a constant, k, the amplitude of the scattered sound wave after propagation over a distance, x, will be expressed as

$$I \propto f^4 e^{2\alpha x} = f^4 e^{-2kfx}$$

Ultrasonic Doppler Flowmeters continued.....

The ultrasonic Doppler flowmeter measures the Doppler shift in the scattered wave due to the moving red blood cells in the blood stream.

When an ultra_x0002_sound transmitter and a receiver are arranged, as shown in Figure, the Doppler shift caused by the scattering object with a velocity U can be calculated in two steps.

First, the frequency observed at the scattered object is calculated. This is a situation where a moving observer receives a sound wave from a stationary source. At the moving observer, the observed frequency of a wave coming from a source at an angle θ to the flow direction is given as

$$f_1 = \frac{c + U\cos\theta}{c} f_s$$

fs is source frequency
c is sound velocity

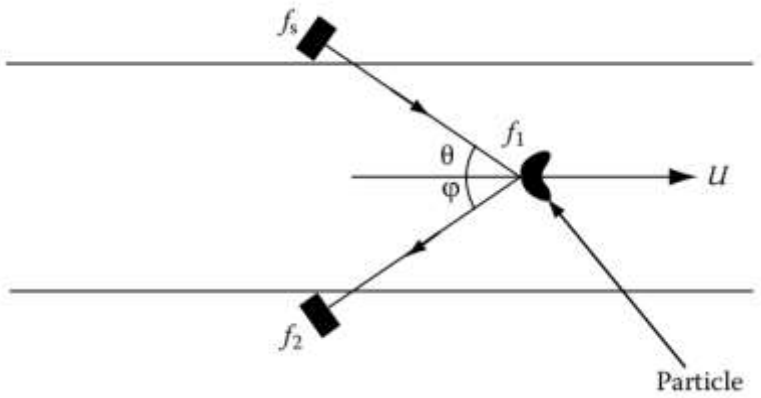


FIGURE : Measurement of the Doppler shift.

Ultrasonic Doppler Flowmeters continued.....

Then, the frequency of the reflected wave observed at a stationary receiver is calculated. This is a situation where a stationary observer receives a sound wave from a moving source.

The receivedfrequency of a wave going toward the observer at an angle ϕ to the flow direction is given as

$$f_2 = \frac{c}{c - U\cos\varphi} f_1$$

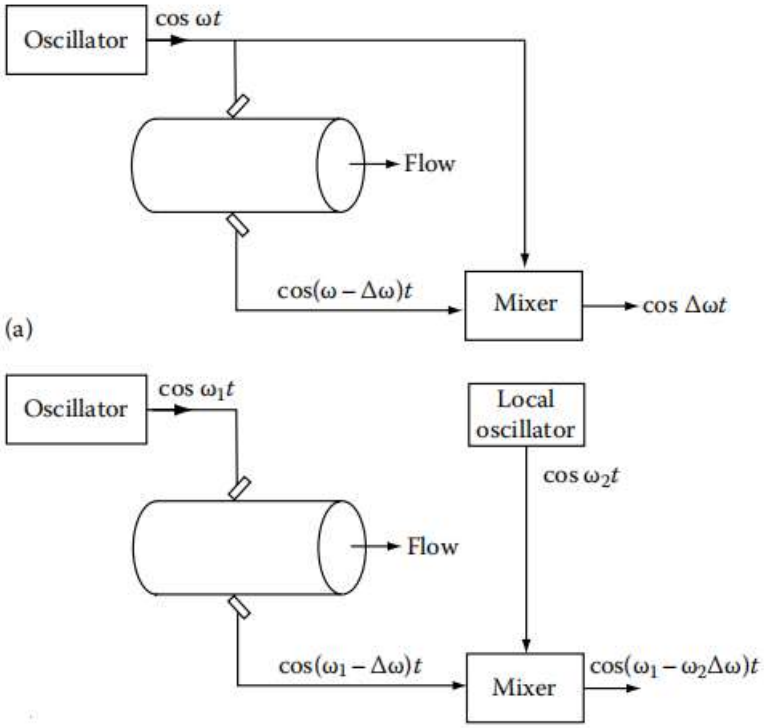
Rearranging the two equations above and assuming that Ucosϕ ≪ c, the Doppler shift, Δf, can be obtained as

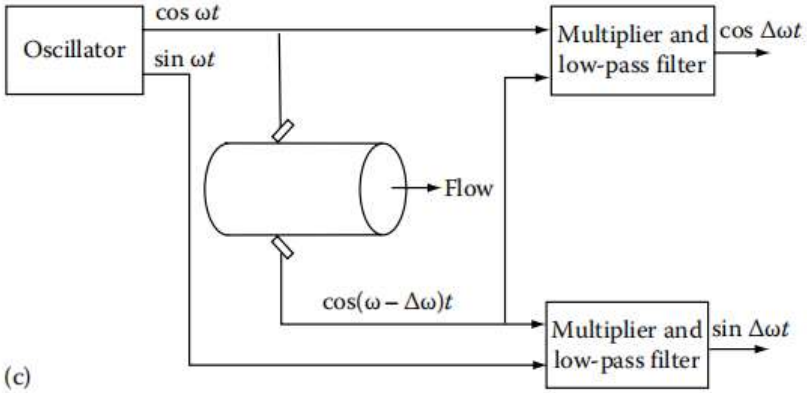
$$\Delta f = f_2 - f_s = \frac{c + U\cos\theta}{c - U\cos\varphi} f_s - f_s \approx \frac{U(\cos\theta + \cos\varphi)}{c} f_s$$

If θ = ϕ,

$$\Delta f \approx \frac{2U\cos\theta}{c} f_s$$

Ultrasonic Doppler Flowmeters continued.....

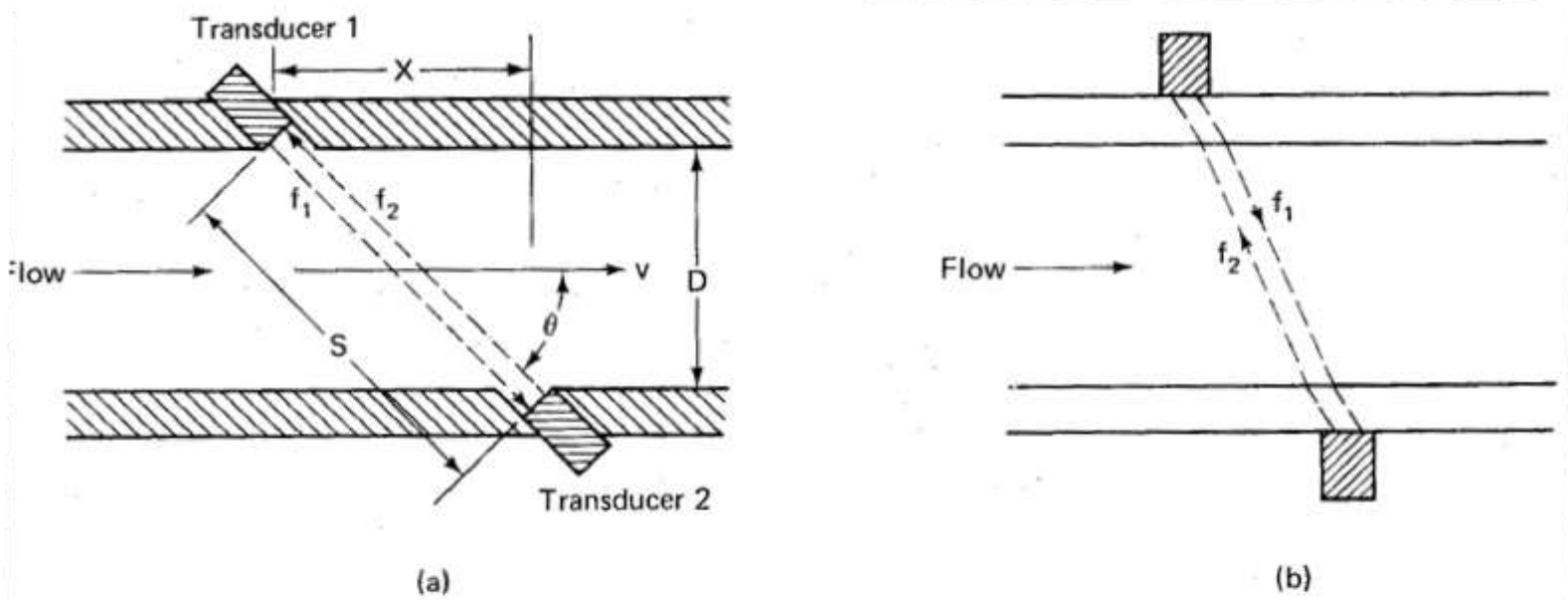




ULTRASONIC FLOWMETERS(Out of Syllabus)

Ultrasonic flow sensors

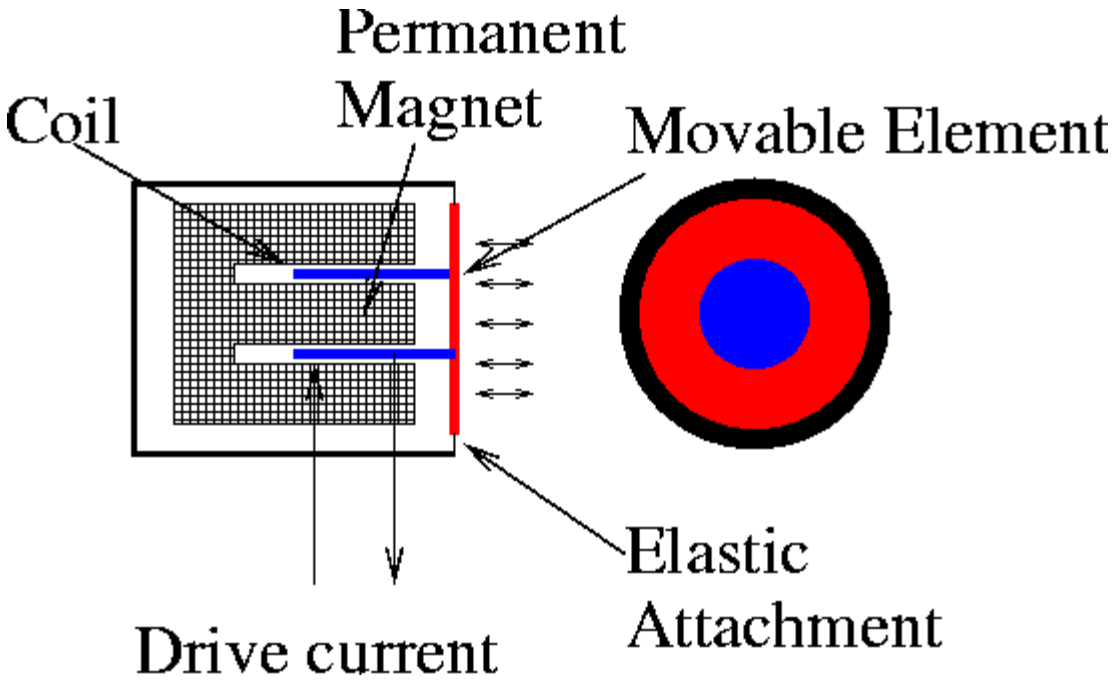
- Many ultrasonic flow sensors consist of pairs of transducers
- Each transducer can operate as either a source or a detector of sound waves



Ultrasonic flow sensing (frequency-difference principle): (a) wetted; (b)

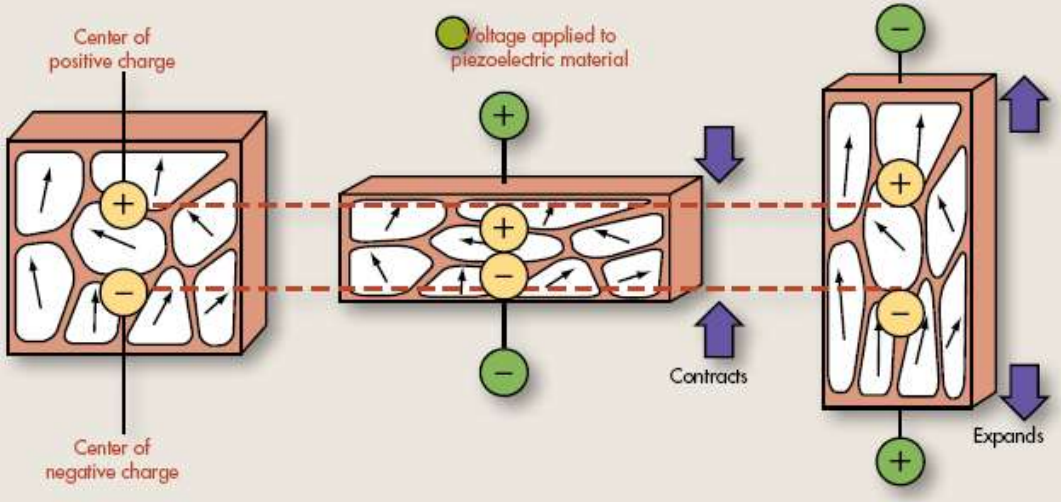
Dynamic Ultrasonic Transducer

- As a generator of ultrasonic waves: the drive current creates a magnetic field which pushes against the permanent magnet.
- As a detector: the motion of the element induces a current in the drive coil

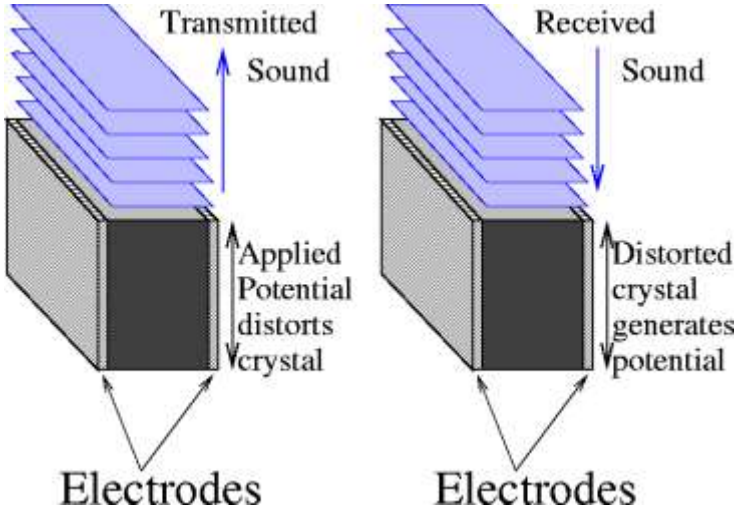


Piezoelectric ultrasonic transducers

- The piezo transmits when an applied potential distorts crystal
- Receives when pressure wave distorts crystal



1. The piezoelectric effect causes crystal materials like quartz to generate an electric charge when the crystal material is compressed, twisted, or pulled. The reverse also is true, as the crystal material compresses or expands when an electric voltage is applied.



Why using ultrasonic types?

•Wide range of applications such as blood, pure water, wash water, sewage, process liquids, oils, and other light homogeneous liquids

•Clamp-on types measure flow through the pipe without any wetted parts, ensuring that corrosion and other effects from the fluid will not deteriorate the sensors.

•Clamp-on types simplify and speed up meter installation and minimize maintenance.

•Ultrasonic flowmeters may be portable.

•Measurement accuracy can be in the range of 1% of flow rate, and speed of response can be as fast as 1 s.

Types of Ultrasonic Flowmeters:

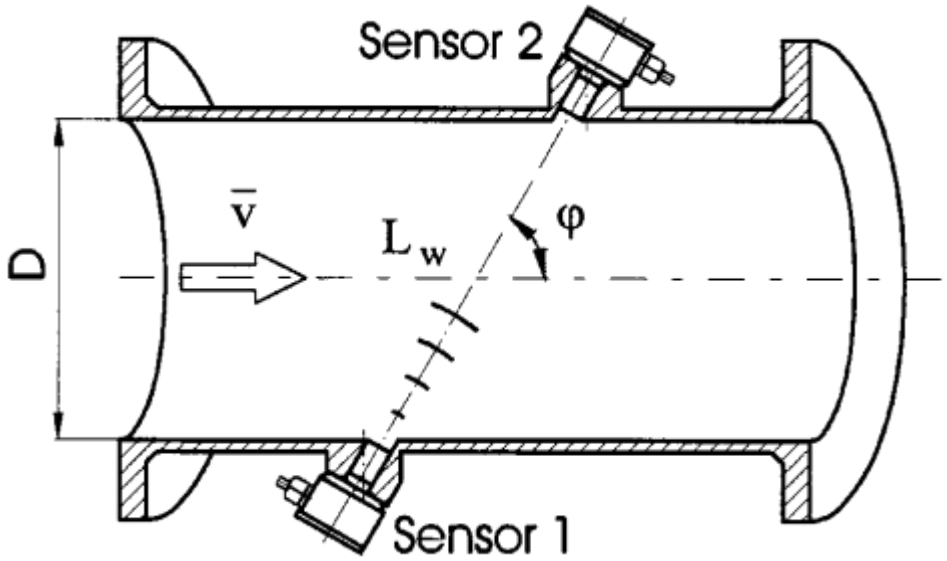
1 Transit time:

This type of ultrasonic flowmeter makes use of the difference in the time for a sonic pulse to travel a fixed distance, first against the flow and then in the direction of flow.

2 Doppler:

It makes use of the Doppler frequency shift caused by sound reflected or scattered from suspensions in the flow path

Transit Time Flowmeter



$$t_{12} = \frac{L_w}{c + v_a \cos\varphi} \quad \text{and} \quad t_{21} = \frac{L_w}{c - v_a \cos\varphi}$$

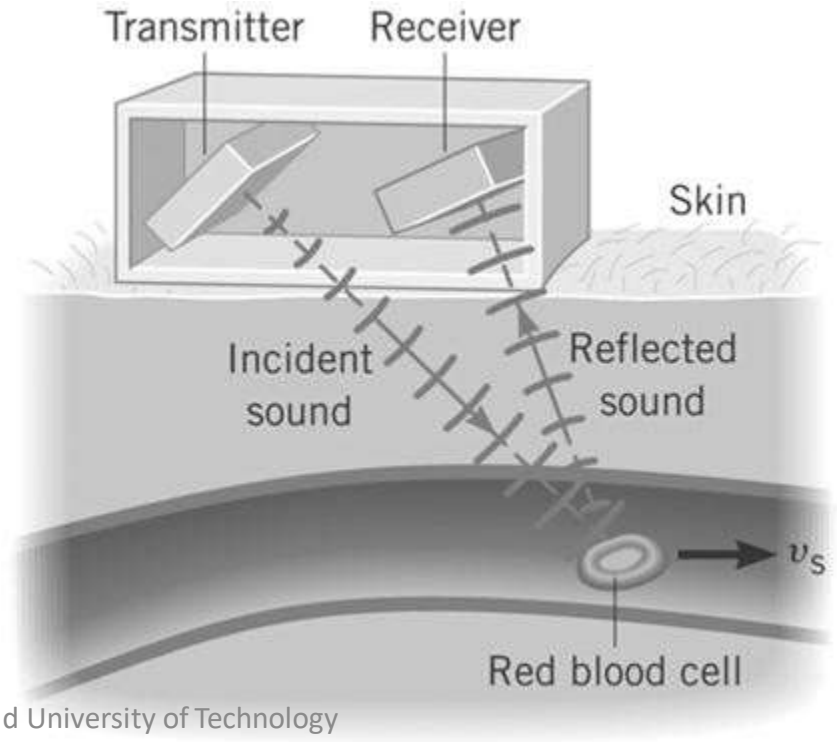
where L_w = Distance in the fluid between the two transducers
c = Speed of sound at the operating conditions
ϕ = Angle between the axis of the conduit and the acoustic path
$\bar{v}_a$ = Axial low velocity averaged along the distance L_w

$$\bar{v}_a = \frac{L_w}{2\cos\varphi}\left(\frac{1}{t_{21}} - \frac{1}{t_{12}}\right) = \frac{D}{2\cos\varphi\sin\varphi}\left(\frac{1}{t_{21}} - \frac{1}{t_{12}}\right)$$

The Doppler effect

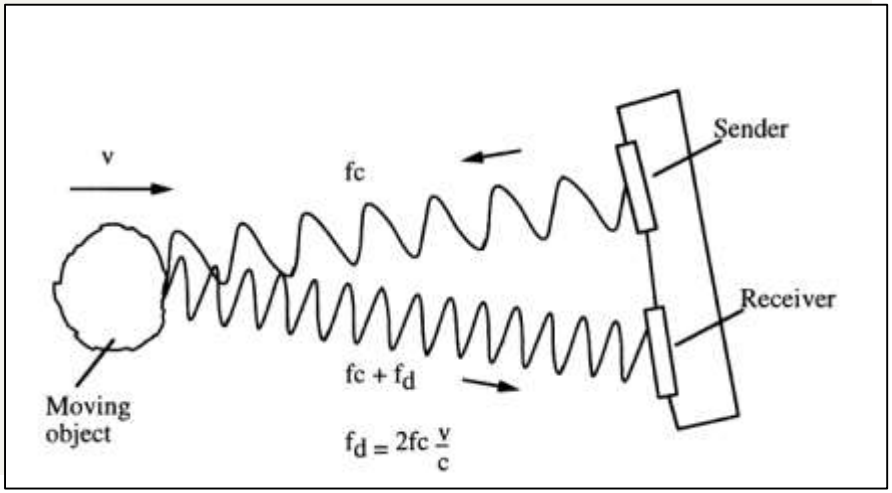
Doppler effect is a shift in frequency from a moving source of waves.

We can use the Doppler effect to measure the velocity of a fluid.



Ultrasound Doppler

Doppler Measurements:



The blood cells in the fluid scatter the Doppler signal diffusively.

In the recent years ultrasound contrast agents have been used in order to increase the echoes.

$$f_d = 2 f_c \frac{v}{c}$$

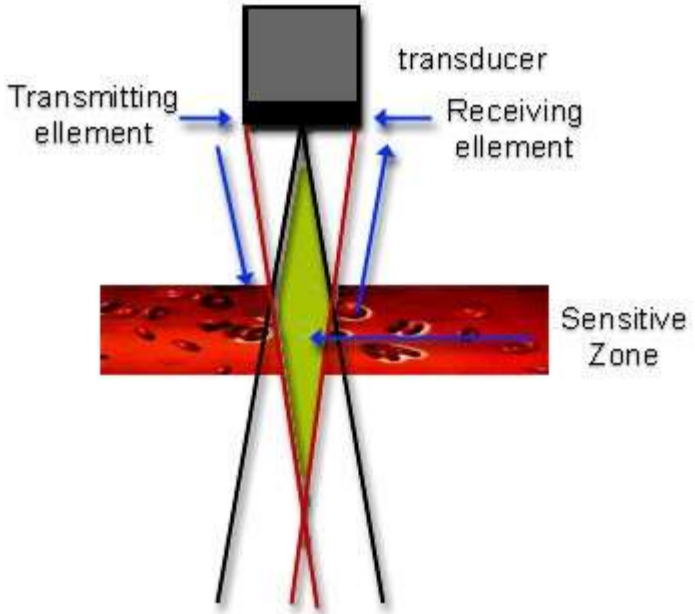
The ultrasound beam is focused by a suitable transducer geometry and a lens

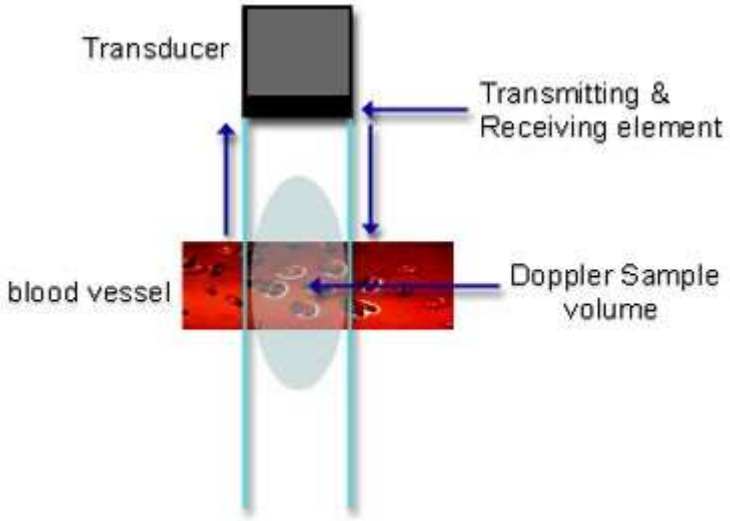
f = 2 – 10 MHz

c = 1500 - 1600 m/s (1540 m/s)

$f_d = 1.3 – 13\ kHz$

Continuous wave and pulse wave





Continuous wave systems use continuous transmission and reception of ultrasound. Doppler signals are obtained from all vessels in the path of the ultrasound beam (high velocity)

The PW transducer both sends and receives the signal. It sends in short bursts and receives in the time when it is not sending(accurate region).

Ultrasound Doppler General Parameters

•the power decays exponentially because of the heating of the tissue. The absorption coefficient ~ proportional to frequency

•the far field operation should be avoided due to beam divergence.

$$d_{nf} = \frac{D^2}{4\lambda}$$

D = Transducer diameter (e.g. 1 – 5 mm)

•the backscattered power is proportional to f^4

the resolution is related to the pulse duration. Improving either one of the parameters always affects inversely to the other

Limitations of Doppler flowmeters

Liquids to be metered must have an excess of approximately 2% suspended solids by volume

Liquid linear velocities must exceed 0.15 m/s

Piping material must be of a homogenous composition

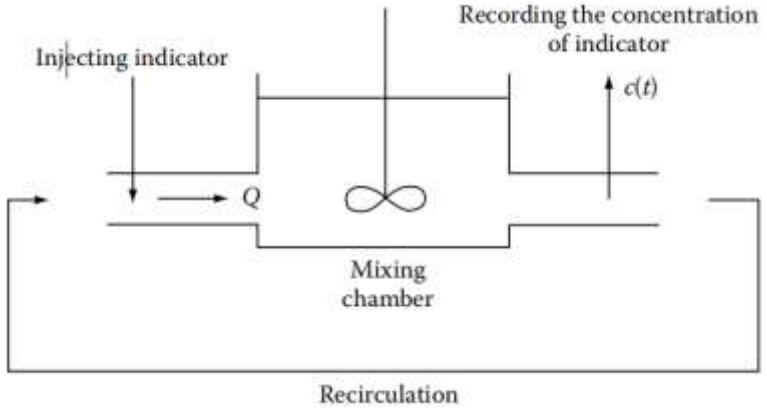
Pipe wall thickness cannot be greater than 1.91 cm

Indicator dilution method – Principle and working

Flow measurement by indicator dilution is a method in which a definite amount of indicator is injected into the blood stream, and the averaged flow rate is estimated from the time course of the concentration of the indicator at the downstream.

Color dyes, radio isotopes, electrolytes, or heat can be used as indicators. Fick's method for cardiac output measurement is also a variation of the indicator dilution method in which oxygen, carbon dioxide, or other gases are used as indicators.

Suppose a part of the circulatory system is simulated by a model as shown in Figur. The indicator is injected upstream, mixed uniformly in the mixing chamber, and then moved downstream where the concentration of the indicator is observed.

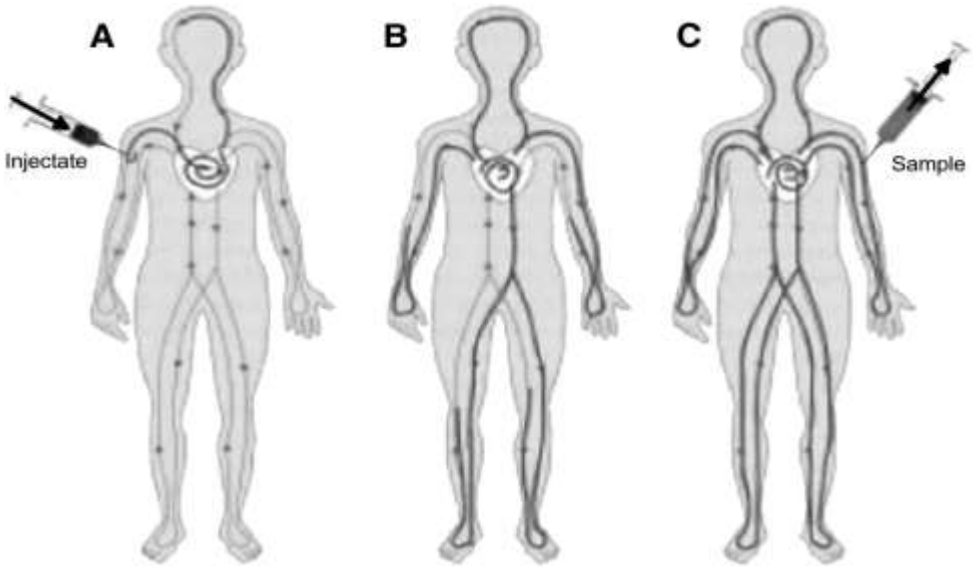


Blood flow measurements

Indicator Dilution Methods:

Continuous Infusion

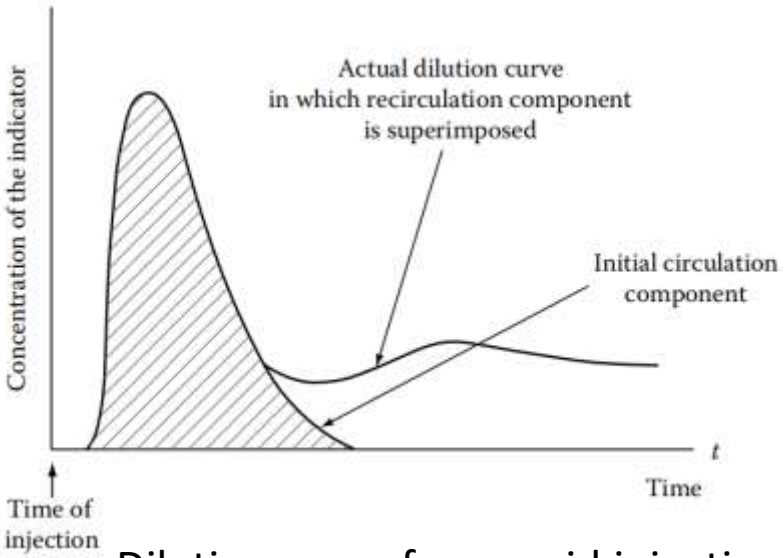
- •Fick Method
- Rapid Injection Methods
  - Dye Dilution
  - Thermo-dilution



In the rapid injection methods, such as slug injection and bolus injection, a definite amount of indicator is injected in a short period of time. After a while,the concentration of the indicator downstream will increase rapidly and then decrease until the indicator appears again by recirculation, as shown in Figure The time course of the concentra_x0002_tion of the indicator is called the indicator dilution curve.

The indicator-dilution methods do not measure **instantaneous pulsatile flow** but, rather, **flow averaged** over a **number** of **heartbeats**.

Indicator Dilution Method continued



Dilution curve for a rapid injection of the indicator.

If the injected indicator does not leak out from the conduit, the entire amount of the injected indicator passes the observation point downstream.
Thus, if the flow rate is Q, and the concentration of the indicator at time t is c(t), the amount of injected indicator, I, should be

$$I = \int_0^\infty Qc(t)dt$$

as long as the indicator does not recirculate. If the flow rate is constant, it can be solved as

$$Q = I \Big/ \int_0^\infty c(t)dt$$

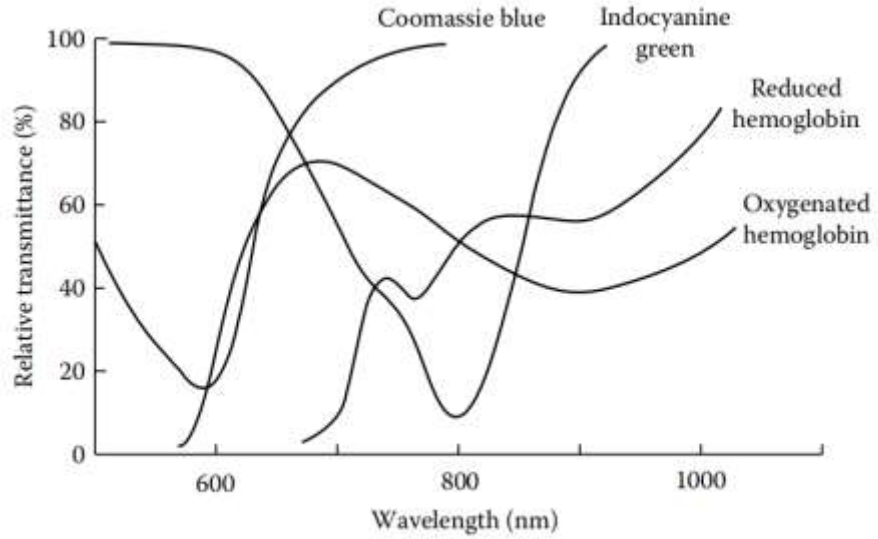
$$Q = I/C$$

Dye Dilution Method

The dye dilution method is a kind of indicator dilution method, in which a dye is used as the indicator and the concentration of dye is usually detected optically.

The dye for blood flow measurement should be water soluble, nontoxic, sterile, and should permit precise determination of its concentration in whole blood or plasma.

Besides this, it should not be lost or degenerated metabolically during initial circulation. It is preferable that the dye leaves the circulatory blood rapidly after initial circulation so that it does not disturb repeated measurements



Relative spectral transmittance of dyes and reduced and oxygenated hemoglobin.

Coomassie blue has its absorption peak at about 600nm

Indocyanine green has two absorption peaks near 800 nm

Dye Dilution Method

When the concentration of dye is low enough, and it is dispersed uniformly in the blood, the Beer–Lambert's law is valid for the optical density, I, and the concentration of dye, c, i.e.,

$$\varepsilon cd = \log\left(\frac{I_0}{I}\right) = \log I_0 - \log I$$

where

ε is the absorption coefficient

d is the thickness of the blood layer

I0 is the optical density when c = 0

Calibration of the earpiece densitometer is performed by taking a blood sample after the dye is uniformly mixed in the circulating blood, and the deflection at the terminal of the dilution curve corresponds to the concentration of dye in the blood sample.

Fiberoptic catheter can be used for recording dye dilution curves in vivo. At the catheter tip, the blood is illuminated by a light beam transmitted through a bundle of optical fibers, and part of the backscattered light is transmitted to the detector through another bundle of optic fibers.

Thermodilution Method

The thermodilution method is a kind of indicator dilution method in which a definite amount of heat is injected into the blood stream, and the corresponding temperature change is recorded down_x0002_stream.

In thermodilution method, a cold fluid is often used as an indicator, because cold fluid is less harmful to the blood and tissue than a hot one.

For cardiac output measurement in a human adult,about 10mL of cold saline or isotonic dextrose solution near 0°C is commonly used. The injected cold indicator is mixed and diluted in the warm blood stream, and causes a slight temperature decrease in the blood downstream.

The thermodilution method has several advantages, i.e., the indicator has no toxic effect so that measurement can be performed repeatedly, the dilution curve can be easily recorded by a thermistor placed in the vessel

Flow rate Q can be calculated from the blood temperature TB as follows. If the volume and the temperature of the injected fluid are VI and TI, and the density and specific heat of the blood andthe injected fluid are ρB, ρI, and CB, CI, respectively, then

$$Q = \frac{\rho_I C_I}{\rho_B C_B} \frac{V_I(T_B - T_I)}{\int_0^\infty \Delta T_B dt}$$

where ΔTB is the change in the blood temperature.
The coefficient ρBCB/ρICI is about 0.93 for 5% dextrose solution and 0.91 for saline

Thermodilution Method continued.....

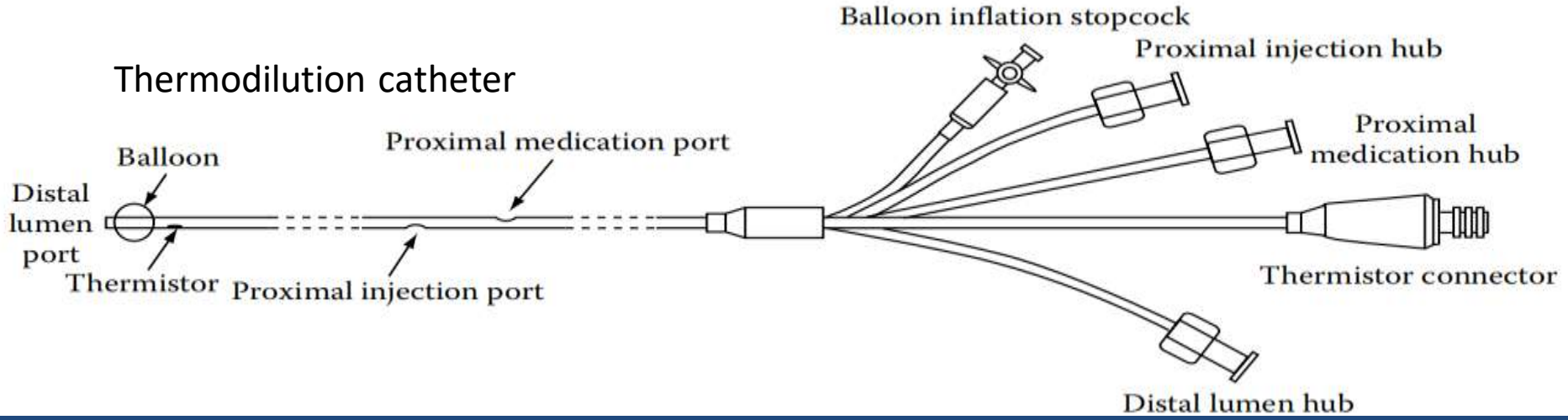
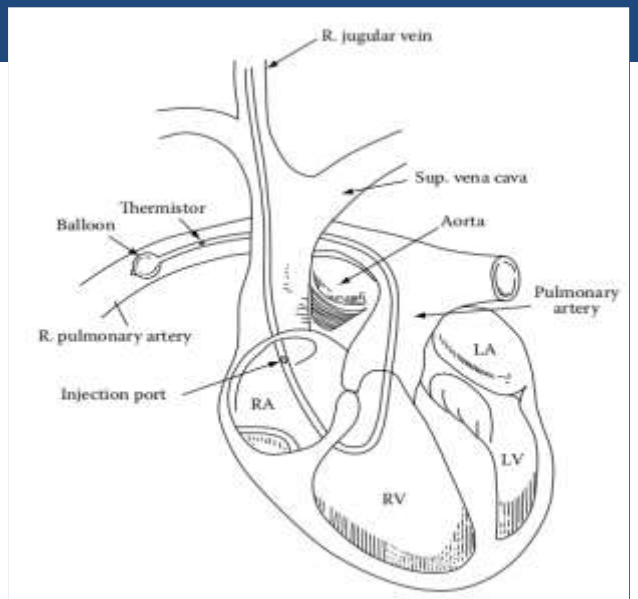


Figure shows the configuration of the catheter. It is 7–7.5F (2.3–2.5mm in outer diameter), has a balloon, a thermistor having time constant 0.2–0.4 is attached near the tip, and an injection port is located 25–30cm from the tip.

Conventionally, the catheter is introduced from a peripheral vein into the pulmonary artery through the right ventricle as shown in Figure.

A bolus of cold saline or dextrose solution is injected into the right atrium, and mixing occurs in the right atrium and the right ventricle and the decrease in the resultant temperature is detected by the thermistor placed in the pulmonary artery.



Thermodilution method for monitoring cardiac output.

Thermodilution Method continued.....

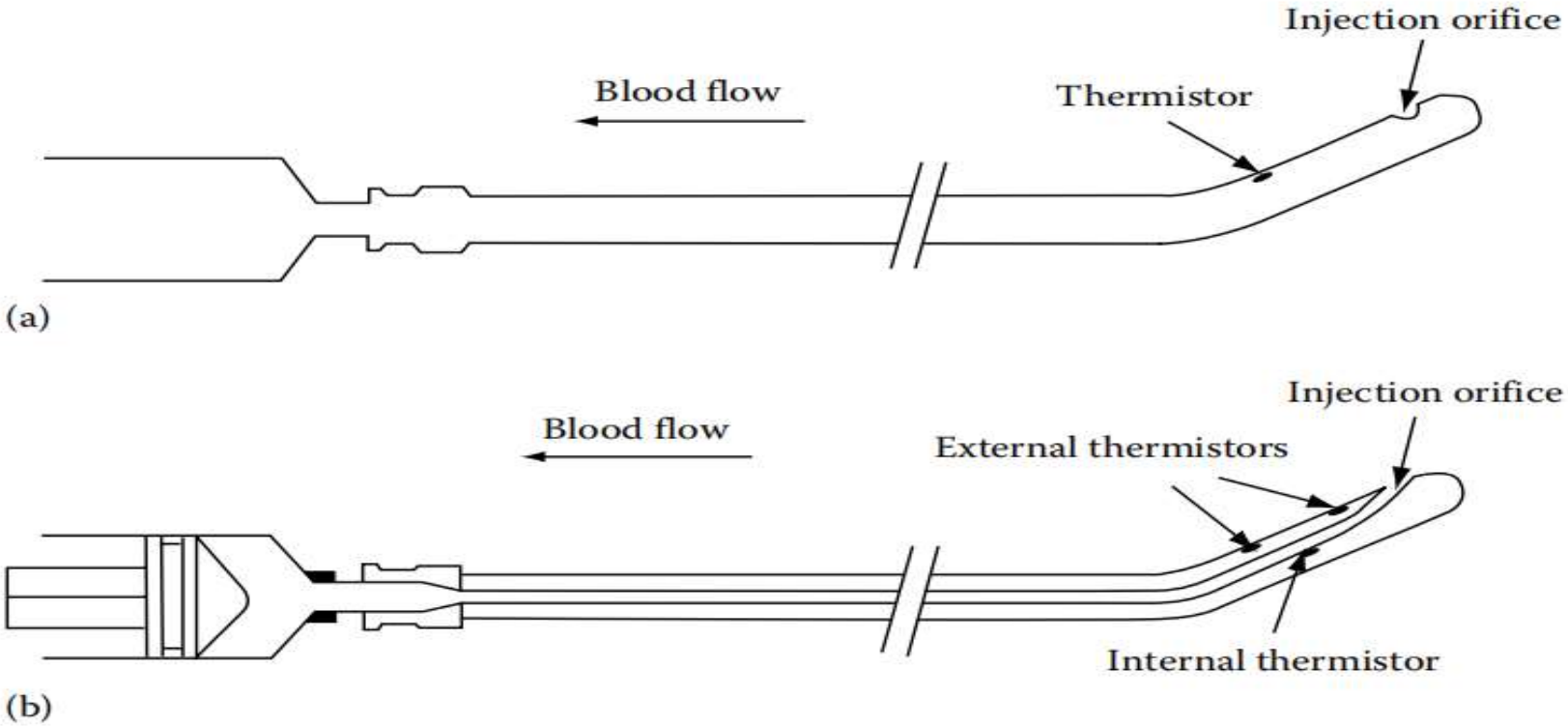


Figure a & b : Thermodilution catheters by continuous injection.

The Fick Method

Pulmonary blood flow can be determined by measuring the rate of gas intake to or extraction from the blood in the lung, and the change in the gas concentration in the blood stream through the lung. This technique is called the Fick method, and has been used as a standard method of cardiac output measurement.

If oxygen uptake, VO2 and oxygen concentration in the pulmonary arterial blood,CaO2, and that in the pulmonary venous blood,CvO2, are known, the blood flow rate through the lung Q can be obtained as

C is considered as the increment of indicator concentration.

$$Q = \frac{\dot{V}_{O_2}}{C_{aO_2} - C_{vO_2}}$$

When a carbon dioxide measurement is employed the pulmonary blood flow is expressed as

$$Q = \frac{\dot{V}_{CO_2}}{C_{vCO_2} - C_{aCO_2}}$$

where
VCO2 is carbon dioxide output
CvCO2 and CaCO2 are carbon dioxide contents in the mixed venous and the arterial blood

Thermistor Velocity Probes

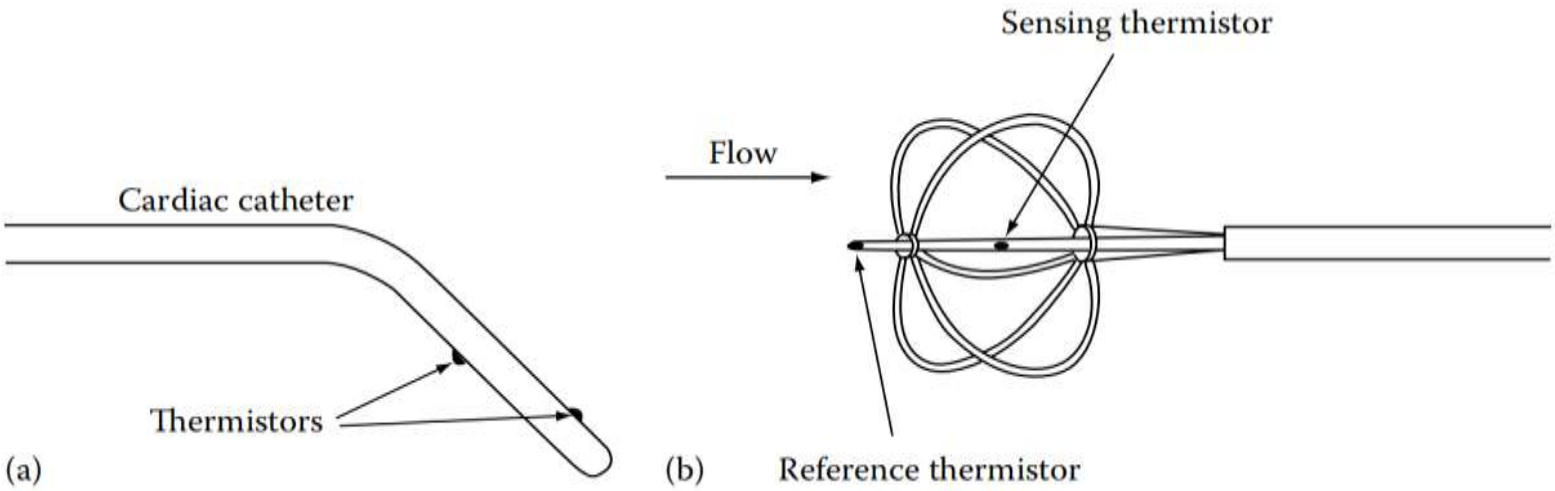
The thermistor is a convenient device for use in thermal velocity probes because it has a large temperature coefficient. Two thermistors are commonly used in the flow probe.

One measures the fluid temperature, and the other is heated to a higher temperature, and the temperature difference between the heated thermistor and the fluid is kept constant.

igure shows two examples of such probes. (a) is a 6F (2mm in outer diameter) cardiac catheter having two thermistors near the tip and, because of a bend, the thermistors are placed near the center of the blood stream when it is introduced into the vessel

The response time was 0.2s for abrupt increase, and 1.5s for abrupt decrease in the temperature of the fluid, and it was used for venous flow measurements.

Thermistor Velocity Probes



Thermistor velocity probes: (a) thermistor velocity probes
(b) venous flow catheter.

Figure b also shows a velocity probe with two thermistors. It has the velocity sensing element at the center of five flexible springs, and the reference thermistor is placed at the tip.

The sensing element consists of a tiny thermistor around which has been wound a fine wire for heating. Its frequency response was about 5Hz, and it could be used for venous flow measurements.

Impedance cardiography

Stroke volume, Cardiac output and heart sounds

1. **Cardiac output** – the volume of blood pumped from each ventricle per minute:

CO	=	SV	x	HR
cardiac output (ml/minute)	=	stroke volume (ml/beat)	X	heart rate (beats/min)

Stroke Volume is the Volume of blood pumped from one ventricle of heart with each beat

Impedance cardiography is a technique in which stroke volume or cardiac output is estimated by the waveforms of transthoracic electric impedance

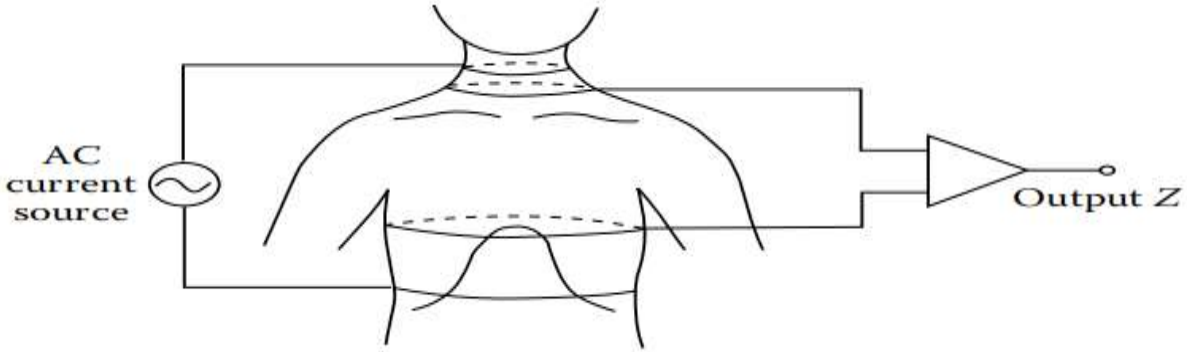
PCG : Phonocardiogram: A chart or record of the sounds made by the heart.

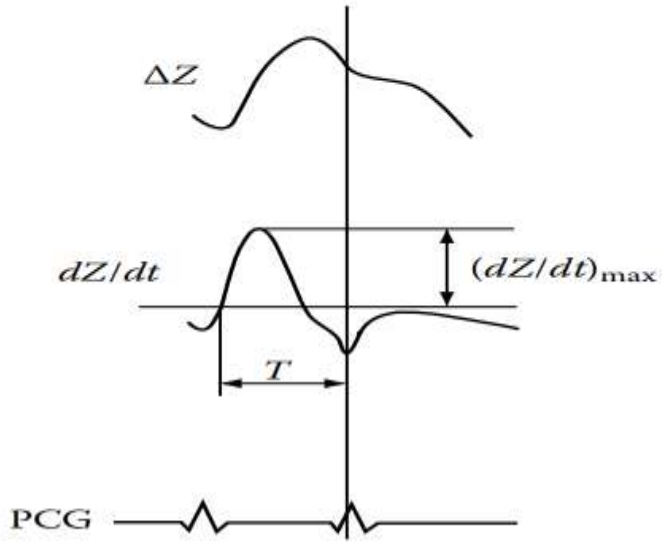
Impedance cardiography Continued

To record transthoracic electrical impedance, the tetrapolar electrode arrangement as shown in Figure 3.27 (left) has generally been used.

An a.c. current within the range of 20–100kHz at a current level within the range of 10μA–10mA is supplied through current electrodes, which are placed at the top of the neck and at the end of the rib cage or distal to it.

Voltage electrodes are placed at the base of the neck and at the level of the xiphisternal joint, and the induced voltage between voltage electrodes is measured. The thoracic impedance, Z, is then defined by the obtained voltage, V, divided by the supplied current, I.





Impedance cardiography Continued

Figure (right) shows a typical tracing of the change in thoracic impedance, ΔZ, its time derivative, dZ/dt, and the phonocardiogram.

$$SV = \frac{\rho_b L^2}{Z_0^2} \Delta Z$$

where
ρb is blood resistivity
L is the distance between voltage electrodes
Z0 is the average thoracic impedance