MODULE 3: PRESSURE MEASUREMENT

3.1 Pressure Transducers

3.1.1 LVDT pressure transducers

3.1.2 Strain gauge pressure transducers

3.2 Physiological pressure ranges and measurement sites

3.3 Direct pressure measurement-catheters for pressure measurement

3.4 Diaphragm Displacement Transducers

3.5 Catheter Tip Pressure Transducers

3.6 Implantable Pressure Transducers and Pressure Telemetering Capsules.

3.7 Indirect Pressure Measurement-Indirect Measurement of Systolic, Diastolic, And Mean

Blood Pressure

3.8 Detection of Korotkoff sounds.

3.1 Pressure Transducers

3.1.1 LVDT pressure transducers

An **LVDT** (linear variable differential transformer) is an electromechanical sensor used to convert mechanical motion or vibrations, specifically rectilinear motion, into a variable electrical current, voltage or electric signals, and the reverse.

An extremely useful phenomenon frequently utilized in designing displacement pick up units is based upon variations in coupling between transformer windings, when the magnetic core of the transformer is displaced with respect to the position of these two windings.

Generally, a differential transformer (Fig. 3.1) designed on this principle is employed for the measurement of physiological pressures. The central coil is the energizing or primary coil connected to a sinewave oscillator. The two other coils (secondary coils) are so connected that their outputs are equal in magnitude but opposite in phase.

With the ferromagnetic core symmetrically placed between the coils, and the two secondary coils connected in series, the induced output voltage across them would be zero.

When the core is moved, the voltage induced in one of the secondary windings exceeds that induced in the other. A net voltage output then results from the two secondaries.

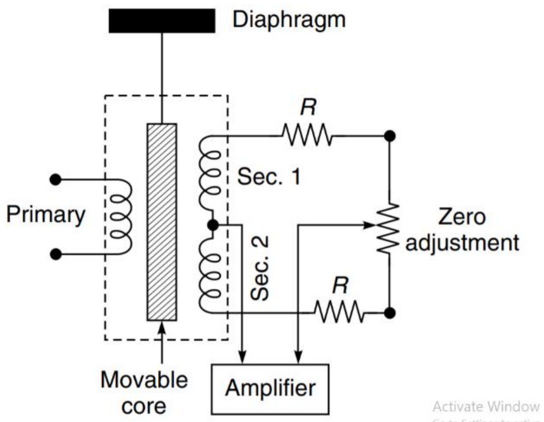


Figure 3.1: Principle of LVDT pressure transducer. The diagram shows the differential transformer and bridge circuit for detection of differential signals

The phase of the output will reverse if the core moves past the central position. A simple bridge circuit can be employed to detect the differential signals thus produced.

The signal can be further processed to directly display a calibrated output in terms of mm of displacement.

Since there is always some capacitive coupling between windings of differential transducers, it produces a quadrature component of induced voltage in the secondary windings. Because of the presence of this component, it is usually not possible to reduce the output voltage to zero unless the phase of the backing voltage is also altered.

Differential transformer displacement transducers generally work in conjunction with carrier amplifiers. Typical operating excitation of these transducers is 6 V at 3.4 kHz.

Since the output voltage of the LVDT is proportional to the excitation voltage, the sensitivity is usually defined for a 1 Volt excitation. Commercial devices typically have a sensitivity of 0.5 to 2.0 mV per 0.001 cm displacement for a 1 Volt input. Full scale displacements of 0.001 to 25 cm with a linearity of ± 0.25 % are available

3.1.2 STRAIN GAUGE PRESSURE TRANSDUCERS

A strain gauge is a device which is used to measure dimensional change on the surface of a structural member under test. Strain gauges give indication of strain at only one point.

To measure such small changes in resistance, strain gages are almost always used in a bridge configuration with a voltage excitation source. The general Wheatstone bridge, illustrated in Figure 3.2, consists of four resistive arms with an excitation voltage, V_{EX}, that is applied across the bridge.

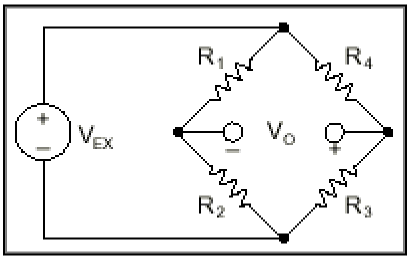


Figure 3.2 Wheatstone Bridge

The output voltage of the bridge, V_O, is equal to:

$$V_O = \left[\frac{R_3}{R_3 + R_4} - \frac{R_2}{R_1 + R_2}\right] \bullet V_{EX}$$

From this equation, it is apparent that when $R_1/R_2 = R_4/R_3$, the voltage output V_O is zero. Under these conditions, the bridge is said to be balanced. Any change in resistance in any arm of the bridge results in a non-zero output voltage.

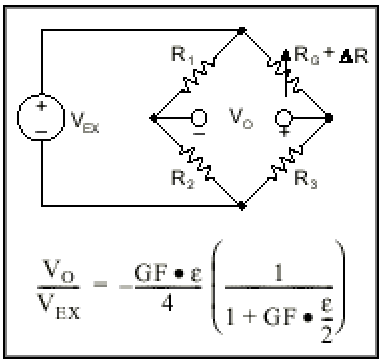


Figure 3.3. Quarter-bridge circuit

Therefore, if you replace R_4 in Figure 3.2 with an active strain gauge, any changes in the strain gage resistance will unbalance the bridge and produce a nonzero output voltage. If the nominal resistance of the strain gage is designated as R_G, then the strain-induced change in resistance, DR, can be expressed as DR = $R_G \cdot GF \cdot e$, from the previously defined Gage Factor equation. Assuming that $R_1 = R_2$ and $R_3 = R_G$, the bridge equation above can be rewritten to express V_O/V_{EX} as a function of strain (see Figure 3.3). Note the presence of the $1/(1+GF\cdot e/2)$ term that indicates the nonlinearity of the quarter-bridge output with respect to strain.

Ideally, you would like the resistance of the strain gage to change only in response to applied strain. However, strain gage material, as well as the specimen material to which the gage is applied, also responds to changes in temperature. Strain gage manufacturers attempt to minimize sensitivity to temperature by processing the gage material to compensate for the thermal expansion of the specimen material for which the gage is intended. While compensated gages reduce the thermal sensitivity, they do not totally remove it.

By using two strain gages in the bridge, you can further minimize the effect of temperature. For example, Figure 3.4 illustrates a strain gage configuration where one gage is active (R_G + DR) and a second gage is placed transverse to the applied strain. Therefore, the strain has little effect on the second gage, called the dummy gage. However, any changes in temperature affect both gages in the same way. Because the temperature changes are identical in the two gages, the ratio of their resistance does not change, the voltage V_O does not change, and the effects of the temperature change are minimized.

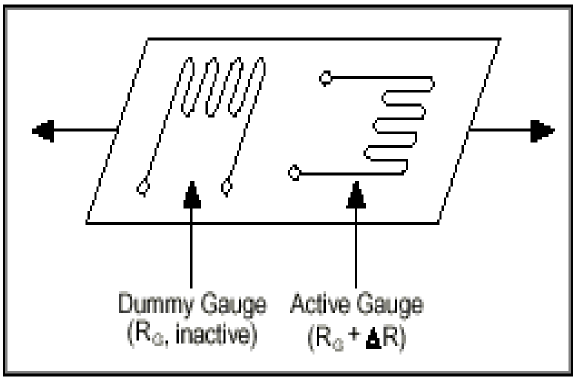


Figure 3.4. Use of Dummy Gage to Eliminate Temperature Effects

The sensitivity of the bridge to strain can be doubled by making both gages active in a half-bridge configuration. For example, Figure 3.5 illustrates a bending beam application with one bridge mounted in tension (R_G + DR) and the other mounted in compression (R_G -

DR). This half-bridge configuration, whose circuit diagram is also illustrated in Figure 3.5, yields an output voltage that is linear and approximately doubles the output of the quarter-bridge circuit.

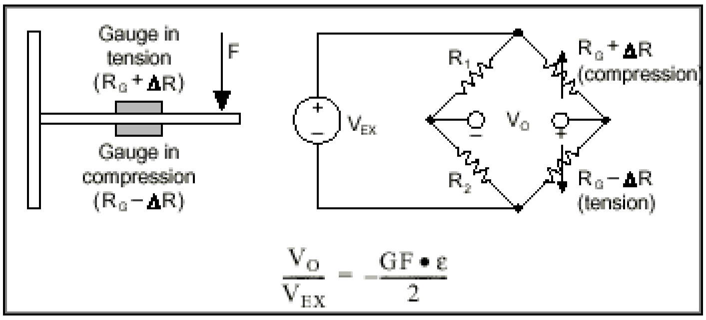


Figure 3.5. Half-Bridge Circuit

Finally, you can further increase the sensitivity of the circuit by making all four of the arms of the bridge active strain gages in a full-bridge configuration. The full-bridge circuit is shown in Figure 3.6.

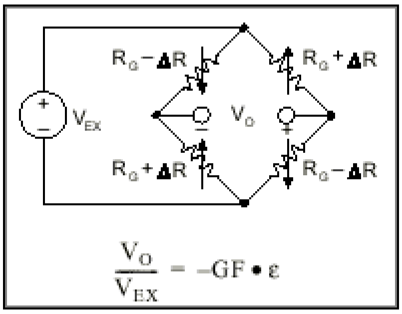


Figure 3.6. Full-Bridge Circuit

The equations given here for the Wheatstone bridge circuits assume an initially balanced bridge that generates zero output when no strain is applied. In practice, however, resistance tolerances and strain induced by gage application generate some initial offset voltage. This initial offset voltage is typically handled in two ways. First, you can use a special offset-nulling, or balancing, circuit to adjust the resistance in the bridge to rebalance the bridge to zero output. Alternatively, you can measure the initial unstrained output of the circuit and compensate in software. This topic is discussed in greater detail later.

The equations given above for quarter-, half-, and full-bridge strain gage configurations assume that the lead wire resistance is negligible. While ignoring the lead resistance may be beneficial to understanding the basics of strain gage measurements, doing so in practice can be a major source of error.

3.2 PHYSIOLOGICAL PRESSURE RANGES AND MEASUREMENT SITES

Pressures in the human body are measured as a part of clinical examinations and for physiological studies. Figure 3.7 shows the ranges of pressure in normal and abnormal situations.

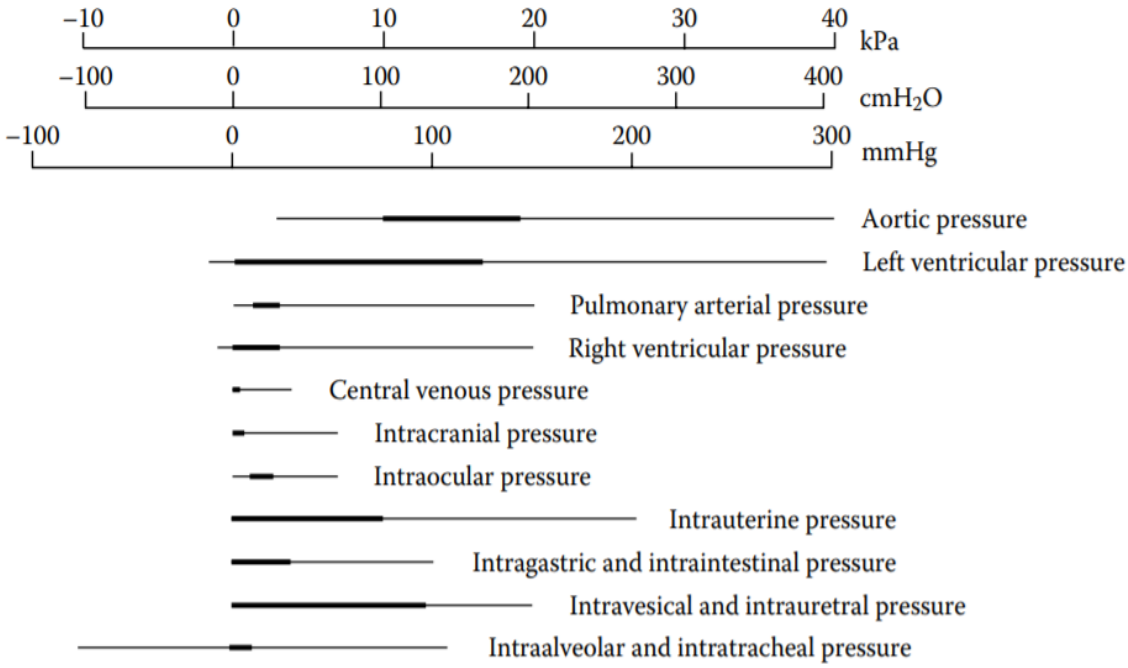


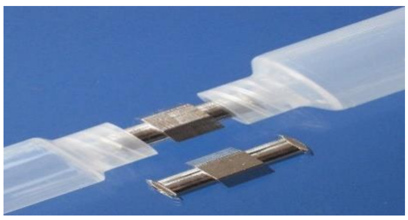
Figure 3.7 Variable ranges of pressures in the body cavities in physiological conditions (thick lines) and unphysiological conditions (thin lines).

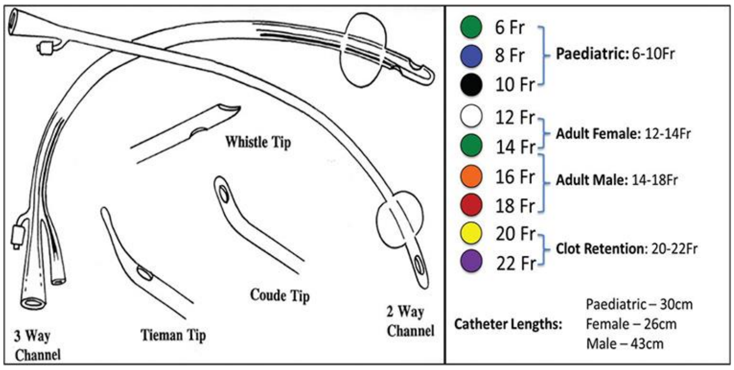
3.3 DIRECT PRESSURE MEASUREMENT

Catheters for pressure measurement

Catheters and needles of different sizes can be used for pressure measurements. A typical setup for cardiovascular pressure measurement is a flexible plastic catheter with a luer-lock connector at the proximal end, so that it can be connected to a stop cock or to other instruments.

The size of the catheter is sometimes denoted in French scale (Fr or F, each unit being equivalent to 0.33mm in outer diameter). Catheters are available in different designs. Some have openings only at the tip, and some have an open or closed tip with one or many side holes, and others have a double lumen with one opening at the tip and the other at some distance from the tip.





A catheter with a balloon near the tip is used for pulmonary arterial measurement. The balloon is carried forward naturally by the bloodstream, and this makes it possible to place the catheter in the pulmonary arteries

X-ray monitoring is sometimes necessary during catheter insertion. In such cases, the catheter should be X-ray opaque. When the catheter is placed in a blood vessel, care should be taken to prevent blood coagulation, which not only disturbs pressure measurement but can also cause serious thromboembolism.

Although catheters used for these purposes are specially made, no material is completely anticoagulant. Thus, a slow infusion of an anticoagulant agent in saline is recommended for long-term measurements. Typically, saline containing about 2000 units of heparin per litre can be continuously infused at a rate of about 3–6mL/h.

3.4 DIAPHRAGM DISPLACEMENT TRANSDUCERS

Most pressure sensors for direct pressure measurements have an elastic diaphragm, and its displacement or strain is detected by a sensing element such as the strain gauge or a variable capacitance. Although the amount of deformation of the diaphragm due to the applied pressure is nonlinear, it can be regarded as linear when the diaphragm is thin and the deformation is small compared with the thickness of the diaphragm. In a circular flat diaphragm with clamped edges, displacement of the diaphragm at a distance r from the center is given as

$$z(r) = \frac{3(1-\mu^2)(R^2 - r^2)\Delta P}{16Et^3}$$

-------------3.1

where μ is Poisson's ratio
R is diaphragm radius
t is diaphragm thickness

ΔP is pressure difference
E is Young's modulus
The displacement is maximum at the center, which can be written as

$$z(0) = \frac{3(1-\mu^2)R^4\Delta P}{16Et^3}$$

-------------3.2

The strain of the diaphragm in a radial component εr and a tangential component εt is expressed as

$$\varepsilon_r(r) = \frac{3\Delta P(1-\mu^2)}{8t^2E}(R^2 - 3r^2)$$

-------------3.3

$$\varepsilon_t(r) = \frac{3\Delta P(1-\mu^2)}{8t^2E}(R^2 - r^2)$$

-------------3.4

These strain components are equal at the center i.e.,

$$\varepsilon_r(0) = \varepsilon_t(0) = \frac{3\Delta P\,(1-\mu^2)R^2}{8t^2E}$$

-------------3.5

Figure 3.4 shows distributions of diaphragm displacement and two components of the strain. Volume displacement, which is defined as the volume change caused by the deformation of the diaphragm, is given as

$$V = \frac{\pi(1-\mu^2)R^6\Delta P}{16Et^3}$$

-------------3.6

In Equations 3.1 through 3.5, the dimension of ΔP and E cancel out each other, and all remaining variables R, r, and t have a dimension of length. Thus, as long as the same unit is used consistently for these variables, any unit such as m, cm, or mm can be used in these equations.

These relations provide the fundamental characteristics for a sensor design. The displacement and strains of the diaphragm for a given pressure depend on the geometry of the diaphragm. As seen in Equations 3.1 through 3.5, the sensitivity of the sensor is determined when the geometry and the components of the diaphragm material are given.

However, if the thickness changes in proportion to the radius, Equation 3.5 does not change. The ratio of displacement at the center to the radius, Z (0)/R, is also unchanged as long as R/t remains constant.

In other words, for geometrically similar diaphragms of different sizes, the strains and the portion of displacement relative to the radius are unchanged, assuming that the material of the diaphragms is the same.

Thus, pressure sensors of different sizes which have equal sensitivity can be constructed using the same material and a geometrically similar design. As long as the sensitivity is the same, a smaller diaphragm is advantageous, since the volume displacement is reduced in proportion to the third power of the radius if R/t is unchanged, as seen in Equation 3.6.

Very small pressure sensors have been made by means of silicon micromachining technology. The lower limit of the size is determined by the noise level due to the Brownian motion of molecules. This effect, however, is insignificant in physiological pressure ranges, even for a diaphragm with a diameter of 0.1mm.

Different principles can be used to detect displacement or strains of the diaphragm. Strain gauge and capacitive methods are most commonly used in pressure sensors used for physiological measurements. In catheter tip pressure sensors, optical methods are also used. The strain gauge type has been widely used.

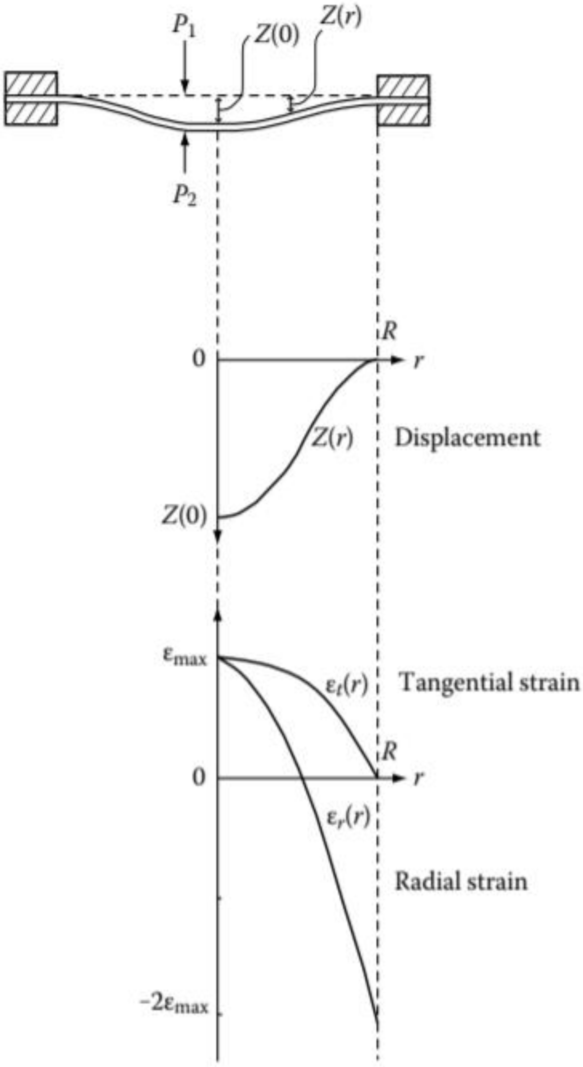


Figure 3.8 Deformation of a thin circular diaphragm with clamped edge, distribution of diaphragm displacement, Z(r), and that of tangential and radial stress components, εt (r), εr(r).

This principle utilizes metal and semiconductor elements in which electrical resistance varies with strain. Although the relation between electrical resistance and strain is nonlinear, the relation can be regarded as linear when the strain is less than 0.5%. If the length and its change are L and ΔL, and the electrical resistance and its change are R and ΔR, their ratio, G, is defined by

$$G = \frac{\Delta R/R}{\Delta L/L}$$

-----------3.7

and it is constant. This is called the gauge factor.

The gauge factor G for metals is about 2.0, and is dependent only on the dimensional change, while semiconductors have larger gauge factors, between −100 and 140, where resistance change due to the piezoresistive effect is added.

As mentioned earlier, the sensitivity of the diaphragm type pressure sensor can be estimated from its geometry and the materials used in it. For example, if a steel diaphragm is used, Young's modulus is roughly 2 × 1011 N/m2, Poisson's ratio is 0.3, and if the radius is 5mm and the thickness of the diaphragm is 0.1mm, then, from Equation 3.2, the displacement at the center is estimated at approximately 0.007mm for an applied pressure of 13kPa (100mmHg).

When a metal strain gauge of 10mm in length is used to detect displacement, the strain is about 0.07%, and if the gauge factor is 2, then the ratio of change in electrical resistance is 0.14%.

A bridge with these gauges gives an output voltage of about 7mV for 13kPa (100mmHg) at 5V excitation. The volume displacement is also estimated at about 0.7mm3 for a pressure change of 13 kPa (100mmHg) from Equation 3.6. When a catheter and pressure sensor are connected to the patient, all surfaces exposed to body fluids should be sterile.

However, it is not convenient to have to frequently sterilize the whole sensor assembly. To avoid having to do so, a disposable dome is used in which a thin plastic membrane separates the diaphragm of the pressure sensor from the body fluids inside the dome. The semiconductor strain gauge is also used in miniature clinical pressure sensors.

The major advantage of this strain gauge-type is that the gauge pattern can be fabricated on the silicon substrate using ordinary integrated circuit processing technology. In addition, the elastic beam or diaphragm can be fabricated in the same silicon substrate. Using silicon micromachining technology, very small sensing elements can be realized.

This technique also diminishes mechanical or thermal instabilities caused by the bonding of strain gauge on an elastic material. Figure 3.5 shows a cardiovascular pressure sensor (Astro-Med P1OEZ, West Warwick, RI) in which a beam-type semiconductor strain gauge element is used to detect displacement of the metal diaphragm.

The silicon diaphragm on which strain gauges are fabricated is also used in clinical pressure sensors such as disposable sensors and catheter-tip sensors. Figure 3.9 shows an example of the disposable pressure sensor (Cobe Laboratories, CDXIII, Inc., Lakewood, CO) with cross sections of the silicon diaphragm and the tube with the sensing element (Spotts and Frank 1982).

To detect diaphragm displacement, a capacitive method can be used. If two plate electrodes are arranged in parallel and close enough to each other so that the effect of fringing electric fields is negligible, the capacitance is expressed as

$$C = \frac{\varepsilon A}{d}$$

--------------3.8

where ε is the permittivity of the medium (8.85 × 10−12 F/m in air)
A is the area
d is the separation

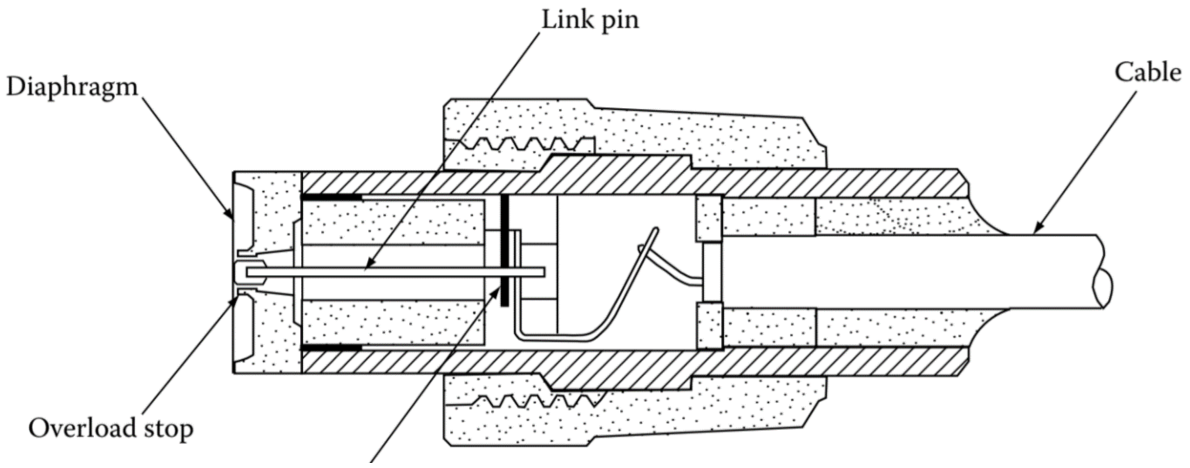


Figure 3.9 A pressure sensor using a beam type silicon strain gauge.

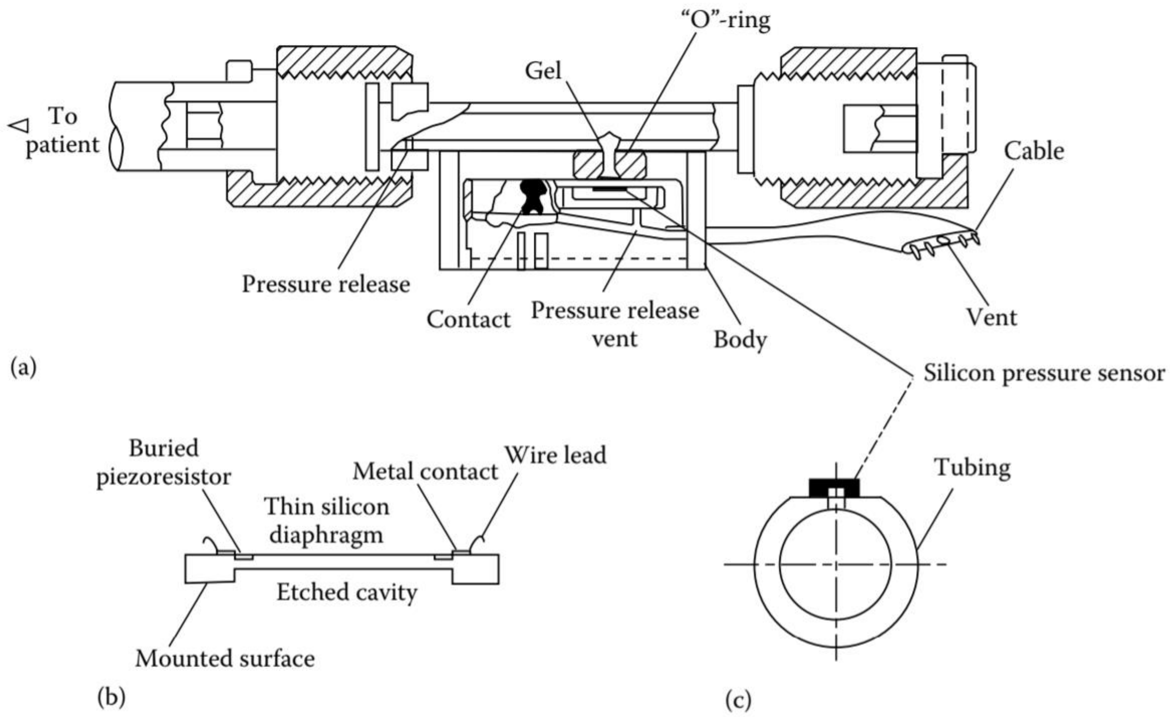


Figure 3.10 A disposable pressure sensor (Cobe CDXIII, Lakewood, CO) (a) the silicon-diaphragm type pressure sensor (b) and its attachment to a tubing

If one electrode is attached to the diaphragm, and the other is fixed to the housing as shown in Figure 3.10 a, the distance d varies according to the displacement of the diaphragm. Although the capacitance is a nonlinear function of d in Equation 1.9, the linear output can be obtained by a simple circuit as shown in Figure 3.10b. When a sinusoidal excitation voltage Vi e jωt is applied, and the gain of the amplifier is large enough, the sum of currents at the input part of the amplifier should be zero and the potential becomes almost zero due to the negative feedback operation. Thus,

$$j\omega C_i V_i e^{j\omega t} + j\omega C_x V_0(t) = 0$$

----------- 3.9

where ω is angular frequency of excitation. Then the output voltage V0(t) is given as

$$V_0(t) = -\frac{C_i}{C_x} V_i e^{j\omega t}$$

--------------3.10

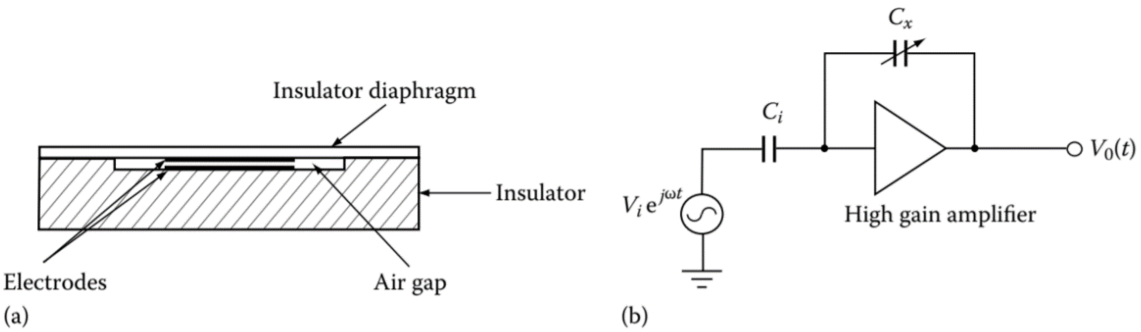


Figure 3.11 A capacitive pressure sensor (a) and a circuit which provides linear output to the displacement (b).

Substituting (3.10), output is expressed as

$$V_0(t) = -\frac{d}{\varepsilon A} C_i V_i e^{j\omega t}$$

---------3.11

which is a linear function of the distance d.

3.5 CATHETER TIP PRESSURE TRANSDUCERS

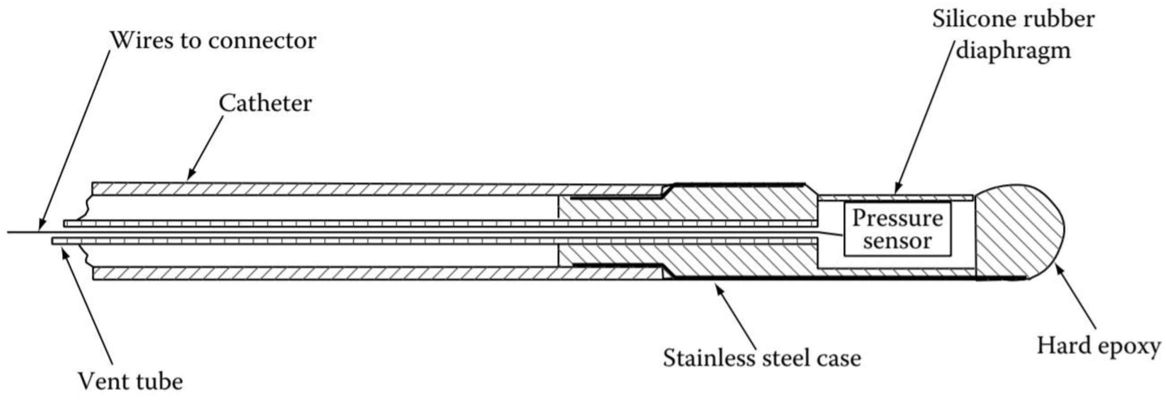


Figure 3.12: A catheter-tip pressure sensor with a beam-type silicon strain gauge.

The catheter-tip pressure sensor, which has been developed primarily for accurate measurement of pressure waveforms in the cardiovascular system, has a pressure sensing element at the tip of a catheter.

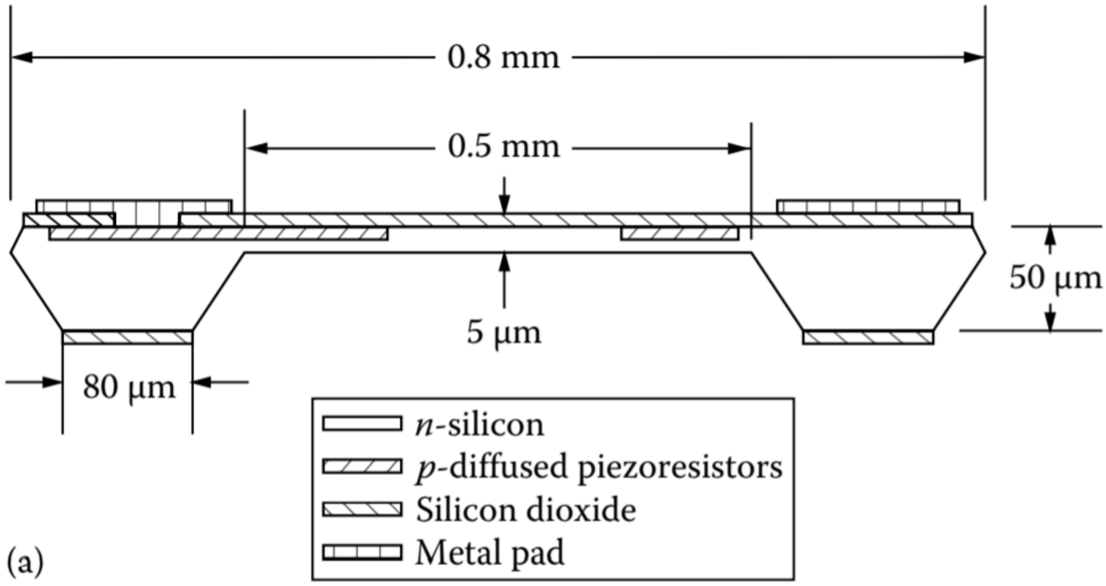
Progress in integrated circuit technology has made possible the fabrication of very small strain gauge elements on the silicon tip. Micromachining can be used to fabricate small, thin diaphragms, beams or cantilevers of different shapes that can be applied to catheter-tip pressure sensors.

Catheter-tip pressure sensors with beam-type silicon strain gauges which detect the side pressure via a silicone-rubber diaphragm (Millar and Baker 1973) have been used extensively. Figure 3.12 shows a cross sectional view of a commercial model.

The catheter has a vent tube which connects the rear side of the diaphragm to the atmosphere so that it can measure pressure relative to the atmospheric pressure.

The catheters are manufactured in different sizes from 3F to 8F (OD 1.0–2.67mm), the nominal pressure range is from −6.5 to 40 kPa (−50 to +300mmHg), and the resonance frequency is from 35 to 50 kHz.

Figure **3.13** a shows a cross section of a silicon diaphragm, and Figure **3.13** b shows the catheter-tip on the end of which the diaphragm is mounted. The silicon diaphragm with diffused strain gauges has also been used in catheter-tip pressure sensors for side pressure or end pressure measurements



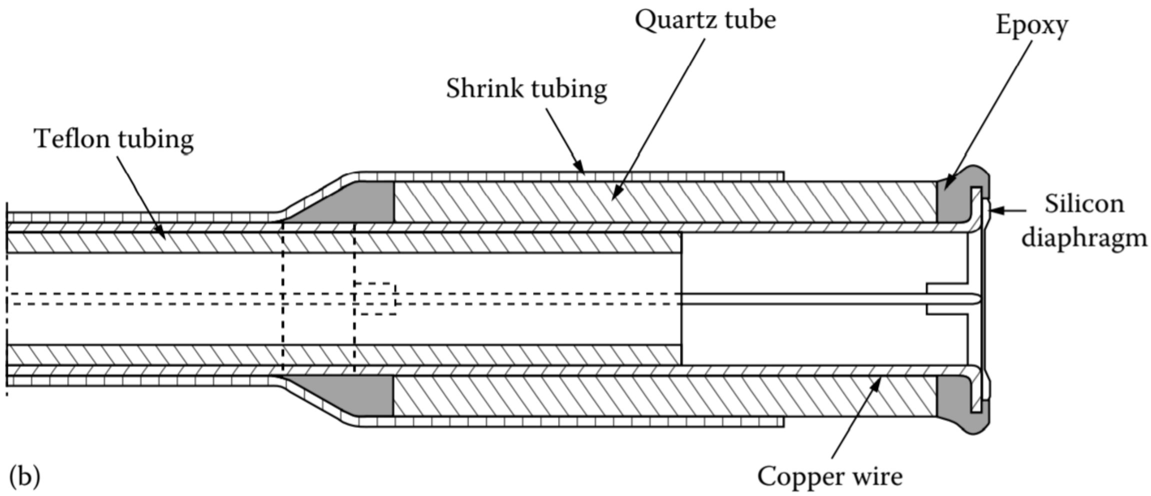


Figure 3.13 Cross section of a silicon diaphragm pressure sensor (a) and a catheter-tip pressure sensor having the diaphragm in its end (b). (Reproduced from Samaun, T. et al., IEEE Trans. Biomed. Eng., 20, 101, 1973.)

Fibreoptic catheters for pressure measurement

Optical fibres transmit light through so-called cladded fibres made with transparent material such as glass or plastic in which fibres with a core having a high reflective index are surrounded by a shell having a low reflective index.

The light transmits through the core by total reflection at the boundary of these two media. In a fibreoptic catheter, light from a source is transmitted to the tip through the optical fibre.

The reflected light from a reflector, which moves in relation to the applied pressure, enters the fibre and the intensity of the reflected light is measured by a photo detector.

There are two different types: one measures the end-pressure, while the other measures the side pressure. Refer the figure for the pictorial View.

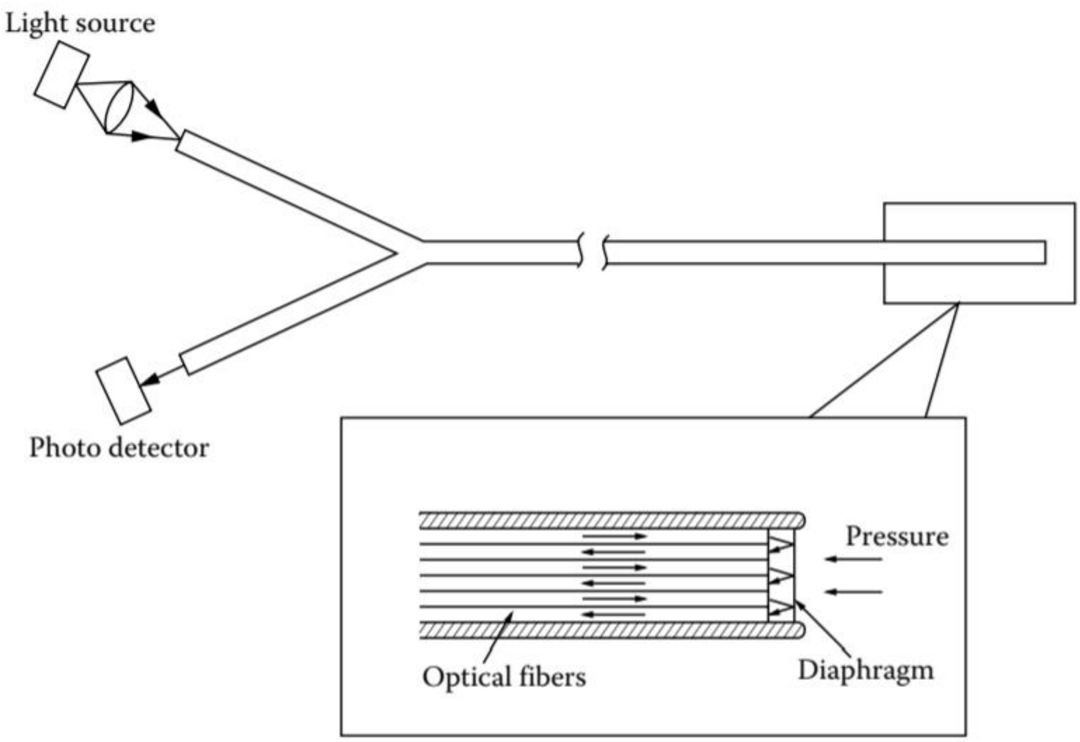


Figure 3.14 A fibre-optic end-pressure sensor.

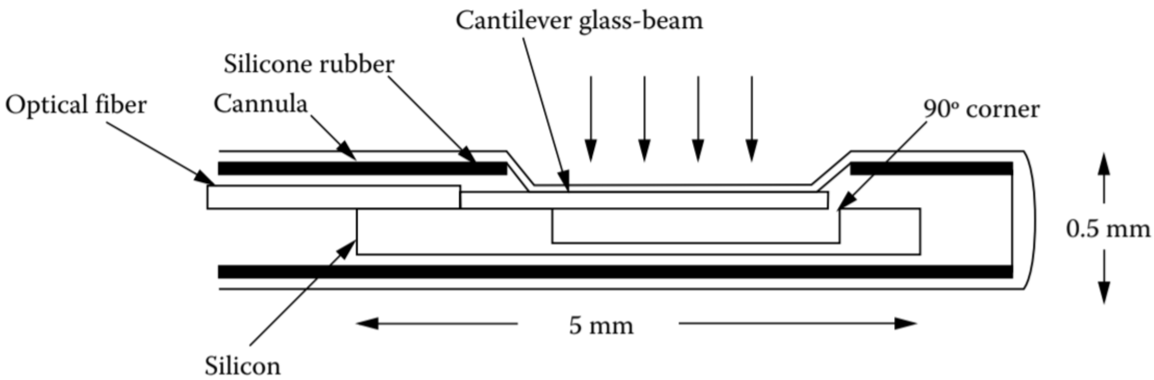


Figure 3.15 A fibre-optic side-pressure sensor

3.6 IMPLANTABLE PRESSURE TRANSDUCERS AND PRESSURE TELEMETERING CAPSULES

For long monitoring of pressure in the body cavities, it is operable to use a pressure transducer that is separated from external instruments so that it can either be surgically implanted (as in the case of the intracranial pressure measurements) or swallowed (for pressure measurements in the gastrointestinal tract).

IMPLANTABLE PRESSURE TRANSDUCER

Implantable pressure transducers are primarily used for intracranial and cardiovascular pressure measurements.

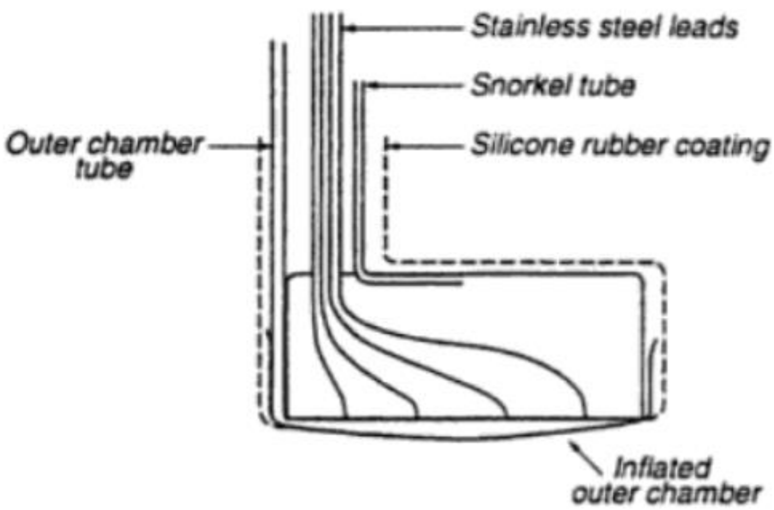
Figure 3.16 a and **3.16 b** are constructions of implantable pressure transducers.

Fig 3.16 a has a diaphragm with strain gauges and an outer chamber made of thin plastic film. By inflating the outer chamber and also applying the same pressure to the rear side of the diaphragm, the baseline reading can be checked in vivo.

In the **3.16 b** transducer, pressure is detected by a capacitive change and it is transmitter and battery for telemetering.

The capacitance chamber is designed so that displacement of the diaphragm is limited by a contact to the opposite wall.

By applying a negative pressure through the vent tube just until the limit of the displacement is reached, the output can be calibrated in vivo



(a)

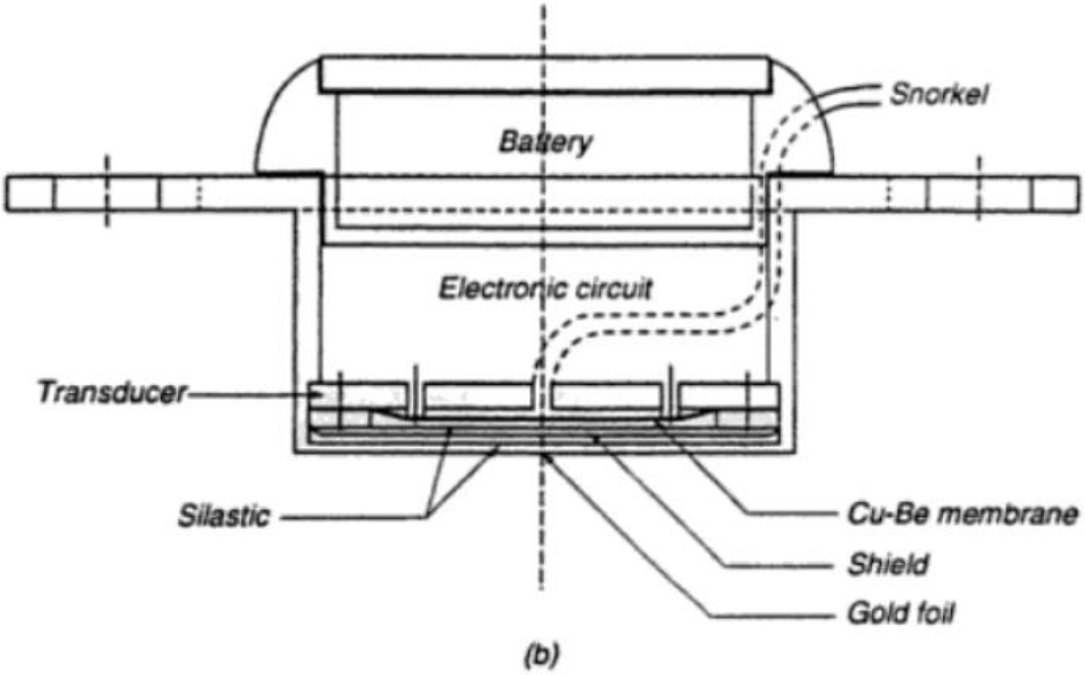


Figure 3.16 a and b

Figure 3.17 shows an example of intracranial pressure transducers which measure absolute pressure. A disc shape absolute pressure transducer is assembled in a stainless-steel housing which fits to the drilled hole in the skull

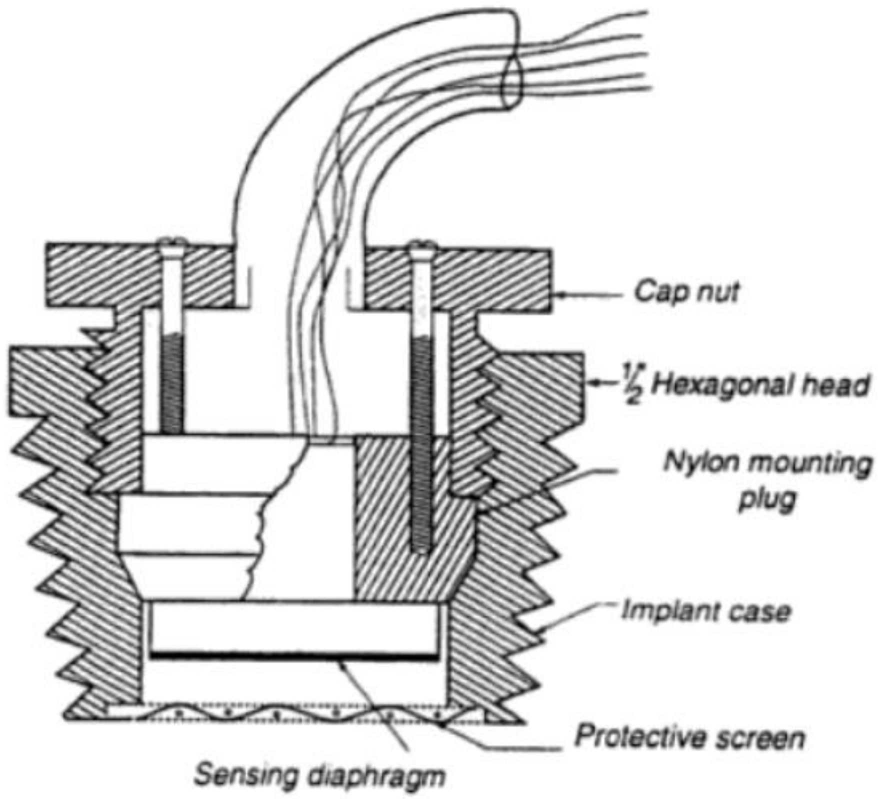


Figure 3.17 Intracranial pressure transducers

PRESSURE TELEMETERING CAPSULES

Swallowable capsule for gastrointestinal pressure telemetry is an example for telemetering capsules.

Most of the capsules consists of pressure transducer, amplifier, radio transmitter and battery.

Passive telemetry, in which electric power is supplied from outside the body by electromagnetic induction, has also been attempted.

Size of these capsules were 0.7 to 10mm in diameter, and 19 to 30mm in length.

Construction of two examples of pressure capsules are shown in the figure 3.18

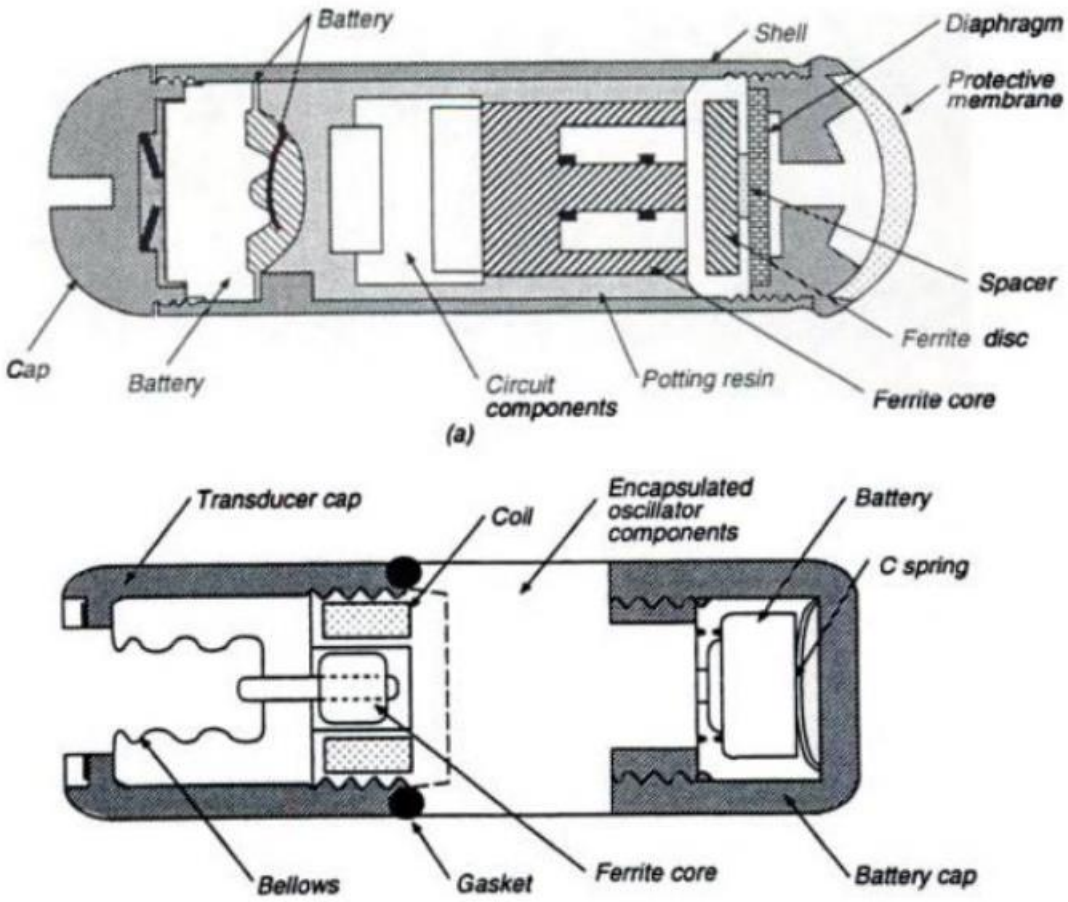


Figure 3.18: Two Pressure Capsules

Pressure is detected by the change in inductance in a resonant circuit and thus it modulates the oscillation frequency.

The capsule can measure intestinal pressure from 0 to 200 kPa for up to about 5 days

The above 2 capsules had difficulty in recovering it for repeated use.

The same problem is solved by an inexpensive single use capsule become available. These pressure measurement capsule has been used for uterine contraction and fetal heart-sound measurements by introducing it into the uterine cavity during labor.

3.7 INDIRECT PRESSURE MEASUREMENT-INDIRECT MEASUREMENT OF SYSTOLIC, DIASTOLIC, AND MEAN BLOOD PRESSURE

All occluding cuff techniques for indirect blood pressure measurement employ the inflatable cuff which is wrapped around an extremity, typically the upper arm.

When the cuff is inflated, full cuff pressure is transmitted to the tissue around the artery, as long as the cuff size is adequate.

The lumen of the artery will open and close following the positive and negative transmural pressure, which is the pressure difference between the inside and outside of the vessel.

Thus, by varying the cuff pressure gradually and detecting the pressure point at which the lumen is just opened or closed, intravascular pressure can be measured from the cuff pressure.

To detect the lumen opening, the auscultatory techniques as well as the automated technique based on the Korotkoff sounds principle have been widely used, while other techniques such as oscillometric methods have also been employed.

In these indirect arterial pressure measurements, the accuracy of measurements is affected mainly by the design of the occluding cuff and the procedure used for detecting the vascular opening.

Cuff Design for Indirect Blood Pressure Measurements

The conventional sphygmomanometer cuff currently used for measuring indirect blood pressure in the upper arm consists of an inflatable bladder within a restrictive cloth sheath as shown in Figure 3.18.

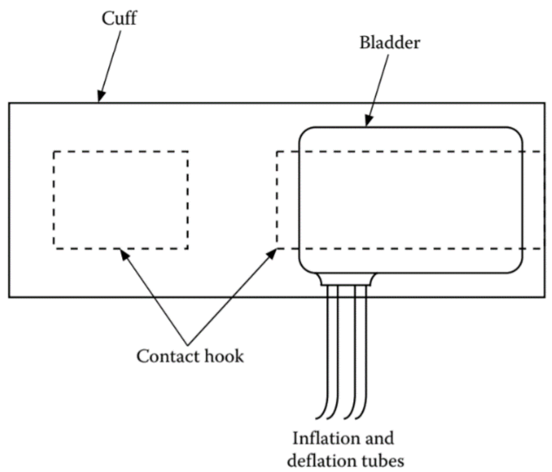


Figure 3.18: The Sphygmomanometer Cuff

The size of the inflatable bladder should be large enough so that the pressure in the bladder fully transmits into the underlying tissue containing the large artery.

According to the standards, the width of the bladder should be 40% of the circumference of the midpoint of the limb (or 20% wider than the diameter), and the length should be twice the recommended width.

Recommendations are also given for bladder diameters for blood pressure cuffs as shown in Table below.

Recommended Bladder Dimensions for Blood Pressure Cuff

Arm Circumference at Midpoint[a] (cm)	Cuff Name	Bladder Width (cm)	Bladder Length (cm)
5–7.5	Newborn	3	5
7.5–13	Infant	5	8
13–20	Child	8	13
17–26	Small adult	11	17
24–32	Adult	13	24
32–42	Large adult	17	32
42–50[b]	Thigh	20	42

3.8 DETECTION OF KORTOKOFF SOUNDS.

The auscultation method of indirect blood pressure measurement is based on the fact that when the occluding cuff pressure is slowly reduced from a pressure above systolic pressure to a pressure below diastolic pressure, acoustic waves called Korotkoff sounds are generated while the cuff pressure remains between the systolic and diastolic pressures.

The origins of Korotkoff sounds are not completely understood, Despite the lack of theoretical basis, Korotkoff's method has been widely accepted and the techniques have been standardized.

In blood pressure measurement using the conventional sphygmomanometer, a stethoscope is placed on the brachial artery close to the distal side of the cuff. The cuff pressure is raised to about 4kPa (30mmHg) above the estimated systolic pressure, and then reduced at a rate of 0.26–0.4 kPa/s (2–3mmHg/s).

The systolic pressure is given as the cuff pressure when clear tapping sounds first appear, and the diastolic pressure is given as the cuff pressure when the sounds disappear. This technique was first proposed by Korotkoff in 1905

Korotkoff sounds can be detected by microphones, instead of a stethoscope, placed at the distal side of the cuff, but more conveniently, the microphone can be placed beneath the cuff or even in the cuff bladder.

Figure 3.19 shows the typical data of relative sound intensities and fundamental frequencies in normotensive subjects when the cuff pressure is reduced according to the conventional auscultatory technique.

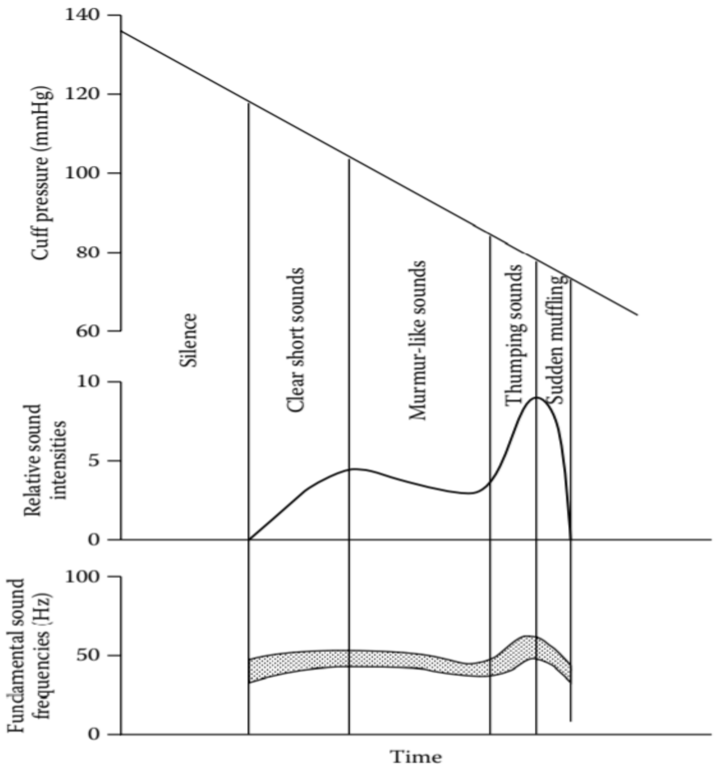


Figure 3.19: Characteristics of Korotkoff sounds in the typical normotensive subject

As seen in the figure, principal components of Korotkoff sounds are low in frequency, with a peak at about 45Hz, but higher harmonic components are also present, and the major frequency components fall in a band from about 20 to 300Hz.

In automated detection systems based on Korotkoff sounds, the effect of noise is a serious problem. While the effect of ambient noise can be reduced by using an appropriate band-pass filter or covering the microphone with the cuff, motion artifact is more difficult to avoid because it can exist in the same frequency band as the major component of Korotkoff sounds.

Motion artifacts can be induced by movements of the whole body, arm, hand, or even finger, and are difficult to avoid completely. To reduce motion artifacts, taping a microphone to the upper arm using micropore surgical tape has been recommended.