# MODULE – 4

# Temperature Measurement, Transducers and Sensors



## *4.1 Requirements for Measurement Ranges:*

- Temperature is measured at many different sites of the body for clinical diagnosis and patient monitoring. In humans and in homeothermic animals, the temperature of the central part of the body is stabilized by a physiological thermoregulatory function, and the deep tissue temperature at the central part of the body is called core temperature or deep body temperature.
- The term "body temperature" is often used to indicate the core temperature, even though the temperature of the body is not uniform but can vary from site to site.
- The core temperature always remains in a range from 35°C to 40°C. Most physiological and pathological temperature variations occur in this range, from the lowest temperature in early morning or in cold weather to the highest one during febrile disease or hard exercise.
- A temperature resolution of 0.1°C is generally required in core temperature measurement, and that of 0.05°C is sometimes

required such as in basal temperature measurement. An absolute accuracy of 0.1°C is acceptable for most purposes.
*(Basal Body Temperature (BBT) is your body's temperature when at complete rest)

- Skin temperature is also measured in physiological studies, clinical diagnosis, and patient monitoring. To estimate heat exchange between the body and its environment, the mean skin temperature is considered. Mean skin temperature– *T*s is defined as the sum of the products of the area of each regional surface element *Aj* and its mean temperature *Tj* divided by the total area of the body surface *A*b

$$\overline{T_S} = \frac{\sum_j (A_j \overline{T_{SJ}})}{A_b}$$

- While many different weighting systems have been proposed in which different measurement sites and corresponding weights *Wj* are assigned, a comparative study showed that most of them agreed with one another within 1°C over 85% of measurements in a wide range of controlled environments.
- The temperature of sweating skin sometimes drops below the ambient temperature and, at the lowest, falls to the dew point. Skin temperature may vary much more widely when the skin is cooled or warmed externally.

*Ambient temperature is the air temperature of an environment or object*

- The resolution, absolute accuracy, and response time of the thermometer required for measuring skin temperature vary according to the purpose, but a comparable performance of the thermometer for core temperature measurement may be acceptable in most applications.
- Local temperature measurements in the tissue are sometimes required. At thermal equilibrium, the temperature of a tissue is determined by local heat production and heat transport into and from the site. Metabolically active tissues have a higher

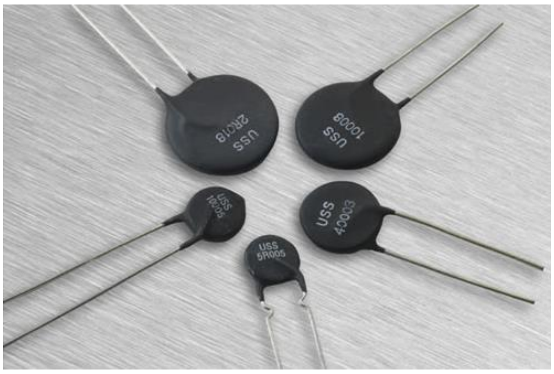
temperature than other sites, and can maintain temperatures higher than the arterial blood temperature.

- In hyperthermia cancer therapy, local temperature measurement of tissue is required. To achieve maximum therapeutic effect, the temperature is maintained at about 43°C, which is close to the limit of survival for normal cells, while on the other hand, cancer cells can survive at a slightly lower temperature.
- In a cold environment, a much higher heat dissipation will occur, and hence the heat dissipation measurement from the body surface requires a range from a few watts per square meter to several hundred watts per square meter.
- The range required for humidity measurement is from zero up to 100% in relative humidity, or saturated water vapor pressure or concentration at the highest temperature suspected during measurement. At 37°C, absolute humidity is 43.83 g/m3 when water vapor is saturated and saturated water vapor pressure is 6279 Pa (47.1 mmHg).
- When humidity is expressed in terms of relative humidity, temperature should also be measured with sufficient accuracy. Near the body temperature, saturated water vapor pressure varies largely with temperature so that a temperature change of 1°C may cause changes of more than 5% in relative humidity.

# *4.2 Temperature Sensors:*

Many different kinds of temperature sensors are available that are used by themselves or installed in surface probes, catheters, or needles making contact with or introduced to the object site of the body. To choose one suitable for medical thermometry, it may be worthwhile to compare different kinds of temperature sensors of different principles. There are always several alternatives in the choice of sensing devices.

## *4.2.1 Thermistors:*



A thermistor is a semiconductor-resistive temperature sensor made by sintered oxides of metals such as manganese, cobalt, nickel, iron, or copper.

*Sintering process of compacting and forming a solid mass of material by heat or pressure without melting it to the point of liquefaction*

The resistance of a thermistor has a negative temperature coefficient, typically about −0.04/K.

Compared with a platinum wire resister, which has a temperature coefficient of about 0.0039/K, the sensitivity of a thermistor is about 10 times that of a platinum wire temperature probe, and hence, a thermistor is suitable for use in physiological temperature measurement, where relatively higher resolution is required in a narrow temperature range.

The resistivity of a thermistor material, ρ, at an absolute temperature *T* is generally expressed as:

$$\rho\ \alpha\ \exp\left(\frac{E_g}{2kT}\right) \tag{4.1}$$

where
*E*g is the band gap energy of the semiconductor
*k* is the Boltzmann constant

Thus, if the resistance of a thermistor is *R*0 at a temperature *T*0, then the resistance at temperature *T* is expressed as:

$$R(T) = R_0 exp\left(\left[\frac{1}{T} - \frac{1}{T_0}\right] B\right) \tag{4.2}$$

Where
*B* = *E*g/2*k* is a constant that depends on the thermistor material, and has the dimension of temperature. *B* always remains in a range from 1500 to 6000 K.

Temperature coefficient α of a thermistor is derived from the above expression as:

$$\alpha = \frac{1}{R}\frac{dR}{dT} = \frac{d}{dT}\left(\frac{B}{T}\right) = -\frac{B}{T^2} \tag{4.3}$$

This means that the temperature coefficient is negative and is temperature dependent. If *B* = 4000 K, α is about −0.0416/K at 37°C. Commercial thermistors for general use have resistance ranging from 6 to 60 kΩ at 0°C, or from 15 to 150 Ω at 37°C.

Thermistors having a much higher resistance are sometimes required for use in instruments to be operated at lower power. For this purpose, thermistors for high-temperature measurement can be used among which thermistors having a resistance of about 1 MΩ at room temperature are available.

Various types of thermistor probes for general use or for medical use are commercially available.

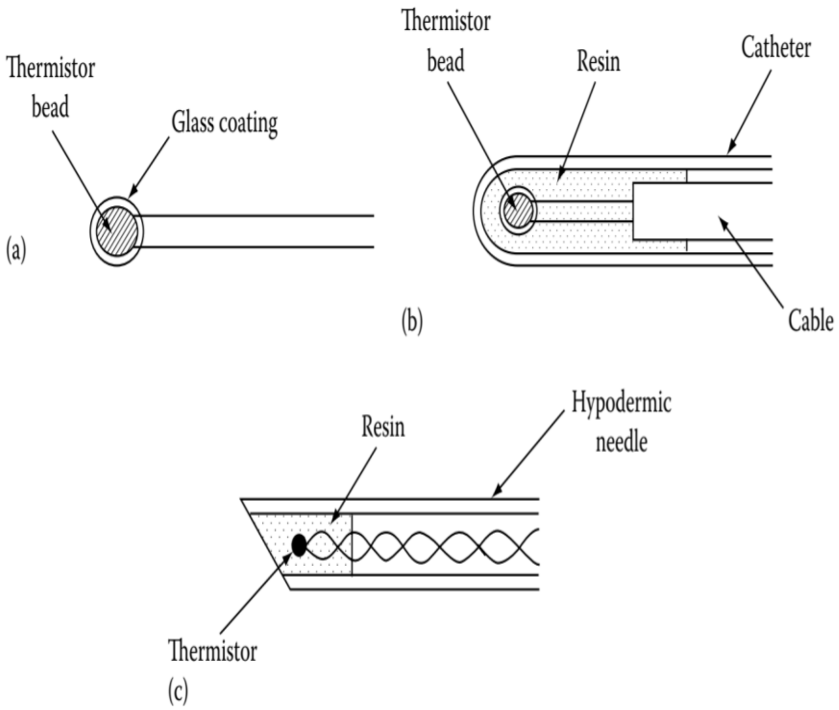



**FIGURE 5.1** Examples of thermistor probes: (a) glass-coated thermistor bead with bare wires, (b) catheter-type probe, and (c) needle-type probe.

Figure 5.1 shows examples of thermistor probes. Figure 5.1a is a glass-coated thermistor bead with bare lead wires. Beads of small diameters down to about 0.3 mm are commonly available.

Figure 5.1b is a catheter-type probe in which the thermistor is connected to a flexible insulated cable, and the connected part is also insulated and completely waterproof. A needle-type probe as shown in Figure 5.1c is also available.

The response time of a thermistor probe depends on its shape, size, and covering material, and the surrounding medium.

Fine catheter-type or needle-type probes have response times of about 0.1 s or less in water, while response times increase to about 3 s or more in air.

While the characteristic of the thermistor is nonlinear, as expressed in Equation 5.8, many techniques have been proposed in order to obtain a linear output to temperature. In a narrow temperature range, linearization can be achieved practically by adding only one resistor as shown in Figure 5.2.

When a constant voltage source is used, a resistor is connected in series as (a), and when a constant current source is used, a shunt resistor is connected as (b).
To minimize measurement error in the required temperature range, resistance $R1$ to be connected to the thermistor as the circuit (a) or (b) is determined as:

$$R_1 = R\left(\frac{B-2T}{B+2T}\right)$$

(5.10)

where
$T$ is the midpoint of the required temperature range
$R$ is the resistance of the thermistor at $T$

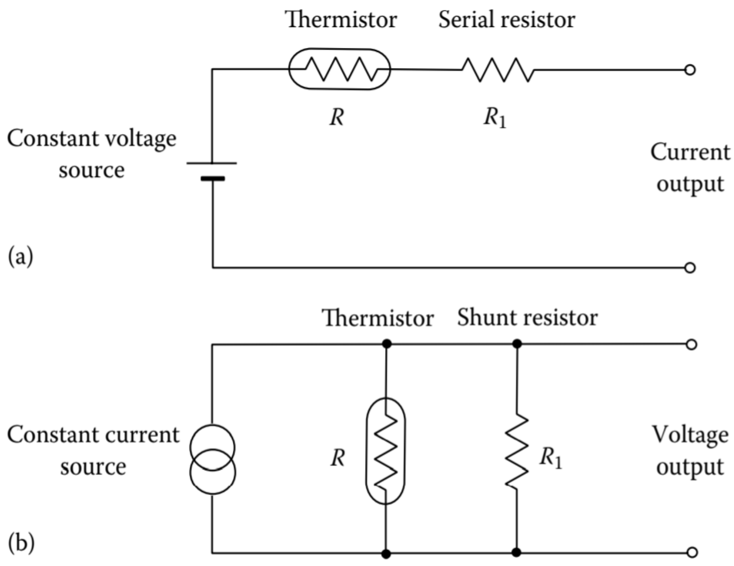



**FIGURE 5.2** Linearization circuits for thermistors: (a) using a serial resistor and a constant voltage source and (b) using a parallel resistor and a constant current source.

For example, when $B$ = 3000 K, and a measurement range from 290 to 310 K is required, the departure from linearity is estimated to be 0.03 K, and if an error of 0.1 K is permissible, a measurement range from 285 to 315 K can be covered.

## ***4.2.2 Thermocouples:***

The thermocouple is a thermoelectric sensor. A circuit composed of two dissimilar metals A and B as shown in Figure 5.3a provides an electromotive force that depends on the temperature difference between two junctions. This phenomenon is known as the Zeebeck effect.

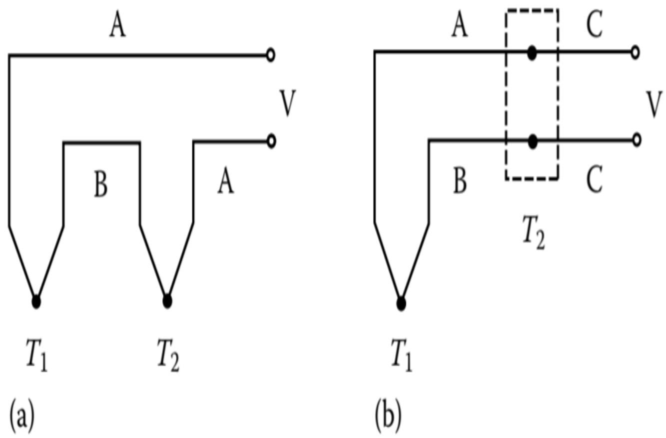



**FIGURE 5.3** Thermocouple circuit with two dissimilar metals A and B (a) and that with an additional third metal C (b).

- In the circuit shown in Figure 5.3a, when the temperature of the reference junction $T2$ is kept constant, electromotive force varies only with the temperature of the measurement junction $T1$. In the circuit shown in Figure 5.3b, a third metal C is connected to both metals A and B, and as long as the two new junctions are at the same temperature, it provides the same electromotive force as that of the former circuit shown in Figure 5.3a, regardless of the material of the third metal.

- Even though the temperature of the reference junction is kept constant, the electromotive force is nonlinear with the temperature of the measurement junction. However, departure from linearity is small within the temperature range used in medical thermometry and the error is always compensated for in commercial thermometers.
- Sensitivities of typical thermocouples in a temperature range of 20°C–40°C are
- about 41 μV/K for copper/constantan,
- about 40 μ V/K for chromel/alumel, and
- about 6.1 μV/K for platinum/platinum rhodium (10%).
- To achieve accurate measurement by a thermocouple, the temperature of the reference junction should be stable enough. The triple point of water can be an accurate reference temperature, which is 0.01°C ± 0.0005°C.
- When high absolute accuracy is not required, the constant temperature bath can be eliminated by employing a method of compensating the reference junction temperature. A convenient method is to use the circuit shown in Figure 5.3b so that the temperature of the input terminals to which the thermocouple is connected is used as the reference.
- The temperature of the measurement junction is measured as the reference junction temperature plus the temperature difference between junctions estimated by the electromotive force. While the sensitivity of a thermocouple is not constant for different reference junction temperatures, the error is small for narrow temperature ranges, and can be easily compensated for.

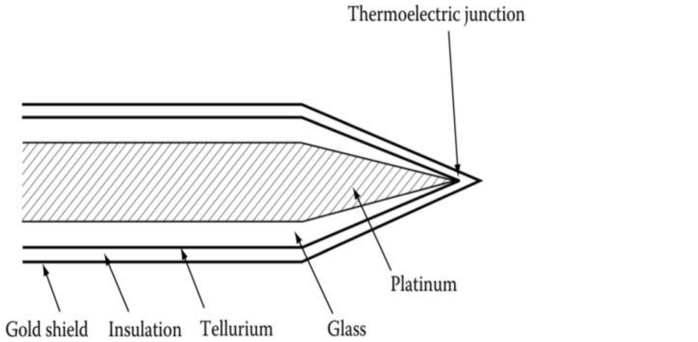





**FIGURE 5.5** Microthermocouple. (After Guilbeau, E.J. and Mayall, B.I., *IEEE Trans. Biomed. Eng.*, BME-28, 301, 1981.)

- Microthermocouples, as shown in Figure 5.5, with a junction size of about 1 μm can be made using the technique of fabricating glass-coated microelectrodes (Guilbeau and Mayall 1981). To fabricate this, a tip of thin platinum wire 25 μm in diameter was tapered by electropolishing, and a very thin glass coating on the platinum was made leaving an exposed cone of platinum.

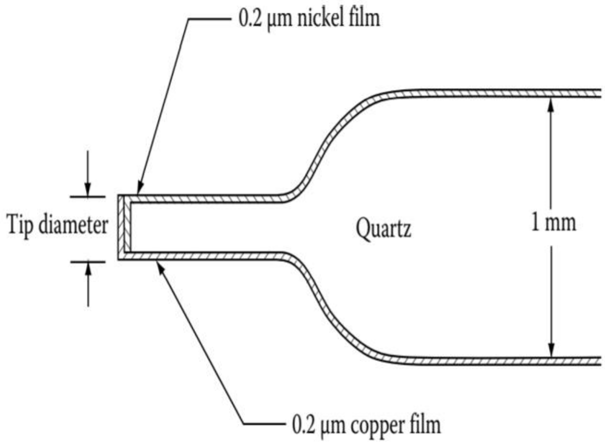



**FIGURE 5.6** An example of a thermocouple tip fabricated by thin film technique. (Reproduced from Cain, C.P. and Welch, A.J., *IEEE Trans. Biomed. Eng.*, BME-21, 421, 1974.)

- Figure 5.6 shows an example of a thermocouple tip fabricated by thin-film technique (Cain and Welch 1974). A quartz rod was tapered and a tip of about 10 μm in diameter was formed. Then a nickel layer was vacuum deposited on one side and a copper layer on the other, with the two metals overlapping only at the tip end.

## *4.2.3 Wire and Thin Film Thermoresistive Elements:*

- A thermoresistive element made from a pure metal has advantages—it has a constant temperature coefficient, linear output can be obtained in a wide temperature range, and a probe having a relatively large contact area with small heat capacity can easily be fabricated—it has been used widely in industrial applications.
- These advantages are less significant in most biomedical applications, where very local temperature has to be measured in a narrow temperature range, while it is useful in special situations where an extremely fast response is required, such as in hot-wire or hot-film anemometry.

- The most common material used as the wire or thin-film thermoresistive element is platinum. The platinum thermoresistive element has a higher stability and smaller nonlinearity than thermistors, but its temperature coefficient of about 0.0039/K is about one-tenth that of the thermistor.
- Various types of probes have been made for surface and fluid temperature measurement. Flat grid winding type probes are used for measuring surface temperature of solids, and probes having winding wires encased in protective tubes are used for fluid temperature measurement.
- The metal wire or film can also be cemented onto the surface at which temperature has to be measured However, the resistance change can occur not only due to a temperature change but also

due to a strain, and hence it may exhibit spurious output due to a differential thermal expansion or mechanical load.

## 4.2.4 p–n Junction Diodes and Transistors:

The voltage across a p–n junction at constant forward-bias current exhibits excellent linear temperature dependency over a wide temperature range, and thus any diode or transistor having a p–n junction can be a temperature sensor. It is also advantageous that a p–n junction can be fabricated on a silicon chip with interfacing circuits by integrated circuitry technology, and many convenient IC temperature sensors are commercially available.

$$I = Aexp\left(\frac{(qV-E_s)}{kT}\right) \tag{5.11}$$

Where,

$I$ is forward bias current
$A$ is constant depending on the geometry of the junction
$q$ is the electron charge
$V$ is the voltage across the junction
$Eg$ is the band gap energy
$k$ is the Boltzmann constant
$T$ is the absolute temperature

When the current $I$ is held constant, $(qV - Eg)/kT$ is kept constant, and thus, the voltage across the junction $V$ is a linear function of the absolute temperature $T$. The temperature coefficient of $V$ is always negative and observed $dV/dT$ in typical small-signal silicon p–n junction diodes ranges from −1.3 to −2.4 mV/K at a current level of about 100 μA (Sclar and Pollock 1972). The base–emitter p–n junction in the transistor in which the base is connected to the collector is also used because of the fact that nonlinearity in temperature dependency is lower than that of most diodes.

If the p–n junction in a diode or transistor is driven by different forward current levels *I*1 and *I*2, and voltages *V*1 and *V*2 are developed at these current levels, then, from Equation 5.11,

$$V_1 - V_2 = \left(\frac{kT}{q}\right) ln\left(\frac{I_1}{I_2}\right) \tag{5.12}$$

Thus, the difference in voltages corresponding to different current levels maintained at a constant ratio is proportional to the absolute

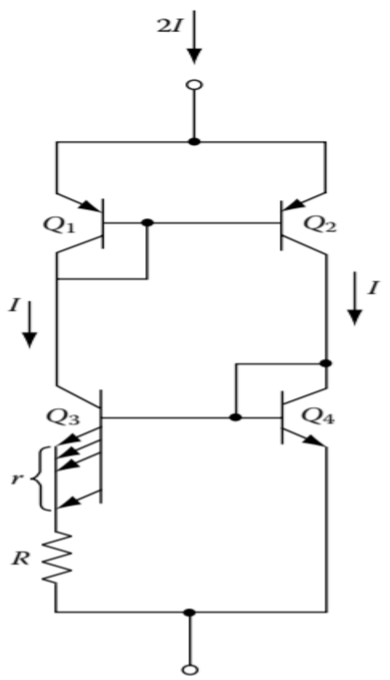



**FIGURE 5.7** Idealized scheme of a two-terminal current-output device. (After Timko, M.P., J. Solid-State Circuits, SC-11, 784, 1976.)

temperature, without any offset. By this principle, a thermometer providing output proportional to the absolute temperature can be realized either by applying a square-wave current to a p–n junction (Vester 1968), or using two matched devices operating at different current levels.

Figure 5.7 shows an idealized scheme of the device. If transistors *Q*1 and *Q*2 are assumed to be equal and have a large gain, their collector currents are equal and constrain the collector current of $Q3$ and $Q4$. $Q3$ has $r$ base–emitter junctions, and each one is identical to that of $Q4$. Then, from Equation 5.12, voltage across $R$ is obtained as:

$$RI = \left(\frac{kT}{q}\right) \ln r \tag{5.13}$$

Thus the total current $2I$ is proportional to the absolute temperature.

## ***4.3 Infrared Radiation Thermometers:***

The radiation thermometer is essentially an instrument that measures thermal radiation power emitted from the object surface. Near the human body temperature, the peak of the thermal radiation is in the far-infrared region, and thus infrared radiometers are used in medical thermometry.
When a radiometer is directed at an object surface as shown in Figure 5.14, both the emission from the object and the ambient radiation reflected at the object surface will enter into the radiometer. Thus, the total power $W$ that enters the thermometer can be expressed as the sum of the emission and reflection components as:

$$W = \varepsilon P(T_s) + (1 - \varepsilon)P(T_a) \tag{5.15}$$

where

ε is the emissivity of the object surface
$T_s$ and $T_a$ are surface and ambient radiation temperatures

$P(T)$ is the Planck's radiation formula at absolute temperature $T$, i.e.,

$$P(T) = \int \frac{C_1\lambda^{-5}}{exp(C_2/\lambda T)-1} d\lambda$$

(5.16)

where

λ is wavelength

*C*1 and *C*2 are universal constants given by:

$$C_1 = 3.74 \times 10^{-16}\, Wm^2$$

$$C_2 = 1.44 \times 10^{-2}\, m\, K$$

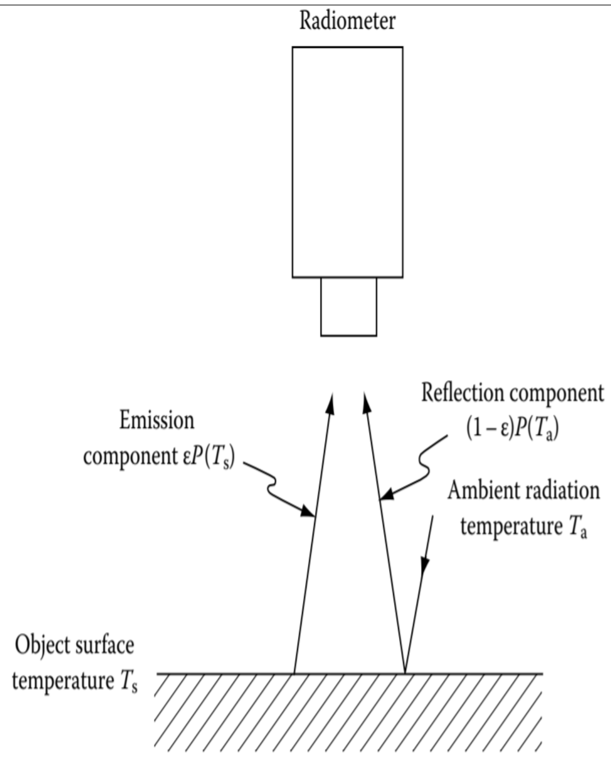



**FIGURE 5.14** Radiation power entering into a radiometer directed to an object surface. It consists of two components: the emission from the object and the reflection of the ambient radiation.

- When the radiometer has sensitivity in a range from λ1 to λ2, the integral in Planck's formula should be taken in that wavelength range. From Equation 5.15, the surface temperature $T_s$ can be obtained when the emissivity ε and the ambient radiation temperature $T_a$ are known.

- While the emissivity of the skin was considered as almost unity, which means that the skin is essentially a black body, it was shown that skin emissivity is around 0.97 in the wavelength range of 8–14 μm.
- This implies that about 3% of reflection may occur at the skin surface, and thus the ambient radiation temperature can affect temperature measurement by a radiation thermometer.
- Temperature readings may vary when the angle between the normal to the skin and the direction of observation is large, but it has been shown that the apparent temperature readings do not fall for angles of inclination less than 70°.
- A significant fall occurs only above 80° when the emissivity is 0.98, λ = 5 μm, surface temperature is 30°C, and background ambient radiation temperature is 20°C.
- When a temperature gradient across the skin exists, the temperature measured by a radiation thermometer will differ from that measured at the surface by means of a contact thermometer probe, because the radiation from the body emanates not only from the external surface but also partly from the underlying layer.
- The amount of radiation from the underlying layer may depend on the transmittance of the epidermis and dermis. Anderson and Parrish (1981) estimated the approximate depth of penetration at which radiation is attenuated to 1/e of the incident energy.
- The depth of penetration for λ = 1.2 μm was estimated as about 2.2 mm. In longer wavelength ranges, the transmittance is somewhat lower than that at 1.2 μm, but a systematic study of skin penetration depth is still lacking.
- Watmough and Oliver (1969) estimated the effective temperature difference in an extreme situation in which blood at 310 K was perfused beneath the epidermis at 305 K, and showed that a temperature reading of about 2.5°C higher than the surface temperature would be expected using a radiation thermometer with a sensitivity peak at 5 μm.

- This analysis suggests that accurate skin surface temperature measurement using a radiation thermometer requires a correction for skin emissivity and penetration depth.
- In a practical situation, such accurate measurement is rarely required and if an error of about 0.5°C is acceptable, a radiation thermometer can be used for skin temperature measurement without such corrections.

## ***4.4 Infrared Thermography:***

- The term thermography generally implies techniques of thermal imaging of the object surface. Practical techniques of thermography are infrared thermography based on radiation measurement, and contact thermography using thermochromic liquid crystals.
- Infrared thermography can provide thermal images in electric signals that are convenient for image processing and data storage, and it can also realize a noncontact, quick, and accurate measurement.
- To obtain a thermal image of an object, surface temperature measurements at many points on the object should be performed. This requirement can be achieved by infrared detectors with either mechanical scanning by means of moving mirrors or prisms, or with electronic scanning of the infrared detectors placed at the focal plane.
- In medical thermography systems, mechanical scanning was used until the 1990s; they were later replaced by electronic scanning. The detector of electronic scanning thermography system consists of a focal plane array, which is a two-dimensional infrared detector array, and a readout circuit, which is commonly a charge-coupled device.
- A typical thermal detector array consists of microbolometers arranged in two dimensions fabricated on a single silicon chip. Configuration of each microbolometer is similar as shown in Figure 5.17.

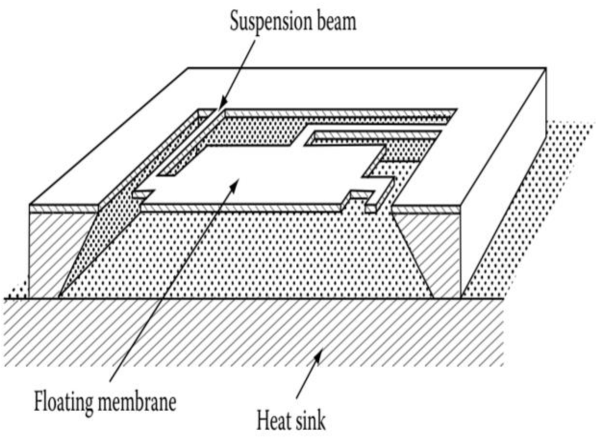



**FIGURE 5.17** A thermistor bolometer of floating plate configuration. (After van Hervaarden, A.W., *Sens. Actuators*, A21–A23, 623, 1989.)

- Sensitivity can be increased by reducing thermal conduction from the floating plate to the substrate. To minimize thermal conduction, the plate is supported by two or four long and thin bridges
- The device is kept in vacuum so as to reduce thermal conduction by surrounding gas molecules. Integrating vacuum packaging is achieved by bonding a silicon top cap wafer to the bolometer array wafer.
- In the infrared window, formed by thin silicon to which an antireflection coating is applied, losses by absorption and reflection are insignificant.
- The number of pixels in an array has been increased with the progress in fabrication technologies and the size of each pixel has been reduced.
- Noise equivalent temperature difference of almost 0.2 K was attained in ambient temperature operation. An example of medical thermography system using microbolometer array has 384 × 288 pixels and its temperature resolution is about 60 mK.
- Photon detector arrays have been used from ultraviolet to far infrared ranges. Photon detector arrays in long wavelength infrared range, typically 8–14 μm, are required for use in medical. thermography.
- In this range, most photon detector arrays have to be cooled to about 77 K to reduce thermal noise. To cool a detector array,

either liquid nitrogen or mechanical cooling engines such as the Thomson cooler or the Stirling engine are used, while the many-staged thermoelectric cooler can be used in 3–7 μm medium wavelength infrared range.

- HgCdTe (Mercury Cadmium Telluride) has been used most widely as the material of photon detector array for long wavelength infrared (8–14 μm).
- By varying fractions of Hg and Cd, so that *x* varies in $Hg1-xCdxTe$, the band gap energy can be varied, and as a result, detectors having different spectral ranges from 1 to 30 μm can be realized.
- It is difficult to fabricate a HgCdTe detector array as a monolithic structure, and thus HgCdTe detector arrays are commonly built with a hybrid structure as shown in Figure 5.19. To interconnect a HgCdTe detector array to a silicon read-out array, indium bump bonding technique is commonly used.
- Conventional thermography systems have an internal temperature reference, so that radiation from the object can be compared with that from the reference body at every line scan. To calibrate a thermography system, a set of discrete temperature references consisting of black bodies at different temperatures , or a continuous temperature calibrator consisting of a metal rod with a uniform axial temperature gradient can be used.

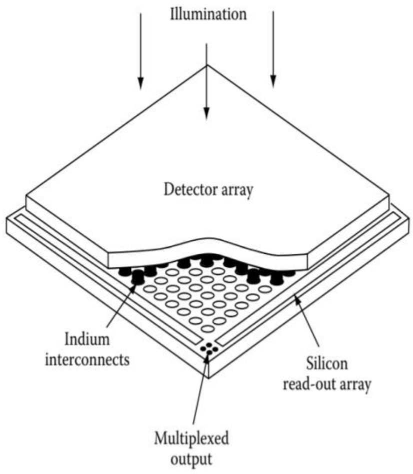



FIGURE 5.19 A hybrid focal plane array.

# ***4.5 Clinical Thermometers:***

- The mercury-in-glass thermometer was introduced in medicine in the early years of the twentieth century, and the measurement of body temperature has since become easy and accurate (Ebstein 1928). The mercury-in-glass clinical thermometer is appreciated in clinical thermometry for its remarkable reliability, as well as its convenience of handling and inexpensiveness. Even today, no other thermometer can fully replace it.
- There are, however, many situations in which the mercury-in-glass clinical thermometer is unacceptable because of its slow response and large heat capacity and size. It is also inconvenient for continuous monitoring of body temperature and data processing by computer. Mercury contamination is also a problem in hospitals (Notani-Sharma et al. 1980). Many kinds of thermometers for medical use have therefore been developed and used successfully.

## *(I) Indwelling Thermometer Probes:*

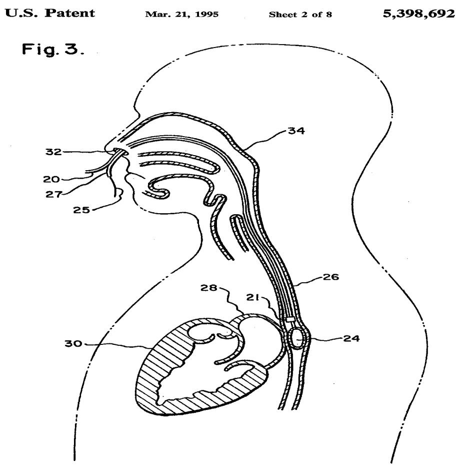
The mouth is the most convenient site for routine measurement of body core temperature. For continuous monitoring, however, placement of the probe in the mouth for a long period of time is uncomfortable, and oral temperature becomes unstable when the mouth is opened. Body core temperature is therefore usually monitored at deeper sites of the body such as the rectum, esophagus, and bladder using indwelling thermometer probes.


## (II) *Rectal Temperature Measurement:*

- The rectal probe has been commonly used for patient monitoring. The probe is simply a flexible catheter having a thermistor at the tip. To avoid the effect of ambient temperature change, the tip of the probe is placed about 8 to 15 cm from the anal sphincter. Rectal temperature has been considered as a reliable index of core temperature, if body temperature is stable.
- Rectal temperature is always 0.2°C–0.3°C higher than that obtained in any other part of the body. The possibility of the effect of bacterial heat production on rectal temperature was pointed out . However, this effect is a matter of doubt as in experimental observations in which temperature change at bowel sterilization due to orally administered antibiotics was insignificant.
- A study in which fresh human stools were placed at 37°C for 6 h showed that the temperature of the stool core increased 0.3°C ± 0.39°C. This result still suggests the possibility of bacterial heat production.

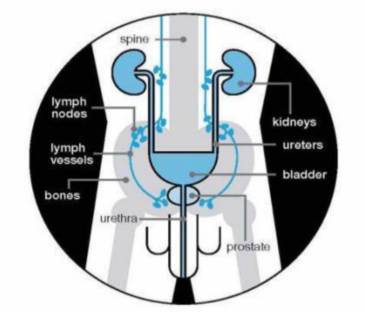
## (III) *Esophageal Temperature Measurement:*

- Esophageal temperature is measured by inserting flexible probes through the mouth or nose, mostly for body temperature monitoring during anesthesia. In the upper part of the esophagus, a significant influence of tracheal air temperature has been observed .It is therefore recommended that measurement of esophageal temperature should be taken 24 to 28 cm below the corniculate cartilages or at heart level.
- At this level, esophageal temperature is intermediate between oral and rectal temperatures, and rapidly follows internal temperature changes. Esophageal measurement can be tolerated postoperatively if the probe is introduced through the nose as a gastric tube.



## (IV) Bladder Temperature Measurement:

- Bladder temperature is monitored commonly by a thermistor-tipped Foley's bladder catheter, i.e., a 16 Fr triple-lumen all-silicone catheter with an attached thermistor.. It has been shown that bladder temperatures are highly correlated with rectal, esophageal, and pulmonary arterial temperatures, and closely follow arterial blood temperature even during rapid cooling and rewarming using extracorporeal circulation Bladder temperature monitoring is recommended particularly for patients in whom Foley's catheterization is indicated.




# 4.6 Telemetering Capsules:

- The telemetering technique is effective when a sensor can be placed near the object, while direct connection by a cable is unfavorable. For body core temperature measurement in the digestive tract, a radio telemetering capsule in which a temperature sensor and radio transmitter are encapsulated in a swallowable capsule has been used.
- Passive telemetering technique without using a battery operated transmitter is also possible and has the advantage that the operating period is not restricted by battery life. Ultrasonic detection of the properties of a crystal resonator is an example of this type.

## (I) Radio Pill:

- Radio pill is a swallowable radio telemetering capsule, which can be used for measurements in the digestive tract. The temperature-sensitive radio pill has been used for body core temperature monitoring. An example of a radio pill is 2.2 cm long, 0.9 cm in diameter, and is operated at about 350 kHz with a temperature coefficient of about 10 kHz/K, and its response time is about 4 min for a 97% response in unstirred water). A passive telemetry system in which the transmitter circuit is powered by an externally applied radio frequency excitation is also possible.

- A radio pill in which a crystal resonator is employed as the temperature sensor has also been developed. Temperature is sensed by a quartz tuning fork having a sensitivity of about 9 Hz/K operated at 262 kHz. The circuit is powered by a 1.2 V rechargeable battery, and the current drain is about 100 μA. The battery can be charged by electromagnetic induction.
- The power consumption can be reduced by command mode operation by which the oscillator is activated for a 10 s readout period when a command signal is applied. A disposable type of this pill is produced commercially. It has a silver oxide battery and operates 200 h.

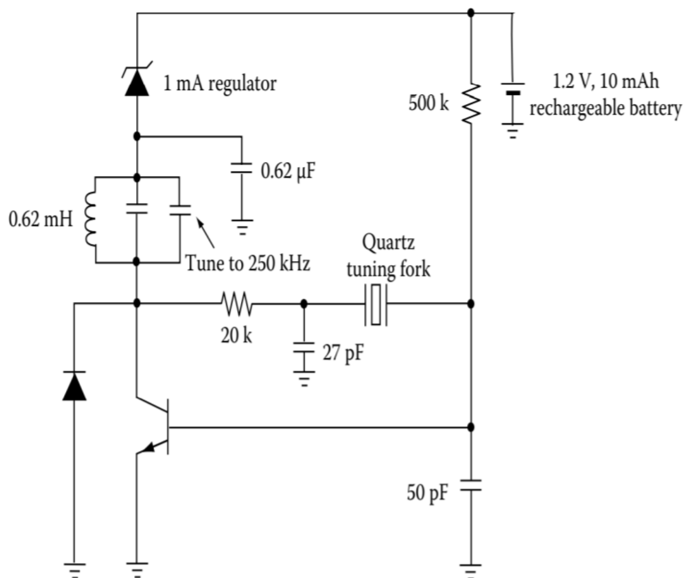



**FIGURE 5.26** The circuit diagram of a radio pill, in which temperature is sensed by a quartz tuning fork. (After Cutchis, P.N. et al., *Johns Hopkins APL Tech. Dig.*, 9, 16, 1988.)

## *(II) <u>Ultrasonic-Coupling Crystal Resonator:</u>*

- The quartz crystal resonator can be excited by applying an ultrasonic wave near the resonance frequency; ultrasonic detection of the resonance signal is also possible as long as the resonator is placed in a medium through which sound propagates.
- In a study, a crystal resonator consisting of a quartz tuning fork encapsulated in small metal cylindrical capsule, 2 mm in diameter and 7 mm in length, as shown in Figure 5.27 was used.
- Its resonant frequency is about 40 kHz with a temperature coefficient of about 3.2 Hz/K. To measure temperature, the resonator is excited for about 0.4 s by applying an ultrasonic wave near the resonant frequency.
- The damped oscillations in the resonant frequency are measured. The absolute accuracy is about 0.1°C, with temperature resolution about 0.01°C. It has been shown that when the capsule is swallowed, temperature measurements can be performed from the abdominal skin surface.

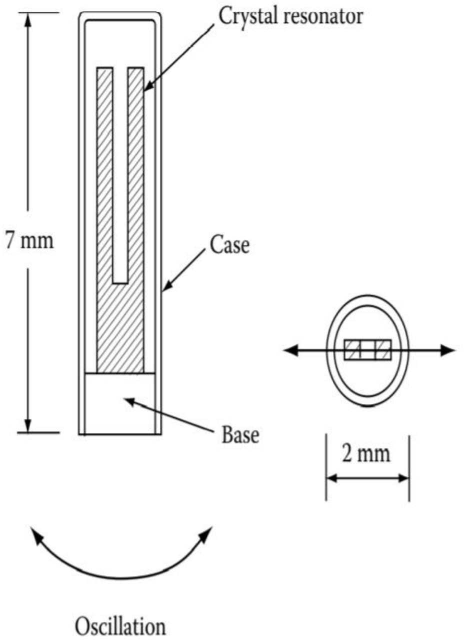



**FIGURE 5.27** An implantable ultrasonic-coupling crystal resonator for local temperature measurement in the body.

## ***4.7 Tympanic Thermometers:***

Tympanic temperature has been considered as a reliable index of core temperature, and it has therefore been used in physiological studies of thermoregulation. It is also used in patient monitoring during anesthesia and in intensive care units. Noncontact tympanic thermometers have an advantage in that body temperature measurement can be performed in a few seconds.

## *(I) Contact Probes:*

- Small thermocouple or thermistor probes have been used for tympanic temperature measurements. Benzinger and Taylor (1963) described tympanic probes in which 36-gauge copper–constantan wires were soldered side by side at the tip and drawn into a fine polyethylene tubing.
- A brush-type probe, as shown in Figure 5.22, was recommended with the free ends of the bristles facing the interior of the meatus. They stated that there was practically no discomfort in wearing it, and minor alterations in hearing were advantageous for ascertaining correct positioning.
- The comparative evaluation of temperatures recorded using tympanic and two nasopharyngeal probes at a region that receives its blood supply from an artery supplying the brain showed that differences between these temperatures were 0.05°C or less in environmental temperatures of 24°C and 45°C.
- Tympanic temperature was also compared with esophageal temperature in clinical situations, and it was recognized that both were closely parallel.
- Although safety and minimum embarrassment have been stressed by many authors, there have been some case reports of tympanic membrane perforation complicating tympanic thermometry using contact probes.

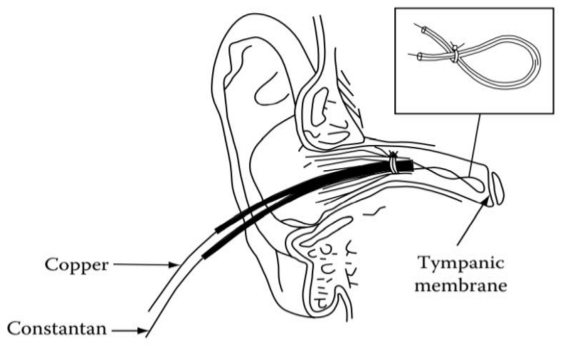



**FIGURE 5.22** Tympanic temperature measurement using a brush-type thermocouple probe. (After Benzinger, T.H. and Taylor, G.W., Cranial measurements of internal temperature in man, in *Temperature, Its Measurement and Control in Science and Industry*, Vol. 3, Hardy, J. D., ed., Reinhold, New York, pp. 111–120, 1963.)

## (II) *Noncontact Tympanic Thermometers:*

- The noncontact tympanic thermometer is essentially an infrared radiation thermometer having a probe that can be directed to the tympanic membrane through the external auditory canal. Noncontact measurement of tympanic temperature was first attempted by a small thermistor with active heating at the proximal end of the probe.
- Since then, instruments employing fast response infrared detectors have been developed, and many different kinds of infrared tympanic thermometers are now supplied commercially.
- A commercial tympanic thermometer (First-Temp®, Intelligent Medical Systems, Carlsbad, CA) employs a thermopile detector with a light pipe installed at the tip of the probe, and the tip is inserted into the external auditory canal of the patient as shown in Figure 5.23.
- When the probe is correctly applied to the ear, tympanic temperature can be measured within 2 s. In the early model, the probe had to be recalibrated after each measurement by placing it on the instrument body on which a reference radiation source is installed. In present-day models, recalibration is unnecessary.
- A clinical observation showed that tympanic temperatures measured by an infrared tympanic thermometer were close to pulmonary arterial temperatures over the temperature range of 34.0°C–39.5°C with a correlation coefficient of 0.98, even including the situation of rapid increase in temperature after open heart surgery.

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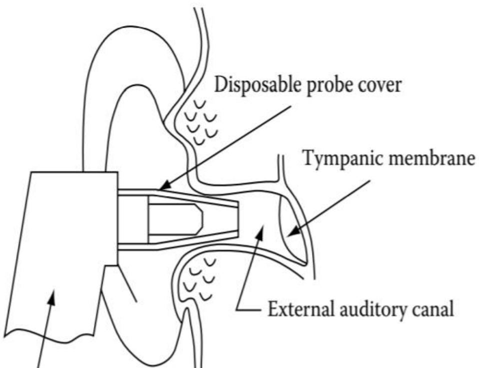


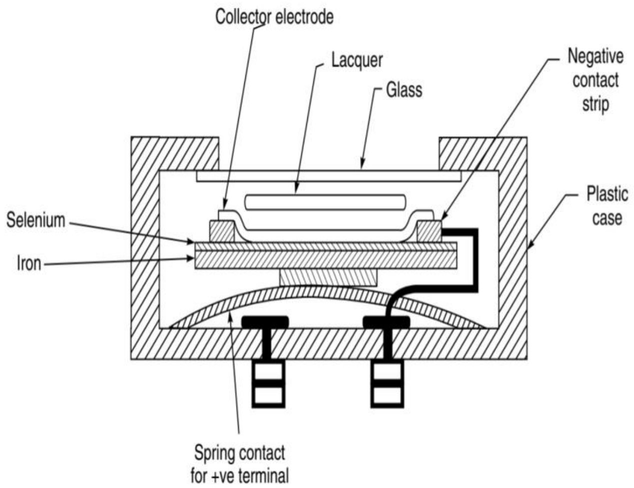
## 4.8 Photoelectric Transducers: Photovoltaic Cells and Photoemissive Cells.

Photoelectric transducers are based on the principle of conversion of light energy into electrical energy. This is done by causing the radiation to fall on a photosensitive element and measuring the electrical current so generated with a sensitive galvanometer directly or after suitable amplification. There are two types of photoelectric cells—photovoltaic cells and photomissive cells.

### *(I) <u>Photovoltaic cells:</u>*

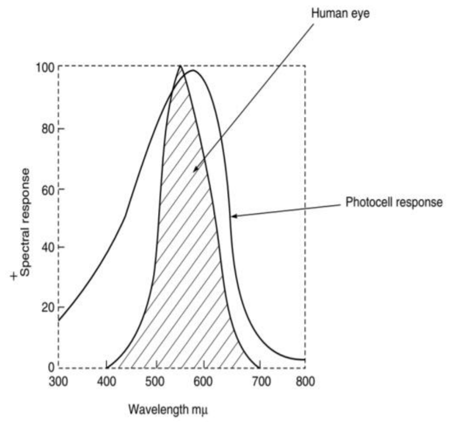
- Photovoltaic or barrier layer cells usually consist of a semiconducting substance, which is generally selenium deposited on a metal base which may be iron and which acts as one of the electrodes.
- The semiconducting substance is covered with a thin layer of silver or gold deposited by cathodic deposition in vacuum. This layer acts as a collecting electrode. Figure 3.16 shows the construction of the barrier layer cell.

- When radiant energy falls upon the semiconductor surface, it excites the electrons at the silver-selenium interface. The electrons are thus released and collected at the collector electrode.
- The cell is enclosed in a housing of insulating material and covered with a sheet of glass. The two electrodes are connected to two terminals which connect the cell with other parts of the electrical circuit.
- Photovoltaic cells are very robust in construction, need no external electrical supply and produce a photocurrent sometimes stronger than other photosensitive elements. Typical photocurrents produced by these cells are as high as 120 mA/lumen. At constant temperature, the current set up in the cell usually shows a linear relationship with the incident light intensity.
- Selenium photocells have very low internal resistance, and therefore, it is difficult to amplify the current they produce by dc amplifiers. The currents are usually measured directly by connecting the terminals of the cell to a very sensitive galvanometer.
- Selenium cells are sensitive to almost the entire range of wavelengths of the spectrum. However, their sensitivity is greater within the visible spectrum and highest in the zones near to the yellow wavelengths. Figure 3.17 shows spectral response of the selenium photocell and the human eye.
- Selenium cells have a high temperature coefficient and therefore, it is very necessary to allow the instrument to warm up before the readings are commenced. They also show fatigue effects. When illuminated, the photocurrent rises to a value several percent above the equilibrium value and then falls off gradually.
- When connected in the optical path of the light rays, care should be taken to block all external light and to see that only the light from the source reaches the cell.




➤ Fig. 3.16 *Construction of a barrier layer cell*

- Selenium cells are not suitable for operations in instruments where the levels of illumination change rapidly, because they fail to respond immediately to those changes. They are thus not suitable where mechanical choppers are used to interrupt light 15–60 times a second.




➤ Fig. 3.17 *Spectral response of a human eye and a selenium photocell*

## *(II) Photoemissive cells:*

Photoemissive cells are of three types:
(a) High vacuum photocells,
(b) Gas-filled photocells and

(c) Photomultiplier tubes.

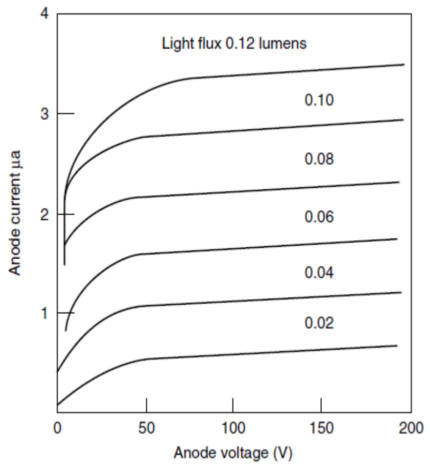
All of these types differ from selenium cells in that they require an external power supply to provide a sufficient potential difference between the electrodes to facilitate the flow of electrons generated at the photosensitive cathode surface. Also, amplifier circuits are invariably employed for the amplification of this current.

***High Vaccum Photoemissive Cells:***

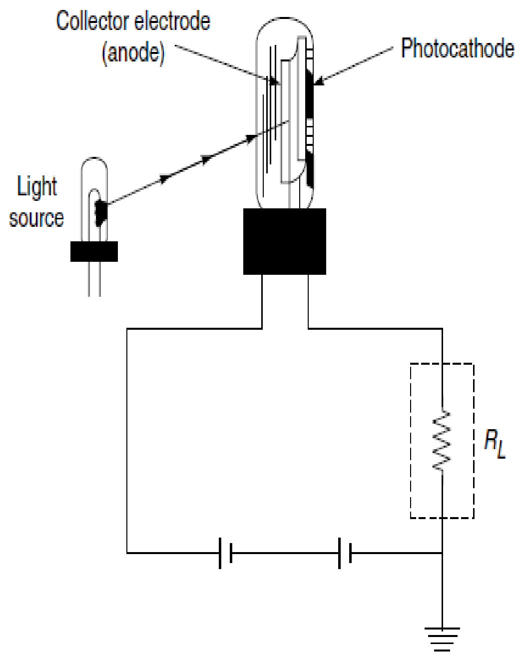
The vaccum photocell consists of two electrodes—the cathode having a photosensitive layer of metallic cesium deposited on a base of silver oxide and the anode which is either an axially centered wire or a rectangular wire that frames the cathode.

The construction of the anode is such that no shadow falls on the cathode. The two electrodes are sealed within an evacuated glass envelope.

Figure 3.18 shows the current-voltage characteristics of a vacuum photoemissive tube at different levels of light flux. They show that as the voltage increases, a point is reached where all the photoelectrons are swept to the anode as soon as they are released which results in a saturation photocurrent. It is not desirable to apply very high voltages, as they would result in a excessive dark current without any gain in response.




➤ Fig. 3.18 *Current voltage characteristics of a vacuum photoemissive tube at different levels of light flux*




➤ Fig. 3.19 *Typical circuit configuration employed with photoemissive tubes*

Figure 3.19 shows a typical circuit configuration usually employed with photoemissive tubes. Large values of phototube load resistor are employed to increase the sensitivity up to the practical limit. Load resistances as high as 10,000 MW have been used. This, however, almost puts a limit, as further increase of sensitivity induces difficulties in the form of noise, non-linearity and slow response. At these high values of load resistors, it is very essential to shield the circuit from moisture and electrostatic effects. Therefore, special type of electrometer tubes, carefully shielded and with a grid cap input are employed in the first stage of the amplifier.

***Gas-filled Photoemissive Cells:***

This type of cell contains small quantities of an inert gas like argon, whose molecules can be ionized when the electrons present in the cell posses sufficient energy.

The presence of small quantities of this gas prevent the phenomenon of saturation current, when higher potential differences are applied between the cathode and anode.

Due to repeated collisions of electrons in the gas-filled tubes, the photoelectric current produced is greater even at low potentials.

***Photomultiplier Tubes:***

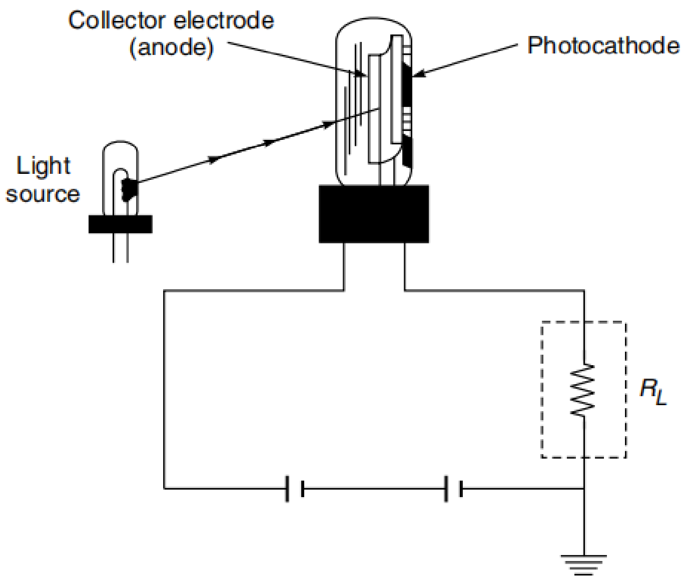
Photomultiplier tubes are used as detectors when it is required to detect very weak light intensities.

The tube consists of a photosensitive cathode and has multiple cascade stages of electron amplification in order to achieve a large amplification of the primary photocurrent within the envelope of the phototube itself.

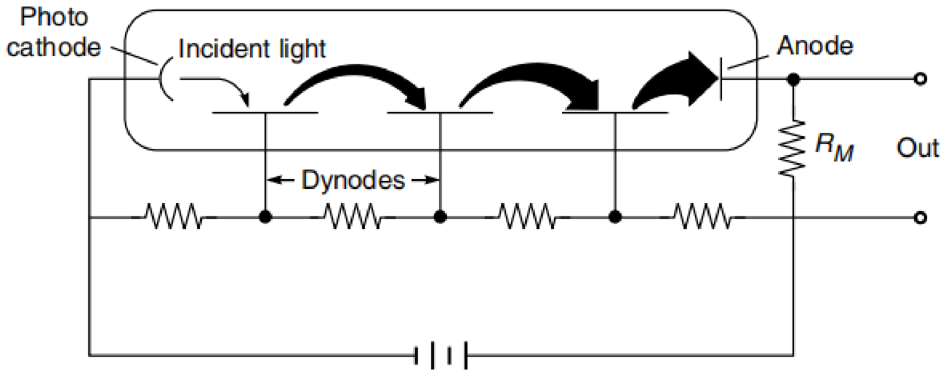
The electrons generated at the photocathode are attracted by the first electrode, called dynode, which gives out secondary electrons. There may be 9–16 dynodes (Fig. 3.20).

The dynode consists of a plate of material coated with a substance having a small force of attraction for the escaping electrons. Each impinging electron dislodges secondary electrons from the dynode.

Under the influence of positive potential, these electrons are accelerated to the second dynode and so on. This process is repeated at the successive dynode, which are operated at voltages increasing in steps of 50–100 V. These electrons are finally collected at the collector electrode.




➤ Fig.3.19 *Typical circuit configuration employed with photoemissive tubes*




➤ Fig.3.20 *Principle of photomultiplier tube*

They can measure light intensities about $10^7$times weaker than those measurable with an ordinary phototube. For this reason, they should be carefully shielded from stray light.

## 4.9 Biosensors

Biosensors combine the exquisite selectivity of biology with the processing power of modern microelectronics and optoelectronics to offer powerful new analytical tools with major applications in medicine, environmental studies, food and processing industries.

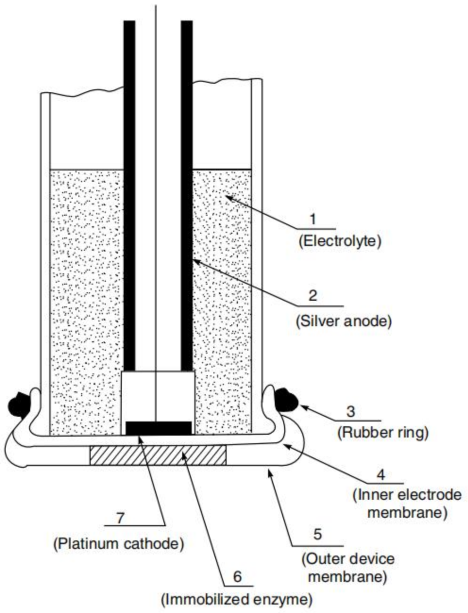
All chemical sensors consist of a sensing layer that interacts with particular chemical substances, and a transducer element that monitors these interactions.
Biosensors are chemical sensors in which the sensing layer is composed of biological macro-molecules, such as antibodies or enzymes.
Today, the term biosensor is used to describe a wide variety of analytical devices based on the union between biological and physico-chemical components.

The biological component can consist of enzymes, antibodies, whole cells or tissue slices and is used to recognize and interact with a specific analyte.

The physico-chemical component, often referred to as the transducer, converts this interaction into a signal, which can be amplified and which has a direct relationship with the concentration of the analyte.

The most successful biosensor to-date is the home blood glucose monitor for use by people suffering from diabetes. The biosensor in this instrument relies upon enzymes that recognise and catalyze reactions of glucose with the generation of redox-active species that are detected electrochemically.

Figure 3.26 shows the construction of this type of sensor. If the immobilized enzyme is soluble glucose oxidase between the two membranes, it becomes a glucose sensor. It works on the principle that in the presence of glucose, oxygen is consumed, providing a change in the signal from a conventional oxygen electrode.




➤ Fig. 3.26 *Constructional details of an enzyme utilizing sensor with oxygen electrode as the underlying analytical tool*

The chemical reaction of glucose with oxygen is catalyzed in the presence of glucose oxidase.
This causes a decrease in the partial pressure of oxygen (pO2), and the production of hydrogen peroxide by the oxidation of glucose to gluconic acid as per the following reaction:

$$\text{Glucose} + O_2 \xrightarrow{\text{Glucose oxidase}} \text{Gluconic acid} + H_2O_2$$

The changes in all of these chemical components can be measured in order to determine the concentration of glucose.

## Smart Sensors

Although an accepted industry definition for a smart sensor does not currently exist, it is generally agreed that they have tight coupling between sensing and computing elements.

Their characteristics, therefore, include: temperature compensation, calibration, amplification, some level of decision-making capability, self-diagnostic and testing capability and the ability to communicate interactively with external digital circuitry.

Currently available smart sensors are actually hybrid assemblies of semiconductor sensors plus other semiconductor devices. In some cases, the coupling between the sensor and computing element is at the chip level on a single piece of silicon in what is referred to as an integrated smart sensor.

The important role of smart sensors are:
Signal Conditioning: The smart sensor serves to convert from a time-dependent analog variable to a digital output.

Functions such as linearization, temperature compensation and signal processing are included in the package.

Tightening Feedback Loops: Communication delays can cause problems for systems that rely on feedback or that must react/adapt to their environment. By reducing the distance between sensor and processor, smart sensors bring about significant advantages to these types of applications.

Monitoring/Diagnosis: Smart sensors that incorporate pattern recognition and statistical techniques can be used to provide data reduction, change detection and compliation of information for monitoring and diagnostic purposes, specially in the health sector.

Smart sensors divert much of the signal processing workload away from the general purpose computers. They offer a reduction in overall package size and improved reliability, both of which are critical for in situ and sample return applications. Achieving a smart sensor depends on integrating the technical resources necessary to design the sensor and the circuitry, developing a manufacturable process and choosing the right technology.110 Handbook of Biomedical Instrumentation
A typical example of a smart sensor is a pressure sensor (MPX5050D) with integrated amplification, calibration and temperature compensation introduced by Motorola (Frank, 1993).

The sensor typically uses pieziorestive effect in silicon and employs bipolar integrated circuit processing techniques to manufacture the sensor.

By laser trimming thin film resistors on the pressure sensor, the device achieves a zero-pressure offset-voltage of 0.5 V nominal and full-scale output voltage of 4.5 V, when connected to a 5.0 V supply.

Therefore, the output dynamic range due to an input pressure swing of 0–375 mmHg is 4.0 V. The performance of the device compares favourably to products that are manufactured with
direct components.