Lecture Four

Basic Sensors and Measurements.

Introduction

The basic mechanisms and principles of the sensors used in several medical instruments will be discussed in this lecture. A transducer is a device that converts energy from one form to another. A sensor converts a physical parameter to an electric output. An actuator converts an electric signal to a physical output. An electric output from the sensor is normally desirable because of its advantages in further signal processing.

There are many methods used to convert physiological events to electric signals. Variations in resistance, inductance, capacitance, and piezoelectric effect may measure dimensional changes. Thermistors and thermocouples are employed to measure body temperatures. Electro-magnetic-radiation sensors include thermal and photon detectors.

DISPLACEMENT MEASUREMENTS

In biomedical engineering fields, it is interesting to measure the size, shape, and position of the organs and tissues of the body. Variations in these parameters are essential in discriminating normal from abnormal function. Displacement sensors can be used in both direct and indirect systems of measurement.

Direct displacement measurements are used to determine the change in the diameter of blood vessels and the changes in the volume and shape of cardiac chambers.

Indirect displacement measurements are used to quantify the movements of liquids through heart valves. An example is the movement of a microphone diaphragm that indirectly detects the heart's movement and the resulting heart murmurs.

RESISTIVE SENSORS: POTENTIOMETERS

Figure 1 shows three types of potentiometric devices for measuring displacement. The potentiometer shown in Figure 1(a) measures translational displacements from 2 to 500 mm. Rotational displacements ranging from 10° to more than 50° are detected, as shown in Figures 1(b) and (c).

The resistance elements (wire-wound, carbon-film, metal-film, conducting plastic, or ceramic material) may be excited by either dc or ac voltages. These potentiometers produce a linear output (within 0.01% of full scale) as a function of displacement, provided that the potentiometer is not electrically loaded. The resolution of these potentiometers is a function of the construction. It is possible to achieve a continuous stepless conversion of resistance for low-resistance values up to 10Ω by utilizing a straight piece of wire.

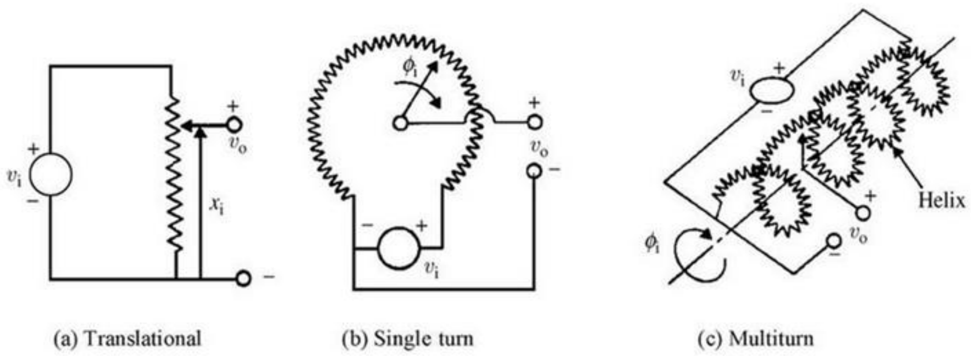
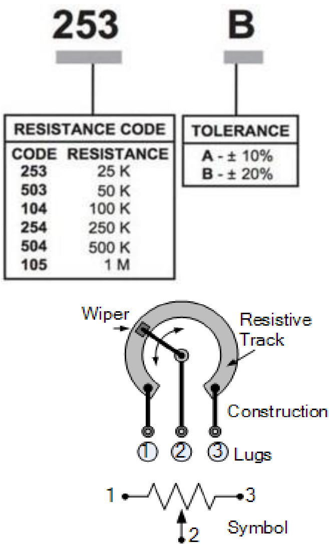


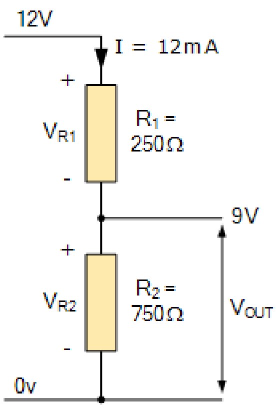
Figure 1 Three types of potentiometric devices for measuring displacement

The resistance wire is wound on a mandrel or card for greater variations in resistance, from several ohms to several megohms. The resistance variation is not continuous but stepwise because the wiper moves from one turn of wire to the next. The fundamental limitation of the resolution is a function of the wire spacing, which may be as small as 20 µm. The frictional and inertial components of these potentiometers should be low to minimize dynamic distortion of the system.



Potentiometer Example

A resistor of 250ohms is connected in series with the second resistor of 750ohms so that the 250ohm resistor is connected to a supply of 12 volts, and the 750ohm resistor is connected to theground (0v). Calculate the total series resistance, the current flowing through the series circuit, and the voltage drop across the 750ohm resistor.



$$R_T = R_1 + R_2$$

$$R_T = 250\Omega + 750\Omega = 1000\Omega \text{ or } \underline{1k\Omega}$$

$$I = \frac{V}{R_T} = \frac{12}{1000} = \underline{12mA}$$

$$V_{OUT} = 12\left(\frac{750}{250 + 750}\right) = \underline{9V}$$

To check:

$$V_{R2} = I \times R_2 = 0.012 \times 750 = \underline{9V}$$

STRAIN GAGES

When a fine wire (25 μm) is strained within its elastic limit, its resistance changes because of the diameter, length, and resistivity changes. The resulting strain gages may be used to measure extremely small displacements on the order of nanometers. The following derivation shows how each of these parameters influences the resistance change. The basic equation for the resistance R of a wire with resistivity ρ (ohmmeter), length L (meters), and cross-sectional area A (meter squared) is given by:

$$R = \frac{\rho L}{A}$$

The differential change in R is found by taking the differential.

$$dR = \frac{\rho dL}{A} - \rho A^{\ 2} L\, dA + L\frac{d\rho}{A}$$

The expression will be modified to represent finite changes in the parameters and is also a function of standard mechanical coefficients. Thus, by dividing the elements in the last expression by resistance equation members and introducing incremental values, we get

$$\frac{\Delta R}{R} = \frac{\Delta L}{L} \quad \frac{\Delta A}{A} + \frac{\Delta \rho}{\rho}$$

And by submitting Poisson's ratio μ relates the change in diameter ΔD to the change in length: ΔD/D= - μ ΔL/L yields

$$\frac{\Delta R}{R} = \underbrace{(1 + 2\mu)\frac{\Delta L}{L}}_{\text{Dimensional effect}} + \underbrace{\frac{\Delta \rho}{\rho}}_{\text{Piezoresistive effect}}$$

Note that the change in resistance is a function of changes in dimension-length (ΔL/L) and area (2μ ΔL/L), plus the piezoresistive effect, which is the change in resistivity due to strain-induced changes in the lattice structure of the material, Δρ/ρ.

The gage factor G, found by dividing the last equation by ΔL/L, is useful in comparing various strain-gage materials.

$$G = \frac{\Delta R/R}{\Delta L/L} = (1 + 2\mu) + \frac{\Delta\rho/\rho}{\Delta L/L}$$

Table 1 gives the gage factors and temperature coefficient of resistivity of various strain-gage materials. Note that the gage factor for semiconductor materials is approximately 50 to 70 times that of metals.

Table 1 Properties of Strain-Gage Materials

Material	Composition (%)	Gage Factor	Temperature Coefficient of Resistivity (°C^{-1}–10^{-5})
Constantan (advance)	Ni_{45}, Cu_{55}	2.1	±2
Isoelastic	Ni_{36}, Cr_8 $(Mn, Si, Mo)_4$ Fe_{52}	3.52 to 3.6	+17
Karma	Ni_{74}, Cr_{20}, Fe_3 Cu_3	2.1	+2
Manganin	Cu_{84}, Mn_{12}, Ni_4	0.3 to 0.47	±2
Alloy 479	Pt_{92},W_8	3.6 to 4.4	+24
Nickel	Pure	−12 to −20	670
Nichrome V	Ni_{80}, Cr_{20}	2.1 to 2.63	10
Silicon	(*p* type)	100 to 170	70 to 700
Silicon	(*n* type)	−100 to −140	70 to 700
Germanium	(*p* type)	102	
Germanium	(*n* type)	−150	

Strain gauges can be classified as either unbonded or bonded. An unbonded strain-gage unit is shown in Figure 2(a). The four sets of strain-sensitive wires are connected to form a Wheatstone bridge, as shown in Figure 2(b). These wires are mounted under stress between the frame and the movable armature, so preload is more significant than any expected external compressive load. This is necessary to avoid putting the wires in compression. This type of sensor may convert blood pressure to diaphragm movement, resistance change, and an electric signal.

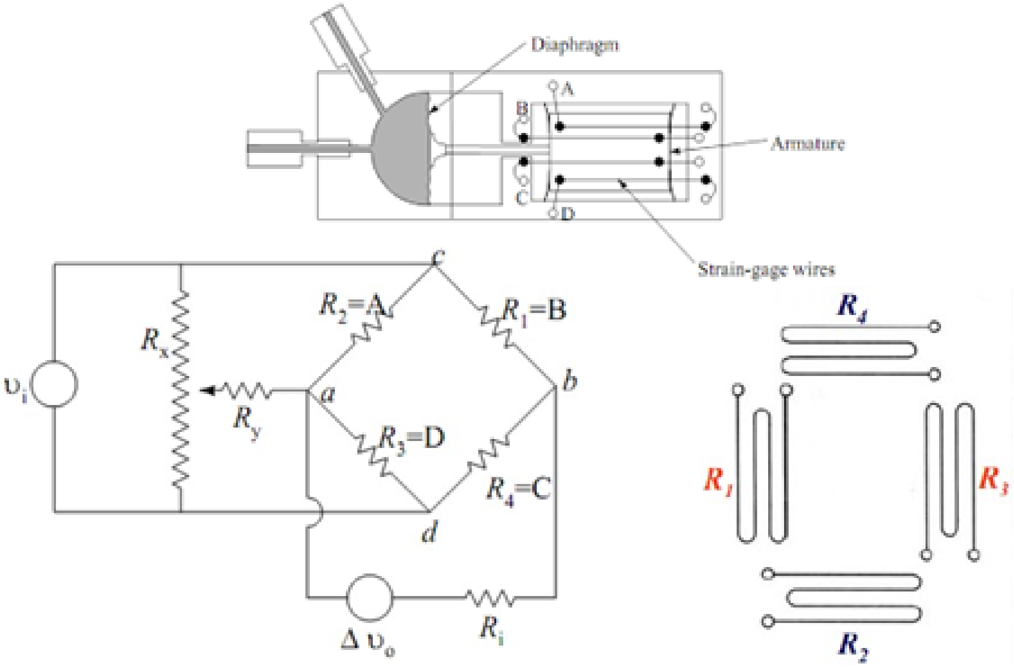


Figure 2 (a) Unbonded strain-gage pressure sensor. The diaphragm is directly coupled by an armature to an unbonded strain-gage system. With increasing pressure, the strain on gage pairs B and C is increased, while that on gage pairs A and D is decreased. (b) Wheatstone bridge with four active elements: Ri = B, R2 = A, R3 = D, and R4 = C when the unbonded strain gauge is connected for translational motion. Resistor Ry and potentiometer Rx are used to initially balance the bridge, Vi is the applied voltage, and Δvo is the output voltage on a voltmeter or similar device with an internal resistance of R1.

According to figure 2, the sensitivity of the Wheatstone bridge can be calculated as follows:

Let R1=R2=R3=R4=R0 at equilibrium (no displacement). After displacement:

R1 =Ro + ΔR , R3 =Ro + ΔR

R2 =Ro - ΔR , R4 =Ro - ΔR, substitute in:

$$v_o = v_i \left(\frac{R_1}{R_1 + R_2} - \frac{R_4}{R_4 + R_3} \right)$$

Then vo = vi (ΔR/R0)

Practically, vo is measured with a voltmeter of an internal resistance Rm. If Rm >> Ro, then the meter resistance has little effect. Otherwise, it produces a nonlinearity.

$$v_o = \frac{2v_i R_m \Delta R}{2R_o R_m + R_o^2 - \Delta R^2} = \frac{2v_i \Delta R / R_o}{2 + \frac{R_o}{R_m}\left(1 - \left(\frac{\Delta R}{R_o}\right)^2\right)}$$

A ***bonded strain-gage*** element, consisting of a metallic wire, etched foil, vacuum-deposited film, or semiconductor bar, is cemented to the strained surface. Figure 3 shows typically bonded strain gages. The deviation from linearity is approximately 1%. One method of temperature compensation for the natural temperature sensitivity of bonded strain gages involves using a second strain gage as a dummy element that is exposed to the temperature variation but not to strain. When possible, the four-arm bridge shown in Figure 2 should be used because it provides temperature compensation and yields four times greater output if all four arms contain active gages. In dental research, four bonded metal strain gages can be used on cantilever beams to measure bite force.

Elastic-resistance strain gages are extensively used in biomedical applications, especially in cardiovascular and respiratory dimensional and plethysmographic (volume-measuring) determinations. These systems usually consist of a narrow silicone-rubber tube [0.5 mm inner diameter (ID), 2 mm outer diameter (OD)] from 3 to 25 cm long and filled with mercury or with an electrolyte or conductive paste. The tube ends are sealed with electrodes (amalgamated copper, silver, or platinum). As the tube stretches, the diameter decreases, and the length increases, causing the resistance to increase. The resistance per unit length of typical gages is approximately 0.02 to 2 Ω/cm.

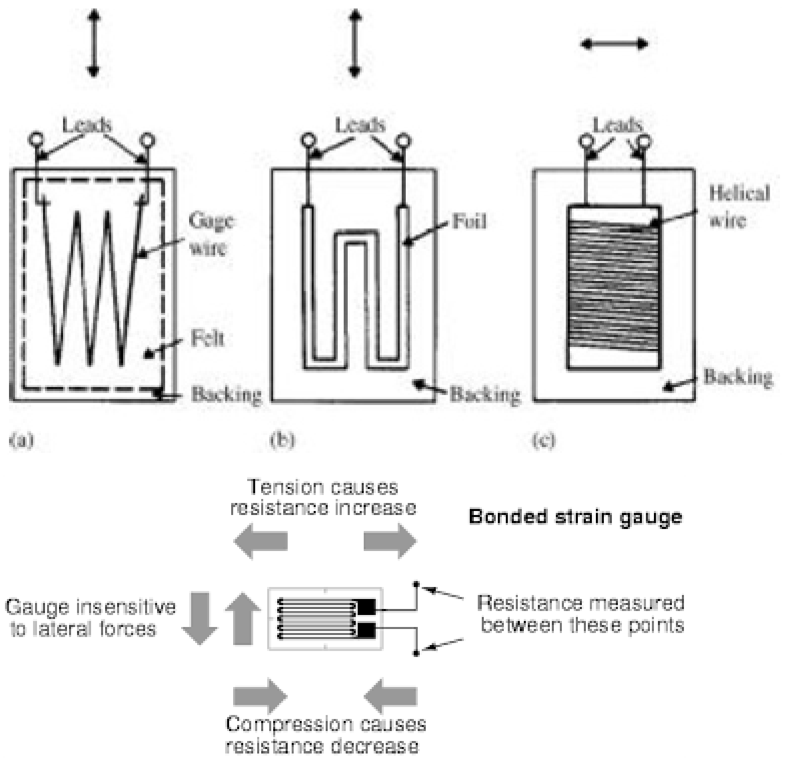


Figure 3 Typical bonded strain-gage units (a) resistance-wire type, (b) foil type, (c) helical-wire type. Arrows above units show direction of maximal sensitivity to strain. [Parts (a) and (b) are modified from Instrumentation in Scientific Research,

These units measure much higher displacements than other gages. The elastic strain gauge is linear within 1% for 10% maximal extension. As the extension is increased to 30% of the maximum, the nonlinearity reaches 4% of the full scale. The initial nonlinearity (dead band) is ascribed to the slackness of the unit. Long-term creep is a property of rubber tubing. This is not a problem for dynamic measurements.

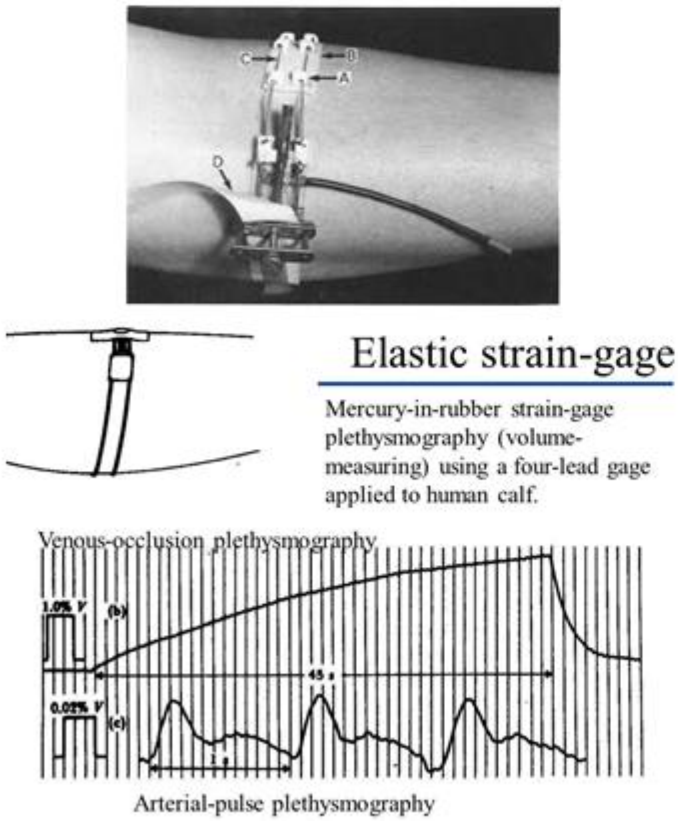


Figure 2.6 Mercury-in-rubber strain-gage plethysmography (a) Four-lead gage applied to human calf, (b) Bridge output for venous-occlusion plethysmography. (c) Bridge output for arterial-pulse plethysmography.

Temperature Measurements

Introduction

A patient's body temperature gives the physician important information about the individual's physiological state. *External body temperature* is one of many parameters used to evaluate patients in shock because the reduced blood pressure of a person in circulatory shock results in low blood flow to the periphery. A drop in *the big-toe temperature* is an excellent early clinical warning of shock.

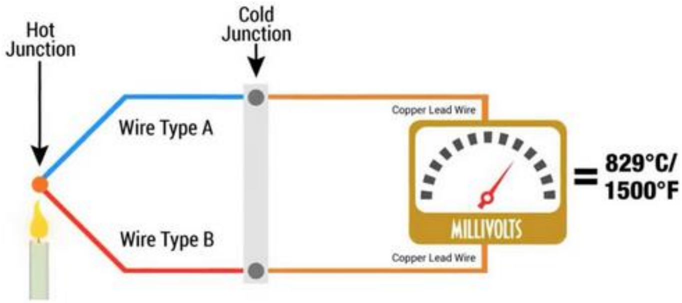
On the other hand, infections are usually reflected by increased body temperature, with hot, flushed skin and loss of fluids. *Anesthesia* decreases body temperature by depressing the thermal regulatory center. Physicians routinely induce hypothermia in surgical cases in which they wish to reduce a patient's metabolic processes and blood circulation.

In pediatrics, *special heated incubators* are used for stabilizing the body temperature of infants. Accurate monitoring of temperature and regulatory control systems are used to maintain a desirable ambient temperature for the infant. In the study of *arthritis*, physicians have shown that the temperatures of joints are closely correlated with the amount of local inflammation. Thermal measurements can detect increased blood flow due to arthritis and chronic inflammation. The specific site of body-temperature recording must be selected carefully to reflect the patient's temperature. The following types of thermally sensitive methods of measurement will be discussed:

Thermocouples	Thermistors
Radiation and fiber-optic detectors	*p-n* junction semiconductor 2 mV/ºC

THERMOCOUPLES

Thermoelectric thermometry is based on the observation that an electromotive force (emf) exists across a junction of two dissimilar metals.



This phenomenon is due to the sum of two independent effects:

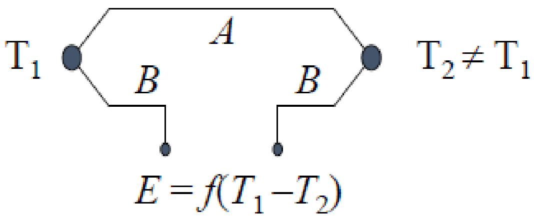
- ***<u>Peltier emf:</u>*** is an emf due solely to the contact of two unlike metals and the junction temperature. The net Peltier emf is roughly proportional to the difference between the temperatures of the two junctions.
- ***<u>Thomson emf:</u>*** is an emf due to the temperature gradients along each single conductor. The net Thomson emf is proportional to the difference between the squares of the absolute junction temperatures (T1 and T2).

The magnitudes of the Peltier and Thomson emfs depend on the metals chosen. Empirical calibration data are usually curve fitted with a power series expansion that yields the Seebeck voltage (appears due to current flows in the circuit, which is caused by the difference in temperature between the two junctions),

$$E = aT + \frac{1}{2}bT^2 + \cdots$$

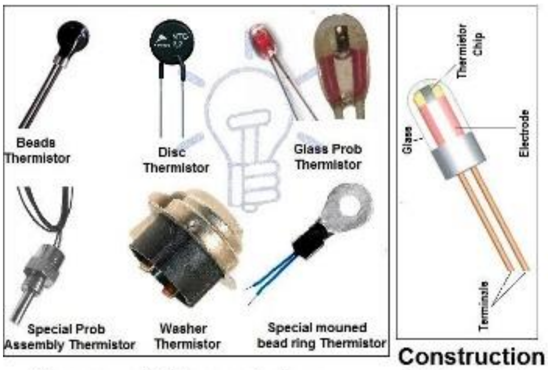
Where: T is in degrees Celsius, and the reference junction is maintained at 0°C.

Figure 1 is a thermocouple circuit with two dissimilar metals, A and B, at two different temperatures, T1 and T2, and *f* is the relative Seebeck coefficient of the thermocouple (V/K). *E* is thermo emf.



Thermistors

Thermistors are semiconductors made of ceramic materials that are thermal resistors with a high negative temperature coefficient (NTC), where their resistance decreases as their temperature increases. These materials react to temperature changes in a way that is opposite to the way metals respond to such changes.



Types of Thermistor   Construction of Thermistor

The resistivity of thermistor semiconductors used for biomedical applications is between 0.1 and 100Ω.m. These devices are small in size (they can be made less than 0.5 mm in diameter), have a relatively large sensitivity to temperature changes (-3 to -5 %/°C), and have excellent long-term stability characteristics (±0.2% of nominal resistance value per year).

The empirical relationship between the thermistor resistance R_t and absolute temperature T in kelvin (K) (the SI unit kelvin does not use a degree sign) is

$$R_t = R_o e^{\left[\frac{\beta (T_0 - T)}{T T_0}\right]}$$

Where: β: material constant for thethermistor, K

T_o: standard reference temperature, K

The value of β increases slightly with temperature. However, this does not present a problem over the limited temperature spans for biomedical work (10 °C to 20 °C). β, also known as the characteristic temperature, is 2500 to 5000 K. It is usually about 4000 K.

The temperature of the thermistor is that of its surroundings. However, above specific current, current flow generates heat that makes the temperature of the thermistor above the ambient temperature.

The temperature coefficient α can be found by differentiating with respect to T and dividing by R_t, Thus:

$$\alpha = \frac{1}{R_t}\frac{dR_1}{dT} = -\frac{\beta}{T^2} = (\%K)$$

The thermistor voltage-versus-current characteristic for a thermistor in air and water are plotted in this figure. The diagonal lines with a positive slope give linear resistance values and show the degree of thermistor linearity at low currents. The intersection of the thermistor curves and the diagonal lines with a negative slope gives the device power dissipation.

- Point A is the maximal current value for no appreciable self-heat.
- Point B is the peak voltage.
- Point C is the maximal safe continuous current in the air.