Principles of pressure transducers, resonance, damping and frequency response

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Learning objectives

After reading this article, you should be able to:

- explain the physical principles that underpin the measurement of pressure
- describe the principles and developments in transducer design
- outline the sources of error and inaccuracy in the clinical use of transducers

Abstract

Blood pressure is a determinant of blood flow, and is the sum of hydrostatic and dynamic pressures. Intravascular pressures can be measured directly using intravascular pressure sensors, or with external transducers connected by a fluid column. Early pressure transducers consisted of wire strain gauges, but these have been superseded by semiconductor devices, which have become increasingly mass-produced and miniaturized, using production techniques common in microelectronics. Performance of pressure-monitoring systems is affected by physical factors including resonance and damping. This article examines the physical principles that underlie transducer design and function, and the sources of error and inaccuracy.

Keywords Damping; frequency; hydrostatic; kinetic; micromachining; piezoresistive; pressure; resonance; strain gauge; transducer

Pressure

Blood pressure is a determinant of blood flow — that is, flow in a vascular bed is linearly related to the pressure gradient across it. Pressure (P) is the magnitude of a force (F) applied to a specific area (A). The SI unit, the Pascal (Pa), is defined as Newtons per square metre (N/m^2).

Static liquid in a container exerts 'hydrostatic pressure' (P_s), which increases with increasing depth from the surface. At the bottom of the container, the relationship between force and area is:

$$P_s = \frac{F}{A} = m \cdot g \cdot \frac{h}{V} = h \cdot w$$

where V = volume of the liquid, h = distance from surface, m = mass of the liquid, g = gravitational acceleration, w = weight per unit volume (mg/V).

Moving liquids exert 'dynamic pressure' (P_d) parallel to the direction of flow due to the kinetic energy of the liquid. The pressure exerted is related to liquid density and flow velocity:

$$P_d = \frac{1}{2}\rho \cdot v^2 = \frac{1}{2}\frac{m}{V} \cdot v^2$$

where ρ = fluid density, m = mass of the liquid, V = volume of the liquid, v = fluid flow velocity.

Bernoulli's equation expresses the total pressure exerted by the liquid (P_0) as $P_0 = P_s + P_d$.[1] The human vascular system is neither a cylinder of fluid, nor a single horizontal tube filled with flowing blood, but it is clear that blood pressure in different parts of the vascular tree varies, according to the height above the lowest point (static pressure), and the velocity of blood flow at the point of measurement (dynamic pressure).[2]

Pressure measurement

Gases expand to fill the volume they occupy, and equilibrate rapidly within a mechanical gauge. Non-compressible fluids, including blood, do not expand like gases, and the viscosity of the fluid precludes the rapid entry into and exit from simple gauges. With the rapid cyclical variation of blood pressure, incomplete equilibration of pressures with a mechanical gauge would impair accurate measurement.

To directly measure blood pressure, therefore, a system must consist of, either,

- a pressure sensor within a blood vessel, or
- a column of fluid in continuity with the blood vessel that transmits the pressure to an external pressure sensor.

General transducer principles

Pressure sensors are transducers — energy converters — which convert the kinetic and potential energy within a fluid to electrical energy, proportional to the magnitude of the pressure within it.

The generic design of pressure transducers quantifies the deflection of a square or circular diaphragm — placed between measurement and reference chambers — across which there is a pressure difference. The deflection of the diaphragm in sensing systems is frequently less than the thickness of the diaphragm itself, and is linearly related to the applied pressure. Pressure is either measured relative to a vacuum (absolute pressure), or relative to ambient pressure (gauge pressure). Pressure transducers in clinical use measure gauge pressure, that is, relative to atmospheric pressure (P_{atm}) (Figure 1).

The deflection of the diaphragm is measured using a form of **strain gauge** (Box 1). The original resistance-wire strain gauge, invented in 1936, measured increased electrical resistance of a wire, bonded to a diaphragm, in response to stretching or 'strain'.[3] **Resistivity** (ρ) (Ωm) is an innate property of a material,

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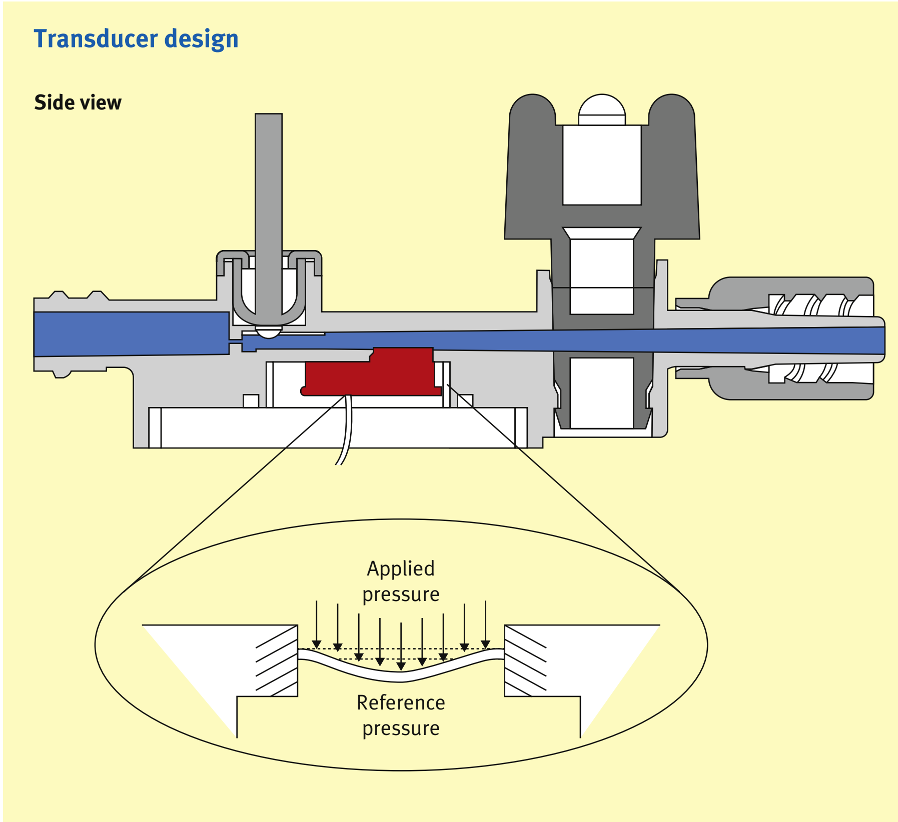

Figure 1 Cross-section of a transducer assembly. The sensor element is shown in red, and magnified schematically below. The fluid channel is shown in blue.

which defines how strongly it opposes conduction of electrical current.[4] A wire bonded to the diaphragm of a pressure transducer, of length l, radius r and resistivity ρ, has resistance (R),

$$R = \frac{\rho l}{\pi r^2}$$

where πr^2 is the cross-sectional area of the wire. If the wire is stretched during deflection of the diaphragm, the **resistance** change in the wire is:

$$\frac{\Delta R}{R} = \frac{\Delta l}{l} - \frac{2\Delta r}{r} + \frac{\Delta \rho}{\rho}$$

The extent to which the wire elongates under pressure is determined by a constant: **Young's modulus** (E). This is the ratio of tensile stress in the wire ($T = F/\pi r^2$) to the relative elongation (strain) of the wire ($\varepsilon = \Delta l/l$), and is expressed in units of pressure.[5]

The extent to which the thickness of the wire contracts under stretch is determined by the **Poisson ratio** (υ), such that $\Delta r/r = -\upsilon\ (\Delta l/l)$.[5] Most materials have a Poisson ratio of around 0.3.[4] So,

$$\frac{\Delta R}{R} = \varepsilon - 2(-\upsilon \cdot \varepsilon) + \frac{\Delta \rho}{\rho} = (1 + 2\upsilon) \cdot \varepsilon + \frac{\Delta \rho}{\rho}$$

Metals are homogeneous materials that are electrically isotropic, that is, their electrical properties are equally distributed within the material, and the values of those properties are identical irrespective of the direction in which they are measured. Therefore, in wire strain gauges, resistance change is essentially independent of material resistivity, which changes negligibly with deformation.

The relationship between stress and strain

Stress (s) (Pa or N/m^2) is force per unit area (i.e. pressure or 'load'). **Strain (e)** (%) is change in length relative to original length. If the relationship between stress and strain is linear, the material is said to obey **Hooke's Law**, and s/e is a constant called **Young's Modulus (E)** (Pa or N/m^2) which describes the 'stiffness' or 'elasticity' of the material.

Box 1

Gauge factor (G) is equivalent to the relative change in resistance ($\Delta R/R$), and is used to describe the performance of a transducer. Resistance is the easiest electrical property to measure accurately at low cost, but with small changes in resistance – in the region of 0.01–0.1% of base resistance – gauge factors for these sensors are low.[2,6]

The change in resistance of strain gauge resistors may be accentuated by arranging them in a 'Wheatstone bridge' configuration. The ratio of voltage change to supply voltage ($\Delta V/V_s$) is linearly related to $\Delta R/R$, and is typically adjusted to have an output of 5 µ V/V_s/mmHg. The use of double-bonded diaphragms exaggerates the output resulting from diaphragm deflection, by

