2.3 BRIDGE CIRCUITS

The Wheatstone bridge circuit is ideal for measuring small changes in resistance. Figure 2.2(b) shows a Wheatstone bridge with an applied dc voltage of v_i and a readout meter Δv_o with internal resistance R_i. It can be shown by the voltage-divider approach that Δv_o is zero—that is, the bridge is balanced—when $R_1/R_2 = R_4/R_3$.

Resistance-type sensors may be connected in one or more arms of a bridge circuit. The variation in resistance can be detected by measuring Δv_o with a differential amplifier feeding an analog-to-digital converter (ADC), which feeds a computer.

Assume that all values of resistance of the bridge are initially equal to R_0 and that $R_0 \ll R_1$. An increase in resistance, ΔR, of all resistances still results in a balanced bridge. However, if R_1 and R_3 increase by ΔR, and R_2 and R_4 decrease by ΔR, then

$$\Delta v_o = \frac{\Delta R}{R_0} v_i \tag{2.6}$$

Because of the symmetry a similar expression results if R_2 and R_4 increase by ΔR and R_1 and R_3 decrease by ΔR. Note that (2.6), for the four-active-arm bridge, shows that Δv_o is linearly related to ΔR. A nonlinearity in $\Delta R/R_0$ is present even when $R_0/R_1 = 0$.

It is common practice to incorporate a balancing scheme in the bridge circuit [see Figure 2.2(b)]. Resistor R_y and potentiometer R_x are used to change the initial resistance of one or more arms. This arrangement brings the bridge into balance so that zero voltage output results from ‘‘zero’’ (or *base-level*) input of the measured parameter.

To minimize loading effects, R_x is approximately 10 times the resistance of the bridge leg, and R_y limits the maximal adjustment. Strain-gage applications normally use a value of $R_y = 25$ times the resistance of the bridge leg. Alternating-current (ac) balancing circuits are more complicated because a reactive as well as a resistive imbalance must be compensated.

2.4 INDUCTIVE SENSORS

An inductance L can be used to measure displacement by varying any three of the coil parameters:

$$L = n^2 G \mu \tag{2.7}$$

where

n = number of turns of coil

G = geometric form factor

μ = effective permeability of the medium

Each of these parameters can be changed by mechanical means.

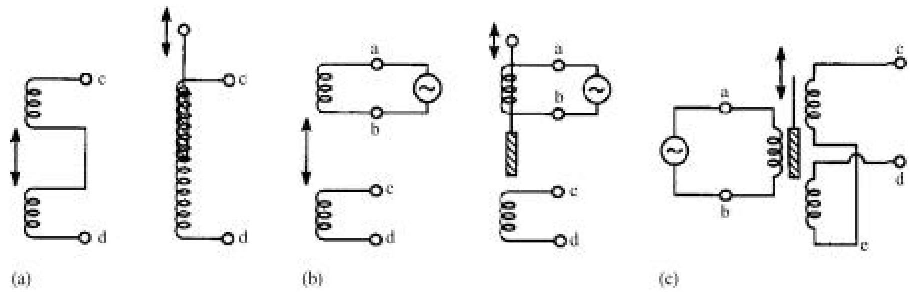


Figure 2.7 Inductive displacement sensors (a) self-inductance, (b) mutual inductance, (c) differential transformer.

Figure 2.7 shows (a) self-inductance, (b) mutual-inductance, and (c) differential transformer types of inductive displacement sensors. It is usually possible to convert a mutual-inductance system into a self-inductance system by series of parallel connections of the coils. Note in Figure 2.7 that the mutual-inductance device (b) becomes a self-inductance device (a) when terminals *b–c* are connected.

An inductive sensor has an advantage in not being affected by the dielectric properties of its environment. However, it may be affected by external magnetic fields due to the proximity of magnetic materials.

The variable-inductance method employing a single displaceable core is shown in Figure 2.7(a). This device works on the principle that alterations in the self-inductance of a coil may be produced by changing the geometric form factor or the movement of a magnetic core within the coil. The change in inductance for this device is not linearly related to displacement. The fact that these devices have low power requirements and produce large variations in inductance makes them attractive for radiotelemetry applications.

The mutual-inductance sensor employs two separate coils and uses the variation in their mutual magnetic coupling to measure displacement [Figure 2.7(b)]. Cobbold (1974) describes the application of these devices in measuring cardiac dimensions, monitoring infant respiration, and ascertaining arterial diameters.

Van Citters (1966) provides a good description of applications of mutual inductance transformers in measuring changes in dimension of internal organs (kidney, major blood vessels, and left ventricle). The induced voltage in the secondary coil is a function of the geometry of the coils (separation and axial alignment), the number of primary and secondary turns, and the frequency and amplitude of the excitation voltage. The induced voltage in the secondary coil is a nonlinear function of the separation of the coils. In order to maximize the output signal, a frequency is selected that causes the secondary coil (tuned circuit) to be in resonance. The output voltage is detected with standard demodulator and amplifier circuits.

The linear variable differential transformer (LVDT) is widely used in physiological research and clinical medicine to measure pressure, displacement, and force (Kesavan and Reddy, 2006). As shown in Figure 2.7(c), the LVDT is composed of a primary coil (terminals a–b) and two secondary coils (c–e and d–e) connected in series. The coupling between these two coils is changed by the motion of a high-permeability alloy slug between them. The two secondary coils are connected in opposition in order to achieve a wider region of linearity.

The primary coil is sinusoidally excited, with a frequency between 60 Hz and 20 kHz. The alternating magnetic field induces nearly equal voltages v_{ce} and v_{de} in the secondary coils. The output voltage $v_{cd} = v_{ce} - v_{de}$. When the slug is symmetrically placed, the two secondary voltages are equal and the output signal is zero.

Linear variable differential transformer characteristics include linearity over a large range, a change of phase by 180° when the core passes through the center position, and saturation on the ends. Specifications of commercially available LVDTs include sensitivities on the order of 0.5 to 2 mV for a displacement of 0.01 mm/V of primary voltage, full-scale displacement of 0.1 to 250 mm, and linearity of ±0.25%. Sensitivity for LVDTs is much higher than that for strain gages.

A disadvantage of the LVDT is that it requires more complex signal-processing instrumentation. Figure 2.8 shows that essentially the same

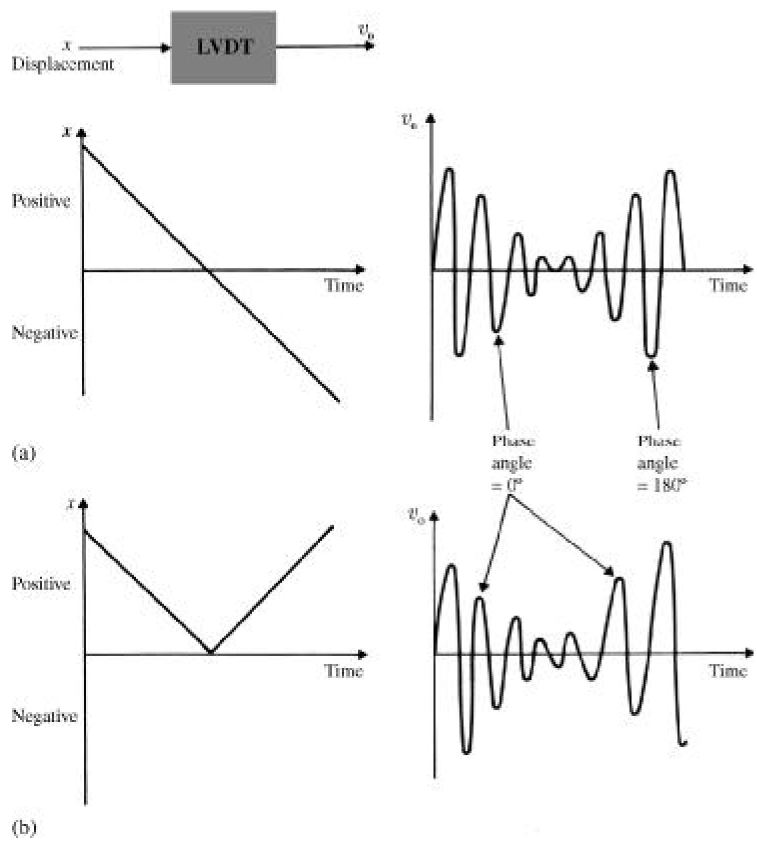


Figure 2.8 (a) As x moves through the null position, the phase changes 180°, while the magnitude of v_o is proportional to the magnitude of x. (b) An ordinary rectifier demodulator cannot distinguish between (a) and (b), so a phase-sensitive demodulator is required.

magnitude of output voltage results from two very different input displacements. The direction of displacement may be determined by using the fact that there is a 180° phase shift when the core passes through the null position. A phase-sensitive demodulator is used to determine the direction of displacement. Figure 3.17 shows a ring-demodulator system that could be used with the LVDT.

2.5 CAPACITIVE SENSORS

The capacitance between two parallel plates of area A separated by distance x is

$$C = \varepsilon_0 \varepsilon_r \frac{A}{x} \tag{2.8}$$

where ε_0 is the dielectric constant of free space (Appendix A.1) and ε_r is the relative dielectric constant of the insulator (1.0 for air) (Bowman and Meindl, 1988). In principle it is possible to monitor displacement by changing any of the three parameters ε_r, A, or x. However, the method that is easiest to implement and that is most commonly used is to change the separation between the plates.

The sensitivity K of a capacitive sensor to changes in plate separation Δx is found by differentiating (2.8).

$$K = \frac{\Delta C}{\Delta x} = -\varepsilon_0 \varepsilon_r \frac{A}{x^2} \tag{2.9}$$

Note that the sensitivity increases as the plate separation decreases.

By substituting (2.8) into (2.9), we can develop an expression showing that the percent change in C about any neutral point is equal to the per-unit change in x for small displacements. Thus

$$\frac{dC}{dx} = \frac{-C}{x} \tag{2.10}$$

or

$$\frac{dC}{C} = \frac{-dx}{x} \tag{2.11}$$

The capacitance microphone shown in Figure 2.9 is an excellent example of a relatively simple method for detecting variation in capacitance (Doebelin, 1990; Cobbold, 1974). This is a dc-excited circuit, so no current flows when the capacitor is stationary (with separation x_0), and thus $v_1 = E$. A change in

8.3 ELECTROMAGNETIC FLOWMETERS

The electromagnetic flowmeter measures instantaneous pulsatile flow of blood and thus has a greater capability than indicator-dilution methods, which measure only average flow. It operates with any conductive liquid, such as saline or blood.

PRINCIPLE

The electric generator in a car generates electricity by induction. Copper wires move through a magnetic field, cutting the lines of magnetic flux and inducing an emf in the wire. This same principle is exploited in a commonly used blood flowmeter, shown in Figure 8.3. Instead of copper wires, the flowmeter depends on the movement of blood, which has a conductance similar to that of saline. Faraday's law of induction gives the formula for the induced emf.

$$e = \int_0^{L_1} \mathbf{u} \times \mathbf{B} \cdot d\mathbf{L}$$

where

$\mathbf{B}$ = magnetic flux density, T

$\mathbf{L}$ = length between electrodes, m

$\mathbf{u}$ = instaneous velocity of blood, m/s

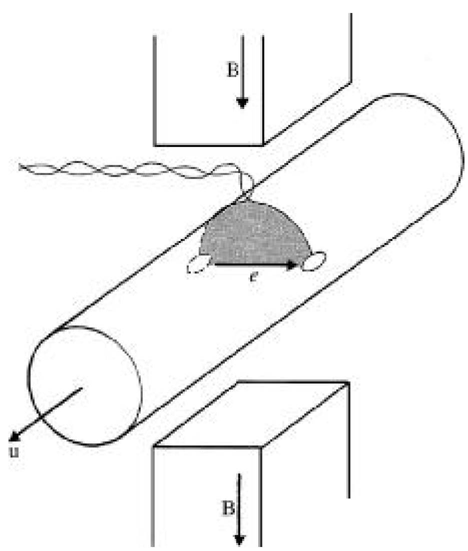


Figure 8.3 Electromagnetic flowmeter When blood flows in the vessel with velocity $\mathbf{u}$ and passes through the magnetic field $\mathbf{B}$, the induced emf e is measured at the electrodes shown. When an ac magnetic field is used, any flux lines cutting the shaded loop induce an undesired transformer voltage.

For a uniform magnetic field B and a uniform velocity profile u, the induced emf is

$$e = BLu \tag{8.8}$$

where these three components are orthogonal.

Let us now consider real flowmeters, several of which exhibit a number of divergences from this ideal case. If the vessel's cross section were square and the electrodes extended the full length of two opposite sides, the flowmeter would measure the correct average flow for any flow profile. The electrodes are small, however, so velocities near them contribute more to the signal than do velocities farther away.

Figure 8.4 shows the weighting function that characterizes this effect for circular geometry. It shows that the problem is less when the electrodes are located outside the vessel wall. The instrument measures correctly for a uniform flow profile. For axisymmetric nonuniform flow profiles, such as the parabolic flow profile resulting from laminar flow, the instrument measurement is correct if u is replaced by $\overline{u}$, the average flow velocity. Because we usually know the cross-sectional area A of the lumen of the vessel, we can multiply A by $\overline{u}$ to obtain F, the volumetric flow. However, in many locations of

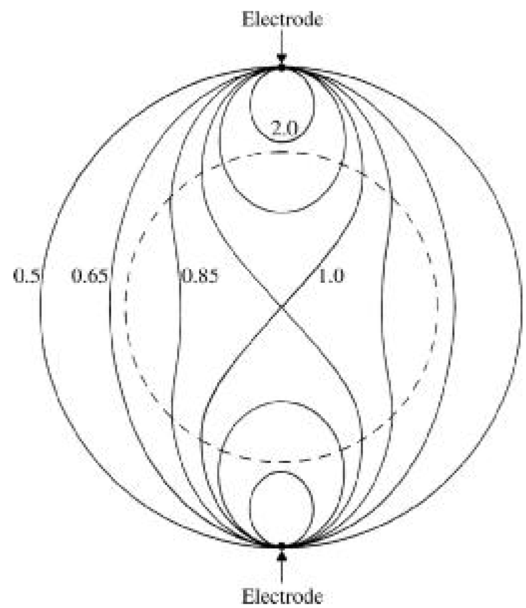


Figure 8.4 Solid lines show the weighting function that represents relative velocity contributions (indicated by numbers) to the total induced voltage for electrodes at the top and bottom of the circular cross section. If the vessel wall extends from the outside circle to the dashed line, the range of the weighting function is reduced. (Adapted from J. A. Shercliff, *The Theory of Electromagnetic Flow Measurement*, © 1962, Cambridge University Press.)

blood vessels in the body, such as around the curve of the aorta and near its branches, the velocity profile is asymmetric, so errors result.

Other factors can also cause error.

1. Regions of high velocity generate higher incremental emfs than regions of low velocity, so circulating currents flow in the transverse plane. These currents cause varying drops in resistance within the conductive blood and surrounding tissues.
2. The ratio of the conductivity of the wall of the blood vessel to that of the blood varies with the *hematocrit* (percentage of cell volume to blood volume), so the shunting effects of the wall cause a variable error.
3. Fluid outside the wall of the vessel has a greater conductivity than the wall, so it shunts the flow signal.
4. The magnetic-flux density is not uniform in the transverse plane; this accentuates the problem of circulating current.
5. The magnetic-flux density is not uniform along the axis, which causes circulating currents to flow in the axial direction.

To minimize these errors, most workers recommend calibration for animal work by using blood from the animal—and, where possible, the animal's own vessels also. Blood or saline is usually collected in a graduated cylinder and timed with a stopwatch.

DIRECT-CURRENT FLOWMETER

The flowmeter shown in Figure 8.3 can use a dc magnetic field, so the output voltage continuously indicates the flow. Although a few early dc flowmeters were built, none were satisfactory, for the following three reasons. (1) The voltage across the electrode's metal-to-solution interface is in series with the flow signal. Even when the flowmeter has nonpolarizable electrodes, the random drift of this voltage is of the same order as the flow signal, and there is no way to separate the two. (2) The ECG has a waveform and frequency content similar to that of the flow signal; near the heart, the ECG's waveform is much larger than that of the flow signal and therefore causes interference. (3) In the frequency range of interest, 0 to 30 Hz, $1/f$ noise in the amplifier is large, which results in a poor SNR.

ALTERNATING-CURRENT FLOWMETER

The clinician can eliminate the problems of the dc flowmeter by operating the system with an ac magnet current of about 400 Hz. Lower frequencies require bulky sensors, whereas higher frequencies cause problems due to stray capacitance. The operation of this carrier system results in the ac flow voltage shown in Figure 8.5. When the flow reverses direction, the voltage changes phase by 180°, so the phase-sensitive demodulator (described in Section 3.15) is required to yield directional output.

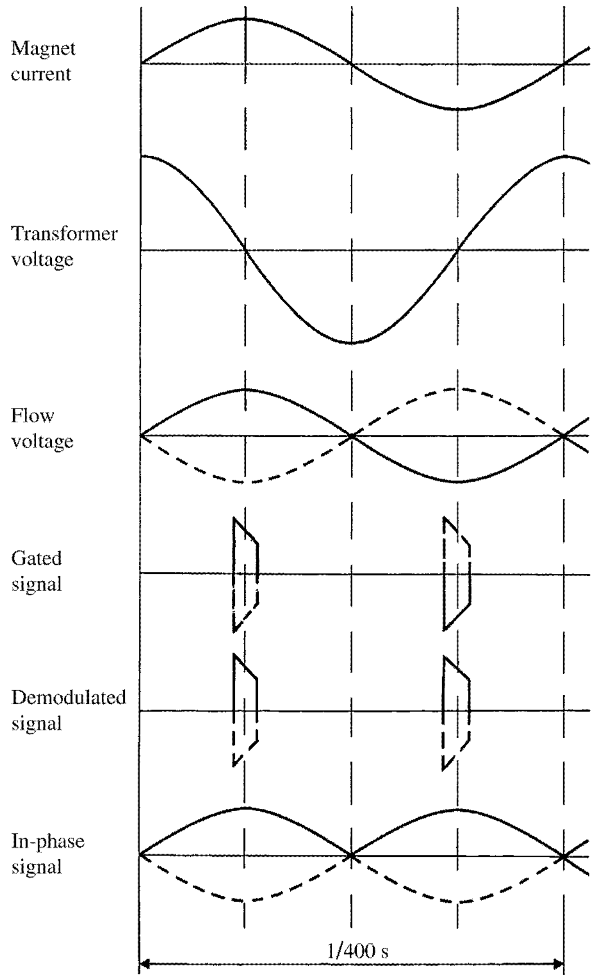


Figure 8.5 Electromagnetic flowmeter waveforms The transformer voltage is 90° out of phase with the magnet current. Other waveforms are shown solid for forward flow and dashed for reverse flow. The gated signal from the gated-sine-wave flowmeter includes less area than the in-phase signal from the quadrature-suppression flowmeter.

Although ac operation is superior to dc operation, the new problem of *transformer voltage* arises. If the shaded loop shown in Figure 8.3 is not exactly parallel to the B field, some ac magnetic flux intersects the loop and induces a transformer voltage proportional to dB/dt in the output voltage. Even when the electrodes and wires are carefully positioned, the transformer voltage is usually many times larger than the flow voltage, as indicated in Figure 8.5. The amplifier voltage is the sum of the transformer voltage and the flow voltage.

There are several solutions to this problem. (1) It may be eliminated at the source by use of a *phantom electrode*. One of the electrodes is separated into

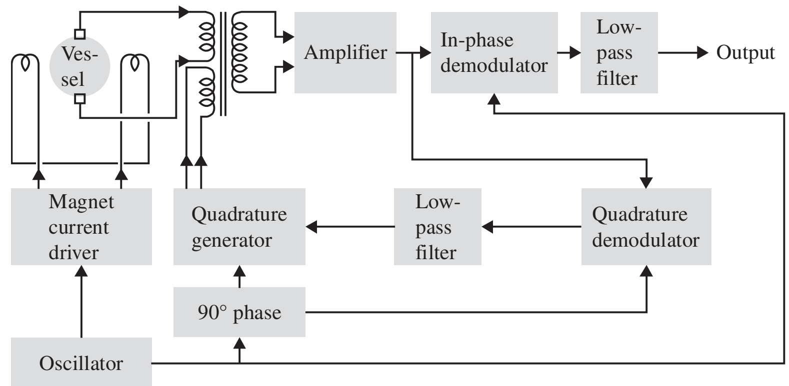


Figure 8.6 The quadrature-suppression flowmeter detects the amplifier quadrature voltage. The quadrature generator feeds back a voltage to balance out the probe-generated transformer voltage.

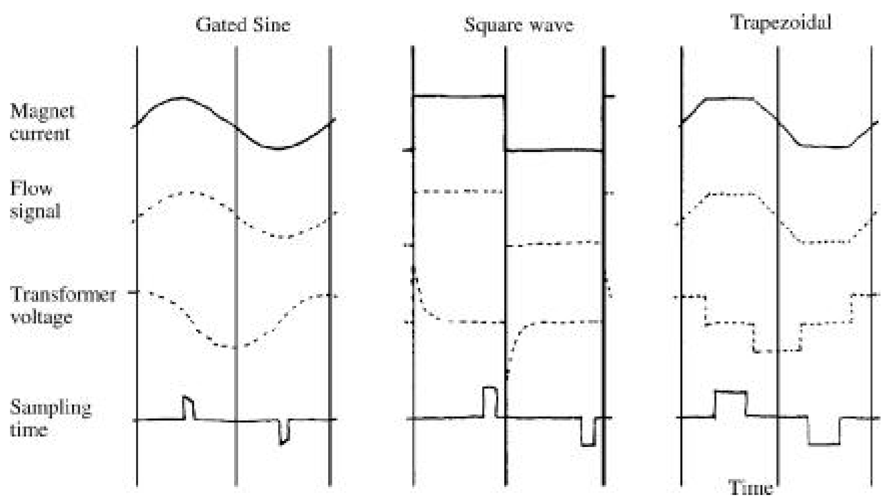
two electrodes in the axial direction. Two wires are led some distance from the electrodes, and a potentiometer is placed between them. The signal from the potentiometer wiper yields a signal corresponding to a ''phantom'' electrode, which can be moved in the axial direction. The shaded loop in Figure 8.3 can thus be tilted forward or backward or placed exactly parallel to the B field. (2) Note in Figure 8.5 that we can sample the composite signal when the transformer voltage is zero. At this time the flow voltage is at its maximum, and the resulting *gated signal* measures only the flow voltage. However, if undesired phase shifts cause the gating to be done even a few degrees away from the proper time, large errors and drifts result. (3) The best method for reducing the effects of transformer voltage is to use the *quadrature-suppression* circuit shown in Figure 8.6.

The magnitude of the voltage in the transformer at the amplifier output is detected by the quadrature demodulator, which has a full-wave-rectified output. This is low-pass-filtered to yield a dc voltage, which is then modulated by the quadrature generator to produce a signal proportional to the transformer voltage. The signal is fed to a balancing coil on the input transformer, thus balancing out the transformer voltage at the input. With enough gain in this negative-feedback loop, the transformer voltage at the amplifier output is reduced by a factor of 50. This low transformer voltage prevents overloading of the in-phase demodulator, which extracts the desired in-phase flow signal shown in Figure 8.5. By choosing low-noise FETs for the amplifier input stage, the proper turns ratio on the step-up transformer (Section 3.13), and full-wave demodulators, we can obtain an excellent SNR.

Some flowmeters, unlike the sine-wave flowmeters described previously, use *square-wave excitation*. In this case the transformer voltage appears as a very large spike, which overloads the amplifier for a short time. After the amplifier recovers, the circuit samples the square-wave flow voltage and processes it to obtain the flow signal. To prevent overload of the amplifier, *trapezoidal excitation* has also been used.

EXAMPLE 8.3 On a common time scale, sketch the waveforms for the magnet current, flow signal, and transformer voltage for the following electromagnetic flowmeters: (1) gated sine wave, (2) square wave, and (3) trapezoidal. Indicate the best time for sampling each flow signal.

ANSWER For gated sine wave, waveforms are exactly like those in Figure 8.5. Sample the composite signal when the transformer voltage is zero. Transformer voltage is proportional to dB/dt. Taking the derivative of square wave B yields spikes at transitions. Because the amplifier is not perfect, these take time to decay. Best time to sample is near the end of transformer voltage $= 0$. Trapezoidal B yields reasonable dB/dt, so sample during time transformer voltage $= 0$.



PROBE DESIGN

A variety of probes to measure blood flow have been used (Cobbold, 1974). The electrodes for these probes are usually made of platinum. Best results are obtained when the electrodes are platinized (electrolytically coated with platinum) to provide low impedance and are recessed in a cavity to minimize the flow of circulating currents through the metal. When the electrodes must be exposed, bright platinum is used, because the platinized coating wears off anyway. Bright platinum electrodes have a higher impedance and a higher noise level than platinized ones.

Some probes do not use a magnetic core, but they have lower sensitivity. A common *perivascular probe* is shown in Figure 8.7, in which a toroidal laminated Permalloy core is wound with two oppositely wound coils. The

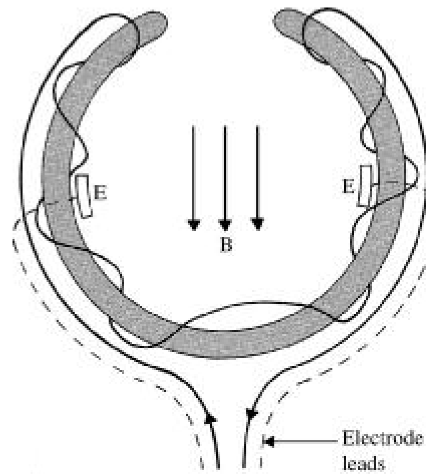


Figure 8.7 The toroidal-type cuff probe has two oppositely wound windings on each half of the core. The magnetic flux thus leaves the top of both sides, flows down in the center of the cuff, enters the base of the toroid, and flows up through both sides.

resulting magnetic field has low leakage flux. To prevent capacitive coupling between the coils of the magnet and the electrodes, an electrostatic shield is placed between them. The probe is insulated with a potting material that has a very high resistivity and impermeability to salt water (blood is similar to saline).

The open slot on one side of the probe makes it possible to slip it over a blood vessel without cutting the vessel. A plastic key may be inserted into the slot so that the probe encircles the vessel. The probe must fit snugly during diastole so that the electrodes make good contact. This requires some constriction of an artery during systole, when the diameter of the artery is about 7% greater. Probes are made in 1 mm increments in the range of 1 to 24 mm to ensure a snug fit on a variety of sizes of arteries. To be able to measure any size of artery requires a considerable expenditure for probes: Individual probes typically cost $500 each. The probes do not operate satisfactorily on veins, because the electrodes do not make good contact when the vein collapses. Special flow-through probes are used outside the body for measuring the output of cardiac-bypass pumps.

8.4 ULTRASONIC FLOWMETERS

The ultrasonic flowmeter, like the electromagnetic flowmeter, can measure instantaneous flow of blood. The ultrasound can be beamed through the skin, thus making transcutaneous flowmeters practical. Advanced types of ultrasonic flowmeters can also measure flow profiles. These advantages are making