cal role is not known (*5*, *13*). Figure 7 shows absorption spectra for red, yellow, and green globules measured *in situ*. The vertical lines indicate the observed variability for each colored globule. These spectra were measured at a magnification of 500 through a specimen area of 2 μ^2. The reference area was immediately adjacent to the globules and contained a slight amount of retinal material. Spectra taken of globules completely isolated from retinal material indicated no significant error resulting from this procedure over the wavelength range under investigation. These results indicate that the different colored globules within the retina can act as bandpass color filters if they are suitably located in relation to a photosensitive pigment (*5*, *15*).

The spectral curves shown in Fig. 7 indicate the presence of carotenoids, a finding in agreement with the biochemical investigation of Wald and Zussman (*15*) on retinal extracts of chicken. The green globules have a major absorption peak near 420 mμ, indicating the presence of galloxanthin, a carotenoid isolated by Wald from the chicken retina (*16*). The yellow globules have a major absorption peak in the region 470 to 480 mμ. Absorption in this portion of the visible spectrum is typical for both xanthophylls and carotenes. Both lutein and zeaxanthin are present in the chicken retina (*15*), and the yellow globules are believed to be composed mainly of these two carotenoids. The red globules have a broad absorption maximum near 500 mμ, indicating the presence of astaxanthin (*15*). The absorption spectra obtained from the colored globules are not from purified compounds but from mixtures of pigments in the natural state. Therefore, some wavelength shifts toward longer wavelengths are to be expected from spectra reported in the literature. These pigmented globules were also investigated in the ultraviolet region of the spectrum. The presence of lipids was indicated by strong absorption at 310 mμ and below (*13*).

The illustrated spectra described are only an indication of the possible usefulness of the instrument. The microspectrophotometer is well suited for study of the effects of chemical and physical environment on the synthesis and concentration of pigments within a single cell or of organelles within a cell (*17*).

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17. This work was aided by grants from the U.S. Public Health Service, National Institute of Neurological Diseases and Blindness (B-397-C5 and B-1874-A), and from the International Lions Club (14 B).

Electromagnetic Blood Flow Meters

Implantable flow transducers facilitate circulatory studies in conscious and free-moving animals.

Alexander Kolin

The interest in a method for determination of blood flow goes beyond the need for such methods in studies of hemodynamics. The wider scope of the potentialities of an effective method lies in the possibility of using it in a general way as an index of the activity of a great variety of organs whose performance is normally studied by quite diverse methods specifically adapted to the function of the organ in question. The rate of blood supply to an organ is a determining factor in regulation of the supply of oxygen, hormones, and nutrient materials. It similarly determines the rate of removal of metabolic products. The blood flow through an organ, when correlated with its rate of activity, could thus be used to follow variations in the organ's activity in response to a variety of stimuli and inhibiting factors.

Ideally, the method of blood-flow determination should permit continuous recording of blood flow in intact undisplaced blood vessels of conscious, freely moving animals. The application of the principle of electromagnetic induction to measurement of blood flow (*1–3*) resulted in a method which, in subsequent modifications, made it possible to record blood flow by means of nonirritating implanted measuring devices (*4–7*). The method is based on the induction of an electromotive force in a fluid flowing transversely through a homogeneous magnetic field. In the case of a circular conduit and axially symmetrical flow, the induced electromotive force is a linear function of the average fluid discharge. In the case of a conductive conduit, the electrical signal can be picked up without making contact with the fluid by establishing contact with two points on the outside wall of the conduit situated at the two ends of a diameter perpendicular to the magnetic field. This configuration is illustrated in Fig. 1.

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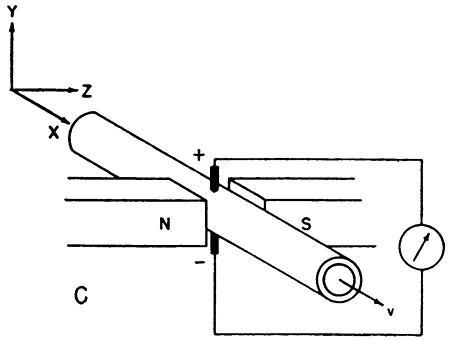
Modifications of the Electromagnetic Flow Meter

The early form of this method was based on use of a constant magnetic field (*1–3*). This required employment of cumbersome nonpolarizable electrodes. The introduction of an alternating magnetic field (*3, 8*) made it possible to use ordinary metal electrodes. This advantage, however, was not gained without sacrifice. Whereas, with a constant magnetic field it is possible to establish a base line corresponding to zero flow by switching off the magnetic field, it is no longer possible to do so without special adjustment in the case of an alternating magnetic field (*9*). A successful attempt was made to combine the advantages of the constant magnetic field with those of the alternating magnetic field by energizing the magnet with a square wave current (*10*). This, however, has been accomplished only through great complexity of circuit design.

The system that I currently prefer utilizes magnets energized by a sinusoidal current and uses phase-sensitive detectors to separate the flow signal from the flow-independent signal induced by transformer action in the input lead circuit (*7*). Phase-sensitive detection was initially based on optical and photoelectric means (*4, 8, 11*). An electronic method of sampling the sine-wave signal derived from an electromagnetic flow meter was suggested for the first time by W. J. James (*12*). The phase-sensitive amplifier used in my laboratory (*7*) provides amplification of signals of less than 1 microvolt (μv) at a noise level of less than 0.2 μv. This high sensitivity makes it possible to de-



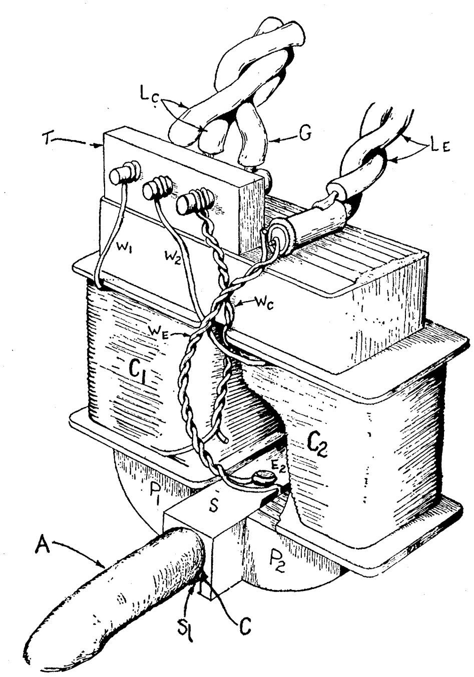


Fig. 1 (left). Configuration of flow (v), magnetic field (magnet poles, N, S), and pickup electrodes (+, —) in an electromagnetic flow meter. Fig. 2 (right). Skeleton of an electromagnetic blood flow meter with inserted artery (A). S, sleeve; Sl, slit for insertion of artery; C, sleeve channel; E_2, one of the pickup electrodes which contact the outer wall of the artery; P_1 and P_2, pole pieces of the magnet; C_1 and C_2, magnet coils; L_E and W_E, electrode leads; G, ground lead; T, terminal board.

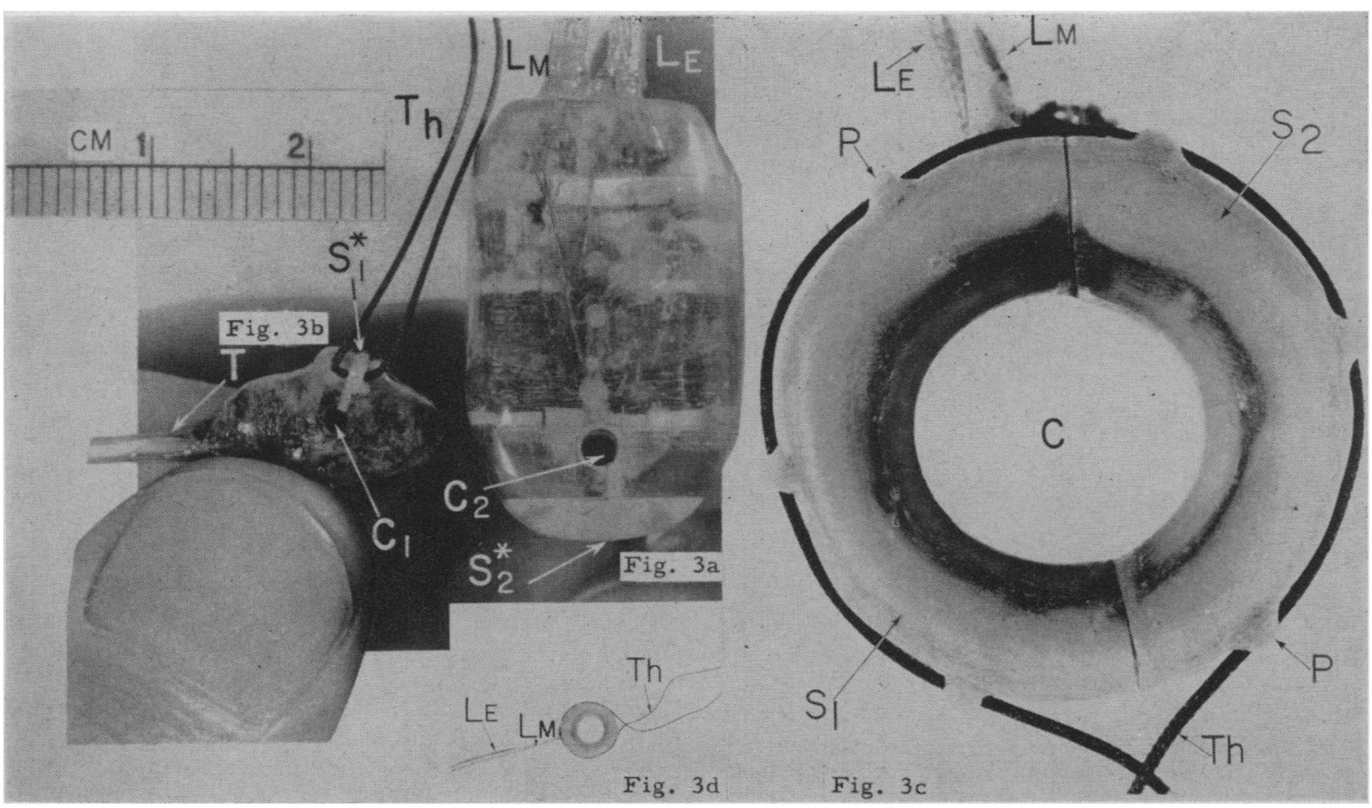


Fig. 3. Implantable electromagnetic blood flow meters. (a) The original implantable "miniature" flow meter for a 2-mm artery. C_2, artery channel; S^*_2, shutter preventing the artery from slipping out; L_M, magnet leads; L_E, electrode leads. (b) Subminiature flow meter for a 1.5-mm artery. T, tubing containing the leads; C_1, artery channel; S^*_1, shutter; Th, thread. (c) Coreless flow meter for a 15-mm artery. This unit has no hinges (compare Fig. 6). The two halves, S_1 and S_2, are tied together by the thread Th. P, thread holes; C, artery channel; L_E, electrode leads; L_M, magnet leads. (d) Flow meter shown in Fig. 3c reduced to caliber of Fig. 3b. The smallness of the space occupied by the magnet as compared to Fig. 3b is conspicuous in this design. [From Kolin and Kado (*7*)]

sign the very small implantable flow meters described below.

The selection of the frequency of the alternating magnetic field depends on several factors. If the details of the flow throughout the cardiac cycle are to be studied, a high-frequency carrier of adequate bandwidth is essential. The selection of a high frequency is also of advantage in minimizing electrode polarization. On the other hand, the spurious flow-independent signals which are introduced into the electrode circuit from the magnet circuit by electromagnetic induction (transformer action) and by electrostatic coupling increase in proportion to the frequency. The pickup due to galvanic coupling (current leakage from the magnetic power source into the electrode circuit due to poor insulation) is independent of frequency.

The foregoing survey classifies the modes of implementation of the electromagnetic flow-meter principle according to the time function of the magnetic field used. The mode of application of the magnetic field and of the electrodes provides an additional important criterion of classification. We can distinguish between the following main modes of application.

1) In the canula-type flow meter (*13*), a short dielectric tube with sensing electrodes is placed in a magnetic field, as shown in Fig. 1, and firmly attached to the flow-meter magnet. The blood vessel is cut, and the flow-meter "canula" is inserted between the two ends of the cut artery. The blood vessel is exteriorized, and the animal must be anesthetized and heparinized. The use of this modification is indicated where the flows to be measured are very minute, so that the magnets which are used with the modifications described below are not intense enough to provide the needed sensitivity.

2) In the flow meter for intact exteriorized arteries (*1–3, 8*), the position of the blood vessel and the state of the animal are the same as in the preceding modification except that the blood vessel is not cut. It is, rather, inserted into a sleeve *S* similar to the one shown in Fig. 2. The sleeve is equipped with electrodes E_1 and E_2, which contact the blood-vessel wall at the end of a diameter perpendicular to the magnetic field maintained between the poles (P_1 and P_2) of the magnet. This approach is made obsolete by the two following modifications.

3) The implanted electromagnetic flow transducer is, in principle, the same as the flow meter described in

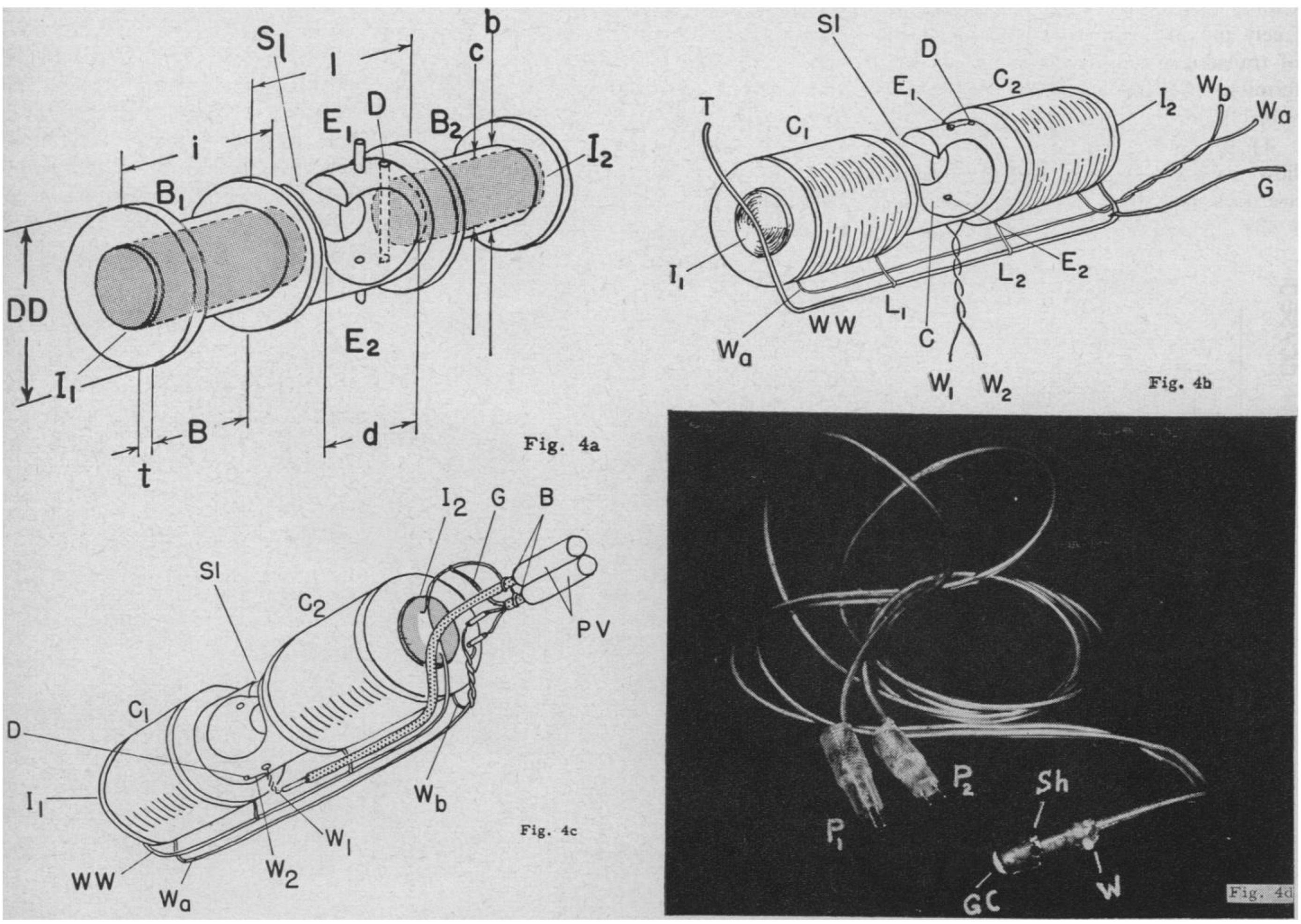


Fig. 4. Implantable subminiature electromagnetic flow meter. (a) Sleeve and bobbins B_1 and B_2, for the subminiature blood flow meter. *Sl*, slit; E_1 and E_2, electrodes; *D*, channel for electrode wire; I_1 and I_2, iron cores. The other letters refer to letters in Table 1. (b) Skeleton of subminiature flow meter. E_1 and E_2, electrodes; *Sl*, slit; *D*, electrode wire channel; W_1 and W_2, electrode wires; C_1 and C_2 magnet coils; *T*, grounding "tail"; W_a and W_b, magnet leads; *G*, ground lead; *WW*, wire connecting the iron core ends; L_1 and L_2, wires connecting the coils in series; *C*, artery channel. (c) Insulation and shielding scheme of the subminiature flow meter. *Sl*, Slot; W_2, electrode wire; W_{12}, electrode lead wires; C_1 and C_2, magnet coils; I_1 and I_2, iron cores; *B*, grounded wire braid; *G*, ground lead; *PV*, polyvinyl chloride tubing; W_a and W_b, coil lead wires; *WW*, wire joining cores I_1 and I_2. (d) Compression-molded subminiature flow meter. The diameter of the sleeve channel is 3 mm. P_1 and P_2, plugs; *GC*, gold cap; *Sh*, Teflon shutter, *W*, Teflon wing (a rod inserted into a sliding bed cast within the TDM body).

modification 2. The modification consists of minimizing the size of the magnet so as to make the flow meter implantable, and of casting the magnet and sleeve into an integral unit. The plastic material secures good electrical insulation between the magnet coils and the electrode circuit. This is very important, since the slightest electrical leakage will give rise to an excessively large spurious signal at zero flow.

The development of high-gain, low-noise amplifiers made it possible to diminish the strength of the magnetic field to a point where the implantable flow meters are small enough to represent a negligible source of mechanical irritation.

The advantage of this modification lies in the possibility of permitting the animal to recover after the implantation of the flow meter and of carrying out observations of blood flow on conscious, essentially normal, and nearly freely moving animals. The same kind of transducer can also be used for determination of blood flow in human subjects during surgery.

4) In spite of the small size of the flow meters of modification 3, there are occasions when it is advisable to go even further in miniaturizing the implant. At the same time, it is desirable to simplify the design of the miniaturized flow transducers, since the fabrication of extremely small transducers consisting of a miniature electromagnet with slotted sleeve, shutter, coils, shields, and so on, is a very complex and delicate job.

Both the simplification and the miniaturization of the implant can be achieved by implanting only the sleeve with pickup electrodes and establishing the magnetic field by means of coils external to the animal body (*4, 14*).

This method possesses the disadvantage of allowing less freedom of movement to the animal, but this is compensated for by the advantage that an implant of smallest possible size, and of the utmost simplicity of construction, can be used. In addition, many simultaneous implants can be used, for simultaneous recording of blood flow in different blood vessels, with a single, large coil to produce the external field.

In this article I have concentrated on discussion of the last two modifications, which, in turn, can be implemented in different ways.

Implanted Blood Flow Meter with Internal Magnet

The flow meter shown in Fig. 2 is actually the skeleton of an improved design (*6*) of a miniature transducer developed for implantation (*5*). Figure 3 shows the evolution of the implantable flow meter incorporating a magnet. Figure 3a shows the original design for a 2-mm artery, and Figure 3b shows, for comparison, the subminiature design of neighboring caliber (1.5 mm) developed subsequently (*6, 7*). Figure 3c shows a flow meter with a coreless magnet, developed for large arteries, and Fig. 3d shows its channel reduced to the same size as that of Fig. 3b. The great reduction in the size of the implant as compared with the channel diameter is quite apparent. The description in this section is limited to the most recent designs of the types shown in Figs. 3b and 3c, which I shall designate the "linear core" unit and the "coreless" unit, respectively. The former design is most effective and convenient for arteries up to about 4 mm in diameter, whereas the second design is used for arteries over 5 mm in diameter.

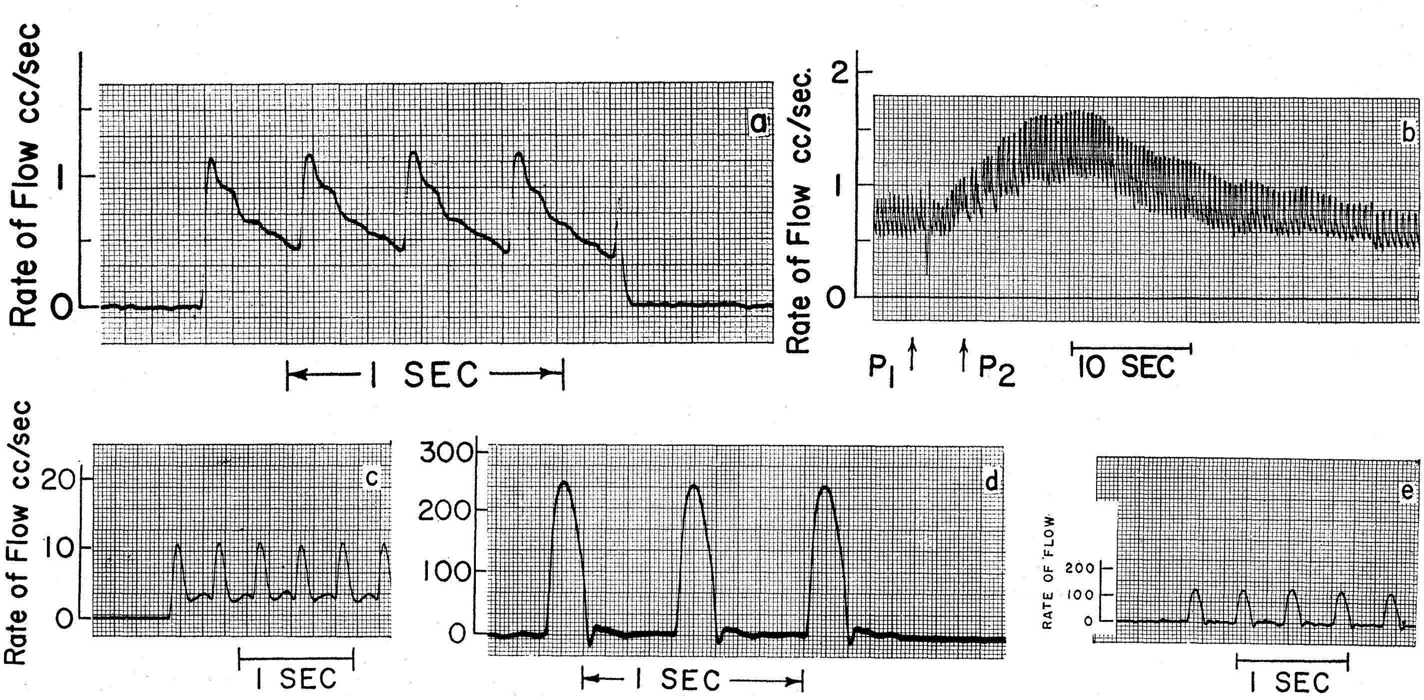


Fig. 5. Blood-flow records obtained (mainly in the course of as yet unpublished studies with L. W. Roth) with different arteries and with different types of implantable flow meters. The ordinate is rate of flow in cubic centimeters per second. The horizontal sections at the left or right ends of the records are base lines obtained by switching off the magnet. (a, b) Blood flow in a conscious cat's carotid artery (1.5 mm in diameter) recorded with a subminiature electromagnetic flow meter; (a) flow contour throughout the cardiac cycle taken at fast recording paper speed; (b), effect of exposure of a conscious cat to ammonia vapors (between the marks P_1 and P_2), taken at slow speed of recording paper. (c) Blood flow in the carotid artery (3 mm in diameter) of a conscious dog, recorded with a transducer of the type shown in Fig. 4d. (d) Record of blood flow in the ascending aorta of a conscious dog, made with a coreless flow meter of the type shown in Fig. 6. (e) Blood flow in the ascending aorta of a conscious dog, recorded by means of the external-field method; a strapped-on magnet (shown in the photograph in Fig. 8) was used, and there was no compensating coil.

"Linear Core" Flow Meter

Figure 4 shows the alignment of the coils (C_1 and C_2), iron cores I_1 and I_2, and sleeve S in the "linear core" flow meter type of transducer (*15*). Figure 4a shows the Lucite form, which can be cast or machined from a Lucite rod. The rod is turned in a lathe so as to produce two bobbins, B_1 and B_2, on which the coil wires are to be wound. The iron cores I_1 and I_2 slip tightly into the bobbins. They should, preferably, be made of laminated material, iron powder, or ferrite, but solid iron cores can be used without excessive heating for units accommodating arteries below 2.5 mm in diameter. Channel C (Fig. 4b) is drilled at right angles to the axis of the Lucite cylinder from which the Lucite skeleton shown in Fig. 4a is machined. The artery is slipped into this channel through the slit *Sl*. A very narrow channel (½ mm) is drilled *precisely* at right angles to channel C. The gold electrodes E_1 and E_2 are introduced through this channel. Their ends are filed flush with the walls of channel C. The exact perpendicularity of the electrode channel to channel C is essential to minimize the potential difference which would arise between electrodes E_1 and E_2 due to currents parallel to the axis of channel C.

A narrow channel D, about 0.13 mm in diameter, between channel C and the end of the iron core I_2, is drilled in the plane passing through the electrodes and the axis of I_1 and I_2. The lead wire W_1 (compare Fig. 4b) connected to E_1 passes through it to join the lead W_2, with which it is twisted (gauge No. 39 wires with heavy Teflon insulation are used for adequate breaking strength).

Figure 4b shows a perspective drawing of the flow-meter skeleton with the coils C_1 and C_2 wound onto the bobbins B_1 and B_2 of Fig. 4a. Wires L_1 and L_2 connect the two coils in series through wire WW, which joins the ends of the cores I_1 and I_2 to each other and to the ground lead G. The "tail" is allowed to emerge to the surface of the flow meter after it has been cast in plastic material (see below). The emerging end of the "tail" is pressed flat against the plastic enclosure of the flow meter after completion of the cast and is painted, as is the adjacent surface at the end of the unit, with silver paint (Du Pont No. 4922). This silver-painted spot is then silver-plated in a potassium silver cyanide solution at 30 ma/cm^2 for 30 minutes and subsequently gold-plated in a potassium gold cyanide solution at 5 ma/cm^2 for ½ hour. The gold cap thus formed at the end of the flow meter grounds the animal.

Figure 4c shows the shielding and insulating scheme of the leads. The electrode leads W_1 and W_2 are shielded by the grounded wire braid B, which is enclosed in a polyvinyl chloride tubing PV. The coil leads W_a and W_b and the ground lead G are similarly enclosed in polyvinyl chloride tubing. The proper performance and durability of the transducer depends on achieving a very good bond between the polyvinyl chlo-

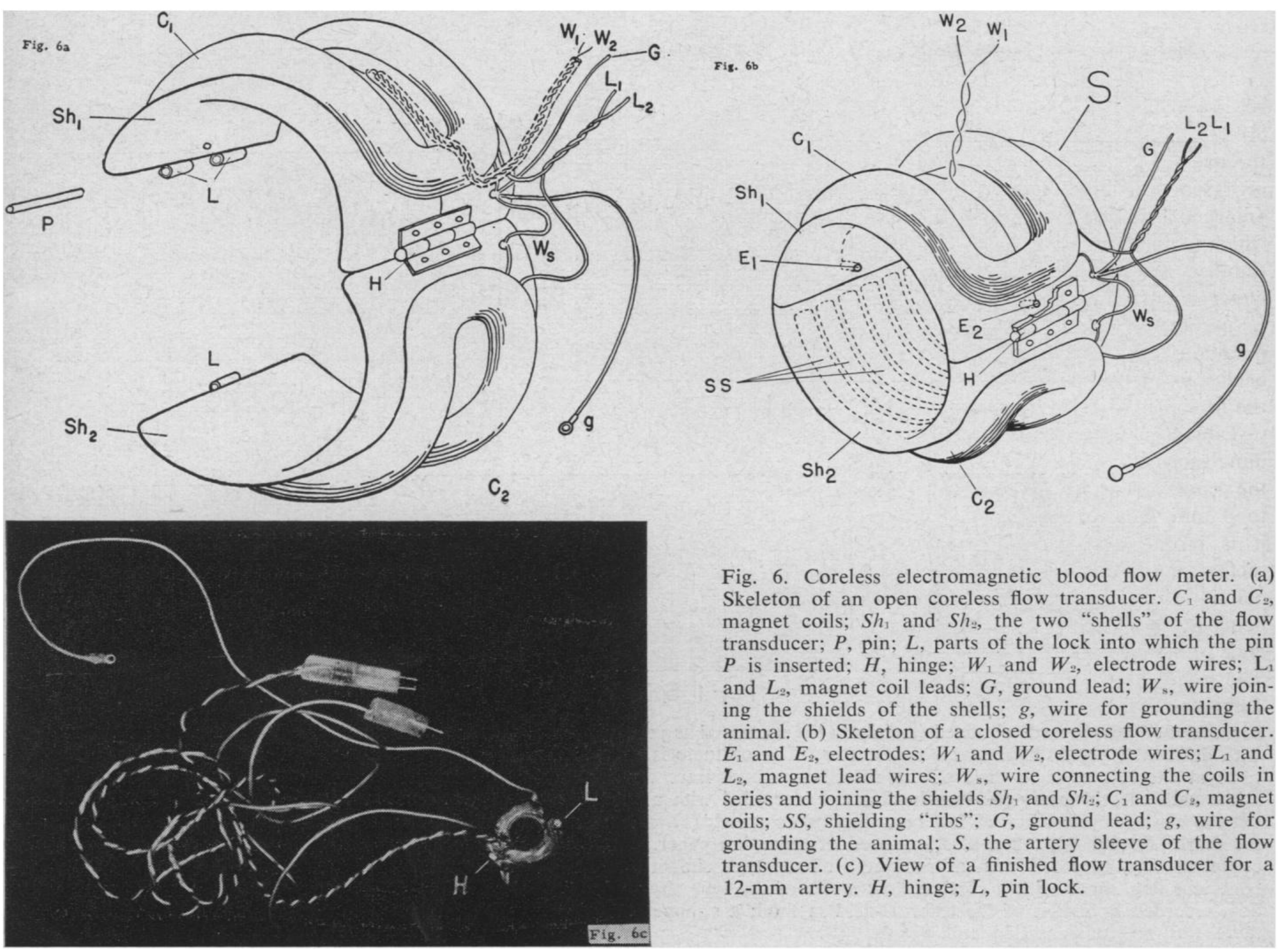


Fig. 6. Coreless electromagnetic blood flow meter. (a) Skeleton of an open coreless flow transducer. C_1 and C_2, magnet coils; Sh_1 and Sh_2, the two "shells" of the flow transducer; P, pin; L, parts of the lock into which the pin P is inserted; H, hinge; W_1 and W_2, electrode wires; L_1 and L_2, magnet coil leads; G, ground lead; W_s, wire joining the shields of the shells; g, wire for grounding the animal. (b) Skeleton of a closed coreless flow transducer. E_1 and E_2, electrodes; W_1 and W_2, electrode wires; L_1 and L_2, magnet lead wires; W_s, wire connecting the coils in series and joining the shields Sh_1 and Sh_2; C_1 and C_2, magnet coils; SS, shielding "ribs"; G, ground lead; g, wire for grounding the animal; S, the artery sleeve of the flow transducer. (c) View of a finished flow transducer for a 12-mm artery. H, hinge; L, pin lock.

ride tubing and the solid plastic. This is achieved by presoaking the ends of the tubing in the monomer of the "Teets denture material" (TDM) from which the plastic body of the flow meter is made for about ½ hour and by applying the polymerizing paste of TDM to the tubing before its surface has dried noticeably. Instead of being presoaked in the monomer, the tubing may be covered with the plastic cement "Dab" (Hollingshead Corp., Camden, N.J.). The best bond is obtained with compression molding of the transducer. The tube becomes very soft at the molding temperature of about 150°C and forms a tight bond with the Lucite which is injected under pressure into the hot mold.

Figure 4d is a photograph of the finished compression-molded flow meter for an artery 3 mm in diameter.

In Fig. 4d, *Sh* is a Teflon shutter which is cast in the plastic material so as to create its own sliding bed. It is removed to admit the artery through the slit into channel *C* and is then reinserted, to prevent the artery from slipping out. A black thread is inserted through a small hole at the end of *Sh* to facilitate handling of the shutter. *W* is a Teflon "wing" which can be inserted into a special gliding bed produced in much the same way that the bed for shutter *Sh* is produced. It is provided with small holes at its ends (marked by black threads) through which it can be sewn onto tissues to stabilize the orientation of the flow meter maintaining the axis of channel *C* parallel to the artery. Failure to maintain this orientation may result in occlusion of the artery by twisting of the flow meter.

Table 1 gives a recommended set of dimensions for flow meters, covering the range of artery diameters from 1 to 4 mm. For diameters above 5 mm it is more advantageous to use flow meters of the coreless design described below. The dimensions are arbitrary, being primarily determined by convenience of standardization of construction. The number of turns of gauge No. 36 Teflon-insulated copper wire is kept constant at 230 turns per coil. A current of 200 ma is used in continuous operation, yielding a magnetic field of 104 gauss, for example, in the case of the 3-mm-channel transducer. The flow signal amplitude V_o, in microvolts, is given by the following expression:

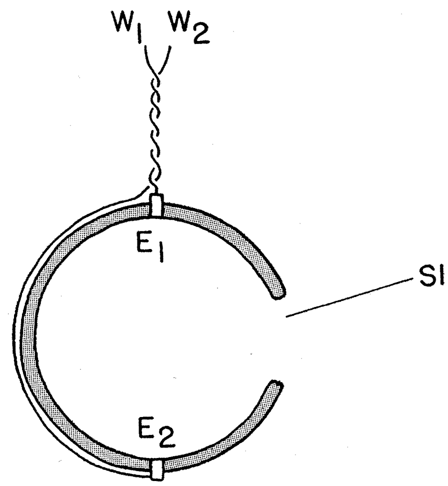
$$V_o=10^{-2}\cdot{}_0Dv \qquad (1)$$

Table 1. A recommended set of dimensions for flow meters of different calibre, covering the range of artery diameters from 1 to 4 millimeters. See Fig. 4a for parts designated by letters in column heads.

d (mm)	*B* (mm)	*l* (mm)	*c* (mm)	*b* (mm)	*i* (mm)	*DD* (mm)	*t* (mm)	No. of turns
1	3	2.5	1	2	4	6.5	0.5	230
1.5	3	3	1.5	2.5	4	6.5	0.5	230
2	3.5	3.5	2	3	4.5	6.5	0.5	230
3	5	4.5	3	4	6	6.5	0.5	230
4	8	5.5	4	5	9	6.5	0.5	230

where B_0 is the magnetic field amplitude, D is the channel diameter, and v is the average blood velocity. For a linear velocity of 30 cm/sec [that is, discharge of $v=(\pi/4)\ D^2 \cong 2.1$ cm³/sec], we thus obtain the flow signal amplitude $V_o=9.4\ \mu v$. This can be easily measured with a phase-sensitive amplifying system (*7, 16*). Figures 5a and 5b show records taken with a unit of this type applied to a cat's carotid artery 1.5 mm in diameter, and Fig. 5c shows a record taken with a unit applied to a dog's carotid artery 3 mm in diameter.

The precision of the alignment of the electrode channels at right angles to channel *C* and wire channel *D* for Fig. 4a within the plane defined by the intersecting axis of the cores I_1 and I_2 and the electrode channel axis is very important to minimize the spurious signal induced at zero flow. When this signal



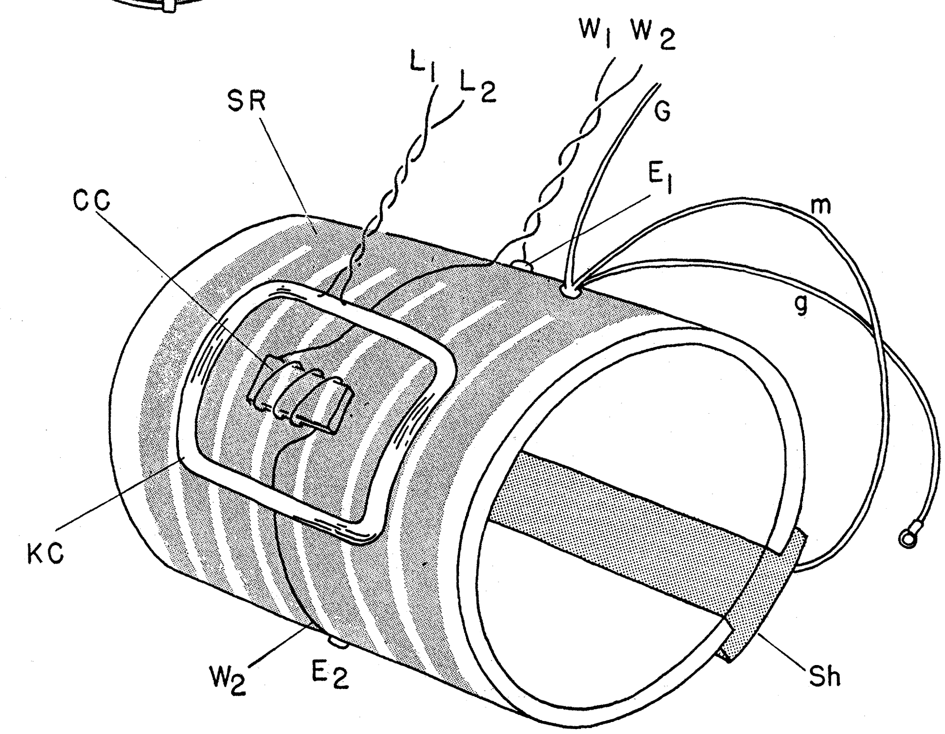


Fig. 7. Sleeve used in conjunction with an external magnet. (Top) Profile of the simplest type of pickup sleeve. *Sl*, slot; E_1 *and* E_2, electrodes; W_1 and W_2, electrode wires. (Bottom) Skeleton of pickup sleeve with auxiliary coils. *Sh*, shutter; *CC*, compensating coil; *KC*, calibrating coil; *SR*, shielding "ribs"; E_1 and E_2, electrodes; W_1 and W_2, electrode lead wires; *m*, wire joining the shields of sleeve and shutter; *G*, ground lead; *g*, wire for grounding the animal.

is small enough, it is ignored by the phase-sensitive recording system, and the base line for zero flow can be ascertained by switching off the magnet current.

Coreless Flow Meter

The large blood flow in the major arteries greatly simplifies the task of measuring flow in them. For example, a magnetic field of only 10 gauss is needed to obtain a 10-mv flow signal from an artery 2 cm in diameter carrying a flow of average linear velocity 50 cm/sec. This signal can be very easily recorded with a phase-sensitive amplifier (*7*). The implanted magnet can thus merely consist of air-core coils of relatively few turns. This greatly reduces the size of the implant, as may be seen from Figs. 3c, 3d, and 6a. The size of the air core magnet relative to the diameter of the artery is very much less than that of a subminiature iron core unit for the same blood vessel (compare Fig. 3d). Figure 6a shows the skeleton of such a flow transducer. C_1 and C_2 are the two coils generating the magnetic field. They are bent to conform to the shape of the cylindrical sleeve *S* (see Fig. 6b) and are cemented in place with TDM. They are connected in series by the wire W_s, which is grounded through the lead *G*. The current is supplied to the coils through the leads L_1 and L_2.

For an artery 15 mm in diameter, for instance, two coils of 50 turns each of gauge No. 29 Teflon insulated wire are used. A current of 1 amp, passed at continuous operation, yields a mean field of 28 gauss. For arteries of different sizes the same number of turns (50 turns per coil) is maintained, and the same current is used. It is desirable to increase the field intensity as the diameter of the artery decreases.

Wires W_1 and W_2 (gauge No. 39 wires with heavy Teflon insulation) are connected to electrodes E_1 and E_2 (see Fig. 6b), which lie in a plane perpendicular to the axis of the cylindrical sleeve channel. The wires are twisted to minimize pickup due to electromagnetic induction. Both electrodes and one coil (C_1) are fixed to one "shell," Sh_1, of the unit. The other, somewhat smaller, shell (Sh_2) carries the second coil C_2. The two shells are connected by a hinge *H*. Thus the unit can be opened to insert an artery and locked by the pin *P* shown in Fig. 6a after closing (Fig. 6b).

The sleeve formed by closing the shells is shielded by a laminated, riblike, silver shield, *SS*, which is grounded through connection to W_s. This shield is produced by painting the sleeve, prior to application of the coils, with silver paint (see above), scratching the silver paint to produce the riblike laminations (shown in Fig. 6b only on one shell), and silver-plating the "ribs" thus obtained. The shields of the two shells are grounded and joined by wire W_s. It is important not to cover too much of the surface with silver, in order to retain enough Lucite area for good adherence of the TDM, with which the unit is finally covered. The wires are shielded with a wire braid and insulated with polyvinyl chloride tubing in a way similar to that described above for the subminiature units. The lead wires are bent toward the sleeve and buried in TDM (after presoaking in TDM monomer). Wire *g* is connected to *G* and serves to ground the animal. Its bare end consists of a small gold ring, which is sewn onto an animal tissue that is a good conductor.

Figure 6c shows a 12-mm flow meter of this type. Before the hinge is attached by TDM to the two parts of the transducer, it is filled with a loose cotton plug soaked in normal saline. The two "shells" are then moved axially relative to each other until the "transformer electromotive force" recorded by the amplifying system has been reduced to zero. With this configuration fixed, hinge *H* and the lock *L* are attached to the "shells" by means of

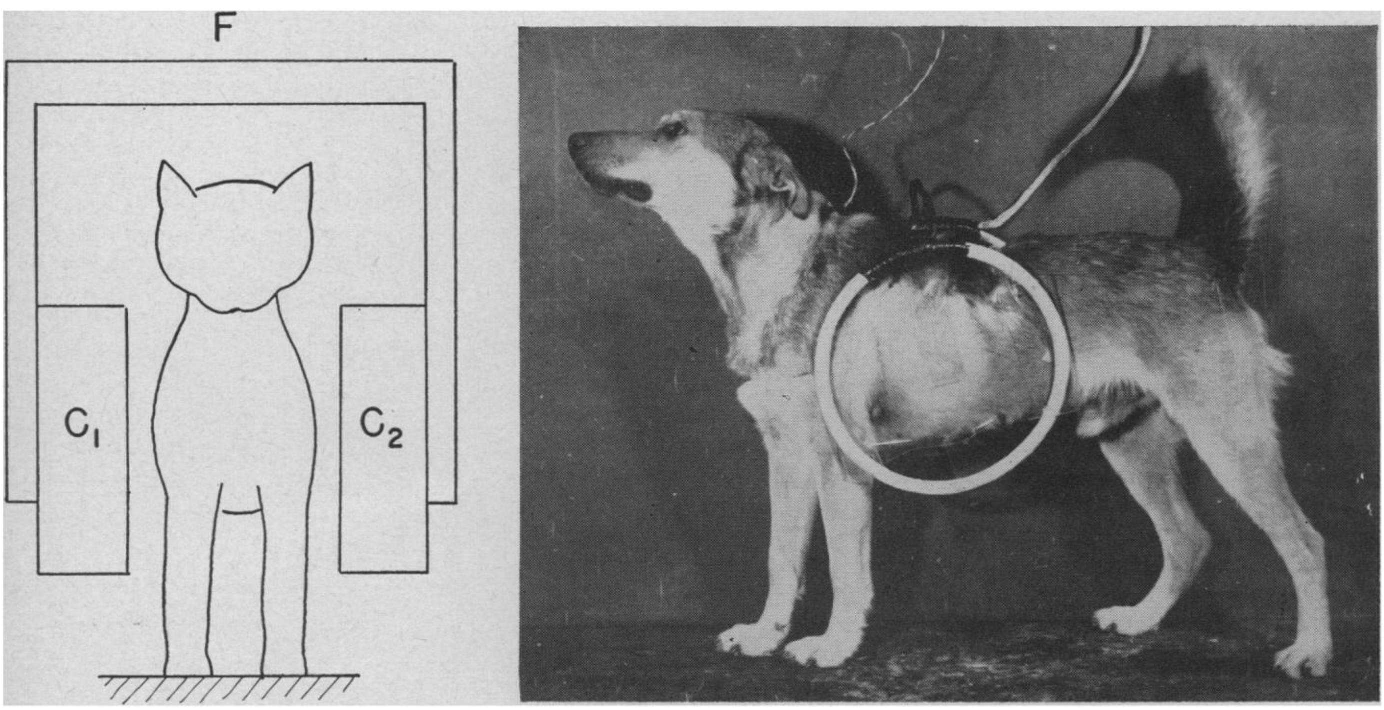


Fig. 8. Uses of the external-magnet method. (Left) Cat between two coils generating a magnetic field, illustrating the external-magnet method based on use of a stationary magnet. (Right) Dog with strapped-on coils, illustrating the external-magnet method based on a magnet attached to the animal.

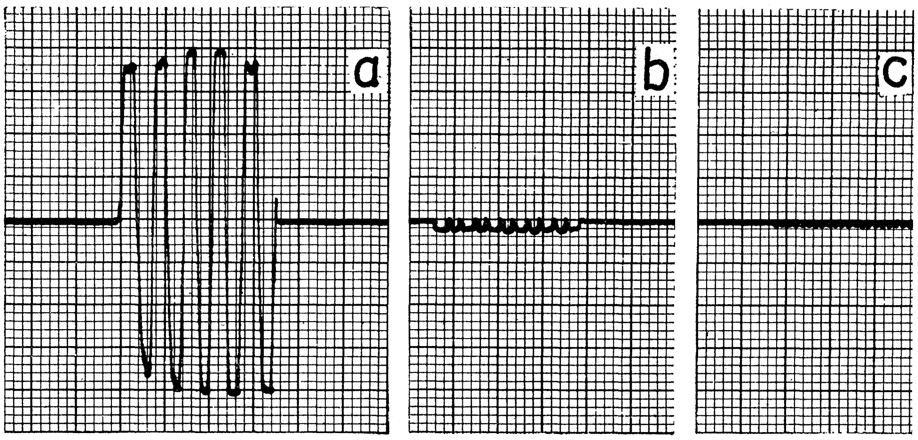


Fig. 9. Records showing the effectiveness of eliminating the transformer electromotive force in the external-magnet method by the use of a compensating coil and by means of a phase-sensitive detector (see text).

TDM. This assures a zero reading at zero flow.

Figure 5d shows a recording of cardiac output taken with a unit of this type applied to a dog's ascending aorta (10 mm in diameter).

External Magnet Method

As can be seen from Fig. 7 (top), the implant required for the external-field method is very much simpler and smaller than the flow transducers described in the preceding section. The principal part of the implant is the pair of electrodes E_1 and E_2, which are embedded in the thin Lucite sleeve. Wire W_2 connected to E_2 runs through a groove in the central plane of the sleeve which passes through the electrode axis E_1E_2. *Sl* is the slit through which the artery is inserted. *Sh* (Fig. 7, bottom) is a shutter which closes *Sl* to prevent the artery from slipping out. Wires W_1 and W_2 are shielded and insulated, as in previous designs. The magnetic field is established by an external magnet, as shown in Fig. 8 (left). For large gaps, such as the one shown in this figure, the gain in field strength through the use of an iron core becomes relatively small, and it is more advantageous to use a somewhat weaker but lighter air-core magnet. A pair of Helmholtz coils 30 cm in radius, of 300 turns per coil, carrying a current of 10 amp, will produce a field of about 100 gauss at the center between the coils. This field is 10 times as strong as the required field for measurement of flow in the major arteries, such as the aorta, and of adequate strength for flow measurements in smaller arteries of the order of magnitude of 1 mm in diameter. Due to the penetration of a large portion of the animal's body by the magnetic field, sleeves can be implanted about many relatively widely dispersed arteries for simultaneous measurements. In cases where the implants are far apart, consecutive measurements are made possible by moving the desired region of the animal's body into the magnetic field.

The stationary coil of Fig. 8 (left) requires immobilization of the animal during the recording of blood flow. This can be avoided by making the coil sufficiently light to be attached to the animal, as shown in Fig. 8 (right). This approach combines the advantage of freedom of motion achieved by the method described in the previous section with the advantages of the smallness and simplicity of the implant shown in Figs. 3c and 3b. The disadvantage lies in the necessity of strapping a coil of about 10 percent of the animal's weight onto the animal.

The main experimental difficulty encountered in this approach lies in the need to eliminate the flow-independent transformer electromotive force induced in the electrode lead wire loops. In principle, this can be done by positioning the animal in the magnetic field so as to reduce to zero the field component parallel to the sleeve axis. In practice, due to the animal's unavoidable movements, this ideal condition cannot be maintained. This problem can be solved, however, by use of a compensating coil (*4*) in series with one axis parallel to the sleeve axis and of the electrode leads, placed with its wound so as to introduce into the electrode circuit an electromotive force of opposite phase and equal magnitude to the transformer electromotive force. Figure 9 shows the effectiveness of this device. In Fig. 9a may be seen a record of the fluctuating signal delivered in the absence of a compensating coil by the electrode leads to the amplifier, due to angular oscillation of the sleeve over an angle of 180°. Figure 9b shows

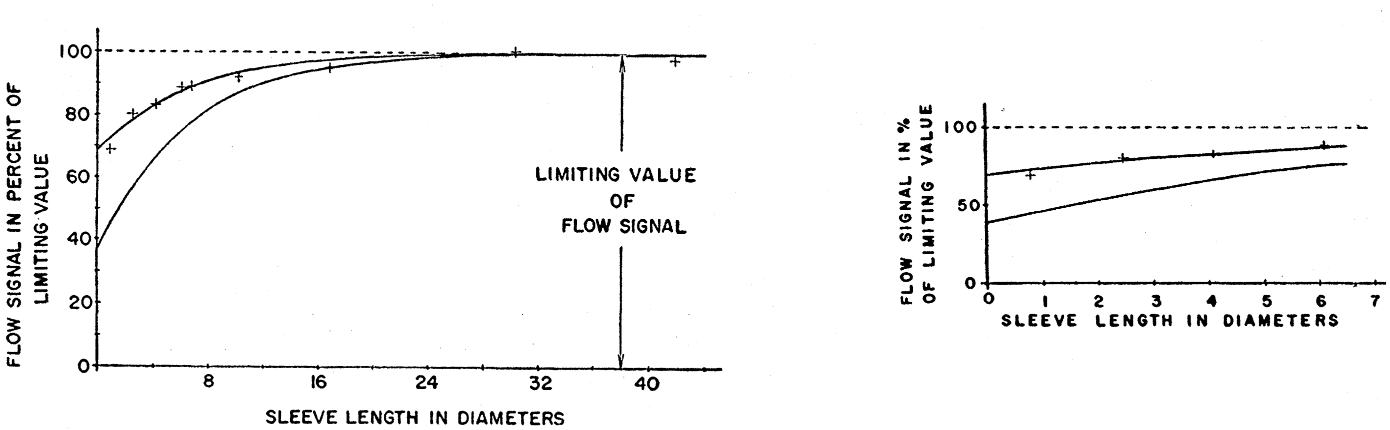


Fig. 10. (Left) Flow signal (in percentage of limiting value) (ordinate) as a function of sleeve length (abscissa). (Right) Enlarged-scale plot of the neighborhood of the origin.

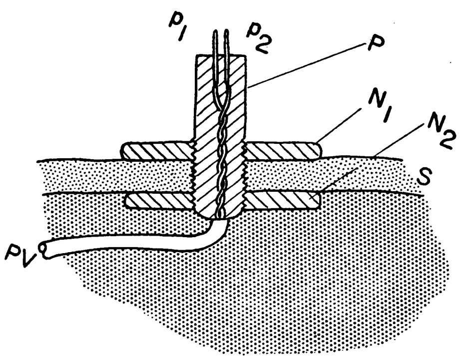


Fig. 11. Method of conveying the flow-meter leads through the animal's skin. *P*, threaded plug (cast from TDM); P_1 and P_2, contact pins; N_1 and N_2, nylon nuts; *S*, skin of the animal; *PV*, polyvinyl chloride tubing containing the lead wires.

a record obtained under the same condition with the compensating coil in the circuit. These records were obtained by deliberately phasing the phase-sensitive detector so as to reject the flow signal and to select the quadrature signal. By phasing the sensing circuit properly for rejection of the quadrature signal, the disturbance can be almost completely eliminated, as shown in Fig. 9c.

Figure 7 (bottom) shows the modification of the sleeve design to include the compensating coil *CC*. This coil is obtained by winding the appropriate number of turns of the lead wire W_2 about a coil form. W_2 is then twisted with W_1 into a single double lead W_1W_2. Coil *CC* is insulated by a layer of TDM, and the outside of the sleeve and of the shutter are gold-plated, as described in the previous section, and their gold shields are connected by wire *m* to a ground wire *G*. The bare gold-plated end of *G* is sutured to a conductive tissue to ground the animal.

A change in the orientation of the flow-meter sleeve introduces a change in the sensitivity of the flow meter. Component *B* of the magnetic field effective in inducing the flow signal is given by $B = B_0 \cos \theta$, where B_0 is the stationary magnetic field originally aligned to give a maximum flow signal and θ is the angular deviation of the sleeve from the orientation of maximum flow signal. Thus, a misalignment of the sleeve over a range of 30° (± 15°) produces an error of only about 4 percent.

It may not always be possible to orient the animal with respect to the magnetic field so as to obtain a maximum flow signal from a given sleeve implant. Calibration is then necessary, to ascertain the magnitude of the magnetic field component effective in inducing the flow signal. This can be accomplished by means of a "calibrating coil" *KC* (*4*), which is shown in Fig. 7 (bottom) and which is disposed so that its area is perpendicular to the magnetic field when the sleeve is in the orientation yielding a maximum flow signal. The terminals of this coil form the twisted lead L_1L_2. The electromotive force induced in this coil is determined once and for all for optimum orientation in a given external magnetic field. When, in a practical situation, we obtain a signal from the coil *KC* from a given site of implantation which is a fraction of this maximum signal, we know that the flow signal observed represents the same fraction of the maximum signal observable at optimum orientation. It is more practical to follow the practice of adjusting the sensitivity of the amplifier so as to obtain a standard reading for the signal derived from the coil *KC*. This will automatically maintain the sensitivity of the flow meter at a constant standard value. Figure 5c shows a record taken of the blood flow in the ascending aorta of a conscious dog with an external attached pair of coils [see Fig. 8 (right)] developing a field of 10 gauss at its center. This record was taken under the most unfavorable conditions—namely, in a moving artery without the use of a compensating coil, at a low magnetic field intensity.

The signal derived from a flow-meter sleeve depends not only on its diameter but also on its length. This is due to the fact that the immobile conductive medium at the two ends of the sleeve represents a load to which the moving blood in the sleeve, acting as a generator, supplies electric power. The current thus engendered gives rise to a drop in the voltage recorded between the electrodes, which becomes more pronounced as the sleeve is shortened. This is illustrated in Fig. 10. The graph at the left was obtained by measuring the flow signal for constant flow with a long flow-meter pipe placed into the magnetic field of a Helmholtz coil. One end of the pipe was connected to a source of saline. The entire pipe was submerged in saline, and its free end was cut shorter periodically until it came to lie next to the electrodes. We see that there is a progressive reduction in sensitivity. The lower curve has been drawn by inference for a sleeve shortened at both ends by doubling the drop in sensitivity below the optimal value observed in the upper curve.

This effect must be taken into account in calibrating a sleeve. If the calibration is accomplished by inserting a pipe into one end of the sleeve, a second calibration should be performed by putting a long pipe (about 15 diameters in length) into the sleeve exit end. The discrepancy between the two readings is to be doubled and subtracted from the reading obtained with the exit pipe in use. This will give a reading corresponding to a sleeve without pipe

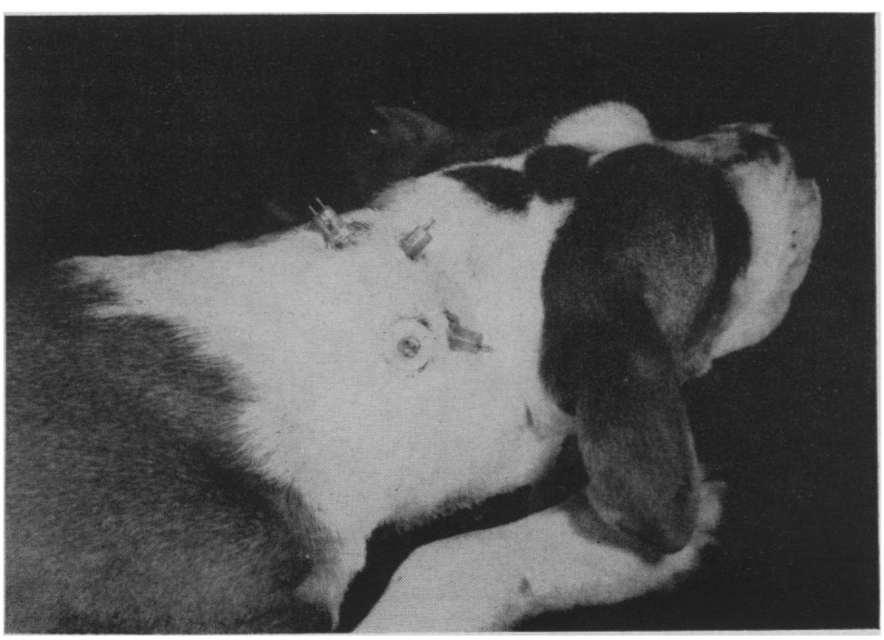


Fig. 12. View of a dog's neck, showing four plugs connected to implanted flow meters.

inserts at either end. It is a more reliable procedure, however, to calibrate the sleeve by perfusing with blood an excised artery placed into the sleeve and submerged under blood.

Connections to the Amplifier and Oscillator

The electrodes and the magnet coils are connected to instruments which are described elsewhere (*6, 7*). The connections are established by suitable plugs. The lead wires are guided from the site of implantation under the animal's skin and are permitted to issue from a small opening in the skin at the nape of the neck. It is very convenient to be able to attach the plugs to the animal's skin at the back of the neck. For this purpose, the outside of the plug is equipped with a thread (cast from TDM), and nylon nuts are used which can be screwed onto this thread. The polyvinyl tubing is bonded well to the Lucite body of the plug to make prolonged submersion in a conductive liquid possible. The entire flow meter, including leads and plugs, is sterilized by submersion in a solution of Zephiran.

Figure 11 shows the manner in which this plug is applied to an animal's skin. A round hole is cut into the skin with a sterile cork borer, and the plug and washers are applied as shown. Figure 12 shows outlets in the neck of a dog, associated with two flow-meter implants. The animals show no discomfort with this method of fixation of outlets. A 10-foot cord connected to the instruments gives the animal ample freedom of motion.

Although the work has been limited so far almost exclusively to animal experimentation, the methods described in this article are suitable for application to human subjects (*17*).

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15. I was greatly assisted in perfecting this type of flow meter by Harry S. Tillson of the Medical School Research and Development Shop. Many important improvements in the manufacturing procedure and design, including the method of molding these units, were suggested and worked out by him. In the latter process the advice and cooperation of Chester Chalberg were very helpful.

16. C. W. Barnes, in preparation.

17. I wish to express my gratitude for the generous support of this work by the Office of Naval Research. The coreless flow meters were tested in the course of cooperative research involving chest surgery with different groups. In this connection, I gratefully acknowledge my indebtedness to Drs. Paul H. Crandall, Peter Gall, James V. Maloney, and Lawrence W. Roth. Thanks are also due to Dr. Bennett J. Cohen for his helpful advice. The aid of the Research and Development Shop of the Medical School was of decisive importance, and I wish to thank Chester Chalberg for his never-failing cooperation and Harry Tillson for his valuable contributions. Last but not least, the tireless assistance of Clarence W. Barnes and Raymond T. Kado in the performance of the experiments and in the preparation of the illustrations is gratefully acknowledged.

Science in the News

Diplomats Ponder Antarctica's Future as Scientific Studies Are Resumed

Representatives of the 12 countries that participated in the antarctic program of the IGY are now in conference in Washington to negotiate a treaty to assure the continued peaceful use of the continent. The call for the conference was made in May 1958 by President Eisenhower when, citing the "splendid example of international cooperation" afforded by the IGY, he sent invitations to the foreign ministers of the 11 nations that had a role in the IGY antarctic program. Among the delegates attending the meeting, which opened 15 October, are First Deputy Minister Vasily Kuznetsov of the Soviet Union and Prime Minister Walter Nash of New Zealand. At the first meeting of the conference Herman Phleger, head of the U.S. delegation, was elected permanent chairman. The conference may last several weeks.

As the diplomats begin their work, scientists from the United States and other countries, with the onset of the summer season, are resuming the intensive continental studies which began on a large scale with the International Geophysical Year in 1957–58. Approximately 70 scientists are now converging at the half dozen U.S. bases in Antarctica to conduct 29 new projects recently announced by the National Science Foundation and to continue others that were begun under earlier NSF grants. Studies of cosmic rays, geomagnetism, aurora and airglow, and glaciology are among the projects, as well as three mapping programs to be conducted by the American Geographical Society and the U.S. Geological Survey.

Scientific Studies Pushed

In announcing the NSF Antarctic Research Program, which the foundation supports with more than $3 million in grants, Alan T. Waterman, director, said: "Greater emphasis will be placed by the United States on geology, cartography, and biology during the next year." The three cartographic projects call for topographic mapping in the area, revision of maps, and the development of an atlas which will include data on snow accumulation, magnetic variation, and the thickness of the ice sheet. Among the geological studies are projects to determine how long low temperatures have existed on the continent by measurement of the thermolumines-