A NEW PRINCIPLE FOR ELECTROMAGNETIC CATHETER FLOW METERS*

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Abstract.—An electromagnetic catheter flow meter is described in which the magnetic field is generated by two parallel bundles of wire carrying equal currents in opposite directions. The electrodes are fixed centrally to the insulated wire bundles that generate the magnetic field. The flow sensor is flexible, resembling a split catheter. The flow transducer is designed to constrict as it is introduced through a branch artery and to expand in the main artery over the span of its diameter. The principle is suitable for branch flow measurement as well as for measurement of flow in a major artery or vein by the same transducer. A special method of guiding the electrode wires results in a zero base line at zero flow for the entire range of diameters accommodating the field generating coil. The electrodes could be used in this configuration with a magnetic field generated by coils external to the patient for blood flow measurements with a catheter of reduced gauge. The transducer can be made smaller in circumference than those employed in other electromagnetic flow measuring catheter devices. This feature is of special value for envisaged clinical uses (percutaneous introduction) to minimize surgical intervention.

The velocity sensitivity of the flow transducer is a logarithmic function of the tube diameter. The flow throughout the entire tube cross section contributes to the flow signal. It is sufficient to calibrate the transducer by one measurement in a dielectric conduit of less than maximum diameter. The sensitivity at other diameters follows from a logarithmic plot. The diameter of the blood vessel is outlined by the transducer in radiograms, thus obviating the need for radiopaque materials. The principle was demonstrated by measurements *in vitro*. Experiments *in vivo*, derivation of equations, and construction details will be published elsewhere.

Introduction.—The volume rate of blood flow can be determined quantitatively by an electromagnetic flow meter applied externally to a blood vessel.[1] Such an application requires surgical exposure of the blood vessel and often involves extensive surgery. This is clearly undesirable, if routine use is to be made of this method for clinical measurement of blood flow in patients. For this purpose, electromagnetic catheter flow meters have been developed consisting of an electromagnetic flow transducer incorporated in a thin flexible tube (catheter) that can be inserted through a branch vessel (e.g., femoral artery or jugular vein), from which it is maneuvered into a major artery (aorta or pulmonary artery) or vein (vena cava).[2–8] Such measuring devices are still rather large, typically 3 to 4.5 mm in diameter, and require a larger opening in the blood vessel than would normally be considered safe. Instruments of this size do not lend themselves to introduction through a small opening in the skin.

The purpose of this communication is to describe a radically different approach that solves the problem of excessive dimensions and remedies some other weaknesses of electromagnetic catheters in current use. Except for the branch flow probe[6, 7] the currently available electromagnetic catheters are intended to measure the local velocity within a conduit at the location of the flow sensor. They are not centered within the blood vessel, and the volume rate of flow is calculated from the velocity measurement on the assumption of a uniform velocity throughout the blood vessel cross section.[2–5] The sensor of the present electromagnetic catheter flow meter is affected by flow throughout the entire conduit cross section and centers itself automatically to fit blood vessels of a wide range of diameters. The most practical property of the new electromagnetic catheter flow meter is its collapsibility to a transverse dimension of about 2.7 mm, which is required for percutaneous introduction. Another advantage is the absence of a rigid transducer.

A particularly interesting and helpful property of the new electromagnetic catheter flow meter is that it remains effective in conduits of different lumens, in which it displays a logarithmic relation between sensitivity and lumen diameter. It can be used to measure branch flow as well as the flow in the main conduit. A new scheme of guiding electrode leads permits achievement of a zero base line error which remains negligible as the sensor dimensions expand and contract to fit conduits of different lumens. The same basic scheme is applicable to pick-up electrodes in a system in which the magnetic field generator is external to the patient.

The flow transducer: Figure 1 shows the scheme of the transducer in which a noncircular flat coil generates a magnetic field that intersects the plane of the coil at right angles. The coil consists of N turns, of which only two turns are

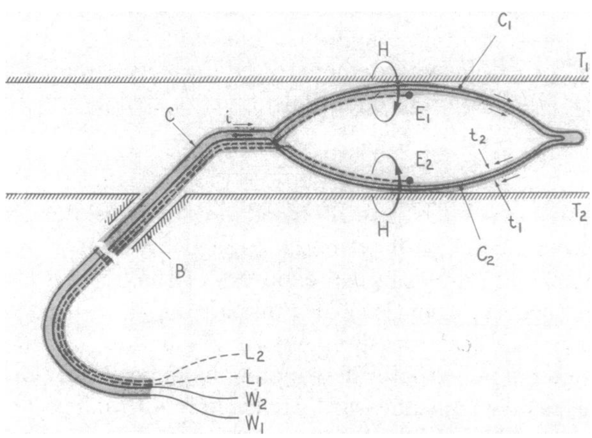

FIG. 1.—Scheme of electromagnetic catheter flow meter. T_1, T_2: walls of the artery. B: artery branch through which the catheter is introduced. C: catheter section harboring the coil lead wires W_1, W_2 and electrode lead wires L_1, L_2. C_1, C_2: legs of the coil generating the magnetic field **H**. Arrows indicate the direction of the current i in the lead wires W_1, W_2, as well as the two turns t_1, t_2 of the coil which are shown.

shown. We can imagine this coil produced from two originally adjacent parallel leads C_1 and C_2 that have been pulled apart to produce the lenticular coil cross section shown in Figure 1. To ensure that the leads maintain the shape shown, a frame of this shape made of an elastic steel wire (e.g., gauge #31 hypodermic tubing or a suitable piano wire) is incorporated into the bundle of wires. The top and bottom strands C_1 and C_2 can be pressed toward each other so as to pass through a narrow opening as parallel strands that spring open again after emergence into a wider space. The two elastic steel wire braces are soldered to each other at each end and to a larger steel tubing (e.g., gauge #18 hypodermic tubing) at the left. This tubing is about 50 cm long and serves as a "back bone" for the catheter (it is not shown in the figure).

The wire bundles C_1 and C_2 are actually threaded through two fine "leggings" of silicone rubber tubing (Dow Corning "Silastic" medical grade 0.025" i.d., 0.047" o.d.) which extend between the hairpin bends in the coil wires on the right and left side. An alternative method of insulation is to paint the wires with General Electric RTV 112 silastic cement. A wider silicone rubber tube that encloses the coil lead wires W_1 and W_2 and electrode lead wires L_1, L_2 forms an unsplit section C of the catheter. All the silicone rubber tubes are cemented to each other by General Electric RTV 112 silastic cement, so that all the coil wires are thoroughly sealed and electrically insulated from the liquid when submerged. The right tip of the coil is encapsulated in RTV 112 silastic cement. A #34 gauge formvar insulated wire has been used to wind the coil of 10 turns.

E_1 and E_2 are electrodes (Pt strips soldered onto fine PVC insulated Cu wires, indicated by dashed lines).

A current (0.5–1 amp) passing through the coil generates a magnetic field, illustrated by two typical field lines (H) encircling the legs of the coil.

Principle of operation: To explain the principle of operation in idealized form, let us imagine the coil to be so long that we can consider the wires in the middle section near the electrodes as portions of two infinitely long parallel wires W_1, W_2 of negligible diameter carrying currents of equal magnitude in opposite directions. Figure 2 shows the magnetic field lines in a plane passing centrally through the electrodes (E_1, E_2) perpendicular to the wires (W_1, W_2). The field lines are not shown completely as closed rings; only the portion of the magnetic field permeating the conduit C is shown.

The magnetic field is not uniform. Its intensity increases from the conduit's central axis toward the electrodes and decreases as we recede from the artery axis along a line perpendicular to the electrode axis E_1-E_2. In practical iron core electromagnets, one also encounters a nonuniformity of the magnetic field that increases from the pipe center toward the magnet pole pieces and decreases from the center outward along the transverse diameter. This type of magnetic field nonuniformity does not preclude satisfactory linear performance of the electromagnetic flow meter over the laminar and turbulent flow regimes,[9, 10] and even greater field nonuniformities can be tolerated without loss of linearity over a wide flow range covering the laminar and turbulent regimes.[11] In the case of pulsating arterial blood flow one actually need not be concerned with transducer performance in the laminar flow regime, since the velocity is practically uniform

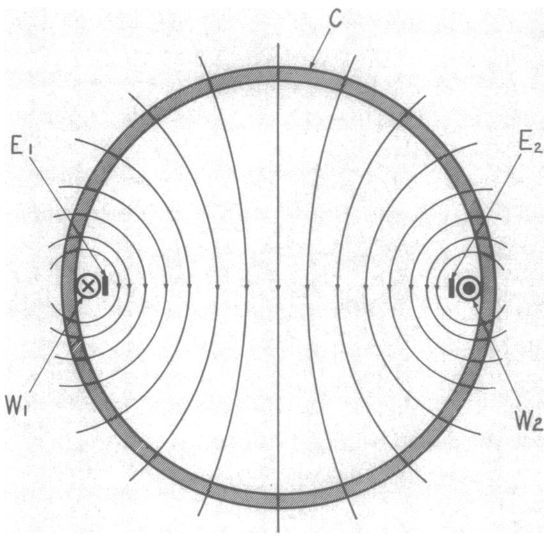

FIG. 2.—Cross section of artery C traversed by magnetic field (arrows). E_1, E_2: electrodes. W_1, W_2: cross sections of coil wires carrying the currents that generate the magnetic field. (The cross indicates direction away from the reader; the dot, direction toward the reader.)

throughout the artery cross section.[3, 8, 12] Our consideration will be limited therefore to the practically important case of turbulent flow across the non-uniform magnetic field of the flow transducer.

We assume a uniform flow of blood to traverse the cross section of conduit C of Figure 2 at right angles to the plane of the drawing. As a result, an electromotive force will be induced in the conductive fluid that will be detected by a potential difference between the electrodes E_1, E_2. This electromotive force can be calculated from the expression for the magnetic field B produced by a straight circular bundle of current carrying wires

$$B = \frac{B_0}{r} \tag{1}$$

(where B_0 is the field at unit distance from the center of the bundle) and the flow meter equation

$$dV = kBvdr \tag{2}$$

where dV is a voltage increment across the radial distance dr, v the local velocity of flow, and k a constant depending on the choice of units. Integration of equation (2) (using the B value of equation (1)) yields a logarithmic expression, the derivation of which will be presented elsewhere. The measured sensitivity ($S = V/v$) of the transducer is represented to a good approximation for large D/d values by a logarithmic function of the ratio of the diameter D of the conduit C of Figure 2 and the diameter d of the wire bundles W_1 and W_2:

$$S = c \ln \frac{D}{2d} \tag{3}$$

where c is a constant depending on the choice of units and intensity of the field-generating current. The sensitivity drops to zero when the electrodes touch each other at $D = 2d$. Figure 3 shows a plot of recorded flow amplitudes as a func-

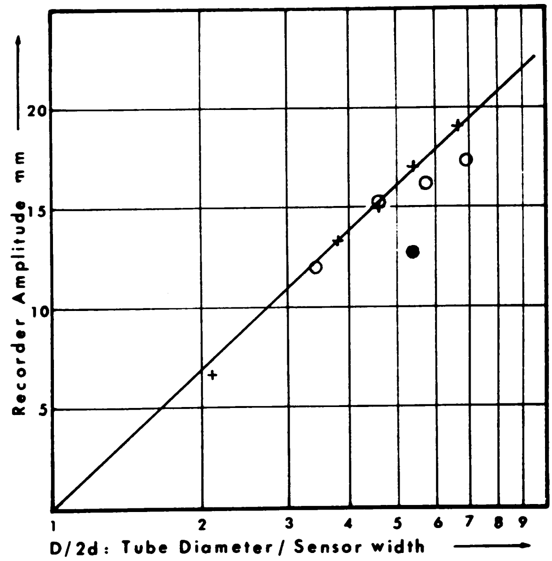

FIG. 3.—Dependence of transducer sensitivity on conduit (artery) diameter. The points shown were obtained by oscillatory movement of a transducer in a 0.9% saline solution.

+: the transducer is enclosed in lucite tubes of different diameters (16.0 mm, 13.0 mm, 11.0 mm 9.1 mm, 5.0 mm).

O: the dimensions of the transducer are altered by a thread tied around it.

●: the transducer is inside a 13.0 mm aluminum tube.

tion of log $D/2d$. The sensitivity of the transducer was measured by oscillating it by an electromechanical device at a uniform rate. D was varied by slipping thin lucite tubes over the transducer to constrict it (points indicated by crosses in the plot). An alternative method of changing the interelectrode distance was to constrict the transducer by wrapping a thin nylon thread once around it (circles). There is no striking change in sensitivity produced by the dielectric tube enclosure.

Because of the flexibility and deformability of the flow transducer, which is essentially a split section of a catheter, we should be able to maneuver it into side branches in accordance with standard angiographic practice and thus be able to measure blood flow through organs supplied by the chosen branch.

We need to know the artery diameter to ascertain the sensitivity of the transducer from a calibration curve and to obtain the rate of volume flow as a product of the measured average velocity v with the cross-sectional area of the blood vessel. This transducer possesses the advantage that it is not necessary to inject radiopaque materials to ascertain the artery diameter. The two insulated wires (or wire bundles) W are in contact with the artery wall. Thus a transverse X-ray exposure will yield a radiogram from which the artery diameter can be easily determined. Our catheter flow meter thus fulfills the collateral function of an arterial diameter gauge.

Calibration: Since the straight line of Figure 3 passes through the origin, it is sufficient to calibrate the transducer at one arbitrary tube diameter. Since a radiogram of the transducer *in situ* gives the artery diameter, the calibration for a given blood vessel follows from a graph like Figure 3.

One important question must be answered in this connection. Does the calibration depend on the electrical conductivity of the conduit? If in the experiments illustrated in Figure 3 we replace a lucite tube by an aluminum tube of equal diameter, there is only a moderate drop in sensitivity, as shown by the black circle in Figure 3. This conductivity change in conduit material from lucite to aluminum represents a practically infinite ratio. In a conventional electromagnetic flow meter, the flow signal would have been reduced to zero. We can safely conclude from this result that the artery wall conductivity will have no significant effect upon the transducer sensitivity.

The absence of an appreciable short-circuiting effect of a conductive tube wall in our catheter device is due to the fact that the particular magnetic field distribution of this kind of flow transducer is derived from two substantially axis-parallel current-carrying wires. It is known that for certain types of magnetic fields the conductivity and thickness of the conduit wall have no effect upon the induced flow signal. This is, for instance, the case when the flow of induced currents is suppressed by using a magnetic field generated by a straight wire that runs along the axis of a circular cylindrical conduit.[9]

Zero flow base line: Another point of importance is the determination of the base-line corresponding to zero flow. If we imagine a line drawn between the electrodes E_1, E_2 in Figure 1 to represent the equivalent resistance of the electrolyte that establishes electrical continuity between the electrodes, we see that the loop thus created in the electrode lead circuit can be considered as the secondary of a transformer whose primary is the coil carrying the a-c supplied by the wires W_1, W_2. The magnetic field of the primary coil is quite nonuniform, and its intensity varies as the distance between the coil legs C_1 and C_2 is varied. Thus, a pulsating artery, by changing the distance between C_1 and C_2, would give rise to a modulated signal of carrier frequency (unrelated to flow) induced in the secondary loop. The signal ("transformer emf") thus induced in the absence, as well as the presence, of flow in the electrode lead circuit can be as much as two orders of magnitude higher than the anticipated flow signal. Its elimination is essential to ensure the practicability of this type of catheter flow meter.

Figure 4 illustrates how the elimination of the transformer emf was achieved. Instead of the electrode lead wires being guided all along the coil legs C_1 and C_2 (as shown in Fig. 1), the lead wires are crossed at a point P before reaching the electrodes. We have thus subdivided the secondary loop of the electrode leads into two sub-loops A and B which are wound in opposition to each other; i.e., a current circling the loop A in the clockwise sense flows counterclockwise around loop B. By shifting the location of the crossover point P of the silicone rubber insulated thin flexible wires, we can adjust the relative areas of the loops A and B. As we do this, the transformer electromotive force induced in the secondary circuit diminishes and can be made to vanish, as is shown in Figure 5. At this point, the wires are fixed by cement to stabilize the position of P. In Figure 5,

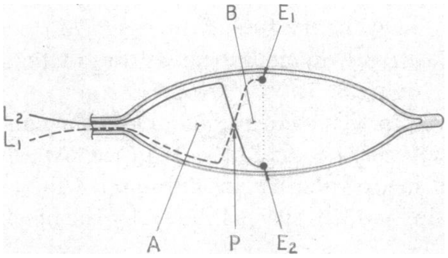

FIG. 4—Scheme for suppression of transformer electromotive force so as to obtain a zero reading at zero flow independently of tube diameter. L_1, L_2: electrode leads crossing at point P. E_1, E_2: electrodes. Relative dimensions of loops A and B are adjusted by moving location of P to obtain a zero reading when instrument is turned on.

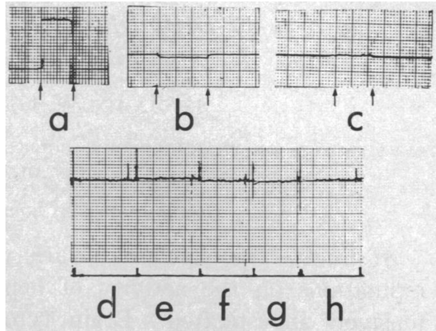

FIG. 5.—Effectiveness of base line adjustment method illustrated in Figure 4. The transducer is stationary in a beaker filled with 0.9% NaCl solution. (*a*) Reading at zero flow when instrument is turned on (between arrows). (*b*) Reading after adjustment of loops A and B. Slight overcompensation leads to downward excursion between arrows. (c) Final step. There is no appreciable base line shift when instrument is turned on (between arrows).

For the following records the transducer was transferred from a beaker into tubes of different diameters filled with the same solution as above: (*d*) beaker; (*e*) 16.0 mm i.d.; (*f*) 11.0 mm i.d.; (*g*) 7.0 mm i.d.; (*h*) beaker.

the initial transformer electromotive force was off scale. Figures 5*a-c* show three subsequent stages in the adjustment for zero base line while the transducer is kept in a saline filled beaker in a stationary fluid. Figures 5*d-h* demonstrate that this adjustment has also rendered the transducer insensitive to the distance between the coil legs C_1 and C_2 (compare Figure 1). We see no appreciable variation in the base line as the transducer is transferred from an unrestricted environment (*d*) (beaker) to a 16.0 mm i.d. tube (*e*), to an 11.0 mm i.d. tube (*f*), to a 7.0 mm i.d. tube (*g*), and back to the beaker (*h*). Pulsatile variation in the distance between the legs C_1, C_2 has consequently no effect upon the signal derived from the electrodes. These tests were performed at an instrument sensitivity capable of detecting flows in the order of 1 cm/sec.

It is worthwhile to mention that the same principle of elimination of the transformer electromotive force could be used even if the legs C_1, C_2 contained no coil for generation of the magnetic field. In this case, the magnetic field could be produced by a large coil located externally to the catheter near the animal or patient. The advantage of this scheme would lie in the possibility of greatly diminishing the thickness of the catheter by eliminating the magnet coil from its structure, thus greatly facilitating its introduction into the patient.

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