A New Approach to Electromagnetic Blood Flow Determination by Means of Catheter in an External Magnetic Field

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Abstract. Maximal reduction in transverse catheter dimension has been achieved for the purpose of creating an intravascular electromagnetic flow sensor capable of percutaneous introduction into the vascular system. The electrodes are mounted on a flexible frame which collapses as it passes through a small branch blood vessel and expands to span the diameter of the main vascular trunk when entering it. Unlike the catheter flow sensors developed previously, which are velocimeters, i.e., sensors of fluid velocity, the present one is capable of measuring the volume rate of flow in branch blood vessels as well as in the major sections of the vascular tree. The magnetic field is provided by a large air core electromagnet placed externally to the animal or patient. A special circuit utilizing two electrodes and three leads permits reduction of the unwanted quadrature signal to zero. A standard sine wave electromagnetic flow meter channel designed for use with conventional electromagnetic flow transducers is adequate for flow measurements as well as for power supply to the large magnet. Illustrations of the performance of the apparatus in vitro and in vivo are presented.

Introduction. The availability of materials like Teflon and polyurethane which can be introduced into the blood stream for considerable periods of time without excessive clot formation, and the development of angiographic techniques of vascular catheterization, created the possibility of eliminating major surgery previously necessary in the application of conventional electromagnetic blood flow sensors which had to be placed around an isolated section of the blood vessel in which the rate of blood flow was to be measured. An intravascular blood flow sensor obviates the need for surgical exposure of the blood vessel and the convenience of this prospect stimulated developments of different approaches to intravascular blood flow measurement by means of electromagnetic catheter blood flow meters.1-5 Such catheter devices were, however, still rather large (at least about 3 mm in diameter) and required exposure and cutting of a branch blood vessel through which they were introduced into the major arteries or veins. The ideal mode of introduction of a catheter flow meter would be through the percutaneous technique which is routinely employed by radiologists for introduction of intravascular catheters through puncture in the patient's skin. A needle is introduced through skin and subcutaneous tissues into a branch artery,
such as the femoral. A flexible guide wire is passed through the needle reaching beyond the branch into the aorta. The needle is then withdrawn leaving the guide wire in the punctured artery. An intravascular catheter (for instance, french 6 in. size) is passed over the guide wire and is guided by it through the puncture in the branch artery into the main artery. The guide wire is then withdrawn and the inserted catheter can then be used for injections of required fluid materials or for passing of measuring devices into the vascular tree.

With the objective of using this mode of introduction, an intravascular electromagnetic flow meter was described\(^6\) which includes a flow transducer wide enough to span the diameter of the aorta but which is transversely collapsible so as to pass through a narrow tube or an artery branch from which it could pass into the aorta or the vena cava. There is a limit below which one could not practically reduce the lateral dimensions of such a flow-measuring catheter because it has to include a coil to generate a magnetic field besides the electrodes needed to pick up the induced flow signal. A drastic reduction in lateral dimensions could be achieved by omitting the magnetic field generating coil. It was pointed out\(^6\) that the basic structure of such a catheter could be used in conjunction with a magnetic field generated by a magnet external to the animal or patient. External magnetic fields have been previously demonstrated to be usable in conjunction with electrode bearing cuffs applied externally to arteries.\(^7\)\(^-\)\(^9\)

This paper describes the use of such an externally generated magnetic field in conjunction with intravascular catheters which permit the placement of pick-up electrodes across the diameter of a major blood vessel or a branch artery or vein in such a fashion that the magnetic field, blood vessel axis, and the vessel diameter passing through the electrodes are approximately mutually perpendicular (this special relationship is optimal but not essential, as pointed out below). Flow sensors of this type can be made not to exceed about 1.5 mm in maximum transverse dimension. Better insulating materials and procedures than those used in this development might permit further reduction in size (below 1 mm).

**Scheme of the apparatus:** Figure 1 shows the general scheme of the apparatus. The intravascular catheter C is introduced into the femoral artery F and passes into the aorta which is shown greatly exaggerated in size (the heart is omitted to better show the sensor). The sensor S is a lens-shaped flexible loop with attached electrodes (not shown in this figure). It collapses transversely so as to pass through the femoral artery and expands in the aorta to make contact with its wall. An X-ray radiogram with a properly oriented sensor provides a measurement of the aorta's diameter.\(^\dagger\) The optimal orientation of the plane of S in the configuration shown is horizontal. The magnetic field is generated by a flat circular coil of 298 turns of gauge no. 14 copper wire (Anaconda Copper Co.'s Anamid M insulation). The winding space is 4-cm wide and the inner diameter of the windings is 30 cm. The field generated at the coil center at a current of 1 amp is approximately 10 gauss.\(^\dagger\) (The coil need neither be circular nor flat.) The bed on the bobbin on which the coil is wound is lined with copper sheeting which is grounded and serves as a shield. A layer of mylar sheeting separates this copper shield from the coil windings. An outer band of copper sheeting soldered to the circular copper trough completes the shield which is not a closed
circle. An interruption of about $\frac{1}{8}$ in. prevents eddy currents from circulating in the copper shield. Electrostatic shielding is very important in this apparatus because of the high self-induced electromotive force in the magnet coil.

Figure 1.—Scheme of external field electromagnetic catheter blood flow meter.

A: aorta (for better visualization of sensor, the heart is omitted and the artery is greatly exaggerated in size). S: Sensor portion of catheter (under optimal conditions, the plane of the loop of S is parallel to the plane of the magnet coil). M: Magnet coil. F: Femoral artery. C: Catheter. L: Leads connecting catheter to amplifier. P: Plug. MCS: Magnet current supply. PSA: Phase-sensitive amplifier. SM: Sensitivity monitor. R: Recorder. CB: Condenser bank resonating the magnet coil at 400 Hz.

Figure 1 shows a magnet current supply MCS, a phase-sensitive amplifier PSA, and a sensitivity monitor SM, as separate interconnected instruments. Actually, one single instrument served all of these functions in our study. It was a phase-sensitive flow meter amplifier built for this laboratory in 1959 by Mr. J. Yee and of the type which was later available by the "Medicon" company. The magnet current supply was adequate to provide a current of 1.15 amp for the magnet coil which was resonated by a series capacitor bank CB of 3.35 $\mu$F. The functions of sensitivity monitor SM were provided by gating the phase-sensitive detector PSA so as to sense optimally a signal in phase quadrature with respect to the magnetic field and the function of the flow detector was provided by PSA in the gating phase sensing optimally a signal in phase with the magnetic field. This instrument is equipped with a dial for continuous change of the gating phase and a switch for a 90° shift in the gating. The recorder R was a Sanborn model 320 two-channel recorder. To avoid undesired effects of the stray magnetic field on the amplifier, the latter was placed at least 6 feet from the magnet coil.
The catheter flow probe: The basic experimental difficulty of this method can be seen by considering a straight-forward flow probe shown in Figure 2.

The resilient frame F (a 0.006 in. Teflon-insulated stainless steel wire) of the probe has been squeezed through the narrow side-tube B and has expanded in the wider tube T to contact diametrically opposite points of the tube wall. The magnetic field (not indicated in the drawing) is perpendicular to the plane of the drawing. The vector v indicates the direction of fluid flow. E₁, E₂ are electrodes which pick up the induced signal and convey it via the leads L₁, L₂ which pass through a catheter C (gauge no. 19 hypodermic steel tubing), to the electronic detection system. A simple probe of this kind is unusable for flow measurements because of an excessive flow-independent signal (quadrature voltage) which is induced by the alternating magnetic field in the loop formed by the lead wires L₁, L₂, and the electrolytic conductor bridging the electrodes E₁ and E₂. In spite of the use of a phase-sensitive amplifier which, in the ideal case, would be sensitive only to a flow signal in phase with the magnetic field, this unwanted emf (which is in phase quadrature with respect to the magnetic field) which may be orders of magnitude larger than the flow-induced emf, should be eliminated, or at least substantially reduced, to insure reliable rejection by the phase-sensitive flow detector. In a previous approach, a sensor utilizing three electrodes with potentiometric adjustment has been used to suppress this error signal. This solution has been abandoned in favor of the simpler system making use of two electrodes only as shown in Figure 3. The elastic insulated stainless steel frame F has two platinized platinum electrodes (E₁, E₂) attached in approximately central locations. E₂ is connected to resistor R by lead wire L₂. E₁ is not connected directly to the other terminal of R. Two leads L₁ and L₁″ are connected to potentiometer P whose wiper is connected to the right terminal of R. The voltage which appears across R is applied to the input of
the flow detector. The leads $L_1'$, $L_2''$ and $L_2$ run along the contour of the frame $F$ and are twisted beyond it. If we observe the signal which appears across $R$ in the quadrature-sensing mode of the detector, we normally detect a large error voltage at zero flow. It is, however, possible to find in a given tube a position for the wiper of the potentiometer $P$ for which this quadrature signal is reduced to zero. This adjustment has no effect upon the signal voltage which appears across $R$ in the flow sensing mode as a result of the potential difference imposed upon the electrodes $E_1$, $E_2$ in a fluid flow traversing a magnetic field.

This potentiometric method of quadrature signal elimination is needed only if the electrodes have not been centered precisely with respect to the loop of the frame $F$. Actually, the potentiometer circuit can be eliminated if the electrodes have been centered very carefully, which is to be recommended. In this case the circuit shown in Figure 3 is simplified as follows. The potentiometer resistance $P$ is eliminated, and the two wires connected to its terminals, as well as the lead wire joining the resistor $R$ to the wiper of $P$, coalesce into a single lead connected to the closed wire loop surrounding the frame $F$ and connecting the lead wires $L_1'$ and $L_2''$. This mode of guiding the electrode lead wires forms two loops (one above and another below the electrodes) in which emfs are induced so as to cancel each other's effects on the induced quadrature voltage appearing between $E_1$ and $E_2$. In the illustrations given below the potentiometer circuit was used in the flow record obtained *in vitro* and was omitted in flow records obtained in the animal.

As a preliminary instrument adjustment it is imperative to find the precise setting for the gating dial of the phase-sensitive detector at which a signal in phase quadrature relative to the magnetic field is not detected. This was done either by adjusting to zero sensitivity to blood flow pulsations and shifting the gating phase by $90^\circ$ or by connecting the amplifier input to the leads $L_1'$ and $L_2''$, and adjusting the phase setting until the detected signal is zero. This phase setting of zero quadrature signal sensitivity is the optimum adjustment for flow detection.

The voltage appearing between the leads $L_1'$ and $L_2''$ can be used for the purpose of determining how the flow meter sensitivity is affected by compression of the flow probe in a narrow artery, by the probe's removal to a point of higher or lower magnetic field strength, and by changes in the probe orientation as the plane of the frame $F$ is rotated relative to the magnetic field vector. Figure 3
shows the terminals of P connected to a "sensitivity monitor." Actually this corresponds to connection to the flow meter channel set for the quadrature sensing mode. It can be shown\(^{11}\) that the sensitivity of this type of flow meter is proportional to the product of the interelectrode distance times the intensity of the magnetic field component perpendicular to the frame area, and that the voltage between the leads \(L_4'\) and \(L_4''\) measures this product for a suitably shaped sensor frame. The voltage between the leads \(L_4', L_4''\) is measured for a standard location at which the probe is calibrated. If this voltage is reduced to \(f\) per cent of the standard reading in a given position inside an animal, we know that we are using the flow meter at \(f\) per cent of its standard sensitivity as established by calibration.

The calibration can be performed by perfusion of a plastic tube, placed diametrically across the magnet coil next to its surface, with saline at a known rate which can be measured volumetrically or by insertion of a calibrated standard electromagnetic flow transducer in series with the tube. The flow probe is placed over the center of the magnet coil with the plane of the frame parallel to the plane of the magnet coil.

It can be shown on the basis of calculations of Gessner\(^{12}\) that the conductivity of the artery wall and the surrounding tissues will cause a 20 per cent sensitivity drop of an external field electromagnetic flow meter in the living animal as compared to a calibration in a plastic tube. This factor must be taken into consideration in absolute blood flow measurements.

![Graph](image)

**Fig. 4A.**—Oscillations of NaCl solution in a 2 cm ID lucite tube, produced by a rubber bulb attached to the lucite tube through a long section of tygon tubing. The first upward peak is caused by compression of the bulb and the following large downward excursion by its release which is followed by damped flow oscillations.

(B) Blood flow in a dog's descending thoracic aorta as recorded with an external field electromagnetic catheter flow meter. The base line on the right was obtained by short-circuiting amplifier output. An ekg artifact is superimposed upon flow record.

Figure 4 shows examples of the performance of the apparatus *in vitro* and *in vivo*. In Figure 4A we see damped flow oscillations produced in a column of 0.9 per cent saline and in 4B, a record of blood flow in a dog’s descending thoracic aorta. The base-line obtained at the end of the record by short-circuiting the amplifier output coincides closely with the diastolic flow level. An ekg artifact is superimposed upon the flow record.
The experiments described above have paved the way for percutaneous measurement of blood flow which will form the next stage of development in which the method will be applied to animals and human subjects.

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† The artery diameter could also be measured purely electrically with a slight modification of the transducer described in reference 6. For this purpose, a wire loop is guided along the transducer frame F as shown in Fig. 3 so as to terminate in the leads $L_1'$ and $L_2'$. This loop is a transformer secondary in which the magnetic field generated by the external coil or by the coil incorporated in the transducer induces an electromotive force determined by the loop area. This signal can serve to measure the artery diameter and its fluctuations caused by pulsating blood pressure.

‡ The idea of the 3-electrode system with potentiometric adjustment was proposed and demonstrated by J. P. Biscar in this laboratory.

3 Kolin, A., these PROCEEDINGS, 57, 1331 (1967).
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