A Constant-Field Interrupted Resonance System for Percutaneous Electromagnetic Measurement of Blood Flow

(electromagnetic/flow meter/catheter)

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ABSTRACT A combination of deformable flow probes of negligible lateral dimensions with an electronic circuit capable of providing a prolonged plateau of $dB/dt = 0$ and of sampling the flow signal at the end of this interval permits electromagnetic measurement of blood flow with a reliable zero base line secured by switching off the magnet. An extracorporeal magnet provides the magnetic field. The flow transducer is introduced into the vascular system percutaneously through a standard angiographic catheter by conventional technique. The idea of the current generator can be described as "principle of interrupted resonance." The current wave form can be described as a sequence of disconnected bisected sine waves joined at the apices by horizontal current plateaus where di/dt is strictly zero.

The original form of the electromagnetic blood-flow meter was based on measurement of the electromotive force (emf) induced in a bloodstream traversing a *constant* magnetic field (1). This system suffered from electrode polarization and could be used only with cumbersome nonpolarizable electrodes (1-3). Introduction of an alternating magnetic field (4) made it possible to utilize ordinary metal electrodes, but at the same time injected a new difficulty: a transformer emf induced in the electrode lead circuit and the interelectrode conductors. This perturbation constituted until now the principal experimental barrier to ideal operation of the method, and a great deal of work has been devoted to a variety of proposed solutions. Without changing the wave form of the original sinusoidal ac, phase-sensitive detection of the flow signal by sampling at the peak of the magnetic field intensity where $dB/dt = 0$ was one approach to the elimination of the transformer emf (4, 5). Another approach sought to improve the suppression of this disturbance on the basis of the above idea by prolonging the interval of $dB/dt = 0$ through use of "square wave" excitation of the magnet (6). That current wave form is, however, an idealization that cannot be obtained in practice even in a small probe using a magnet of small inductance. The transformer emf induced in this scheme declines exponentially, and there is, strictly speaking, never a period of $dB/dt = 0$ and thus no elimination of the flow-independent error signal (7). This problem would, of course, be greatly magnified, should the small flow-probe magnet of the conventional approach be replaced by our three to four orders

of magnitude larger external field magnet with a correspondingly greater inductance.

In the present "external field method" where the entire torso of an animal or patient can be exposed to an alternating magnetic field, the zero-flow error signal is greatly augmented by eddy currents induced in the heterogeneous volume conductor (torso). This unwanted emf can be hundreds of times as large as the flow signal, imposing an overload on the phase-sensitive detector. Moreover, this error signal may not be strictly in phase quadrature with respect to the magnetic field‡ so that the error signal component in phase with the magnetic field will introduce a flow-independent reading indistinguishable from a flow signal. Such disturbances are particularly severe when conductivity inhomogeneities are located in the eddy current field in the vicinity of the flow-sensing electrodes‡. Thus, the motion of the diaphragm and inflation and deflation of the lungs can cause severe periodic disturbances when flow is measured near them in the aorta.

We have set ourselves the objective of eliminating perturbations due to eddy currents and polarization and of achieving the capability of establishing a *reliable* zero-flow base line by turning off the magnet current without causing a base-line shift. The solution of this problem is based on a new approach, which can be described as an "interrupted resonant circuit," that is outlined below.

The magnet current supply

The two keys to the elimination of the zero-flow error signal in any implementation of the electromagnetic flow-meter principle are: (1) the achievement of an interval of constant magnetic field during which to sample the flow signal ($dB/dt = 0$) and (2) periodic inversion of this magnetic field to suppress emfs due to polarization at interfaces of heterogeneous conductors. The latter disturbance becomes particularly prominent in the present external field modification since the electrodes are not shielded (as in cuff flow meters) from external currents and because the exposure of large segments of the body generates widespread eddy currents. To maximize attenuation by discharge of polarization emfs, the period of $dB/dt = 0$ must be made sufficiently long and the flow signal must be sampled at the end of this period. These requirements have guided us in the design of the present circuit.

In principle, we could pass through the external magnet coil alternating currents with prolonged plateaus of $di/dt = 0 = dB/dt$ by feeding the coil with a current derived from an

Abbreviation: emf, electromotive force.

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appropriately switched constant current supply. In view of the large inductance and appreciable ac resistance of the large magnet coil, this conventional approach would entail great practical difficulties due to large power losses in the system and high voltages induced in the coil during rapid current reversals. These obstacles have been overcome by the idea of temporarily switching the magnet coil into a series resonant circuit to store the energy of the collapsing magnetic field in a capacitor that would return it to the coil, with compensation for energy losses from the constant current source. Because of resonance the circulating currents can be extremely large even with a small power supply. As a result, one can use in the present circuit a power supply of 36 V to maintain an "alternating constant-plateau-current" (see i_L in Fig. 2) of 1 amp (plateau value) instead of an 800-V supply that would be required in a conventional approach. The excess power in the latter case would have been dissipated as wasted heat.

Fig. 1 shows the scheme of the magnet current supply. V is a dc voltage supply of adequate power rating, and I is a switched precision constant-current source that stabilizes the crucial period of $di/dt = dB/dt = 0$. In practice, for simplicity of design, two identical storage capacitors are alternately switched to opposite ends of the magnet coil to allow the use of a single current source.

The current can be switched to pass either through the upper left switch (UL) or the upper right (UR). Crossed pairs of switches are activated so that either UL and LR or UR and LL are on. During these periods a constant current passes through the magnet coil inductance L either to the right or to the left while the switches S_1 and S_2 are disconnected. When, for example, UL and UR are opened and simultaneously UR and LL are closed, S_2 is closed at the same time for a brief period. This creates a resonant LC circuit in which a sinusoidal current begins to flow in the following path as the magnetic field of L collapses: L, C_2, S_2, LL . The condenser is charged to a high voltage as the magnetic field energy is stored as electrostatic field energy in the condenser C_2 . As C_2 begins to discharge through L , the current in it reverses direction, and at the point where the current in L reaches the value of the programmed constant current, S_2 is turned off.

FIG. 1. Current supply for magnet coil L . V : dc power source; I : constant-current source; UL, UR, LL, LR, S_1, S_2 : switching devices; D_1, D_2 : diodes; C_1, C_2 : capacitors.

The analogous process is repeated when the current in L flows in the reversed direction, whereafter S_1 is turned on to form the resonant current path L, C_1, S_1, LR . The variable switching interval was adjusted to approximately 5 msec (compare Fig. 3).

The switching diagram of Fig. 2 shows the time relationships in the switching sequence, which is controlled by a master clock relying on standard digital techniques to derive all switching functions. The figure shows the relationship between the "on" and "off" states of UL, LR and UR, LL and the relationships of the "on" intervals of S_1 and S_2 to those of the aforementioned switches. We note especially that the current in L varies sinusoidally (in the $L-C$ circuit) during the closure periods of S_1 and S_2 . We shall make use of these sinusoidal current periods for calibration as described below. The critical constant current period ($di/dt = 0 = dB/dt$) lies between the interruptions of S_2 and LL for one direction of i_L and the interruption of S_1 and closure of UR for the reversed direction of the coil current.

The slopes of the composite current function [$i_L = f(t)$ of Fig. 2] correspond thus to a sine wave that has been cut at the apices and drawn apart, joining the maxima and minima by horizontal lines which represent the vital periods of current plateaus. The bottom diagram of the sequence of Fig. 2 shows the periods during which the amplifier samples the flow signal at the end of the current plateaus after subsidence of transient

FIG. 2. Time relationships in the switching sequence. The labels on the left specify the devices whose state is depicted in the time sequence diagrams. The lowest (i_L) diagram depicts the time course of the magnet current.

FIG. 3. Oscillograms of the time relationships between the voltages V_1 , V_2 at the coil terminals with respect to ground and the rate of change of the magnet current (di_L/dt), which is proportional to the rate of change of the magnetic field (dB/dt). The sinusoidal peaks are recorded during the resonance phase, and the flat portions during the open switch phase and after stabilization of the constant current. The brief sections linking the above two phases occur while the constant current supply is making up power losses before achievement of the constant current phase. The low constant voltage plateaus of V_1 and V_2 are due to the iR drop in coil L (Fig. 1) during the constant current phase. The flat portion of the dB_L/dt tracing is zero, implying a period of strictly constant magnetic field.

responses evoked by the sinusoidal ac intervals. Fig. 3 shows oscilloscope tracings of the voltages (V_1 and V_2) between the coil terminals and ground as well as the rate of change di_L/dt of the coil current. The achievement of a constant current interval ($di_L/dt = 0$), and hence of a constant magnetic field, is clearly in evidence.

The signal amplifier

The signal amplifier scheme is shown in Fig. 4. It is controlled by the master switching circuit in such a way that it is activated only for a short interval at the end of the $dB/dt = 0 = di/dt$ period when the flow-induced emf provides the only measurable signal unperturbed by spurious transients.

The input switching devices $FS1$ and $FS2$ serve the function of connecting the primary of the input transformer T to the flow-probe electrodes. The grounded center tap is connected to the probe's ground electrode G of Fig. 5A. The switches

FIG. 4. Signal amplifier. T : Transformer; $FS1$, $FS2$, $FS3$: switching devices; A : operational amplifier; D : phase-sensitive detector; R_1 , R_2 , R_3 , $FS3$: elements of feed-back circuit; TC : time constant network; DC : dc-coupled operational amplifier; M : flow-monitoring voltmeter; R : recorder.

FIG. 5. (A) AR : artery; M : extracorporeal magnet coil; L_1 , L_2 : loops of the bifilar wire whose strands W_1 , W_2 (shown as solid and dashed lines, respectively) form the sensor loop L . E_1 , E_2 : flow-sensing electrodes; G : ground electrode; ST : probe stem formed by strand of wires; P : plug; FP : triplet of prongs connected to electrodes E_1 , E_2 and G for flow measurement; LP : pair of prongs for measurement of emf induced in loop L_2 used in calibration. (B) Loop probe in 3-mm (inner diameter) Teflon tube (T). (C) Loop probe resting on a millimeter scale after emerging from a 1-mm (inner diameter) Teflon tube (t).

remain open to isolate the amplifier from the disturbances of transient emfs except for a 500- μ sec period at the end of each current plateau. The output of the transformer secondary is fed to the input of a gain-controlled low-noise operational amplifier A . The feedback circuit, consisting of R_1 , R_2 , R_3 , and switch $FS3$, is connected to change the gain of the amplifier from approximately 10 to 1000 when the switch is closed (simultaneously with $FS1$ and $FS2$).

The detector D is a phase-sensitive detector whose output is linked through a switch-adjustable time constant network TC to a conventional dc-coupled operational amplifier DC , which is connected to the input of recorder R and a flow-monitoring voltmeter M . The unconventional aspect of this circuit is the aforementioned input and amplifier gain change. Details of the electronic circuitry will be published elsewhere. §

The flow transducer

Fig. 5A depicts the scheme for the electromagnetic measurement of blood flow. The magnet coil M is to be imagined outside the patient's or animal's body. The flow records shown in Fig. 6 have been obtained with a 13-cm outer diameter (7 cm inner diameter), 852-turn, gauge-18, formvar-insulated copper wire coil (3-cm thick) insulated by encapsulation in "Teets Denture Material" (Co-oral-ite Manufact. Comp., Santa Monica, Calif.). It was driven by a current of 1-amp plateau value in all experiments. The artery AR harbors the oval loop probe L , which is made of a bifilar pair of polyurethane-insulated "Elgiloy" wires 0.1 mm in diameter (Elgiloy Comp., Elgin, Ill.). The two wires of the bifilar pair are shown separated for clarity, one of them represented by a solid line and the other by a dashed line. Actually we should imagine the

§ Steele and Imai: in preparation.

FIG. 6. Verifications of the magnetic zero-flow base line. (The timer marks intervals of 1 sec. The timer "black-outs" at *M* mark the periods when magnet is off.) (A) Diastolic flow is arrested by occlusion balloon in dog's carotid artery at *left arrow*, and occlusion is removed at *right arrow*. The magnet is switched off during interval *M*. The "magnet-off base line" coincides with the diastolic occlusion base line. (B) At high-gain and high-time constant (to measure average flow), the dog's carotid artery is occluded at *left arrow* and released at *right arrow*. While the artery is occluded, the magnet is turned off during interval *M* (note switching transient). The magnetic and mechanical base lines coincide perfectly. (C) Phasic record of blood pressure *p* and blood flow *F* in dog's femoral artery. The flow is occluded by balloon at *left arrow* and released at *right arrow*. The magnet is off during interval *M* (note switching transient). The magnetic and mechanical base lines coincide. (D) Illustration of normal mode of securing magnetic base line for flow records. *P* is a record of average blood pressure and *F* of average blood flow in a dog's femoral artery. The magnet is turned on at *arrow* after recording a base-line section. The pressure artifact and the arrow at *IN* mark intra-arterial injection of the drug into femoral artery (0.25 mg of acetylcholine). The initial rise in flow is accompanied by a drop in blood pressure. (E) Base-line verification in dog's abdominal aorta by cardiac arrest. A magnetic base line is recorded at *M*, whereupon the magnet is turned on. During the "timer black-out" at *IN*, 6 mg of acetylcholine are injected intravenously. This is followed by a period of drop in blood pressure (*p*) and cessation of flow (*F*), which yields a mechanical base line identical with the magnetic base line at *M*. After resumption of the heart beat, the diastolic flow level between the first five beats coincides with the magnetic base line.

dashed wire to be under the solid one, i.e., behind the page. The electrodes E_1, E_2 are obtained by scraping off the insulation of a short section of each of the two wires W_1, W_2 , which form two congruent loops L_1 and L_2 as well as the stem ST which terminates in the plug P . An additional polyurethane-insulated copper wire about 0.08 mm in diameter runs parallel to the W_1 and W_2 wires terminating in a section G , bared of insulation, which serves to ground the subject by connection to the center (ground) pin of the pin triplet FP of the plug P . The outer pins of this "flow pins" triplet convey the flow signal from the electrodes to the flow detector-amplifier. The pin doublet LP (loop pins) conveys the emf induced in the loop L_2 to a VTVM for calibration purposes (see below). The procedure of manufacturing such probes (with different electrode connection schemes) has been

described elsewhere (8). Fig. 5*B* and *C* gives an idea of the actual dimensions of such a flow transducer as viewed against a metric scale. Fig. 5*C* shows a loop probe expanded to maximum width, whereas Fig. 5*B* shows it in a 3-mm (inner diameter) Teflon tube compressed as it would be in a branch artery.

The induced emf signal derived from the loop L via the pin doublet LP is a measure of the product of the area of the loop L (which is nearly proportional to the internal diameter of the artery) and the magnetic field component perpendicular to the plane of the loop. This product is used for calibration of the flow transducer as described in a previous publication (9), which reviews additional aspects of the calibration. This calibration signal is induced in the loop during the sinusoidal portions of the current i_L shown in Fig. 2.

Verification of the magnetic zero-flow base line in animal experiments

It would not be practical to make blood-flow measurements with this method without the capability of establishing the base line corresponding to zero flow without flow arrest. At the root of the aorta such a base line is provided automatically by the zero flow level in mid-diastole (10). In other locations, using previous methods, the base line had to be secured by compression of the blood vessel (11). The present apparatus yields a *reliable* base line by simply switching off the magnet, as was possible originally with the constant magnetic field (1-3). To verify this base line, hundreds of experiments were performed in different arteries of dogs. In small arteries, the flow of blood was stopped by means of occlusion cuffs (12, 13). While the blood flow was arrested, the magnetic field was switched on and off to demonstrate that its presence did not affect the zero-flow base line. The sensitivity of this test was increased by watching a smoothed base line at a somewhat higher time constant (250 msec instead of the value of 25 msec used in phasic flow measurements). In the aorta the flow was arrested by stopping the heart for several beats by drug action.

The degree to which a satisfactory base line can now be obtained is demonstrated in Fig. 6. The periods during which the magnet is switched off are indicated by blacking out of the time signals at *M*. The arrows indicate occlusion and release of the blood flow.

Fig. 6*A* and *B* have been obtained in a dog's carotid artery. The occlusion cuff was inflated so as to stop the flow during diastole but not during systole. The diastolic level of this occlusion portion of the phasic flow record clearly coincides with the "magnet-off base line." The characteristic transient flow increase on reestablishment of blood flow is pronounced. (The peak at the end of the "magnet off" period is a switching transient.)

In Fig. 6*B* the same experiment is repeated at 4-fold gain and a higher time constant (12.5 sec) used in average flow measurements. (There is a switching transient similar to that observed in Fig. 6*A*). In spite of the high gain, there is no perceptible disparity between the mechanical and magnetic base lines.

Fig. 6*C* shows a simultaneous record of blood pressure (top) and femoral blood flow in a dog[†]. The lower record is interrupted during the base-line check, which exhibits perfect correlation between the mechanical and magnetic zero.

Having gained confidence in the magnetic base line, we normally use it exclusively as shown in Fig. 6*D*. A period

during which the magnet is turned off before the flow recording (or at its end) establishes the base line. Fig. 6*D* illustrates a typical experiment in which the effect on average blood flow and pressure of an intra-arterial injection of 0.25 mg of acetylcholine into a dog's femoral artery is tested. A sharp increase in blood flow is accompanied by a drop in arterial blood pressure.

Fig. 6*E* illustrates a verification in a dog's aorta (just below the renal arteries) of the magnetic base line (at *M*) with a stop-flow base line obtained by intravenous injection of 6 mg of acetylcholine. The *precise* correspondence between the magnetic base line and the base line during the prolonged cardiac arrest, as well as during the extended diastolic intervals following it, is quite apparent.

Conclusion

The electromagnetic method of blood-flow determination has reached a point where it is possible to record blood flow in the aorta and branch arteries by means of percutaneously introduced flow probes of negligible dimensions. The magnetic field is provided by an extracorporeal magnet. A suitably designed electronic system energizes the magnet with an unconventional current wave form that yields a reliable base line by turning off the magnetic field.

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[†] The blood pressure was measured in all cases about 4 cm below the electrodes (through the same angiographic 7F catheter that introduces the probe) by means of a Statham pressure transducer.