Effect of Hematocrit Value upon Electromagnetic Flowmeter Sensitivity


ABSTRACT

An electromagnetic flowmeter system that has an accuracy of 0.1% and is insensitive to fluid conductivity was used to study the sensitivity to steady flow of a 6-mm cannular transducer. All measurements were carried out at 37°C. Sensitivity decreased with increasing hematocrit value whether flow was laminar or otherwise. With laminar flow and a given hematocrit value, the sensitivity also decreased with increasing flow rate. Sensitivity started to rise when increasing flow rate caused departure from the laminar regime. At the lowest flow used in the laminar regime (250 ml/min), the sensitivity fell 4% when the hematocrit value was changed from 0 to 66%; at 39.5% the reduction in sensitivity was 2%. With laminar flow at hematocrit 39.5% the sensitivity fell 1% as flow increased from 250 to 1500 ml/min. At hematocrit 29.5% the sensitivity rose 1.2% when the flow regime changed from laminar to turbulent. The changes may be explained in terms of a radial distribution of cells, of which a cell-free boundary layer is a particular example. Theoretical calculations show that the thickness of an apparent cell-free boundary layer must be tens of microns to explain the observations with laminar flow and that the apparent thickness is greatly reduced with onset of turbulence. A possible mechanism is described by which the results could be explained in terms of cell orientation.

ADDITIONAL KEY WORDS

laminar flow turbulent flow radial distribution of red cells plasma cell-free plasma cell orientation

Observations made in this laboratory (unpublished) and elsewhere (see Discussion) indicate that the cell content of blood may have a small but definite effect upon the sensitivity of electromagnetic flowmeters. We set out to establish the magnitude of the effect under controlled conditions, and if possible to explain it. We were aware that possible effects due to hematocrit in a correctly designed flowmeter system could be explained by an axisymmetric radial variation of electrical conductivity in the blood, such as might be caused by a peripheral cell-poor or cell-free zone some tens of μ thick. It also seemed possible that any effect would depend upon the flow regime (laminar or turbulent). We therefore included in our measurements the electrical conductivities of the fluids under examination and the pressure difference along the tube through which they flowed.

Methods

THE TRANSDUCER

An x-ray photograph of a flowmeter of the cannular type used is shown in Figure 1. The tube was made of epoxy with a central region 6 mm in diameter and 11 mm long, which tapered over a further distance of 10 mm at each end to a 7.5-mm diameter. The field strength at the center of the cannula was 136 gauss rms and the nominal interelectrode voltage was 11.6 μv rms with a flow of 250 ml/min. The magnet was excited at 0.3 amp rms ± 1%, at which value the peak flux density in the iron circuit was 35% less than the saturation value. The power dissipation was 0.25 w. At the side walls, the field strength was 6% greater, at one electrode 1.5% greater, and at the other, 4% less, than that at the center. The reduction in sensitivity due to circulating currents at the end boundaries of the magnetic field ("end-shorting," Shercliff [1]) was calculated to be a factor 0.958, and found experimentally to be 0.968. No ground electrodes were fitted, and the construction was in accordance with principles previously described (2). The magnet was

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energized with sinusoidal current at 240 Hz. The quadrature voltage between the electrodes was less than $10 \mu \text{V} \text{rms}$ (see below). Cavity electrodes were used, 1 mm in diameter and 1 mm deep. The electrical conductivity of fluids used in the experiments ranged from 1 to 50 mohm$^{-1}$ cm$^{-1}$, from which it may be found that the transducer source resistance ranged approximately from 2 to 40 kilohms (3).

**ELECTRONIC UNIT**

The electronic unit (4, 5) used an input circuit having an impedance from each terminal to ground exceeding 100 megohms in parallel with 20 $\mu$F, and between terminals of 0.9 megohms. After amplification $\times 4000$ the signal passed through a suitable band-pass filter to a further gain stage $\times 250$ and thence to two full-wave phase-sensitive detectors, one of which responded to the signal caused by flow (phase voltage), the other to quadrature voltage. The latter response was used to generate an opposing quadrature signal which was fed into the input circuit in series with the electrodes, thus reducing the net quadrature voltage in the input circuit by a factor approximately 50.

**ELECTRICAL MEASURING SYSTEM**

The voltage between the transducer electrodes at $e,c$ (Fig. 2) was balanced against a manually adjusted potential derived from a substandard resistance $R$ through which the magnet current $i$ was passed. Balance was obtained to within 1% of the flow signal, and the out-of-balance voltage integrated over a measured period of time. This system had the following advantages: the amplifier gain variations affected the measurement of flow voltage only in the order of 0.01%; the input impedance seen by the transducer was increased.

![Image of transducer, 6-mm bore.](image)

**FIGURE 2**

Balance and integrating circuit for measuring flow voltage. $A$, amplifier; $V_{ao}$, automatic quadrature suppressing signal; $M$, magnet coils; $R_1, R_3$ decade resistance boxes.

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by a factor at least 100, thus reducing error with the lowest conductivity fluid used to less than 0.04%; it eliminated the difficulty experienced when setting the bridge balance to an accuracy of better than 1% due to small fluctuations in flow caused by imperfections in the flow generator; the transducer and the balancing potential were equally affected by any small variation in magnet current which might occur (less than 1%), thus avoiding any error from this cause and resulting in a measure of transducer response per unit exciting current; and readings consistent to better than 0.1% were possible except at the lowest flow, where on occasion the difference between consecutive readings was 0.15%. The integrating circuit had a time constant of 1150 seconds and was fed by the amplified flow signal via an electromechanical detector (Carpenter relay), thus avoiding errors due to drift in the d-c output stage of the electronic unit, which had been designed for high stability at the 1% not 0.1% level. The integrated voltage across the capacitor was always less than 10 mv, which was the full-scale reading of the Vibron electrometer used for measuring it. A stable phase voltage less than 0.2 μV rms was present between the electrodes with no flow, and this was balanced out with a d-c supply in series with the integrator until the integrating voltmeter showed a movement of less than 0.2 mv/min, corresponding to a net baseline drift no flow, and this was balanced out with a d-c supply in series with the integrator until the integrating voltmeter showed a movement of less than 0.2 mv/min, corresponding to a net baseline drift. Baseline drift was most noticeable with the lowest conductivity fluid, since noise, including that at very low frequency, was then greatest. The relative sensitivity \( S \), normalized to a flow of 250 ml/min, was found to be 0.005 μV rms. Baseline drift was most noticeable with the lowest conductivity fluid, since noise, including that at very low frequency, was then greatest. The relative sensitivity \( S \), normalized to a flow of 250 ml/min, was found (using the experimental data) from the following formula (see Appendix 2):

\[
S = \left( \frac{250}{Q} \right) \left( \frac{50}{f} \right) \left[ \frac{R_2}{1 + 0.0011(10 - R_2)} + \frac{\alpha V_Q}{22700} \left( 1 + \frac{9.8}{Q} \right) \right],
\]

where \( Q \) is the flow in ml/min, \( f \) the line frequency, \( R_2 \) the bridge setting in kilohms, \( \alpha \) a calibration factor embracing the amplifier gain and integrating circuit time constant, and \( V \) the integrated voltage in mv. The value of \( S \) was near 1. The voltage \( V_Q \) (Fig. 2) represents the automatic quadrature suppression signal.

**FLOW SYSTEM**

A flow generator (6) was used to produce steady flows ranging from 250 to 2000 ml/min. The mechanism consisted of a piston driven by a lead screw. The latter could be moved in either direction by a rotating nut driven by a synchronous motor via a gear box. Microswitches were used to terminate the piston excursion, and to switch on and off the flow integrator. The relationship between one flow and another was exact, being dependent only upon toothed gear ratios. The repetitive accuracy at a given flow depended upon the mean rate of rotation of the motor, which depended upon the mean value of the line frequency during an observation. A frequency meter was used which could be read to better than 0.02 Hz, or 0.04%. The frequency rarely varied as much as this during an observation, the maximum period of which (at 250 ml/min) was 91 seconds. The piston (made of Perspex tube) had a nominal diameter of 7.5 cm, and its maximum variation in cross-sectional area, which occurred at one point along the length used, was 0.25%. This produced no measurable error, since the flow voltage was measured between fixed points of movement of the piston (approximately 12 cm apart). The cylinder within which the piston moved was made of Perspex and filled with tap water. The space around it was maintained thermostatically at 44°C, at which temperature it was found that the temperature of the flowing liquid remained closely at 37°C. The fluid under test was contained in a system consisting of a pliable plastic bag connected to a reservoir by a length of Perspex tube in which the transducer was included (Fig. 3). The bag was firmly secured to a rim on the underside of the lid of an upright cylindrical vessel. The reservoir (Fig. 3) contained a perforated stainless steel plate fitted with a handle, which served both for stirring the blood and as a ground electrode. On each side of the transducer were Perspex tubes 20 cm long bored to 6-mm diameter; their ends adjacent to the transducer were flared to a 7.5-mm diameter over a length of 10 mm and slightly bell mouthed. The ends of the transducer were slightly rounded and fitted into the bell-mouthed regions of the adjacent tubes to form a coaxial system. There was a further 30-cm length of nominal \( \frac{3}{8} \)-inch (6.35 mm) tube between the flowmeter and the reservoir. The whole length was clamped to a platform in a water bath. The tube was covered to a depth of 2 cm with water maintained at 37°C. Two side tubes, each 17 cm from the transducer, were provided for measuring pressure difference.

**PRESSURE MEASUREMENT**

The pressure difference was measured with an Elema-Schöndner differential air manometer (Type EMT 490 Bd) with ranges 0-10, 0-20 and 0-30 cm water. The air-liquid interfaces were obtained in glass tubes 2 cm in diameter and...
ELECTRICAL CONDUCTIVITY MEASUREMENT

A Radiometer conductivity meter Type CDM 2c and conductivity cell Type CDM 104 were used at a frequency of 3000 Hz. The apparatus was always checked with a standard KCl solution before use. Difficulty arose with blood of high hematocrit because fibrin formation rapidly coated the surface of the conductivity cell. This was prevented by adding heparin to all blood used (see Experimental Procedure).

HEMATOCRIT MEASUREMENT

A Hawkesley microhematocrit centrifuge was used for estimate of hematocrit on duplicate samples which were spun for 5 minutes. These observations are prone to the error recently demonstrated by Sirs (7).

Experimental Procedure

The effect of hematocrit, temperature, and hemolysis on the electrical conductivity of stored bank blood and the effect of temperature and hemolysis on that of plasma was investigated initially. Hemolysis was produced by shaking in a Warburg bath at 37°C for 5 hours.

The main experiments relating flowmeter sensitivity to hematocrit were conducted as follows. Cells and plasma of group O blood (blood bank blood, 420 ml/acid-citrate dextrose, 120 ml) were recombined on the day prior to the experiments to make up seven 700-ml volumes of blood with measured hematocrits in the range 9.5% to 66%. Heparin, 30 mg/600 ml, was added, and the conductivity of an aliquot of each hematocrit was measured at 37°C. The blood was stored overnight at 6°C, and the flow experiments were performed on the following day. The volumes were used in ascending order of hematocrit, and a clean dry bag was used for each volume and for plasma. The fluid to be tested was placed initially in the bag and, in the case of blood, was pumped to the reservoir and back into the bag several times to ensure that the blood finally in the reservoir would not differ significantly from that originally in the bag. Measurements of transducer sensitivity were made only with the fluid flow directed from the reservoir to the bag (Fig. 3). When the fluid was changed it was drained and blown out from the horizontal tube, and the reservoir was drained and dried. The tube remained submerged and warm throughout, and the blood and new bag were prewarmed to 37°C. Between the final observation of a series using blood and the following observations using plasma, the system was washed out with a glucose-saline solution having a conductivity of 14.5 mohm⁻¹ cm⁻¹.

Pairs of observations of sensitivity at each flow in the sequence 250, 500, 1000, 1500, 2000, and 250 ml/min were recorded for each value of hematocrit. The pressure difference was recorded with each observation. The temperature of fluid in the reservoir was maintained at 37 ± 0.2°C. The hematocrit of the well-mixed blood in the reservoir was measured at the beginning of each

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set of flow observations. If different from that of the previous day (1% difference being typical) the value of conductivity taken the previous day was corrected using the curve of Figure 4.

The plasma and whole blood conductivities were corrected for any temperature deviation (e.g. 0.2°C) using the data of Figure 5. The conductivity of the plasma (obtained by centrifuging the whole blood) was measured at the end of each set. The blood was stirred by gently moving the perforated steel plate in the reservoir up and down a few times during each observation. (We noted that rate of stirring had no perceptible effect upon the results, probably because the integrating system tended to obscure any small nonuniformity of hematocrit by recording the response to the passage of a certain number of cells in a certain amount of plasma.) Pooled plasma of unknown group was used as a standard before and after a series of observations including all hematocrits and flows. Similar observations were also made with saline having conductivities of 51, 19.2, 9.6 and 1.06 mho\(\text{m}^{-1}\text{cm}^{-1}\). When the electrode diameter of the transducer was vertical, the variation of sensitivity at the lowest flow in

Electrical conductivity of human blood bank blood approximately 25 days old. Heparin, 30 mg/600 ml, 37°C.

Electrical conductivity of human blood bank blood vs. temperature at different hematocrits.
Results

ELECTRICAL CONDUCTIVITY

The results are illustrated in Figures 4 and 5. The conductivities of plasma from blood that was centrifuged before and after hemolysis (see Experimental Procedure) were 13.96 and 14.34 millimho respectively. The change indicated that the degree of hemolysis encountered in the flow experiments was likely to be only of marginal significance.

FLOW EXPERIMENTS USING SALINE

Results were obtained using saline of different conductivities (51, 19.2, 9.6 and 1.06 mohm$^{-1}$ cm$^{-1}$), for each of which the sensitivity was measured using several flow rates. Each point (Fig. 6, top) represents the mean of at least 2 observations made at each flow rate and conductivity, relative to the mean value of all readings taken at the lowest flow (250 ml/min). Where four points are not distinguishable at a given flow, one or more are superimposed. At the lowest flow (250 ml/min) the effect of noise (especially with the lowest conductivity) was reflected by an overall variation of 0.14%. Otherwise, but excluding observations at 500 ml/min, the largest variation between consecutive observations at any flow and conductivity was less than 0.1%. The pressure difference curve shows that the sign of departure from the laminar regime occurs near 500 ml/min ($N_r = 2300$). It was at this flow that a wide nonsystematic variation of flow signal was observed. We consider that at flows of 1000 ml/min and above, turbulence was fully established, the 0.75% increase of flow signal in this range being near that expected due to nonuniformity of magnetic field (8) (see Methods). We classified the flow at 500 ml/min as of an unstable regime intermediate between laminar and turbulent. The results established that the system was insensitive to conductivity over a range far larger than that of the blood used in the experiments, and would yield consistent results to an accuracy of at least 0.14% and, in most cases, of about half this value.
FLOW EXPERIMENTS USING BLOOD

Pressure Difference Data.—It was desirable to define the type of flow (laminar, intermediate, or turbulent) at a given flow and hematocrit value. The usual method of plotting flow as a function of pressure yields an indefinite transition point between laminar and turbulent flow, and this has resulted in different artifices for clarifying the information available from the data (9, 10). The present data are illustrated in Figure 7, where pressure difference is plotted as a function of viscosity relative to the highest value obtained (at 66% hematocrit) with flow as parameter. The relative viscosity was obtained from the slope of the pressure vs. flow curves over the linear regions at each hematocrit. We have defined those points in Figure 7 not on the linear regions but on regions of positive slope, as turbulent; points on regions of negative slope as intermediate; and points on the linear regions (to the right of the dotted line) as laminar.

Sensitivity versus Flow.—Figure 8 shows the relationship between sensitivity S and flow at different hematocrits, with the type of flow indicated from the data of Figure 7. The sensitivity was calculated from equation 1 and is seen to fall with increasing value of hematocrit at all flows. When the flow is laminar, the sensitivity falls with increasing
flow at all hematocrit values. Figure 9 shows the results corrected for magnetic field nonuniformity on the assumption that sensitivity variations observed with plasma were due only to this effect, taking due note of the flow regime. The figures are given relative to the mean value of S for the initial and final observations with plasma at a flow of 250 ml/min. Figure 10 shows the sensitivity relative to that with plasma, averaged over all observations made with laminar flow at each hematocrit value.

**Apparent Cell-Free Peripheral Layer.—** The thickness \( t \) of the apparent cell-free layer was calculated (Appendix 1, equation 1) and plotted against the wall shear stress \( T_w \) for different hematocrits and laminar flows (Fig. 11).

### Discussion

Examination of the effect of hematocrit upon electromagnetic flowmeter sensitivity has resulted in many variable and sometimes conflicting results. Khouri and Gregg (11) reported a fall in sensitivity of 15% when using, within an artery, blood of 45% hematocrit instead of saline or plasma; Spencer and Denison (12) reported a fall of 2% in sensitivity for each 10% increase in hematocrit when using extracorporeal transducers; Brunsting and ten Boor (13) observed a fall in sensitivity of 14% when using blood of 55% hematocrit in a cannular transducer, and about 50% when using a cuff flowmeter on a blood vessel; Bond (14) found no significant effect with cannular transducers; Beck et al., (15) using a cannular transducer, found a fall of 5% with blood of 70% hematocrit.

The literature is confusing for two reasons. Generally there has been no distinction between the effect of hematocrit with and without a blood vessel present, and only rarely has the effect of fluid conductivity (as distinct from hematocrit) been examined or stated. The introduction of a blood vessel greatly alters the experimental conditions, and the way in which sensitivity depends upon hematocrit can be expected to vary with vessel geometry and conductivity (16, 17). If, in the absence of a blood vessel, the instrument response is dependent upon fluid conductivity, as Beck et al. (15) have shown the pulsed-field flowmeter to be, this effect and any due to hematocrit will be difficult to separate.
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since conductivity changes with hematocrit value (Fig. 4).

The cause of variation in sensitivity with hematocrit value is not clear. We have shown (Fig. 11) that our results with laminar flow are consistent with a cell-free peripheral layer some tens of microns thick. Flow in such a layer tends to reduce the shunting effect of the layer (see Appendix 1) and if the velocity profile becomes more uniform, as in turbulent flow, the sensitivity of a transducer is less affected by the presence of the layer. When the flow is turbulent the apparent layer thickness deduced from our results represents a least value. At a hematocrit value of 39.5%, for example, the apparent layer thickness fell from a maximum value of 85 μ with laminar flow to a value not less than 35 μ with turbulent flow. There is no direct experimental evidence either for or against the existence of such thick cell-free layers in tubes of diameter about 6 mm, although there is direct evidence for their existence in smaller tubes in thicknesses up to 14 μ (18-21). The presence of a cell-free zone has been postulated to explain the non-Newtonian behavior of blood flowing in tubes, and the apparent thickness of such zones has been calculated from data relating flow and pressure gradient. The values obtained differ widely. Haynes (22) found effective cell-free zones varying in width from zero to a few microns, and stated "the effect can be considered to be virtually non-existent in tubes whose diameter is 1 mm or more." Charm and Kurland (23) deduced the existence of zone widths as great as 48 μ in tubes up to 1 mm diameter; Hershey and Cho (24) used tubes from 4.1 mm to 7.4 mm in diameter and calculated effective zone widths up to 46 μ; recently Charm et al. (25) postulated values as great as 120 μ in tubes of the order of mm in diameter, dependent upon tube size, hematocrit and flow velocity. The same authors conclude that no cell-free layer forms with blood of hematocrit value greater than 40% in tubes of diameter greater than 0.3 mm. Various refinements of the two-phase flow model have been proposed and tested, such as the existence of finite unsheared laminae in the liquid (22, 26) and radially distributed cells rather than a peripheral cell-free zone (25). Our results do not demand for their explanation a simple peripheral cell-free zone; they can be interpreted equally well as the consequence of some rotationally symmetric distribution of cells with its attendant radial variation of electrical conductivity. Alternatively, our results can be explained by anisotropic conductivity of the blood, particularly by a difference between the conductivities in the radial and circumferential directions, which may or may not be accompanied by some nonuniform radial distribution of cells. Suppose that in laminar flow the erythrocytes are generally aligned with their main surfaces parallel to the direction of flow. As a consequence, the electrical resistance of the blood would be greater in a diametral direction than in a circumferential one. When laminar flow occurs in an electromagnetic transducer, symmetrical circulating currents are set up in the tube cross section (1), and these are directed in part along that tube diameter on which the electrodes lie and in part circumferentially. The postulated difference in resistivity in these directions could so affect the potential drop between the electrodes as to account for our observations. Phibbs and Burton (27) have shown that orientation of erythrocytes with their biconcave surfaces parallel to the shear plane of laminar flow tend to be more pronounced in the peripheral half of the vessel, so there is some direct evidence for this mechanism.

Appendix 1

EFFECT OF A PERIPHERAL CELL-FREE LAYER

The thickness t of a stationary cell-free peripheral layer of conductivity σ₂ surrounding a core of flowing blood of conductivity σ₁ is given by

\[ t = \frac{1}{U^2 - 1} A \left( \frac{\sigma_2}{\sigma_1} - 1 \right) \]

where \( U \) is the ratio of the transducer response when the layer is present to that when it is not and \( A \) is the tube radius of the transducer (1). It follows from the use of equation 1 and of
conductivity data that the change in $U$ corresponding to 10 $\mu$ in $t$ is 0.06% at 9.5% hematocrit, 0.55% at 49% hematocrit and 1.13% at 66% hematocrit. In fact, any cell-free or cell-poor peripheral layer which existed would not be stationary, and the more exact equation relating the tube radius $A$, the inner radius $a$ of the layer, the mean fluid velocity $V$ in the tube and the mean fluid velocity $V_a$ in the layer can be shown to be (Wyatt, unpublished)

$$U = \left[1 - \frac{1}{2} \left(1 - \frac{r}{\sigma_1} \right) \left(1 - \frac{a^2}{A^2} \right) \frac{V_{A-a}}{V_A} \right] \cdot \frac{2}{2 - \left(1 - \frac{r}{\sigma_1} \right) \left(1 - \frac{a^2}{A^2} \right)}$$

(2)

If the flow is everywhere laminar and the fluid has viscosity $\mu_2$ in the layer and $\mu_1$ elsewhere,

$$U = \left[1 - \frac{1}{2} \left(1 - \frac{r}{\sigma_1} \right) \left(1 - \frac{a^2}{A^2} \right)^2 \right] \cdot \frac{2}{2 - \left(1 - \frac{r}{\sigma_1} \right) \left(1 - \frac{a^2}{A^2} \right)}$$

(3)

If a cell-free layer formed, it would lead to a hematocrit value ($H_c$) in the central core different from that in the whole blood ($H$) given by

$$H_c = \frac{A^2}{a^2} H.$$  

(4)

In addition, the end-shorting correction of the transducer would be affected by the presence of a peripheral cell-free layer. The net result of these corrections is not significant in the present context. For example, with blood of hematocrit value 49%, neglect of flow in the layer would lead to an underestimate of the layer thickness by 10 $\mu$; neglect of the hematocrit correction (equation 4) would cause an overestimate of layer thickness of 6 $\mu$; and neglect of the end-shorting correction (Wyatt, unpublished) would cause an over-estimate of 3 $\mu$. The net error would thus be an underestimate of 1 $\mu$, or 1.5%. At other hematocrits, the percent error would be about the same.

**Appendix 2**

The time $t$ let the flow generator displace fluid at a rate $Q_t$. Let the magnet current be $i_t$, and the flowmeter interelectrode potential $\Sigma_t$, where $\Sigma_t = ki_t Q_t$ and $k$ is the constant we wish to determine. Let the instantaneous balance voltage from the resistance network, operating in the amplifier input circuit, be $E_{t}$, and the amplifier gain $a_t$. The amplifier output is

$$e_t = a_t \left( ki_t Q_t - E_t \right).$$

Let $E_{t} = \theta \beta R_{t,i_t}$,

where $\theta = [1 + 0.2R_2(10 - R_2)/(R_4 + R_4' + R_3 + R_3')]^{-1}$

$$\beta = \frac{R}{130} (R_4 + R_4')/(R_4 + R_4' + R_3 + R_3').$$

The meaning of the symbols is given in Figure 2. Integrating throughout we obtain

$$k = \frac{\beta}{T} \int_{0}^{T} \frac{1}{Q_t} \int_{0}^{T} e_t \, dt.$$ 

Let the nominal flow rate, when the synchronous driving motor is supplied from the line at 50 cps, be $Q$. Then

$$\frac{1}{T} \int_{0}^{T} Q_t \, dt = \left( \frac{f}{50} \right) Q,$$

where $f$ is the recorded line frequency, assumed constant during the period $T$.

The value of $Q$ is dependent upon the mean value of the piston diameter during the period of integration, but this is always the same since the integrator is switched on and off by microswitches fitted to the flow generator. Since the second term in square brackets is $\sim 1\%$ of the first term, we can treat $a_t i_t$ as constants. (They each vary by less than 1% over a period of 1
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hour.) Then

\[ k = \frac{BT}{Q} \left( \frac{50}{T} \right) \left( \theta R_2 + \frac{1}{\beta T_{ai}} \int_0^T e_t \, dt \right). \]

When \( Q = 250 \) ml/min the value of \( T \) was 91 seconds, thus

\[ T = \frac{250}{Q} \times 91 = \frac{22700}{Q}. \]

The integration was carried out with a simple R-C circuit of value 1150 seconds giving

\[ \int_0^T e_t \, dt \approx VRC \left( 1 + \frac{T}{2RC} \right) = VRC \left( 1 + \frac{9.8}{Q} \right). \]

where \( V \) is the integrated voltage in millivolts, and \( \alpha \) is a calibration constant equal to \( CR/\beta a i \). The value of \( \alpha \) was determined by observing \( V \) in millivolts over a timed period \( T \) with no flow \( (k = 0) \), and with \( Q/22700 \) replaced by \( 1/T \) in the equation; the value of \( R_2 \) used was 0.1 kilohm. \( R_1 \) and \( R_2 \) were adjusted together so that \( R_1 + R_2 = 10 \) kilohm. The value of \( \beta \) could be determined by feeding into the amplifier input (in lieu of the flowmeter) a voltage \( E \) derived by passing the magnet current \( i \) through a potentiometer made up of 0.01% resistances. The value found was 1.691.10^{-3}. The relative sensitivity \( S \) is conveniently given by

\[ S = \frac{250k}{\beta}, \]

\[ = \left( \frac{250}{Q} \right) \left( \frac{50}{T} \right) \left[ \theta R_2 + \frac{\alpha VQ}{22700} \left( 1 + \frac{9.8}{Q} \right) \right]. \]

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References


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