A THERMOELECTRIC PROBE FOR TISSUE BLOOD FLOW MEASUREMENTS.

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M. Sc. (Eng.)

AFFIDAVIT

I hereby empower the University of Capetown to produce for the purpose of research either the whole or any portion of the contents of this thesis in any manner whatsoever.

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Section 1

TITLE: A THERMOELECTRIC PROBE FOR TISSUE BLOOD FLOW MEASUREMENTS.

SECTION 1.A.

SUMMARY.

A study was made of a method of determining blood flow velocity in vascular tissues by relating the difference in power dissipated by a heated thermistor placed in the vascular tissue matrix to the flow velocity of the blood past the thermistor.

A detailed theoretical model, based on the general solution to this forced convection heat problem, was evolved and results predicted were verified by means of practical experimentation. It was concluded that the above relationship is of the form $P = K\Delta T \ (A+B\sqrt{V})$ and the relationship is valid for laminar flow in tubes and vessels but the validity is severely reduced for tissue flow because of the heterogeneity and complexity of the flow on a macroscopic scale in tissues.

A flowmeter based on this principle was built and details of the probe and circuit design were discussed. An appraisal of the performance of the flowmeter and guidelines for further research were included.

/... SECTION 1.B.

SECTION 1.B.

INTRODUCTION. The measurement of flow in the circulatory system presents a most challenging problem in instrumentation. A knowledge of the quantities of blood flowing per unit time or even a knowledge of the relative changes of the flow rate through vessels, organs, tissue and bone is of tremendous value for the assessment of the effects of drugs, physical exercise, physiological stimuli and changes of various kinds.

Unlike the measurement of pressure in the circulatory system, flow measurement is complicated by the great differences in the physical nature and location of the components of the circulatory system, which impose severe constraints on the size and shape of transducers.

There are many methods in use for determining flow rates or velocities of blood and among these are a few which show great potential as highly practical tools. The electromagnetic flowmeter and the ultrasonic flowmeter for instance, although severely limited in a number of ways, have been developed and refined and several commercial models are available.

This work stems from the urgent need for a device which can measure the velocity of flow of blood through organs and tissue in particular. An instrument claimed to measure tissue and vessel flow qualitatively was built prior to this work. It was based upon the principle of the hot-wire anemometer so useful for determining fluid and gas flow rates. The instrument consists of an alternating current excited Wheatstone bridge with thermistor elements, a high gain carrier amplifier and detector, and a metered output.

Several attempts to obtain consistent

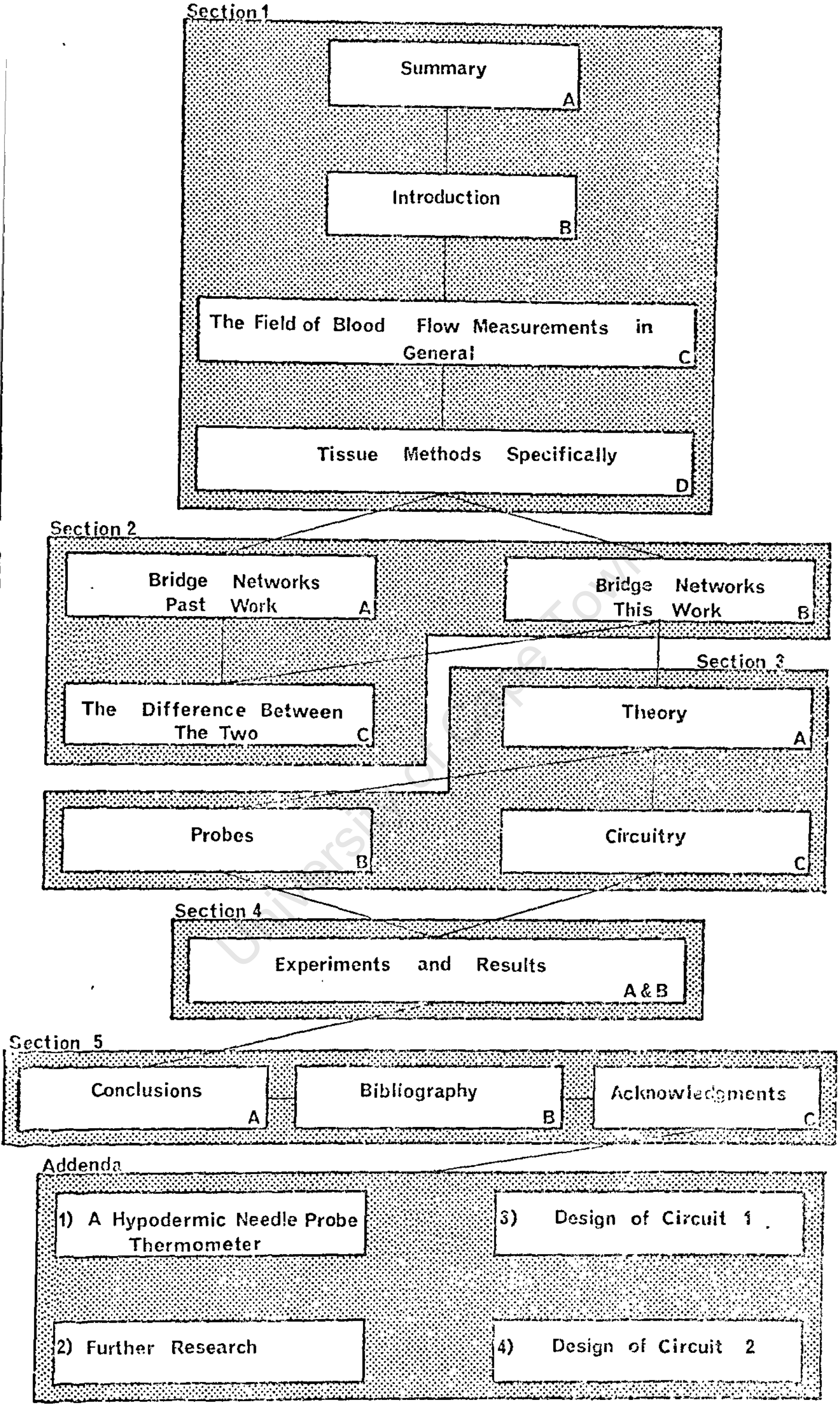
/... results

results with this instrument under laboratory conditions of simulated flow were unsuccesful. Hence, it was decided to further investigate the validity of the method and, if proved to be viable, to modify, improve or redesign the system in order to produce a reasonably reliable means of measurement.

The principle of the flowmeter is as follows:-

A thermistor sited at the tip of a hypodermic needle and located in the flowing medium, is self-heated by a fixed current. The temperature to which it is raised by virtue of this I^2R heating is a function of the velocity of the flowing medium. The temperature of the thermistor is measured and related to the flow velocity by means of a calibration.

Figure 1.B. best illustrates the logical development of the project. It serves as a guide to the reader and has been designed to minimise the frequency of cross-references. In several cases the sub-sections are independent and refer upwards only.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1.B.

SECTION 1.C.

THE FIELD OF BLOOD FLOW MEASUREMENTS IN GENERAL.

1.0. OBJECTIVE. In order to gain a background and perspective of the subject, a study was made of the methods used to date for the qualitative and quantitative assessment of tissue, vessel, organ and bone blood flow. Emphasis was laid on the electronic and thermo-electronic methods.

1.1. ELECTROMAGNETIC FLOWMETERS.

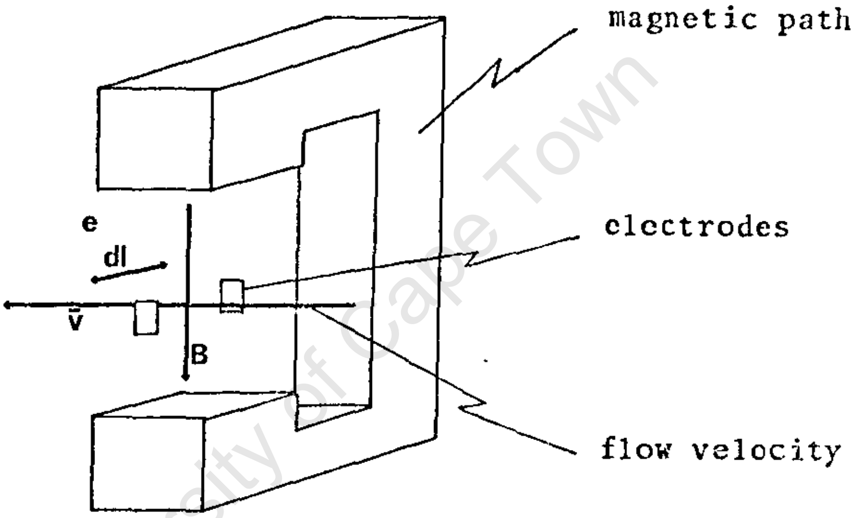
1.1.1. INTRODUCTION. Electromagnetic flowmeters have found use in many widely differing applications. Common uses include the measurement of ionised gas flow and measurement of the flow of a variety of electrically conducting fluids - sewerage pumping stations are often controlled by the data from electromagnetic flowmeters. Only recently have they found the particular application in blood flow measurement. This has largely been due to the work of Kolin [1] Clarke [2] Wyatt [3] Beck [4] Ferguson [5] Landahl [6] Gessner [7] Mills [8] Shercliffe [9] Spencer [10] and Dennison [11]. The instruments have been refined from an engineering point of view and qualitative measurements of pulsatile and average blood flow through vessels can be made. Accuracy is of a high order - quoted in sales literature to be 5% to 10% of full scale deflections.

1.1.2. METHOD. The method relies upon the fact that a conductor of length l moving with a velocity $\bar{v}$ in a magnetic field of uniform and constant magnetic intensity B, generates along its length a potential difference e according to the mathematical expression:

$$e = \int_{0}^{l} \bar{v}. \ B dl$$

/... The

The blood flow represents the moving conductor and it is of interest to note here that the conductivity of blood remains substantially constant under all normal physiological and biochemical changes. However, it does change with velocity [12] and hematocrit value. [13] (the red blood corpuscle concentration per unit volume). Another interesting point is that this process is reversible! Compensation can be made for the latter variation to an extent which preserves the accuracy of the method.



The theory has been found to apply irrespective of the velocity profile and is equally valid for both laminar and turbulent flow. Turbulence in the circulatory system is found only at the root of the Aorta [14] . For reasons of preservation of electric field symmetry the flow must have circular symmetry. This is true for all vessels other than the small capillaries at the normal viscosity of blood.

The instruments are high gain carrier amplifiers. Noise problems have been a constant cause of refinement of the technology. The standard arrangement is to have alternating current field excitation and quadrature signal

/... rejection

rejection in the voltage delecting circuitry thus eliminating the obvious problem of cross-talk from the excitation network to the detection network (excitation voltage is in quadrature with magnetising current and magnetising current is in phase with voltage detected.) This problem becomes quite acute when probe size decreases and this, together with the familiar problem in high gain low frequency amplifiers of baseline drift and low frequency noise, has led to experimentation with the excitation waveforms.[15] One of the most recent methods uses squarewave excitation and appropriate gating of the detection circuitry to almost totally eliminate the cross-talk problem [15]. Developments such as this one have enabled the construction of very small non-ferrous magnetic path probes allowing catheterisation of probes.

There are three different types of probe currently available for use with the electromagnetic flowmeters. These are:

1) Peri-vascular. As the name suggests these are of the Cuff-type, Clip-on-type and U-type. The larger ones are provided with slot pieces for completion of the magnetic path. These probes are used on vessels which range from 3mm o.d. to 20mm o.d. The electrodes which make contact with the outside of the vessel wall, are made of gold or platinum.

2) Cannular. Surgery is required in order to use this type as the vessel must either be passed through the probe or fitted over each end of the probe. They normally have the same diametric range as the perivascular probe.

3) Catheter. Commercial types 3mm o.d. to 1.5mm o.d. are currently available. Their construction is delicate and complicated and hence is highly costly. They are difficult to position correctly but have a small added

/... advantage

advantage that provision is always made for simultaneous blood pressure measurement.

1.1.3. EVALUATION. The electromagnetic flowmeter is limited to the measurement of flow in vessels ranging from 3mm o.d. to 20mm o.d. and, in the experience of the writer and others who have used these meters, it has a reasonable accuracy and stability only above 1 cm/sec which corresponds to minimum volumetric flow rate of 17 ml/min and an input voltage of the order of 1 microvolt. This is somewhat restrictive.

The perivascular probes are easy to work with, - the catheter and cannular types very much more difficult. The instrument, once having been initially set up, is easy to operate. Great care must, however, be exercised in earthing.

The greatest single advantage of the electromagnetic flowmeter is its high frequency response $\leq$ 500 Hz. It is also direction sensitive and so is most valuable in cardiac research for waveform observation. It is a linear device with an arbitrary scale reading and has provision for external monitoring.

The disadvantages include sensitivity to vessel wall composition, external electromagnetic fields and physical movement of the probe. These, in fact, are found to be so frustrating that in many cases the instruments are not used.

Some preliminary experiments have been carried out using extra-corporeal magnetic fields and probes located in tissue for tissue flow experiments [5]. These have met with very little success. The discreet differences in electrical conductivity of blood, tissue, fat and bone

/... together

together with the difficulty of forming uniform magnetic fields across the large gaps required seems to be the problem.

1.2. ULTRASONIC FLOWMETERS.

1.2.1. INTRODUCTION.

As in the case of the electromagnetic flowmeter, the ultrasonic flowmeter has found use in a variety of industrial applications. It was first used on laboratory animals in 1961 by Franklin Schlegel and Rushner [16]. It has evolved and several commercial models are available. Until recently these had been limited to transcutaneous measurement only and usually only comparative flow for clinical purposes to evaluate results of remedial action for peripheral, arterial and venous disease [17]. Experimental models of probes of the perivascular and cannular types have been made and found to be reasonably successful. This technique warrants further scrutiny.

1.2.2. METHOD.

The method relies simply upon the Doppler Effect. A Piezzo-electric crystal of a fixed natural frequency (of the order of 10 MHz in the commercial varieties) is excited externally and focussed at an acute angle against the direction of blood flow. A detector crystal is focussed so that it forms a virtual detection point with the exciter crystal, large in volume (because of deliberate low resolution) below the surface of the skin. The frequency of the detected signal is decreased by the velocity of the blood flow according to the following reasoning:

If f_e is the frequency of the exciter crystal V is the propagation velocity of the ultrasonic signal in the medium and v is the velocity of the blood flow, then

/...the

the detected frequency f_d is given by the expression

$$f_d = f_e V/V + v$$

In the earlier flowmeters the signal frequency $f_d - f_e$, arranged to fall in the audio range, was amplified and applied to a loudspeaker. It is claimed that clear tones are heard corresponding to Systole and Diastole [18]. Venous flow, because of its lesser phasic nature, is more difficult to record, being a distribution of frequencies centred around a mean frequency. Frequency to voltage conversion circuitry is used for recording purposes. Probes available are totally encapsulated transmitter/detectors and are applied to the surface of the skin with an aqueous coupling gel.

1.2.3. EVALUATION. The ultrasonic flowmeter does not suffer the severe limitations of the electromagnetic flowmeter in so far as physiological variables and external interference is concerned. There is a remote possibility that it could be used in tissue flow although filtering and interpretation of results would be a formidable task. Because of the unbounded detection area, volumetric results are impossible.

In its present form for transcutaneous measurements it is a highly practical clinical tool but if it is used for venous flow measurement it will probably suffer similar advantages and disadvantages as the electromagnetic flowmeter.

1.3. THERMODILUTION.

1.3.1. INTRODUCTION. Although severely limited this is an important and useful method of measuring average flow. It is normally only used in measurements of a specialised

/...nature

nature and has not, as far as the writer knows, found common application in industry. It was first used in 1960 [19] for coronary sinus blood flow measurements in dogs. There are no commercial "kits" available. The method has so far only been used in vessels and in heart chambers for estimating stroke volumes and charge and discharge rates. It is doubtful whether it could be used in tissue.

1.3.2. METHOD. The technique involves the injection of a thermal indicator (saline is normally used) with a fixed specific heat S_i and density d_i in a fixed quantity Q_i at a temperature T_i (room temperature normally) - into the blood stream. The indicator is injected at a rate high enough to cause homogeneous mixing of the indicator and the blood but low enough for its momentum not to substantially affect the flow velocity of the blood.

For thermal balance:-

(subscript b means blood and subscript m means the mixture of the indicator and the blood).

$$Q_b = d_i \ S_i \ Q_i \ (T_m - T_i) \ / \ d_b \ S_b \ (T_b - T_m)$$

$$= KQ_i \ (T_m - T_i) \ / \ (T_b - T_m)$$

T_i, T_m and T_b are monitored by thermistor elements sited at the injection orifice. K varies with the hematocrit value of blood but predictably and, hence, can be compensated for in computation. An analogue computer for the instantaneous solution of this equation has been devised. It features accepted subtraction, division and multiplication circuits and sample and hold circuits for storing T_i and T_b pre computation.

It was highly successful for measuring Aortic flow in dogs. Probes differ quite radically in design but all

/... are

are of the catheter type. The probe of Roberts[20] was constructed as follows:-

Multilayer walls were used to insulate the indicator from the warmer surrounding tissue. Spacing of the tubes was achieved by spiral winding of the thermistor leads between the walls. The indicator was passed down the centre tube under high pressure, hence mechanical strength of the probe was a very important consideration. The mixing orifice was sited a short distance from the end of the probe and was bored at an angle of 30 to 40 deg. against the direction of blood flow and for hydrodynamic reasons was required to have a particular length to diameter ratio to achieve the important homogeneous mixing. The probe was constructed of Nylon. The ratio of catheter diameter to vessel diameter has to be kept small so that the flow rate at the mixing site is not greatly affected by the presence of the catheter.

1.3.3 EVALUATION.

This method has three major disadvantages.

1) It can probably only be used for large vessel flow.
2) It has a low frequency response and can only be used for average flow.
3) It can only be used in controlled experiments.

It is most certainly not a practical tool but it does have the advantage of high accuracy claimed to be ± 5% of actual flow.[21] It can only be used for flow rates in excess of 20ml/min in large vessels > 7mm i.d.

1.4 IMPEDANCE METHODS.

1.4.1. INTRODUCTION.

As long ago as 1937 experiments were performed using two closely spaced platinum or gold electrodes in vessels to correlate the resulting impedance

/... variations

variations with flow velocity [22]. Results were highly variable and no useful information was obtained.

There has recently been renewed interst in the method and while it is still very much in the experimental stage some promising results have been reported [23].

1.4.2. METHOD. It has been found generally that a relationship of the form

$$Z = f(\sqrt[3]{v})$$

exists where Z is the interelectrode impedance and v the velocity of the blood. There are a number of different methods used to measure Z but they are usually alternating current excitation of Wheatstone bridges where frequencies in the high audio spectrum are used. The bridges are often arranged to be periodically rebalanced by means of servo-mechanisms or feedback networks so as to compensate for long time variations in the conductivity of blood and differing inter-electrode spacing.

One good recent paper [24] described a method using one electrode in the vessel and another electrode on the surface of the skin. The electrode placed perpendicular to the vessel through the wall was made long enough to cancel the velocity profile effects. An interesting theory offered in this paper was that the region of ionisation surrounding the red blood corpuscles increases and decreases its thickness depending upon the velocity of the blood. A red blood corpuscle is a depressed sphere. This might help to explain the cubed root relationship.

1.4.3. EVALUATION. This method is still in its infancy. There has been no agreement in regard to mechanisms involved in the change in the conductance properties of blood in motion. Some work is in progress at Groote Schuur Hospital

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and reasonable results have been claimed for bolus blood movement in the chest.

If there should prove to be any validity in this method it will show great promise for blood flow measurement in general. It would have very few limitations, be highly practical and should prove to be valid for arterial, venous and tissue measurements.

1.5. OTHER METHODS.

In the past many methods have been developed and used for specialised measurements of blood flow. A number of these rely upon the washout of inert gases, radio isotopes and diffusible indications of various types. Differential pressure measurements have been used for some time now and probes of very small dimensions have been constructed. These methods are all slow and impractical for use during surgery or in the clinic and require long and involved observation techniques. Most of these methods, however, are very useful for tissue flow measurements.

1.6 BRIDGE NETWORKS

1.6.1. INTRODUCTION.

The hot-wire anemometer has long been in use for the measurement of flow of gases and liquids and there are many industrial applications. Commercial flowmeters are in common use by air conditioning design engineers and for process control and a number of varieties are available.[26] As early as 1933 [25] a thermo-electric blood flow recorder in the form of a needle was described and was used for comparative measurement of the flow in the jugular veins of dogs. The writer knows of only one commercial instrument designed specifically for blood flow measurements.[27]

1.6.2. METHOD.

The method relies upon the convective cooling of a heated element suspended in the flowing medium. The theory may be simply stated as follows:-

In general $T = f(v^x)$

Where T is the temperature of the heated element and v is the velocity of the flowing medium. The relationship depends upon the physics of the heated element, the velocity of the fluid stream, the physical properties of the fluid and hence is different for different applications.

The industrial instrument consists basically of a thin (0.001 to 0.005" diameter) electrically heated wire stretched across the ends of two prongs. When the wire is exposed to the fluid stream it loses heat principally by forced convection. The temperature of the wire which is measured, with all other variables fixed, depends upon the velocity of the fluid stream. Probes are either fixtures in a system or can be introduced at a particular point. The temperature element for the blood flow recorder consisted of a heated thermistor pair encapsulated in the end of a hypodermic needle.

1.6.3. EVALUATION. The evaluation of this technique is of necessity carried throughout this work and shall, therefore, not be treated specifically at this stage. It will serve, however, to state that it shows the greatest possibilities for the measurement of tissue blood flow.

1.7. RESUMÉ. The objective was achieved - the writer was introduced to all the published techniques for the measurement of blood flow and gained a broad understanding of all the aspects, requirements and difficulties of the problem.

SECTION 1.D.

TISSUE METHODS SPECIFICALLY.

1.0. INTRODUCTION. From the above introduction to blood flow it can be seen that of all the methods in use, there are probably only three of these which could be used for tissue flow measurements. Washout of indicators is a slow impractical method. Impedance methods show great promise but there is still a great deal of groundwork to be done before any useful work for its application to tissue measurements could be realised. This leaves bridge networks which is the subject of this work.

1.1. WHAT IS REQUIRED? To broadly say that it is required to measure tissue blood flow is rather an illusive statement. One tends to ask why and for what purpose. Most organs and tissues have one or more input and one or more output which supply and deliver blood. Is it not sufficient to make measurements at these points? The answer to this question is two-fold.

a) It is often not possible to make measurements on the inputs and outputs of organs, muscle and tissue and with the crude tools at the disposal of the physician it is most certainly not possible in the clinical situation, because of the requirements of access to these remote locations. However, if the right tools were available it would be possible for practical spot comparative tissue measurements to be made.

b) In all organs, muscle and tissue it would be of enormous value to be able to make a field plot of the topography of flow.

In particular, for instance, it was hoped that this work would result in a device which could be used to measure relative flow of blood in the uterine wall (which is a highly vascular tissue) during pregnancy and, in particular, relative flow in the region of the placenta through the course of childbirth and to observe the flow event sequence.

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Unfortunately different tissues have vastly different physical properties and flow rates so it cannot be hoped to obtain anything other than pure qualitative results from any device based on the principle of the bridge network method. However, even qualitative results would be highly valuable.

1.2 THE CIRCULATORY SYSTEM. Electrical engineers have a habit of converting all other phenomena into electrical analogies and the circulatory system is no exception. The system may be considered to be a highly complex hybrid transmission line consisting of both distributed and lumped parameters. The voltage source generates a peculiar waveform which is propagated down this transmission medium into the multifarious distributed and lumped impedances. The system can be completely specified at any point in the network if the two variables, pressure and flow can be measured. The theory of blood flow is being formalised and the work encountered in this study was that of Attinger[28] and Evans[14]. At this stage little is known about the flow of blood through vascular beds because of the complexity of its geometry and the non-uniformity of its physical properties.

1.3. SOME PROPERTIES OF TISSUE. Some very thorough statistical work on the thermal properties of biological fluids has been done by Spells [29] and listed below are the thermal conductivities of the biological substances of particular interest:

Human blood	$12.1X10^{-4}$	cal/cm/sec/deg C
Normal deviation ±	$2.0X10^{-4}$	cal/cm/sec/deg C
Human muscle	$10.5X10^{-4}$	cal/cm/sec/deg C
Normal deviation ±	$2.0X10^{-4}$	cal/cm/sec/deg C
Human fat	$4.77X10^{-4}$	cal/cm/sec/deg C

(all measurements at 37°C)

/... The

The percentage of water contained in blood and tissue by volume as listed by Best and Taylor [30] is:

Human blood	83%
Human muscle	75%
Human fat	20%

From this it can be seen that the thermal conductivities of blood and tissue are very close. The average percentage composition of muscle tissue by volume is 83% blood and 17% tissue by volume and hence it is reasonable to assume that if we look at a macroscopic cube of, say. 1mm in side dimension in pure tissue with no fat and further, that on this scale it is highly vascular, then at least half of this volume will represent blood diffusing in random directions. The diffusion velocities will be very small but if vectorially summed could be represented by a net flow in one direction at a fixed velocity. There is no doubt that this velocity will differ depending upon the location of the macroscopic cube but the physical properties of the cube will vary by at least an order of magnitude less. The problem is to isolate this variable and measure it.

Section 2

SECTION 2.A.

BRIDGE NETWORKS - PAST WORK.

1.0 INTRODUCTION. The purpose of this section is to present a summary of past work using this method specifically and to indicate the salient advances that have been made.

1.1. THERMOCOUPLES. As early as 1933 Gibbs [25] described a blood flow recorder in the form of a needle which consisted of an iron constantan thermocouple heated by passing a current through the constantan wire. Comparative measurements of the flow in the jugular vein of dogs were made and a few experiments performed in tissue - such as observations of the effects of intravenous injection of adrenalin on the flow in the cortex of a cat's kidney and changes in temporal muscle flow produced by the stimulation of sympathetic nerves. He used no temperature compensation but suggested that a temperature reference thermocouple be placed in adjacent tissue. He claimed that satisfactory sensitivity could be obtained without raising the temperature of the thermocouple above 2°C above the body temperature. It must be noted, however, that his calibration in-vivo, he admits, allows for a needle temperature rise of 10°C above body temperature.

An examination on a physical basis of a logical development of Gibb's recorder was performed by Grayson [31] in 1952 following previous experimentation in 1951. His basic objective was not to obtain experimental results of a medical nature but to theorise the method.

Grayson established an equation of the form

$$I^2R = k\Delta T$$

/...where

where

R is the resistance of the heating element.
I is the current through the heating element.
ΔT is the temperature of the heated junction.
k is the thermal conductivity of the surrounding medium.

The investigation of this equation was performed by using substances with differing values of k, holding ΔT constant and measuring I^2 required to achieve this. The linear relationships he obtained are shown below:-

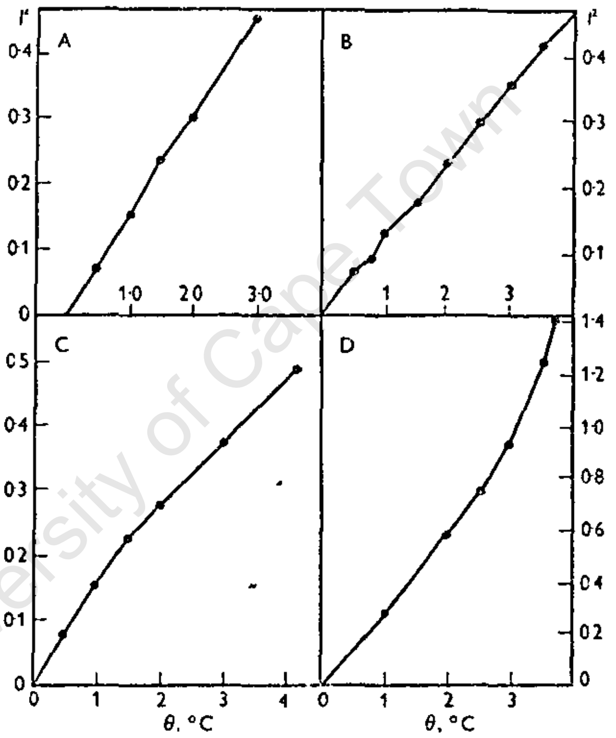


Fig. 3. The relation between I^2 and θ. A, recorder embedded in gel (10% gelatin in water); B, recorder in dead liver (implanted during life); C, recorder in inadequate quantity of gel (1 mm below surface); D, recorder in water, convection effects.

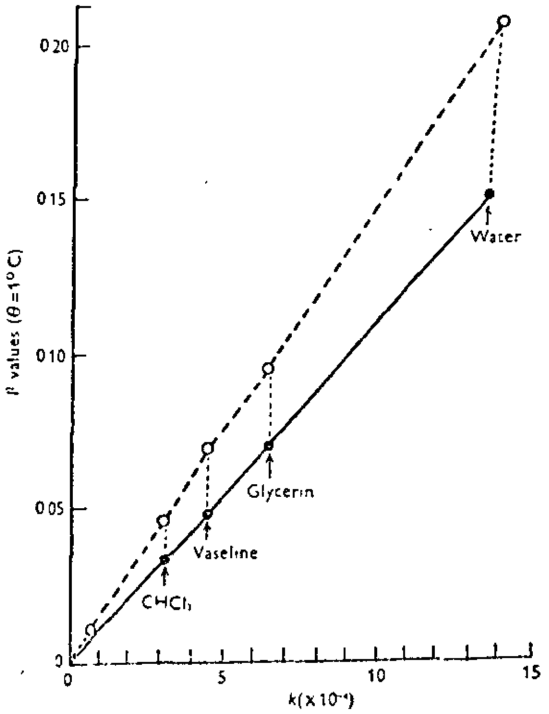


Fig. 4. The relation between I^2 and thermal conductivity (k). Interrupted line, needle recorder; continuous line, standard recorder as used for implantation.

Grayson maintained further that the thermal conductivity of the surrounding medium was raised by the presence of flow and he established that "the thermal conductivity increment" which he called it was linearly related to volumetric flowrate. His results are shown below:-

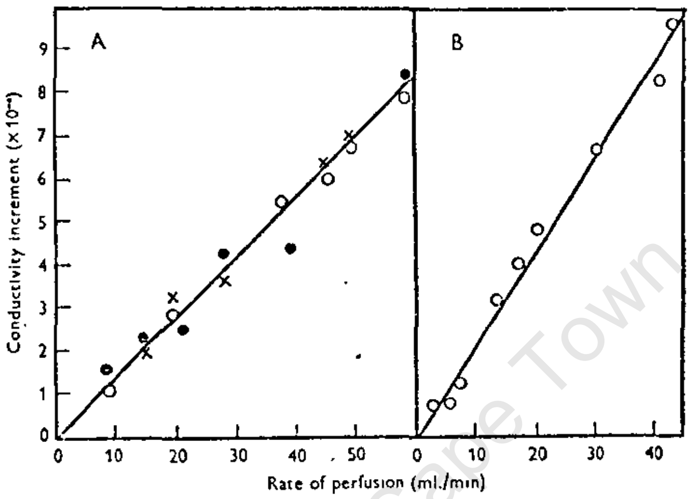


Fig. 11. Relation between conductivity increment and rate of flow. A, sheep spleen perfused with saline, three experiments; B, sheep kidney perfused with whole blood from same animal.

He also described a method of standardisation of probes by submerging them in fluids of known thermal conductivities. This is the only report of a linear relationship between the square of the current and the flow velocity.

Linzell[32] in 1953 followed up the work of Grayson and established that the variation of I^2 with flow velocity was indeed not a straight line but a curve which approximated three straight lines as shown below:-

$$\left.\begin{array}{l} v \quad \alpha \quad I \\ v \quad \alpha \ \sqrt{I} \\ v \quad \alpha \quad I^2 \end{array}\right\downarrow$$

for increasing flow rate.

/...His

His results are shown below:-

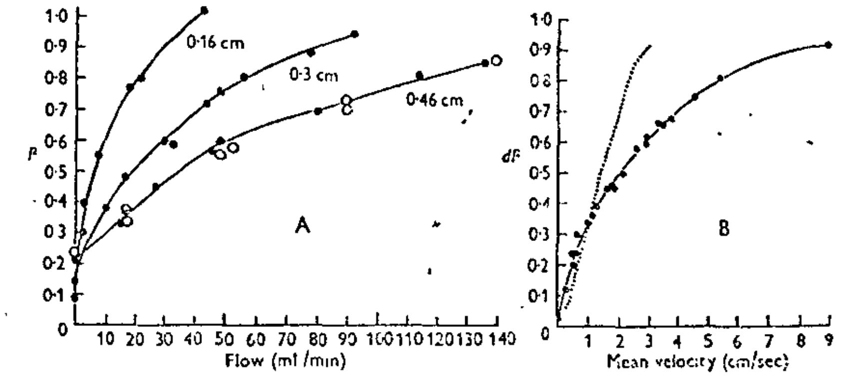


Fig. 3. A: the relation between I^2 ($\theta = 1°$ C) and the flow of water through glass tubes of the radii indicated. ⊙, steady flow; ○, pulsatile flow. B: the calculated mean velocities from all three tubes plotted against the increase in I^2 over that at zero flow. The dotted curve is obtained by plotting dI^2 against the square root of the mean velocity.

Linzell was critical of Grayson's findings and claimed that, under the conditions of his experiments, the apparent local blood flow as indicated by the measured increase in thermal conductivity could not be used as an accurate measure of total blood flow.

Mowbray [33] in 1958 modified the thermocouple and placed it in a hypodermic needle of 12 s.w.g. The end of the needle was sealed with solder. Results claimed were good but it is thought that the raised thermal inertia of the soldered end tended to dampen random fluctuations experienced earlier.

In all the work thus far encountered, tolerance on quantitative results were no better then 20%. It was only by virtue of the fact that high flow rates were used, that even this degree of consistency was obtained. All the while, however, experiments were being performed where the chief interest was results rather than technique, and so the thermocouple method, identical in all respects to that of Gibbs was finding valuable use.

/...In

In 1961, Bill[34] took a theoretical look at validity of the method for tissue flow. In the past, many interesting model organs were proposed for investigating tissue flow with little success or follow-up. However, that of Bill's is rather novel.

He performed a number of experiments in blocks of gelatin with polyethylene tubes running through them.

The basic equation used was

$$K_o = aI^2/\Delta T$$

where K_o was the thermal conductivity of the needle surrounds i.e. gel. He defined ΔK_o as the apparent increase in thermal conductivity and found that for an isolated polyethylene tube in gelatin

$$\Delta K_o = f\ (v) = f\ (\text{o.d. of the tube})$$

The results are shown below:-

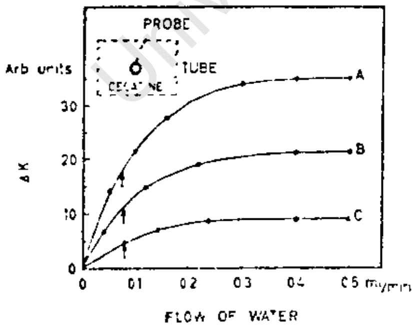


Fig. 2. The relationship between ΔK and flow of water. The model used is shown inset in the figure. The probe was placed close to and in parallel with the polyethylene tube, which had an inner diameter of 0.36 mm, wall thickness 0.20 mm. Curve A was obtained when there was no extra layer of polyethylene added to the tube, curve B when one extra layer, thickness 0.20 mm, had been added to the tube wall, and C when two extra layers had been added. Arrows at value f_{50}.

Asymptotes were reached at a value of flow of the order of 5ml/min. He also defined a value of $f_{50} = f(v)$ at the point where ΔK_o = 50% of its overall change in value and related f_{50} to the inside diameter of the tube.

/...A

A very useful set of results with a number of tubes in gelatin showed that

$$\Delta K_o = \sum_{m}^{n} (\Delta K_m + \Delta K_n)$$

i.e. the apparent increase in thermal conductivity was equal to the sum of the apparent increases due to each tube taken in separately. Bill concluded that there was no standard type of relationship between ΔK and the flow in any tissue into which the probe is blindly introduced and a standardisation has to be made in each experiment if more qualitative information is sought.

1.1. THERMISTORS. The next major step was taken by Juhasz[35] in 1967 and Grahn[36] in 1968. The work of Juhasz was based upon the following analysis.

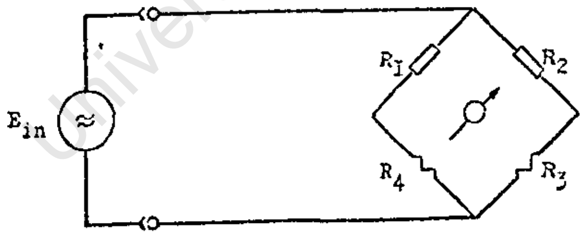


Fig. 1.

Assuming a constant excitation voltage E_{in}, the functional expression for the velocity sensor R1 and for the compensation sensor R2 can be written as follows:

$$R1 = f\ (\Delta T\ R1;\ P1)$$

$$R2 = f\ (\Delta T\ R2;\ P2)$$

where P1 and P2 represent the self dissipation

/...of

of the semiconductors and ΔTR1 and ΔTR2 represent the resistance changes due to the temperature variations. When R3 and R4 are made equal the balance condition for the bridge is expressed as

$$F\ (\Delta T\ R1;\ P1) = F\ (\Delta T\ R2;\ P2)$$

Assuming both semiconductos experience identical temperature influence then ΔTR1 = ΔTR2 and hence

$$F\ (P1) = F\ (P2)$$

It can be seen that, in the case of matched semiconductor pairs, the balance point for the bridge is mainly by the dissipated power of the individual semiconductors.

The flow probe, a hypodermic needle containing two thermistors, is shown below in electrical schematic and physical layout:

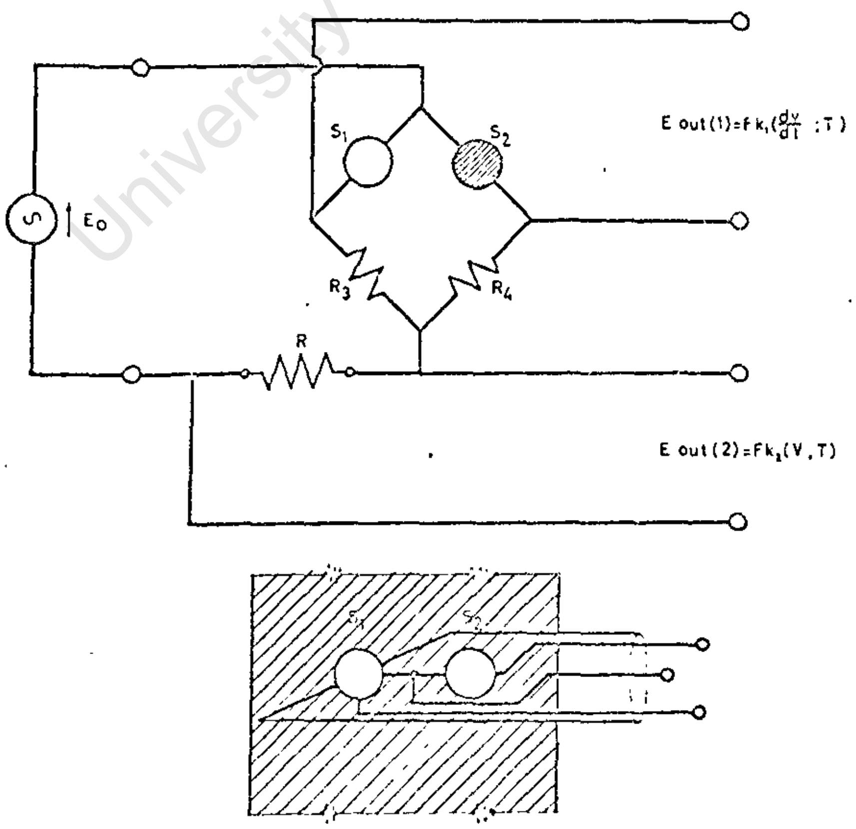


Fig. 8. Thermistor-flow transducer.

In general the operating principle of the velocity

/...probe

probe was based upon the thermal balance between the velocity sensor R1 and the compensating sensor R2 where the semi-conductors have equilibrium time constant t1 and t2 respectively, - the difference in equilibrium time constants being caused by the exposure of the flow sensor to the flowing medium and the shrouding effect of the needle and its encapsulation to the temperature sensor.

The argument is that sensor 1 will respond to both oscillatory and mean flow and sensor 2 with its greater equilibrium time constant will not respond to flow but will respond to changes of T with respect to time.

However, $R1 = R2$ in this case and $I1 = I2$ so, initially, $P1 = P2$ and a balance exists. In situ the temperature sensor temperature rises to its equilibrium temperature, say T1 and the temperature of the flow sensor rises to its temperature T2 by virtue of the influence of the constant velocity of the cooling blood past it.

$T2 \neq T1$ because of the difference in heat dissipations, therefore $\Delta T1 \neq \Delta T2$ which defies the above analysis. Clearly

$$^{dR1}/_{dT1} \neq {}^{dR2}/_{dT2}$$ since the differential is a function of temperature as will be shown. Hence, temperature compensation cannot work accurately. This statement will become quite clear when the section on theory is read.

The experiments performed by Grahn were of a similar nature. He used three thermistors - the third for directional sensitivity. His probes were catheters and the directional sensitivity achieved by profiling the end so that the third thermistor was sensitive to direction only. He used the method of monitoring the current required to heat the thermistor and hence maintain the balance of the bridge.

SECTION 2.B.

BRIDGE NETWORKS - THIS WORK.

1.0. INTRODUCTION. Having studied the past approaches adopted for this method, it became obvious that no serious attempt was made to scrutinise the theoretical basis of the method, develop a mathematical model and use this as a basis for clearing the confusion and conflicting findings of past work. All too often in a great variety of situations, work is duplicated and even triplicated for lack of information, lack of communication and lack of recognition of fundamental similarities.

Firstly, the forced convection heat dissipation problem - the fundamental concept in this work - has been studied by many mechanical engineers in the past and a great deal of knowledge built up. Secondly, the physical properties of tissue and vessels have been thoroughly investigated and statistically analysed. Thirdly, many of the problems associated with the design of bridge networks have been resolved.

Armed, therefore, with this wealth of knowledge, the following procedure for solving the problem was adopted.

1.1. APPROACH.

(1) To evolve a mathematical model of the heat flow problem - an ideal theoretical model of a heated sphere placed in a cooling fluid stream. Then to modify the ideal theoretical model so that it eventually represents the actual physical case of a heated thermistor encapsulated in an epoxy resin and placed in a tissue cellular matrix.

(2) To investigate the model to predict results and

/...estimate

estimate ideal operating conditions for the thermistor. Also to estimate the effect of parametric spreads and find out the most efficient way to measure, in terms of which parameters to hold constant and which to measure.

(3) To construct a probe which, subject to manufacturing limitations and medical requirements, satisfies to the greatest extent possible the requirements of the modified mathematical model.

(4) To develop circuits which will provide optimum operating conditions for the thermistors as predicted by the theory and to enable the parameters to be changed so that theoretical predictions could be verified.

(5) To set up a series of experiments designed to verify the theory and elucidate possible practical problems so that a practical flowmeter may be developed.

(6) To build a practical flowmeter and assess its utility.

(7) With the above work as a background to establish guidelines for future work.

SECTION 2.C.

THE DIFFERENCE BETWEEN PAST WORK AND THIS WORK.

There were a number of differences between this work and past work. These are summarised below:-

(1) A formal scientific approach to the problem was adopted. The writer knows of no previous work which, used as its beginning, points a mathematical model.

(2) Since the work was not initiated by the requirement for any results specific to a particular medical experiment, a general engineering approach could be adopted and a broad application was envisaged. Accent was placed on the possibility of manufacture of probes and instruments and standardisation.

(3) The ultimate objective was the development of a practical device which could be used in a variety of applications including the possibility of adaption for industrial use.

(4) From the thermistor theory it can be shown that:-

$$^{V}/_{I} = f\ (\Delta T)$$

From the heat flow theory it can be shown that:-

$$VI = f\ (v;\ ^{V}/_{I})$$

Hence it is quite possible to measure any of the variables and obtain a "function of a function" relationship between V, I, VI and $^{V}/I$ and the flow velocity.

In the case of the thermocouple work VI was measured for $^{V}/_{I}$ constant and in the case of the thermistors $^{V}/I$

was measured allowing VI and v to determine its value.

In this work I was held constant and V was measured hence

$$Vk = f\left(v; \ {}^{V}/_{k}\right)$$

with the proviso that ${}^{V}/_{I}$ has a limiting value so that ΔT remains in the region of

$$0 \leq \Delta T \leq 5°C$$

The current through the temperature sensor was also several orders smaller than the current through the flow sensor. This enabled a very much greater temperature compensation accuracy than those previously obtained.

Section 3

SECTION 3.A.

THEORY

1.0. INTRODUCTION. In order to design a thermistor probe and to apply the correct bias conditions to the thermistor a study was made of the forced convection heat dissipation from a heated sphere located in a cooling fluid stream.

In the past, studies of forced convection heat dissipation have been made under vastly differing physical conditions - from large copper heated spheres located in air flow ducts to molten metal shot cooling in oil quench baths and even on a macroscopic scale, cooling of hot dust particles in the atmosphere following nuclear explosions.

1.1. ASSUMPTIONS. Two fairly sweeping assumptions were made in order to simplify the study. These are as follows:-

1.1.1.a) Assumption 1. That the sphere is located freely in the fluid stream without the proximity of any material which will distort the heat flow pattern. This means that all the heat dissipated by the sphere over and above that required to maintain the temperature of the sphere at that of the fluid is freely dissipated to the fluid.

1.1.1.b) Discussion. This is obviously not the case since the thermistor is necessarily partially encapsulated in the resin and sited at the end of a stainless steel hypodermic needle which is itself a good heat conductor. However, it can be shown that there exists an apparent diameter $D_{o\ app}$ such that

$$D_{o\ app} = k(D_o)$$

where D_o is the actual diameter of the thermistor. K is a

/... constant

constant factor and may be found from

a) The thermal properties of the resin.
b) The general geometry of the probe.

The separation between the thermistor and the stainless steel is large enough for the quantity of heat conducted away by the needle to be negligible. More will be said about the factor of k at a later stage.

1.1.2. a) Assumption 2. That the fluid flowing freely past the sphere is at a fixed temperature T_f lower than that of the sphere T_s and that it is homogeneous on the same scale i.e. that the largest fluid particle is at least a factor of 10^2 smaller than the sphere. A further assumption is that its velocity is constant for a period greater than the time constant of the sphere. This, too, will be discussed at a later stage.

1.1.2. b) Discussion. In Section 1.D. some of the statistical physical properties of tissue were discussed and it seems reasonable to assume that the thermistor will be exposed to an average velocity $\bar{v}$ which is the vectorial resultant of n small diffusion velocities $\bar{v}_n$ such that

$$\bar{v} = \sum_{n=1}^{n} \bar{v}_n$$

The effect of tissue in contact with the thermistor will probably only have a small effect on the heat dissipation. Provided that the cell structure surrounding the thermistor does not occupy more than, say, 50% of the surface area of the thermistor then it seems reasonable to assume that this will, in fact, only modulate the value of k by a factor just smaller than unity. If, perchance, the probe is located in fat, however, the theory will no longer hold.

The ambient temperature fluctuations of the blood are very small indeed, probably of the order of 0.25°C max. in 1 minute. This fluctuation will be compensated for adequately in Section 3.C. Circuitry.

Homogeneity is obviously not true in tissue, the properties of which vary radically from point to point. Blood, on the other hand, is homogeneous in comparison with the size of the thermistor. (mean diameter of red blood corpuscle - 7.5μ). [73] Blood coagulation would destroy the validity of the method and is an important consideration as there are few, if any, substances which, when placed in contact with blood in the circulatory system, do not cause an immediate coagulation reaction. (Teflon (R) is particularly "medically inert"). If, however, coagulation is found to present a problem, administration of Heparin reduces the coagulation tendency and does not simultaneously radically alter the thermal properties of blood.

1.2. THE PHYSICS.

Consider a simplified model, a heated sphere placed in a cooling fluid stream. This is an example of forced convection heat dissipation. The quantity of heat dissipated from the sphere under these conditions is given by the equation:

$$q = \bar{h} A \Delta T \qquad 1.0$$

$$\text{cal} = \frac{\text{cals} \quad \text{cm}^2 \quad \text{deg C}}{\text{deg C} \quad \text{cm}^2 \quad \text{sec}}$$

where

q = the heat dissipation of the sphere.

A = the surface area of the sphere.

ΔT = the temperature difference between the surface of the sphere and the flowing medium. - $T_s - T_f$.

$\bar{h}$ = the combined unit surface conductance

$$= \bar{h}_c + \bar{h}_r \qquad 2.0$$

= the average unit convective conductance + the average unit radiation conductance.

/... Examining

Examining the expression 1.0 above in the light of the actual physical case we see that q is the power dissipated by the thermistor which we must assume is convected to the fluid hence:

$$4.186\ q = I^2R \qquad 3.0$$

$$\frac{\text{cal}}{\text{sec}} = \text{watts}$$

A is the effective surface area of the thermistor and is related to $D_{o\ app}$ by the expression:

$$A = \Pi\ D^2_{o\ app} \qquad 4.0$$

$$\text{cm}^2 = \text{cm}^2$$

ΔT is the temperature difference between the blood and the thermistor.

$$\Delta T = T_t - T_b \qquad 5.0$$

$$\text{deg C} = \text{deg C}$$

$\bar{h}$ is rather more complicated. However, there are some simplifications that can be made. The first is that $\bar{h}_r$ is negligible in comparison with $\bar{h}_c$

so

$$\bar{h}_c = \bar{h} \qquad 6.0$$

$\bar{h}_c$ clearly consists of two parts. The first is the static convection and the second the dynamic convection

/... or

or forced convection. So:

$$\bar{h}_c = A + B \qquad 7.0$$

The unit surface convective conductance is a function of the thermal and physical properties of the fluid and the sphere and a number of these variables may conveniently be represented by two dimensionless combinations known as the Reynolds number and the Prandtl number.

The Prandtl number has no velocity dependence and is expressed as follows:

$$Pr_f = \frac{C_p \mu_f}{k_f} \qquad 8.0$$

$$- = \frac{\text{cal}}{\text{gm deg C}} \times \frac{\text{cm deg C sec}}{\text{cal}} \times \frac{\text{centipoise}}{1}$$

$$= \frac{\text{cal}}{\text{gm degC}} \times \frac{\text{cm deg C sec}}{\text{cal}} \times \frac{\text{gm}}{\text{cm sec}}$$

$$= -$$

where:

Cp = the specific heat of the fluid.

μf = the viscosity of the fluid.

k_f = the thermal conductivity of the fluid.

The Reynolds number has a velocity dependence and is expressed as follows:-

/...Re_f

$$Re_f = \frac{D_o \quad \rho_f \ \bar{v}}{\mu_f} \qquad \text{\_\_\_\_\_\_\_\_\_\_\_\_\_\_\_\_\_\_\_\_} 9.0$$

$$- \quad = \quad \frac{cm}{1} \ X \ \frac{gm}{cm^3} \ X \ \frac{cm}{sec} \ X \ \frac{1}{centipoise}$$

$$= \quad \frac{cm}{1} \ X \ \frac{gm}{cm^3} \ X \ \frac{cm}{sec} \ X \ \frac{cm \ sec}{gm}$$

$$= \quad -$$

where:

D_o = the diameter of the sphere.
ρ_f = the density of the fluid.
$\bar{v}$ = the velocity of the fluid.

Now from the work of Cary [37], Kramers [38], Holman [39], Ranz & Marshall [40], and Kreith [41], it was established by careful comparison that for the physical conditions in this study, the dimensions of the sphere and the orders of Reynolds and Prandtl number ($Pr \doteq 2 \times 10^3$ & $Re \doteq 10$) that an equation of the form

$$\frac{\bar{h}_c \ D_o}{k_f} (Pr_f)^{-0.33} = 2.0 + 0.6 \left[\frac{\bar{v} \ \rho_f \ D_o}{\mu_f} \right]^{0.5} \qquad \text{\_\_\_\_\_\_\_\_\_\_\_\_} 10.0$$

holds true. It is surprising to note here that there is only a very small change in the value of the constants and the form

/...of

of the equation for very large changes in the physical properties, dimensions, velocities and thermal properties of the fluid and sphere.

Dimensional analysis of this equation yields the result:

$$\frac{\text{cal}}{\text{sec cm}^2\ \text{degC}} \times \frac{\text{cm}}{1} \times \frac{\text{cm deg C sec}}{\text{cal}} \times \frac{\text{cm}}{\text{sec}} \times \frac{\text{gm}}{\text{cm}^3} \times \frac{\text{cm}}{1} \times \frac{\text{cm sec}}{\text{gm}}$$

$$- = -$$

Substituting 10.0 into 1.0 and adjusting the subscripts we get an equation which represents the actual physical case.

$$I^2R = 4.186\ \Pi\ D_o{}^2\text{app}\ \Delta T \left[\frac{k_b \quad 2.0+0.6\left[\dfrac{D_o\ \text{app}\ \bar{v}\rho_b}{\mu_b}\right]^{0.5}\left[\dfrac{C_p\ \mu_b}{k}\right]^{0.33}}{D_o\ \text{app}} \right] \qquad \text{11.0}$$

which when simplified is of the form:

$$P = (A^1 + B^1\sqrt{\bar{v}})\quad \Delta TA \qquad \text{12. 0}$$

where:

A^1 = the static unit surface convective conductance.

$B^1\sqrt{\bar{v}}$ = the dynamic unit surface convective conductance.

/... Hence

Hence, in words, the power dissipated from the thermistor to the blood is proportional to the temperature difference and proportional to a constant A¹ plus B¹ times the square root of the mean velocity $\bar{v}$ of the stream.

That P will, in fact,vary with the physical and thermal properties of the blood and the thermistor cannot be disputed but the variations expected are of several orders of magnitude smaller than the contribution by $\sqrt{\bar{v}}$, ΔT and A.

1.3. PARAMETERS.

I^2R is the power dissipation in the thermistor and is expressed in watts. It is the unknown quantity.

$D_{o\ app}$ is the apparent diameter of the thermistor. It is related to the actual diameter D_o by a constant k which is nominally unity but is modulated by two factors viz. the thermistor encapsulation which will tend to increase k and the possibility of the tissue cellular matrix partially covering the thermistor which will tend to reduce k.

ΔT is the temperature difference between the blood and the thermistor and must be made as high as possible. There are, however, three factors which limit the magnitude of ΔT. These are:

a) pain limit due to heating of the surrounding tissue
b) cellular damage caused by the high temperature
c) possible consequences on coagulation time.

All three of these effects become apparent at temperatures in excess of 42°C which corresponds to a ΔT of 5°C. This was accepted as true for the purposes of this work but the writer believes that some simple subjective and objective tests should be performed before future work is attempted. As it will later become apparent, it would be very much more convenient if ΔT could be raised above 5°C.

k_b is the thermal conductivity of the blood and has a value of 12.1×10^{-4} cal/cm sec deg C with a normal deviation of $\pm 2.0 \times 10^{-4}$ cal/cm sec deg C.

μ_b is the viscosity of the blood and has a value of 2.7 centipoise with a normal deviation of ± 0.2 centipoise [72].

ρ_b is the density of the blood and has a value of 1.060 gm/cm^3 with a normal deviation of 0.040 gm/cm^3 [73].

Cp_b is the specific heat of the blood and has a value of 1.060 cal/gm deg C, with a normal deviation of 0.040 cal/gm deg. C.

$\bar{v}$ is the velocity of the blood in the region of the thermistor.

1.4. THEORETICAL INVESTIGATION OF THE EQUATION.

A. The first investigation of equation 11.0 was performed in order to study the relationship between P the power dissipation in the thermistor and $\bar{v}$ the average velocity of the fluid past the thermistor for different values of ΔT.

All other variables were held constant. $\bar{v}$ was incremented from 0.1 to 10.0 cm/sec (this was a 'guesstimate' since the average tissue flow rate in the body per weight of tissue is 100 ml/kg sec). This was repeated for ΔT = 0.5°C to 4.0°C.

The value of $D_{o\ app}$ was chosen as 0.05 cm (Standard Telephones & Cables Ltd., U23US thermistors – D_o = 0.040, ± 0.030.cms).

Equation 11.0 was programmed on an IBM 1130 computer

/... and

and solved by means of nested 'do-loops'. The results are plotted on graph 3.A.G.1. This graph shows fairly clearly the optimum bias conditions for the thermistor since the pain limit bounds the workable area from above and for ΔT low, the value of $dP/d\bar{v}$ approaches a constant value for high flow rates.

B. The second investigation of equation 11.0 was performed in order to assess the effect of the parametric spreads. (normal deviations). The results are plotted on graph 3.A.G.2. Each was taken through its normal deviation with all the others fixed at their nominal values.

a) The change in power with density and specific heat is very small, each at its maximum deviation producing a change of not more than 2% in the value of P.

b) The change in power with viscosity is also small, a maximum change of viscosity producing 2.5% change in the value of P.

c) Changes in the normal thermal conductivity of blood produce a far greater effect, however - up to 20% in P. This variation, however, is linear over the range of interest and if deemed necessary it may be possible to relate this to some measurable - such as hematocrit value and apply a linear correction factor should quantitative results be desired.

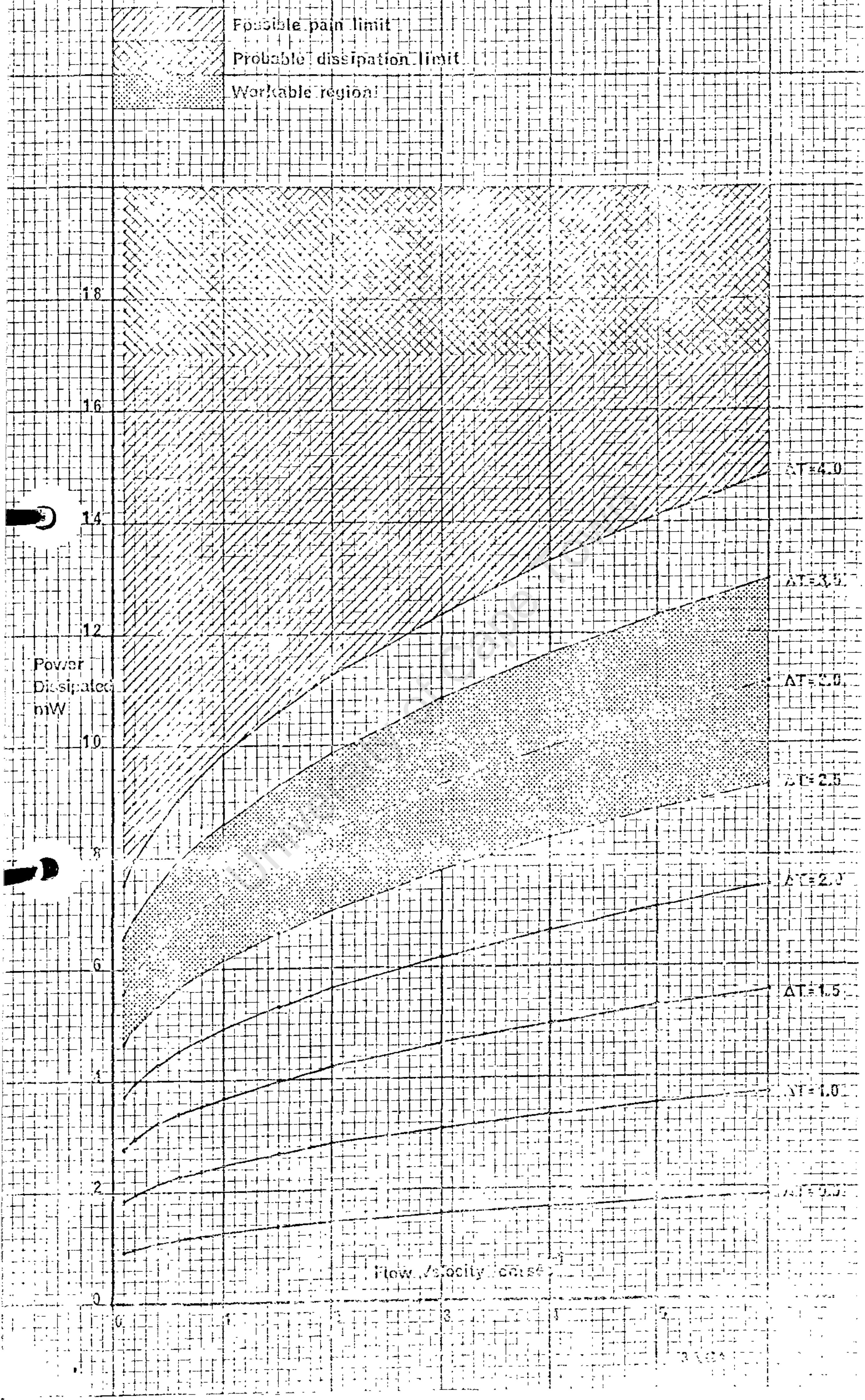
d) The thermistor diameter has, of course, the greatest effect. Again the important point is that this variation is very nearly linear and hence lends itself to compensation by gain adjustment.

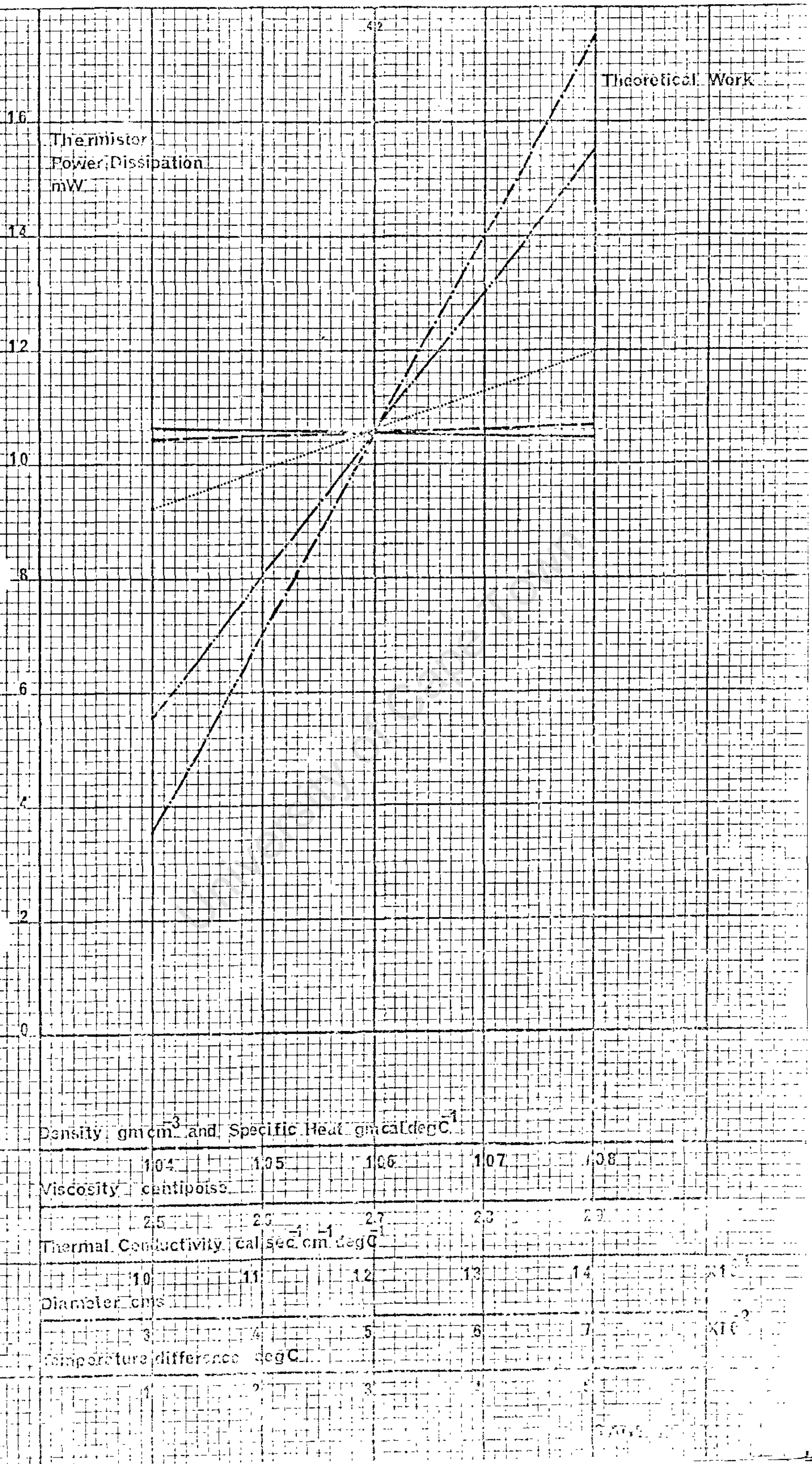
From a) Discussion:

$$D_{o\ app} = k\, D_o$$

and from the graph:

/... P=


Possible pain limit
Probable dissipation limit
Workable region
Power Dissipated mW
0
2
4
6
8
10
12
14
16
18
ΔT=4.0
ΔT=3.5
ΔT=3.0
ΔT=2.5
ΔT=2.0
ΔT=1.5
ΔT=1.0
ΔT=0.5
Flow Velocity cm/sec
0
1
2
3
4
5


Theoretical Work
Thermistor Power Dissipation mW
16
14
12
10
8
6
4
2
0
Density gm cm⁻³ and Specific Heat gm cal deg C⁻¹
1.04
1.05
1.06
1.07
1.08
Viscosity centipoise
2.5
2.6
2.7
2.8
2.9
Thermal Conductivity cal sec⁻¹ cm⁻¹ deg C⁻¹
10
11
12
13
14
X10
Diameter cms
3
4
5
6
7
X10⁻²
Temperature difference deg C
1
2
3
4
5

$$P = a + k^1 D_{o\ app}$$
$$= a + k^1 k D_o \qquad 13.0$$

for a fixed flow velocity.

If Pin is the input power of an amplifier and Pout is the output power such that Pin/Pout = α then if P = Pin we have that

$$\text{Pout} = \alpha (a + k^1 k D_o) \qquad 14.0$$

and it will be possible to suitably calibrate a probe with any value of $D_{o\ app}$ by adjusting the power gain of its associated amplifier. This could be used as a technique for standardisation of probes.

Another interesting point to note is that, as the apparent diameter of the thermistor increases, so the slope of the P vrs $\bar{v}$ for fixed ΔT also increases, making measurements more sensitive. However, the pain level limit together with the increasing the thermal inertia of the thermistor dictates that there must be an optimum $D_{o\ app}$.

c) The relationship of power and temperature difference was plotted for reference.

1.5. FURTHER THEORETICAL INVESTIGATION.

1.5.1. THERMISTORS.

N.T.C. Thermistors are prepared from the high resistivity oxides of the iron group of transition elements:

Cr Mn Fe Co or Ni

Usually they are made of 50:50 percent mole

/...compositions

compositions of $MgCr_2O_4$ and $FeFe_2O_4$. They are transformed into semi-conductors by the addition of small amounts of foreign ions which have a different valency and are provided with compensating ions in order to preserve their electro-neutrality.

The resistance variation of these thermistors in a broad temperature region is expressed by the simple equation:

$$R = Ae^{B/T} \tag{15. 0}$$

Where A and B are constants which depend entirely upon the nature of the semiconductor material. α is defined as the temperature coefficient and is expressed as follows:

$$\alpha = \frac{I}{R}\,\frac{dR}{dT} = -\frac{B}{T^2} \tag{16. 0}$$

and is in the region of 1 to 5 % per deg. C.
3.4% for S.T.C. U23US types.

<u>1.5.2. POWER DISSIPATION.</u> Equation 15.0 may be written:

$$\log_e R = \log_e A + \frac{B}{T} \tag{17. 0}$$

if:

T_b = the temperature of the blood.
T_t = the temperature of the thermistor.
D = the dissipation factor - i.e. the power required to raise the temperature of the thermistor by 1°C then

$$P = D\,(T_t - T_b) \tag{15. 0}$$

/... now

now

$$P = VI \tag{19. 0}$$

$$\therefore \quad \log_e V + \log_e I = \log_e D + \log_e (T_t - T_b)$$

also from 17.0

$$\log_e V - \log_e I = \log_e A + {}^{B}/_{T_t}$$

$$\therefore \quad \log_e V = {}^{1}/_{2} \log_e AD + {}^{1}/_{2} \log_e (T_t - T_b) + {}^{B}/_{2\,T_t} \tag{20. 0}$$

This equation has an extreme value as a function of T if:

$$d \log_e {}^{V}/_{d\,T_t} = 0 \tag{21. 0}$$

in which case

$$\frac{1}{2(T_t - T_b)} = \frac{B}{2\,T_t^{\,2}}$$

$$\text{i.e.} \qquad 2B\,(T_t - T_e) = 2\,T_t^{\,2} \tag{22. 0}$$

T_t max can be obtained from the formula:

$$\frac{{}^{1}/_{2}\, b \pm \sqrt{b^2 - 4\,ac}}{2a}$$

(The value of the minus sign gives the temperature corresponding to the maximum value of voltage). Only if $B > 4\,T_t$ will this maximum be present and for $B = 2900$ which is the case for the U23US thermistor then

$$T_t \max = 85^\circ C \tag{23.0}$$

1.5.3. THERMAL TIME CONSTANT.

If a thermistor has a uniform temperature during cooling then the following equation holds for its cooling during a time dt

$$- H dt = D\,(T - T_c)\ dT \tag{24.0}$$

where H is the heat capacity of the thermistor in joules/deg C

This yields

$$(T - T_o) = (T_1 - T_o)\ e^{-t/\Gamma} \tag{25.0}$$

where $\Gamma = H/D$. Γ is termed the thermal time constant and represents the time required for the thermistor to change to $1/e$ of the total difference between its initial and final temperature when subjected to a step function in temperature.

1.6 EQUATION 11.0 REWRITTEN.

Equation 11.0 is of no great value in its present form since, in order to measure $\bar{v}$ it would be necessary to hold V/I constant and measure VI which is technically possible but more difficult.

Rewriting 11.0 we get

/... VI =

$$VI = 4.186\Pi\, D_{oapp}\, \Delta T \left[\frac{k_b \{ 2.0+0.6\left[\frac{D_{oapp}\,\bar{v}\rho_b}{\mu_b}\right]^{0.5}\left[\frac{C_{P_b}\,\mu_b}{k_b}\right]^{0.33}}{D_{o\,app}}\right] \qquad 26.0$$

Therefore:

$$\left[\frac{VI}{4.186\Pi D_{oapp}\,\Delta T\; k_b}\right] = 2.0+0.6 \left[\frac{D_{oapp}\bar{v}\rho_b}{\mu_b}\right]^{0.5} \left[\frac{C_{P_b}\,\mu_b}{k_b}\right]^{0.33}$$

and so

$$\frac{\left[\frac{VI}{4.186\Pi D_{oapp}\Delta T\; k_b} - 2.0\right]^2}{0.6\left[\frac{C_{P_b}\,\mu_b}{k_b}\right]^{0.33}} = \left[\frac{D_{oapp}\,\bar{v}\;\rho_b}{\mu_b}\right]$$

$\therefore$

$$\bar{v} = \frac{\mu_b}{D_{oapp}\rho_b}\;\frac{\left[\frac{VI}{4.186\; D_{oapp}\Pi.6\Delta T\; k_b} - \frac{2.0}{0.6}\right]^2}{\left[\frac{C_{P_b}\;\mu_b}{k_b}\right]^{0.33}} \qquad 27.0$$

It now remains for ΔT to be rewritten in terms of V and I in compliance with the characteristics of the thermistors.

/...From

From equation 15. 0 we get

$$T_t = B\, e^{V/IA} \qquad \text{28. 0}$$

$$\text{and } \Delta T = T - T_o \qquad \text{29. 0}$$

$$\text{hence } \Delta T = B\, e^{\frac{V}{IA}} - T_b \qquad \text{30. 0}$$

Now furnished with equations 27. 0 and 30. 0 it is possible to obtain the relationship between the flow velocity $\bar{v}$ and the voltage and current V & I of the thermistor.

If I is fixed in value then ΔT will vary and it is, therefore, necessary to add the proviso that

$$0 \leq \Delta T \leq 5°C$$

Because of the difficulty of separating the variables in equations 30. 0 and 27. 0, the following procedure was adopted for solution by means of the computer:

a) Choose a value of I

b) For the restriction that $\Delta T = (T_t - T_b)$ where $0°C \leq \Delta T \leq 5°C$ and $T_b = 37°C$ determine from 30. 0 the values of V which correspond. i.e. the set of V which suit.

c) Now put these values of V into equation 27. 0 and determine the corresponding range of flow velocities which suit.

1.7. FINAL THEORETICAL RESULTS.

I was varied from 0.1 mA in steps of 0.1 mA to 3 mA. For these values of I, V falls in the region 0.80 to 3.40 volts, and the corresponding flow velocities were 0 to 100cm/sec.

/... Graph

Graph 3.A.G.3. shows an extract from these curves - that of I = 2.0mA to I = 2.8 mA. Clearly of all the values of I, this range is best suited for this work. The full range of flow velocities is covered and the rate of change of V with $\bar{v}$ is high.

Due to the nature of equation 27. 0 the relationship is not simple. In equation 27. 0 the term

$$\frac{VI}{4.186\Pi\, D_{o\ app}{}^{0.60}\, \Delta T\, k_b} - \frac{2.0}{0.60} \qquad 31.\ 0$$

has a value of zero when

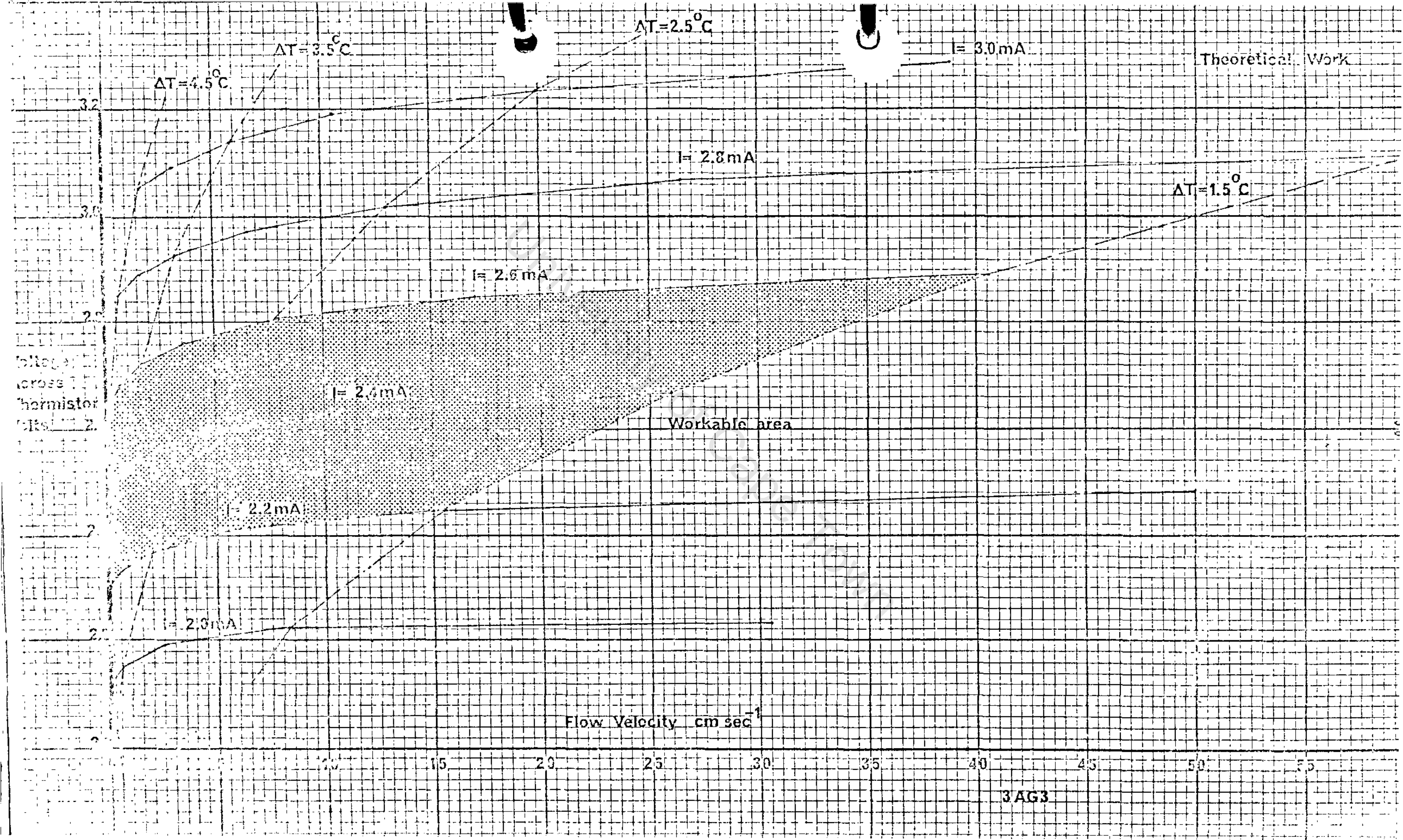
$$VI = 2.0.4.186\Pi D_{oapp}\Delta Tk_b \qquad 32.\ 0$$

and is negative when

$$\frac{VI}{4.186\Pi D_{oapp}{}^{0.60}\, \Delta T\, k_b} < \frac{2.0}{0.60} \qquad 33.\ 0$$

Condition 32. 0 arises when the equilibrium is obtained by virtue of the static convectional heat dissipation only. Condition 33. 0 corresponds to a negative direction of power flow i.e. when the blood heats the thermistor and, under normal conditions, the convection is by means of both the static and dynamic mechanisms.

This concludes the theoretical predictions based on the mathematical model.


Theoretical Work
ΔT=4.5°C
ΔT=3.5°C
ΔT=2.5°C
ΔT=1.5°C
I= 3.0 mA
I= 2.8 mA
I= 2.6 mA
I= 2.4 mA
I= 2.2 mA
I= 2.0 mA
Workable area
Voltage across Thermistor Volts
3.2
3.0
Flow Velocity cm sec^{-1}
10
15
20
25
30
35
40
45
50
55
3 AG3

SECTION 3.B.

PROBES.

1.0 OBJECTIVE. Given the criteria that the technique described above dictates, the objective was to design and build a thermistor probe in such a way that it closest resembled the theoretical ideals and at the same time was simple and easy to make.

1.1.0. THEORETICAL IDEALS. The ideal model of a thermistor probe for use with the above technique is as follows:-

1.1.1. PHYSICAL IDEALS.

a) The thermistor must be of such a shape that it has the minimum possible volume for the greatest possible surface area. It must also be physically small. This means that its thermal inertia will be low and the power dissipated, by virtue of its self-heating, to its surrounds, will be high.

b) The thermistor must have the greatest possible value of $^{dR}/_{dT}$ so that minimum amplification is required.

c) The electrical connections to the thermistor must be good electrical conductors and poor heat conductors so that heat cannot be removed along the conductors.

d) The thermistor must be capable of being placed freely in the flowing medium so that it is separated from any material which could distort the heat flow pattern, insulate from the flowing medium or raise its thermal inertia.

e) The thermistor must have a low impedance, be well screened from stray electromagnetic radiation and be electrically insulated from the flowing medium.

f) The probe must preferably not be direction sensitive

/... so

so that it can be sited in the flowing medium from any angle with little or no difference in performance.

1.1.2. MEDICAL REQUIREMENTS.

a) The thermistor must be of such a shape and size that it will reasonably average the effects (on a macroscopic scale) of the local tissue composition variations.

b) The probe must be small enough to minimise trauma associated with its insertion.

c) The probe must be able to withstand repeated insertions without damage.

d) The probe must be easy to handle and insert and must be physically robust.

e) It must be boilable for sterilisation or must be able to be sterilised with gas.

f) It must be made of materials which will not contaminate or be contaminated by any of the biological fluids. In particular, it must not cause clotting of blood as this would seriously impair operation.

1.2. PRACTICAL LIMITATIONS. Naturally the practical limitations will inhibit realisation of the ideal. Consider the following limitations and their effect on the ideal probe.

Commercially available bead thermistors are always spherical or nearly spherical in shape. This is simply a manufacturing limitation. Sizes are variable and are determined by the manufacturing processes and are generally a function of the viscosity and surface tension of the semiconductor material at the temperature of formation of the bead. The dimensions are in the region of 0.30mm to 1.50mm in diameter with a poor tolerance of ± 50% to ± 20% [42]. While the spherical shape and physical size

/... defeats

defeats the requirements of 1.1.1. a), it contributes somewhat towards that of 1.1.2 a). Closer tolerance thermistors are available but are costly. They are simply sorted with regard to size, shape or resistance value.

The value of $^{dR}/dT$ for commercial thermistors is between 1% and 5% per °C at 20°C and can be tailored by the manufacturers as requested. This was found to be quite acceptable for the biasing of the thermistors in this application. i.e. at 37°C and with the temperature variations expected. (condition 1.1.1.b)).

The conductors are made of material which, at most, contributes towards satisfying condition 1.1.1.c). Platinum alloys and, in particular, Platinum Ruthenium wires are used. These are extremely thin. (0,025mm to 0,075mm in dia.)

Thermal conductivity of Pt	1.77 watts/in°C
Thermal conductivity of Cu	9.77 watts/in°C
Electrical conductivity of Pt	10.0 $\mu\,\Omega\,cm^3$
Electrical conductivity of Cu	1.72 $\mu\,\Omega\,cm^3$

The thermistors are usually coated in an extremely thin layer of lead glass to prevent them from disintegrating and since they are normally very delicate even with this coating, they must further necessarily be encapsulated in a resin for mechanical rigidity. In order, however, to least impair the heat dissipation qualities a minimum of 50% of the surface area of the thermistor must be covered in a very thin layer of the resin. Further, investigations into the nature of muscle tissue, generally, have shown that stainless steel hypodermic needles of 15g to 18g are suitable for repeated insertion with the minimum of trauma and tissue damage. They are also, fortunately, large enough to accommodate thermistors within their bores. The thermistor must be placed as far forward in the needle as possible without bearing the major stress of parting the tissue.

/...The

The resin has now to satisfy three conditions:-

1) It must have a high mechanical tensile strength.

2) It must have a low thermal conductivity so that it will not substantially increase the thermal inertia of the thermistor.

3) It must have a high thermal conductivity so that it does not thermally insulate the thermistor from the flowing medium.

Conditions 2) and 3) above are in direct conflict but, as it happens, epoxy resins (chosen for their hardness and adhesive qualities) have neither a high nor low thermal conductivity - typically 1 to 4 X 10^{-3} cal/cm/°C/sec - nor has lead glass - typically 2 X 10^{-3} cal/cm/°C/sec. Condition 1) is also adequately satisfied - typically 350 to 750Kg/cm^2 .[43] Epoxy resins have a high dielectric strength 1 to 10 X 10^{12} ohm cms [44] and they are often recommended for implantable electrodes [45] so, thus far, conditions 1.1.1. d) and e) and 1.1.2. b), c) and f) are reasonably satisfied.

The thermistors are boilable T max = 180°C but it is preferable to use gas sterilisation. (condition 1.1.2.e).

The stainless steel hypodermic needle is naturally direction sensitive but provided that it can be held rigidly in one position relative to the flowing medium, the effect of different orientations can only be to increase or decrease sensitivity. Since the purpose was to develop an instrument for qualitative measurements only, this presents no severe limitation. (condition 1.1.1. f).

The probe body could probably best be made of

/... Teflon

Teflon (R) but, although Teflon (R) is extremely inert and has a high tensile strength, it suffers from severe plastic deformation and is difficult to machine. Nylatron (R) was chosen for ease. (condition 1.1.2.d)

Having reasonably satisfied all the ideal requirements, the probe design will now be discussed.

1.3. DESIGN.

Thermistors. Fig. 3.B.1., shows the specification sheets for two thermistor types that could be used. Because of their closer tolerance, smaller physical size, greater $^{dR}/_{dT}$ and specific design purpose, the Standard Telephone & Cables Ltd., type U23US (R) were used.

Hypodermic Needles. Galaxy (R) disposable 18g stainless steel syringe type hypodermic needles were used. Fig. 3.B.2. The moulded plastic syringe fitting was removed and the needle belled out slightly at its squared-off end. This was done on a vertical drill press with a long conical machine tool bit. The stainless steel was found to be very hard and care had to be taken so as not to tear the metal. A very shallow angle was used to prevent the needle from sliding through its guide hole.

Probe Body. Fig. 3.B.2. This was machined from Nylatron(R) stock on a centre lathe with a three jaw self centering chuck. Threads were cut with taps and dies. Section C was placed in a circular collet to prevent distortion when the thread was tapped and the drill bit left in section B when its thread was cut. A shallow rake angle and a deep cut gives the best finish.

Cables and Plugs. Single core braided screened cable and RCA (R) plugs and sockets were used.

Thermistor extension wires. 48g (0.0072") tinned copper wires were used for the extension and connection wires.

/...Epoxy

STC

THERMISTORS

Type U

Unmounted bead

DESCRIPTION

The type U is a very sensitive thermistor responding rapidly to changes of environment and temperature. It is intended primarily for use in katharometry and anemometry, but can be used for temperature measurement. Owing to its small size it is particularly useful for special medical applications where space is at a premium.

The type U thermistor is manufactured by forming a bead of thermistor material on two platinum ruthenium wires of 0,025 mm (0·001 in) diameter. The bead is then coated with a layer of lead glass approximately 0,025 mm (0·001 in) thick to protect the surface of the thermistor from the surrounding medium. The thermistor so formed is lemon-shaped and of approximate diameter 0,4 mm (0·015 in). It is available in single ended (U23US) or double ended (U23UD) configurations.

The standard U thermistor is available in only one value of resistance. However, other sizes and resistance values are available to special order.

STANDARD TYPE (U23)

Resistance at 20 °C (R_{20})	2	kΩ
Resistance at 25 °C (approximately)	1·7	kΩ
Minimum operating resistance (approximately)	60	Ω
B nominal	2 900	°K
Temperature coefficient at 20 °C (approximately)	−3·4	%/°C
E_{max} in free air at 20 °C (typical)	1·5	V
Maximum operating current in free air at 20 °C	16	mA

Type U

CONTINUED

GENERAL DATA

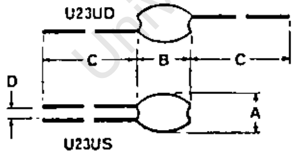
Maximum ambient temperature, T_{Amax}	180	°C
Maximum bead temperature	180	°C
Maximum mean power dissipation in free air at 20 °C averaged over any 20 ms period	14	mW
Maximum instantaneous power dissipation in free air at 20 °C	28	mW
Standard resistance tolerance	±20	%
Nominal thermal cooling time constant	1·0	s
Dissipation constant	0·087	mW/°C

SPECIAL REQUIREMENTS

A large increase in the cost of a thermistor is often produced by only a small change in the specification. Since the cost is also dependent upon the quantity, it is advisable to contact the Technical Sales department whenever a non-standard thermistor is required.

Type U thermistors are available as matched pairs to within 5 per cent at 20 °C (i.e. the resistances of the two thermistors at 20 °C are such that the value of the lower resistance thermistor is within 5 per cent of that of the higher resistance thermistor). The suffix letters MP should be added to the code number when ordering matched pairs, e.g. U23US becomes U23USMP.

OUTLINES

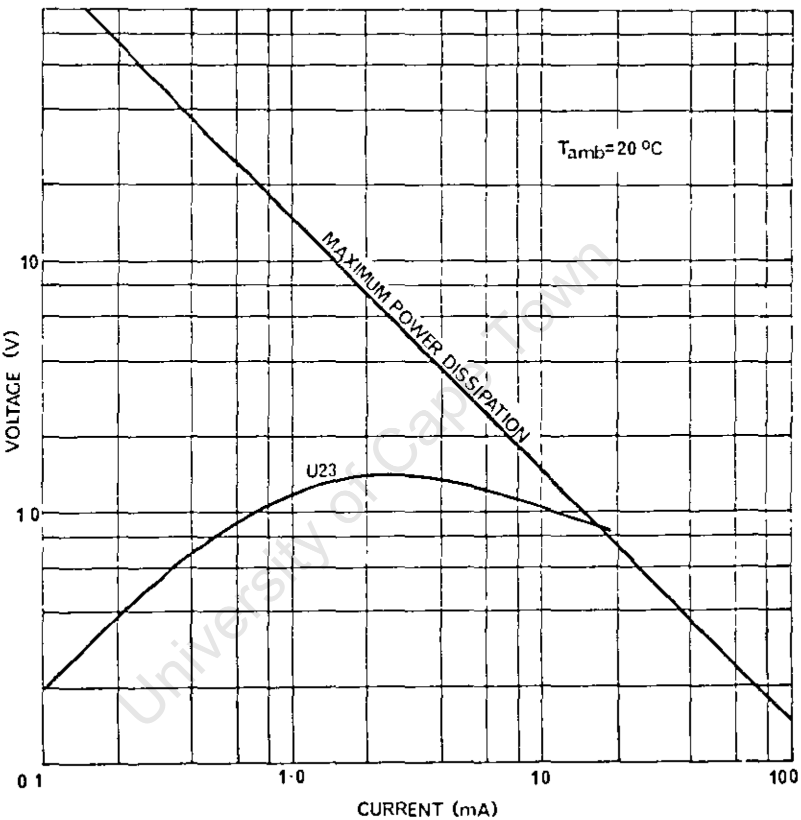


Dimension	mm	in
A (nom.)	0,4	0·016
B (approx.)	0,5	0·020
C (min.)	25,4	1·0
D (approx.)	0,13	0·005

Type U

CONTINUED

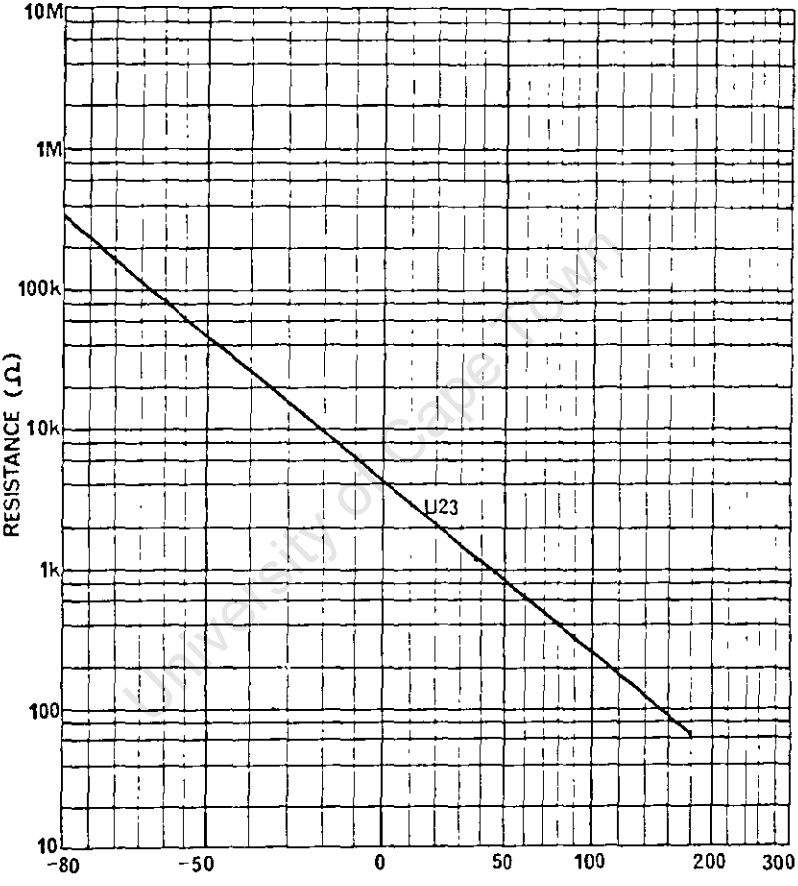
Voltage v. Current in Free Air



Type U

CONTINUED

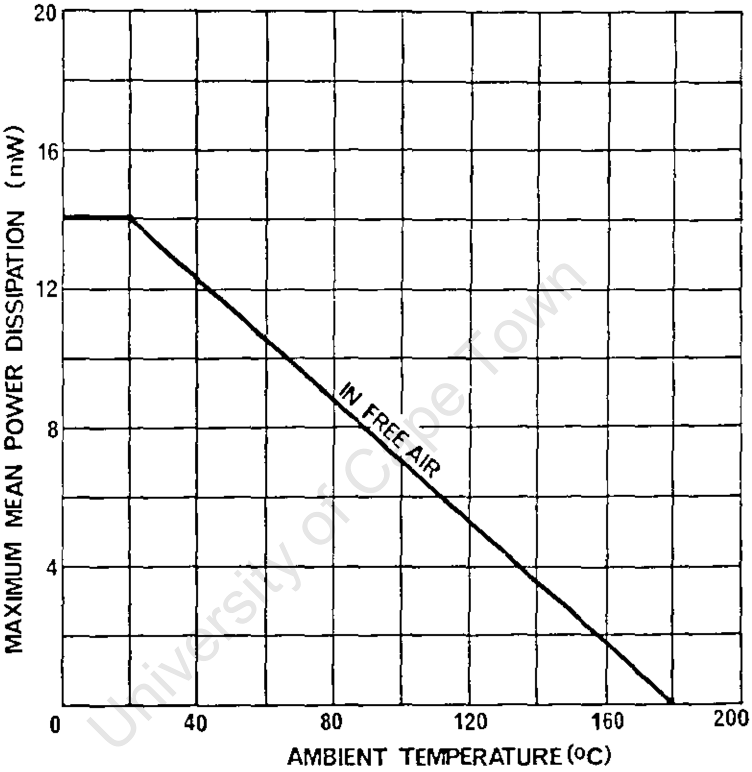
Resistance v. Temperature



Type U

CONTINUED

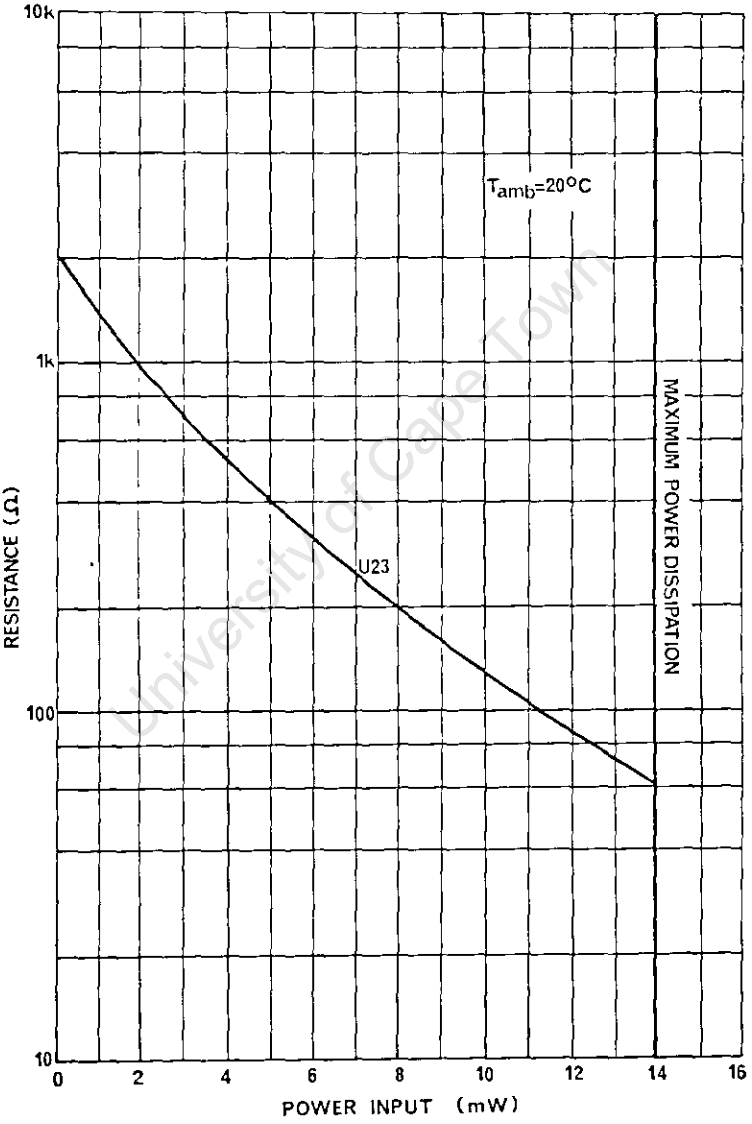
Power Derating



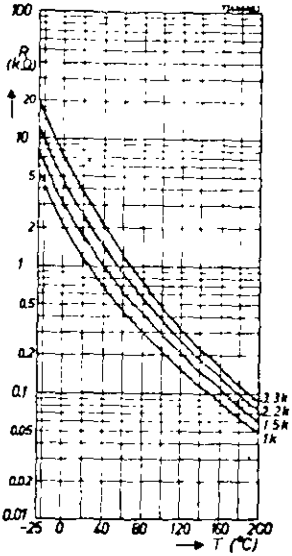
Type U

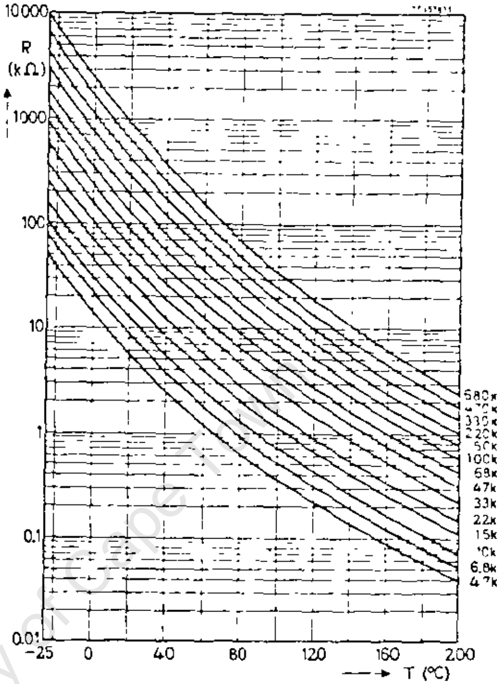
CONTINUED

Resistance v. Input Power



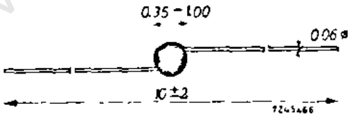
Resistance/temperature characteristics



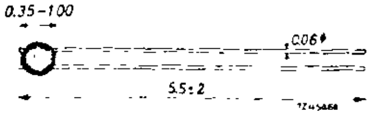


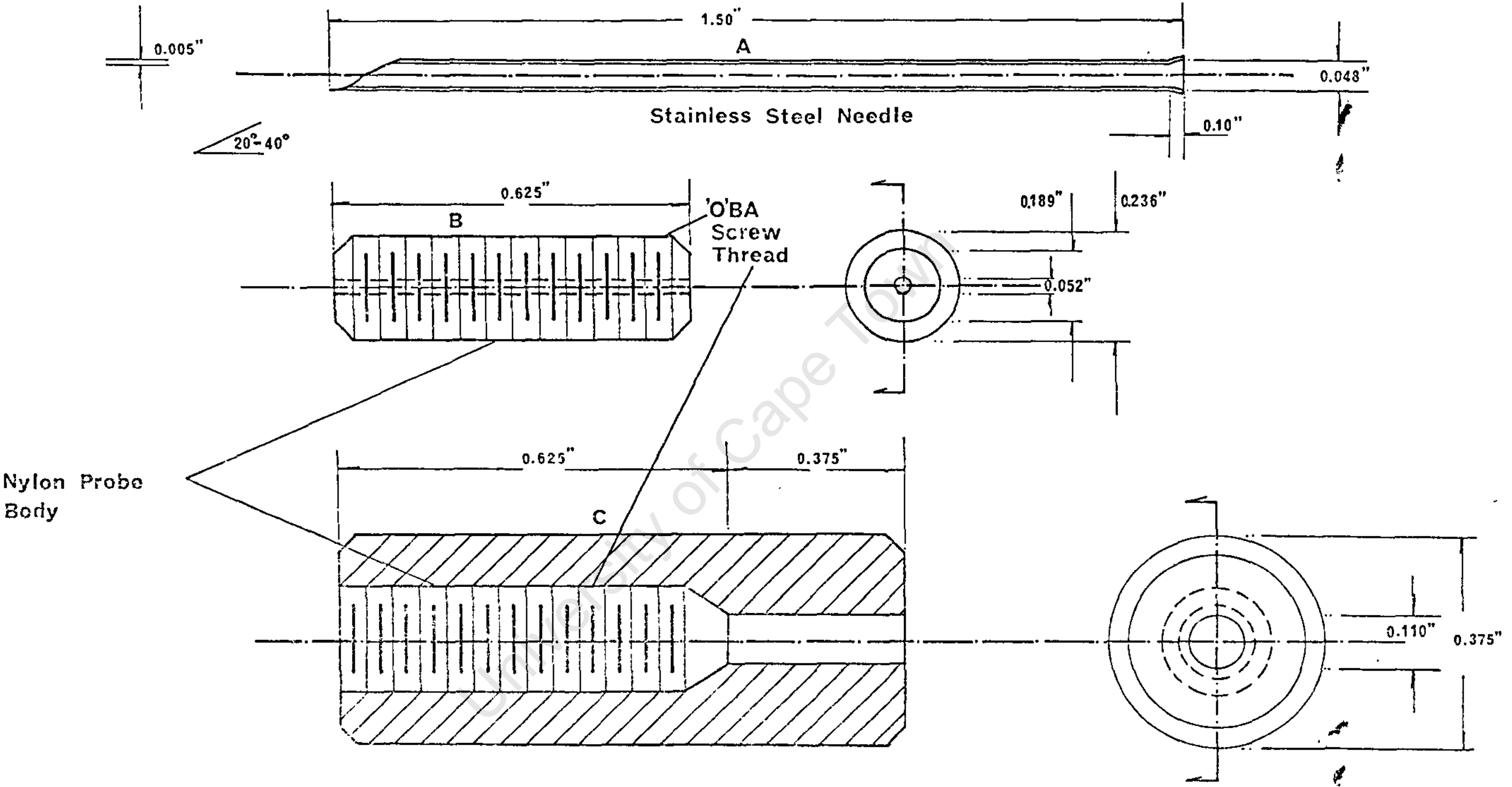
VERSIONS

2322 634 01... Naked bead



2322 634 11... Naked bead




1.50"
A
0.005"
0.048"
Stainless Steel Needle
0.10"
20°-40°
0.625"
B
'O'BA
Screw
Thread
0.189"
0.236"
0.052"
Nylon Probe
Body
0.625"
0.375"
C
0.110"
0.375"

3.B.2.

Epoxy Resins. Five varieties of epoxy resin were tried. They are listed together with the purpose and evaluation in Fig. 3.B.3.

1.4 ASSEMBLY. In the design of a probe of such small dimensions great emphasis must be placed on simplicity of construction so that a number of identical probes may be made without a large variation in the characteristics of each. The following probe design and assembly was developed through numerous trials. It is simple and effective.

The major problem areas were as follows:-

a) Creep of Resins.

b) Soldering of thermistors to extension wires.

c) Achieving a high insulation resistance between thermistor leads and the metal of the hypodermic needles.

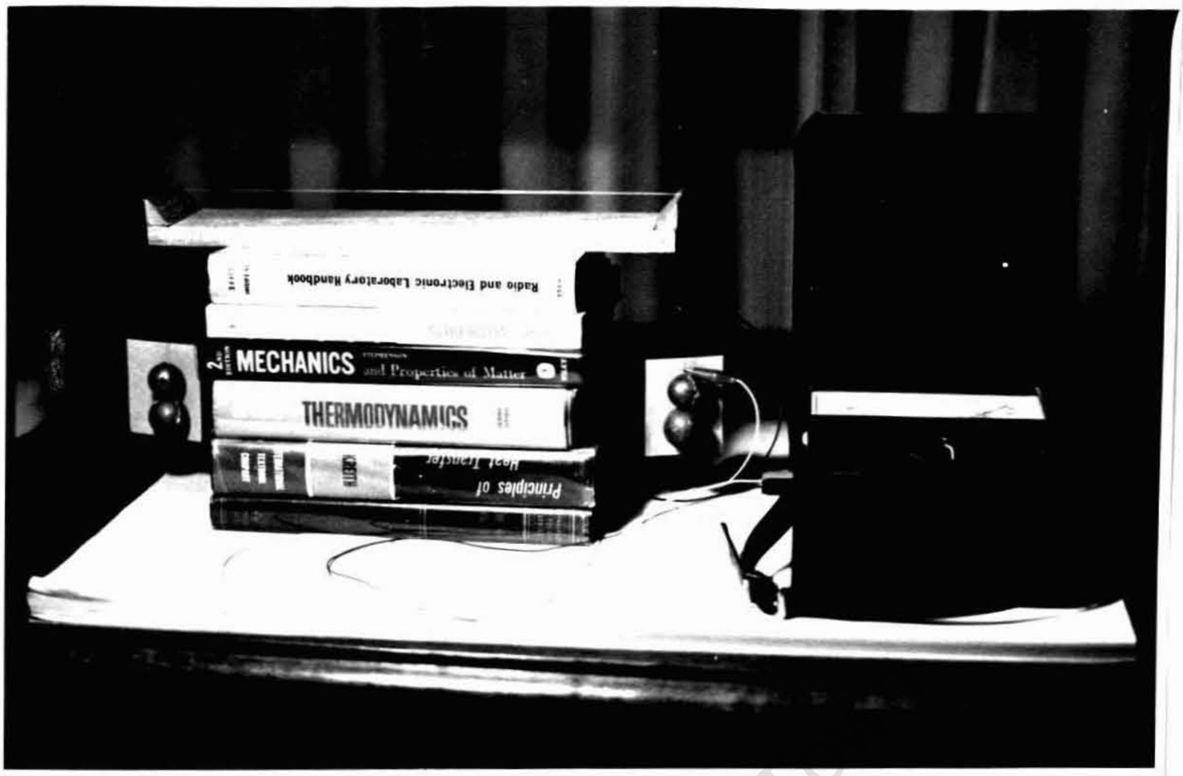
d) Centering the thermistor in the bores of the needles.

The solution to these problems is contained in the assembly method which now follows:

Step 1. Photographs P.1., P3 & P.4. show the method of preparation of the extension wires. P.1. shows how the 48 gauge (0.0072") wires were extended between the knife supports by means of lead fishing sinkers. The wires were connected to an ohmmeter to serve as a check for short circuits and they were separated where the sinkers naturally touched by means of two pieces of cardboard. P.3. shows how the wires were edged up to each other to a separation of ± 0.005". A drop of Pratley's (R) white epoxy glue was

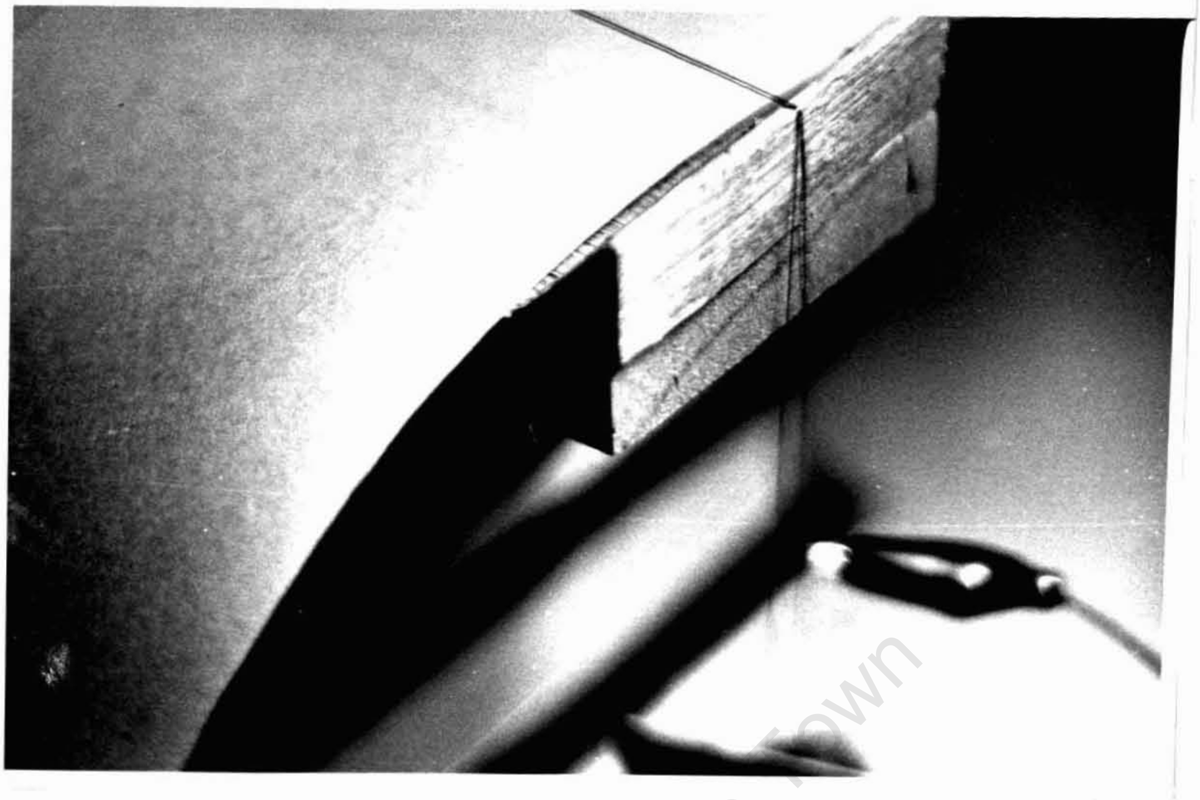
Name of Product	Araldite	Pratleys Clear Epoxy Quickset Glue	Pratleys White Epoxy Glue	Polylite 710 A Polyester Resin	Clear Epoxy 365 Activator and Base
Supplier or Maker	Ciba-Geigy Ltd. Basle Switzerland	Pratleys Mfg. & Eng. Co. Ltd. Transvaal	Pratleys Mfg. & Eng. Co. Ltd. Transvaal	Poly-Resin Products Ltd. Durban	
Encapsulation of Thermistors					
Rating	Excellent	Excellent	Good *	Poor	Poor
Evaluation	Slightly slow to cure This may easily be accelerated by the application of heat Hard and has high tensile strength Recommended for encapsulation of implantable electrodes High adhesion to stainless steel Difficult to work because of low viscosity at accelerating temperatures	Slightly fast to cure Hard and has high tensile strength High adhesive strength to stainless steel Difficult to work with because of high viscosity and quick curing time — sets in 10 minutes at room temperature	* For temperature sensor probes only Hard, breaks up under thermal expansion caused by heating of thermistor for the flow sensor High tensile strength High adhesion to stainless steel Easy to work because of moderate curing time and colour – very important * Water absorption is high because of chalk content (± 0·2% by vol.)	Low adhesion to stainless steel prevented use Peeling on the bevelled edge of the needle	Softening when in contact with saline for protracted periods prevented use
Adhesion of Extension Wires					
Rating	Poor	Good	Excellent		
Evaluation	Ultimate strength in 3 - 4 days Too slow to use in practise Jig is likely to be bumped by someone before ultimate strength is acheived	Does not cause separation of wires from stainless steel needle	Cures overnight and chalk particles cause good separation of outsides of extension wires from inside wall of needle	Not used	Not used

3.B.3

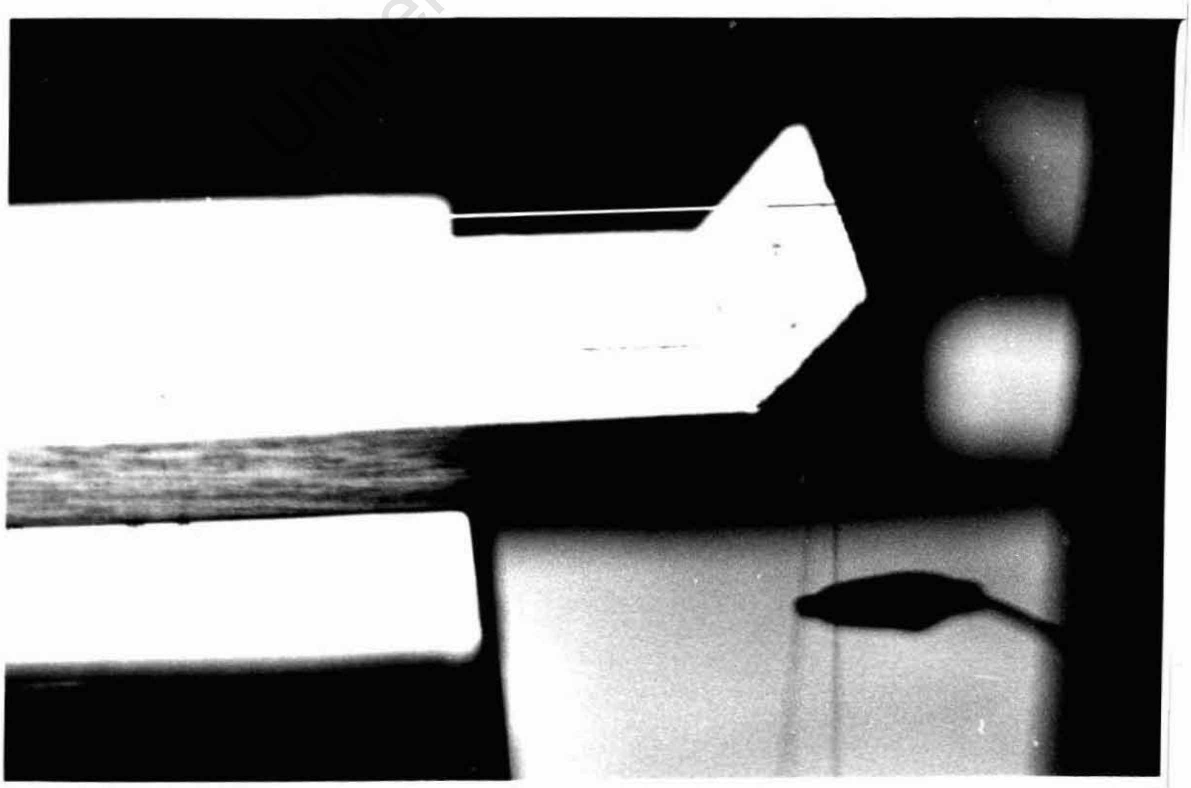

Radio and Electronic Laboratory Handbook
2nd EDITION MECHANICS and Properties of Matter
THERMODYNAMICS
Principles of Heat Transfer

P1

P2



P3



P4

placed between the forefinger and thumb and run the length of the extension wires. The result was a smooth clean bonding between and on the outside edges of the wires as shown in P.4. After overnight curing the bonded wires were cut into appropriate lengths and the epoxy removed at each end. (Fig. 3.B.4.D.)

It was found that the chalk granules of the white epoxy provided a good key for the lacquer coat so that the extension wires were well insulated from the inside of the hypodermic needle.

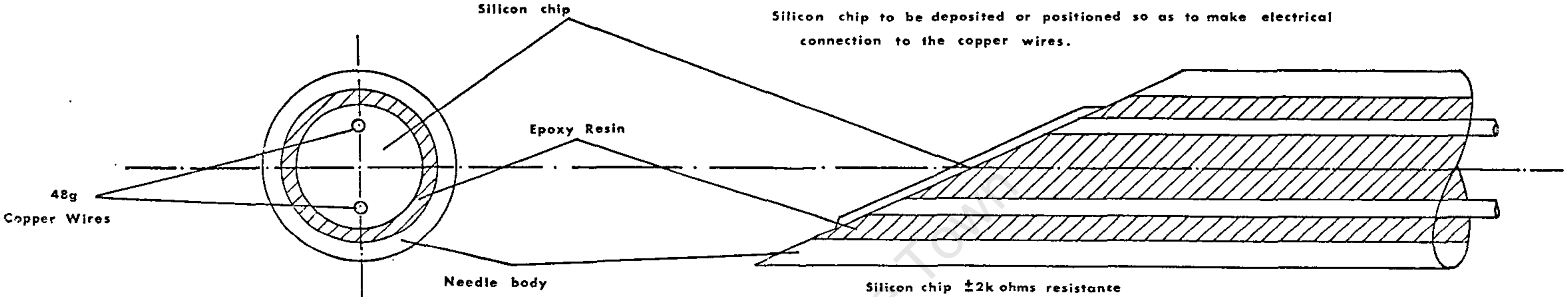
Step 2. The thermistor shown in P.6. was then soldered to the two extension wires. (Fig. 3.B.4. A, B & C.) This was by far the most difficult task in assembling the probes. The best approach was found to be to secure the extension wires by means of Cellotape (R) and hold the thermistor with shortened leads, by its leads, in the jaws of a pair of tweezers. The whole assembly was then coated in clear lacquer several times along its length in order to build up insulation and strength. As a check, the thermistor resistance was monitored continuously during this stage. The lacquered assembly is shown in P.5.

Step 3. The cable end (Fig. 3.B.4. G & H) was prepared and passed through the cable end of the probe body. Fig. 3.B.2.C.) and shown in P.2. The cable end was soldered to the extension wires (Fig. 3.B.4. E & F) and the cable extracted so that the soldered joints were well inside the probe body.

Step 4. Fig. 3.B.5. shows the probe assembly jig which was made of Perspex (R). The hypodermic needle, complete with probe body screw was clamped in the probe assembly jig. The assembled cable, extension wire, thermistor and probe body was passed through the hole for the probe body and up the clamped hypodermic needle until the thermistor was sited

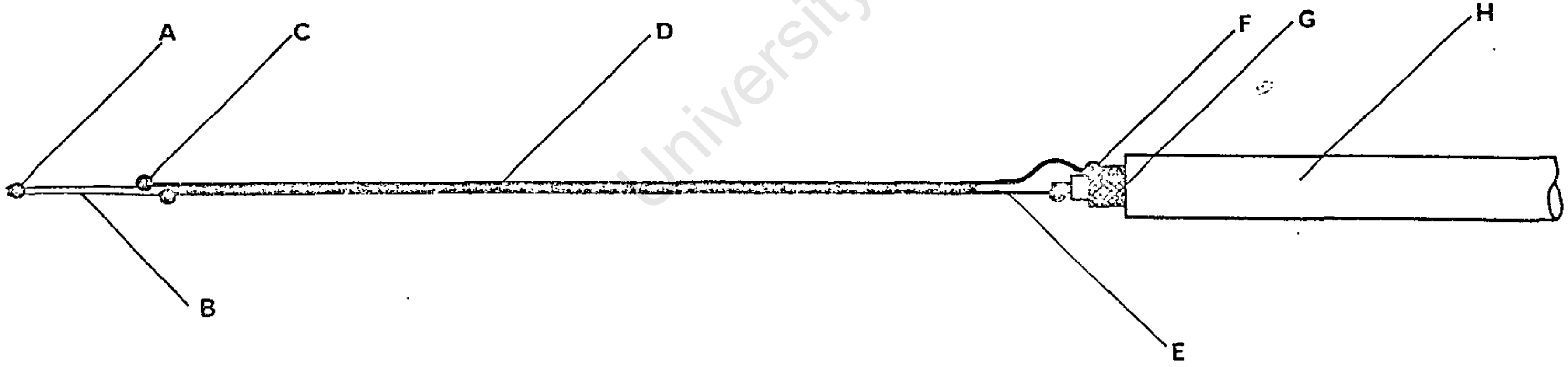
Wires to be threaded through needle, needle to be filled with epoxy resin and ground off at an angle of 25° to the centre line.

Silicon chip to be deposited or positioned so as to make electrical connection to the copper wires.

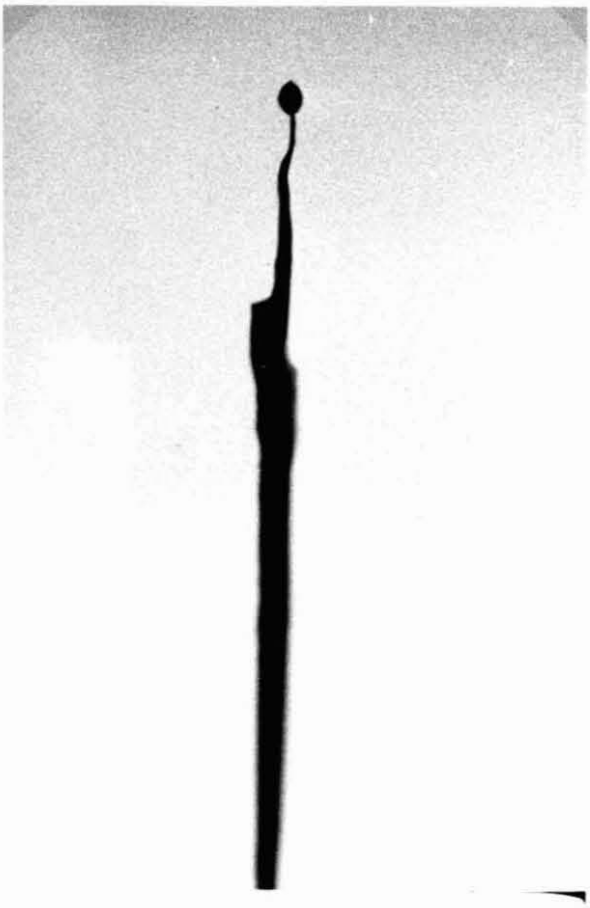


Silicon chip ±2k ohms resistance and 2 to 3 thou. thick, to be protected with very thin layer of resin or glass (oxidised silicon)

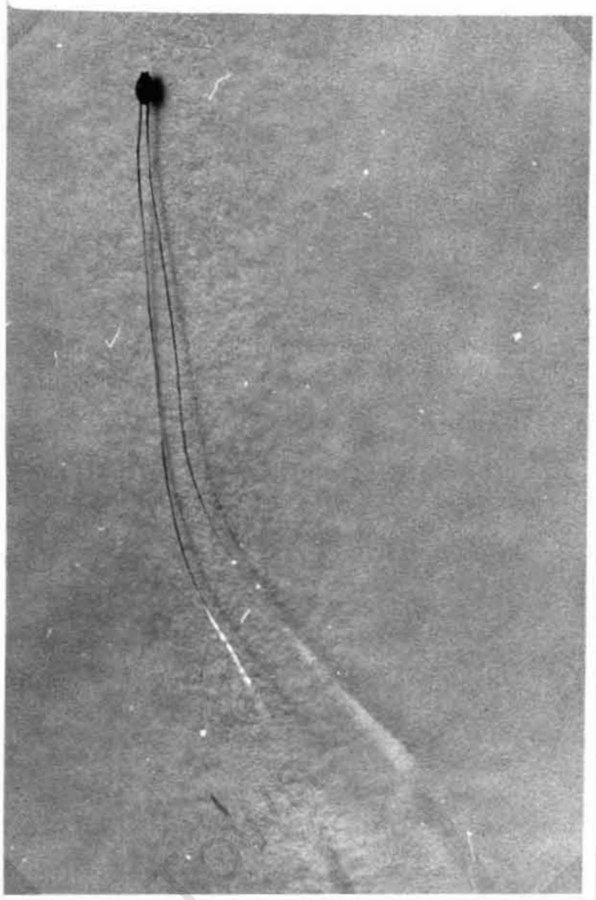
3.B.6



3.B.4

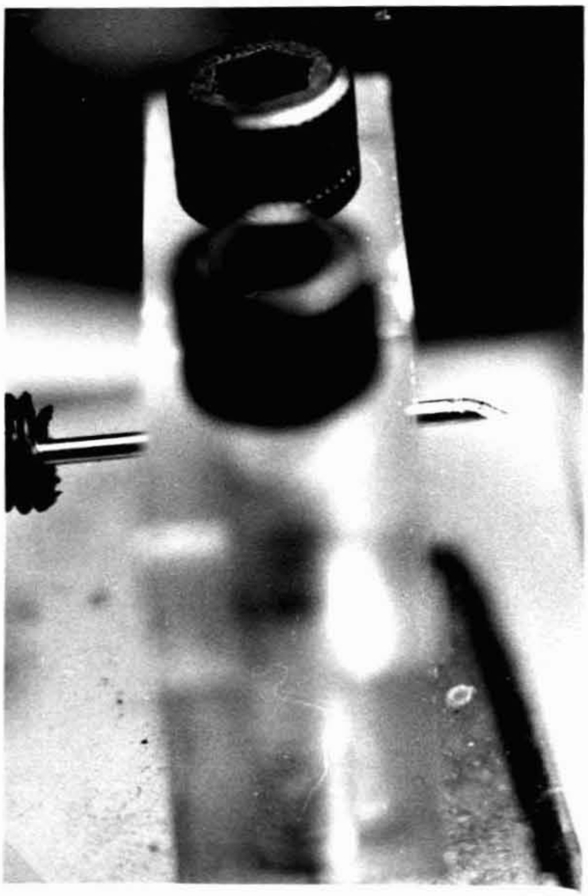


P5



P6

P7



P8

in its correct position at the bevelled end of the needle. To obtain the correct siting it was often necessary to rotate the assembly and gently prod the thermistor into the correct position. The probe body was then clamped in the probe assembly jig. (shown in P. 7.)

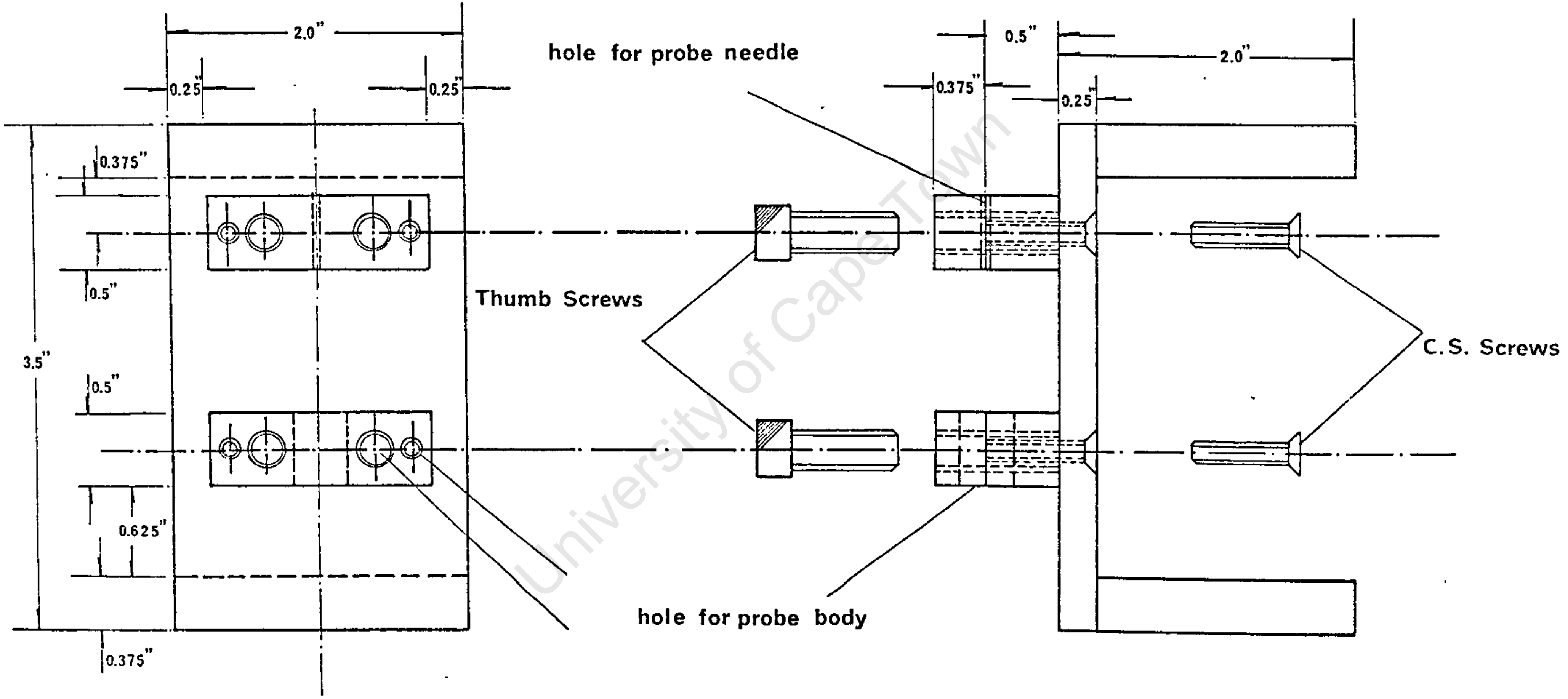
Step 5. The jig was then positioned vertically and the probe body filled with the encapsulating epoxy resin by means of a hypodermic needle and disposable syringe. The probe body screw was then screwed into the probe body to a depth of approximately 0.250". This forced the resin up the hypodermic needle past the soldered joints and into the bevelled end. When the thermistor was judged to be sufficiently covered, the jig was canted so that the bevelled end was horizontal. The screw was given a slight tightening to allow for the contraction creep of the resin, and the assembly was left to cure. (Shown in P.8.) Accelerated curing was obtained by the application of heat. (blower heater). Again the resistance of the thermistor was monitored during this process.

Step 6. The curing complete, the protruding screw was machined flush in a lathe. The finished probe is shown in P.9.

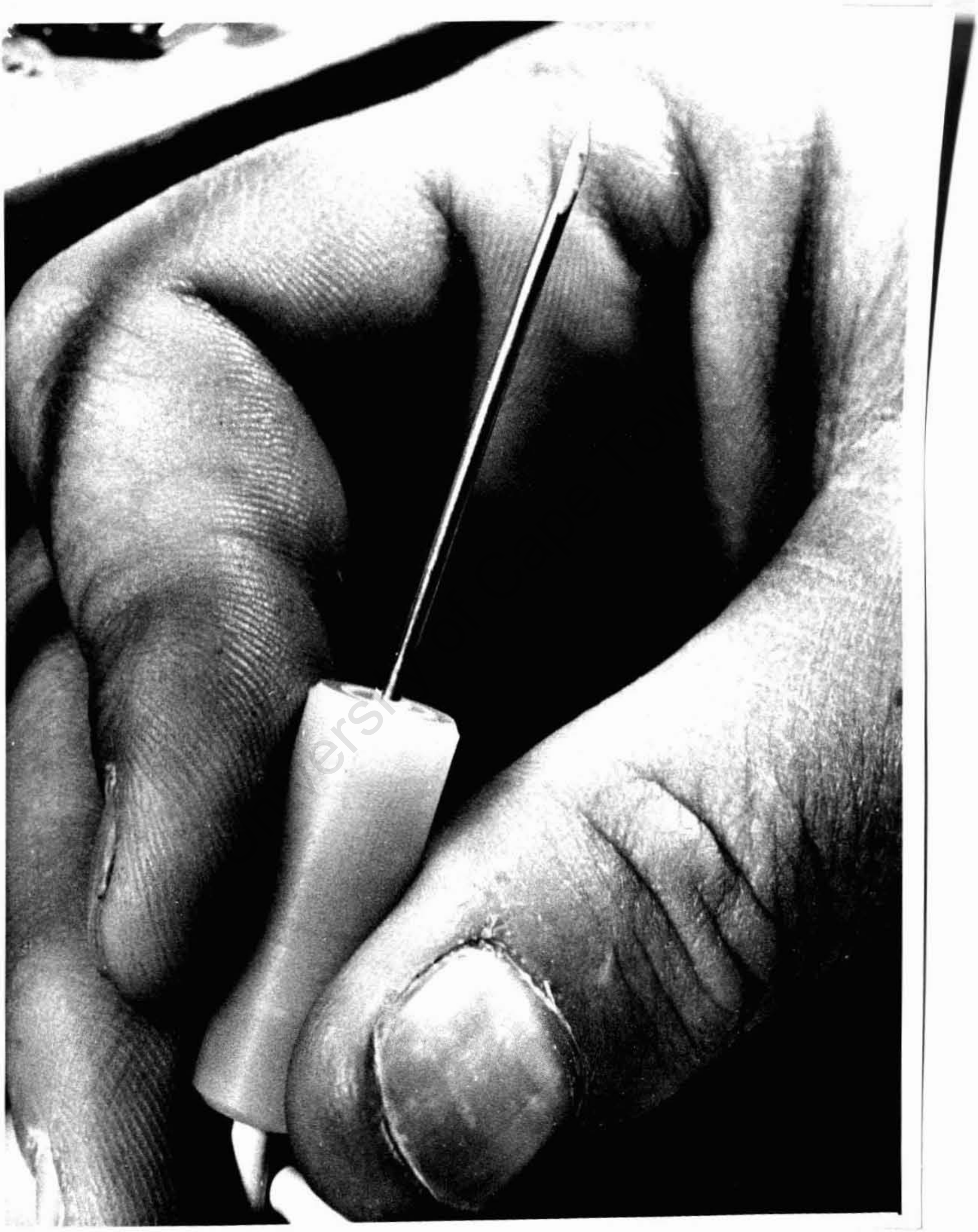
1.5 EVALUATION. There is no doubt that expertise is built up through experience. Probes were built on three occasions and the following page contains a brief summary of the successes. Only on the last two occasions was the method as outlined above used.

/... Summary

Probe Assembly Jig



3.B.5



P9

Occasion	Number of Probes attempted	Number of Thermistors broken or damaged	Number of Probes completed	Faults if any	Insulation Resist
1	2	5	2	Resin softened when in contact with saline for protracted periods. Poor cleavage to stainless steel	5Mohm
2	3	2	2	White epoxy used hence probes were useful for temperature measurements only. (white epoxy absorbs water (0.2)%)	5Mohm
3	5	0	5	Slightly variable resin thickness between thermistor and fluid flow otherwise none	10Mohm

It would appear that the technique is good and could probably be introduced on to a production basis with very few aids. In the following section an evaluation of the effectiveness of the encapsulation will be discussed.

1.6. FUTURE WORK The prospect of using some sort of pulsed technique (which will be outlined under ("FURTHER RESEARCH") necessitates:

a) Increasing the heat conduction to the flowing medium.

b) Decreasing the thermal inertia of the probe.

c) Increasing the resistance tolerance of the temperature element.

d) Decreasing the spread of the thermal characteristics of the probes.

Correspondence was entered into with the National Electrical Engineering Research Institute of the South African Council for Scientific and Industrial Research on the possibility of their being able to make a probe as shown in Fig. 3.B.6.

Indications were that it would present no great problem and the method adopted would have been to prepare a thin metal-diffused resistor in a silicon slice, then to evaporate convenient metalised contact areas and, after dicing, mount one upside down on a prepared needle tip by means of a conducting epoxy.

Such a resistor would have a positive temperature coefficient of just less than 1% per °C and provided that it could be made small enough, it would satisfy all the requirements above.

/...SECTION 3.C.

SECTION 3.C.

CIRCUITRY.

1.0. INTRODUCTION. Two circuits were developed for the experimental work. The first, called circuit 1., consisted ot two single constant current networks - one for the flow sensor thermistor and the other for the temperature sensor thermistor - and a comparitor and amplifier. This circuit was designed so that each constant current network and the comparitor was highly flexible with regard to the biasing of the thermistors and the ratio of comparison of the voltage signals. This was deliberately done so that the performance could be explored over a large range of flow velocities and a large range of temperature differences. The circuit was used for most of the experimental work although it was decidedly not simple and easy to use.

The second circuit, called circuit 2, was developed in order to overcome a number of problems associated with the first. It consisted of a constant current bridge network with thermistors in opposite arms of the bridge. The bridge was designed to be symmetrical in all respects except that of the current flowing in each arm. The bridge was followed by a comparitor network, the ratio of comparison being rather a function of the bridge network than the comparitor. The number of circuit variables was reduced to an absolute minimum and the circuit was designed to be simple and easy to use. Metering was self-contained.

As previously noted the resistance of a thermistor obeys the law

$$R = Ae^{B/T}$$

For the thermistors used B is quoted

/... in

in the specifications as being 2900°K and $R_{(20)}$, the resistance of the thermistor at 20°C = 2kΩ.

$$\therefore \quad A = \frac{2000}{e^{\frac{2900}{293}}} = \frac{2000}{e^{9.9}}$$

$$= 0.1050$$

Now 50°C was chosen as the upper limit for the flow sensor and at this temperature

$$R = 0.1050\, e^{\frac{2900}{323}} \doteqdot 800\Omega$$

This was verified by exposing the thermistor to flowing water held at fixed temperatures. At temperatures above 50°C the rate of flow affected the resistance of the thermistor. Below 50°C no change of resistance was observed for changes in flow rate.

At this point, the bias conditions were:

Flow Sensor Maximum

RTh = 800 Ω   ITh = 7.5 mA
VTh = 6.0 volts   PTh = 17.75 mW

37°C was chosen as the upper limit for the temperature sensor so that the heat flow would always be from the blood to the thermistor. At 37°C

$$R = 0.1050 e^{\frac{2900}{310}} \doteqdot 1.2\ \text{k}\Omega$$

A current of 0.60 mA was chosen and hence the

/...bias

bias conditions were:

Temperature Sensor Maximum

RTh = 1.2 kΩ
VTh = 0.72 volts
ITh = 0.60 mA
PTh = 0.43 mW

Further,

$$^{dR}/_{dT} = - Re^{B/T^2}$$

This works out for small excursions from a fixed bias point (50°C and 37°C in this example) to be 23.2 Ω/°K for the flow sensor and 35.2 Ω/°K for the temperature sensor.

$$^{dV}/_{dT} = {}^{dV}/_{dR} \cdot {}^{dR}/_{dT} = I \, {}^{dR}/_{dT}$$

which is: 174 mV/°K for the flow sensor and 21.1 mV/°K for the temperature sensor.

Therefore, for small excursions about T the gain difference in the two paths must be 174/21.1 = 8.25 for approximate voltage tracking.

1.1. CIRCUIT 1.

1.1.1. DESIGN PHILOSOPHY.

The constant current networks were designed around a simple single transistor current source which, despite all its inherent temperature dependence limitations, was deemed sufficiently accurate for a first experimental network.

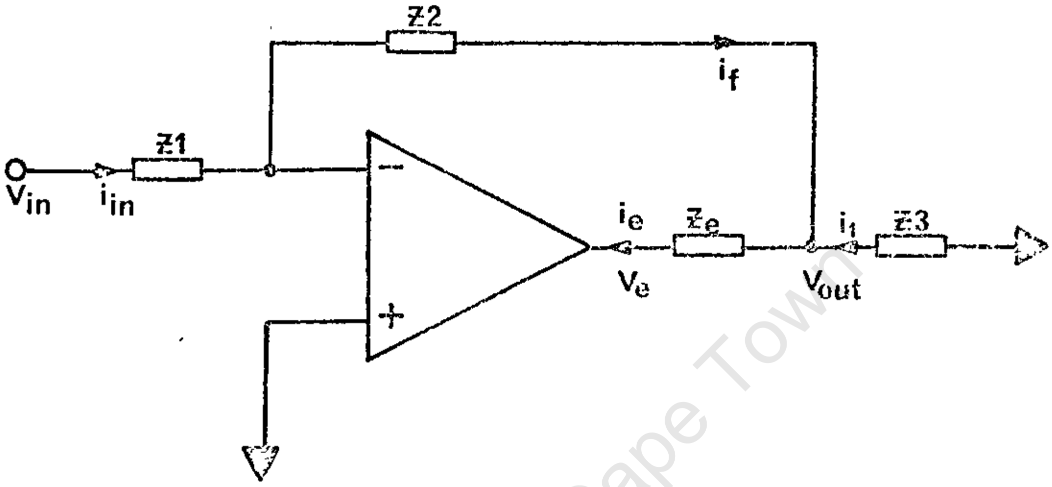
The circuit is shown in Figure 3.C.1., and the design details are included at the back of this work in Addendum 3.

/... 1.2. CIRCUIT 2.

1.2. CIRCUIT 2.

1.2.1. DESIGN PHILOSOPHY.

Consider the ideal circuit below:



$$V\,out = \alpha_1\,V\,in$$

$$= Z2/Z1.V\,in$$

$$\frac{V\,in}{Z1} = i_f \text{ and } \frac{V\,out}{Z3} = i_1$$

$$i_1 + i_f = i_e = \frac{V\,in}{Z1} + \frac{V\,out}{Z3}$$

$$= \frac{V\,in}{Z1} + \frac{Z2\,V\,in}{Z1Z3}$$

$$= V\,in\left[\frac{1}{Z1} + \frac{Z2}{Z1Z3}\right]$$

/...hence

hence i_e is independent of Z_e and is a function of V in and Z1, Z2 and Z3 only. This circuit provides a very accurate current source and is realisable in practice if:

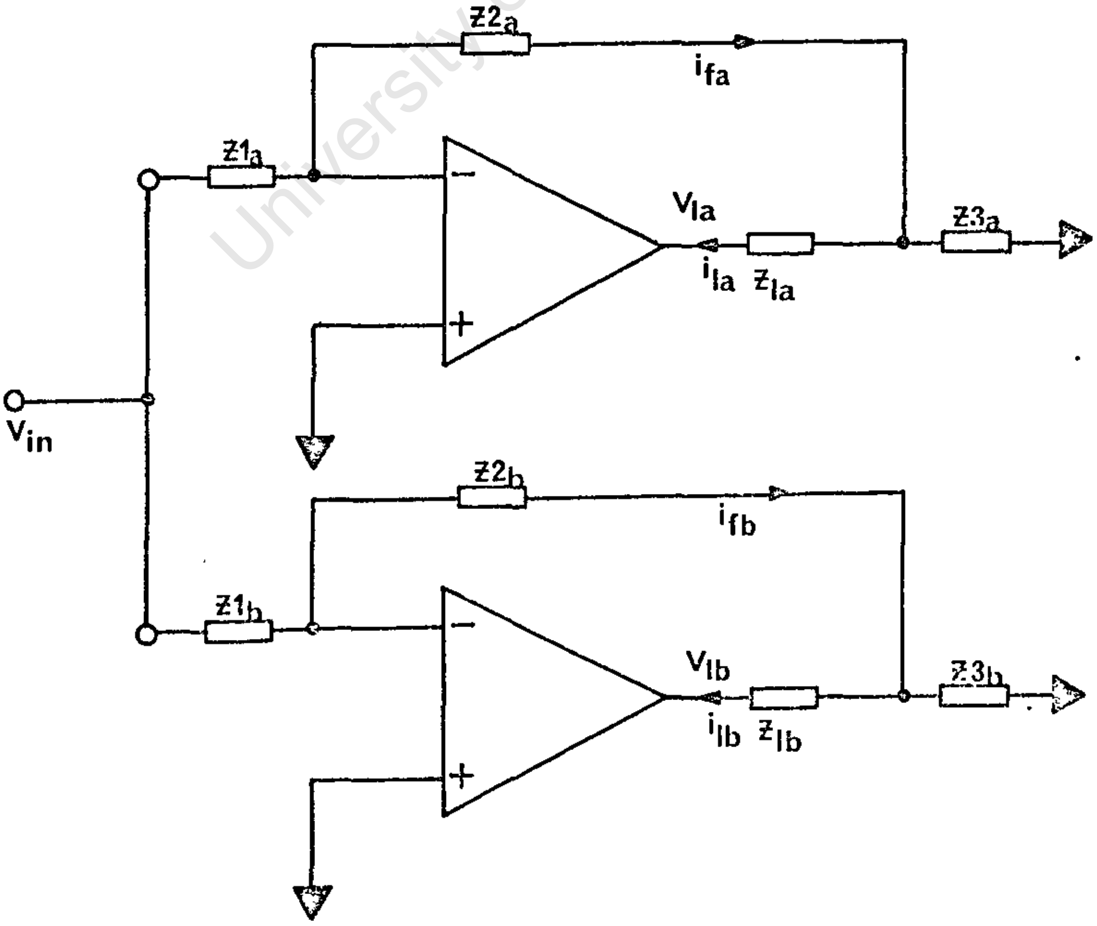
a) The input offset voltage is very small in comparison with V in.

b) The input offset current is very small in comparison with i_f.

c) The voltages i_1 Z3 + $(i_f + i_1)$ Z_e is within the linear capabilities of the amplifiers.

d) The impedances Z1, Z2 and Z3 are of such values as not to violate the stability criteria.

Consider further the following bridge circuit:



/...if

if i_{fa} and i_{fb} are small compared with i_{1a} and i_{1b} and if

$$\frac{1}{Z_{1a}} + \frac{Z_{2a}}{Z_{1a}Z_{3a}} = \frac{1}{Z_{1b}} + \frac{Z_{2b}}{Z_{1b}Z_{3b}}$$

then $i_{1a} = i_{1b}$

and if $Z_{1a} = Z_{1b}$

then $V_{1a} = V_{1b}$

and the bridge is in balance irrespective of the values of $Z_{1a} = Z_{1b}$ and V in

Further, if

$Z_{1a} = Z_{1b}$

$Z_{2a} = Z_{2b}$

$Z_{3a} = Z_{3b}$

then the output of the bridge

$$V_{1a} - V_{1b} = (i_{1a}\ Z_{1a} + i_{1a}\ Z3_{a} - i_{1b}\ Z_{1b} + i_{1b}\ Z3_{b})$$

$$= i_{1a}\ Z_{1a} + \frac{Vin\ Z_{2a}}{Z1_{a}\ Z_{3a}} - i_{1b}\ Z_{1b} + \frac{VinZ2_{b}}{Z1_{b}Z3_{b}}$$

$$= Z_{1a} - Z_{1b}$$

If the two amplifiers share the same substrate there is a decided further advantage to this circuit and, that is, that any thermal drift is likely to reflect in a common mode voltage i.e. if the amplifiers have a thermal

/...drift

drift, say, V_{td} then

$$V_{1a} + V_{td} = V_{1b} + V_{td}$$

and provided that the differential amplifier which follows has a high common mode rejection

then $(V_{1a} + V_{td} - (V_{1b} + V_{td})$

$$= V_{1a} - V_{1b}$$

The circuit is shown in Figure 3.C.2. & Figure 3.C.3. and the design details are included at the back of this work in Addendum 4.

Picture (P1) shows the assembled bridge. Addendum 4.

1.3. TEMPERATURE COMPENSATION.

From Section 3.A. Theory it was shown that if V is measured with I constant ΔT will vary with the flow rate. Hence, it is not valid to fix α, the gain difference between the flow sensor and temperature sensor for a fixed current I in the flow sensor.

Generally

if $$\left.dV_1/dT_1\right|_{T_1} = \alpha\, dV_2/dT_2$$

then $$dV_1/d\ T_1 + \Delta T \neq \alpha\, dV_2/dT_2$$

where V_1 is the voltage across the flow sensor and. T_1 and $T_1 + \Delta T$ are two temperatures which the flow sensor might experience due to the cooling fluid flow. V_2 is the voltage across the temperature sensor, fixed at a temperature of T_2.

There is only one way to achieve accurate temperature

/...compensation.

compensation and that is to hold the resistance of the flow sensor constant and to measure its power dissipation

i.e. $\frac{V}{I} = k$ measures V X I

However, let us consider Circuit 2.

For I = 4.7 mA for the flow sensor,

$$\frac{dV}{dT} \doteq 140\ \text{mV}/{}^{o}\text{K} \quad @ \quad \Delta T = 5°\text{C}$$

and

$$\frac{dV}{dT} \doteq 160\ \text{mV}/{}^{o}\text{K} \quad @ \quad \Delta T = 0°\text{C}$$

For I = 1.5 mA for the temperature sensor,

$$\frac{dV}{dT} = 53\ \text{mV}/{}^{o}\text{K} \quad @ \quad 37°\text{C}$$

The gain difference is 3, hence:

$$\frac{dV}{dT}\ \text{out} = 140 - 160 = 20\ \text{mV}/{}^{o}\text{K}$$

$$\text{or} \quad 160 - 160 = 0\ \text{mV}/{}^{o}\text{K}$$

hence the error is 20 mV/^{o}K which almost represents a first order approximation to accurate temperature compensation. Further, for any value of I for the flow sensor lower than 4.5 mA the temperature sensor current may be altered to preserve the first order compensation.

Fortunately, the body regulates its temperature via a number of control loop mechanisms rather well. In fact, in large internal organs and tissue well below the surface of the skin, it is debatable whether temperature compensation is or is not required.

/... 1.4. EVALUATION

1.4 EVALUATION. As an experimental circuit, the first was highly effective. The constant current regulation was found to be accurate to within 2%. Temperature compensation could be adjusted by means of the variable gain of its associated amplifier. All experiments with this circuit were made under conditions of simulated flow only. A constant temperature water bath was used, and because all the fluid flow was passed through a 2m 0.8cm i.d. copper tubular heat exchanger, temperature compensation was generally found to be unnecessary. The temperature compensation amplifier was operated at very low gain. Output drift (long term) was found to be high. However this must have been due to amplifier thermal drift and was of the order of 5% of F.S.D. in 15 minutes - tested by substituting fixed resistors in place of the thermistors.

Circuit 2 was found to be stable to within 1% of F.S.D. at highest sensitivity and was simple and easy to use.

Pictures and circuits used are shown at the back of this thesis together with detailed design data (see Addendum 3 and Addendum 4.)

Section 4

SECTION 4. A. & B.

EXPERIMENTS AND RESULTS.

1.0. OBJECTIVE.

In this section four experiments and their results are described. The primary objective of the first three of these was to determine practically, the relationship between the flow velocity past a heated thermistor situated in a cooling fluid stream and the voltage across the thermistor for differing values of constant current through the thermistor. The secondary objective was to investigate the practicality of the method with special emphasis on frequency response, drift, stability and the features desirable for a practical instrument. The final experiment was carried out on the practical bridge network which was designed to embody the requirements and features brought to light by the previous experiments.

1.1. INITIAL APPARATUS.

1.1.1. THE FLOW CIRCUIT.

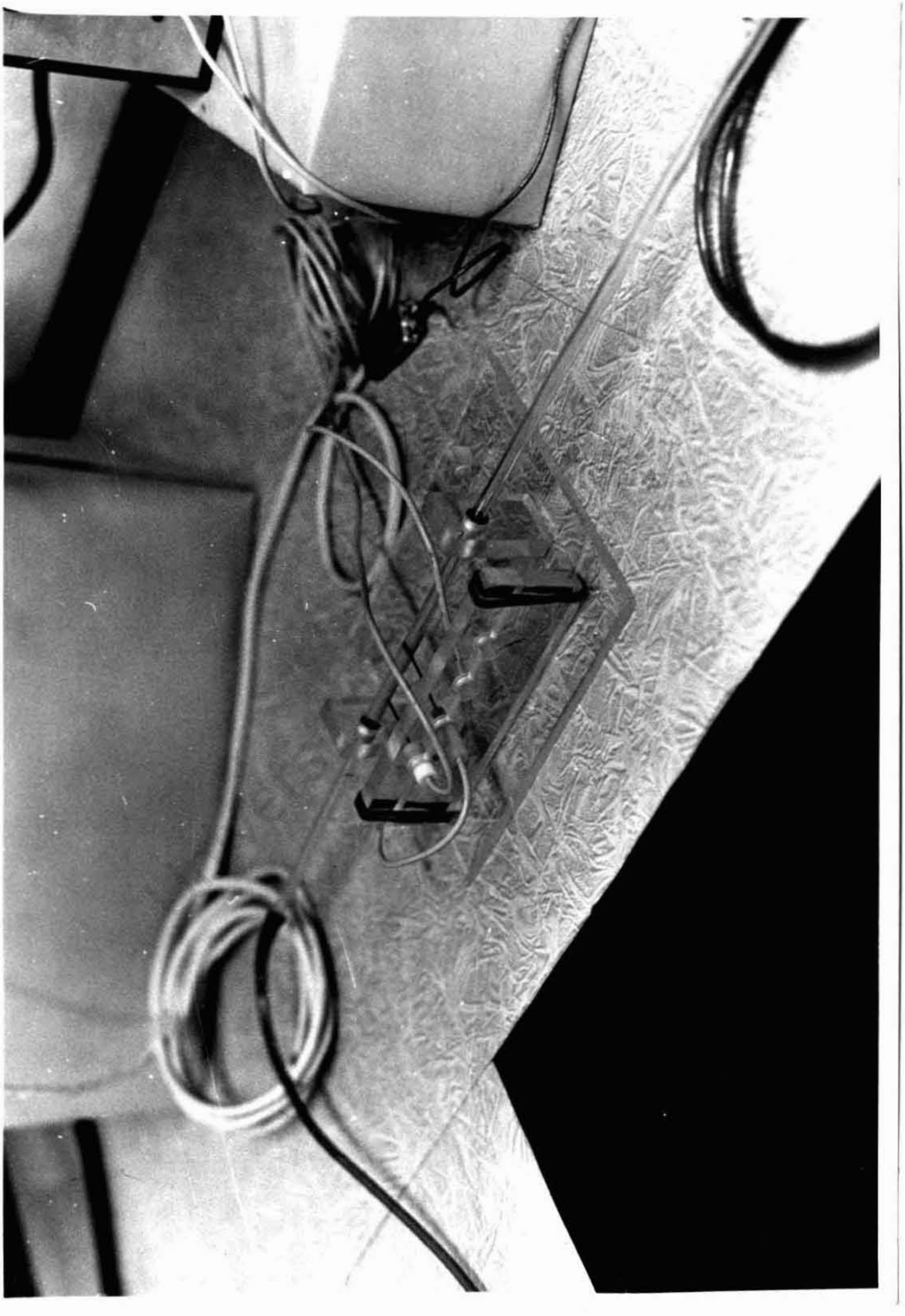
The flow circuit was an open loop fed initially by a saline drip-set. Saline was used as the flowing medium because of the electromagnetic flowmeter used in the circuit and because saline simulates the thermal properties of blood as discussed before. The flow adjustment was made by means of the knurled wheel-cock provided with the drip-set which was found to be far superior to a pinch-cock for linearly regulating the flow velocity. A heat exchanger made by winding a 0.8mm i.d. 2 meter long copper tube into a coil of 10cm o.d. was incorporated into the flow circuit well before the measuring tube so that the whole apparatus could be submerged in a constant temperature water bath in order to simulate the in-vivo temperatures. The heat exchanger was designed to be adequate for a flow velocity of 25cm per sec and a 25°C rise in temperature for normal saline. [41]

An 8mm i.d. cannular electromagnetic flowmeter probe was incorporated into the flow circuit external to the water bath and with a long section of flexible tubing either side of it. (P2) The measuring tube consisted of a 20cm straight section of $^3/8$ inch i.d. uniform bore perspex tubing. 18 s.w.g. holes were bored at right angles to the axis of the tube, a third of the overall length from the output end, to accommodate the thermistor probe needles. The needles were inserted so that the thermistors were positioned approximately on the axis of the measuring tube. A perspex jig held the measuring tube, probe bodies and heat exchanger in rigid relative positions. The flexible tubing of the output of the flow circuit was clamped in a retort stand at a fixed height relative to the input so as to maintain a constant head of saline. The output fed a measuring cylinder in a drain bath. All joints were made to be very tight fits and were chamfered to limit the possibility of turbulence. Under all conditions, theoretically, the flow was laminar.

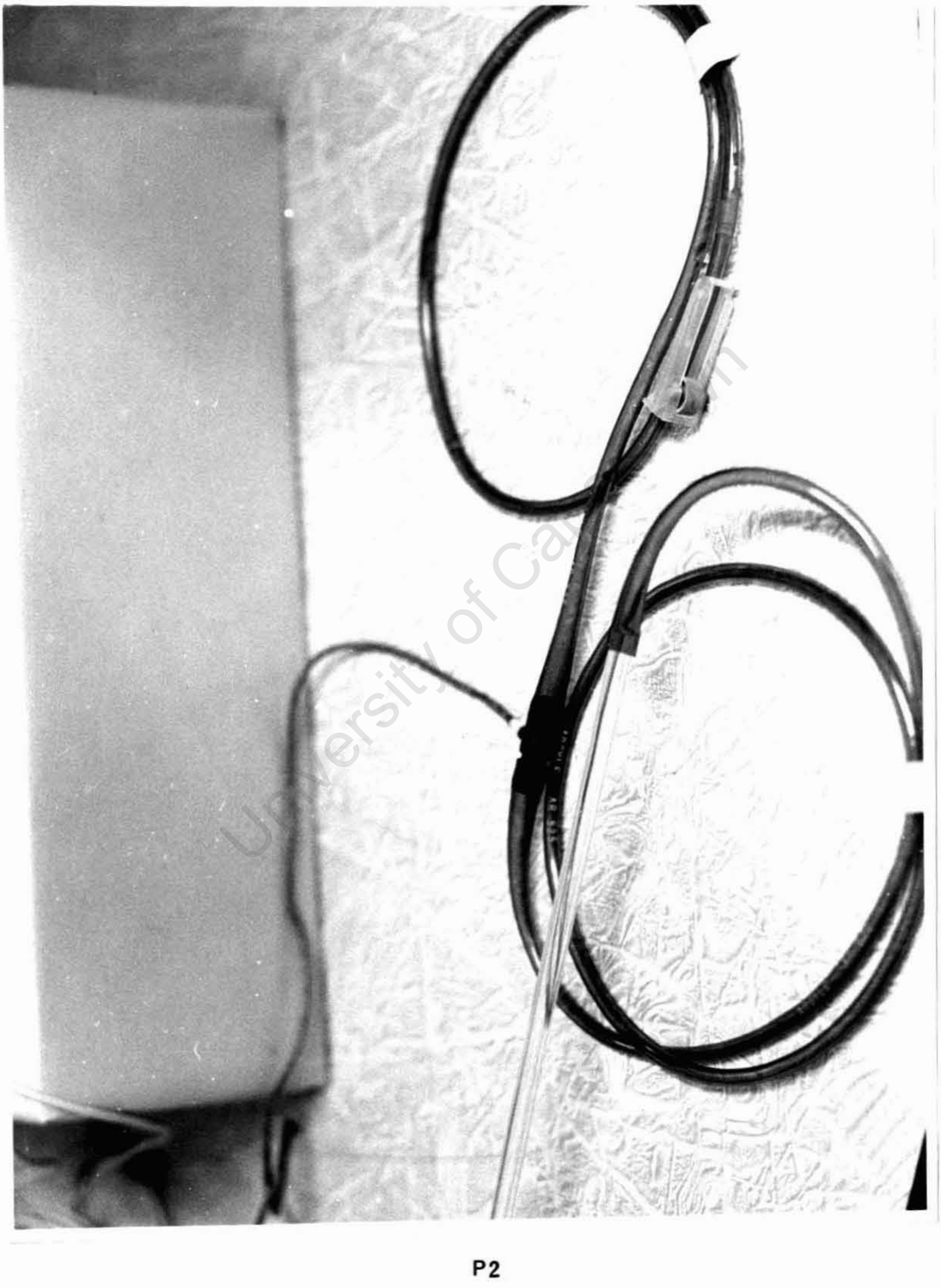
Before each use of the apparatus it had to be purged with saline so as to remove all traces of air bubbles as these affect both the electromagnetic flowmeter and the thermistor flowmeter so much that there was a possibility of damaging the galvanometers in the U.V. Recorder. (P 1) shows the perspex jig with the flow probes in position.

1.1.2. THE ELECTRIC CIRCUIT. Circuit 1 was used throughout the first three sets of experiments. The output was matched to one of the U.V. Recorder galvanometers and the special galvanometer output on the electromagnetic flowmeter was matched to a second galvanometer on the U.V. Recorder. Circuit 1 output was monitored on an external multimeter (10MΩ 30pF) and the electromagnetic flowmeter was monitored by means of its internal monitor meter. Care

/...was



P1



P2

was taken to ensure correct matching of the signals to the galvanometers for critical damping.

1.2. EXPERIMENT (1). This was the first attempt to relate the flow past the thermistor to the voltage across the thermistor for fixed excitation currents. Both the electromagnetic flowmeter and the thermistor bridge were placed in the flow circuit and absolute flow rates were determined by taking the time for a discharged quantity of saline. The volumetric flow rate was converted into a linear flowrate. Voltage - an arbitrary meter reading - was recorded against absolute flow velocity for three increasing values of constant current

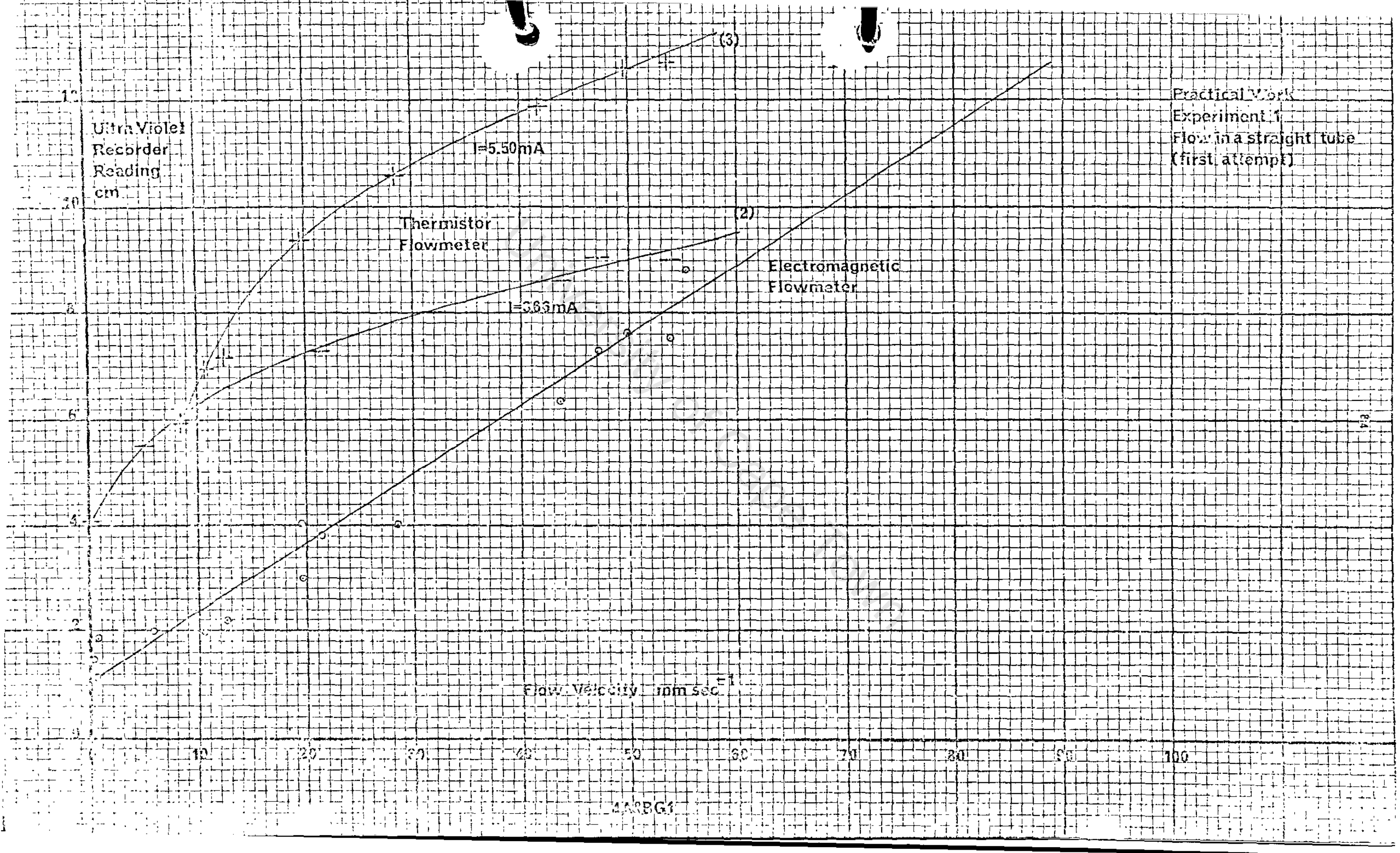
i.e. 2.50mA
3.66mA
5.50mA

Also, a strip-chart was obtained for a slow stepped increase in flow from zero to 2cm/sec for comparison of the two flowmeters.

U.V. strip-charts were taken for all three graphs and the flow was closed off after each reading. Information for the graphs plotted was obtained from the U.V. charts, since the relationship between the voltage across the thermistor and the galvanometer deflection is linear.

1.3.1. RESULTS. Two of the above three graphs were plotted and are shown on 4 A & B G1 together with the graph for the electromagnetic flowmeter. S1 and S2 show three sections of strip-chart. The first (A) is a section from the chart used for plotting the curve of I = 5.50mA and shows the electromagnetic flowmeter vrs. the thermistor flowmeter for two values of flow viz. 41.5mm/sec and 49.5mm/sec. The second (B) strip-chart shows a section from the chart used for plotting the curve of I = 3.66mA

/...and


Practical Work
Experiment 1
Flow in a straight tube
(first attempt)
Ultra Violet
Recorder
Reading
cm
12
10
8
6
4
2
0
(3)
I=5.50mA
(2)
Thermistor
Flowmeter
I=3.63mA
Electromagnetic
Flowmeter
Flow Velocity mm sec^{-1}
0
10
20
30
40
50
60
70
80
90
100
4ARBG1

and shows both flowmeters for two values of flow viz. 31.7mm/sec and 42.6mm/sec. The third chart (C) shows the slow stepped increase in flowrate for the two flowmeters.

1.3.2. DISCUSSION. As a first experiment these results were gratifying to a certain extent. The writer was surprised to find that the frequency response of the thermistor flowmeter was as high as is shown in Chart 1. The results were not used for any quantitative assessment of the flowmeter but did, however, point out a number of modifications which would improve further results. These were as follows:-

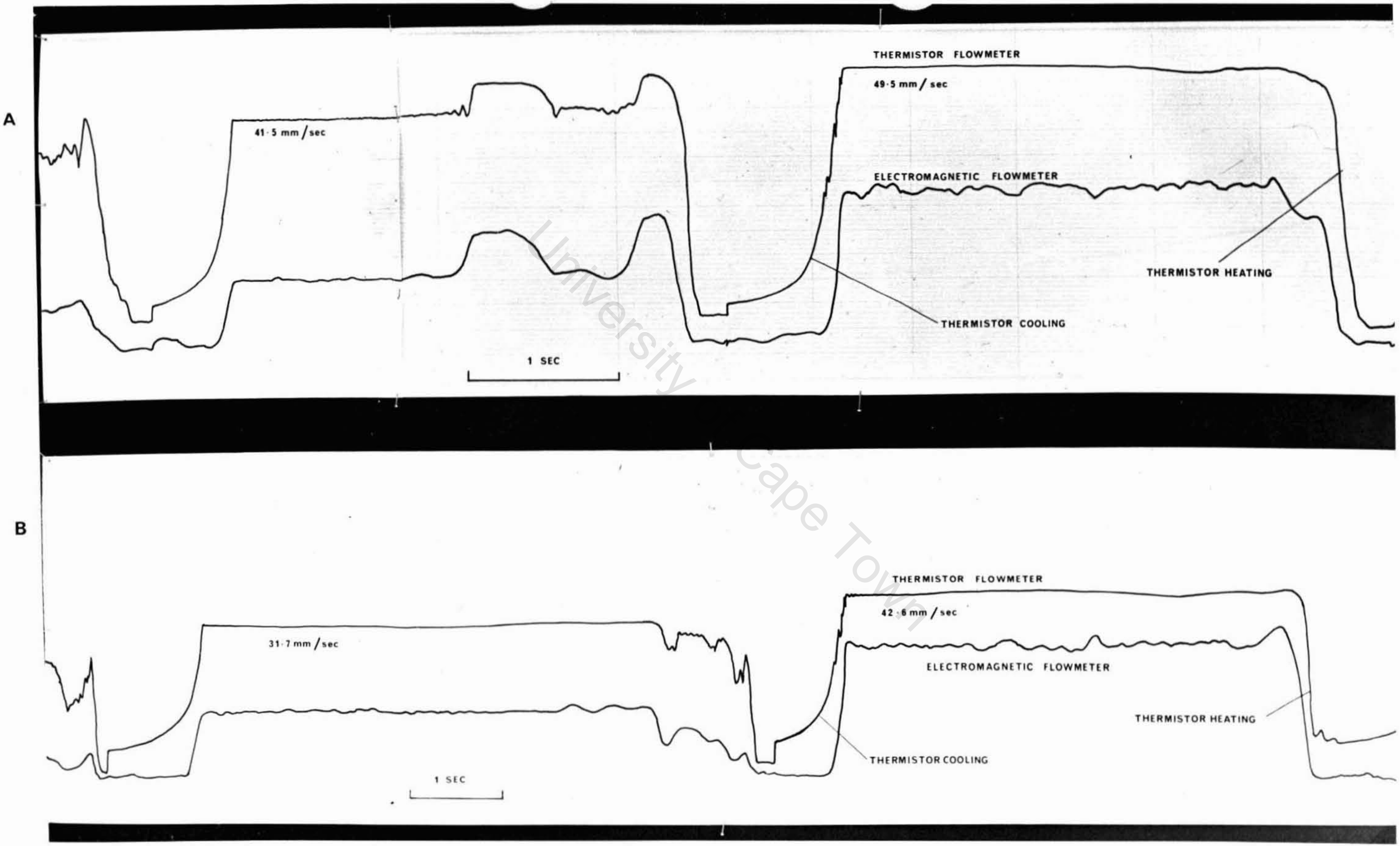
A) It was found that slight movements of the interconnecting rubber tubing would reflect along the measuring tube and interfere with results. The soft rubber tubing was eliminated and "oxygen bubble tubing" which is far more rigid was used. Further, the tubing was secured in place so that the stirrer motor of the constant temperature water bath could not move it. Oxygen bubble tubing can, of course, also be cut to a selected diameter for convenience in interconnecting.

B) The measuring tube was replaced by a similar tube 50cm long and 0.6cm i.d. so that suspected turbulence effects would be reduced and the range of linear flow velocities increased.

C) The probes were sealed in position so as to prevent the saline from discharging slowly from the 18 s.w.g. press-fit holes bored into the measuring tube.

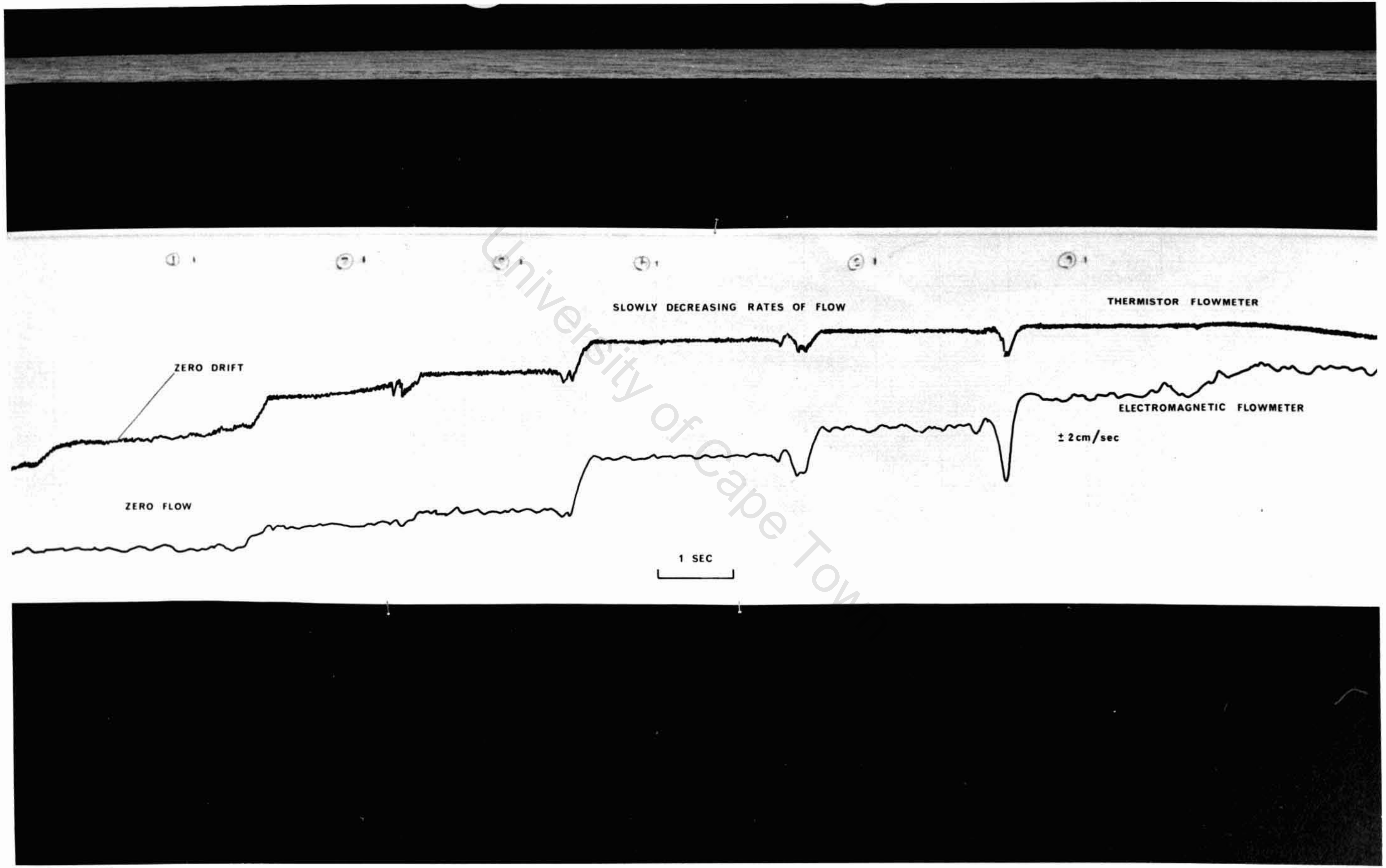
D) The 8mm i.d. cannular electromagnetic flowmeter probe was found to be too large for use at such low flowrates. (At this stage of the project, the 8mm probe was on loan pending the arrival of the 5mmprobe ordered).

/... E) It


A
THERMISTOR FLOWMETER
41·5 mm / sec
49·5 mm / sec
ELECTROMAGNETIC FLOWMETER
THERMISTOR HEATING
THERMISTOR COOLING
1 SEC
B
THERMISTOR FLOWMETER
31·7 mm / sec
42·6 mm / sec
ELECTROMAGNETIC FLOWMETER
THERMISTOR HEATING
THERMISTOR COOLING
1 SEC

S1

C


①
SLOWLY DECREASING RATES OF FLOW
THERMISTOR FLOWMETER
ZERO DRIFT
ELECTROMAGNETIC FLOWMETER
± 2cm/sec
ZERO FLOW
1 SEC

S2

E) It was found to be quite obvious that there was little hope of obtaining a zero on the thermistor bridge for zero flowrate since the heated thermistor continues to heat its surrounds long after the flow has stopped and, in fact, it requires several minutes before an equilibrium state is achieved. Hence it was decided that, for differing values of constant current, the thermistor bridge would be balanced to zero at a particular value of flow - small - so that the families of curves could be compared on this basis.

F) Cables were screened to eliminate noise.

<u>1.3 EXPERIMENT (2).</u> This was the first serious attempt to obtain a family of curves relating the flow velocity to the voltage output of the bridge. The electromagnetic flowmeter was temporarily removed from the circuit as it was of no value with the 8 mm i.d. cannular probe.

The U. V. chart recorder was run only during the period in which readings were taken, - (± 1 to 3 minutes each). The five increasing values of constant current were as follows:-

7.32mA
6.10mA
4.97mA
3.66mA
2.44mA

The flowrate with the new measuring tube was 0 to 2cm/sec., and the bridge was balanced at 0.4cm/sec for each value of current.

<u>1.3.1. RESULTS.</u> As the first set of useful results obtained these were highly gratifying. The graph 4 A & B G2 shows the family of curves extracted from the strip-charts. The shape of the curve is strikingly similar to that predicted by the theory. Two sections of strip-chart are shown on S3. (A) is the chart from

which the curve of 3.66mA was obtained - this shows that the readings were substantially constant over a long time period. (B) is the chart from which the curve of 7.32mA was obtained and is included to show the sensitivity and response at low flowrates in the form of the ripple on the graph which was caused by the drops falling from the discharge end of the flow circuit. In this particular case the droplet frequency was 0.66 and 1.2/sec.

1.3.2. DISCUSSION. These results were certainly good enough to warrant an attempt to correlate them with the theory. However, before this was attempted, it was decided to further modify the technique so as to obtain a wider range. The further modifications were as follows:-

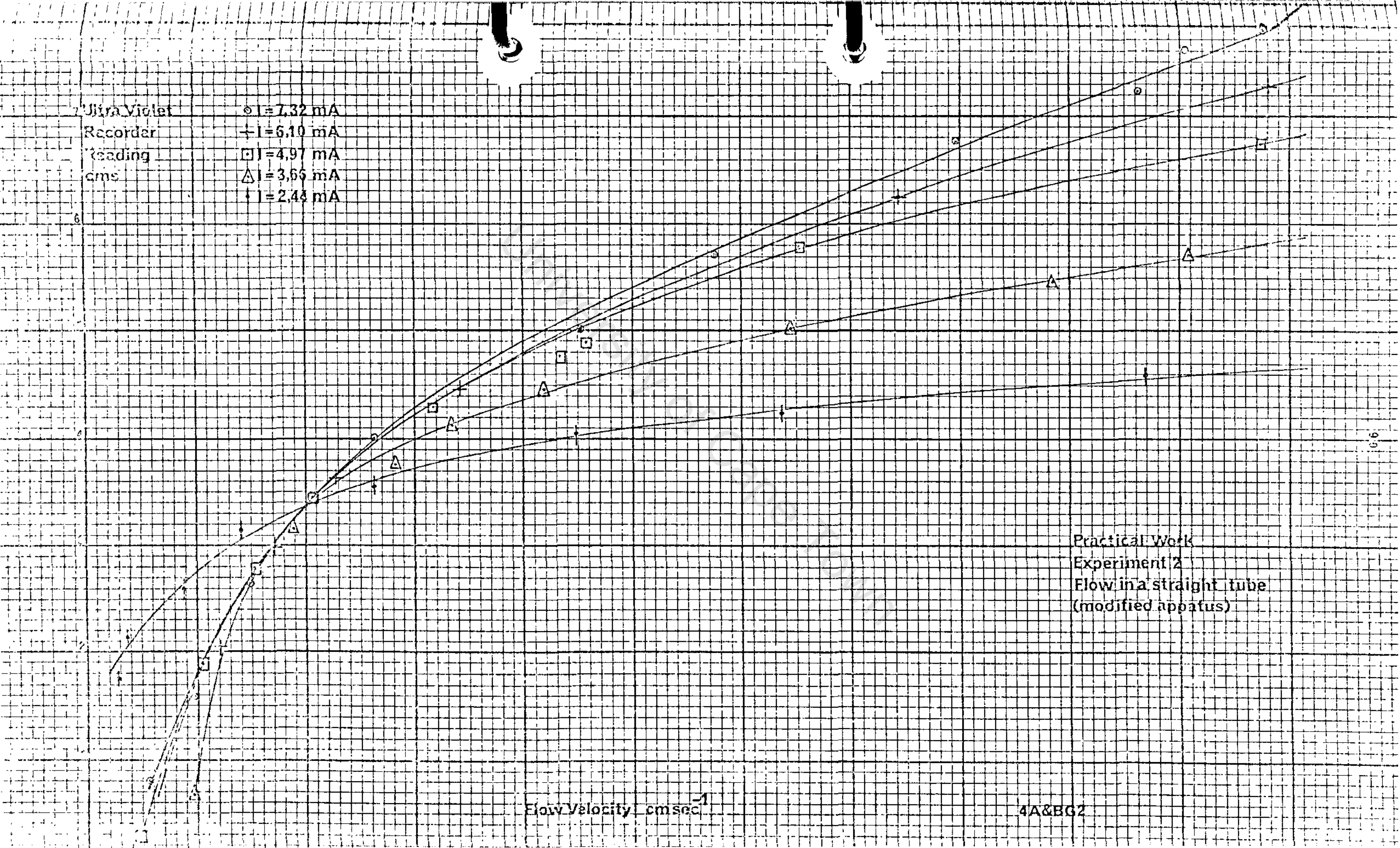
A) The 8mm i.d. cannular probe was returned and the 5mm i.d. probe, which was ordered for this work, used.

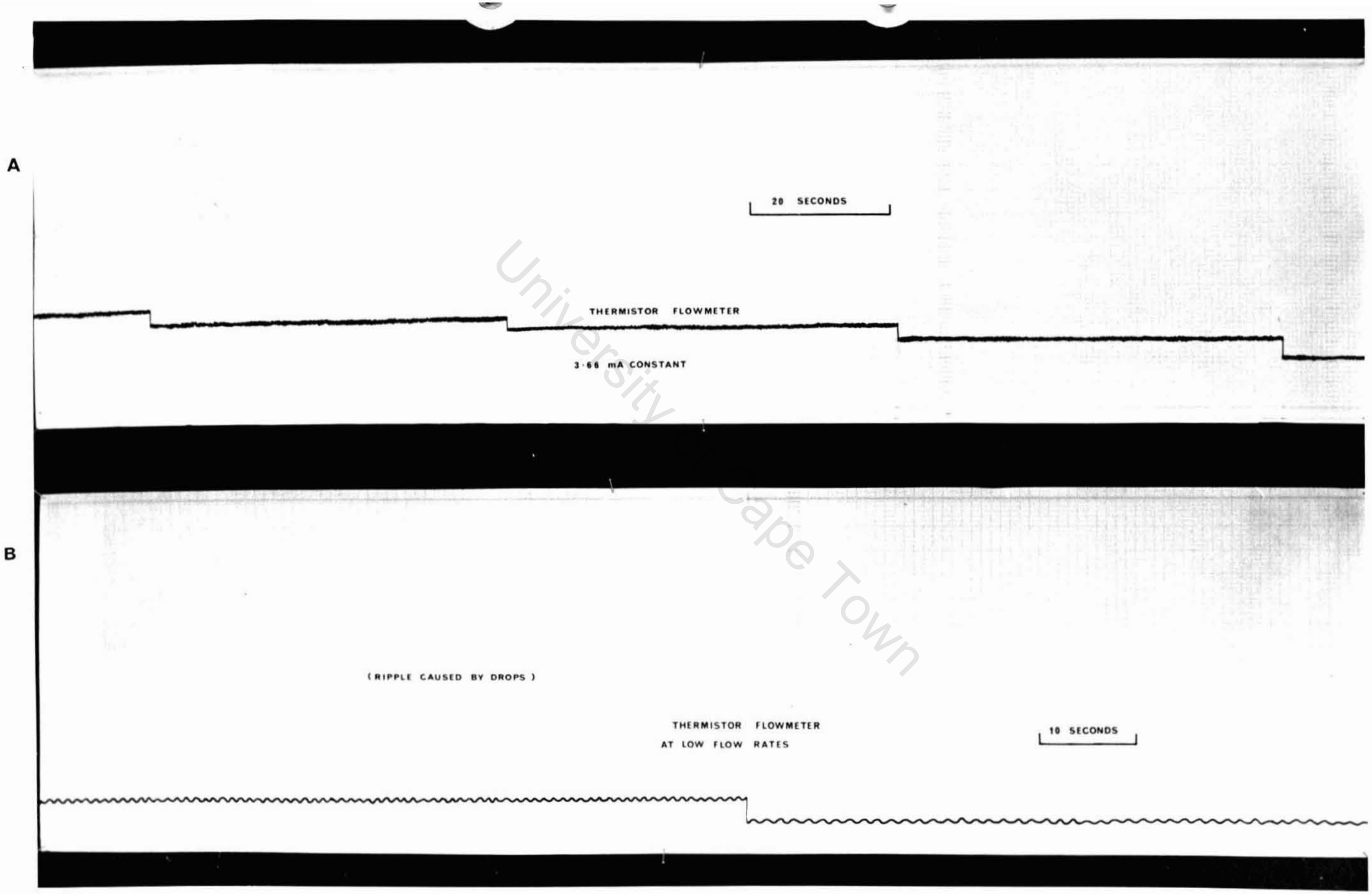
B) A baffle plate was placed between the stirrer and the flow apparatus to prevent the still present effect of the water movement in the water bath. This precaution was found to be quite adequate.

C) In order to achieve the higher flowrate sought and to investigate the frequency response, the measuring tube was replaced by a 50cm long 4mm i.d. uniform bone perspex tube and the saline drip-set was replaced by a 10 litre aspirator placed nominally 1 meter above the flow apparatus. (The knurled wheel-cock was retained because of the ease of regulating the flow).

1.4 EXPERIMENT (3). Again a family of curves was obtained relating the flow velocity to the bridge output voltage. This time the electromagnetic flowmeter was used and found to be excellent at the higher flowrates. It was not, however, used as the standard and absolute flowrates were again taken. It did, however, serve the very useful purpose of a reference for the frequency response estimation and it, in turn, was calibrated against

/...absolute


Ultra Violet
Recorder
Reading
cms
⊙ I = 7,32 mA
+ I = 6,10 mA
⊡ I = 4,97 mA
△ I = 3,66 mA
ǂ I = 2,44 mA
6
4
2
Flow Velocity cm sec^{-1}
Practical Work
Experiment 2
Flow in a straight tube
(modified appatus)
4A&BG2


A
20 SECONDS
THERMISTOR FLOWMETER
3·66 mA CONSTANT
B
(RIPPLE CAUSED BY DROPS)
THERMISTOR FLOWMETER
AT LOW FLOW RATES
10 SECONDS

S3

absolute flow for this purpose. The chart recorder was operated for all measurements and the results extracted from it. Four values of constant current were taken as follows:-

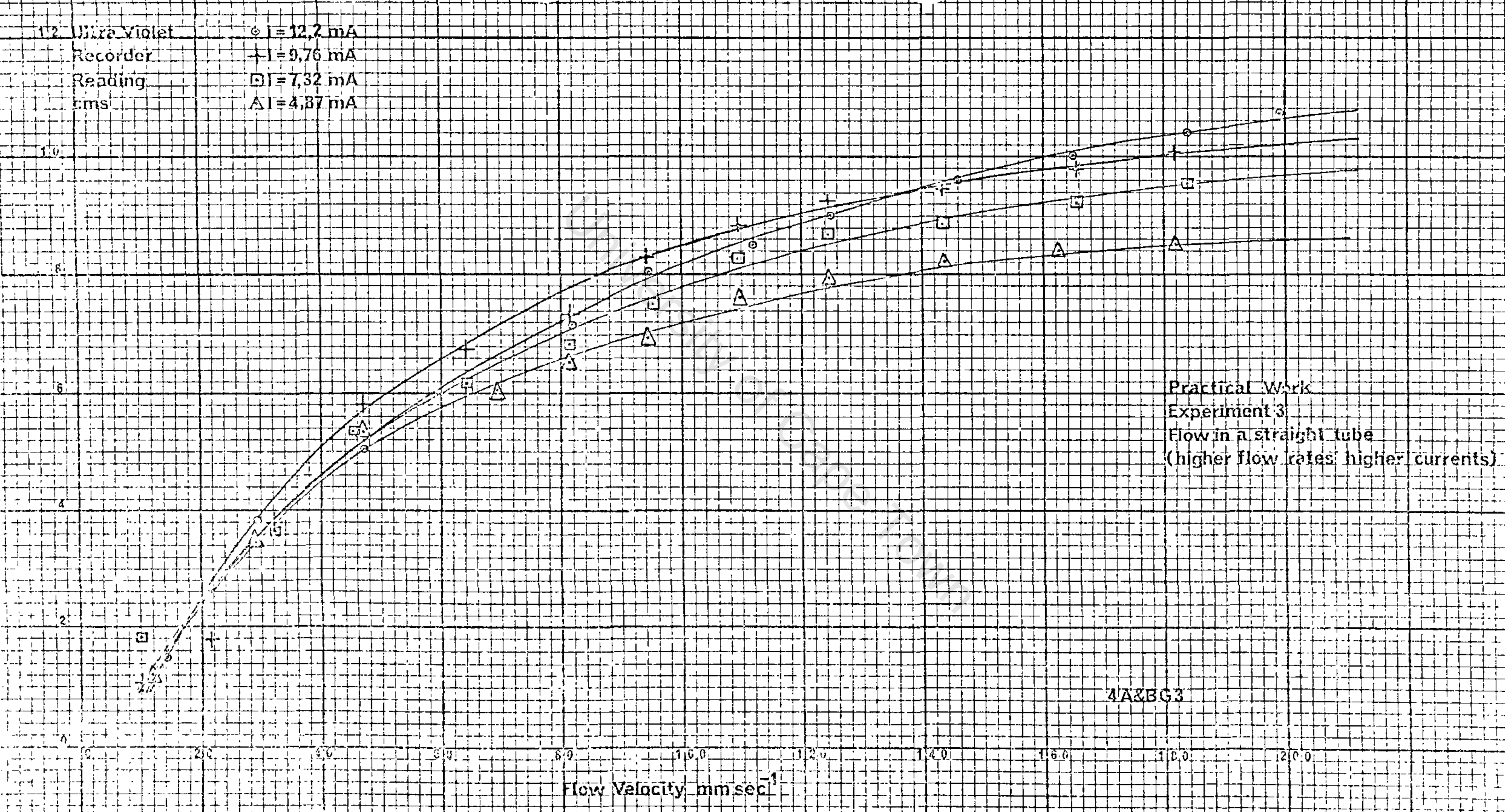
12.20mA
9.76mA
7.32mA
4.87mA

The flowrate with the new measuring tube and 10 litre aspirator was from 0 to 20cm/sec.

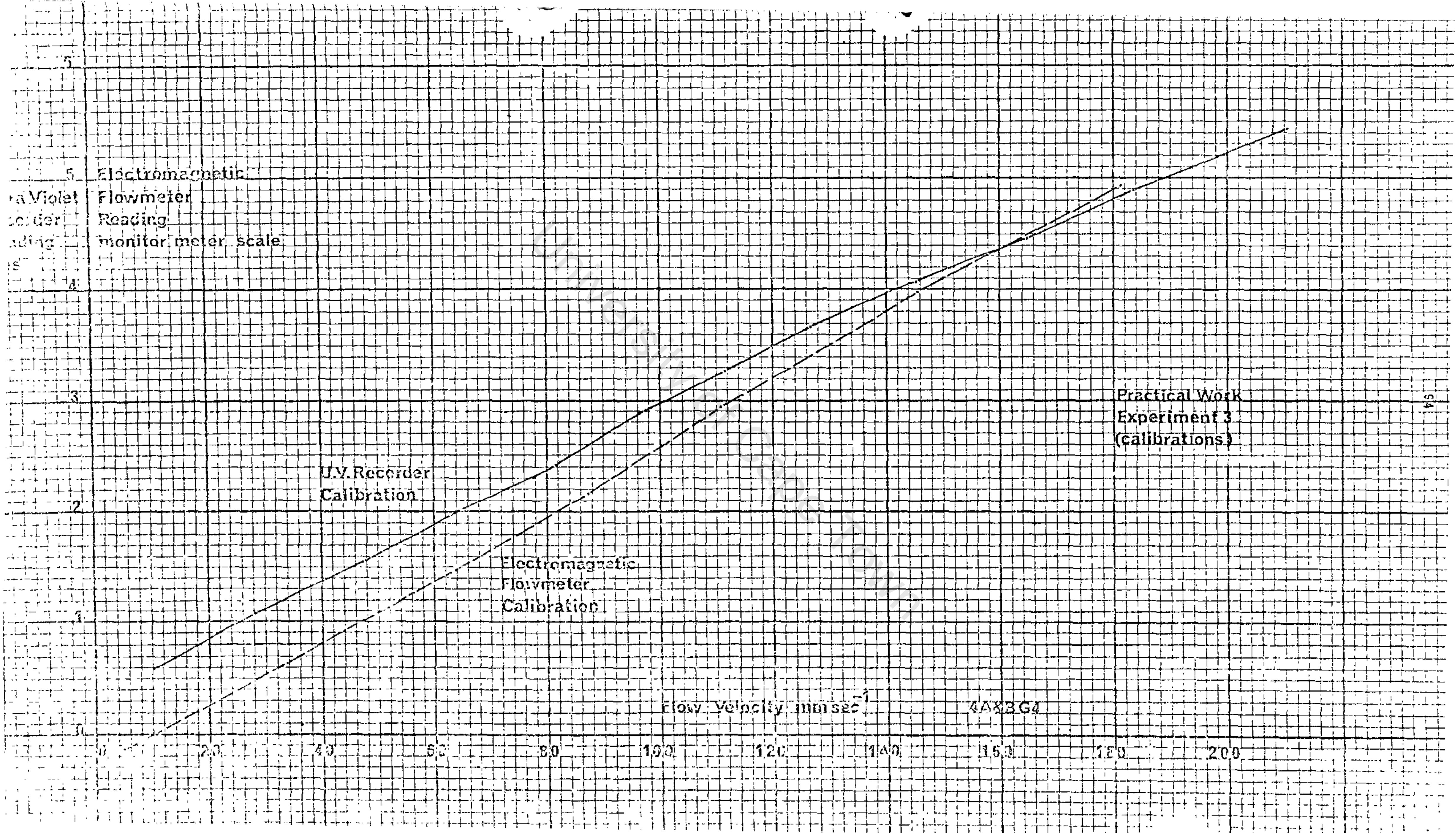
1.4.1. RESULTS. Again results were pleasing. Graph 4 A & B G3 shows the family of curves obtained for the higher flowrates and greater current - the amplification of the bridge was left constant but the signals had to be slightly attenuated to match the U.V. Chart recorder. Graph 4 A & B G4 shows both the U.V. Recorder calibration and the electromagnetic flowmeter calibration. S1 shows the charts for the U.V. Recorder calibration and for the electromagnetic flowmeter calibration. Superimposed on the latter chart is the thermistor bridge response at an arbitrary current setting. This is a particularly good comparison for both meters at fixed flowrates. S.2 shows excerpts from the charts taken to determine the frequency response of the thermistor flowmeter.

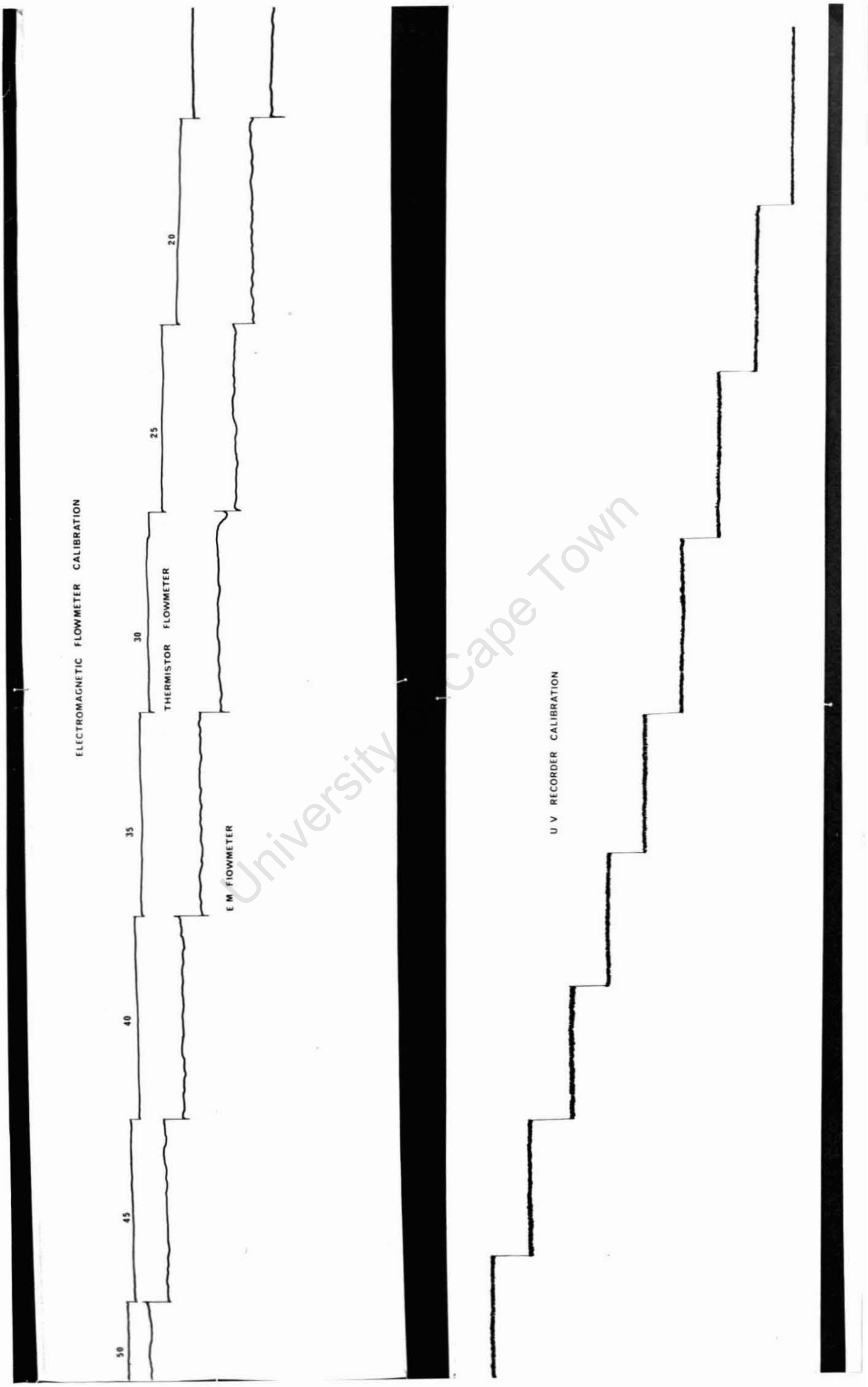
1.4.2. DISCUSSION. These results were deemed sufficiently consistent to attempt the correlation of practical and theoretical results, and the design and construction of circuit 2 was started. No further modifications to the flow circuit were considered necessary with the exception of two simulated tissue flow experiments.

A) A 10cm length of foam sponge was introduced into a measuring tube at the measuring point. (shown in P3). The sponge was constricted to approximately $^1/4$ of its natural volume by the measuring tube. It was

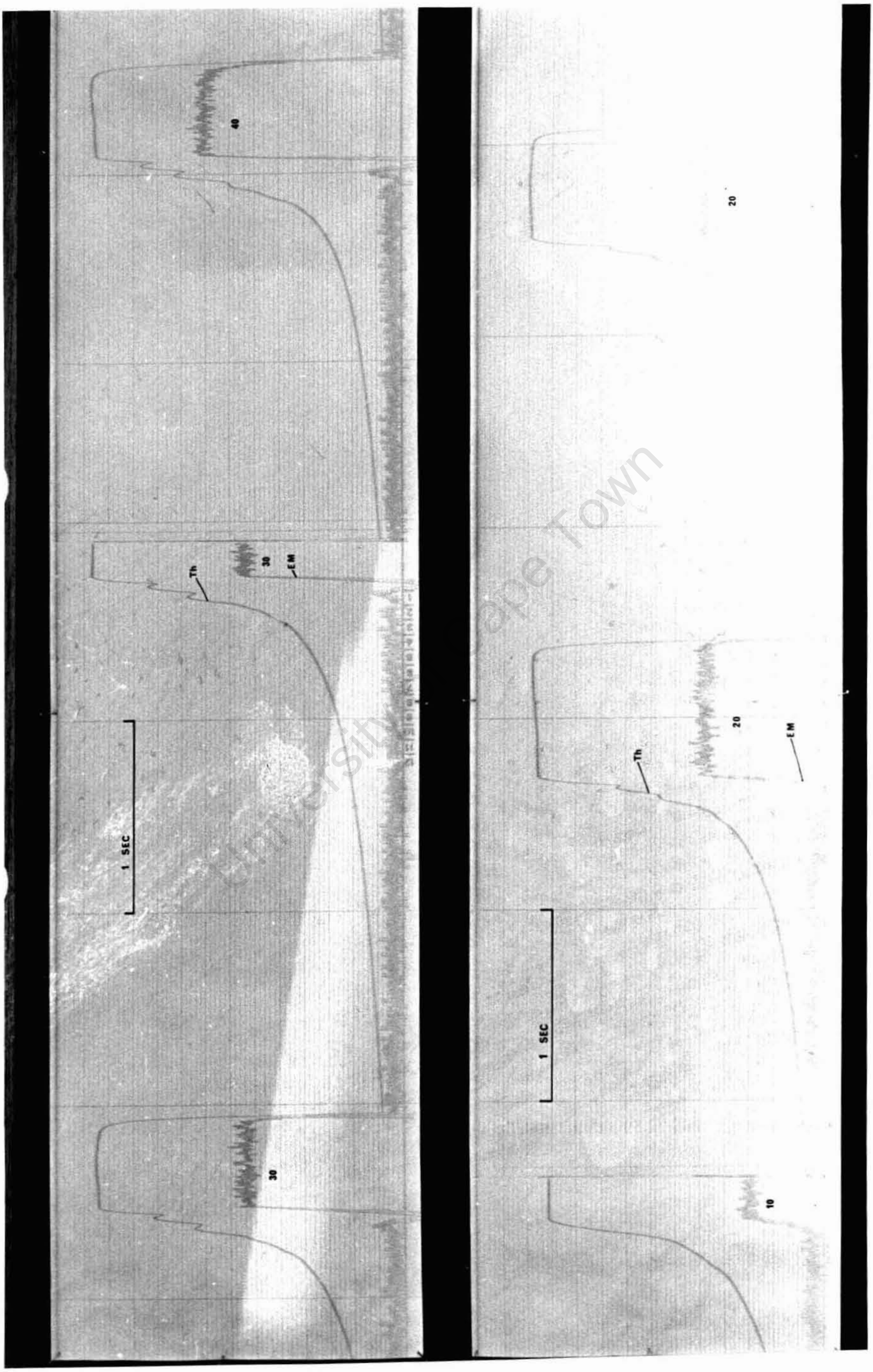


Practical Work
Experiment 3
Flow in a straight tube
(higher flow rates: higher currents)


Practical Work
Experiment 3
(calibrations)
a Violet
corder
ading
s
Electromagnetic
Flowmeter
Reading
monitor meter scale
U.V. Recorder
Calibration
Electromagnetic
Flowmeter
Calibration
Flow Velocity mm sec^{-1}
4A&BG4
0
1
2
3
4
5
6
0
20
40
60
80
100
120
140
160
180
200


ELECTROMAGNETIC FLOWMETER CALIBRATION
50
45
40
35
30
25
20
THERMISTOR FLOWMETER
E M FlOWMETER
U V RECORDER CALIBRATION

S1


1 SEC
30
Th
30
EM
40
10
1 SEC
Th
20
EM
20

estimated that this would increase the density of the medium to that of tissue and that approximately 50% of the thermistor would be covered by the sponge matrix. The diffusion through the sponge would be random and similar to the diffusion of blood through highly vascular tissue.

B) In order to estimate the capabilities of the instrument with regard to the topographical flow in tissue, a rectangular sponge was placed inside a rectangular perspex box of 2" X 3" X 4" internal dimensions, - (shown in P4). The box was purged with saline, placed vertically in a retort stand and perfused with saline at a constant rate. An 8 X 8 matrix of 18 s.w.g. holes was bored in one face of the box and the holes were covered with Elastoplast (R). The temperature sensor probe was placed in reference 1,1 and the flow sensor probe introduced into each hole in turn with the bevelled end of the hypodermic needle facing upwards. It was hoped that a flow pattern of the distribution of flow in the box could be obtained.

1.5 EXPERIMENT (4).

The new practical flowmeter having been built, a set of four tests was performed:-

(1) A family of curves for absolute flow versus thermistor flowmeter meter reading for three different values of constant current were taken viz.

2.5mA
3.0mA
3.5mA

The flow velocity range was 0 to 15cm/sec.

(2) A family of curves for absolute flow versus thermistor flowmeter meter reading for the above same three constant currents were taken for the flow velocity range 0 to 5cm/sec.

/...(3) The

(3) The sponge filled measuring tube was perfused with saline and a further family of curves obtained for the same three values of constant current. The output flow velocity range was 0 to 2cm/sec.

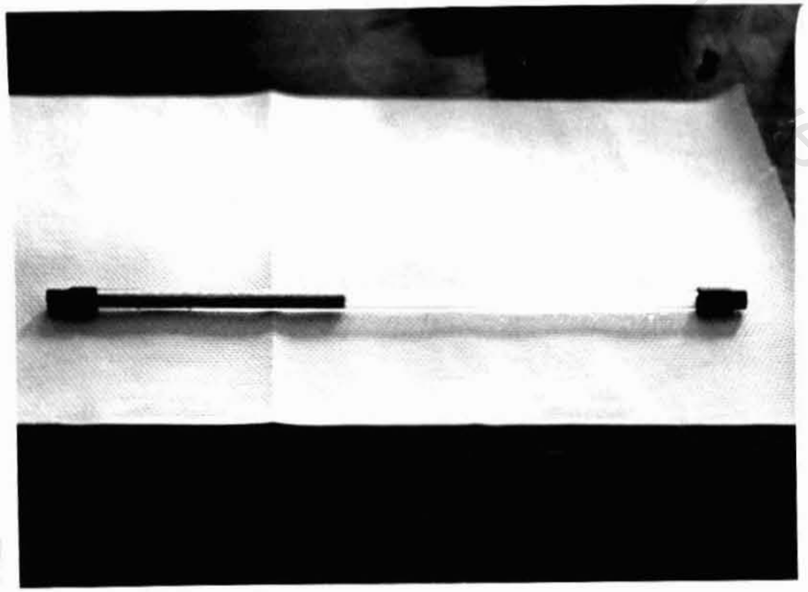
(4) The sponge box was perfused at a fixed rate of 20ml/min and the flow sensor current fixed at 3.0mA and a field plot of the flow magnitude was obtained.

The constant temperature water bath was unobtainable so all experimentation was performed at 23°C. The temperature compensation was adjusted to suit. Hence the proportionately lower constant currents. All readings were compared with absolute flow rates and the electromagnetic flowmeter was included in the flow circuit as a monitor.

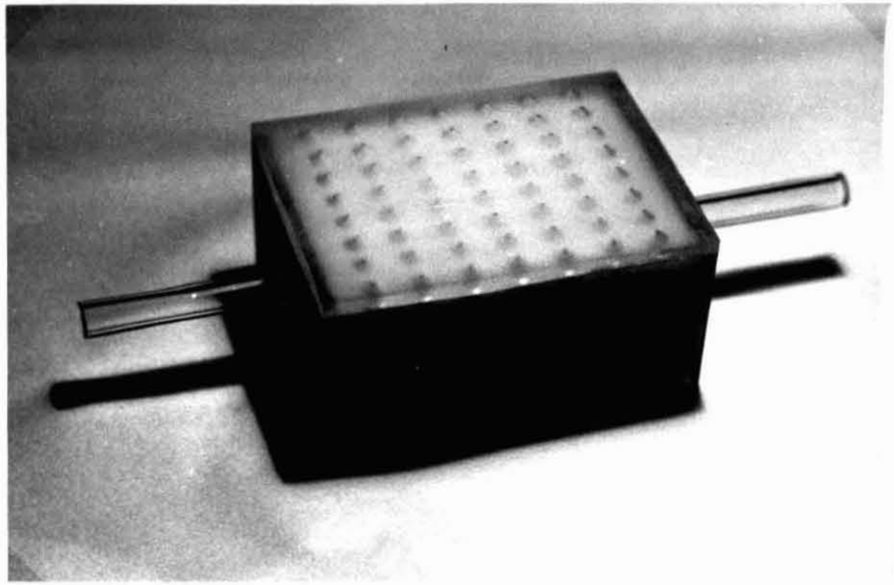
1.5.1. RESULTS. The results of the first test which are plotted on 4 A & B G5 are pleasing and were referred to 2cm/sec as zero. The gain of the flowmeter was set to "low". The results of the second test which are plotted on 4 A & B G6 were very much more erratic. The gain was set to "medium". These were taken each as a straight line and plotted on the left of the graph so that the axes crossed. This is obviously not strictly valid but does give an indication of the nature of the variation of slope with current. The results of the third test plotted on 4 A & B G7 are again erratic. However, the regular decreasing slope with decreasing current holds. The gain was set to the "high" position for this set. The flow rates were very low because of the constriction of the sponge. Graph 4 A & B G8 shows the results of the field plot and will be discussed in the next section.

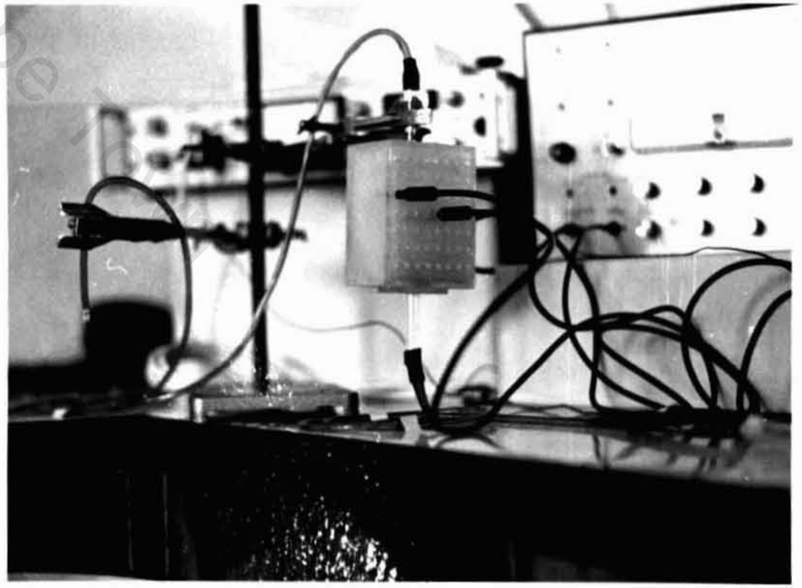
1.5.2. DISCUSSION. While the results of the four tests on the practical flowmeter were a trifle disappointing in the light of earlier results, it must be borne in mind that the flowrates used, except in the case of test 1, were

/...abnormally



P3





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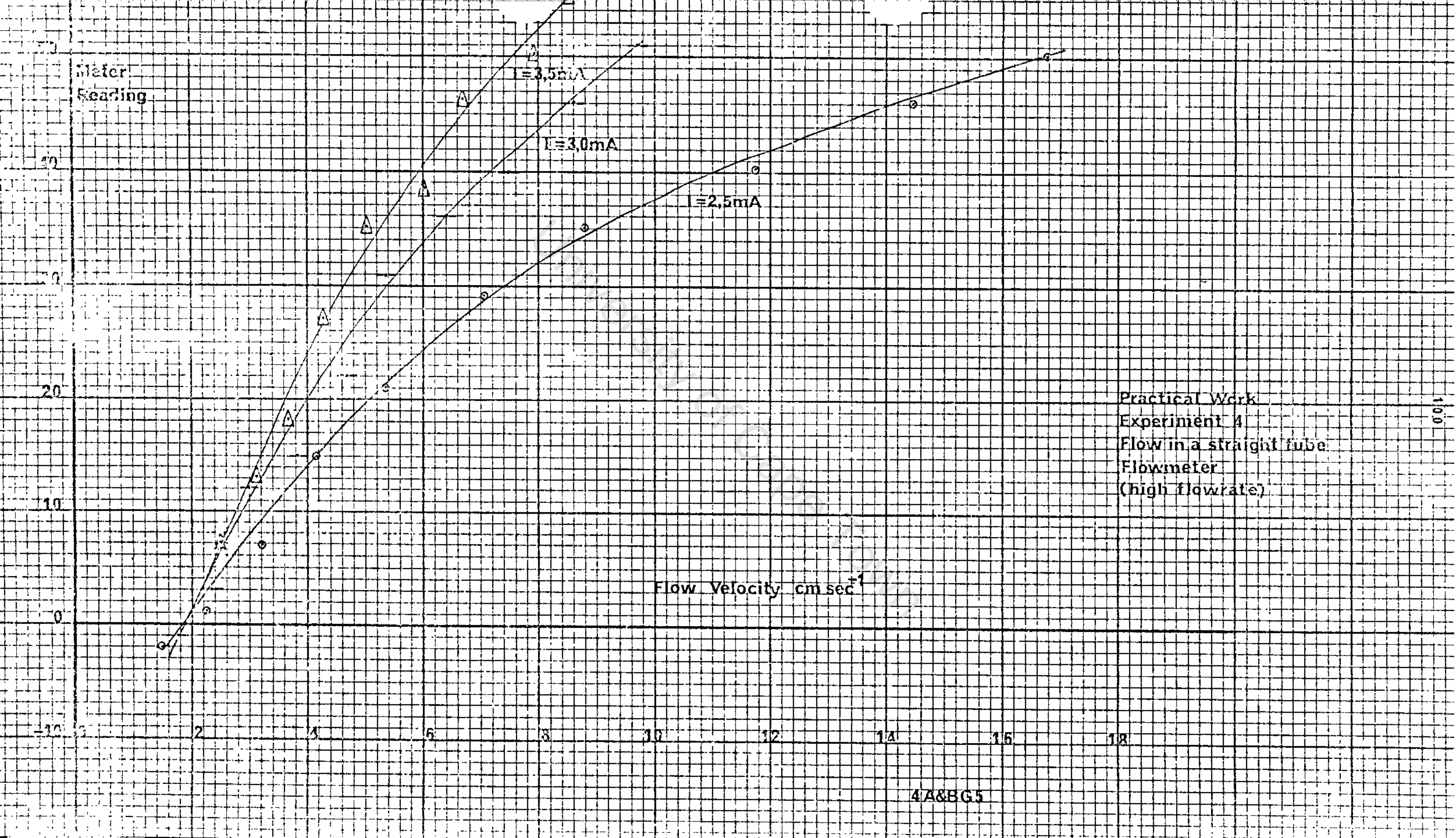
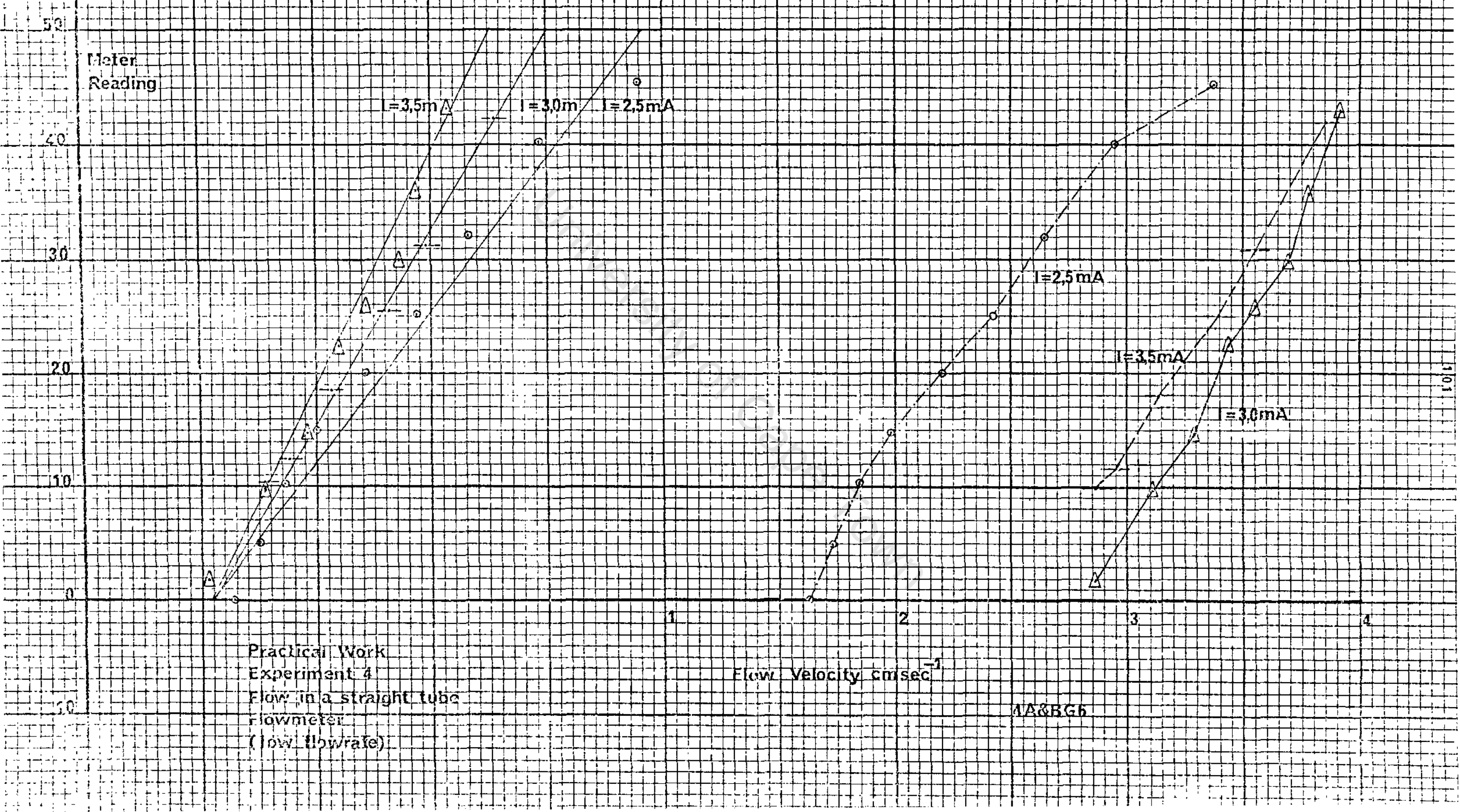
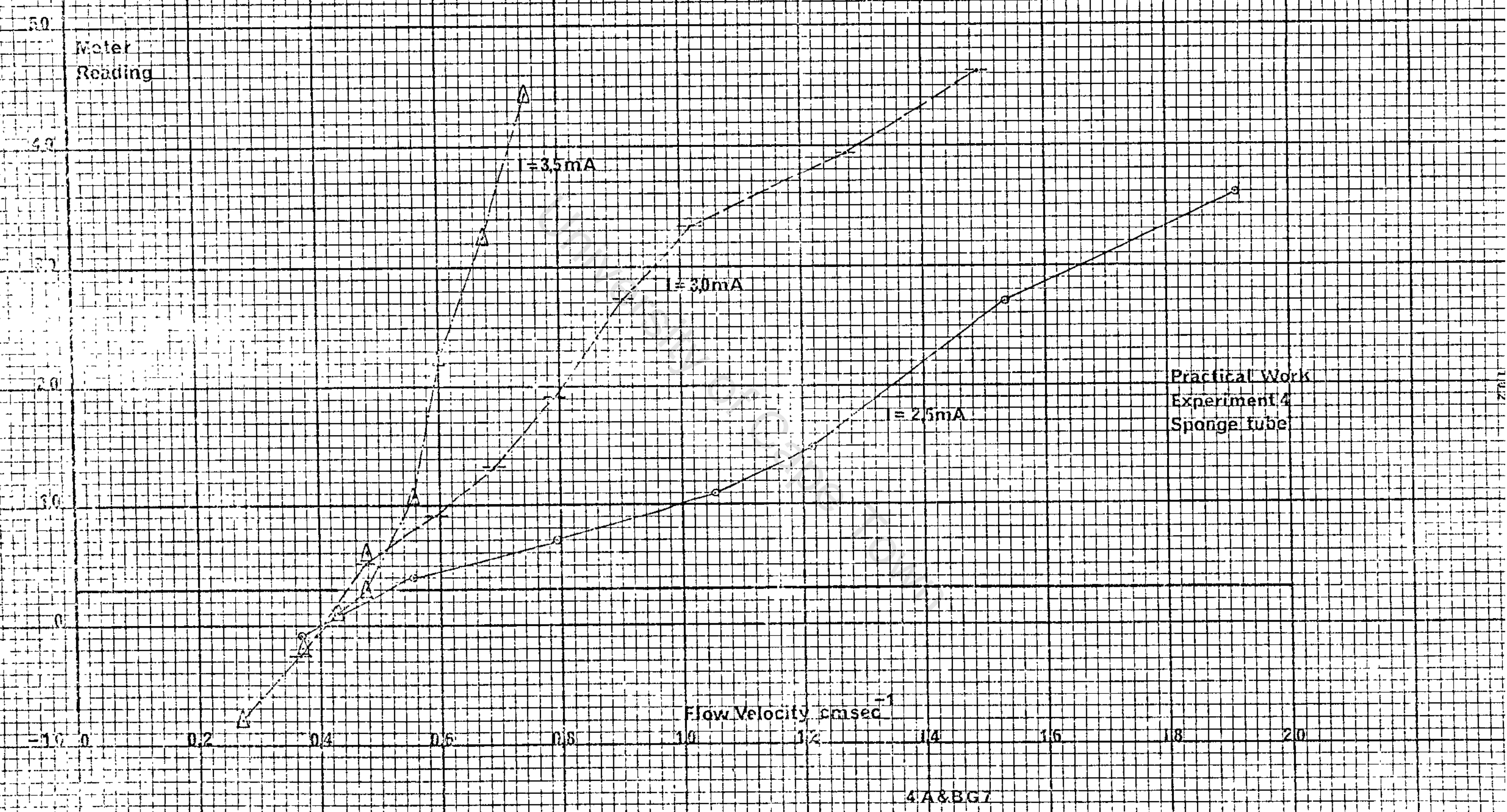


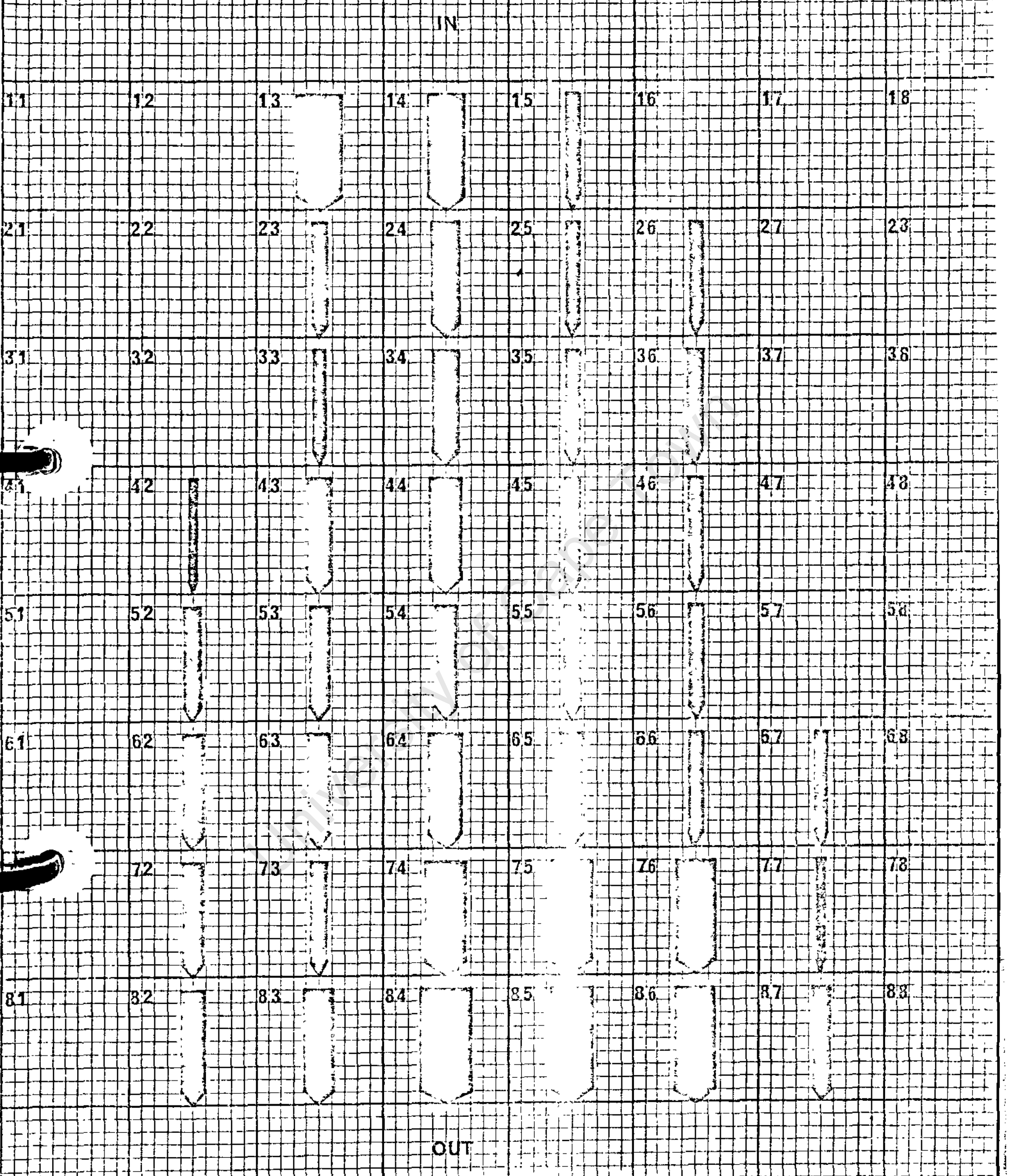
Practical Work
Experiment 4
Flow in a straight tube
Flowmeter
(high flowrate)



Practical Work
Experiment 4
Flow in a straight tube
Flowmeter
(low flowrate)



Practical Work
Experiment 4
Sponge tube



Practical Work
Experiment 4
Sponge box

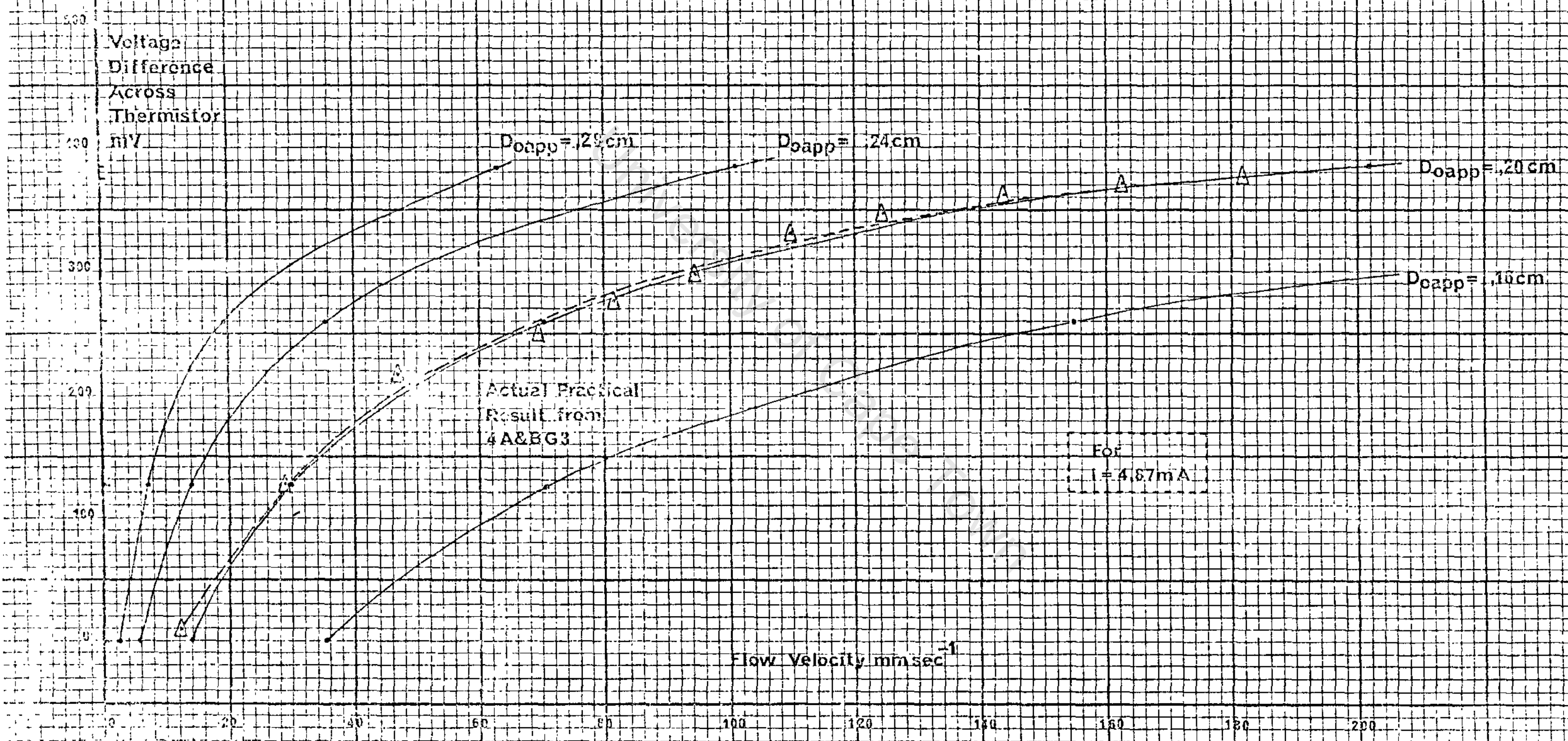
[illegible]

abnormally low and the currents set to preserve the 5°C limitation were also low. It was clear from previous experimentation that the response is very much better as flowrate and constant current increase. The electromagnetic flowmeter results were, as a matter of interest, no better than those of the thermistor flowmeter. Results in simulated in-vivo temperatures might well have proved to be more consistent.

From the flow plot shown in graph 4 A & B G8, it would appear that the greatest flowrates in the sponge box were at the bottom. In fact, in theory the reverse is true. However, it is certainly true that the saline diffusion in the upper half will be more sparse than in the lower half, - i.e. widely spaced small flow channels could easily form in the top half. The probability of obtaining a reasonable average in the top half of the box is, therefore, very low. The saline fills the bottom half of the box and therefore the probability of obtaining an average value anywhere in the lower half is high. This reasoning would certainly help to clarify the results. The experiment did prove, however, that a field plot could be obtained.

1.6 CORRELATION OF RESULTS. An attempt was made to correlate the practical results in Graph 4 A & B G3 with the theoretical predictions. If it is assumed that the basic equation as derived in the theory is correct, then the factor which alone would cause the greatest difference between the theoretical and practical results would be a great difference in $D_{o\ app}$. This seems almost intuitively understandable. Consider Graph 4 A & B G9. This is a theoretical plot of the voltage difference across the thermistor against flow velocity for four different values of $D_{o\ app}$ with the current through the thermistor held at a constant value of 4,87mA i.e. that of the fourth curve on Graph 4 A & B G3 of the practical results. It can be seen that the curves coincide for a value of $D_{o\ app}$ of 0.20cms - i.e. approximately 4 times the theoretically assumed value. It

/...would



Correlation of Results

4A&BG9

would, however, not be justifiable to assume, therefore, that the positioning and encapsulation of the thermistor increased its effective diameter by a factor of four to 0.20cm because, although the curves can be matched for all four of the practical results i.e. 7.32mA, 9.76mA and 12.2mA, the gain required to amplify the voltage difference so that the output voltages correspond is different.

To perform the correlation properly it would be necessary to relate the galvanometer deflections directly to a voltage difference across the thermistor in each current case and then home in on the $D_{o\ app}$ required to coincide the curves. At this stage, it is beyond the scope of this work.

1.7. TEMPERATURE COMPENSATION. The temperature compensation for both of the circuits was checked by randomly changing the environmental temperature of both probes simultaneously. In both cases it was found that the output voltage change for common mode temperature changes smaller than 1°C was at least an order of magnitude less with the temperature compensation in.

1.6. A LIVER PERFUSION. Although, at this stage, there was very little time left for experimentation, it was decided to perform two experiments in order to assess the potential of the flowmeter for use in actual tissues.

By courtesy of the Medical School of the University of Capetown, a constant head rat's liver perfusion apparatus was set up. The liver was cannulated in such a way that it was fed with 'Plasmalyte B', through the portal vein and drained through the hepatic vein. The whole apparatus was housed in a constant temperature enclosure with a temperature monitor. Unfortunately, the apparatus was designed specifically for constant head perfusions and it was not easily possible to change the flow rate. In addition, the

/...perfusion

perfusion life of the liver under these conditions is of the order of two hours and the flowrate decreases regularly throughout this period. It was, therefore, not possible to obtain balanced conditions over a period greater than 1 to 2 minutes and so, since it was difficult and time-consuming to change the flowrate within this period and regain a balance point, it was impossible to obtain a valid set of readings.

1.7. IN VIVO USE.

Again by courtesy of the Medical School of the University of Capetown the flowmeter was able to be tested during a pig kidney transplant. The probe was placed in the medulla of the kidney and balanced. The balance point showed a random drift of approximately 20% of the F.S.D. on the lowest sensitivity. The renal artery was then occluded and the flowmeter showed a sharp decrease in flow. The occlusion was then removed and the flowmeter registered an increase with overshoot to a new average balance point higher than before. The probe was then placed in the renal artery and, although the balance point was again unstable (less- approximately 10% of F.S.D. because of the lower sensitivity - medium) the pulsatile nature of the flow was quite evident.

In both the above experiments it was noted that the partial reason for the high random drift at balance was due to movement of the probe. It should have been secured in position.

/...Section 5.A.

Section 5

SECTION 5.A.

CONCLUSIONS. It was concluded from the work done that:-

1) The method is valid for laminar flow of a homogeneous medium past the thermistor and that the relationship for this type of flow is of the form

$$VI = K\,\Delta T\,(A + B\sqrt{V})$$

The value of the constants as developed in the theory hold true and depend on the physical properties of the flowing medium.

2) The validity of the method does not necessarily hold for use in an heterogeneous medium such as a constricted sponge or vascular tissue because conditions in these media, on a macroscopic scale are highly variable. Tissue damage due to insertion of the probe or movement of the probe completely alters the expected properties of the flow in tissues and renders variable results. Repeatable accuracy in the sponge experiments was found to be ± 25%.

3) The method could prove to be very valuable for the analysis of pulsatile flow - especially if the instrument were a.c. decoupled and the probe thermal inertia reduced. Repeatable accuracy in tubular flow was found to be ± 10%.

4) Probes could be improved considerably by:-

   a) Reduction of physical size.
   b) Reduction of thermal inertia.
   c) Increasing the heat dissipation rate to the fluid.

/...Instruments

d) Instruments should be simplified by limiting the number of controls. This could be achieved by use of a self-balancing servomechanism.

e) Further research is necessary and could usefully be channelled along the lines laid out in Addendum 2. Further Research.

f) The method shows great promise for industrial applications.

/... SECTION 5.B.

SECTION 5.B.

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SECTION 5.C.

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ADDENDUM 1.

A THERMISTOR THERMOMETER WITH HYPODERMIC PROBE.

1.0 INTRODUCTION. Soon after the thermistor hypodermic probes were built, there was an enquiry as to whether a sensitive thermometer could be built around the probe. The particular application was in the removal of stones from the kidney. In this operation the kidney is cooled to 18°C and is held at this temperature during the course of the operation. The difficulty experienced is to determine the internal temperature of the kidney and to maintain this constant.

The basic requirements of the instrument were as follows:-

a) It was to be calibrated in degrees C.

b) It was to cover the range 15°C to 37°C with an accuracy of ± 1%.

c) It was to be battery operated and hence tolerant of supply voltage.

d) It was to be simple and easy to use.

The application was found interesting and the following instrument developed in an attempt to satisfy the requirements.

1.1. CIRCUIT. Consider Figure A1, the circuit of the thermometer. In this drawing the supply is shown centre-tapped and the amplifiers are loaded to the centre-tap.

/... This

This is not necessary and in the actual circuits built, the supply and amplifier outputs were left floating, the amplifiers being loaded only by the metering circuit.

The battery, a 22.5 volt hearing aid type, was connected in series with "Push-to-Read" normally open - push button S1. This ensures that the battery cannot inadvertently be left connected and hence discharge during storage. The supply is connected through a series resistor VR4, 50kΩ, and through a "Read/ Battery Check", Switch S2, to the meter as a supply voltage check. S2 is a self centering switch to the "Read" position.

The zener diode ZD1 is biassed through the resistor chain R1 and R2 and supplies the base current for the complimentary transistors Q1 and Q2. The voltage between points $Q1_e$ and $Q2_e$ is, therefore, substantially constant.

$$V_e\ Q1 - V_e\ Q2 = VZD1 - (0.6X2)$$

This remains true for a very large supply voltage variation. The bridge circuit is placed in parallel with diode D1 which is forward biassed and fed through the chain R3, Q1, R4, R5, Q2 and R6. The dynamic resistance of D1 is low compared with the resistance of the bridge hence the voltage across the bridge is constant, - 0.6 volts. V_{be} thermal effects of Q1 and Q2 will tend to be cancelled by D1 and ZD1.

The preset bridge arm VR2 was set for balance at the centre scale of the calibration on the meter and the gain of the differential amplifier - set by R16 and R17, ganged presets - was arranged to give a full scale

/...deflection

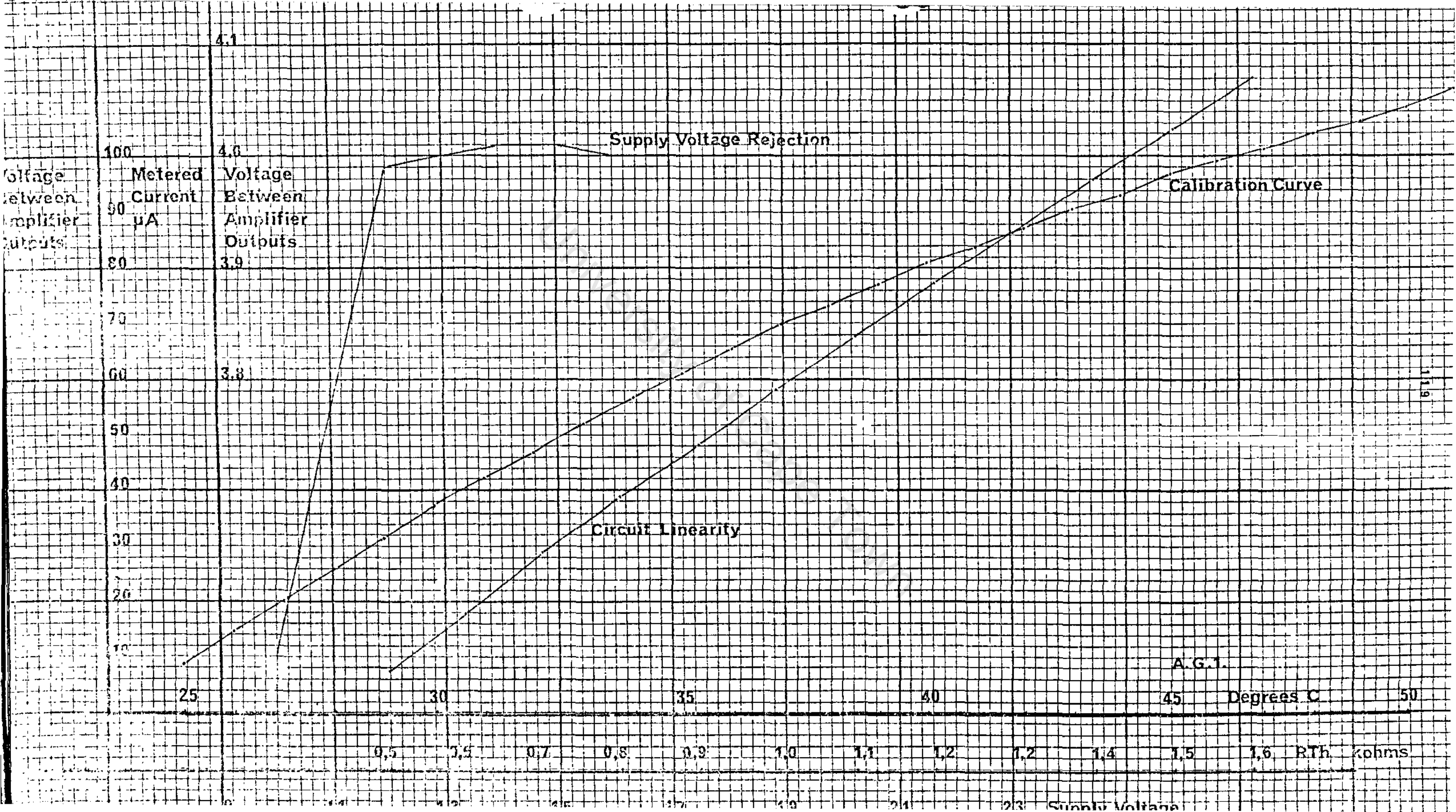
deflection either side of the meter for the region of interest. The circuit was designed so that it could be set to read a minimum of 1°C for full scale deflection at any temperature between 15°C and 50°C.

The current drain of the whole circuit is 2mA maximum i.e. 1.5mA for the operational amplifiers which again consisted of two amplifiers on the same substrate and 0.5mA for the excitation circuit. Graph A.G.1 shows the calibration of the instrument and was taken as an average over three readings. The calibration was performed in a calorimeter twice with cooling water and once with heated water. The standard was a mercury thermometer. Graph A.G.1 also shows the supply voltage rejection - the worst case of the whole design range, and the circuit linearity.

Two instruments were built - the first to operate from 20°C to 50°C and the other from 32°C to 42°C. They were found to be equally successful.

<u>1.3 RESULTS.</u> Several of these instruments have been built by the Dept. of Bio-Engineering of the University of Capetown and are proving to be useful. The original is in the possession of the Dept. of Surgery in the Medical School of the University of Capetown. The latter has been used so far with great success during a number of kidney stone operations. The probes are gas sterilised.

Picture P1 is a photograph of the original instrument.


Supply Voltage Rejection
Calibration Curve
Circuit Linearity
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Voltage Between Amplifier Outputs
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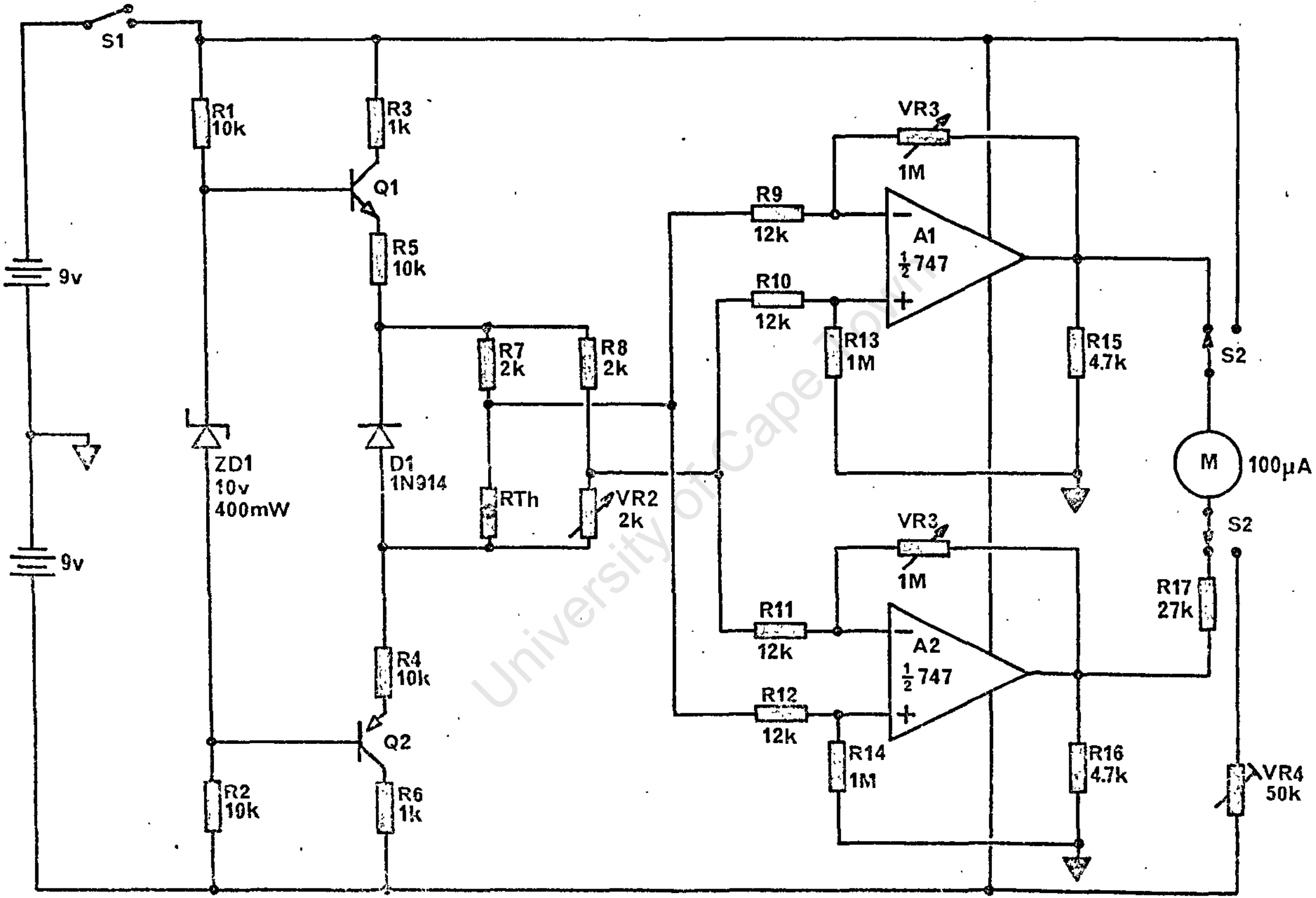
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R8
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RTh
VR2
2k
R4
10k
Q2
R2
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VR3
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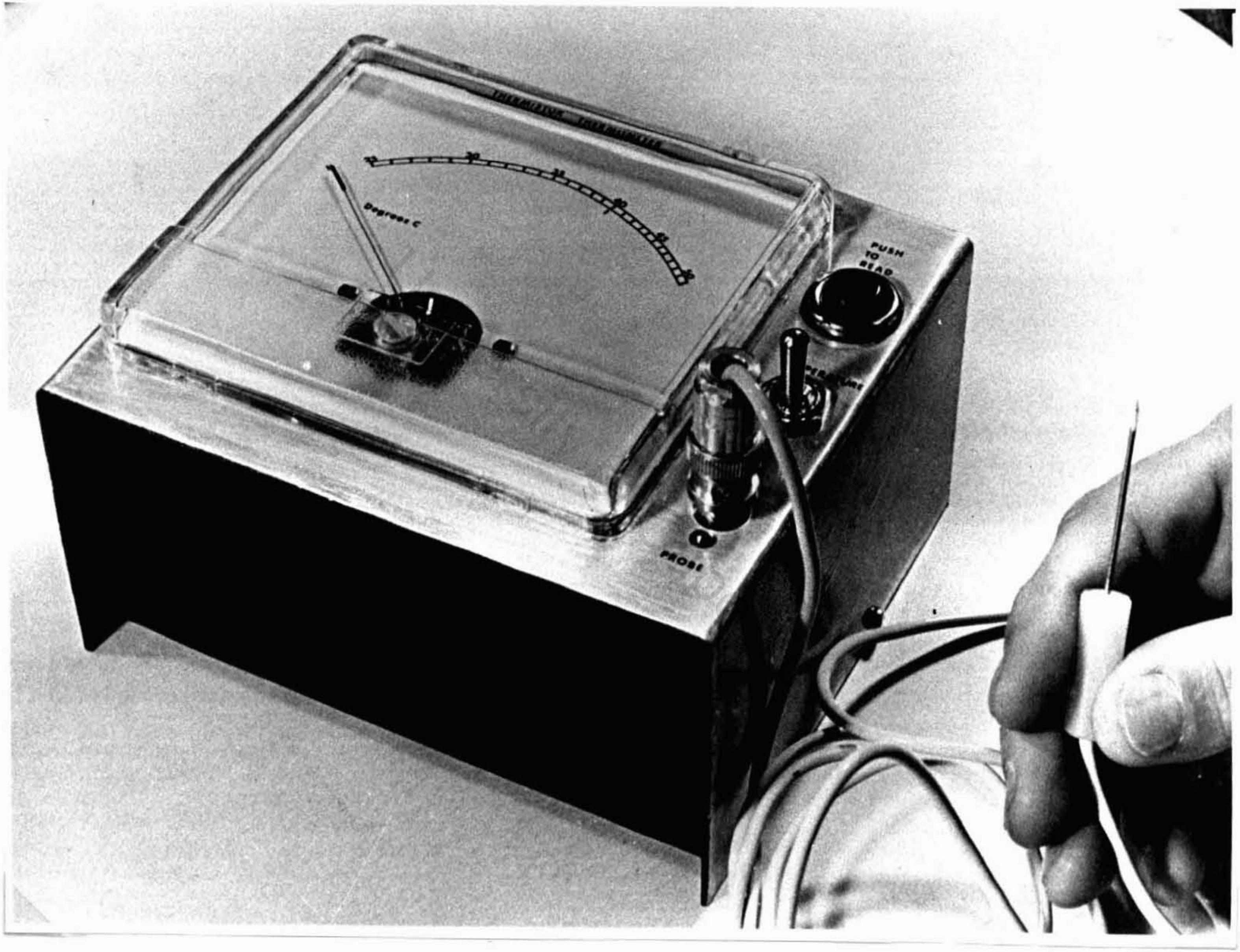
A.1.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PUSH
TO
READ
PROBE

ADDENDUM 2.

FURTHER RESEARCH.

1.0. BRIDGE NETWORK. The first simple mathematical model showed that:-

$$P = f\ (v;\ \Delta T)$$

If ΔT is held constant i.e.

$$RTh_{\text{flow sensor}} - RTh_{\text{temperature sensor}} = k$$

Then if the ambient temperature changes by, say, δT small in comparison with ΔT then:-

$$(T + \delta T) - (T + \Delta T + \delta T) = \Delta T$$

and the temperature sensor changes its value resistance to $RTh_{\text{temperature sensor}} - \delta RTh_{\text{temperature sensor}}$ and the flow sensor changes its value of resistance to RTh flow sensor – δ^1 RTh flow sensor.

Provided that δT is small

$$\delta RTh_{\text{temperature sensor}} / \delta T = \delta RTh_{\text{flow sensor}} / \delta T$$

and α is a function of T but is substantially constant for δT small.

Therefore if we measure the power VI and hold the resistance $^V/_I$ constant we can improve the temperature compensation. There would, however, be little value in going to this end to resolve a detail which represents no great practical problem since temperature compensation, as shown, is not of prime importance.

A slightly different approach might well prove to be more fruitful.

/... 1.1.

<u>1.1. A PULSE TECHNIQUE.</u> Consider applying a fixed power VI to a thermistor for a fixed period of time t, say. The thermistor, at ambient temperature T_o will heat up to a temperature T, say. The temperature difference $T - T_o$ through which the thermistor will heat in that fixed time period will be dependent upon the velocity of the cooling fluid stream. When the power source is removed, the temperature of the thermistor will fall back to ambient temperature T_o. The difference in the resistance of the thermistor at time t_o and time t will be a measure of the flow velocity.

$$RTh_{t_o} - RTh_t = f(v)$$

If the thermal inertia and heat dissipation qualities of the sensor element can be improved, then the cycling rate could be fairly high and preliminary tests have shown that it would be quite possible to track pulsatile flow. The theoretical basis of this method could easily be investigated by stimulating the model with a square wave of power and studying the transient response.

It could prove to be more useful to monitor the rate of decay of the temperature during the unheated period and use this as the measured variable.

$$dRTh/_{dt}\ _{\text{(unheated cycle)}} = f(v)$$

The writer feels that this method contains the key to the solution of the two major problems associated with bridge networks. i.e.

(a) Temperature compensation.
(b) The zero flow zero balance problem.

This method is akin to thermodilution as outlined in the first section.

/... Consider

Consider the circuit shown in 5.C.1.

A. DESIGN OF THE NON SYMMETRICAL MULTIVIBRATOR.

It can be shown that the period of a multivibrator of the form of A1 is given by the expression

$$t = 2CR \log_e \left(1 + 2\frac{R1}{R2}\right)$$

Now $R1/R2$ was chosen to be $\frac{3.3}{5.6} = 0.59$

(R1 = 3.3kΩ)

(R2 = 5.6kΩ)

$\log_e (1 + 2 \times 0.59) = \log_e 2.18 = .78$

Now C1 was fixed at 47mmF

(C1 = 47mmF)

and RF1 and RF2 were chosen so that the thermistor would be heated for a period variable over the range

t = 0 sec. to t = 2 sec.

This gives a value for RF1 = 10kΩ

(RF1 = 10kΩ)

(variable)

and cooled for a period variable over the range

t = 0 sec. to t = 10 sec.

This gives a value of a RF2 = 100kΩ

(RF2 = 100kΩ)

(variable)

Diodes D1, D2 and D4 are IN914's.

(D1, D2 & D4 = IN914)

D4 was included so as to allow only the short period heating cycle to pass to the buffer amplifier A2.

/...B Design

B. DESIGN OF THE BUFFER STAGE AND CONSTANT CURRENT DRIVE FOR THE THERMISTOR.

The network R3, R4 & R5 is an attenuating network to prevent the input of amplifier A2 from being overloaded.

(R3 = 15kΩ)
(R4 = 2.2kΩ)
(R5 = 1.0kΩ)

RF3 was chosen so that during the heating period A2 would go into positive saturation i.e. + 15 V

(RF3 = 10kΩ)

C2 was included to prevent the amplifier A2 from breaking out into sporadic oscillation. This was probably due to component layout.

The constant current network was designed in exactly the same way as the previous two but, in order to retain flexibility, the emitter resistor of Q1 was made variable hence the constant supply current could be varied at any stage.

(2D1 = 6.8 V)
(R7 = 56kΩ)
(R6 = 5kΩ)
(variable)

C. PEAK DETECTOR CIRCUIT A3.

In this network the feedback diode D3

(D3 = IN 914)

simply charges the capacitor C3 to the peak value of the voltage at the positive input of A3.

The rate of charge is RC3 where R is the output resistance of the amplifier A3 and the rate of discharge is

/... RC3

RC3 where R is the input impedance of the amplifier in parallel with the reverse bias resistance of the diode D3 the ratio is

$$\frac{30 \times 100}{300{,}000 \times 100} = 10{,}000$$

C3 was chosen to be 100mmF

(C3 = 100mmF)

so that the rate of discharge would be very much larger than the cooling period i.e. ± 100 secs. R9 is a load for A3.

(R9 = 15kΩ)

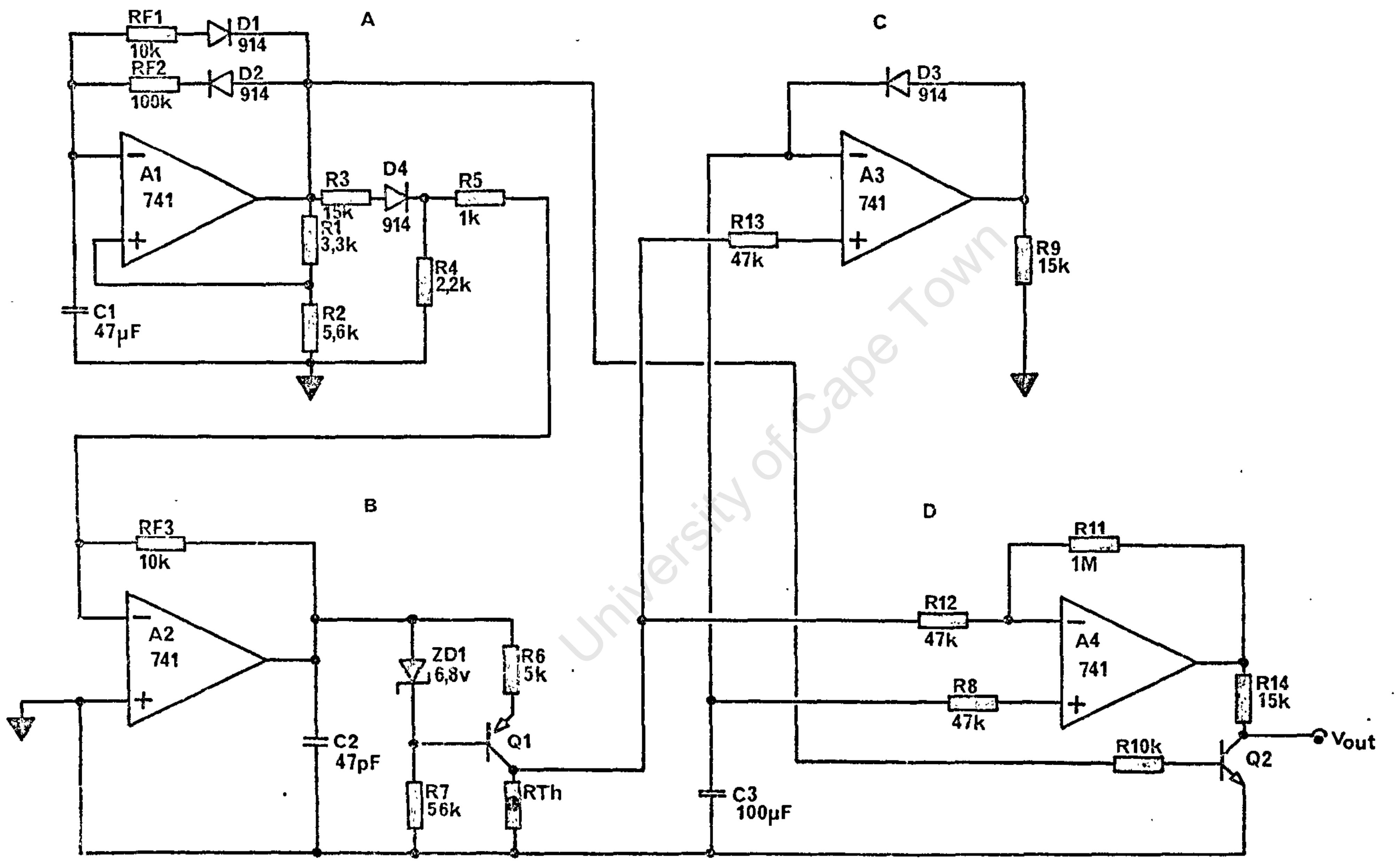
D. DIFFERENTIAL COMPARITOR AND GATING.

The output of the differential amplifier A4 is the difference between the voltage across C3 and the thermistor RTh multiplied by the gain of A4 i.e. R11 ÷ R12.

At the beginning of the heating cycle the output is zero. It then increases as the thermistor heats up until the end of the heating cycle.

When the output of A2 goes to zero after the heating cycle (for the cooling cycle) the output of A4 would go to positive to a value of the voltage on C3 multiplied by the gain of A4. This saturates A4 but during this period Q2 is switched on, shortcircuiting the output of A4, through a load resistor R9 to zero volts.

Results with this circuit - in breadboard layout - were hopeful. The writer believes that this approach warrants further work.


A
RF1
10k
D1
914
RF2
100k
D2
914
A1
741
R3
15k
D4
914
R5
1k
R1
3,3k
R4
2,2k
C1
47µF
R2
5,6k
B
RF3
10k
A2
741
ZD1
6,8v
R6
5k
Q1
C2
47pF
R7
56k
RTh
C
D3
914
A3
741
R13
47k
R9
15k
C3
100µF
D
R11
1M
R12
47k
A4
741
R8
47k
R14
15k
R10k
Q2
Vout

5.C.1.

ADDENDUM 3.

DESIGN OF CIRCUIT 1.

1.1.2. DESIGN OF THE CONSTANT CURRENT NETWORK FOR THE FLOW SENSOR.

(1) Consider the circuit in Figure 3.C.1.A.

Let ZD1 = 2.7 volts (400mW Si Type)

In order that the zener diode operates at a point where dVz/dI is minimal, the current must be chosen to be high.

$$\text{If } R3 = 10\text{k}\Omega \text{ then } I1 = \frac{27.3}{10\text{X}10^{3}} = 2.73\text{X}10^{-3} \text{ amp.}$$

The base current of Q2 will be negligible in comparison with this current.

(2) At 7.5mA the thermistor RTh1 has a resistance of ± 800Ω and heats up to a temperature of 50°C in pure water at 37°C.

So since V_e Q2 = 2.7 – 0.7 = 2.0 volts

$$R5 = \frac{2}{7.5\text{X}10^{-3}} \doteqdot 270\Omega$$

This resistance was made variable so that the current through the flow sensor could be set at will to any value from 0.5mA to 15.0mA.

The voltage drop across R5 is 2.0 volts and across RTh1 is $7.5 \text{ X } 10^{-3} \text{ X } 800 = 6.0$ volts.

(3) We require the voltage at V1 ≑ 0 volts

hence: V_{ce} Q2 + V_e Q2 = 15.0 volts

so V_{ce} Q2 = 15.0 – 2.0 = 13.0 volts

/...but

but V_b Q1 = V_e Q1 + 0.70 volts

and V_e Q1 = V(R5) + V_{ce} Q2 + V (RTh1)

= 2.0 + 13.0 + 6.0 = 21.0 volts

hence V_b Q1 = 21.7 volts

(4) Now in order that V1 can be nulled for RTh1 between say 500 and 2000 Ω, the bias network for Q1 was chosen as follows:

R1 = 8.2kΩ RV1 = 50kΩ R2 = 27kΩ

(5) So that Q1 does not saturate R4 was made equal to 470Ω.

1.1.3. DESIGN OF THE CONSTANT CURRENT NETWORK FOR THE TEMPERATURE SENSOR.

(1) Consider the circuit in Figure 3.C.1.B.

Let ZD2 = 2.7 volts (400 mW Si Type).

In order that the zener diode operates at a point where $d^{V}z/dI$ is minimal, the current must be chosen to be high.

If R8 = 10kΩ then 12 = $\dfrac{27.3}{10\text{X}10^{3}}$ = $2.73\text{X}10^{-3}$ amp.

Again the base current will be negligible in comparison with this current.

(2) At 0.6 mA the thermistor RTh2 has a resistance of ± 1.2kΩ when immersed in pure water at 37°C

/... so

So since V_e Q4 = 2.7 – 0.7 = 2.0 volts

$$R10 = \frac{2}{0.6 \times 10^{-3}} = 3.3\text{k}\Omega$$

Since at this value of current, the thermistor will be heated by its surrounds - even with an ambient temperature as low as 20°C, - it was not necessary to make R10 variable.

The voltage drop across R10 is 2.0 volts and across RTh2 is $0.6 \times 10^{-3} \times 1.2\text{k} = 0.72$ volts.

(3) We require the voltage at V1 $\doteqdot$ 0 volts

hence V_{ce} Q4 + V_e Q4 = 15.0 volts

so V_{ce} Q4 = 15.0 – 2.0 = 13.0 volts

but V_b Q3 = V_e Q3 + 0.7 volts

and V_e Q3 = V(R10) + V_{ce} Q4 + V(RTh2)

= 2.0 + 13.0 + 0.7 = 15.7 volts

hence V_b Q3 = 15.7 + 0.7 = 16.4 volts

(4) Now in order that V2 can be nulled for RTh2 between say 1000 and 1500Ω the bias network for Q3 was chosen as follows:

R6 = 22kΩ RV2 = 10kΩ R7 = 27kΩ

(5) So that Q3 does not saturate R9 was made equal to 4.7kΩ.

1.1.4 DESIGN OF THE COMPARITOR AND AMPLIFIER.

(1) For a thermistor, as before,

$$R = Ae^{B/T}$$

/... and

and for the thermistors used in this work the equation $^{dR}/_{dT}$ may be approximated to a linear function over a small range of temperatures - typically for linear tracking of thermistors within 10%, the order of the range T is 1°C.

From the introduction

$$dV1_{/dT} = \alpha \; ^{dV2}/dT$$

where α is constant for $T - 1°C < T < T + 1°C$ say

(2) Amplifier A1

$RF1_{/R11} = 1$ (RF1 = 12kΩ R11 = 12kΩ)

$R12 = RF1XR11/RF1 + R11 \doteqdot 5.6\Omega$

(3) Amplifier A2

$RVF1_{/13} = \alpha = 8.25$

For flexibility let $0 < \alpha < 10$

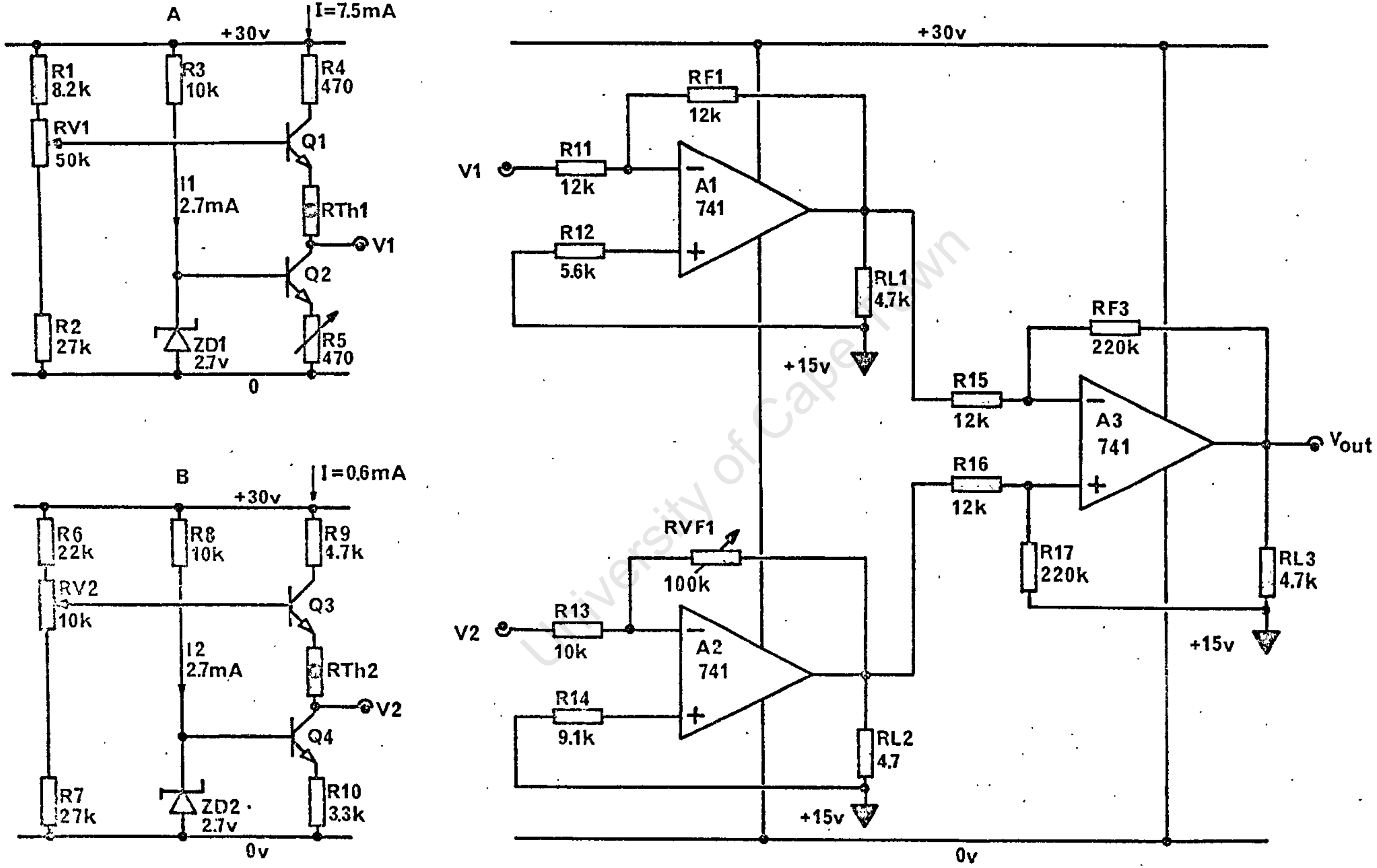
hence (RVF1 = 100kΩ R13 = 10kΩ)

R14 = RVF nominalxR13/RVF1 nominal + R13 $\doteqdot$ 8.2kΩ

(4) The load of all three amplifiers was chosen to be 4.7kΩ and the comparitor amplifier was designed with a gain of 18

RF3 = 220kΩ
R15 = 12kΩ
R16 = 12kΩ
R17 = 220kΩ

All resistors were high stability 1% metal oxide type. External metering and power supplies were used.


A
+30v
I=7.5mA
R1
8.2k
R3
10k
R4
470
RV1
50k
Q1
I1
2.7mA
RTh1
V1
Q2
R2
27k
ZD1
2.7v
R5
470
0
B
+30v
I=0.6mA
R6
22k
R8
10k
R9
4.7k
RV2
10k
Q3
I2
2.7mA
RTh2
V2
Q4
R7
27k
ZD2
2.7v
R10
3.3k
0v
+30v
RF1
12k
V1
R11
12k
A1
741
R12
5.6k
RL1
4.7k
+15v
RF3
220k
R15
12k
A3
741
Vout
R16
12k
RVF1
100k
R17
220k
RL3
4.7k
V2
R13
10k
A2
741
+15v
R14
9.1k
RL2
4.7
+15v
0v

3.C.1.

ADDENDUM 4.

DESIGN OF CIRCUIT 2.

1.2.2. DESIGN OF THE VOLTAGE SUPPLIES.

V_s V_e $V_{in}1$ AND $V_{in}2$.

(1) Refer to the circuit in Figure 3.C.2. A convectional full wave rectified and smoothed power supply was used. Two 15 volt 1 Watt zener diodes in series provided the sources necessary to power the operational amplifiers. i.e. Zero V_s and V_e. This form of supply while not ideal, was found to be adequate. The supply voltage rejection of the amplifiers is 150µV/V of the supply.

(2) The input voltage $V_{in}1$ of amplifiers A2 and A3 was derived from zener diode Z2 which feeds a fixed current into amplifier A1. RVF1 varies the output voltage $V_{in}1$ from 0 to 10 volts without altering the input current and hence the zener diode bias.

(3) The input voltage $V_{in}2$ of amplifiers A4 and A5 was derived from the resistor chain R4 and RV1 in parallel with the zener diode ZD2. It can clearly be seen that the shunting effect of the input current to A4 and A5 will not substantially affect the bias of zener diode ZD2. In any case RV1 is preset for

/...any

any fixed position of RV1 and hence the zener current is constant. Should the supply voltage vary, however, it is clear from the design philosophy that the effect can only be a common mode voltage at points $V1_a$ and $V1_b$ and $V2_a$ and $V2_b$.

1.2.3. DESIGN OF THE CONSTANT CURRENT BRIDGE NETWORK FOR THE FLOW SENSOR.

(1) For this practical circuit it was necessary to reduce the upper temperature limit for the flow sensor to 42°C since 5°C above 37°C was regarded in the theory to be the maximum permissible temperature difference.

Now since

$$R = Ae^{B/T}$$

$$RTh1 \doteqdot 1k\Omega \text{ minimum for the flow sensor}$$

Consider amplifier A3. When $V_{in}1$ is at its maximum value i.e. 10 volts, $V_{out}1$ is $^{1k}/47k \times 10$ volts = 0.212 volts. This voltage appears directly across R15 hence the current through RTh1 is 0.212/47 = 4.5 mA. (i_f in this case is extremely small). Amplifier A2 was designed in exactly the same way and so when RV2 and RV3 are in their centre positions $V1_a \doteqdot V1_b$. The actual voltage at $V1_a \doteqdot V1_b$ is $0.212 + 4.7 \times 10^{-3} \times 1 \times 10^{3} = 5$ volts, which under no circumstances exceeds the output capabilities of the amplifiers.

Amplifiers A2 and A3 are on the same substrate. Output V_{in} was provided for metering the current through RTh1.

(2) From the introduction, it is clear that

$$^{dR}/_{dT} = -Re^{B/T^2}$$

/... and

and for the new temperature of 42°C this works out to $\pm$ 31.0Ω/^{o}K maximum.

1.2.4. DESIGN OF THE CONSTANT CURRENT BRIDGE FOR THE TEMPERATURE SENSOR.

(1) Consider amplifier A5. When $V_{in}2$ is at its maximum value, i.e. 3.3 volts, $V_{out}2$ is $^{1k}/_{47k}$ X3.3 volts = 0.070 volts.

This voltage appears directly across R17, hence the current through RTh2 is 0.070/47 = 1.5 mA (i_f in this case is extremely small). Amplifier A4 was designed in exactly the same way so, when RV4 and RV5 are in their centre positions $V2_a \doteqdot V2_b$. The actual voltage at $V2_a \doteqdot V2_b$ is 0.070 + 1.5×10^{-3} X $1.5 \times 10^{3} \doteqdot$ 2.3 volts which, under no circumstances exceeds the output capabilities of the amplifiers.

Amplifiers A4 and A5 are on the same substrate.

(2) There is no change in the nominal value of RTh2 hence $^{dR}/dT = 35.2\Omega\ 1^{o}K$ for small excursions from 37°C.

1.2.5. THE DIFFERENTIAL AMPLIFIERS AND METERING.

(See Figure 3.C.3.)

(1) From above

$$\frac{dV}{dT} = I \frac{dR}{dT}$$

So for the flow sensor, the maximum value is $4.5 \times 10^{-3} \times 31$ = 140 mV/^{o}K and for the temperature sensor it is $1.5 \times 10^{-3} \times 35.2$ = 53 mV/^{o}K

The amplifier gain difference must be a factor of 3

/...in

in order to achieve reasonable temperature tracking.

(2) The design of the two differential amplifiers, also sharing the same substrate, is straight forward - $\alpha_1 / \alpha_2 = {}^1/_3$ and metering was provided between the two outputs - again eliminating common mode errors.

(3) Switch S2 has three positions:

1) Current monitor ($^V m$)
2) Flow bridge balance
3) Temperature bridge balance and Read.

Series resistors were provided for scale adjustment.

(4) Switch S3 has three positions which simply alter the sensitivity of the bridge. On the lowest sensitivity setting the bridge was designed to show F.S.D. for a 5°C change in RTh1.

(5) The centre-zero meter was protected by two Germanium diodes and because of their poor characteristics a protection on/off switch was provided - Switch S4. Amplifier A8 was included to give a visual indication of imbalance in the form of two L.E.D.s, but was not strictly necessary since a centre zero meter was used.

Picture P1 shows this bridge assembled, Picture P2 the printed circuit and Figure 3.C.4. gives the specification of the amplifiers used.

The operating procedure is as follows:-

/... a) Connect

a) Connect and site electrodes. (Red - right Black - left.)
b) Check that "Meter Protection" is "ON".
c) Switch "Mains On/Off" switch "ON".
d) "Set Current" fully clockwise.
e) Set "Function" switch to "Balance Flow".
f) Balance bridge by means of "Flow Sensor Balance" - "Coarse" and "Fine".
g) Select Sensitivity.
   Venous or arterial flow - "Coarse" and "Medium".
   Tissue flow - "Fine".
h) Set Function switch to "Measure Flow".
i) Balance bridge by means of "Bridge Balance" - "Coarse" and "Fine".
j) Switch "Meter Protection" OFF.

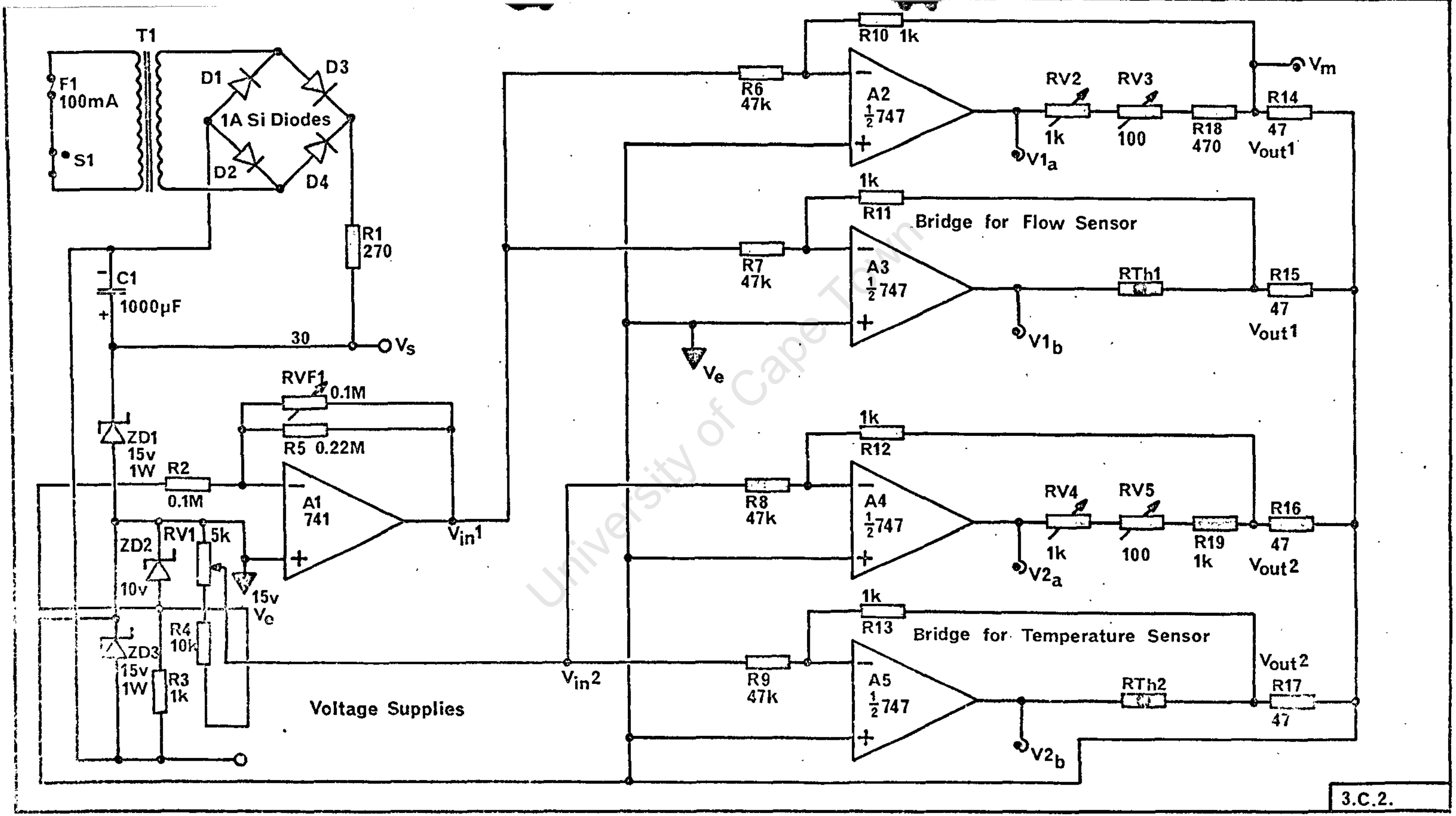
The Bridge is now in operation.

There are some fairly obvious improvements which could be effected in the circuit. These are:-

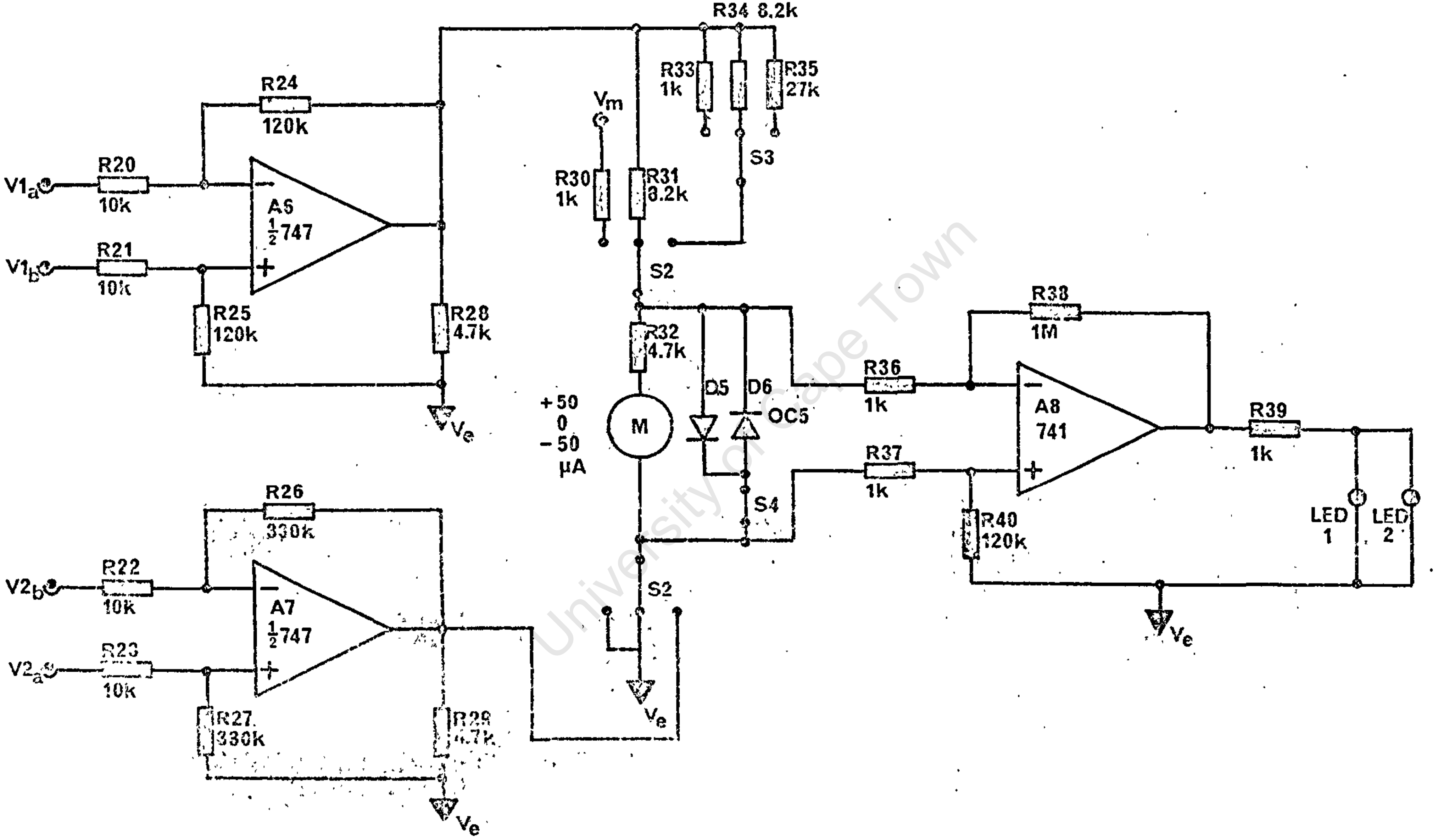
1) A grounded output for external monitoring.
2) $V_{in}2$ linked to V_{in} or $V_{in\ 1}$ via a constant factor circuit to achieve better temperature compensation.

and

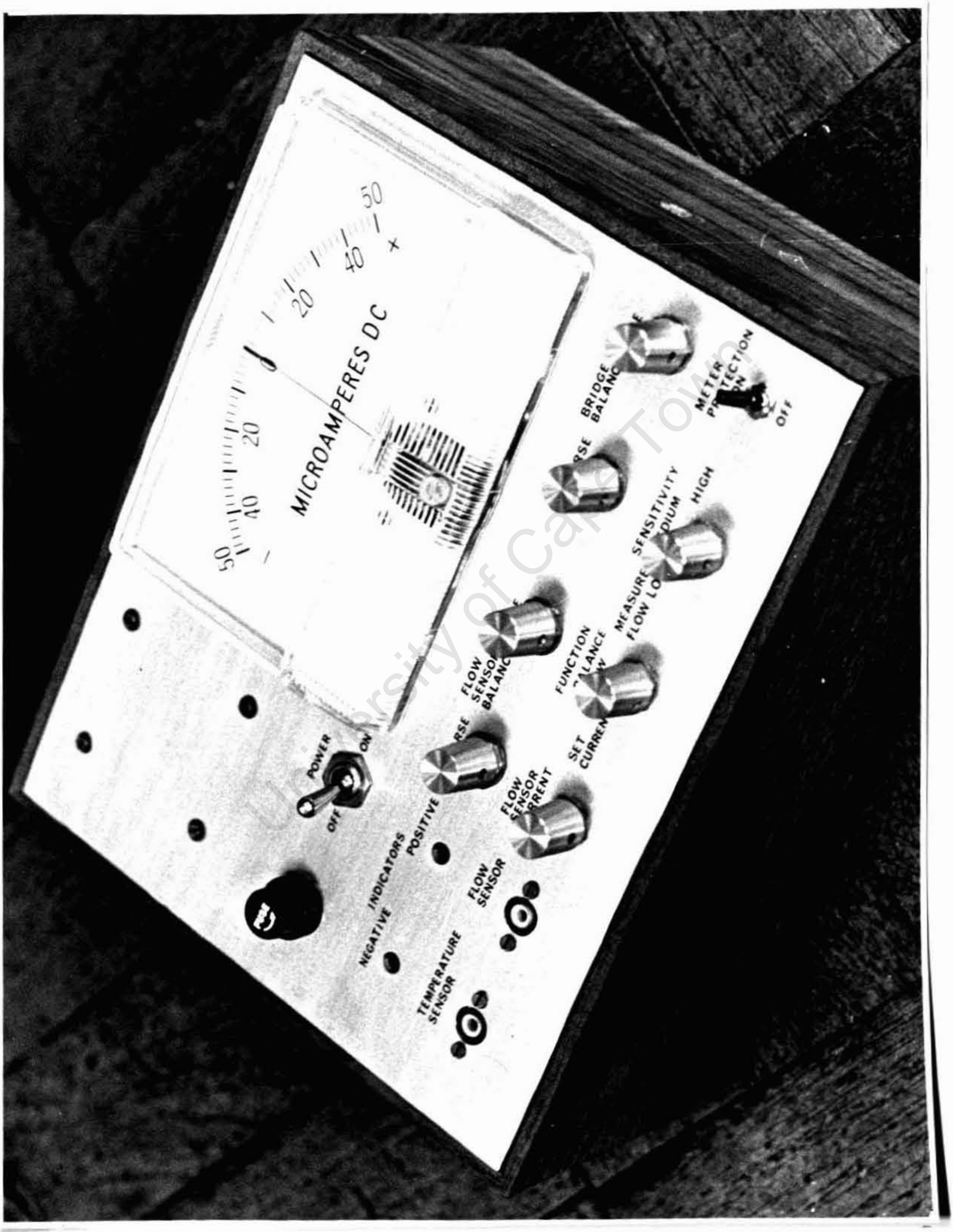
3) Low frequency AC coupling for monitoring pulsatile flow.

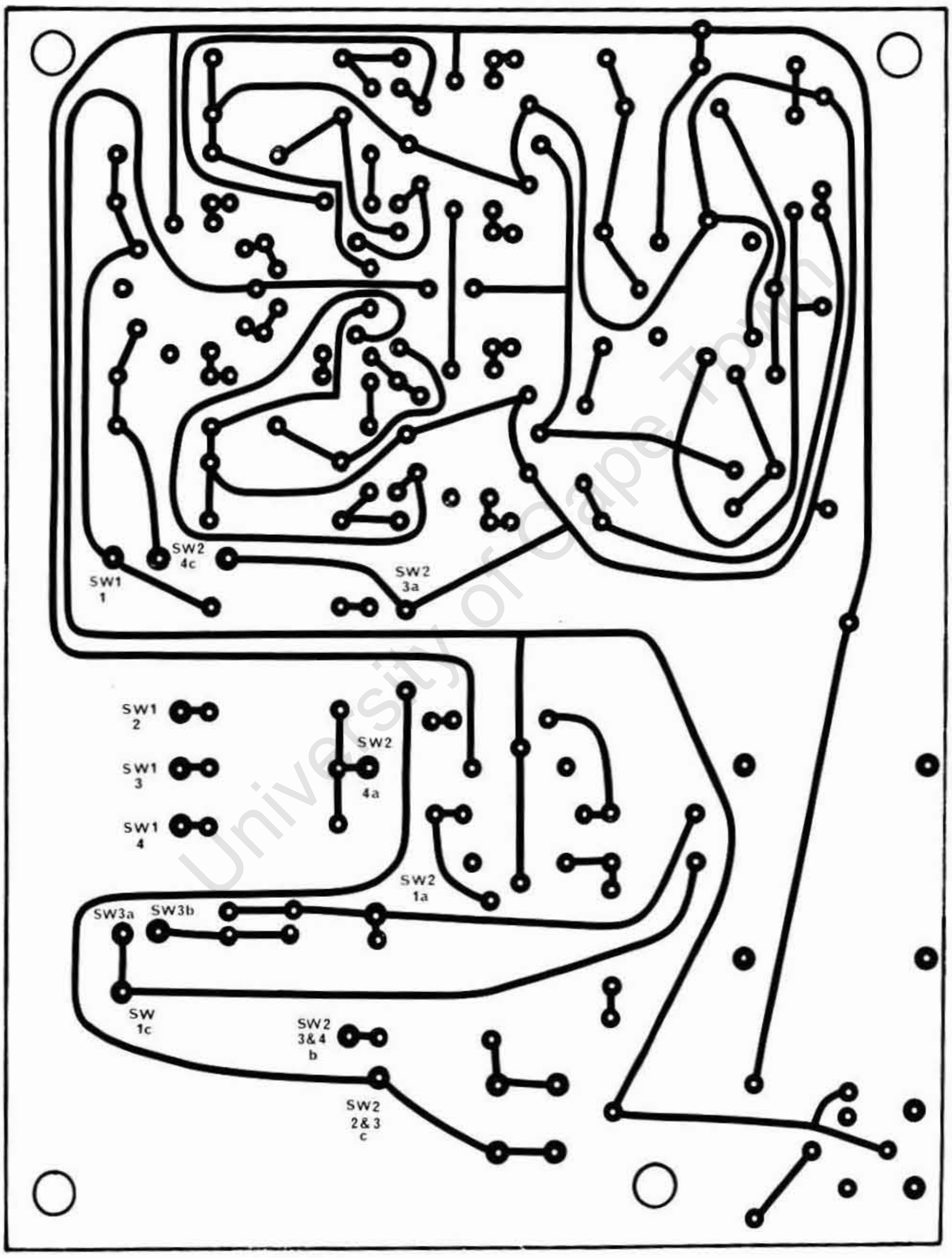

T1
F1
100mA
S1
D1
D2
D3
D4
1A Si Diodes
R1
270
C1
1000µF
30
Vs
ZD1
15v
1W
R2
0.1M
RVF1
0.1M
R5 0.22M
A1
741
Vin1
ZD2
10v
RV1
5k
15v
Ve
R4
10k
ZD3
15v
1W
R3
1k
Voltage Supplies
Vin2
R10 1k
R6
47k
A2
½ 747
RV2
1k
RV3
100
R18
470
R14
47
Vout1
Vm
V1a
1k
R11
Bridge for Flow Sensor
R7
47k
A3
½ 747
RTh1
R15
47
Vout1
V1b
Ve
1k
R12
R8
47k
A4
½ 747
RV4
1k
RV5
100
R19
1k
R16
47
Vout2
V2a
1k
R13
Bridge for Temperature Sensor
R9
47k
A5
½ 747
RTh2
Vout2
R17
47
V2b

3.C.2.


R34 8.2k
R33
1k
R35
27k
R24
120k
Vm
R20
10k
V1a
R21
10k
V1b
A6
½ 747
R25
120k
R28
4.7k
Ve
R30
1k
R31
8.2k
S3
S2
R32
4.7k
D5
D6
OC5
+50
0
−50
µA
M
S4
R36
1k
R37
1k
R38
1M
A8
741
R39
1k
R40
120k
LED 1
LED 2
Ve
R26
330k
R22
10k
V2b
R23
10k
V2a
A7
½ 747
R27
330k
R29
4.7k
S2
Ve
Ve

3.C.3.


MICROAMPERES DC
50
40
20
0
20
40
50
−
+
BRIDGE BALANC
RSE
METER PROTECTION
ON
OFF
SENSITIVITY
DIUM
HIGH
MEASURE
FLOW LO
FUNCTION
BALANCE
FLOW SENSO BALANC
RSE
SET CURREN
FLOW SENSOR CURRENT
POWER
ON
OFF
INDICATORS
POSITIVE
NEGATIVE
FLOW SENSOR
TEMPERATURE SENSOR


SW2
4c
SW1
1
SW2
3a
SW1
2
SW1
3
SW1
4
SW2
4a
SW2
1a
SW3a
SW3b
SW
1c
SW2
3&4
b
SW2
2&3
c

CIRCUIT TYPES SN52741, SN72741
HIGH-PERFORMANCE OPERATIONAL AMPLIFIERS

- Short-Circuit Protection
- Offset-Voltage Null Capability
- Large Common-Mode and Differential Voltage Ranges
- No Frequency Compensation Required
- Low Power Consumption
- No Latch-up
- Same Pin Assignments as SN52709/SN72709

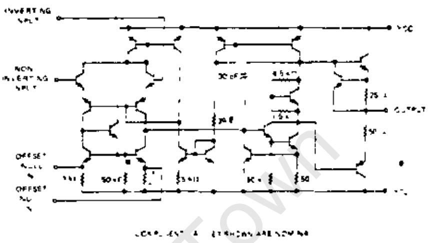
description

The SN52741 and SN72741 are high-performance operational amplifiers, featuring offset-voltage null capability.

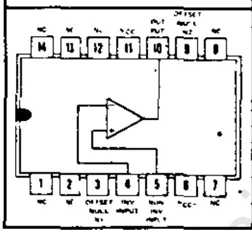
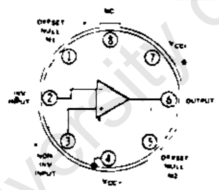
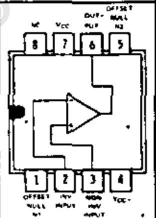
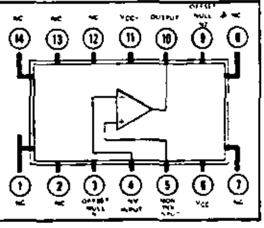
The high common-mode input voltage range and the absence of latch-up make the amplifier ideal for voltage follower applications. The devices are short circuit protected and the internal frequency compensation ensures stability without external components. A low-value potentiometer may be connected between the offset null inputs to null out the offset voltage as shown in Figure 11.

The SN52741 is characterized for operation over the full military temperature range of −55°C to 125°C, the SN72741 is characterized for operation from 0°C to 70°C.

schematic



terminal assignments

J OR N DUAL IN LINE PACKAGE (TOP VIEW)	L PLUG IN PACKAGE (TOP VIEW)	P DUAL IN-LINE PACKAGE (TOP VIEW)	Z FLAT PACKAGE (TOP VIEW)
	 PIN 4 IS IN ELECTRICAL CONTACT WITH THE CASE		

NC—No internal connection

3.C.4.

absolute maximum ratings over operating free-air temperature range (unless otherwise noted)

		SN52741	SN72741	UNIT
Supply voltage V_{CC+} (see Note 1)		22	18	V
Supply voltage V_{CC-} (see Note 1)		−22	18	V
Differential input voltage (see Note 2)		±30	±30	V
Input voltage (either input, see Notes 1 and 3)		±15	±15	V
Voltage between either offset null terminal (N1/N2) and V_{CC-}		±0.5	±0.5	V
Duration of output short-circuit (see Note 4)		unlimited	unlimited	
Continuous total power dissipation at (or below) 55°C free air temperature (see Note 5)		500	500	mW
Operating free-air temperature range		−55 to 125	0 to 70	°C
Storage temperature range		−65 to 150	−65 to 150	°C
Lead temperature 1/16 inch from case for 60 seconds	J, L, or Z Package	300	300	°C
Lead temperature 1/16 inch from case for 10 seconds	N or P Package	260	260	°C

NOTES
1. All voltage values, unless otherwise noted, are with respect to the zero reference level (ground) of the supply voltages where the zero reference level is the midpoint between V_{CC+} and V_{CC-}
2. Differential voltages are at the noninverting input terminal with respect to the inverting input terminal.
3. The magnitude of the input voltage must never exceed the magnitude of the supply voltage or 15 volts, whichever is less.
4. The output may be shorted to ground or either power supply. For the SN52741 only, the unlimited duration of the short circuit applies at (or below) 125°C case temperature or 75°C free air temperature
5. For operation above 55°C free air temperature, refer to Dissipation Derating Curve Figure 12

3

electrical characteristics at specified free-air temperature, V_{CC+} = 15 V, V_{CC-} = −15 V

PARAMETER		TEST CONDITIONS†		SN52741			SN72741			UNIT
				MIN	TYP	MAX	MIN	TYP	MAX	
V_{IO}	Input offset voltage	R_S ≤ 10 kΩ	25°C		1	5		1	6	mV
			Full range			6			7.5	
$\Delta V_{IO(adj)}$	Offset voltage adjust range		25°C		±15			±15		mV
I_{IO}	Input offset current		25°C		20	200		20	200	nA
			Full range			500			300	
I_{IB}	Input bias current		25°C		80	500		80	500	nA
			Full range			1500			800	
V_I	Input voltage range		25°C	±12	±13		±12	±13		V
			Full range	±12			±12			
V_{OPP}	Maximum peak-to-peak output voltage swing	R_L = 10 kΩ	25°C	24	28		24	28		V
		R_L ≥ 10 kΩ	Full range	24			24			
		R_L = 2 kΩ	25°C	20	26		20	26		
		R_L ≥ 2 kΩ	Full range	20			20			
A_{VD}	Large-signal differential voltage amplification	R_L ≥ 2 kΩ, V_O = ±10 V	25°C	50,000	200,000		20,000	200,000		
			Full range	25,000			15,000			
r_i	Input resistance		25°C	0.3	2		0.3	2		MΩ
r_o	Output resistance	V_O = 0 V, See Note 5	25°C		75			75		Ω
C_i	Input capacitance		25°C		1.4			1.4		pF
CMRR	Common-mode rejection ratio	R_S ≤ 10 kΩ	25°C	70	90		70	90		dB
			Full range	70			70			
$\Delta V_{IO}/\Delta V_{CC}$	Power supply sensitivity	R_S ≤ 10 kΩ	25°C		30	150		30	150	μV/V
			Full range			150			150	
I_{OS}	Short-circuit output current		25°C		±25	±40		±25	±40	mA
I_{CC}	Supply current	No load, No signal	25°C		1.7	2.8		1.7	2.8	mA
			Full range			3.3			3.3	
P_D	Total power dissipation	No load No signal	25°C		50	85		50	85	mW
			Full range			100			100	

†All characteristics are specified under open loop operation. Full range for SN52741 is −55°C to 125°C and for SN72741 is 0°C to 70°C.

NOTE 5 This typical value applies only at frequencies above a few hundred hertz because of the effects of drift and thermal feedback operating characteristics, $V_{CC+} = 15$ V, $V_{CC-} = -15$ V, $T_A = 25^\circ$C

PARAMETER		TEST CONDITIONS	SN52741			SN72741			UNIT
			MIN	TYP	MAX	MIN	TYP	MAX	
t_r	Rise time	$V_I = 20$ mV, $R_L = 2$ kΩ,		0.3			0.3		μs
	Overshoot	$C_L = 100$ pF, See Figure 1		5%			5%		
SR	Slew rate at unity gain	$V_I = 10$ V $R_L = 2$ kΩ, $C_L = 100$ pF, See Figure 1		0.5			0.5		V/μs

DEFINITION OF TERMS

Input Offset Voltage (V_{IO}) The d-c voltage which must be applied between the input terminals to force the quiescent d-c output voltage to zero. The input offset voltage may also be defined for the case where two equal resistances (R_S) are inserted in series with the input leads.

Input Offset Current (I_{IO}) The difference between the currents into the two input terminals with the output at zero volts.

Input Bias Current (I_{IB}) The average of the currents into the two input terminals with the output at zero volts.

Input Voltage Range (V_I) The range of voltage which, if exceeded at either input terminal, will cause the amplifier to cease functioning properly

Maximum Peak-to-Peak Output Voltage Swing (V_{OPP}) The maximum peak-to-peak output voltage which can be obtained without waveform clipping when the quiescent d-c output voltage is zero

Large-Signal Differential Voltage Amplification (A_{VD}) The ratio of the peak to peak output voltage swing to the change in differential input voltage required to drive the output

Input Resistance (r_i) The resistance between the input terminals with either input grounded.

Output Resistance (r_o) The resistance between the output terminal and ground

Input Capacitance (C_i) The capacitance between the input terminals with either input grounded.

Common-Mode Rejection Ratio (CMRR) The ratio of differential voltage amplification to common-mode voltage amplification. This is measured by determining the ratio of a change in input common-mode voltage to the resulting change in input offset voltage

Power Supply Sensitivity ($\Delta V_{IO}/\Delta V_{CC}$) The ratio of the change in input offset voltage to the change in supply voltages producing it. For these devices, both supply voltages are varied symmetrically

Short-Circuit Output Current (I_{OS}) The maximum output current available from the amplifier with the output shorted to ground or to either supply.

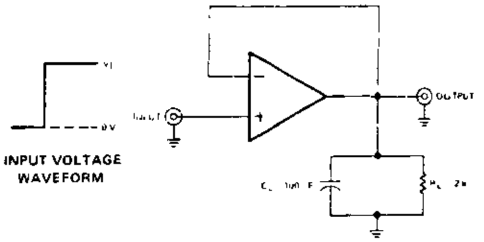
Total Power Dissipation (P_D) The total d-c power supplied to the device less any power delivered from the device to a load. At no load: $P_D = V_{CC+} \cdot I_{CC+} + V_{CC-} \cdot I_{CC-}$.

Rise Time (t_r) The time required for an output voltage step to change from 10% to 90% of its final value.

Overshoot The quotient of (1) the largest deviation of the output signal value from its steady-state value after a step function change of the input signal, and (2) the difference between the output signal values in the steady state before and after the step-function change of the input signal.

Slew Rate (SR) The average time rate of change of the closed-loop amplifier output voltage for a step-signal input. Slew rate is measured between specified output levels (0 and 10 volts for this device) with feedback adjusted for unity gain

PARAMETER MEASUREMENT INFORMATION



INPUT VOLTAGE WAVEFORM

TEST CIRCUIT

FIGURE —RISE TIME, OVERSHOOT, AND SLEW RATE

TYPICAL CHARACTERISTICS

INPUT OFFSET CURRENT
vs
FREE AIR TEMPERATURE

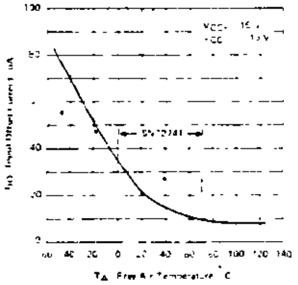


FIGURE 2

INPUT BIAS CURRENT
vs
FREE AIR TEMPERATURE

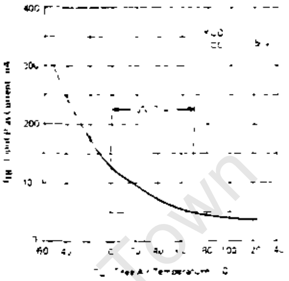


FIGURE 3

MAXIMUM PEAK TO PEAK OUTPUT VOLTAGE
vs
LOAD RESISTANCE

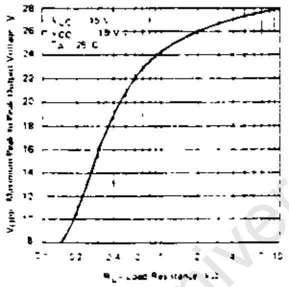


FIGURE 4

MAXIMUM PEAK-TO PEAK OUTPUT VOLTAGE
vs
FREQUENCY

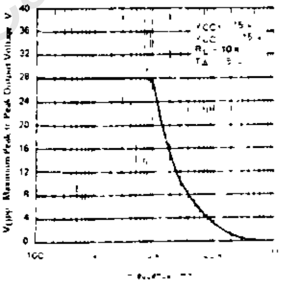


FIGURE 5

TYPICAL CHARACTERISTICS

OPEN LOOP LARGE SIGNAL DIFFERENTIAL VOLTAGE AMPLIFICATION vs SUPPLY VOLTAGE

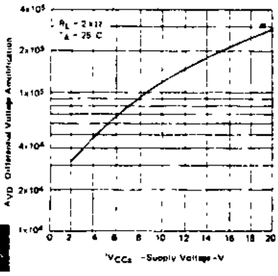


FIGURE 6

OPEN LOOP LARGE-SIGNAL DIFFERENTIAL VOLTAGE AMPLIFICATION vs FREQUENCY

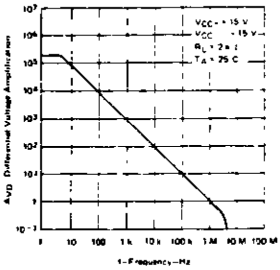


FIGURE 7

COMMON-MODE REJECTION RATIO vs FREQUENCY

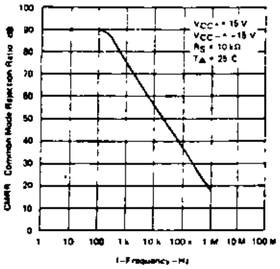


FIGURE 8

OUTPUT VOLTAGE vs ELAPSED TIME

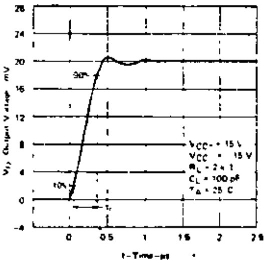


FIGURE 9

VOLTAGE-FOLLOWER LARGE-SIGNAL PULSE RESPONSE

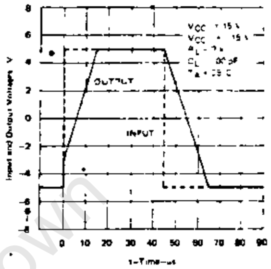


FIGURE 10

TYPICAL APPLICATION DATA

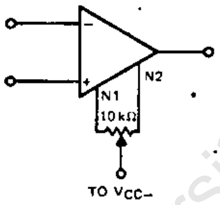


FIGURE 11—INPUT OFFSET VOLTAGE NULL CIRCUIT

THERMAL INFORMATION

DISSIPATION DERATING CURVE

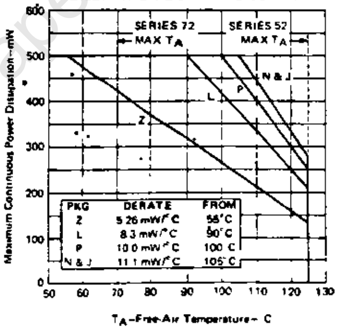


FIGURE 12

CIRCUIT TYPES SN52747, SN72747
DUAL HIGH-PERFORMANCE OPERATIONAL AMPLIFIERS

- No frequency Compensation Required
- Low Power Consumption
- Short-Circuit Protection
- Offset-Voltage Null Capability
- Large Common-Mode and Differential Voltage Ranges
- No Latch-up
- Designed to be Interchangeable with Fairchild μA747 and μA747C

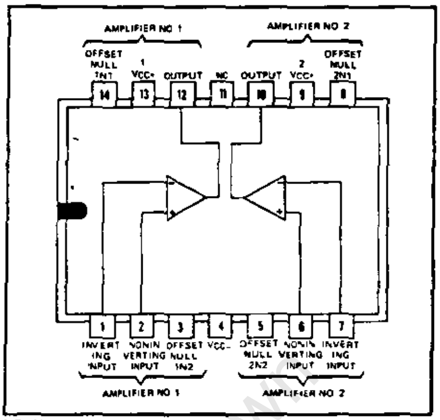
description

The SN52747 and SN72747 are dual high-performance operational amplifiers, featuring offset-voltage null capability. Each half is electrically similar to SN52741/SN72741.

The high common-mode input voltage range and the absence of latch-up make the amplifiers ideal for voltage-follower applications. The devices are short-circuit protected and the internal frequency compensation ensures stability without external components. A low-value potentiometer may be connected between the offset null inputs to null out the offset voltage as shown in Figure 3.

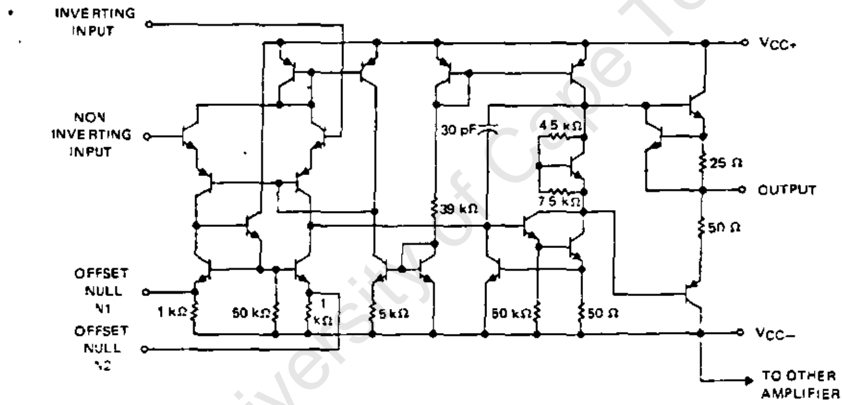
The SN52747 is characterized for operation over the full military temperature range of −55°C to 125°C; the SN72747 is characterized for operation from 0°C to 70°C.

J OR N DUAL-IN-LINE PACKAGE (TOP VIEW)



NC—No internal connection

schematic (each amplifier)



Component values shown are nominal.

CIRCUIT TYPES SN52747, SN72747
DUAL HIGH-PERFORMANCE OPERATIONAL AMPLIFIERS

absolute maximum ratings over operating free-air temperature range (unless otherwise noted)

		SN52747	SN72747	UNIT
Supply voltage V_{CC+} (see Note 1)		22	18	V
Supply voltage V_{CC-} (see Note 1)		−22	−18	V
Differential input voltage (see Note 2)		±30	±30	V
Input voltage (either input, see Notes 1 and 3)		±15	±15	V
Voltage between either offset null terminal (N1/N2) and V_{CC-}		0.5	±0.5	V
Duration of output short circuit (see Note 4)		unlimited	unlimited	
Continuous total dissipation at (or below) 70°C free-air temperature (see Note 5)	Each amplifier	500	500	mW
	Total package	800	800	
Operating free-air temperature range		−55 to 125	0 to 70	°C
Storage temperature range		−65 to 150	−65 to 150	°C
Lead temperature 1/16 inch from case for 60 seconds	J package	300	300	°C
Lead temperature 1/16 inch from case for 10 seconds	N package	260	260	°C

NOTES
1. All voltage values, unless otherwise noted, are with respect to the zero reference level (ground) of the supply voltages where the zero reference level is the midpoint between V_{CC+} and V_{CC-}.
2. Differential voltages are at the noninverting input terminal with respect to the inverting input terminal.
3. The magnitude of the input voltage must never exceed the magnitude of the supply voltage or 15 volts, whichever is less.
4. The output may be shorted to ground or either power supply. For the SN52747 only, the unlimited duration of the short circuit applies at (or below) 125°C case temperature or 75°C free-air temperature.
5. For operation of SN52747 above 70°C free-air temperature, refer to Dissipation Derating Curve, Figure 2.

electrical characteristics at specified free-air temperature, $V_{CC+} = 15$ V, $V_{CC-} = -15$ V

PARAMETER		TEST CONDITIONS†		SN52747			SN72747			UNIT
				MIN	TYP	MAX	MIN	TYP	MAX	
V_{IO}	Input offset voltage	$R_S \leq 10$ kΩ	25°C		1	5		1	6	mV
			Full range			6			7.5	
$\Delta V_{IO(adj)}$	Offset voltage adjust range		25°C		±15			±15		mV
I_{IO}	Input offset current		25°C		20	200		20	200	nA
			Full range			500			300	
I_{IB}	Input bias current		25°C		80	500		80	500	nA
			Full range			1500			800	
V_I	Input voltage range		25°C	±12	±13		±12	±13		V
			Full range	±12			±12			
V_{OPP}	Maximum peak-to-peak output voltage swing	$R_L = 10$ kΩ	25°C	24	28		24	28		V
		$R_L \geq 10$ kΩ	Full range	24			24			
		$R_L = 2$ kΩ	25°C	20	26		20	26		
		$R_L \geq 2$ kΩ	Full range	20			20			
A_{VD}	Large-signal differential voltage amplification	$R_L \geq 2$ kΩ, $V_O = \pm 10$ V	25°C	50,000	200,000		50,000	200,000		
			Full range	25,000			25,000			
r_i	Input resistance		25°C	0.3	2		0.3	2		MΩ
r_o	Output resistance	$V_O = 0$ V, See Note 5	25°C		75			75		Ω
C_i	Input capacitance		25°C		1.4			1.4		pF
CMRR	Common-mode rejection ratio	$R_S \leq 10$ kΩ	25°C	70	90		70	90		dB
			Full range	70			70			
$\Delta V_{IO}/\Delta V_{CC}$	Power supply sensitivity	$R_S \leq 10$ kΩ	25°C		30	150		30	150	μV/V
			Full range			150			150	
I_{OS}	Short-circuit output current		25°C		±25	±40		±25	±40	mA
I_{CC}	Supply current	No load, No signal	25°C		1.7	2.8		1.7	2.8	mA
			Full range			3.3			3.3	
P_D	Power dissipation (each amplifier)	No load, No signal	25°C		50	85		50	85	mW
			Full range			100			100	
V_{o1}/V_{o2}	Channel separation		25°C		120			120		dB

† All characteristics are specified under open-loop operation. Full range for SN52747 is −55°C to 125°C and for SN72747 is 0°C to 70°C.

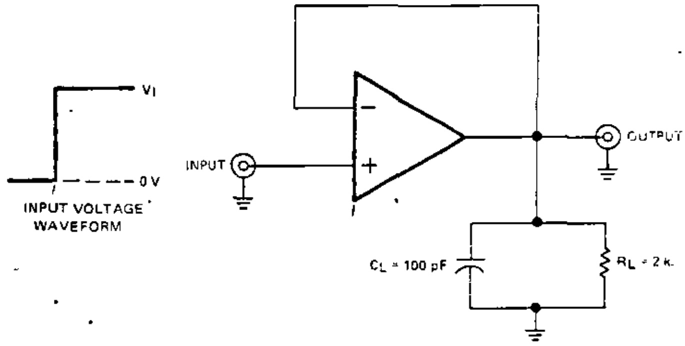
NOTE 5. This typical value applies only at frequencies above a few hundred hertz because of the effects of drift and thermal feedback.

For definitions of terms, mechanical data, and ordering instructions, see the SN52741/SN72741 data sheet dated November 1970.

operating characteristics, $V_{CC+} = 15$ V, $V_{CC-} = -15$ V, $T_A = 25°C$

PARAMETER		TEST CONDITIONS	SN52747			SN72747			UNIT
			MIN	TYP	MAX	MIN	TYP	MAX	
t_r	Rise time	$V_I = 20$ mV, $R_L = 2$ kΩ, $C_L = 100$ pF, See Figure 1		0.3			0.3		μs
	Overshoot			5%			5%		
SR	Slew rate at unity gain	$V_I = 10$ V, $R_L = 2$ kΩ, $C_L = 100$ pF, See Figure 1		0.5			0.5		V/μs

PARAMETER MEASUREMENT INFORMATION



TEST CIRCUIT

FIGURE 1–RISE TIME, OVERSHOOT, AND SLEW RATE

THERMAL INFORMATION

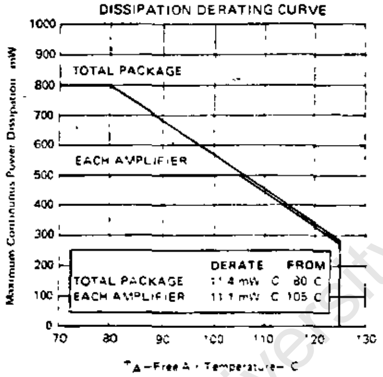


FIGURE 2

TYPICAL APPLICATION DATA

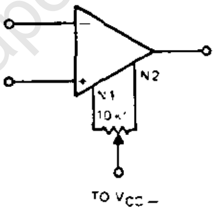


FIGURE 3–INPUT OFFSET VOLTAGE NULL CIRCUIT

TYPICAL CHARACTERISTICS

INPUT OFFSET CURRENT
vs
FREE AIR TEMPERATURE

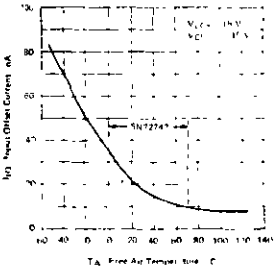


FIGURE 4

INPUT BIAS CURRENT
vs
FREE AIR TEMPERATURE

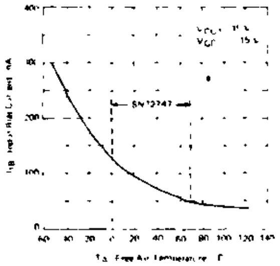


FIGURE 5

MAXIMUM PEAK-TO-PEAK
OUTPUT VOLTAGE
vs
LOAD RESISTANCE

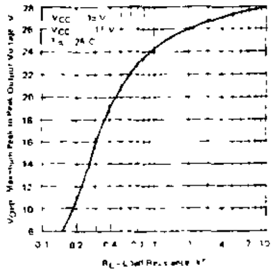


FIGURE 6

MAXIMUM PEAK TO PEAK
OUTPUT VOLTAGE
vs
FREQUENCY

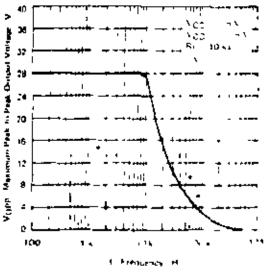


FIGURE 7

OPEN LOOP LARGE SIGNAL
DIFFERENTIAL
VOLTAGE AMPLIFICATION
vs
SUPPLY VOLTAGE

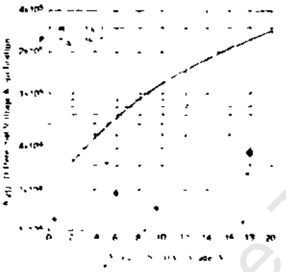


FIGURE 8

OPEN LOOP LARGE SIGNAL
DIFFERENTIAL
VOLTAGE AMPLIFICATION
vs
FREQUENCY

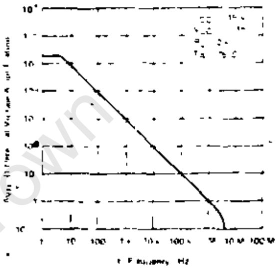


FIGURE 9

COMMON-MODE REJECTION RATIO
vs
FREQUENCY

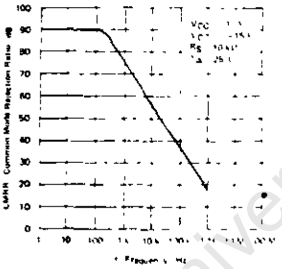


FIGURE 10

OUTPUT VOLTAGE
vs
ELAPSED TIME

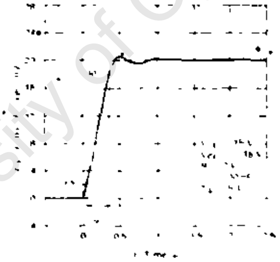


FIGURE 11

VOLTAGE-FOLLOWER
LARGE SIGNAL PULSE RESPONSE

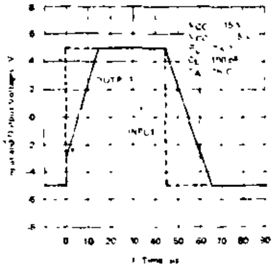


FIGURE 12