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A Self-Heated Thermistor Technique to Measure Effective Thermal Properties From the Tissue Surface

A microcomputer based instrument to measure effective thermal conductivity and diffusivity at the surface of a tissue has been developed. Self-heated spherical thermistors, partially embedded in an insulator, are used to simultaneously heat tissue and measure the resulting temperature rise. The temperature increase of the thermistor for a given applied power is a function of the combined thermal properties of the insulator, the thermistor, and the tissue. Once the probe is calibrated, the instrument accurately measures the thermal properties of tissue. Conductivity measurements are accurate to 2 percent and diffusivity measurements are accurate to 4 percent. A simplified bioheat equation is used which assumes the effective tissue thermal conductivity is a linear function of perfusion. Since tissue blood flow strongly affects heat transfer, the surface thermistor probe is quite sensitive to perfusion.

Introduction

This paper presents a technique to measure the effective thermal properties of tissue using surface thermistor probes. The advantage of the surface probe over previous thermistor techniques is that the measurement can be made without penetrating the tissue. A simplified bioheat equation is used which assumes the effective tissue thermal conductivity is a linear function of perfusion. The finite element numerical method was used to study the measurement errors caused by the presence of a decoupler between the probe and the tissue. A numerical study showed that the baseline tissue temperature drift must be less than 0.03° C/min to ensure reliable thermal conductivity measurements.

The accuracy of the surface probe for measuring thermal conductivity and diffusivity was evaluated using liquid standards. Experiments in a two layer liquid were used to determine the effective measurement depth.

The perfusion measurements were evaluated using isolated and *in vivo* rat liver preparations. The experiments show that the surface thermistor probe is quite sensitive to perfusion. The accuracy of the perfusion measurement could not be determined due to the nonuniform perfusion at the surface of the isolated rat liver. Perfusion can be quantified only when uniform perfusion extends completely to the tissue surface.

Background

Since the eighteenth century, there have been efforts to improve the understanding of the human thermoregulatory system and to develop a means of measuring thermal energy and thermal energy transfer *in vivo* (Jain and Chato 1983). The rate at which thermal energy is dissipated in biological tissue is a function of the tissue's thermal conductivity, thermal diffusivity and blood flow. Chato (1968), Bowman et al. (1977), Jain (1979), Chen et al. (1981), Valvano et al. (1984a), and Walsh (1984) have shown that self-heated thermistors are capable of measuring thermal properties and perfusion.

Knowlege of the thermal properties of tissue is important for both diagnostic and therapeutic medicine. Thermal properties are required to model thermal transport phenomena in tissue. Such models allow better interpretation of heat transfer processes in thermography, organ preservation, hyperthermia, hypothermia and various peripheral vascular diseases.

Tissue perfusion is a measure of the local blood flow through the capillary network of a tissue. While flow in a single vessel is a vector quantity, the convoluted nature of the capillary bed forces blood flow in a macroscopic region to be considered nondirectional. Perfusion is the volume flow rate of blood per unit mass of tissue (mL/100 g-min). Blood flow in arteries and veins is directional and is measured in units of volume of blood per time (mL/min). Blood flow and tissue perfusion are clearly related, but there may be significant perfusion abnormalities even in the presence of normal blood flow (e.g., a myocardial infarction). Tissue perfusion provides a reliable estimate of the viability of an ischemic or reimplanted tissue.

Thermal Model

Chen et al. (1980), Weinbaum and Jiji (1985), and Williams

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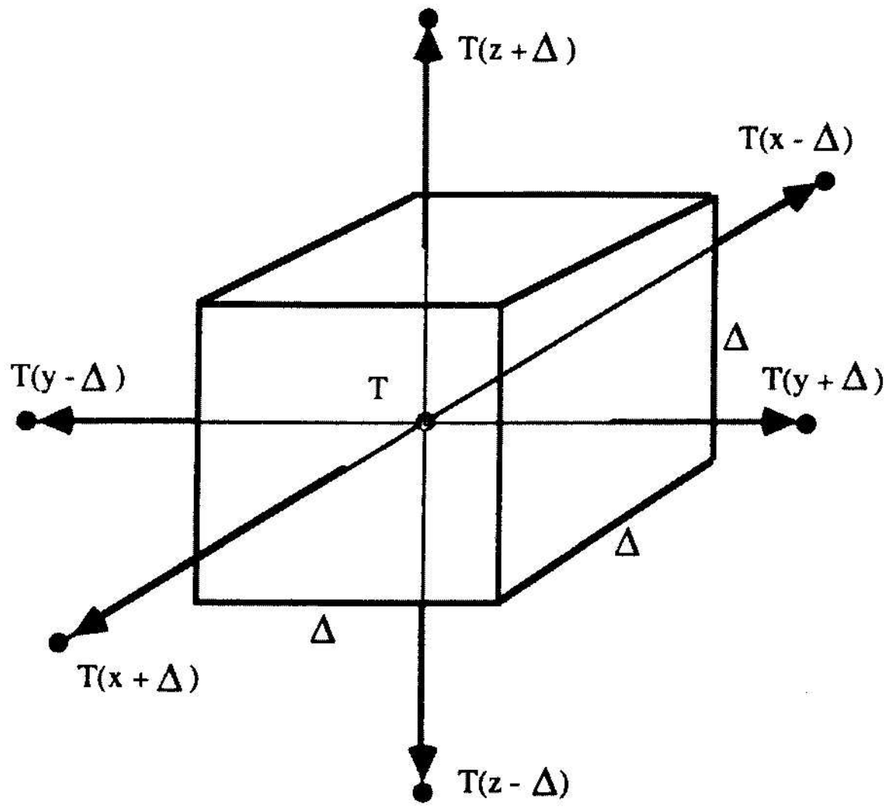

Fig. 1 Tissue control volume and temperatures at adjacent volumes

and Roemer (1986) have shown that Pennes' equation (1948) does not describe heat transfer due to perfusion. Although Weinbaum's model does accurately predict temperature fields in perfused tissue (Song et al. 1987), it is difficult to use because it requires many tissue parameters. A simplified bioheat equation has been developed to model how perfusion affects the self-heated thermistor. These equations are not intended to be a general purpose model to describe heat transfer in all situations.

Figure 1 shows a tissue control volume with equal dimensions, Δ. The invasive thermistor probe is located in the center of the control volume, and Δ^3 is the effective measurement volume of the thermistor. The following conditions are assumed:

1) The temperature within each control volume is uniform;
2) Temperature discontinuities exist at the boundaries between control volumes;
3) Blood leaving a control volume is in equilibrium with that control volume;
4) The volume and density of a control volume remain constant;
5) The fluid is incompressible;
6) There are no thermally significant vessels in the control volume.

Assumptions 4 and 5 dictate that the total blood flow into the control volume will equal the total blood flow out of the control volume. The "Bio-heat Equation" is:

$$\rho_m c_m \partial T/\partial t = K_m \nabla^2 T + q_{\text{perf}} + q_{\text{ext}} + q_{\text{met}} \tag{1}$$

where q_{ext} is the volumetric heat added by the thermistor. One temperature is assigned to each control volume, as shown in Fig. 1, analogous to the finite difference technique. The volumetric heat due to arterial flow, $F_a(z+\Delta)$, into the top of the control volume is:

$$q_a(z+\Delta) = F_a(z+\Delta)\rho_b c_b T(z+\Delta)/\Delta^3 \tag{2}$$

The volumetric heat due to venous blood flow, $F_v(z+\Delta)$, out of the top of the control volume is:

$$q_v(z+\Delta) = -F_v(z+\Delta)\rho_b c_b T/\Delta^3 \tag{3}$$

The net volumetric heat due to perfusion is the sum of the heat gains and losses into and out of the six faces of the control volume.

$$\begin{aligned} q_{\text{perf}} = \rho_b c_b \{ & F_a(x+\Delta)T(x+\Delta) + F_a(x-\Delta)T(x-\Delta) \\ & + F_a(y+\Delta)T(y+\Delta) + F_a(y-\Delta)T(y-\Delta) \\ & + F_a(z+\Delta)T(z+\Delta) + F_a(z-\Delta)T(z-\Delta) \\ & - T[F_v(x+\Delta) + F_v(x-\Delta) + F_v(y+\Delta) \\ & + F_v(y-\Delta) + F_v(z+\Delta) + F_v(z-\Delta)]\}/\Delta^3 \end{aligned} \tag{4}$$

Three additional assumptions allow the heat transfer due to perfusion to be modeled as an effective thermal conductivity:

7) Perfusion is a scalar field;
8) Perfusion is uniform in this control volume and in the adjacent volumes;
9) Perfusion in the control volume is randomly directed.

Since perfusion is uniform and randomly directed, the flow across each surface of the control volume may be taken to be 1/6 of the total flow. Thus:

$$F_a(x+\Delta) = F_v(x+\Delta) = \ldots = F_v(z-\Delta) = w\Delta^3/6\rho_b \tag{5}$$

Since the flows are equal, equation (4) can be simplified.

$$\begin{aligned} q_{\text{perf}} = wc_b\Delta^2[\{ & T(x+\Delta) - 2T + T(x-\Delta) + T(y+\Delta \\ & - 2T + T(y-\Delta) + T(z+\Delta) - 2T + T(z-\Delta)\}/\Delta^2]/6 \end{aligned} \tag{6}$$

The expression within the brackets [ ] is the finite difference form of the Laplacian $\nabla^2 T$. So the perfusion heat can be approximated by the Laplacian operator:

$$q_{\text{perf}} \approx w\Delta^2 c_b \nabla^2 T/6 \tag{7}$$

The simple bioheat equation becomes

$$\rho_m c_m \partial T/\partial t \approx K_{\text{eff}} \nabla^2 T + q_{\text{ext}} + q_{\text{met}} \tag{8}$$

Nomenclature

a = thermistor radius (cm)
c = specific heat (mW-s/g-°K)
d = thickness of the decoupler between probe and tissue (cm)
F = blood flow to/from the control volume across one side (cm^3/s)
I = steady state $P/\Delta T$ (mW/°K)
k = empirical calibration coefficients
K = thermal conductivity (mW/cm-°K)
p = mass fraction of glycerol/agar-gelled water mixture
P = applied thermistor power (mW)
q = distributed heat source (mW/cm^3)
S = slope of $P/\Delta T$ ($mW/°K\text{-}s^{-1/2}$)
t = time (s)
T = temperature (°K)
V = electrical voltage (volts)
w = tissue perfusion (g/cm^3-s)
x, y, z = coordinates (cm)
α = thermal diffusivity (cm^2/s)
ρ = density (g/cm^3)
Δ = dimension of the cubic control volume (cm)
ΔT = $T_h - T_0$ volume average thermistor temperature rise (°K)

Subscripts

0 = initial time, baseline
a = arterial or in flow
b = blood
d = decoupler between probe and tissue
eff = effective (with perfusion)
g = glycerol
h = heated
m = intrinsic tissue (without perfusion)
met = tissue metabolism
mix = glycerol/agar-gelled water mixture
perf = due to perfusion
v = venous or out flow
w = agar-gelled water

where

$$K_{\text{eff}} \approx K_m + w\Delta^2 c_b/6 \tag{9}$$

Numerical and experimental studies have shown that the thermistor is most sensitive to the tissue thermal properties within about 2 thermistor radii (Tables 2 and 3). Let a be the radius of the thermistor. By matching the volume of a sphere with radius $2a$ with the volume of a cube with side Δ, an estimation of Δ can be found,

$$\Delta^3 = 4/3\pi(2a)^3 \tag{10}$$

Typical values were used in equations (9) and (10) to simulate the data shown in Table 1. These values are consistent with *in vivo* measurements (Bowman et al. 1977, Valvano et al. 1984a).

Instrumentation

Probe Design. The surface probe consists of a thermistor, partially embedded in an insulator, placed on the surface of a tissue. A Thermometrics P60DA102 thermistor, with a radius of 0.08 cm, is used as the active transducer. The electrical resistance is about 1 KΩ at 25°C. An insulator with low thermal conductivity and diffusivity is used to direct the heat supplied by the thermistor into the tissue. The insulator also shields the thermistor from air currents. Although smaller thermistors have faster responses, the sensitivity to conductivity and perfusion increases with probe size (Valvano 1984a).

Measurement Protocol. A microcomputer based instrument has been developed to control the thermistor probe and perform the calculations to measure effective thermal properties. Figure 2 is a block diagram of the analog instrumentation. The measurement of thermal properties and perfusion involves placing the thermal probe in contact with the perfused tissue, which is initially at a constant baseline temperature T_0. The probe is allowed to equilibrate thermally with the medium. By selecting the 0.1mA precision current source, the instrument first passively measures T_0. The thermistor is then self-heated, and the applied power is dissipated into the tissue medium. The 5.00V precision voltage source is applied across the series combination of a 783Ω precision resistor and the 1KΩ thermistor for 60 seconds. By measuring V_1 and V_2, the microcomputer can calculate both the applied power and the resulting thermistor temperature. The total power is about 10 mW. This power results in a volume average thermistor temperature rise of about 4°C. The temperature of the thermistor surface increases about 1.5°C. The heated thermistor temperature and applied power are recorded during the 60 second heating interval. The applied power reaches steady state within 10 seconds, and the thermistor temperature gradually increases. A linear regression is used to estimate the steady state response from the transient data. $P(t)/(T_h(t)-T_0)$ follows $t^{-1/2}$ (Chato 1968, Bowman et al. 1977, Valvano et al. 1984a).

$$P(t)/(T_h(t)-T_0) = I + St^{-1/2} \tag{11}$$

The heated temperature $T_h(t)$ is assumed to be the average temperature in the bead over the entire volume of the bead. The power $P(t)$ is the total power applied to the bead. The steady state response, I, depends on the thermal conductivity of the medium. The transient response, S/I, depends on the thermal diffusivity of the medium. Perfusion increases I and decreases S/I.

The parametric equation used to measure conductivity

$$K_{\text{eff}} = 1/(k_1/I + k_2) \tag{12}$$

is identical to that used for the invasive thermistor (Bowman et al. 1977). The parametric equation for diffusivity

$$\alpha_{\text{eff}} = 1/(k_3 S/I + k_4)^2 \tag{13}$$

was modified from the equation used for the invasive thermistor (Valvano et al. 1984a). It is necessary to calibrate the thermal probe before it can be used. The calibration constants k_1, k_2, k_3, k_4 are determined from measurements in glycerol and agar-gelled water. Since equations (12) and (13) independently measure K and α, the specific heat of the medium can also be determined.

The tissue control volume under a surface probe has less blood flow than the tissue control volume for an invasive probe. If the insulator of the surface probe is perfect, K_{eff} will be still a linear function of perfusion but with a different slope, and equation (9) can be rearranged to allow calculation of perfusion,

$$w = k_5(K_{\text{eff}} - K_m)/c_b \tag{14}$$

In order to quantify perfusion, the calibration coefficient k_5 must be determined empirically. The probe is placed on a tissue with known K_m, c_b, w, and K_{eff} is measured. k_5 incorporates the geometry and thermal properties of the probe. The quantification of perfusion is limited by the uncertainty of K_m *in vivo* (Valvano et al. 1984a).

Numerical Studies

Effect of a Decoupler Between the Probe and the Tissue. Numerical studies were used to estimate the error caused by an imperfect thermal/physical contact between the thermistor and the tissue probe. An imperfect thermal contact could result from a thin air film or a thin liquid film between the probe and tissue. The sensitivity to perfusion will also be

Table 1 Simulated values of K_{eff} versus perfusion for various thermistor sizes

Thermometrics Thermistor Type	Radius a(cm)	Δ(cm)	K_{eff}(mW/cm-°K) at $w=0$	at $w=0.01$	at $w=0.02$ g/cm^3-s
P20BA102	0.03	0.08	5.00	5.04	5.09
P60BA102	0.08	0.25	5.00	5.44	5.88
P100DA102	0.13	0.41	5.00	6.18	7.35

Table 2 Conductivity errors due to a decoupler between the probe and tissue

decoupler conductivity	decoupler thickness	Percent errors in conductivity measurement					
		$a=0.03$ cm		$a=0.08$ cm		$a=0.13$ cm	
K_d(mW/cm-°C)	d(cm)	Surface	Invasive	Surface	Invasive	Surface	Invasive
3	0.005	9	11	4	4	2	3
3	0.01	15	17	6	7	4	6
3	0.05	30	30	21	21	17	18
4	0.005	4	4	1	2	1	1
4	0.01	7	7	3	3	2	2
4	0.05	14	14	9	9	7	7

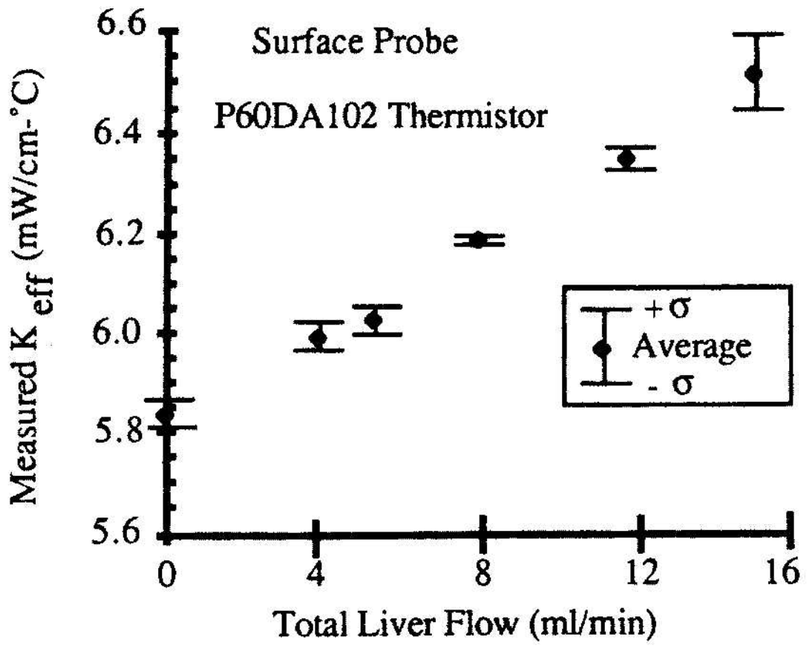

Fig. 8 Measured K_{eff} versus total flow in an isolated rat liver

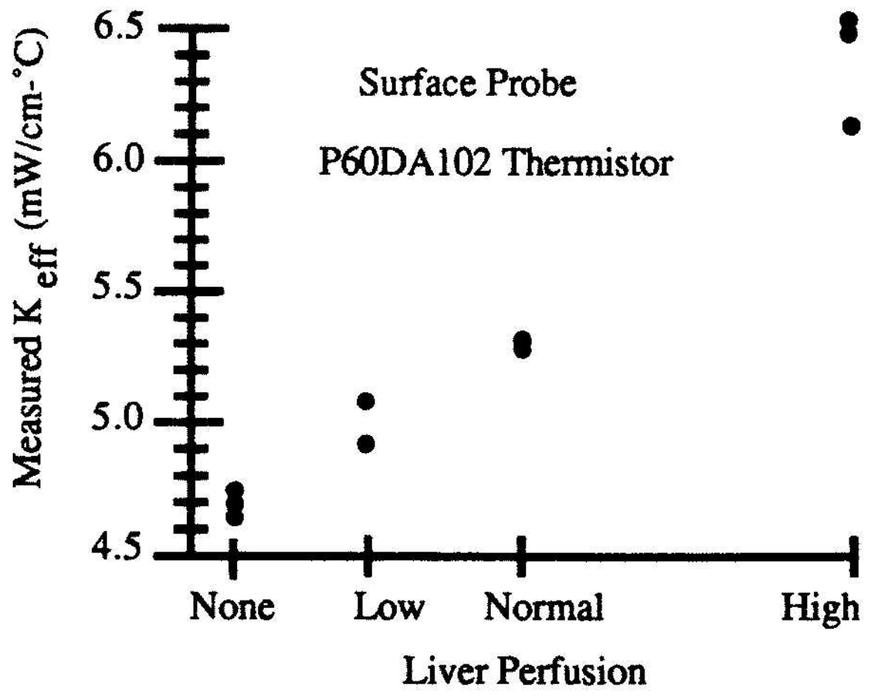

Fig. 9 Measured K_{eff} versus perfusion in an *in vivo* rat liver

tivity of the decoupler, the thickness of the decoupler, the heating interval, and the size of the probe. A numerical study showed that the baseline tissue temperature drift must be less than 0.03°C/min to ensure reliable thermal conductivity measurements.

The surface measurements of the thermal conductivity and thermal diffusivity were accurate to 2 to 4 percent respectively. *In vitro* experiments to determine the effective depth of measurement showed that the probe was sensitive to the thermal conductivity of a medium located at distances less than 3 mm (4 thermistor radii). This effective depth increases with probe size and length of heating interval. Unfortunately, the error due to baseline tissue temperature drift increases with the length of heating interval. The effects of poor contact were found to cause significant errors in the measured thermal properties.

The isolated rat liver apparatus design is simple and provides adequate control over baseline temperature, total flow rate and pH. Unfortunately, nonuniform perfusion fields in the isolated rat liver prevented quantification of perfusion using the surface probes. Both the isolated and *in vivo* rat liver experiments demonstrate the surface probes are sensitive to changes in perfusion.

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decreased by the presence of an unperfused layer at the tissue surface.

The finite element numerical method was used to evaluate errors caused by imperfect contact. Figure 3 contains the geometries for the finite element models of the invasive and noninvasive probes. Equation (8) was used to model the time-varying heat transfer in the tissue. Three-dimensional results were obtained by using a two-dimensional finite element code and rotating the grid about the vertical axis. The validity of the finite element model was established by comparing the numerical results of the invasive probe without a decoupler ($d = 0$) with the analytic solution (Valvano et al. 1984a). The transient finite element code was first run with ($d=0$, K_m = 3mW/cm-°C), and ($d=0$, K_m = 5mW/cm-°C) simulating the calibration procedure. Calibration k_1 and k_2 were calculated using equation (12). The transient finite element code was then run with K_m fixed at 5mW/cm°C and various a, d, and K_d. The errors in the surface probe were slightly less than those in the invasive probe. The conductivity error increases as $K_m - K_d$ increases, as the thickness of the decoupler increases, and as the size of the probe decreases. The error due to the decoupler also decreases as the heating interval increases.

Errors in the Conductivity Measurement Due to Drifts in Tissue Temperature. Reproducible measurements can be obtained in the laboratory because the thermal conditions can be controlled. One significant source of error, found clinically, is the instability of the baseline temperature. Consider the situation where the baseline tissue temperature, T_0, is falling due a process independent from the probe. The single thermistor is used to measure T_0 before the heating interval. Since T_0 is not measured during the 60 second heating interval, there will be an error in the measured ΔT.

A numerical simulation was performed to evaluate the effect of temporal variation in tissue temperature on the measurement of thermal properties. Typical values of $P(t)$ and $T_h(t)$ were used. The effect of temperature drift was simulated by adding a time varying drift to the $T_0(t)$ data. The thermal conductivity was then recalculated using equations (11) and (12). The numerical experiment was repeated for various heating intervals and values of dT/dt. From the results, shown in the Fig. 4, it can be seen that the conductivity error increases with dT/dt and heating interval. With a 60 second heating interval, it is our experience that the baseline temperature drift must be less than 0.03°C/min to ensure accurate thermal conductivity measurements.

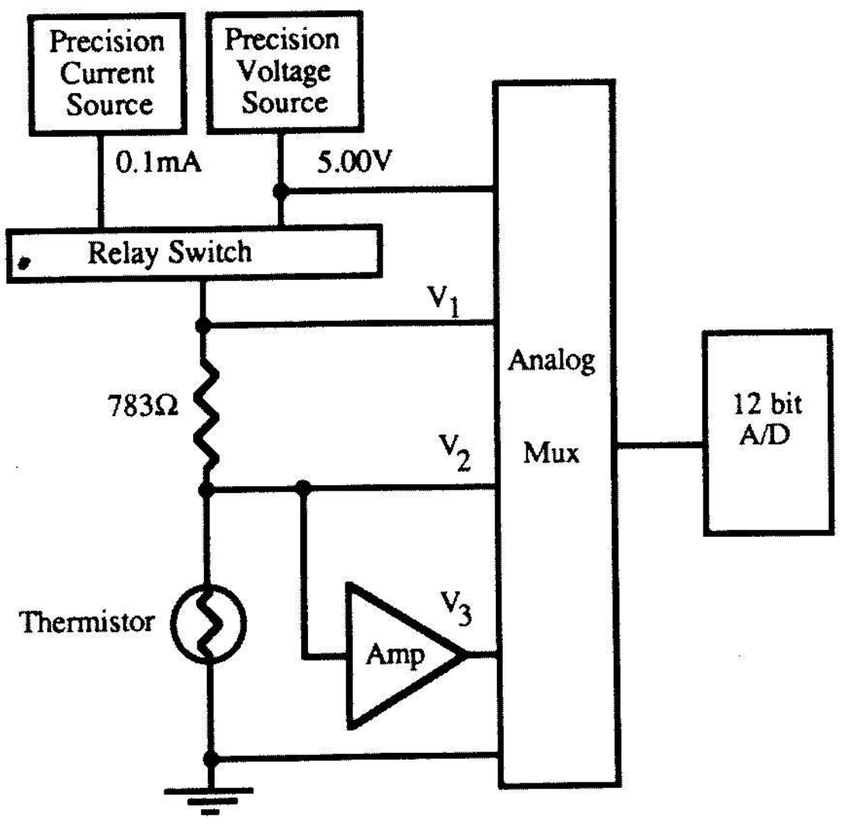

Fig. 2 Analog instrumentation

In Vitro Experiments

Effective Depth of Measurement. The effective depth of measurement is defined as the distance from the probe at which the probe becomes insensitive to changes in the thermal properties of the medium. This study used a composite medium of agar-gelled water and glycerol (Fig. 5). The thickness of the glycerol layer was varied from 0 to 0.5 cm. The measured conductivity can be considered as a volume average conductivity of the two liquids. As the thickness of the glycerol increases, the effect of the agar-gelled water on the measurement decreases.

The effective depth of measurement was estimated to be the thickness of glycerol at which the measured conductivity was 3.06 mW/cm-°C. This conductivity value (3.06 mW/cm-°C) represents 5 percent agar-gelled water and 95 percent glycerol. The thermal conductivities were measured for 20, 40, and 80 second heating intervals to estimate the effect of the heating interval on the effective depth of measurement. A Thermometrics P60DA102 thermistor (a = 0.98 cm) was used and the experimental results are presented in Fig. 6. From Table 3 it can be seen that the effective measurement increases as the duration of the heating interval increases. Effective depth of measurement also increases with thermistor size. These results can not be generalized to all situations because they are dependent on the geometry and the thermal properties of the two liquids. Nevertheless, the results are consistent with the numerial studies.

Accuracy of the Thermal Property Measurements. The accuracy of the thermal conductivity and diffusivity measurements was tested using seven liquid standards. The seven liquid standards used were glycerol, agar-gelled water,

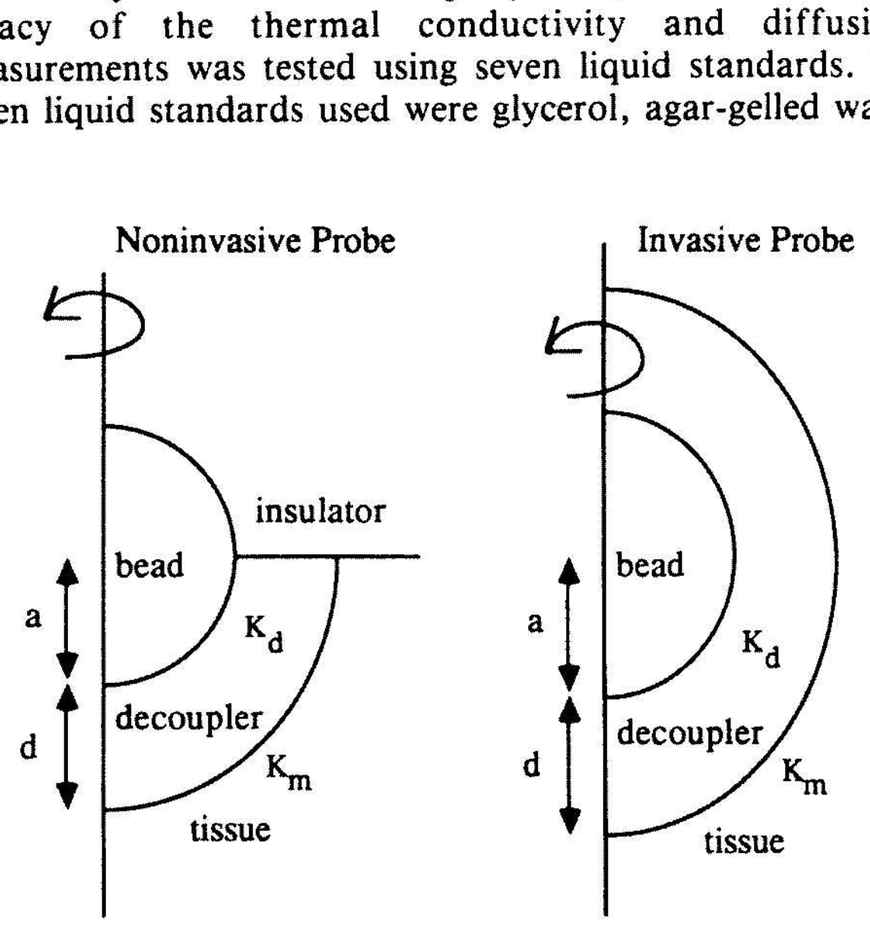

Fig. 3 Block diagram of the finite element model of the thermistor probe

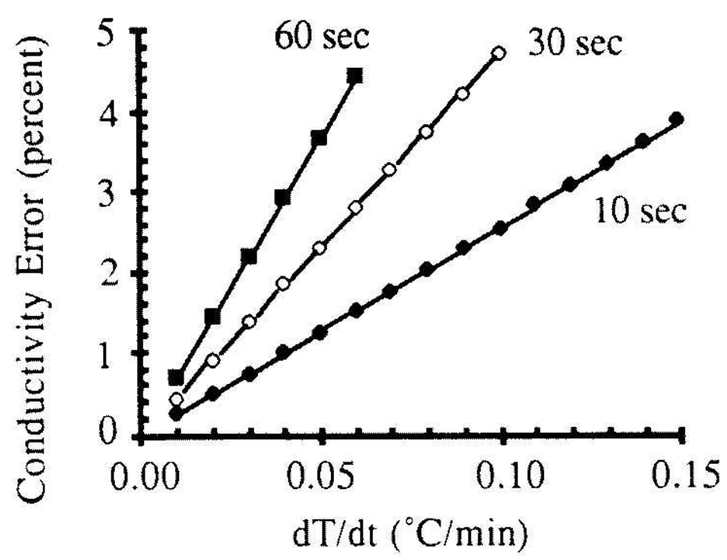

Fig. 4 Measurement error versus baseline temperature drift

and five mixtures of glycerol and agar-gelled water. The water was gelled with agar to prevent convection. The thermal properties of the water are not significantly altered by the presence of agar (Patel 1986). The thermal conductivity and thermal diffusivity of the mixtures were calculated from the mass fractions p_g and p_w (Valvano et al. 1985),

$$K_{mix} = K_g p_g + K_w p_w - 1.4 p_g p_w [K_w - K_g - 2] + 0.014 p_w p_g (T_0 - 20) \quad (15)$$

$$\alpha_{mix} = \alpha_g p_g + \alpha_w p_w \quad (16)$$

Twelve experiments were conducted in each of seven media and the measured I and S/I were recorded. After discarding highest and lowest readings, the average I and S/I were calculated. Using average values of I and S/I, the thermal conductivity and diffusivity were calculated for each medium. The average conductivity and diffusivity errors for the probe were 0.7 and 2.1 percent, respectively and the maximum errors were 1.4 and 3.9 percent, respectively. The average errors are similar to the 1.5 and 0.7 percent accuracy of Bowman's invasive probe (Valvano et al. 1984a).

Isolated Rat Liver Experiments

To determine the accuracy of the perfusion measurement, it is necessary to know the true perfusion of the tissue on which the probe is placed. It is difficult to control both the perfusion and temperature of a tissue *in vivo*. Hence, an isolated rat liver apparatus (Fig. 7) was used to evaluate the perfusion measurements (Brauer et al. 1951, Bartosek et al. 1973, Brunengraber et al. 1973, Valvano et al. 1984b). Figure 8 plots the measured effective thermal conductivity versus total liver flow during a typical isolated rat liver experiment.

Probe placement was a critical factor. Poor or imperfect contact between the probe and the tissue surface caused widely scattered measurements even when the baseline temperature was stable. When the contact was very poor, the measurements were insensitive to perfusion. Repeated measurements with the same flow demonstrated the reproducibility of the measurement. Perfusion could not be quantified because of the nonuniform perfusion field in the isolated rat liver. These results suggest a linear relation between K_{eff} and perfusion.

In Vivo Rat Liver Experiments

The objective of the *in vivo* rat liver experiments was to demonstrate the sensitivity of the surface probe to changes in perfusion. The results, summarized in Fig. 9, also demonstrate the sensitivity of the surface probe. The following experimental protocol was used:

1. Anesthetize the rat with sodium pentabarbitol IP;
2. Make a small incision in the abdomen to expose the liver;
3. Place the surface probe on the liver;
4. Place the rat in an isothermal chamber at 37°C;
5. Wait for the temperature stability;
6. Measure K_{eff} with the microcomputer based instrument under conditions of normal flow
   use intropin to increase perfusion
   use epinepherine to decrease perfusion;
7. Sacrifice the rat with an overdose of sodium pentabarbitol;
8. Measure K_m.

Conclusion

A simplified bioheat equation has been presented in which the effective tissue thermal conductivity is a linear function of perfusion. The measurement errors caused by the presence of a decoupler were slightly less in the surface probe than in the invasive probe. The conductivity error is a function of conduc-

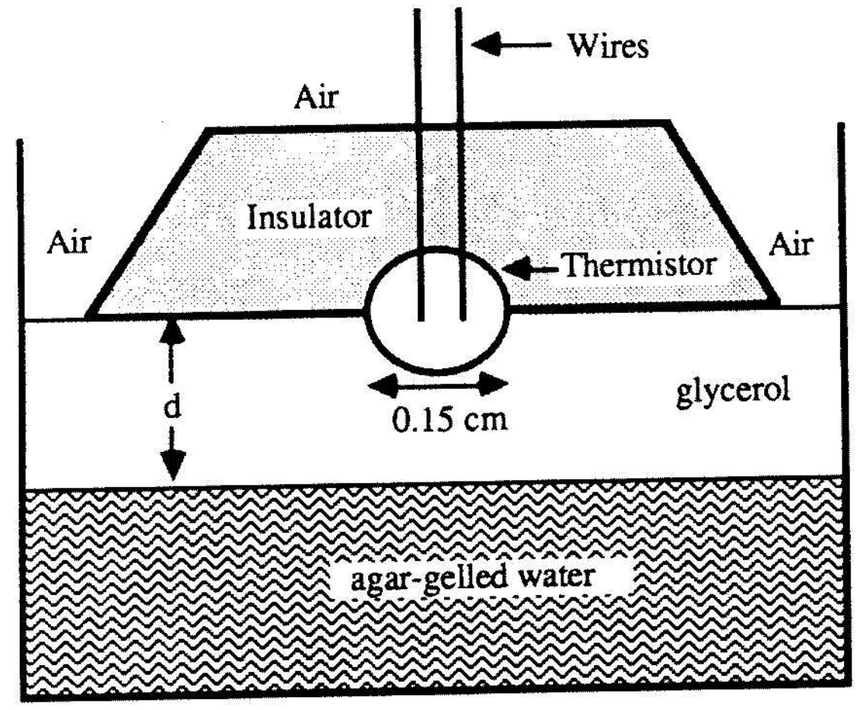

Fig. 5 Apparatus to determine effective depth of measurement

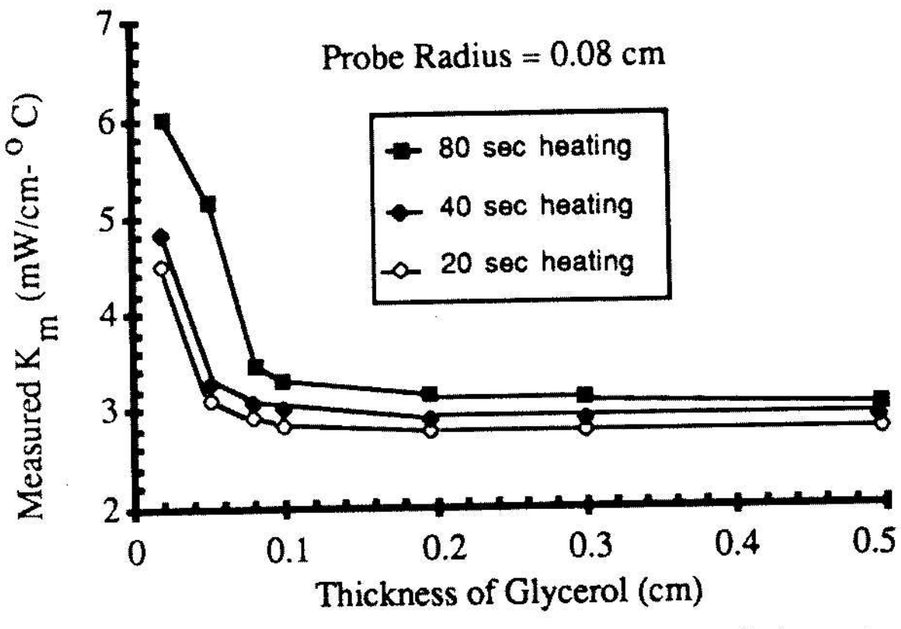

Fig. 6 Measured conductivity versus thickness of glycerol

Table 3 Effective depth of measurement

Heating interval	Depth of measurement
20 s	0.9 radii 0.07 cm
40 s	1.3 radii 0.10 cm
80 s	4.3 radii 0.34 cm

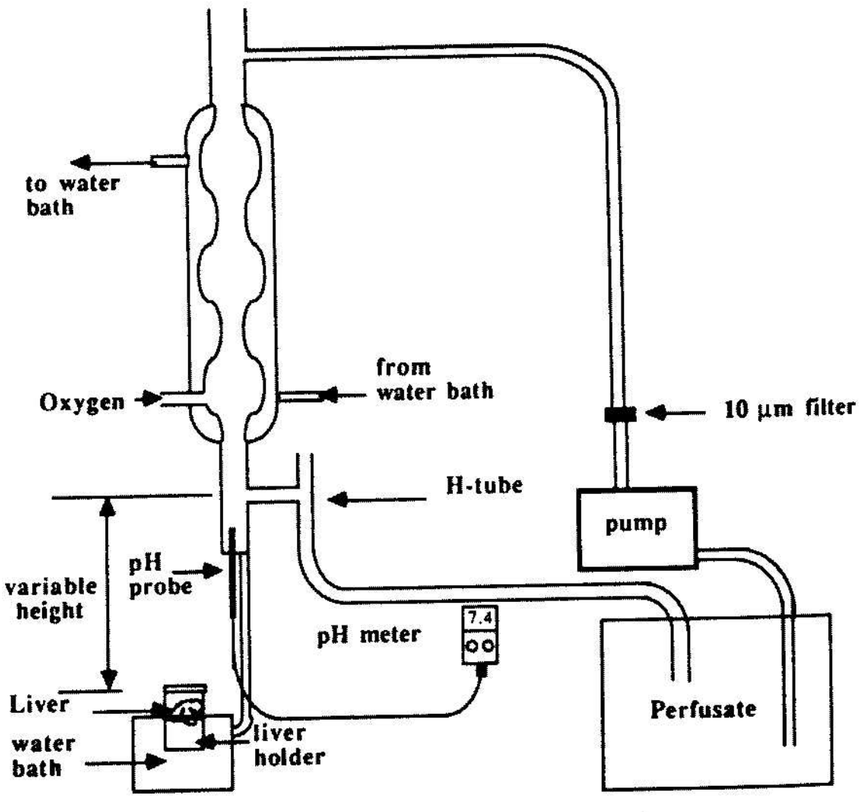

Fig. 7 Isolated rat liver apparatus

and five mixtures of glycerol and agar-gelled water. The water was gelled with agar to prevent convection. The thermal properties of the water are not significantly altered by the presence of agar (Patel 1986). The thermal conductivity and thermal diffusivity of the mixtures were calculated from the mass fractions p_g and p_w (Valvano et al. 1985),

$$K_{\text{mix}} = K_g p_g + K_w p_w - 1.4 p_g p_w [K_w - K_g - 2] + 0.014 p_w p_g (T_0 - 20) \quad (15)$$

$$\alpha_{\text{mix}} = \alpha_g p_g + \alpha_w p_w \quad (16)$$

Twelve experiments were conducted in each of seven media and the measured I and S/I were recorded. After discarding highest and lowest readings, the average I and S/I were calculated. Using average values of I and S/I, the thermal conductivity and diffusivity were calculated for each medium. The average conductivity and diffusivity errors for the probe were 0.7 and 2.1 percent, respectively and the maximum errors were 1.4 and 3.9 percent, respectively. The average errors are similar to the 1.5 and 0.7 percent accuracy of Bowman's invasive probe (Valvano et al. 1984a).

Isolated Rat Liver Experiments

To determine the accuracy of the perfusion measurement, it is necessary to know the true perfusion of the tissue on which the probe is placed. It is difficult to control both the perfusion and temperature of a tissue *in vivo*. Hence, an isolated rat liver apparatus (Fig. 7) was used to evaluate the perfusion measurements (Brauer et al. 1951, Bartosek et al. 1973, Brunengraber et al. 1973, Valvano et al. 1984b). Figure 8 plots the measured effective thermal conductivity versus total liver flow during a typical isolated rat liver experiment.

Probe placement was a critical factor. Poor or imperfect contact between the probe and the tissue surface caused widely scattered measurements even when the baseline temperature was stable. When the contact was very poor, the measurements were insensitive to perfusion. Repeated measurements with the same flow demonstrated the reproducibility of the measurement. Perfusion could not be quantified because of the nonuniform perfusion field in the isolated rat liver. These results suggest a linear relation between K_{eff} and perfusion.

In Vivo Rat Liver Experiments

The objective of the *in vivo* rat liver experiments was to demonstrate the sensitivity of the surface probe to changes in perfusion. The results, summarized in Fig. 9, also demonstrate the sensitivity of the surface probe. The following experimental protocol was used:

1. Anesthetize the rat with sodium pentabarbitol IP;
2. Make a small incision in the abdomen to expose the liver;
3. Place the surface probe on the liver;
4. Place the rat in an isothermal chamber at 37°C;
5. Wait for the temperature stability;
6. Measure K_{eff} with the microcomputer based instrument under conditions of normal flow
   use intropin to increase perfusion
   use epinepherine to decrease perfusion;
7. Sacrifice the rat with an overdose of sodium pentabarbitol;
8. Measure K_m.

Conclusion

A simplified bioheat equation has been presented in which the effective tissue thermal conductivity is a linear function of perfusion. The measurement errors caused by the presence of a decoupler were slightly less in the surface probe than in the invasive probe. The conductivity error is a function of conduc-

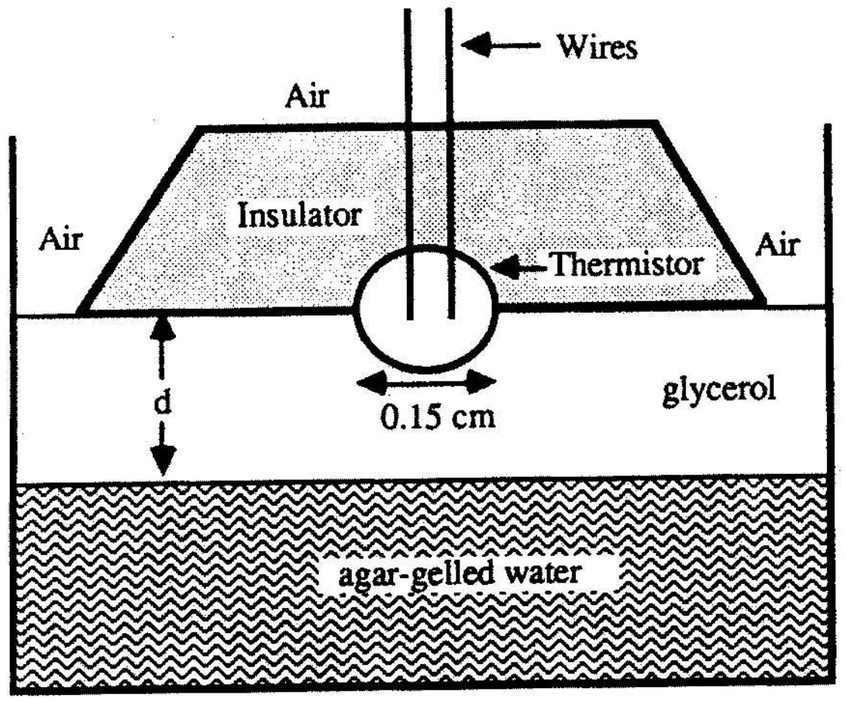

Fig. 5 Apparatus to determine effective depth of measurement

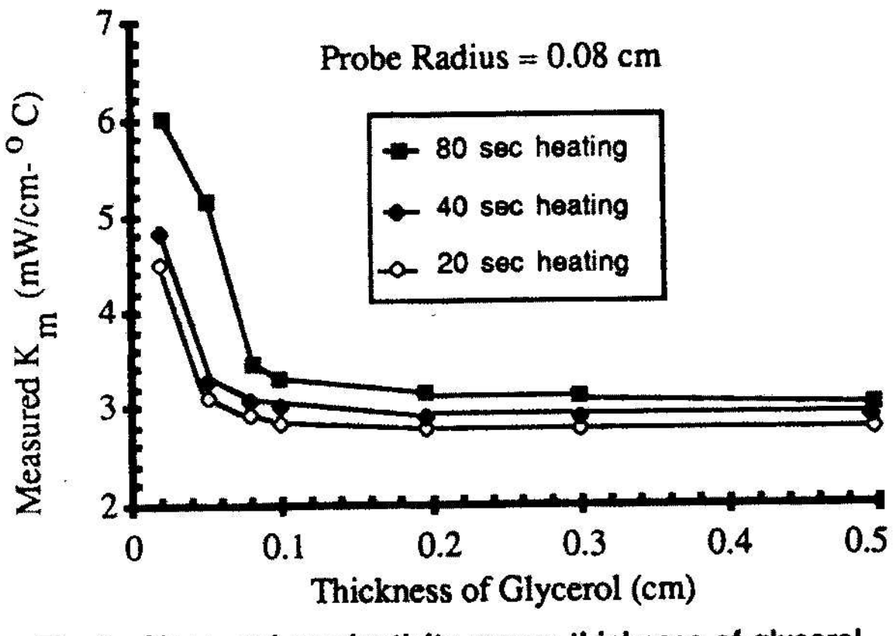

Fig. 6 Measured conductivity versus thickness of glycerol

Table 3 Effective depth of measurement

Heating interval	Depth of measurement
20 s	0.9 radii 0.07 cm
40 s	1.3 radii 0.10 cm
80 s	4.3 radii 0.34 cm

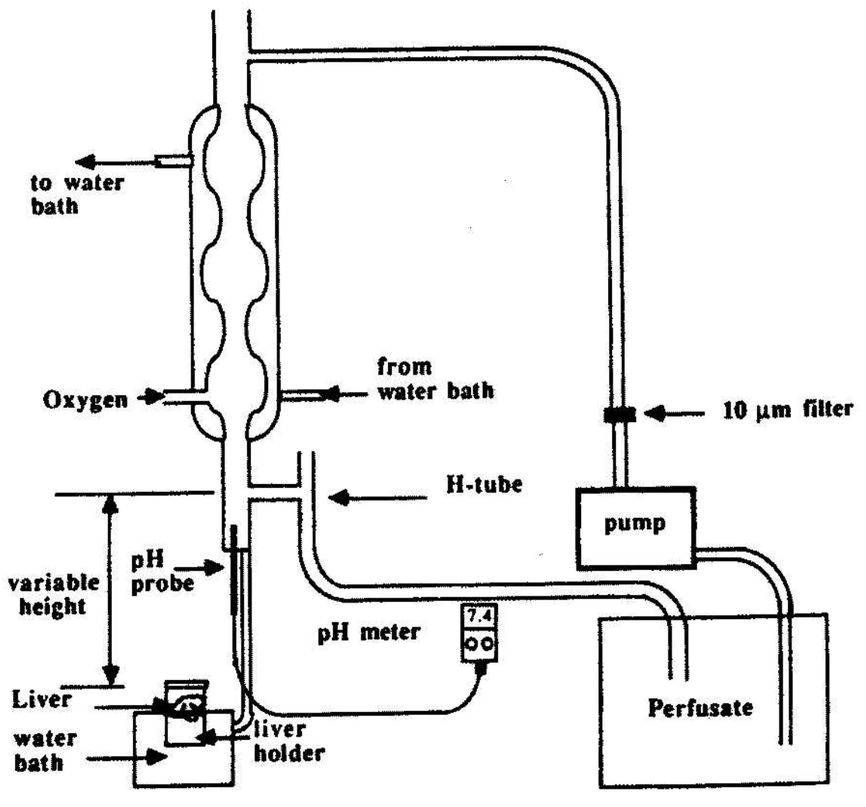

Fig. 7 Isolated rat liver apparatus