IN VITRO ANALYSIS OF EXTRACORPOREAL BLOOD HEAT EXCHANGE DEVICES.

J. B. Riley and B. A. Winn

Ohio State University, Columbus, Ohio.

"Fellowship Award Winner"

PURPOSE

An original protocol, utilizing a dual recirculating equilibrium circuit, was designed to evaluate heat transfer in four internal, three external and one prototype blood heat exchange devices. The results of this study set present standards for the evolution of new blood heat exchange devices.

BACKGROUND

Before considering the quantitation of heat transfer in an extracorporeal circuit blood heat exchange device, the process of heat exchange must be analyzed. Figure 1 lists the abbreviations that will be employed in the presentation.

ABBREVIATION	TERM	UNITS
A	surface area	cm^2
c	specific heat	cal/gm · °C
cal	calorie	amount of heat required to raise 1 gram of H_2O from 14.5°C to 15.5°C
C_{HE}	Coefficient of Heat Exchange	
C_{HA}	Coefficient of Heat Accountability	
°C	degree Centigrade	registers freezing pt of H_2O at 0°C and boiling pt of H_2O at 100°C
D	density	gram/milliliter
E.C.C.	extracorporeal circuit	
F_B	blood flow	milliliters/minute
F_W	water flow	liters/minute
H	heat capacity	specific heat x mass = cal/°C
Hct	hematocrit	% volume Red Blood cells
k	thermal conductivity	cal/sec · cm^2 · °C/cm
Kcal	kilocalorie	1000 calories
m	mass	grams
$\dot{m}$	mass flow	mass/unit time = gms/min
Q	heat	calories or kilocalories
$\dot{Q}$	heat flow	heat/unit time = Kcal/min
T_x	temperature	°C, refers to measuring point
ΔT	temperature difference	°C
t	time	seconds or minutes
Δx	thickness	cm

Figure 1

In E.C.C. blood heat exchange devices the majority of heat transfer is through the process of conduction, where thermal energy is passed from molecule to neighboring molecule in the conductor. Convection, defined as heat transfer from one point to another by mass motion of a gas or liquid medium, and ratiation, heat transfer by means of electromagnetic waves, will not be considered or quantitated in this protocol.[1-2-4]

A single temperature measurement is independent of the size or mass of a system and reflects only the degree of the absence or presence of heat. The relationship between temperature change, heat exchanged and the mass of a body is stated: Eq. 1

$$Q = c \cdot m \cdot \Delta T$$

If any three of the above variables are known then the fourth may be calculated. For example, if a container is filled with 500 ml (gms) of water and resting at 25°C, adding 500 cal of heat to the water will change the temperature as follows.

$$500 \text{ cal} = (1 \text{ . cal/gm°C}) \times (500 \text{ gm}) \times (\Delta T) \quad \Delta T = 1°C$$

The, 500 cal added to the water raised the temperature from 25°C to 26°C. This process is reversed to quantiate the amount of heat added to the blood as it passes through an E.C.C. heat exchange device. Mass flow rate is substituted for m and heat flow is derived instead of Q. Eq. 2

$$Q = c \times m \times \Delta T$$

For example, if the T_{Bi} (T blood-in) = 28°C and T_{Bo} (T blood-out) = 33°C in a heat exchange device for blood of Hct = 25% at a F_B = 2500 ml/min, the heat flow would be calculated in the following manner:

$$m = D_{25\%\ Hct} \times F_B = (1.051 \text{ gm/ml}) \times (2500 \text{ ml/min})$$

$$m = 2628 \text{ gm/min}$$

and since

$$c_{25\%\ Hct} = .905 \text{ cal/gm°C}$$

then

$$Q = (.905 \text{ cal/gm°C}) \times (2628 \text{ gm/min}) \times (33\text{-}28°C)$$

$$Q = 2378 \text{ cal/min} = 23.78 \text{ Kcal/min}$$

Therefore the E.C.C. heat exchange device is delivering approximately 24 Kcal/min to the blood as it passes.

Figure 2 relates the same Q to the physical characteristics of an E.C.C. blood heat exchange device.[7,8,10]

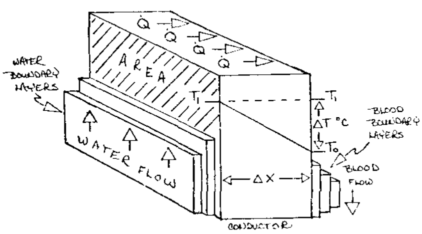


Figure 2

Equation 3 quantitates Figure 2 and the effects of changing surface area, medium residence time, conductor thickness, thermal conductivity and temperature difference across the conductor.

$$Q = A \times \frac{\Delta T}{\Delta x} \times k \times t \qquad \text{Eq. 3}$$

A change in any one design characteristic will result in a change in Q. It is evident from Figure 2 that it is the flow of heat from one side of the conductor to the other that affects a temperature change in the medium on either side at the time of equilibrium. Observing the temperature change between the entering and exiting medium (T_{Bi} - T_{Bo}) and comparing it to the temperature difference between the warming medium, T_{Wi} (T water-in) and medium to be warmed at their inlets, is the rationale for the Coefficient of Heat Exchange.[2] Equation 4 relates the comparison.

$$C_{HE} = \frac{T_{Bo} - T_{Bi}}{T_{Wi} - T_{Bo}} \qquad \text{Eq. 4}$$

The C_{HE} is dependent on several variables including heat exchanger water and blood flow, surface area, conductor thermal conductivity, thickness and boundary layer phenomenon. The C_{HE} is independent of T_{Wi} - T_{Bi}. When the C_{HE} is reported, the water flow and blood flow should be reported simultaneously. Typically, the C_{HE} decreases with increased blood flow yet the blood heat flow increases. At the same blood flow, increasing water flow increases the C_{HE} and blood heat flow, however, the measurements tend to level off as water flow further increases.

Figure 3 depicts the First Law of Thermodynamics — heat in equals heat out.[2]

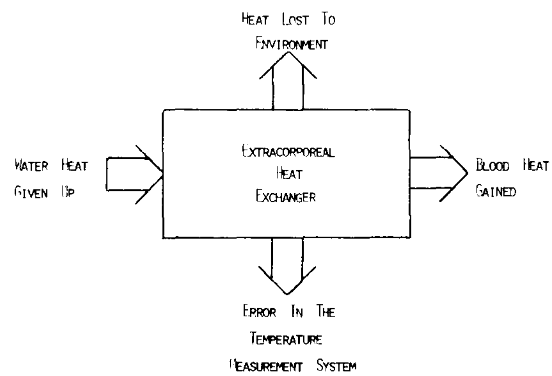


Figure 3

The water heat given up may be calculated using Equation 2, substituting values for the water path. Blood heat gained may be compared to water heat given up with the use of the Coefficient of Heat Accountability.

$$C_{HA} = \frac{\text{Blood heat gained}}{\text{Water heat given up}} \qquad \text{Eq. 5}$$

Any deviation from unit in the C_{HA} may be accounted for in heat loss to the environment (including gas phase in an integral unit) or in measurement error.

CIRCUIT

The *in vitro* circuit centers around a 15 L reservoir containing a glycerol in water solution. Figure 4 relates the physical properties of 21.6% glycerol to 25% hematocrit blood.[5,9,12,13]

Parameter in Order of importance to protocol	Units	21.6% glycerin	25% HCT Blood
Viscosity	dyne sec/cm^2 x 10^{-2}	2.18 @ 30°C	2.25 @ 37°C
Specific heat	cal/gm · °C	.915 @ 37°C	.93 @ 37°C
Surface tension	dyne/cm	72.5*	64.5
Thermal conductivity	cal/sec·cm·°C x 10^{-3}	1.27 @ 20°C	1.30 @ 24°C (Dog blood)
Density	gm/cm^3	1.051 @ 20°C	1.045 @ 37°C

* The only major discrepency, but equal to the surface tension of whole blood with a HCT $\cong$ 10%

Figure 4

Two twin roller pumps recirculate the reservoir contents through two metal external heat exchangers to maintain the reservoir at a constant temperature. A separate water delivery unit controls the water flow through and temperature in the maintenance heat exchangers. The maintenance heat exchanger water paths are connected in series. A calibrated twin roller pump delivers constant temperature blood (glycerol solution) from the reservoir and pumps it through the test heat exchanger blood path. The temperatures of the blood-in and blood-out are measured in close proximity to the test heat exchanger with 1/10°C precision matched mercury thermometers.* Two calibrated twin roller pumps in parallel draw water from a controlled temperature mixing valve and pump water through the test heat exchanger water path. Water-in and water-out temperatures are measured in close proximity to the test heat exchanger with 1/10°C precision thermometers matched to the blood path thermometers. Heat exchanger blood flow, water flow, $T_{Wi} - T_{Bi}$, $T_{Bi} - T_{room}$ and reservoir temperature are easily controlled or varied.

METHODS

Heat exchanger blood flow, water flow, temperature of blood-in and water-in were adjusted and simultaneously recorded with the measured room, blood-out and water-out temperatures at each equilibrium point. The gas to blood flow ratio was maintained at 1:1 in all integral units and $T_{Wi} - T_{Bi}$ held constant at 10°C unless otherwise stated. Once equilibrium was established five data sets were collected with every parameter change. Each heat exchange device was subjected to the following conditions.

1. $T_{Wi} - T_{Bi}$ was varied at 4°C, 8°C, and 12°C maintaining the heat exchanger water flow at 15 L/min and blood flow at 4000 ml/min.
2. Heat exchanger water flow was varied at 7.5, 10.0, 12.5, and 15.0 L/min at a blood flow of 5000 ml/min.[6]
3. Heat exchanger blood flows were varied at 1000, 2000, 3000, 4000, and 5000 ml/min at constant water flow of 15 L/min.
4. $T_{Wi} - T_{Bi}$ was maintained at 0°C and $T_{Bi} - T_R$ was adjusted to 15°C at a water flow of 15 L/min and a blood flow of 2000 ml/min.[11]

The Coefficient of Heat Exchange, water heat given up, blood heat gained and the Coefficient of Heat Accountability were calculated using the formulae (See BACKGROUND) for each data set. The mean and its standard deviation of C_{HE}, Q_{Blood} and $T_{Wi} - T_{Bi}$ were calculated for the five repetitions at each parameter change.

The heat exchanger water path resistance was quantitated by measuring the pressure drop across the water path at water flows of 7.5, 10.0, 12.5 and 15.0 L/min. Five measurements of pressure drops with a calibrated electronic pressure measurement system were taken at each water flow and the average of all the calculated resistances were reported.

In external heat exchange devices the blood path resistance was measured in a similar fashion to the water path resistance at blood flows of 1000 through 5000 ml/min.

RESULTS

Physical Characteristics

Prior to analyzing the results, it is appropriate to quantitate various physical properties which effect heat transfer. The importance of the physical characteristics listed in Figure 5 have been reported previously.[4,6,7,8]

Heat Exchanger Manufact. & Model #	- - - CONDUCTOR - - - Type Metal	Thick- ness	Therm. Cond. cal/sec ·cm·°C @ 20°C	Surface Area cm^2	Unit Priming Volume *mL	Mean Blood Path Thick- ness cm.	Blood Residence Time @ 4 L/min sec*
Bentley Q200A	Tin plated steel, urethane coated	15 mil	.57 .08	452	270	.597	4.05
Cobe 42-208	Alum. Polyurethane coating	(1/8 in.)	.48	1053	495	.470	7.42
Harvey H 1000	Alum. epoxy coating	10 mil	.48	2385	365	.153	5.48
Olson T-1003	Stainless steel #304	15 mil	.08	1750	135	.077	2.03
Olson Prototype	Nickel plated copper	.0381 cm	.86 .76	667	235	.353	3.53
Sarns 8348	Stainless steel #304	30 mil	.08	1844	125	.068	1.88
Shiley S100	Alumin. Polyurethane coating	18 mil	.48	2000	390	.195	5.85
Travenol 5M0338	Stainless steel #304	10 mil	.08	968	90	.093	1.35

Measurements made or estimated by authors in absence of availability of manufacturer's specifications.

* at a gas to blood flow ratio = 1:1

Figure 5

The values in Figure 5 were generated by the authors at normal operating conditions. Values from Figure 5 may be inserted into Equation 3 to predict the amount of heat flow. Knowledge of the physical characteristics of an individual unit aids in understanding that unit's performance in the methodology.

Effects of Temperature Gradients

The performance of each heat exchange device as the temperature gradient between water-in and blood-in is varied at a constant blood flow of 4 L/min and a constant water flow of 15 L/min is presented in Figure 6.

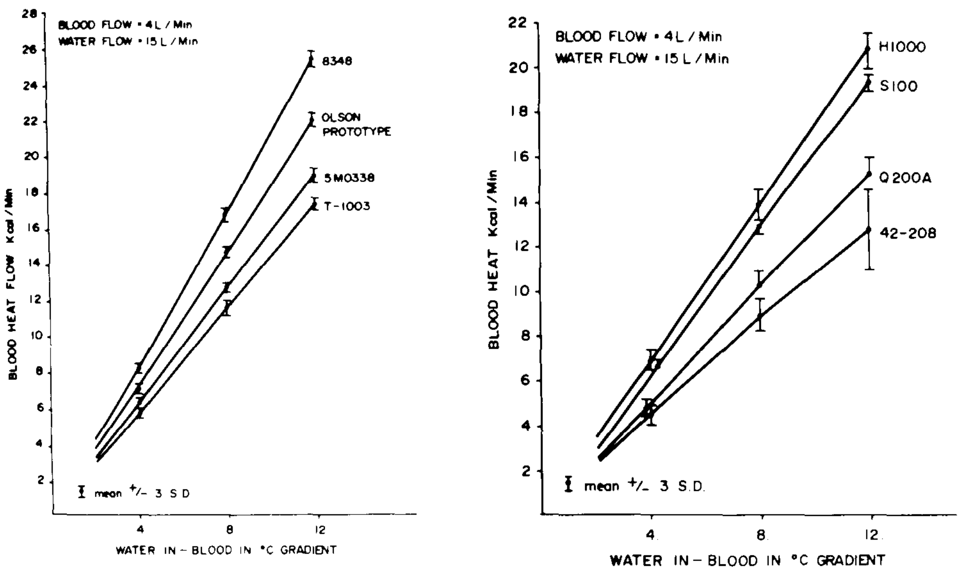


Figure 6

Blood heat flow is a linear function of temperature gradient between the water-in and the blood-in to the heat exchange device. As the blood-in temperature, approaches the water-in temperature in an E.C.C. heat exchanger, decreasing the gradient, blood heat flow diminishes. A decreased blood heat flow may be observed in a normal cardiopulmonary bypass procedure in that it takes longer to warm a patient from 35 - 37°C than it does from 28 - 30°C if the inlet water temperature is held constant at 40°C.

The Coefficient of Heat Exchange was found to be independent of the magnitude of $T_{Wi} - T_{Bi}$. Although blood heat flow dramatically increases with increased $T_{Wi} - T_{Bi}$, the C_{HE} does not reflect the change.

Effects of Altering Water Flow

The effects of altering water flow at a constant blood flow of 5 L/min and a constant $T_{Wi} - T_{Bi}$ of 10°C are depicted in Figure 7.

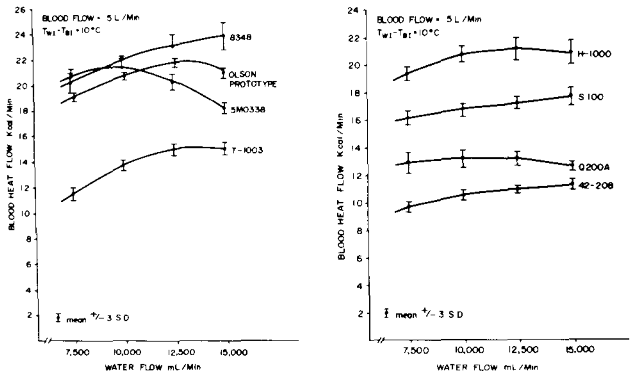
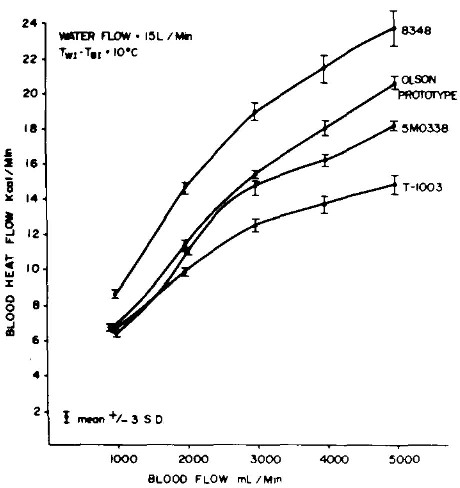


Figure 7

In general the blood heat flow and the Coefficient of Heat Exchange increased with increased water flow. Most of the units reached 95% of their maximum blood heat flow at a water flow of 12.5 L/min. Increasing the water flow decreases the water side boundary layers which form on the conductor and also maintains a greater temperature gradient down the length of the conductor leading to increased heat flow. The water paths of some units are designed to create turbulence to increase water heat delivery to the conductor via convection. Except for the 5M0338, the units with less turbulent water paths, (S100, Olson Prototype, H1000, Q200A) demonstrated a decreased blood heat flow at water flows greater than 12.5 L/min.

Blood Heat Flow Versus Blood Flow

Figure 8 presents the affects that increasing the blood flow has on blood heat flow at a constant water flow of 15 L/min and T_{Wi} - T_{Bi} = 10°C.



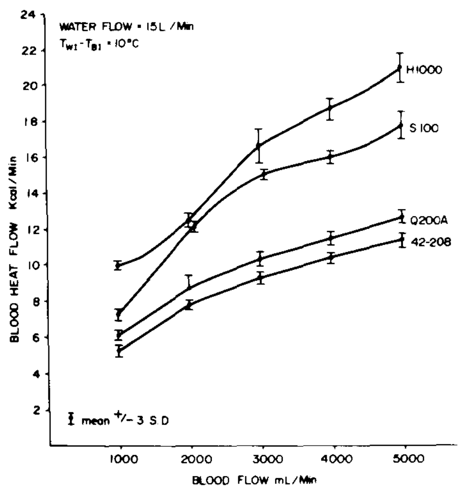


Figure 8a

Figure 8b

Blood heat flow rises exponentially as blood flow is increased. The increased blood velocity decreases the blood boundary layer and decreases the magnitude of the numerator of the C_{HE}, T_{Bo} - T_{Bi}. Thus the C_{HE} decreases whereas the number of kilocalories of heat delivered by the blood per minute increases. Increasing the blood flow during total cardiopulmonary bypass will increase heat delivery to the patient and decrease the warming time, although not linearly.

Heat Loss to Environment

Figure 9 presents the heat loss to the environment at a blood flow of 2000 ml/min, a water flow of 15 L/min, a $T_{Wi} - T_{Bi} = 0°C$ and a gradient between blood-in and room temperature = 15°C.

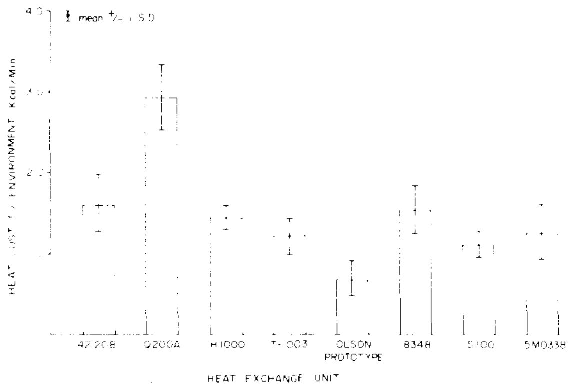


Figure 9

The surface area, thermal conductivity and thickness of the device materials that are in contact with blood and exposed to the environment dictate the heat loss. Increased length of the oxygenating column, increased gas to blood flow ratios, decreased room temperatures and increased room air flow contribute to environmental heat loss. In most units, the heat not gained by the blood path that was given up by the water path, approximated the heat loss to the environment under identical operating conditions.

Blood and Water Path Resistances

The opposition to water flow that the heat exchange water paths offer is presented in Figure 10.

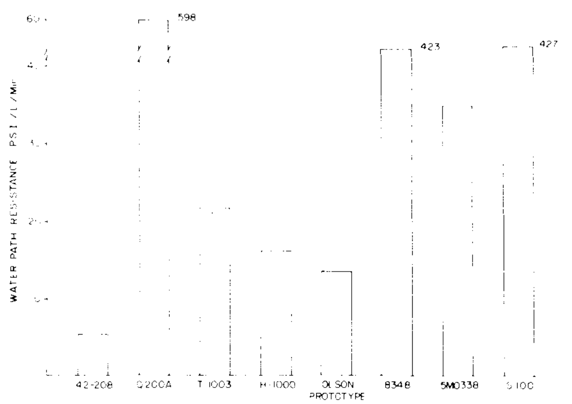


Figure 10

If the lengths of the wall water delivery and drain hoses are minimized and the diameters of all water carrying components, including connectors and temperature mixing valve, are maximized, then the resistance of the heat exchanger water path becomes the major determinant of water flow at a constant wall pressure. Assuming the pressure at the drain and water conduit resistances are negligible, Equation 6 may be used to predict the heat exchanger water flow.

$$\text{Water Flow} = \text{Wall Pressure} \quad \text{Water Path Resistance} \qquad \text{Eq. 6}$$

Figure 11 quantitates the blood path resistances of the external heat exchange devices.

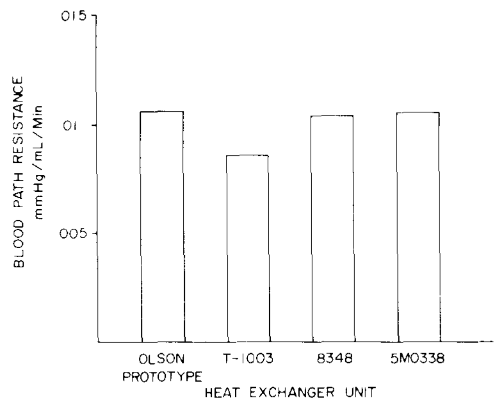


Figure 11

When the external unit is employed in the E.C.C., the blood path resistance contributes to the total patient arterial re-entry line resistance. The arterial line pressure contribution may be calculated using Equation 7.

$$\begin{array}{l} \text{Arterial line} \\ \text{pressure con-} \\ \text{tribution mmHg} \end{array} = \begin{array}{l} \text{Blood} \\ \text{Flow} \\ \text{ml/min} \end{array} \times \begin{array}{l} \text{Heat exchange} \\ \text{Blood Path} \\ \text{Resistance} \\ \text{mmHg/ml/min} \end{array} \qquad \text{Eq. 7}$$

If the external unit is employed in the venous siphon drainage return line, the necessary height gradient between the patient's right atrium and the effective level of the oxygenating device may be calculated for a given blood flow. For example the blood path resistance for the Olson Prototype is .0106 mmHg/ml/min and if a venous return flow of 5000 ml/min is required, then a column of blood exerting a pressure of 53 mmHg is necessary. The column converts to 73.2 cm of blood or a 2.4 foot height difference.

DISCUSSION

If total body hypothermia is utilized during cardiac surgery then the main objective of an E.C.C. blood heat exchange device is to subtract and add calories to the patient rapidly and safely. If normothermia is employed, a device with less capacity for caloric exchange is suitable, however, hypothermia can not be readily instituted in a patient crisis demanding decreased metabolic rates.

Clinical warming times may be predicted from the results generated by the methodology presented.[7] The number of kilocalories required to raise the temperature of a patient of mass m, assuming a body specific heat of 0.9 cal/gm • °C, may be calculated using Equation 1.

$$Q = 0.9 \times m \times \Delta T$$

For example, if a 75 Kg patient is on total heart lung bypass at 28°C and an H-1000 is employed in the E.C.C., the warming time may be approximated by locating the blood heat flow on the previous graphs appropriate for the present E.C.C. operating conditions. If the blood flow is 5.0 L/min, the water flow is 15 L/min, and a T_{Wi} - T_{Bi} gradient of 10°C is maintained except when the water temperature begins to exceed 42°C, the H-1000 will deliver 20.9 Kcal/min to the blood. To raise the patient temperature from 28°C to 37°C, Equation 1 yields 607.5 Kcal. At a blood heat flow of 20.9 Kcal/min, it will take 29 minutes to deliver 607.5 Kcal to the patient. The above approximation breaks down considering not all 75Kg of the patient are at 28°C after cooling, due to variations in the distribution of the blood and heat flow with changing cardiac indices and organ vasomotion.

SUMMARY

The following conclusion may be drawn from the methodology presented.

1. Heat exchange device blood heat flow is easily quantitated and reproducible *in vitro*.
2. Blood heat flow is a linear function of T_{Wi} - T_{Bi}.
3. Most heat exchange devices reach 95% of their maximum blood heat flow at a water flow of 12.5 L/min.
4. Blood heat flow increases with increasing blood flow as the Coefficient of Heat Exchange diminishes.
5. Not all of the water heat given up is gained by the blood and as much heat as 15% of the water heat given up may be lost to the environment.
6. Heat exchanger water flows may be predicted at a given wall water pressure.
7. Blood path resistance measurements make the operator aware of the physical operating limitations.
8. Approximate warming times are predictable using the results generated in this method.

An *in vivo* study is the next logical step in a complete heat exchange device evaluation. An *in vivo* comparison must quantitate the various organ temperatures and attempt to control the distribution of blood flow before ranking device warming times.

The authors with to thank Louis S. Toth, James E. Snyder, James P. Dearing, Joseph P. Mandl and Marilyn G. Ratcliff for their technical assistance.

REFERENCES

1. Biot, Maurice A., *Variational Principles in Heat Transfer*. Oxford at Clarendon Press, 1970.
2. Stoever, Herman J., *Applied Heat Transmission* (Conduction), McGraw Hill, 1941.
3. Galletti, P. and Brecher, G. *Heart Lung Bypass Principles and Techniques of Extracorporeal Technology*. Grune and Stratton, 1961.
4. Esmond, William G. *et al.*, Profound Hypothermia with Simplified Equipment: A Disposable Stainless Steel Heat Exchanger of High Efficiency, *J. of Thor. & Cardiovasc. Surg 42*(5): 563-574, 1961.
5. Ahuja, Avtar Singh, Measurement of Thermal Conductivity of Stationary Blood by Unsteady-State Method. *J. Appl. Physiol. 37*(5): 569, 1975.
7. Esmond, William G. *et al.*, Optimization of the Medical Blood Heat Exchanger: Plastic Film Heat Exchanger, *Trans. Amer. Soc. Artif. Int. Organs, IX*, 1963.
8. Peirce, E. Converse, II, A Simplified Heat Exchanger for Perfusion Hypothermia, *Arch. Surg. 84:* 329-333, 1962.

9. Ahuja, Avtar Singh, Heat Transport in Laminar Flow of Erythrocyte Suspensions, *J. Appl. Physiol.* *39*(1), 1975.
10. Grimson, J., *Advanced Fluid Dynamics and Heat Transfer.* McGraw Hill Book Co., 1971.
11. Personal communication with Donn D. Lobdell, Cobe Laboratories, Inc. October, 1976.
12. *Respiration and Circulation*, Philip L. Altman and Dorothy S. Dittmer, Eds. Federation of American Societies for Experimental Biology, 1971.
13. *Handbook of Chemistry and Physics*, 47th Edition, Robert C. Weast, Editor. The Chemical Rubber Co., 1966.