Some Engineering Problems Associated With Open-Heart Surgery

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by

[Signature]

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ABSTRACT

This thesis discusses four problems associated with open-heart surgery; namely:

(a) The design of a level control device to automatically replenish blood in an oxygenator.

(b) The development of a rotating-disc heat exchanger which serves the dual purpose of oxygenating the blood and providing rapid heat transfer.

(c) The design of a temperature-regulation unit for hypothermia techniques.

(d) The design of an efficient heat exchanger to investigate the advantages of the "Drew" technique.
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NOMENCLATURE

Unless otherwise stated, the following nomenclature and units will be used in this thesis.

\( C_p \) -- heat capacity k-cal/kg-°C.
\( d_1 \) -- inner diameter of an annulus (meters)
\( d_2 \) -- outer diameter of an annulus (meters)
\( D \) -- diameter (as specified)(meters)
\( D_e \) -- equivalent diameter of an annulus \((d_2 - d_1)\)
\( f \) -- hydraulic friction factor
\( g \) -- gravitational constant \(= \text{m/hr}^2\)
\( h \) -- film coefficient of heat transfer (subscript b = blood film,
  \( c \) = coolant film)(k-cal/m\(^2\)-hr-°C.)
\( k \) -- thermal conductivity \((\text{k-cal/hr-m-°C.})\)
\( L \) -- length of annulus (meters)
\( P_f \) -- pressure drop due to friction (meters of fluid)
\( Q \) -- rate of heat transfer \((\text{k-cal/hr})\)
\( R_e \) -- Reynolds number \( \frac{D_e V}{\nu} \)
\( t \) -- temperature of fluid to be cooled or heated (subscripts 1 = inlet,
  2 = outlet, b = blood)°C.
\( \Delta t \) -- temperature difference \( t_{\text{sat}} - \frac{(t_1 + t_2)}{2} \) where sat refers to the mean surface temperature.
\( T \) -- temperature of coolant or rewarm fluid (subscripts as above)°C.
\( U \) -- overall coefficient of heat transfer \((\text{k-cal/hr-sq.m-°C.})\)
\( V \) -- velocity of the fluid \((\text{m/hr})\)
\( w \) -- mass flow rate (kg/hr)

\( X \) -- thickness of material (meters)

\( \beta \) -- coefficient of cubical expansion (1/°C.)

\( \nu \) -- dynamic viscosity (subscripts, a – at bulk fluid temperature

\[ \eta_a = \left\{ \frac{t_{ms} + t_1 - t_2/2}{2t_{ms}} \right\} \] at the mean surface temperature.

\( \omega \) -- angular velocity (r.p.m.)

\( \rho \) -- density (kg/m\(^3\))

\( \theta_{m} \) -- log mean temperature difference
I INTRODUCTION

Remarkable advances in surgery of the heart have been made in the last decade. The surgeon today is able to open up the heart, empty it entirely of blood, and under direct vision make unhurried repair of the valves and chambers inside. This has been accomplished despite the fact that only a brief interruption of the normal functions of the heart and lungs exposes to irreversible damage such oxygen-sensitive tissues as the central nervous system or kidneys; that a single emboli of air in the arterial system may prove fatal; that the blood cells themselves are extremely fragile and subject to easy damage.

The key to these medical achievements has been the application of bio-physical and engineering principles to surgery in the development of heart-lung simulators for temporary by-pass of the central circulatory system. In order to work within the heart at normal body temperatures surgeons have had to find a way to circulate the blood outside the body (extracorporeal circulation) to by-pass not only the heart but the lungs as well. Thus, these heart-lung "machines" not only do the pumping for the heart but also provide for the pulmonary exchange of oxygen and carbon dioxide, and supply the heart itself with oxygenated blood during these operations. With these machines surgeons are correcting many formerly hopeless congenital heart defects which condemn to premature death one out of every one hundred children born.

The clinical use of the heart-lung machine is quite recent. Although the first successful operation was reported in 1953, it was not until after 1954 that operations using this apparatus were regularly
successful. Consequently, many divergent techniques and operating principles in "open-heart" surgery are currently being investigated. Some of these techniques will stand the test of time; others will be discarded.

Three extracorporeal circulation techniques which have been used successfully are:

(a) By-passing the heart and lungs while maintaining normal body temperatures (normothermia) and normal blood flow rates.

(b) Lowering the body temperature (hypothermia) during heart-lung by-pass by cooling the blood as it re-enters the body. This lowers the body metabolism and reduces the blood flow rate required.

(c) By-passing only the heart while the body temperature is reduced to very low levels. When the desired temperature is reached, the circulation is completely interrupted while the heart is operated on. This relatively new technique which utilizes the patient's lungs to oxygenate the blood (autogenous oxygenation) is often referred to as the "Drew" technique.

Members of the Department of Surgery at the University of Saskatchewan Hospital have employed normothermia-extracorporeal circulation techniques clinically for some time. Research is presently under way to investigate hypothermia techniques in conjunction with extracorporeal circulation for the repair of congenital and acquired heart defects. After the laboratory phase of this program has been completed, it is intended to investigate the advantages of the "Drew" technique.

The design of equipment for this research presents many new and interesting engineering problems. Many of these problems are readily solved by the direct application of engineering theory and available equipment.
Others, because of their special nature and lack of sufficient design information, require considerable development work before they may be adequately solved. In many cases, the exact specifications for a particular apparatus cannot be given until medical research has more clearly defined the requirements. Some of the engineering problems associated with open-heart surgery discussed in this thesis may be briefly itemised as follows:

(a) The design and construction of a level control system to automatically replenish blood in an oxygenator.

(b) The development of a rotating disc heat exchanger which serves the dual purpose of oxygenating the blood and providing rapid heat transfer.

(c) The design of a temperature regulation unit for hypothermia techniques.

(d) The design of an efficient heat exchanger to investigate the advantages of the "Drew" technique.
II DEVELOPMENTS IN HEART SURGERY

Developments in heart surgery have been almost entirely confined to the twentieth century, with major advances in the last decade. Although the first successful heart surgery was reported in 1896(1), operations within the chest were considered unsafe until the late 1930's. As better antibiotics, anaesthetics, and techniques of blood transfusion made prolonged operations within the chest possible, considerable success was then achieved in correcting those congenital defects which did not necessitate opening the heart.

Operations within the heart itself were not regularly successful until 1948 when a Philadelphia surgeon(2) perfected a technique to dilate a stenosed or constricted heart valve with a finger or instrument. This "blind" or "closed" procedure is successful only in those cases uncomplicated by other abnormalities.

It became increasingly apparent that if complicated defects within the heart were to be repaired the surgeon must be provided with the advantages of direct vision and a dry surgical field. One approach, suggested by W.G. Bigelow of the University of Toronto(3), was to decrease the metabolic requirements of the vital organs by externally cooling the body to 30°C., allowing the circulation to be safely interrupted for brief periods. In 1952, these findings were used by F.J. Lewis of the University of Minnesota, who performed the first successful "open-heart" operation.

This hypothermia technique has been used to correct many simple heart defects, but its application is limited by the fact that the surgeon is restricted to five or six minutes within the heart. The circulation may be interrupted for longer periods at lower body temperatures, but only at
a greater risk of ventricular fibrillation. In this state, the ventricles contract weakly and spasmodically and are virtually ineffective as pumps.

F.H. Gibbon of Philadelphia and other investigators began as early as 1930 to develop a mechanical substitute for the heart. A substitute for the heart alone must pump the blood to the lungs for the pulmonary exchange of oxygen and carbon dioxide and then from the lungs to the other organs of the body. To provide a dry operating field, this double pump must be connected to the inlet and outlet vessels on both sides of the heart. It was evident from animal experiments\(^{(4)}\) that the difficulties of incising and later repairing these vessels, and the obstruction of the surgeon's vision by the many tubes required, made this scheme impractical.

A machine which could substitute for the basic functions of the lungs as well as the heart would require the insertion of only three collecting tubes or cannulas, and one of these could be placed out of the surgical field. Such a machine was developed and successfully used clinically for the first time by Gibbon in 1953\(^{(4)}\). However, further attempts on humans at this time failed, and it appeared that patients suffering from cardiac defects could not tolerate the extra burden of extracorporeal circulation.

New impetus was given to work in this field late in 1954 by reports of successful open-heart operations performed by a team of top-ranking surgeons (E. Warden, R.L. Varco, W.C. Lillehei, and M. Cohen)\(^{(2)}\) of the University of Minnesota. This team used a cross-circulation technique with a healthy individual acting as donor to oxygenate the blood for the patient. They were successful in thirty-four operations, but the method was immediately discarded after the death of one of the donors.
The cross-circulation technique demonstrated the feasibility of heart-lung by-pass and the advantages of direct vision, unhurried, open-heart surgery. Improvements in mechanical heart-lung machines have, since 1954, made this technique the method of choice for most open-heart operations. Many different types of pumps and oxygenators (artificial lungs) have been used successfully. Figure 1 illustrates this technique diagrammatically.

As shown in Figure 1, two catheters are inserted through the wall of the right atrium to tap the superior and inferior vena cava. Blood flows by gravity into a collecting and defoaming vessel. The blood then passes through some form of oxygenator which substitutes for the basic action of the lungs. The oxygenated blood is then pumped through a filter into the femoral artery. Blood flows in the reverse to normal direction in the aorta and then into the many branches of the circulatory system to be collected again by the catheters in the vena cavae. Unless the heart is deliberately stopped it continues to beat although its chambers are empty except for the small amount of blood that returns from the coronary system. This blood is gently aspirated by an auxiliary pump.

Although heart-lung by-pass at normal body temperatures has been used with dramatic success, there are a number of limitations to this technique. Present machines cannot cope with high flow rates without resulting in inadequate oxygenation control, increasing the risk of emboli entering the system or increasing the traumatic damage to blood constituents. Consequently, safe perfusion time is limited. In addition, the heart itself must at all times be assured of an adequate supply of oxygenated blood. The entrance to the coronaries is just above the aortic valve. Thus it is difficult to perfuse the heart during such operations as the
repair of the aortic valve and the correction of abnormalities in the aortic arch. The length of time available for corrective action should the mechanical circulation be accidentally interrupted is extremely limited.

Some of these limitations may be overcome by combining hypothermia and heart-lung by-pass. In 1952, F. Gollan(5) included an efficient heat exchanger in the extracorporeal circuit and experimentally demonstrated the possibility of rapid internal cooling. This method has a number of advantages over external methods of cooling. The vital organs are cooled rapidly, and it is of little consequence if the heart does fibrillate because adequate perfusion is maintained by the external system. Lower temperatures are thus possible. In addition, when rewarming the patient by this method the cold heart is not called upon to supply the increasing metabolic demands of warm peripheral tissues as it is in the case of external rewarming.

Many investigators have reported that for a 10°C drop in body temperature the oxygen requirements of the body decrease by one-half. Since blood also has a greater affinity for oxygen at low temperatures, the blood flow rate in the extracorporeal circuit may be greatly reduced. Many centres are utilizing moderate hypothermia (30°C) during heart-lung by-pass as a standard procedure.

A number of investigators (Drew(6), Gordon(7), Bjork(8)) have recently demonstrated that it is possible to dispense with an oxygenator by what has become known as the "Drew" technique. The extracorporeal circuit for this technique is shown in Figure 2. Perfusion is started by inserting a cannula into the left atrium and the femoral artery. The
Figure 1. Normothermia Extracorporeal Flow Circuit

Figure 2. Autogenous Oxygenation-Deep Hypothermia Extracorporeal Flow Circuit (Drew Technique)
left atrium is drained at a rate of about 100-250 milligrams per minute into a reservoir. The blood is then pumped through a heat exchanger and filter, back into the femoral artery. Oxygenation is carried out by the lungs in the normal manner. When an esophageal temperature of 16-20°C is reached (or the heart has started to fibrillate) this left side perfusion is stopped and cannulas are inserted into each vena cava. A third cannula is inserted into the pulmonary artery through a stab wound in the right ventricular wall. Blood is drained from the vena cavae into a reservoir and pumped back into the pulmonary artery. Both systems are then perfused until the esophageal temperature reaches 10°C. At this temperature the oxygen requirements of the body are approximately 6% of normal. Circulation may then be interrupted completely while corrective surgery is undertaken. After the operation has been performed, great care is taken to remove any air from the system and perfusion of both systems re-started with slow rewarming. When an esophageal temperature of 30°C is reached, the heart is defibrillated and the right side perfusion stopped. Left side perfusion is continued until normal body temperatures are attained.

The advantages claimed for this technique are:

1. There is less blood damage. Bjork reports 19 mg.% free hemoglobin (a measurable indication of mechanical damage) with this system as compared to 83 mg.% using the same type of pump with a rotating-disc oxygenator.

2. The lungs oxygenate the blood much better than any presently available oxygenator. This is borne out by the fact that all referenced investigators report no post-operative acidosis (low pH) with this technique.
3. Surgical procedure is facilitated by providing up to forty-five minutes of operating time on a completely flaccid heart.

4. The margin of safety is increased.

5. No coronary perfusion is required.

One of the disadvantages of this technique is the limited time available without re-perfusion. There is also additional complexity of equipment since the two pumps required must be operated at interdependent rates. The method is also more time-consuming than normothermia methods. However, perhaps the most serious disadvantage is the report by Bjork\(^9\) of cerebral damage in a number of cases. At low temperatures, the number of thrombocytes and white corpuscles decrease but return to normal after rewarming. It has been hypothesized that these cells aggregate at low temperatures and occlude the circulation in certain areas of the brain for some time, resulting in localized brain damage.

Each of the three techniques described above has definite advantages over the others for a particular type of open-heart operation.

Many simple cases can be safely treated without the introduction of hypothermia. Facilities for immediately introducing moderate hypothermia are desirable in the event that unpredicted abnormalities are discovered after the heart is opened. The "Drew" technique shows promise as a method for repair of the aortic valve and some abnormalities of the aortic arch. A modern hospital must, therefore, be equipped to enable the surgeon to employ his method of choice.
III A LEVEL CONTROL SYSTEM TO AUTOMATICALLY REPLENISH BLOOD IN AN
OXYGENATOR

1. Heart-Lung Apparatus Currently Used Clinically at the University
   of Saskatchewan Hospital

   The heart-lung apparatus used clinically at the University of
   Saskatchewan Hospital is illustrated in Figure 3 and the diagrammatic sketch
   of Figure 4. Venous blood flows by gravity and is collected by aspiration
   control vessels (B). The blood then enters a rotating-disc oxygenator.
   This consists of a horizontal pyrex cylinder, 5-1/8 inches in diameter,
   and 5, 9, 13, 21, or 25 inches long (depending upon the estimated oxygen
   requirements of the patient) mounted between stainless steel end plates.
   Thin corrugated stainless steel discs 4.7 inches in diameter and axially
   spaced .2 inches apart are rotated within this chamber. The blood level
   in this chamber is maintained to immerse the discs to a depth of 1-5/8
   inches while a constant flow of oxygen (or mixture of oxygen and carbon
   dioxide) is passed through the upper region. As the discs rotate, a thin
   film of blood clings to each plate, providing a large surface area between
   the blood and oxygen. The surface area exposed per minute may be
   controlled by varying the disc speed by a mechanical variable speed control.
   The temperature of the blood is maintained by a heating cable wrapped around
   the cylinder.

   The mechanical heart of the system is a Sigmanotor pump (F).
   This pump consists of rods made to compress a rubber tube in a sequential
   manner by a rotating cam. If desired, the rubber tube may be so positioned
   that each rod completely compresses it. Thus the pump may be made to
   exhibit positive displacement characteristics and may be precalibrated to
Figure 3
Photograph illustrating the normothermic heart-lung apparatus currently used at the University of Saskatchewan Hospital.

Figure 4
Diagrammatic sketch of apparatus illustrated in Figure 3.
deliver a given flow. The pump forces the blood through a filter and back into the arterial system of the patient.

2. Need for a Level Control

Animal experiments with this apparatus pointed out the need for a control device to maintain blood levels in the oxygenator. During these experiments, blood lost through hemorrhage was replaced by manually removing a mechanical clamp on the connecting tube between the reserve blood supply and the oxygenator. The priming volume of the largest oxygenator is two liters, and flow rates up to seven liters per minute may be realized. If the blood level in the oxygenator drops 5/8 of an inch when operating at these high flow rates, there is a very great risk of bubbles entering the arterial system. In addition, a change in blood level alters the surface area exposed to oxygen, and hence the rate of oxygenation. It was, therefore, necessary for the perfusionist to constantly watch the blood level and act very quickly should a major hemorrhage occur. A simple system for detecting a drop in level and automatically replenishing blood was considered essential before the machine could be used clinically.

A number of level control systems for maintaining the blood level in a rotating-disc oxygenator have been reported. Kantrowitz(10) described a photocell level detecting system with control provided by an auxiliary pump which added or removed blood from the oxygenator as required. A weight detecting system also has been used. A pump speed control system using an electrode level sensing device has been described by Olstead(11). Since it was necessary only to replenish blood lost, these level control systems were considered unduly complex. Figure 3 illustrates the system which was subsequently designed and built for use.
at this University.

This system consists of an electrode level detecting assembly (J), (see Figure 4), an electronic control unit (I), and two solenoid valves (I). The two solenoid valves permit reserve blood to be added manually or automatically at two different rates. When operating automatically, a drop in blood level opens a solenoid valve controlling flow through quarter-inch tygon tubing. This flow is usually sufficient to maintain the blood level within the oxygenator to within one-sixteenth of an inch of normal. If the blood level continues to drop, the second valve also opens, permitting additional blood to be added via half-inch tubing. Should the level continue to drop to the point where there is a risk of bubbles entering the system, the pump is stopped and will not re-start until a safe level is automatically established or the operation switched to manual control.

3. The Electrode Assembly

The electrode assembly designed for this level control system is detailed in Figure 5.

The electrode assembly consists of a stainless steel "level glass" fitting attached to the outlet end of the oxygenator. A five-inch section of one-inch I.D. transparent tygon tubing serves as the "glass". The tubing is slipped over the fitting and is rigidly supported at the top by a bracket. This bracket is graduated to serve as an aid in estimating blood volumes in the oxygenator. Into the tygon tube are inserted five platinum electrodes mounted in a threaded teflon rod. The four outer electrodes are vertically spaced to detect desired levels. The fifth and central electrode extends below the other four to serve as a common ground.
Figure 5
ELECTRODE ASSEMBLY
LEVEL CONTROL
HEART-LUNG APPARATUS
The electrodes are mounted in such a way that moisture condensed during autoclaving will not result in a conducting path between electrodes. The vertical position of the teflon rod and hence the control level may be varied by an adjusting nut. The unit is easily cleaned, and is constructed of materials that are non-toxic.

4. **Electronic Control Unit**

   The four outer electrodes of the assembly are directly connected by cable to the grids of two double triodes. These are represented in the circuit diagram of Figure 6 by terminals 1, 2, 3, and 4. The fifth central electrode (G) is connected to a common ground. With normal blood level in the oxygenator, electrodes 2, 3, and 4 are immersed resulting in a conducting path through the blood to electrode (G). Tubes T₂, T₃, and T₄ are normally conducting, energizing their respective plate relays. Electrode 1 is above the level of the blood; T₁ is therefore held at cut-off by the 15-volt negative bias. If the level drops so that the conducting path between electrode 2 and (G) is broken, T₂ will be cut-off and relay 2ₚ will drop out, putting on pilot light 1ₚ̅₁. If mode switch 4ₕ is on "automatic" solenoid valve V₁ will open, permitting reserve blood to flow by gravity through quarter-inch tubing into the oxygenator. Similarly, if the level drops below electrode 3, a second warning light 2ₚ̅₁ will be put on and solenoid valve V₂ will open. If the blood level drops below electrode 4, relay 5ₚ will be energized, stopping the main pump motor. As soon as flow ceases, the level in the oxygenator immediately rises above electrode 4. To prevent the pump from oscillating on and off, relay 4ₚ is interlocked with relay 3ₚ. The pump will not automatically re-start until the level has risen above electrode 3.
SUPPLY
PILOT LIGHTS
PUMP CONTROL

LEVEL DETECTOR

POWER SUPPLY

Fig. 6

LEVEL CONTROL CIRCUIT HEART-LUNG APPARATUS
or the mode switch changed to manual control.

Electrode 1 provides a "high level" indication and facilities for future "high level" control. Individual manual control of the valves is provided by switches 2S and 3S.

5. **Solenoid Valves**

The automatic control of the addition of reserve blood required some form of valve that was physically small, easily inserted in the flow line, and which would be atraumatic to blood constituents when open. In addition, the parts in direct contact with the blood would have to be easily disassembled to be cleaned and sterilized. The simplest solution appeared to be a device which would automatically pinch or clamp the supply tubing. Tests were subsequently made on the size and type of tubing normally used for this purpose. These indicated that an actuator capable of delivering a four-pound pull at a stroke of three-eighths of an inch would satisfy the requirements.

Air actuators were considered for this purpose. These can be made physically smaller than electrical actuators with equivalent pull and stroke; however, this did not appear to justify the added complexity of an electrically-controlled, air-actuated system.

The electrical actuator designed to meet the requirements outlined above is shown to scale in Figure 7. The actuator is of water-proof construction and can be easily cleaned. One side of the adjustable clamping bar is slotted so that the bar may be twisted out of the way to insert the tubing without contaminating it. Adjustment is provided to accept all commonly-used tubing.
Adjustable clamp
Brass cover plate
Adjustable clamp
Brass cover plate
Tygon tubing
Air-damped armature
Brass bushing
Soft iron
Connecting plug
28,000 turns
#33 AWG

Figure 7
Solenoid Valve
Heart-lung Apparatus
6. Performance of the Level Control System

The control system as described has been used clinically for over a year and one-half, and has consistently prevented the level from dropping more than one-sixteenth of an inch. The initial electrode assembly was fitted with an electrical connector which used bakelite insulation. After repeated autoclaving moisture condensed within this fitting, resulting in a conducting path between electrodes. The electrode assembly described has been completely satisfactory.

The level control system does not provide a method of automatically removing excess blood from the oxygenator.

During an operation flow rates are often checked by clamping the venous inflow to the oxygenator for a fixed interval of time and measuring the rise in level in the aspiration control vessel. If the operator fails to switch the electronic controller to manual, reserve blood will automatically be added to the oxygenator during this interval. When the clamp is released, the level in the oxygenator rises above normal. This would indicate a need for a "high level" control. However, a reliable flow meter is the basic requirement to alleviate the cause.
IV THE DEVELOPMENT OF A ROTATING-DISC HEAT EXCHANGER AND THERMOREGULATION
UNIT FOR COMBINED HYPOTHERMIA AND EXTRACORPOREAL CIRCULATION

1. Requirements of the System

Hypothermia has been successfully employed in various forms as an adjunct to cardiac surgery—by external surface cooling and more recently, by extracorporeal circulation. The combination of hypothermia with extracorporeal circulation reduces the metabolic requirements and blood flow rates, increases the efficiency of oxygenation, and enables longer safe perfusion. It also permits the perfusion of large patients whose flow rates would otherwise exceed the capacity of a given oxygenator at normal body temperature.

A means of controlling the temperature of the blood is also required for normothermia techniques. The pump-oxygenator consists of many large stainless steel, glass, and plastic surfaces which are responsible for the rapid heat loss from the circulating blood. Many methods have been used to compensate for this loss. The method which is finding increasing acceptance because it permits the surgeon to select either hypothermia or normothermia, is the direct addition or removal of heat by an efficient heat exchanger.

The requirements of an ideal thermal regulation system for extracorporeal circulation are: (1) maximum efficiency of heat exchange, (2) no additional impedance to the flow of blood, (3) no additional priming volume, (4) the equipment must not increase the hazard of emboli entering the circulatory system, (5) material in contact with the blood must be non-toxic, (6) precise and simple control of blood temperature, (7) there must be minimal mechanical damage to blood constituents.
2. **Advantages of a Rotating Heat Exchanger**

The efficiency of heat exchange and oxygenation both depend upon a maximum exchange surface per unit of blood flow. Consideration was therefore given to the possibility of performing both these functions in a single unit. The rotating-disc oxygenator has proven a simple and reliable device for oxygenating the blood. A number of heat exchangers have been designed specifically for this. Urschel(12) reported an efficient heat exchanger consisting of a series of one-eighth-inch diameter tubes semi-circularly molded to fit within the pyrex cylinder. A reservoir one-half inch larger diameter than standard was required to accommodate this coil, resulting in considerable increase in priming volume. Circulating coolant under the glass reservoir has also been tried but this does not result in a very efficient heat exchange.

A heat exchanger which took advantage of the large surface area presented by the rotating discs themselves would appear to satisfy all of the requirements of the ideal system listed in the previous section.

3. **The Need for More Design Information**

The cooling discs of a rotating heat exchanger would have to be thicker than the standard discs used to oxygenate the blood. Also, the cooling discs would have to be permanently sealed in place, because it is undesirable to have a gasket seal between the coolant and blood. The minimum spacing would, therefore, be that necessary for easy cleaning. The maximum number of cooling discs per inch of shaft was estimated to be approximately 3.5, whereas the number of convoluted oxygenating discs
per inch of shaft is approximately 5. Thus the oxygenating surface would be reduced by about 1.5 discs for every axial inch of heat exchange incorporated. The minimum number of discs for adequate cooling was thus desired.

The problem of forced convection has long defied mathematical solution. Only for the simplest cases and corresponding simplified assumptions has a mathematical solution been successful. In the case of a rotating heat exchanger a mathematical treatment would be extremely difficult because of the geometrical complications, the complex flow pattern, and the difficulty in assessing the effect of the film of fluid that adheres to each disc as it rotates. Although convective heat transfer in rotating systems is of importance in the thermal design of many types of industrial equipment, relatively little information on heat transfer in such systems is available in the literature, especially for fluids. At the time this project was started, no quantitative information could be found which could be directly applied to this problem.

The equivalent heat exchange surface of other heat exchangers designed for this purpose provided a rough estimate of the number of discs required. On this basis, seventeen discs would be required assuming that heat exchange takes place only at the submerged surface. If, on the

After the tests on the prototype were completed the author became aware of an investigation by Kuo(13) of heat transfer through rotating ducts. The assumptions which prompted the prototype tests were confirmed in this report.
other hand, the effective surface area of a rotating disc could be assumed as the total wetted surface, only seven discs would equal the average surface area provided by other heat exchangers on which data had been published.

The effectiveness of that portion of the disc not submerged in providing heat exchange was open to question. A thin film of blood adheres to each disc as it rotates. This film will be cooled by conduction from the disc, but will be effectively insulated on the side which is in contact with the air. If this layer of blood effectively mixes with the main stream after rotation submerges it again, the effective area of transfer would be increased.

Consider for simplicity a hollow disc of diameter $D$, rotating at 120 r.p.m. and immersed in blood up to its axis of rotation. Assume that the surface of the disc is heated at a constant temperature $T_0$ by an infinite coolant flow through the disc. If one assumes that a film of blood of thickness $X$ and temperature $T_b$ adheres to the disc and is cooled to temperature $T_0$ before it is effectively mixed with the main stream of blood again, the rate of heat transfer from both sides of the disc may be given by:

$$ Q = U \left( \frac{\pi D^2}{4} \right) \left( T_0 - T_b \right) + \omega c_p X \rho \left( \frac{\pi D^2}{4} \right) \left( T_b - T_0 \right) $$

If this disc were totally immersed, the rate of heat transfer would be given by:

$$ Q = \frac{\pi D^2}{2} \left( T_b - T_0 \right) $$

The thickness of the adhering film for equal rate of heat transfer from both systems may be given by equating these two expressions resulting in:

$$ X = \frac{U}{\omega \rho c_p} $$
The overall coefficient of heat transfer \( U \) is a function of the type and thickness of the disc material. It also depends upon the velocity, viscosity, density, specific heat, and thermal conductivity of the coolant on the inside of the disc and on these same parameters of the blood in which the disc is immersed. However, as an approximation, a relatively high overall coefficient of heat transfer for conventional water to water heat exchangers of 1,450 kcal/hr-m\(^{-2}\)-\(^{\circ}\)C. may be assumed. The thickness of the film required on the partially submersed disc to have the same rate of heat transfer as a completely submersed disc based on the above assumptions would be:

\[
X = \frac{1450}{(120 \times 60) \times 0.94 \times 1.06 \times 10^3} = 2 \times 10^{-4} \text{m or } .2 \text{ mm}
\]

Admittedly this is a gross approximation, but it served to indicate that considerable heat transfer might be realized from the adhering film. The faster the speed of rotation, the greater would be the film thickness, since within the range of interest the centrifugal force could be considered negligible. However, at greater disc speeds, the film has less time to cool before re-entry. The only tangible information that this form of analysis gave, therefore, was that the number of cooling discs required would be between seven and seventeen.

The assumption that even seventeen discs would provide adequate cooling would only be valid if adequate coolant flow could be maintained through each disc. This entails passing coolant through a rotating coupling, circulating it effectively through each disc, and collecting it via another rotating coupling. To have the coolant pass through each disc in series would restrict the flow and result in inefficient heat exchange. It would be desirable to have the coolant flow in parallel
paths through each disc. However, this requires two headers (one for inflow and one for outflow) common to each disc. These headers must be designed so as to cause no additional turbulence as the discs rotate.

The material used in the construction of a heat exchanger which would be in contact with the blood must be non-toxic. Nickel-plated materials may be used. However, apparatus constructed of brass or other easily-worked and platable material is also easily damaged in handling. Stainless steel, therefore, appeared to be the only choice. This presented many fabrication problems which would have to be solved before a clinically usable, rotating-disc heat exchanger could be constructed.

In view of the uncertainty of design information and the technical problems involved in construction and operation, it was decided to construct a prototype model of easily fabricated material to test the workability and efficiency of the proposed design.

4. The Prototype Rotating Heat Exchanger

The basic idea conceived for providing adequate coolant flow through each disc is illustrated in simplified form in Figure 8.

Cooling discs were fabricated by milling the flow pattern illustrated in Figure 9, 3/32 of an inch deep in brass plates, 1/8 of an inch thick. The pattern was designed primarily for ease of milling and minimum resistance to coolant flow. A brass plate 1/32 of an inch thick was then silver soldered to the milled plate. Collars were added and the assembly machined to form the completed disc illustrated in Figure 10. The central shaft illustrated in Figure 11 was machined. Gaskets and spacers of various lengths were made so that from one to eleven discs could be mounted with various spacings on the shaft.
Figure 8

Principle of Rotating-disc Heat Exchanger
Figure 9
Photograph of coolant flow pattern milled in brass plate.

Figure 10
Photograph of completed cooling discs for prototype heat exchanger.
Figure 11
Milled central shaft for prototype heat exchanger.

Figure 12
Central shaft for prototype heat exchanger modified to be used in clinical oxygenator.
5. Procedure for Testing the Prototype Model

Figure 13 shows a schematic of the apparatus used to test the prototype heat exchanger. A supply of water at a constant head and approximately 37°C was maintained by overflowing tap water in a large container. The flow of this water into a simulated oxygenator was regulated with a clamp on the inlet tubing. A clamp on the outlet tubing was adjusted to maintain normal levels within the oxygenator. A constant flow of ice water coolant was pumped through the heat exchanger via two commercial rotating couplings. Mercury thermometers were inserted in the flow stream to measure the inlet and outlet temperature of the coolant and of the water flowing through the oxygenator. Flow rates were determined using a calibrated vessel and a stop watch.

Tests were made to determine the relative effect of the following factors on the rate of heat transfer:

(a) number of cooling discs
(b) rate of flow of water through the oxygenator
(c) speed of disc rotation
(d) number of oxygenating discs
(e) rate of coolant flow

No attempt was made to determine heat transfer coefficients or to set up equations relating these factors. The object of the tests was to obtain relative design information only.

* The author wishes to express his appreciation to the Prairie Regional Laboratories of the National Research Council for the use of these couplings.*
Schematic of Apparatus used to test
Prototype Rotating - Disc Heat Exchanger

Figure 13
The efficiency of a heat exchanger as defined by Dodge (14) was selected as the standard of comparison for these tests, since other heat exchangers designed for hypothermia have been compared on this basis. This definition states that the efficiency of a heat exchanger is the ratio of the rate at which heat is actually removed from a fluid to the rate at which it might be removed by a heat exchanger with infinite cooling surface area and infinite coolant flow. This may be expressed in equation form as:

\[
\text{Heat exchanger efficiency} = \frac{w_p (t_1 - t_2)}{w_p (t_1 - T_1)} \quad \text{or}
\]

\[
= \frac{t_1 - t_2}{t_1 - T_1} \quad \text{which is the temperature change of the fluid per degree inlet temperature difference.}
\]

Although it is known that the heat exchange properties of blood differ from those of water, water was used in these tests to permit comparison with similar data published on other heat exchangers.

6. Results of Tests

Figure 14 shows a plot of efficiency versus simulated blood flow rate through the oxygenator for a heat exchanger with 0, 2, 4, 6, 8, 10 and 11 cooling discs and 58 oxygenating discs. Also shown in this figure are the maximum recommended flow rates for each size oxygenator. The information contained in this graph is replotted in Figure 15 to show more clearly the effect of varying the number of cooling discs.

The change in heat exchanger efficiency as a function of disc speed is shown in Figure 16. Eleven cooling discs only, made up the heat exchanger for this test. The simulated blood and coolant flow rates were
Efficiency vs Flow Rate
for Prototype Rotating Disc Heat Exchanger

Figure No. 14
MAX. RECOMMENDED FLOW RATE FOR VARIOUS SIZES OXYGENATORS.

NO. OXYGENATING DISCS 98
DISC SPEED 120 RPM
FLOW RATE NOTED
COOLANT FLOW 4 GAL/MIN

EFFICIENCY vs NO. COOLING DISCS
FOR PROTOTYPE ROTATING DISC HEAT EXCHANGER

FLOW RATE 3 LITRES/MIN
COOLANT FLOW 1.3 GAL/MIN
NO. COOLING DISCS 11
NO. OXYGENATING DISCS 0

EFFICIENCY vs DISC SPEED
FOR PROTOTYPE ROTATING DISC HEAT EXCHANGER
three and six liters per minute respectively. The efficiency of the heat exchanger increased from .16 at zero disc speed to .39 at 180 r.p.m. No attempt was made to ascertain if this increase was due to a thicker film adhering to the discs or simply due to better mixing of the main flow. The efficiency varied approximately 6% over the normal operating range from 100 to 140 r.p.m.

Figure 17 shows the effect of adding standard oxygenating discs to an eleven-cooling-disc heat exchanger. A heat exchanger with ten cooling discs and fifty-eight oxygenating discs provided approximately the same heat exchange as an exchanger with eleven cooling discs only. Very little increase in efficiency was obtained by increasing the number of oxygenating discs from twenty-one to fifty-eight. The maximum number of oxygenating discs that could be placed on the prototype model was fifty-eight. The limited number of tests, however, indicated that very little increase in efficiency would be obtained from the additional sixty-five that would normally be used in the twenty-five-inch oxygenator.

Also shown on this graph is the effect of increasing the coolant flow rate from 1.3 to 4 gallons per minute. When the coolant flow rate was reduced to 1.3 gallons per minute the efficiency of the eleven-cooling-disc heat exchanger dropped to approximately that of a mix-cooling-disc exchanger with a coolant flow rate of four gallons per minute. Coolant flow rates above four gallons per minute were not obtainable with the pump used for the tests. However, the inlet and outlet coolant temperatures differed by less than two degrees at this flow rate. Thus very little additional efficiency could be expected by increasing the coolant flow rate above this figure.
VARIABLES

<table>
<thead>
<tr>
<th>NO. OXYGENATING DISCS</th>
<th>CURVE</th>
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<tbody>
<tr>
<td>58</td>
<td>1</td>
</tr>
<tr>
<td>21</td>
<td>2</td>
</tr>
<tr>
<td>0</td>
<td>3</td>
</tr>
<tr>
<td>0</td>
<td>4</td>
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<table>
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<tr>
<th>DISC SPEED (RPM)</th>
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<tr>
<td>120</td>
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<td>120</td>
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<tr>
<td>120</td>
<td>3</td>
</tr>
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<td>120</td>
<td>4</td>
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</table>

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<th>COOLANT FLOW (GPM)</th>
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</tr>
</thead>
<tbody>
<tr>
<td>4</td>
<td>1</td>
</tr>
<tr>
<td>4</td>
<td>2</td>
</tr>
<tr>
<td>4</td>
<td>3</td>
</tr>
<tr>
<td>1.3</td>
<td>4</td>
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</tbody>
</table>

<table>
<thead>
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<th>NO. COOLING DISCS.</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>11</td>
<td>1</td>
</tr>
<tr>
<td>11</td>
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<tr>
<td>11</td>
<td>3</td>
</tr>
<tr>
<td>11</td>
<td>4</td>
</tr>
</tbody>
</table>

EFFICIENCY VS FLOW RATE
FOR PROTOTYPE ROTATING DISC HEAT EXCHANGER

FIGURE NO. 17
A comparison of the various heat exchangers is made in Figure 13. It will be noted that the eleven-cooling-disc prototype with fifty-eight oxygenating discs indicated an efficiency somewhat greater than two commercially available heat exchangers connected in series.

In addition to the tests described above, three hypothermia experiments were conducted on animals to further evaluate the workability of the unit. For these experiments the prototype was modified so that a six-disc unit could be used without changing the clinical oxygenator. Also, simple rotating couplings were designed to be evaluated during these tests.

The results of these tests indicated that satisfactory cooling rates could be obtained. The increased width of the cooling discs did not appear to generate any more bubbles than the standard oxygenating discs. The rotating couplings proved to be completely satisfactory. It was, therefore, decided to proceed with the design and construction of a clinically-usable, rotating heat exchanger.

7. The Design of a Clinically-Usable, Rotating Heat Exchanger

On the basis of the preceding tests it was decided to construct a seven-cooling-disc heat exchanger for the nine- and thirteen-inch oxygenators and a thirteen-cooling-disc unit for the larger sizes. Tabulated in Table I are the predicted heat exchanger efficiencies for these combinations. Also shown is the estimated decrease in oxygenating efficiency based on the decreased number of discs. This estimate also takes into account a report by Mendelsohn¹⁵ that convoluted discs are 30% more efficient than flat discs in oxygenating the blood. Four methods for compensating for the decrease in oxygenating efficiency are listed in Table II.
CURVE NO.
1. BROWN-HARRISON (TUBULAR (SINGLE))
2. DR. ESMONDS (DISPOSABLE)
3. 7 DISC STAINLESS STEEL
4. NMRI (TUBULAR)
5. 2 BROWN HARRISON IN SERIES
6. 11 DISC BRASS PROTOTYPE

WATER (SIMULATED BLOOD) FLOW RATE - LITERS/MIN.
* ACTUAL BLOOD USED

COMPARISON OF VARIOUS HEAT EXCHANGERS

FIGURE NO. 18
TABLE I
PREDICTED HEAT EXCHANGER EFFICIENCY AND PER CENT DECREASE IN OXYGENATION EFFICIENCY FOR VARIOUS SIZE OXYGENATORS

<table>
<thead>
<tr>
<th>Oxygenator size</th>
<th>9&quot;</th>
<th>13&quot;</th>
<th>21&quot;</th>
<th>25&quot;</th>
</tr>
</thead>
<tbody>
<tr>
<td>Number of cooling discs</td>
<td>7</td>
<td>7</td>
<td>13</td>
<td>13</td>
</tr>
<tr>
<td>Approximate number of convoluted oxygenating discs at standard spacing (.2&quot;)</td>
<td>33</td>
<td>52</td>
<td>82</td>
<td>102</td>
</tr>
<tr>
<td>Predicted heat exchanger efficiency/maximum recommended flow (water used as simulated blood)</td>
<td>.70</td>
<td>.60</td>
<td>.52</td>
<td>.42</td>
</tr>
<tr>
<td>Estimated decrease in oxygenating efficiency</td>
<td>11.4%</td>
<td>7.8%</td>
<td>9.6%</td>
<td>8.0%</td>
</tr>
</tbody>
</table>

TABLE II
METHODS OF COMPENSATING FOR DECREASE IN OXYGENATING EFFICIENCY

<table>
<thead>
<tr>
<th>Oxygenator size</th>
<th>9&quot;</th>
<th>13&quot;</th>
<th>21&quot;</th>
<th>25&quot;</th>
</tr>
</thead>
<tbody>
<tr>
<td>1) Decrease maximum recommended flow rate</td>
<td>1.0 L/min</td>
<td>1.6 L/min</td>
<td>3.0 L/min</td>
<td>4.5 L/min</td>
</tr>
<tr>
<td>from-</td>
<td>900</td>
<td>15</td>
<td>2.7</td>
<td>4.1</td>
</tr>
<tr>
<td>to-</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>2) When operating within the range of flows stated in 1) increase disc speed from 120 r.p.m. to</td>
<td>133</td>
<td>129</td>
<td>131</td>
<td>130</td>
</tr>
<tr>
<td>3) Decrease spacing of oxygenating discs from .2&quot; to</td>
<td>.185&quot;</td>
<td>.195&quot;</td>
<td>.183&quot;</td>
<td>.192&quot;</td>
</tr>
<tr>
<td>4) Increase the length of the oxygenator to</td>
<td>10&quot;</td>
<td>14&quot;</td>
<td>23&quot;</td>
<td>27&quot;</td>
</tr>
<tr>
<td>Net increase in priming volume of 4)</td>
<td>33 ml.</td>
<td>33 ml.</td>
<td>83 ml.</td>
<td>83 ml.</td>
</tr>
</tbody>
</table>
It should be pointed out that the decrease in oxygenating surface is of significance only between the range of flows stated in 1) of Table II. That is, the thirteen-inch oxygenator equipped with convoluted oxygenating discs would normally be used when the predicted flow rate was between 1000 and 1600 ml/min. The thirteen-inch unit with a seven-cooling-disc heat exchanger incorporated would be used when the predicted flow was between 900 and 1500 ml/min. A more serious disadvantage is the decreased oxygenating capacity of the twenty-five-inch oxygenator.

Disc speeds up to 140 r.p.m. have been used clinically for short periods of time to increase the oxygen saturation of the perfused blood. However, speeds over 120 r.p.m. are not recommended because of increased foaming, particularly at high blood-flow rates. Since cooling decreases the oxygen requirements of the body by approximately one-half for every\footnote{10^\circ{c}} drop in temperature and since blood has a greater affinity for oxygen at lower temperatures, it was considered that this method of compensation could be used for the short period of time the patient was on by-pass at normal temperatures.

A more desirable method of compensating for the decrease in oxygenation surface, caused by the cooling discs, would be to decrease the spacing of the oxygenating discs to \(1.15\) in. It has been reported that disc spacing may be reduced to \(1.145\) in without adverse effects.

The length of the oxygenator may be increased to accommodate more discs only with the accompanying disadvantage of increasing the priming volume. However, the additional priming volume necessary is considerably less than that required by other heat exchangers with
equivalent heat exchange (e.g. 175 ml for a single Brown-Harrison).

It thus appeared that the disadvantage of decreased oxygenating surface would not be a major one.

8. The Construction of a Stainless Steel, Rotating-disc Heat Exchanger

A number of problems were encountered in constructing a satisfactory cooling disc of stainless steel. It would be impractical to mill the coolant flow pattern in stainless steel as was so easily done in constructing the brass prototype. The simplest method of fabricating the discs would have been to stamp the pattern in thin sheets and resistance weld the plates so formed together. Unfortunately, facilities for doing this were not locally available. Attempts to solder deflecting fins and thin plates together were initially unsuccessful due to the shifting of the fins and warping of the plates. An inert gas furnace was not available. As a result corrosion and oxidation of the stainless steel during the soldering process became a major problem. Figures 19 and 20 illustrate some of the early attempts at forming a stainless steel disc.

A method whereby corrosion-free, unwarped, cooling discs could be simply produced was developed. Figure 21 shows two formed plates and the equipment used to form them. An 11/16 inch hole was punched in a sheet of #28 gauge, type 304 stainless steel. The sheet was then clamped between the face plate and a special saddle inserted in the center mount of a lathe as shown in Figure 22. A tapered shaft more clearly shown in Figure 23 was drawn through the hole to form an inner lip. The sheet was then trimmed and the outer edge rolled as illustrated in Figure 24. Deflection fins of 1/16 inch stainless steel wire were soldered into a single unit using a high temperature silver solder. The deflecting fins and the
Figure 19
Illustration of fin movement during soldering.

Figure 20
Corrosion and oxidation of stainless steel cooling discs during soldering.
Figure 21
Equipment designed to form stainless steel plates, and two completed plates.

Figure 22
Forming central lip of plates.
Figure 23
Tapered shaft in position in lathe to form central lip.

Figure 24
Rolling outer lip of stainless steel plate.
inner surfaces of both plates were then completely coated with regular soldering flux. Silver solder wire was formed around the fins and outer seam as shown in Figure 25. To prevent corrosion it was necessary to completely coat the outside surface with a special high-temperature flux. Figure 26 shows the assembly ready for the electric oven.

Very rigid discs were obtained in this manner. The discs were pressure tested up to 60 psi without visible signs of failure. Tested with twice normal operating pressure of 20 psi, .020 inches of non-permanent deflection was noted.

The oxygenator bearings and convoluted discs were modified to accept a 5/8 inch hollow coolant shaft. Each cooling disc was individually soldered to this shaft using a low temperature, non-toxic solder. A .185-inch spacer placed over the inner lips of each disc formed a secure, overlapping joint between discs. The completed seven-disc heat exchanger is shown in Figures 27 and 28.

The rotating couplings simply consist of "O" ring seals placed on both sides of a brass housing. Coolant is passed into the housing and through a number of holes drilled into the shaft of the heat exchanger. Each coupling requires approximately twenty-seven oz.-in. of torque unlubricated. The "O" rings are normally coolant lubricated or may be lubricated with petroleum jelly if desired. These couplings need not be autoclavable since they do not come in contact with the blood.
Figure 25
Assembling of plates with solder wire in place.

Figure 26
Assembly ready for soldering.
Figure 27
Completed seven-disc rotating heat exchanger.

Figure 26
Completed seven-disc rotating heat exchanger.
V THE DESIGN OF A THERMOREGULATION UNIT FOR HYPOTHERMIA

1. Guiding Specifications

A means of controlling the temperature of the coolant passing through the rotating heat exchanger was required. The precise specifications of the type of control required could not be given until considerable experimental work had been carried out on animals. Tentative guiding specifications could be stated as follows:

a) The unit should have sufficient cooling capacity to lower the esophageal temperature of a 100 kg. man from 37.5°C to 10°C as rapidly as possible.

b) The coolant temperature must not exceed the range of from 0°C to 44°C.

c) The unit should provide for rapid rewarming.

d) The temperature of the esophagus should be controlled within 1/2°C over the range of from 37.5°C to 10°C.

e) The unit should be as small and compact as possible so as to take up a minimum of floor space.

f) Facilities should be provided for monitoring esophageal, rectal, arterial blood, and coolant temperatures.

g) Consideration should be given to making the unit explosion-proof, although this requirement was not considered a major one since non-explosive anaesthetics are normally used for this type of operation.

2. Determination of Cooling and Rewarm Capacity Required

When assigned the problem to design equipment to lower the
temperature of a 100 kg. man to 10°C., the first reaction was to
calculate the body heat that must be removed if the total body mass is
reduced to this temperature. This may be calculated to be approximately
3000 k-cal. To remove this amount of heat in half an hour would require
at least a 2 H.P. refrigeration unit. Fortunately, however, when the
body is cooled by cooling the blood, only the so-called "core zone" comprised
essentially of the vital organs are appreciably cooled. For example, when
the esophageal temperature is lowered to 10°C., the brain and heart will
be at approximately the same temperature, but the thigh muscle may be as
high as 20°C.

The capacity of the cooling unit was determined by calculating
the total body heat removed when the esophageal temperature was lowered
to 10°C. Based upon published results of other investigators,(16) this
was calculated to be roughly 1/3 of that which must be removed if the
total body mass were reduced to that temperature. Thus, to cool a 100 kg.
man so that his esophageal temperature is 10°C., approximately 1000 k-cal.
must be removed. If a high rate of cooling is required, this still entails
a fairly large refrigeration unit.

Since it would be undesirable to have the coolant temperature
go below the freezing temperature of blood, ice water was selected as
the coolant medium. Crushed ice affords maximum cooling capacity with
minimum space and expense. Assuming that 1000 k-cal. are to be removed
from the patient and allowing a 1.5 safety factor for other heat losses,
storage facilities for approximately 20 kgs. of crushed ice must be provided.
This requires only about one cubic foot of storage volume.

A simple means of rewarming would be to circulate tap water
through the heat exchanger. However, since pumping facilities would be required for the cooling process, electric heaters would provide a completely self-contained unit.

It has been shown (12) that the maximum temperature of the rewarmed fluid should not exceed 44°C. Thus assuming that it is desired to rewarmed a patient from 10°C as quickly as possible, using a heat exchanger with an efficiency of .4, heat must be supplied at a rate of 4750 watts if the blood flow rate is 5000 ml/min. This cannot be supplied from a standard 110-volt outlet.

The rate at which heat must be supplied will decrease in some logarithmic manner because of the decreasing temperature gradient between the blood and coolant. This function cannot be determined since it depends upon the heat exchange efficiency of the body. For a first approximation, it was assumed that the rate at which heat must be supplied decreases linearly with time until the blood is returned from the body at normal body temperature. The rate at which heat must be supplied at this time will be approximately 900 watts. If one further assumes that at this time the total heat removed from the body during the cooling process has been returned, the time required for rewarming may be calculated to be twenty-five minutes. This figure is optimistic, but using this approximated power curve one can estimate the result of using smaller heaters with an initial come-down capacity. The combination finally selected was a 2000-watt heater with an initial come-down capacity of 20 liters of water heated to 44°C. With this combination it may be shown that the rewarn fluid temperature will not drop below 30°C. Thus little advantage would be gained from additional come-down capacity heater output.
3. **Temperature Control Apparatus**

There has been considerable debate over the desirability of fast rewarming. Blood has a greater affinity for oxygen at low temperatures. Some authorities hypothesized that because of this there may be danger of emboli forming in the arterial system during the rewarming process. Edmark\(^{(17)}\) and other have reported the occurrence of "rewarming shock" and difficulty in maintaining normal pH during fast rewarm. Other investigators\(^{(12)}\) have reported successful experiments with fast rewarming. Thus facilities for both fast and controlled rewarming would be desirable for experimental trial.

There have been no reports of any adverse effects due to fast cooling, nor due to the resulting large temperature gradients within the body. Controlled rates of cooling, therefore, did seem necessary. There are many forms of automatic temperature control that may prove essential for this application. These may be the control of rates of cooling and rewarming; of maximum temperature gradient between blood and coolant or between the esophagus and the arterial blood. It was, therefore, decided to provide only the essential automatic safety controls. Manual controls would be provided whereby the operator could control the rate of cooling and rewarming if this proved necessary.

A schematic of the thermoregulation unit and its associated control circuit is shown in Figure 29.

The unit is equipped with an ice storage tank with approximately 20 kg. crushed ice capacity for rapid cooling. A rewarm tank
REWARM VESSEL.

COOLING VESSEL

CONTROL VESSEL

HEAT EXCHANGER

REWARM VESSEL 20 LITRE CAPACITY
COOLING VESSEL 20 KG. CRUSHED ICE CAPACITY
CONTROL VESSEL 4 LITRE CAPACITY

LEGEND

Ti & Ts — 415A. Honeywell Temperature Controller
T2 — Fenwall Thermistor Controller
Pw & Pc — Little Giant Type 4-12R Pumps
H1 & H2 — 100 W. 115 V. Heaters
V1, V2 & V4 — Sporlan Type 73 Solenoid Valves
V3 — “ ” Type 12 “ ”
V5 — 3/8 Gate Valve
PRV — Pressure Relief Valve 0-15 PSI

Schematic of Thermoregulation Unit and Control Circuit for Hypothermia Apparatus

Fig. 20
is maintained at the desired rewarm fluid temperature by thermostat T₁ which controls a 1000-watt heater. A third control vessel of about four liters capacity is equipped with a control thermostat, a safety thermostat, and a second 1000-watt heater.

The mode switch MS is placed in position 1 for normothermia operation. In this position, thermostat T₂ controls the 1000-watt heater inserted in the control vessel. Coolant is circulated by pump P₆ through the control vessel, the heat exchanger, and solenoid V₁.

To rapidly cool the blood, the mode switch is placed in position 2. The coolant is then circulated by pump P₆ through the crushed ice via valve V₂ (and also through valve V₃ if thermostat T₂ has been adjusted low enough). The coolant may be maintained at a desired low temperature by adjusting thermostat T₂ to that temperature and switching the mode switch to position 3. In this position, the thermostat controls V₂ to add ice water at a rate preset by the manual valve V₅. By-pass valve V₁ is constantly open in this position and pump P₆ maintains a high rate of coolant flow through the heat exchanger.

Rewarming is initiated by placing the mode switch to position 1 or 4. In position 1, the 1000-watt control heater will reheat the rewarm fluid to the temperature controlled by thermostat T₂. In position 4, fast rewarming is provided by pump P₆ circulating the rewarm fluid through the rewarm tank via valve V₄. If thermostat T₂ is set to the maximum rewarm temperature, both heaters will function if the rewarm fluid temperature drops below this value.

Slower rates of rewarming than that provided by the 1000-watt
heater operating continuously may be manually obtained by adjusting thermostat $T_2$ in steps. Similarly, faster rates can be induced by switching between positions 1 and 4. This has proven quite satisfactory in experiments, however, stepless control could easily be provided at a later date if this proves desirable.

The response requirements of the temperature controller $T_2$ were determined in the following manner. The 1000-watt heater provides approximately 14.3 k-cal/min. The volume of coolant involved is approximately 4 kgs. Assuming negligible heat loss for the worst condition, the temperature of the water is increasing at a rate of .06°C/sec. Again, although this function is not linear, over the limited range of interest one may approximate it as such. The temperature probe of the controller will indicate a temperature which will lag the temperature of the coolant depending upon the time constant of the probe and the rate at which the temperature is increasing. The time constant of temperature probes are given by the manufacturer as the time required for a probe to read 63.2% of a step function change in temperature. The assumed ramp function change in coolant temperature may be approximated by a series of step functions. At the end of each step interval, the indicated temperature of the probe will have increased by:

$$\Delta T_1 = T_{bs} (1-e^{-\frac{t}{T_p}})$$

where:

$T_{bs}$ is the temperature difference between the actual temperature of the fluid and the temperature indicated by the probe,

$t$ is the step interval in seconds,
and $T_p$ is the time constant of the temperature probe in seconds.

Curves similar to those plotted in Figure 30 may be constructed if the time constant of the probe is known. From these curves the differential of the controller may be specified, depending upon the degree of control desired. Thus, using a 1000-watt heater and a temperature probe with a time constant of 3.4 seconds, a controller with a differential of $0.3^\circ C$ would be required to control the coolant temperature to within $0.5^\circ C$; a controller with a differential of $0.7^\circ C$ would control to $1^\circ C$.

The same temperature controller and probe are used to automatically maintain the $30^\circ$ temperature necessary for moderate hypothermia. Although the blood temperature is above room temperature during these operations, more heat is added by the metabolic functions of the body than is dissipated in the external circuit. Therefore, ice-water must be periodically added to decrease the coolant temperature circuit. The coolant system is purged in twenty seconds; a temperature probe with a 3.4-second time constant will register a step function change in temperature in approximately seventeen seconds. It is only necessary then to restrict the change in coolant temperature in accordance with the differential of the controller selected and the degree of control desired.

To ensure control of the esophageal temperature within $0.5^\circ C$, a probe with a 3.4-second time constant and a controller with a differential of $0.5^\circ C$ would be required. As indicated by Figure 30, this would
1 - Fluid temperature
2 - Reading of Probe with 3.4 sec time constant
3 - Reading of Probe with 7 sec time constant

To Control to 1° C
Minimum Controller Differential Req'd

Temperature probe characteristics for a linear increase in temperature.
readily control the temperature of the coolant to within .5°C. when the
mode switch (MS) is in position 1 during normothermia. To maintain
hypothermia (mode switch in position 3), manual control valve V₃ (Figure
29) is preset so that when V₃ automatically opens, the temperature of
the coolant entering P₀ will be no less than 29.5°C. when the coolant
entering V₁ is at 30°C.

Since the form of temperature control required could not be
exactly specified, only two of the three temperature controllers of
the tentative design were purchased. These were T₁ and T₂, the
specifications of which were not critical. The controller purchased
for eventual installation as T₂ was installed as T₂ to test the work-
ability of the proposed design. This controller is a Honeywell vapor-
tension type with a tested differential of .5°F. and a relatively long
time constant. Tests indicated, however, that the controller and probes
meeting the specifications as determined above should provide the
desired degree of control.

4. **Use of the Apparatus in Explosive Atmospheres**

A great deal of consideration was given to the possibility
of designing the unit so that it would be C.S.A. approved for use
where explosive anaesthetics are being administered. The first design
was laid out so that all equipment which was not explosion-proof was
located above the approved five-foot level. This required the operator
to change from a sitting position (which is necessary to watch the blood
levels and flow rates) to a standing position to control the electrical
apparatus. In addition, this design was not acceptable because of its
relatively cumbersome appearance.
The present unit is not intended to be used with explosive anaesthetics and will bear a warning clearly indicating this. The use of these gases is precluded for open-heart surgery by the use of such non-explosive apparatus as electrocautery. It is intended to equip the unit with a continuously operated fan which will draw air from the five-foot level through a snorkel tube to pressurize the equipment housing. The fan will be started while the unit is outside the operating theatre. This will provide reasonably safe, though not C.S.A. approved, operation in the event that explosive anaesthetics are deemed necessary in some emergency. This method has been approved in the United States. (18)
VI PERFORMANCE OF THE ROTATING-DISC HEAT EXCHANGER AND THERMOREGULATION UNIT

1. Experimental Results

The unit described in this section is shown in Figures 31, 32, and 33. Figure 33 shows the unit during an experimental run. The control unit on top of the storage tank will be recognized as the control unit for the normothermia apparatus. Provision has been made to house this equipment on the control panel of the hypothermia unit, thus making it completely self-contained.

The efficiency of the heat exchanger using water as simulated blood is shown in Figure 18. The efficiency of the unit is approximately the same as that of the six-disc prototype. The reason for this apparent decrease in efficiency is probably due to the restricted coolant flow through the heat exchanger couplings. With the pumps selected for the thermoregulation unit the rate of coolant flow through the seven-disc heat exchanger is approximately 12 kg/minute as compared to 18 kg/minute obtained during the tests on the prototype.

The thermoregulation unit and stainless steel heat exchanger have been used in twelve hypothermia experiments on animals. Figure 34 shows a plot of esophageal, rectal, inlet blood, and coolant temperatures during a typical experiment in which fast rewarming was used. None of the three dogs which were rewarmed quickly, survived. Considerable difficulty was experienced in defibrillating these animals. As a result, in subsequent tests an attempt was made to control the rate of esophageal rewarming to .6°C. rise per minute. No difficulty was experienced in accomplishing this with manual control, as is illustrated in Figure 35. All of the dogs
Figure 31
Thermoregulation unit.

Figure 32
Thermoregulation unit.
Figure 23
Color photo of hypothermia unit in use in laboratory.
Temperature, pH and blood flow rate during hypothermia experiment with fast rewarming.

![Graph showing temperature, pH, and blood flow rate during hypothermia experiment with fast rewarming.](image)

**Figure 34**

Deep Hypothermia Test 1-1

Dog #975 wt. 20.9 kg.

- Esophageal temp. x-x
- Arterial Blood temp.
- Rectal temp.
- Coolant temp.

Perfusion Time (minutes)
As in Figure 34, with controlled rate of rewarming

Flow rate ml/min.

Arterial blood pH

Perfusion time (minutes)

- Deep Hypothermia Test 1-4
- Dog #1049 wt. 22 kg.
- Esophageal temp.
- Arterial blood temp.
- Rectal temp.
- Coolant temp.
rewarmed slowly were defibrillated easily.

Figure 36 shows the great variability in the time required to cool animals of approximately the same weight to various esophageal temperatures. Although the blood flow varied from 1000 to 1500 ml/min, the variability in rates of cooling could not be explained on this basis.

Also shown in Figure 36 is the average rate of cooling for these tests. This compares very favorably with published results of similar experiments with other heat exchangers.

Figure 37 shows an experiment using moderate hypothermia (30°C). This was manually achieved by switching the coolant pump on as required.

The apparatus in general has operated satisfactorily in all experiments.
Average rate of cooling with rotating seven-disc heat exchanger.

Curve 1 - Time required to cool with 7-disc-rotating heat exchanger
Weight of dogs 19-22.3 kg. (Avge. 21.5 kg.)
Blood flow rate 1.0 - 1.5 L/m

Curve 2 - Average cooling rate for series of tests

Figure 36
As in Figure 34, with moderate hypothermia.

- **Moderate Hypothermia**
- **Test 2-1**

Dog #1161   wt. 20.9kg.

Esophageal temp.  •••
Arterial blood temp.  ••

Perfusion time (minutes)
VII IMPROVEMENTS IN DESIGN PROPOSED FOR THE THIRTEEN-DISC HEAT EXCHANGER

1. Improved Coolant Flow Patterns Through the Discs

Although the efficiency of the six-disc heat exchanger compares favorably with other reported exchangers, considerable time is required to cool to low temperatures. The fact that the outlet blood temperature does not approach that of the coolant would indicate that an even more efficient heat exchanger might be possible. A re-evaluation was made of the design with the object of improving the efficiency of the proposed thirteen-disc unit.

The efficiency of the exchanger can only be increased by decreasing the resistance to heat flow. This may be done by:

a) increasing the cross-sectional area in the direction of heat flow,

b) decreasing the thickness of the metal, or

c) decreasing the film resistance of the blood or coolant.

The characteristics of the blood film heat resistance are fixed by the physical properties of the blood and by the flow pattern through the oxygenator. Further study of the coolant flow pattern, however, revealed that considerable improvement in the heat transfer coefficient between coolant and metal might be made. The heat transfer coefficient can be increased by increasing the effective velocity of the coolant as it passes through the discs. Figure 38 shows the flow baffle design used in the seven-disc exchanger on the left and the proposed design for the thirteen-disc unit on the right. The probable flow pattern for each disc is also indicated. The area dotted in the figure on the left
would be a region of very low flow. The baffle design on the right provides a much better flow pattern with high velocity flow maintained in that portion of the disc where heat transfer is to take place. In addition, this design provides a more rigid disc.

2. **Estimate of the Minimum Thickness of Stainless Steel**

The resistance to heat flow of the metal is small compared to the resistances offered by the fluid films. However, since the resistance of the stainless steel varies directly as the thickness, an investigation was made to determine what minimum thickness would be permissible without exceeding safe stress limits.

The deflection of very thin plates may be larger than one-half the thickness of the material. Under these conditions the middle surface becomes appreciably strained and the stress in it cannot be ignored. This stress, called the diagonal or direct stress, enables the plate to carry part of the load as a diagonal in direct tension. Thus the plate exhibits stiffer characteristics than would be predicted by ordinary theory of flat plates. Stresses and deflection for a given load are consequently less.

The determination of the deflection and stress at any point in a configuration shown in Figure 39 would be extremely difficult. The section of largest unsupported area is that indicated as area A. It would seem reasonable to make a simplifying assumption that this is the area of maximum deflection and stress.

A second assumption must be made as to the type of support that is offered by the soldered baffles. Formulas have necessarily been derived for two ideal conditions of support - held, or held and
Figure 38
Flow baffle design and probable flow pattern. On left, seven-disc; on right, proposed baffle for thirteen-disc unit.

Approximated Areas For Maximum Stress and Reflection Calculations

Figure 39
fixed - both of which are never truly realized in practice. A slight yielding of a nominally fixed edge greatly relieves the stresses there but increases the deflection and stress at the center. Considering the area referred to above, edge 1 could perhaps be considered fixed. Sides 2 and 3 cannot be so easily defined, since there will, no doubt, be considerable yielding at these points. Side 4 should perhaps be considered free with a point support at point X. This imposes formidable complications to the problem \(19\)\(20\) and further simplifications must be made.

The approach taken to the problem was to first assume all sides held (not fixed) to obtain an estimate of the maximum deflection and stress at the centre. Secondly, all sides were assumed fixed to obtain a figure for the maximum stress that might be realized. Thirdly, it was recognized that point X would be a point of high stress concentration. This can be partly relieved in construction by increasing the area of this point. However, to obtain the worst conditions calculations were made assuming area B held, and fixed, on four sides. Using the formulae given in reference \(20\) the results tabulated in Table III were obtained.
TABLE III - STRESS AND DEFLECTION FOR VARIOUS THICKNESS OF STAINLESS STEEL FOR CONFIGURATION SHOWN IN FIGURE 39

<table>
<thead>
<tr>
<th>Thickness of metal (in.)</th>
<th>Assumed Pressure psi</th>
<th>All sides held not fixed</th>
<th>(1) All sides held &amp; fixed</th>
<th>Max. Defln. (in.)</th>
<th>Max. Stress (psi)</th>
<th>Max. Defln. (in.)</th>
<th>Max. Stress (psi)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>Area A</td>
<td>Area B</td>
<td>Area A</td>
<td>Area B</td>
<td>Area A</td>
<td>Area B</td>
</tr>
<tr>
<td>.018&quot;</td>
<td>10</td>
<td>.001&quot;</td>
<td>.003&quot;</td>
<td>6.8</td>
<td>15.8</td>
<td>.0005&quot;</td>
<td>8.0</td>
</tr>
<tr>
<td>.015&quot;</td>
<td>10</td>
<td>.002&quot;</td>
<td>.012&quot;</td>
<td>9.2</td>
<td>23.0</td>
<td>.0007&quot;</td>
<td>10.8</td>
</tr>
<tr>
<td>.012&quot;</td>
<td>10</td>
<td>.004&quot;</td>
<td>.014&quot;</td>
<td>15.3</td>
<td>24.3</td>
<td>.0015&quot;</td>
<td>18.5</td>
</tr>
<tr>
<td>.010&quot;</td>
<td>10</td>
<td>.005&quot;</td>
<td>.015&quot;</td>
<td>22.0</td>
<td>24.6</td>
<td>.003&quot;</td>
<td>26.0</td>
</tr>
<tr>
<td>.008&quot;</td>
<td>10</td>
<td>.008&quot;</td>
<td>.016&quot;</td>
<td>24.2</td>
<td>25.2</td>
<td>.004&quot;</td>
<td>39.0</td>
</tr>
</tbody>
</table>

(1) Maximum stress and deflection occur at the centre of the rectangle.
(2) Maximum stress occurs along the centre of the longest edge.

It would appear from the analysis that one could use a thickness of .012" and have a safety factor of greater than 2 for the particular stainless steel and silver solder used.
VIII DESIGN OF A HEAT EXCHANGER OF AUTOGNOSIS OXYGENATION

1. Need for Another Heat Exchanger

The rotating heat exchanger has a number of advantages when the technique for by-pass utilizes an oxygenator. It has, however, a number of distinct disadvantages when used in such autogenous oxygenation techniques such as that used by Drew. For this relatively new technique, an efficient heat exchanger with minimum priming volume is required. The rotating-disc heat exchanger would introduce unnecessary traumatic damage, and since the oxygenator is not required, would add considerable priming volume to the system.

- A number of heat exchanger designs were again considered. In general, those reported utilize a large number of small bore tubing. One such design employs 112 tubes with a bore of .093 inches.\(^{21}\) This requires a priming volume of 195 ml. and is difficult to clean. Another design inserts nylon filter rods in 3/16 inch I.D. tubing to reduce the priming volume.

The maximum cooling surface and minimum priming volume are realized when the blood is passed through the exchanger in a thin film of rectangular cross-section. A design with rectangular flow pattern was completed, but discarded because of the anticipated fabrication and sealing problems. Consideration was, therefore, given to thin laminar films as the next best pattern. A heat exchanger utilizing two concentric annuli was designed. However, to obtain the required surface area an annulus of 3.5 inches mean diameter was required to keep the pressure drop within tolerable limits. In addition, this design required a special order of stainless steel in lieu of having the unit constructed from nickel-plated
brass.

Figure 40 shows the details of a design which was more acceptable. This design uses to good advantage surplus stainless steel tubing originally purchased for another purpose. To obtain maximum heat exchange with minimum priming volume, blood is passed through an annulus in a film 0.053 inches thick. This annulus is formed by the inner and the outer surfaces of a 1/2 inch OD and 3/8 inch OD tubing respectively. Coolant is circulated on the outside of the 1/2 inch OD and on the inside of the 3/8 inch OD tubing. The unit provides 400 square inches of surface area with a priming volume of 170 ml. The blood flow pattern is laminar and should be atraumatic. The unit may also be easily cleaned and assembled. In contrast to the rotating heat exchanger, sufficient information is given in the literature to predict its performance with a reasonable degree of accuracy.

2. **Design Procedure**

The design of a heat exchanger is a trial and error process. The surface area required and the geometry of the design must be assumed and then the performance can be calculated. If the performance is not that required for the application the surface area must be increased or other steps taken to increase the over-all coefficient of heat transfer.

The number and length of tubes may be determined for an assumed surface area required, by the allowable pressure drop. The isothermal pressure drop for an annulus if given by

\[ P_f = \frac{2f}{g D_e} \sqrt{\frac{g}{\rho}} \text{meters of fluid, where} \]

\[ f = \frac{16 \rho}{\frac{16 \rho}{R_e} \left( \frac{d_1}{d_2} \right) V_f (d_2 - d_1) (d_2)} \text{and} \]
COOLANT INLET HEADER

COOLANT OUTLET HEADER

Inset For Gasket

$\frac{1}{4}$ OD x .02 Wall Al. Tubing.

$\frac{1}{2}$ Tubing

$\frac{3}{8}$ OD x .01 Wall SS. Tubing.

$\frac{1}{2}$ Male Quick Coupling.

Upper Baffle, 3 Req'd.

Clamps, 3 Req'd.

Dowls

$2\frac{7}{8}$

$3$

$3\frac{1}{2}$
\[
\phi(d_1) = \frac{(1 - \frac{d_1}{d_2})^2}{1 + \left(\frac{d_1}{d_2}\right)^2 + \left[ 1 - \frac{(d_1)^2}{d_2} \right] \ln\left(\frac{d_1}{d_2}\right)}
\]

All other symbols are defined in the list of nomenclature.

Below a Reynolds number of 2000 fair agreement has been obtained between these theoretical equations and experimental data except for values of \(d_1/d_2\) between .64 and .91. Within this range Nootbar and Kinter (23) have obtained a value of 1.35 for \(\phi(d_1/d_2)\) instead of the theoretical values of 1.43 and 1.50.

The Reynolds number for a blood flow of five liters/minute at a temperature of 10°C. through the proposed heat exchanger was calculated to be 42.4, well down in the laminar region. This was calculated on the basis that the relative viscosity of blood (4.7 at 37.5°C.) rises 15% at 10°C. (24) (25) The pressure drop at this flow rate and temperature was calculated to be .49 ft. of blood or approximately 8 mm.Hg. The pressure drop at the same flow rate but at normal body temperatures would be .19 ft. of blood or 3 mm. Hg. This pressure drop is quite acceptable.

3. Predicted Performance of the Proposed Design

The performance of the heat exchanger may be predicted assuming an efficiency, calculating the inlet and outlet temperature of the blood and the rate of heat removal. If the rate of flow and inlet temperature of the coolant is known, the outlet temperature of the coolant may be determined. The rate of heat transfer for these established conditions may then be calculated from:

\[ Q = UA\theta_l m \]
The rate of heat transfer calculated from this equation should equal the rate of heat removal from the blood. If it does not, another efficiency is assumed and the process repeated.

The overall coefficient of heat transfer $U$ may be determined from the expression:

$$U = \frac{1}{\frac{1}{h_d} + \frac{1}{h_c} + \frac{\nu}{k}}$$

The determination of $h_d$ and of $h_c$ for the inner coolant film involves the calculation of these coefficients for laminar flow in an annulus. Little theoretical work has been done on laminar flow heat transfer in annuli compared to that done on circular tubes. The complex inter-relationships associated with heat transfer make theoretical analysis of little value without experimental verification of the various coefficients.

This fact may be better appreciated by considering the factors involved. First, there are the physical factors of the fluid flowing past the transfer surface. Fluid density, viscosity, heat capacity, and thermal conductivity are all considered to have an effect on the heat transfer coefficient. The dimensional variables of the system also must be considered. These include the equivalent diameter, the distance from the fluid entrance, and the curvature of the flow path. In addition, the velocity and mechanism of flow (laminar or turbulent) have a significant effect.

Investigators have related these variables (using dimensional analysis) by a series of dimensionless numbers. This provides a means of predicting the performance of a proposed system using the experimental results of another. For an annulus these factors have been correlated
experimentally by Chen, Hawkins, and Solberg(26) with the following
expression:
\[ \frac{h\Delta T}{k} = 1.02 \left( \frac{WRe}{D_a} \right)^{0.45} \left( \frac{D_a}{L} \right)^{0.14} \left( \frac{D_a}{d_1} \right)^{0.4} \left( \frac{D_a}{d_2} \right)^{0.8} \left( \frac{d_2}{d_1} \right)^{0.05} \]

Various diameter annuli and water flow rates were used by Chen to
verify this expression resulting in an average deviation of 6.6% over
a range of Reynolds numbers of 200 to 2000.

This equation was used to determine the transfer coefficients
and thus predict the efficiency of the proposed heat exchanger. The
Reynolds number for the water (simulated blood) was 340 for the highes
flow rate, and 2100 for the coolant flow in the annuli. The \( L/D_s \) and
d_2/d_1 ratios of the proposed exchanger were also within the range used
by Chen although longer annuli of much larger diameter were used by him
in the verification of this equation.

The following assumptions were made to predict the performance
of the heat exchanger:

- Water (simulated blood) flow rate: 5 liters/min
- Water inlet temperature: 37.5°C
- Inlet coolant temperature: 1°C
- Coolant flow rate: 18 kg/min
- Heat exchanger efficiency: 0.45

The rate at which heat is removed from the blood based on
these assumptions is 4800 k-cal/hr. The film transfer coefficients
for the simulated blood and coolant were found to be 1430 and 2260 k-cal/hr
(sq-m)(°C.) respectively. Using Table 10-6 and Figure 10-20 of reference (27)
the film coefficient for the outer coolant surface of the main annulus
assembly was calculated to be 4200 k-cal/(hr)(sq-m)(°C.). This results
in an overall coefficient of heat transfer, using .01" stainless steel
of 855 and 1020 for the inner and outer surfaces of the annuli respectively.
The total rate of heat transfer through the tubing walls was determined to be 6200 k-cal/hr. Thus the efficiency of the heat exchanger with a simulated blood flow of five liters/minute will be at least 0.45; this is considerably greater than hypothermia exchangers previously reported. Because of the favorable results of these calculations, the construction of the proposed heat exchanger will be proceeded with as soon as technician help is available.
CONCLUSION

This thesis has discussed only those problems associated with open-heart surgery on which the engineering design has been completed by the author. There are a number of other problems which are under current investigation.

The most immediate requirement is to provide instrumentation so that the normal blood chemistry may be maintained as closely as possible during these operations. To properly control perfusion, the perfusionist must know the arterial and venous oxygen tension, and the arterial carbon dioxide tension. The present method of determining these properties is to take spot samples of blood to be analyzed. As a result, there is considerable delay in obtaining information necessary for corrective control. Polarographic cells have been developed to continuously monitor oxygen tension. Instrumentation is not presently available to continuously monitor carbon dioxide tension. However, this parameter is related to pH, which may be monitored to the required degree of accuracy. There remains the problem of safely introducing the polarographic cells and the rather delicate pH electrode into the extracorporeal circuit to ensure accurate monitoring without causing additional turbulence. These cells must also be adapted for easy removal and sterilization.

The oxygen and carbon dioxide tension of the blood may be maintained within normal physiological limits during extracorporeal circulation by controlling the blood flow rate, the oxygenator efficiency, and the carbon dioxide content of the gas in the oxygenator. Increasing the blood flow rate (pump speed) increases the venous oxygen tension,
but decreases the efficiency of the oxygenator, and results in a drop in arterial blood oxygen tension. The efficiency of the oxygenator, and hence the arterial oxygen tension, may be increased by increasing the disc speed. The pH of the arterial blood may be controlled by varying the amount of carbon dioxide introduced into the oxygenator or by simply controlling the rate of flow of pure oxygen through the oxygenator. (29)(30) A simple method of automatically controlling these variables is an interesting possibility which has not, to the author's knowledge, been developed.

In addition to the desirability of having instantaneous information of the disc speed, oxygen and blood flow rates, and of the blood parameters previously mentioned, the perfusionist must know the mean arterial and venous pressures, and the blood temperature. When hypothermia is introduced, esophageal, rectal, and coolant temperatures must also be observed. The anaesthetist and attending cardiologist must have access to most of this information and must also observe the electrocardiogram and electroencephalogram. It would also be desirable to present all these variables in such a way that the surgeon could assess the condition of the patient at a glance. This presents a problem of designing a unified system to accurately and simply display and record this information.

There are many other engineering problems associated with heart surgery which have not been adequately solved. A completely satisfactory artificial valve to replace unrepairable, diseased heart valves is not presently available. A method of adequately perfusing the coronaries during repair of the aortic valve would eliminate the need
to reduce the body to very low temperatures for this operation. A coronary cannula has been designed for this purpose and will be tested in the near future.

Perfusion time is still limited, largely because of the traumatic damage the blood constituents incur passing through the extracorporeal circuit. The recent development of carcinocidal drugs which must be administered by the restricted perfusion of the cancerous limb or organ has emphasized the need for a pump-oxygenator which would permit long-term perfusion.

There is also a need for basic engineering research to provide design information in this relatively new field. The viscosity of blood at low temperatures has only recently been investigated.\(^{31}\) The thermal conductivity of blood has not been determined. In co-operation with Professor Green of the Mechanical Engineering Department, apparatus has been designed to measure this parameter for blood and other biological fluids. There is considerable information on the toxic effects of various metals and other construction materials when ingested or inhaled. A quantitative report of the toxic effects of exposing the blood to various construction and soldering materials would be invaluable in the design of equipment for open-heart surgery.

While a vast number of engineering problems associated with open-heart surgery remain to be solved, a small contribution has been made with the successful completion of those problems discussed in this thesis.
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