A DISCUSSION OF THE PROBLEMS OF HEAT EXCHANGE BLOOD WARMING DEVICES

BY
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SUMMARY
Effective warming and infusion of blood is governed by factors which can be theoretically analysed. How effectively blood can be heated depends on the dimensions of the tube and its ability to transfer heat. The output temperature also depends on the specific heat and thermal conductance of the liquid. However, the temperature of the infusion into the patient will be decreased by heat loss in the output line. The maximum flow which can be achieved is determined by the liquid's viscosity and the resistance of the tubing and veins. Each of these factors is discussed and practical examples are shown. A summary of the points in the design of a good heat exchanger is given.

During the development of a warming unit at Sydney Hospital (Russell, 1969) it became apparent that a single unit would not suit a wide range of demands. To see more clearly the problems and consequences of changes in design, some theoretical approaches were used. This paper discusses briefly the major problems of a tube heat exchange device used for warming blood. The main factors involved are the heat exchange, the thermal conductance and specific heat of the blood, the flow through the unit, and the heat loss in the output line. Each of these factors is discussed in turn.

HEAT EXCHANGE

General.
The heat exchange is governed by the temperature gradients, the dimensions of the unit and the thermal conductivity of the various layers through which heat must flow.

The total temperature gradient is almost constant, as the blood temperature, and therefore the input temperature must be kept at about 4°C before infusion to minimize deterioration. The bath temperature is not limited except by the blood temperature which may be reached at zero output. Unless a circulating volume is created which is independent of the input and output (fig. 1), the blood temperature at zero output after unlimited time will equal the temperature of the bath. That is, the curve of the blood output temperature versus the reciprocal of the flow rate (time in contact) is asymptotic to the bath temperature. For a unit such as the Sydney Hospital one, an overall coefficient of heat transfer may be calculated. This is the mean heat transfer per unit time and it is calculated directly from:

$$h_c = \frac{(\Delta \theta_1 C w)}{(\Delta \theta_2 A)} \quad \text{(1)}$$

where

- $h_c =$ coefficient of heat transfer in cal/sec/sq.cm/deg. C;
- $\Delta \theta_1 =$ temperature difference between the output and the input fluid in deg. C;
- $\Delta \theta_2 =$ temperature difference between the bath temperature and the mean blood temperature in deg. C;
- $C =$ specific heat in cal/g/deg. C (vide infra p. 348);
- $A =$ area of heat exchange in sq.cm (for Sydney Hospital unit approx. 450 sq.cm);
- $w =$ mass flow rate in g/sec.

For the Sydney Hospital unit using the constants mentioned above and the values of $\Delta \theta_1$ and $\Delta \theta_2$ derived previously for a flow rate of 95 ml/min:

$$h_c = 4 \times 10^{-3} \text{ cal/sec/sq.cm/deg. C for water.}$$


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Awarded the Gilbert Troup-Australian Society of Anaesthetists Prize, 1967, for the best essay written during the year, by an anaesthetic registrar or junior specialist who is within five years of obtaining a postgraduate qualification.
Diagram of two types of heat exchange circuits.

This unit and similar heat exchangers each can be regarded simply as a long tube. If a tube which is long compared to its diameter is used as a heat exchanger, some predictions can be made of the effect of increasing the priming volume. The priming volume can be increased in two ways, either by increasing the radius of the lumen or by increasing the length of the tube.

**Radius and length.**

The mathematical treatment of changes in the dimensions of the unit is complex. However, in an ideal case, where the thermal resistance of the wall is much less than the thermal resistance of the blood, it can be shown that the output temperature is independent of the radius (see Appendix 1). The radius will still affect the unit in terms of the resistance to flow and the priming volume.

The final temperature is dependent on the term $L/w$ in a non-linear fashion (see Appendix 1). Thus if both $L$ and $w$ are varied in proportion, the final output temperature will not alter. Once an output temperature is accepted as being suitable, the corresponding flow in a tube of particular length can be found quickly experimentally. The length required to achieve the accepted output temperature with any desired maximum flow now can be calculated directly.

**Example:**

The desired output temperature is 38°C.

In the present unit, an output temperature of 38°C is found at a flow rate of 55 ml/min (Russell, 1969; fig. 3).

The present length of the unit is 220 cm.

The desired maximum flow is 175 ml/min (with 38°C temperature), i.e. $L/w = 220/55$

Thus the required length ($L_r$) = $220 \times (175/55) = 660$ cm

It should be remembered that series lengthening of the channels would have a proportional effect on the resistance to flow. However, parallel lengthening of the channels would improve similarly the output temperature for a given flow, with the advantage of a decrease in resistance (fig. 2).

The truth of this can be seen if the increased length is regarded as three channels of 220 cm which will have the required output flow of 175 ml/min and output temperature of 38°C (i.e. each channel still delivers only 55 ml). It does not matter how the required length of tubing is disposed, the same output temperature still will be achieved, even if a single long tube is used.

**Layers.**

The heat exchanger passes heat from a source at high temperature to the input fluid at low temperature. To achieve this the heat must penetrate all the intermediate layers and the skin layers of the source and fluid. In recent designs (Knight and Wellard, 1962; Burke, 1964; Burton and Holderness, 1964; Williams and Abrams, 1964; Holmes, 1965) heat must pass from the water through the p.v.c. plastic and through the layers of blood if laminar flow is present. Thus the conductivity of the plastic and the blood, as well as the thickness of these layers, will affect the efficiency of the heat exchanger.

To see the effect on heat transfer, the layers can be simplified to a plane wall, provided the thickness of the wall is small compared to the
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For the present unit this reduces to:

\[ 250 = (1/h_w) + 107 + 130 \]

i.e. \( (1/h_w) \) is very small

and \[ 1/h_c = (x/k_p) + (1/h_a) \] approximately \( \ldots \) (3)

The high value of \( h_w \) is an expression of a good coefficient of heat transfer between the water and the surface of the unit.

Equation (3) can be valid only if good circulation is present in the bath and thus the effect of the water layer is minimized by agitation. This replaces heat conduction by forced convection (Knight and Wellard, 1962; Williams and Abrams, 1964; Russell, 1969).

The blood layers can be regarded as constant with laminar flow and thus only the plastic remains variable. Reduction of the thickness of the plastic layer is not possible at present. The tensile strength of the unit must be adequate to stand pumping pressure and the relative importance of the plastic layer would decrease as it became thinner. Below a certain point a thinner wall would have very little effect on total thermal conductance. This would not appear to be true until a wall thickness of less than 0.01 cm is reached, i.e. if \( x/k_p \) in equation (3) is less than about 30, so the effect on \( 1/h_c \) will be small, which means if \( x \) is less than 0.012 cm. This very thin p.v.c. plastic would be difficult to use, but it is apparent that the plastic should be kept as thin as is practicable.

Substitution with materials of better conductance would not seem an advantage at present. The blood cannot be substituted and the water bath provides a cheap and large heat reservoir which is transparent and also avoids the problem of an air layer with its very low conductance. The plastic has the advantage that it is cheap and thus can be disposable. It is also transparent, so that air bubbles can be seen and cleared before the warmer is used. However, it does not have a good coefficient of heat transfer and the performance of the warmer would be improved if a substance of better conductance were used. Disposable units made from, say, aluminium foil would have a better conductance, but the technical difficulties of manufacture and sterilization appear greater. Certainly, it would seem advisable to use material cheap enough to make each unit disposable. This would increase the safety for the patient, by
avoiding the possibility of contamination from previous use.

The blood layers could be destroyed by creating turbulent flow within the unit. However, the Sydney Hospital unit has a theoretical mean critical flow of approximately 370 ml/min for water and a figure of about four times this for blood would be predicted from Reynold's formula (Appendix 3). Thus, significant turbulent flow is unlikely to occur in clinical practice. The critical flow figures above are for 37°C.

However, with the lower temperature initially present and thus the higher viscosity this figure will be closer to 1000 ml/min (2.5 × 370 ml/min) (Burton, 1965) and the equivalent would be 1400 to 3700 ml/min for the critical flows with blood. The critical flow would reduce in proportion to any decrease in radius, and smaller channels could be made but this would affect the resistance. Inducing turbulence by irregularities in the channel wall at various points would provide some reduction in laminar flow but again there would be an increase in resistance.

Some mixing effect is present at all flows because of the modified flow effect caused by changes in viscosity with temperature (McAdams, 1954). This results in radial mixing and distortion of the parabolic profile of the laminar flow. This effect is extremely small, however, over the temperature range and flow used here.

THERMAL CONDUCTANCE AND SPECIFIC HEAT

Some units have been tested by warming ice water (Burton and Holderness, 1964; Holmes, 1965; Russell, 1969), which has a specific heat of 1 cal/g/deg. C. As the specific heat of blood is lower (about 0.87 cal/g/deg. C; Mendlowitz, 1948), the amount of heat required is less to achieve a given temperature, i.e. the output temperature should be higher at a given flow.

\[ H = wC \Delta \theta, \]  
where \( H \) = quantity of heat/unit time; cal/sec.

However, the thermal conductivity of blood appears to be lower than that of water and so the heat transfer is less efficient. The net result is a similar performance for both blood and water as shown in Russell (1969), table II.

FLOW

General.

The improved flow with warmed blood would appear to be related to two factors. If laminar flow is present the Poiseuille-Hagen equation can be applied.

\[ Q = (\pi(P_1 - P_2)r^4)/8L\mu \]  
where \( P_1 - P_2 = \) pressure drop.

Thus the flow is proportional to the 4th power of the radius of the tube and inversely proportional to the viscosity.

Viscosity.

If the warm infusion is compared with a cold infusion without the unit in the circuit, a better flow occurs if the total resistance of the unit plus the cannula or needle and the patient's veins to about the subclavian in the warmed circuit, is less than the resistance of the cannula and the patient's veins in the cold circuit. The two most important factors would appear to be the change in the viscosity of the blood with temperature, and the change in the diameter of the veins. The change in viscosity as blood is warmed is about 1.7 times from 22°C to 37°C (Rand et al., 1964) or 2.5 times from 0°C to 37°C (Burton, 1965). If the viscosity effect was the main cause of improvement, it should be present in laboratory trials and furthermore the improvement in flow would be roughly the same in each of the paired patient infusions (Russell (1969); tables III and IV). This latter is not so.

The laboratory trials suggest that the viscosity effect is usually not prominent as the increased resistance to flow from the longer circuit approximately balances the viscosity improvement (Russell, 1969; table III).

Veins.

The radius of the vein will play a very important part, as a small change in radius will affect markedly the resistance. Other writers have considered this to be the reason for better flows with warmer blood (Mollison, 1961; Holmes, 1965). When a variable improvement in flow occurs with different patients, this would seem to be the most likely explanation. The degree of improvement will be greater if the patient is one who will react to cold with marked vasoconstriction (venous spasm). If the ambient temperature is high or the patient's sympathetic tone is depressed, e.g. by
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general anaesthetic, then this effect will be less marked. From table IV (Russell, 1969) it would seem that venous spasm does reduce the flow of cold blood and the changes in flow were probably due entirely to changes in the calibre of the veins. However, the paired patient trial did not take into account any changes in the patient’s venous pressure, nor changes in the ambient temperature. Isothermal viscosity differences between the bloods from the paired bottles were also not investigated. Although none of these factors would seem adequate to account for the wide variation in improvement, it cannot be said conclusively that venous spasm alone has produced the variation observed.

**Table I**

<table>
<thead>
<tr>
<th>Flow (ml/min)</th>
<th>Output temp. (°C)</th>
<th>Infusion temp. (°C)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Thick wall</td>
<td>Thin wall</td>
</tr>
<tr>
<td>20.6</td>
<td>39.7</td>
<td>37.2</td>
</tr>
<tr>
<td>20.8</td>
<td>39.8</td>
<td>37.3</td>
</tr>
<tr>
<td>5.7</td>
<td>39.7</td>
<td>31.0</td>
</tr>
<tr>
<td>5.7</td>
<td>39.6</td>
<td>31.0</td>
</tr>
</tbody>
</table>

Internal diameter 3.2 mm.
Thick wall 1.65 mm; thin wall 0.76 mm.
Ambient temperature 23.5°C.
Output temperature corrected to a bath temperature of 40°C. Each infusion temperature is the average of 10 readings; these readings were done with the Sydney Hospital unit using iced water (0–2°C).

**Tubing.**

The resistance to flow through the unit could be improved by increasing the diameters of the output and input tubes as these make up 150 cm of the 370 cm total length. If the internal diameters of the tubing and cannula were increased by 50 per cent, a decrease in resistance of about 5 times, i.e. $(1.5)^4$, could be expected in this tubing and this would result in a fall in total resistance of about 58 per cent. A free flow by gravity in excess of 100 ml/min should be possible.

Decreased resistance would follow also if the unit were produced with its own drip set to eliminate the narrowing in the line caused by male and female Luer fittings.

Flow resistance would decline proportionally if the channels were formed in parallel instead of in series (fig. 2). This has been done in the warming units designed by Burton and Holderness (1964) and Williams and Abrams (1964). These should be low resistance units although no figures are given for their performance. However, probably priming would be more difficult.

**HEAT LOSS IN THE OUTPUT LINE**

Heat loss during transmission of the warmed blood from the exchange unit to the patient depends on factors similar to those involved in the heat transfer within the exchanger itself. The four most important factors are flow rate, length of tubing, the conductivity of the layers and the temperature gradient. With a constant heat gradient, the heat loss increases as the flow rate declines. However, the heat gradient is not constant but declines as the flow increases because of a lesser temperature rise in the heat exchanger, and also as the blood cools in the line. If the gradient is minimized by warming the environment, less heat loss will occur. It is thus doubly advisable to warm the room during paediatric exchange transfusion in order to keep the patient and the blood from cooling. The rate of infusion cannot be varied to suit low heat loss but depends on the patient's requirements. Usually, at low flows the amount of blood infused is small relative to the patient's total body water and the cold blood will not cause a significant drop in body temperature.

The length, as before, is very important and to conserve heat the shortest possible lead should be used. The short lead will have also a lower resistance to flow.

The conductivity of the layers is the reverse of the previous problem. The layers are now blood, plastic and air. If the ambient temperature is markedly lower than the output temperature, the air removes the heat by convection, which usually is very effective. The p.v.c. plastic can be increased in thickness but this does not affect significantly the loss of heat in the output lead (see table I). A stationary air layer is a very poor conductor of heat, and insulation which adds such an air layer greatly reduces the heat loss. Probably the best method of producing a stationary air layer is by plastic foam or similar material about 2 cm thick. The difficulty with any insulation on the output lead is that it will increase the bulk and weight of the lead and reduce the flexibility.
CONCLUSIONS

Increasing the length will improve the output temperature for a given flow. The effects of change in length can be predicted by consideration of the output temperature and flow within the present length.

All tubes should be as large as possible to minimize the flow resistance within the limits of an acceptable priming volume. The ideal is a multichannel parallel flow pattern which would have a low flow resistance and a good heat exchange.

The conducting layers are defined for warming blood but agitation of the water bath is important to minimize the water layer and facilitate the heat transfer by forced convection.

Turbulent blood flow would improve the heat transfer but for the flows used in practice a very small tube radius would be necessary to achieve turbulence and the very high resistance to flow would be prohibitive.

Heat loss in the output line will be greatest at low flow rates. This loss would be minimized by a short insulated output lead.

In the present designs the p.v.c. plastic is a significant barrier to good heat transfer at flows of about 100 ml/min. The plastic should be no thicker than 0.01 cm to allow optimum heat transfer. However, at present, mechanical considerations preclude the use of plastic as thin as this.

APPENDIX 1

Presuming a laminar flow, with a parabolic velocity profile and uniform wall temperature, Graetz obtained the following formula to describe heat transfer (McAdams, 1954).

\[ \frac{t_2 - t_1}{t_w - t_1} = 1 - 8\psi(n_t) \]

where \( t_1 \) is the input temperature in deg. C; \( t_2 \) is the output temperature in deg. C; \( t_w \) is the wall temperature in deg. C; and \( \psi(n_t) \) represents the convergent infinite series.

\[ \psi(n_t) = 0.10238e^{-1.627n_t} + 0.01220e^{-89.22n_t} + 0.00237e^{-212n_t} + \ldots \]

where \( n_t \) represents the dimensionless term \( \pi kL/4wc_p \);

\( k \) = thermal conductivity of liquid in cal/sec/sq.cm/deg. C per cm;

\( c_p \) = specific heat in cal/g/deg. C.

Thus (1) the output temperature is independent of the radius;

(2) for a given liquid, the output temperature is a function of \( L/w \).

APPENDIX 2

McAdams' equation (9-23b) which utilizes Graetz' equation and the heat balance is:

\[ h_wD/k = [2/\pi][wc_p/kL][1 - 8\psi(n_t)/1 + 8\psi(n_t)] \]

The graphical representation of this (curve ACD, fig. 9-17, p. 231) can be solved for 95 ml/min in the present unit. \( L = 220 \) cm.

\[ wc_p/kL = 5.0 \]

and thus from McAdams' graph

\[ h_w = 7.7 \times 10^{-3} \text{ cal/sec/sq.cm/deg. C per cm} \]

and \( 1/h_w = 130; \)

\( k \) water = \( 1.45 \times 10^{-3} \text{ cal/sec/sq.cm/deg. C per cm} \);

\( c_v \) water = \( 1.0 \text{ cal/g/deg. C} \);

\( D \) = diameter of tube;

\( = 0.56 \) cm for the present unit.

APPENDIX 3

Reynold's formula:

\[ v_c = K\mu/\rho \]

where \( v_c \) = critical velocity cm/sec;

\( \mu \) = viscosity in poise; approximately 0.007 for water at 37°C and 0.026 for blood at 37°C (Coulter and Pappenheimer, 1949); high shear;

\( K \) = critical Reynold's number (dimensionless); approximately 1000 for water and blood (Coulter and Pappenheimer, 1949);

\( d \) = density;

\( r \) = radius;

but \( Q_c = v_c \pi r^2 \) for a tube,

i.e. \( Q_c = (K \mu \pi r)/d \),

where \( Q_c \) = critical flow in ml/sec.

The mean radius of the Sydney Hospital unit in the heat exchange region is 0.282 cm. Therefore the critical flow is 6.16 ml/sec or 370 ml/min.
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ACKNOWLEDGEMENTS

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REFERENCES


UNE REVUE DES PROBLEMES D'ECHANGE DE CHALEUR DANS LES APPAREILS POUR RECHAUFFER LE SANG

SOMMAIRE

Le réchauffement efficace et l'infusion de sang dépendent de facteurs, qui peuvent être analysés théoriquement. L'efficacité du réchauffement de sang dépend des dimensions du tube et de sa capacité de transmettre la chaleur. La température de débit dépend encore de la chaleur spécifique et de la conduction thermique du liquide. Mais la température du liquide infusé au malade est réduite par la perte de chaleur qui se produit au niveau du tube de débit. Le flux maximal qu'on peut obtenir, est déterminé par la viscosité du liquide et la résistance des tubes et veines. Chacun de ces facteurs est discuté et des exemples pratiques sont cités. Un aperçu des facteurs dont on doit tenir compte lors de la mise au point d'un bon échangeur de chaleur est donné.

EINE DISKUSSION ZU DEN PROBLEMEN DER GERATE ZUR BLUTERWARMUNG DURCH WARMEOBERTRAGUNG

ZUSAMMENFASSUNG


BRISTOL ANAESTHETIC CLUB

Programme for Summer Term 1969

WEDNESDAY, MAY 7. 8.00 p.m. School of Nursing, Bristol Royal Infirmary.
“ A New Muscle Relaxant” (Dr. B. A. Sellick).

WEDNESDAY, JUNE 4. 8.00 p.m. The Sisters’ House, Frenchay Hospital, Bristol.
“From Arrowhead to Ampoule” (Dr. K. Bryn Thomas).

WEDNESDAY, JULY 2. 8.00 p.m.
The Blood Transfusion Centre (Reception Room), Southmead Hospital, Bristol.
“Anaesthesia and the Reduction of Surgical Haemorrhage” (Dr. W. N. Rollason).