BRITISH MEDICAL JOURNAL

LONDON SATURDAY JULY 11 1953

A MECHANICAL HEART-LUNG FOR USE IN MAN

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Foreword

Some years ago the department of surgery in the Postgraduate Medical School of London was concerned with research in the perfusion of isolated human organs. The perfusion of isolated organs by blood artificially oxygenated and propelled by a mechanical pump cannot be continued for more than a few hours. After some hours the perfused organ swells visibly and the outflow of blood from it progressively and rapidly diminishes. After a short and not very successful examination of the problem of prolongation of the survival time of isolated organs in these conditions attention was turned to the extracorporeal heart-lung circulation for the maintenance of the circulation of the experimental animal by a total replacement of its own heart and lungs, comparable to that which would be required in the application of an apparatus of this kind to intracardiac surgery. It seemed to us, after some experience in this field, that a machine to undertake the whole function of heart and lungs in a patient subjected to intracardiac surgery was not yet an immediately attainable aim.

Animals maintained entirely on an extracorporeal heart-lung circulation sometimes survive for indefinite periods, but a high proportion fail to survive, perhaps for reasons not dissimilar to those which make prolonged perfusion of the isolated organ impossible. The natural lung may have other functions than oxygenation. In any case, there are immediate practical difficulties in employing a heart-lung machine to replace the heart and lungs of a patient subjected to cardiac surgery. Heparinization must be complete, and the difficulty of obtaining a bloodless field even in a heart to which blood is not returned by the venae cavae has already been demonstrated at least once in man. Our attention accordingly passed to the use of a heart-lung machine as a form of "assisted circulation."

Our aim in this field became the ultimate use of an artificial heart-lung machine for the treatment of certain acute forms of cardiac or pulmonary failure. The value of artificial oxygenation of the blood for even a short period in acute pulmonary oedema or status asthmaticus seemed obvious enough, for, provided the patient can be kept oxygenated for a time, these pathological circumstances are often self-limiting. In coronary thrombosis there are patients in whom it might be expected that the

temporary provision of a high aortic pressure might tide the patient over a fatal low-pressure period, and might improve such coronary circulation as was still open. In congestive heart failure, when failure has been due to a relatively small fall in a previously low cardiac output, to withdraw blood from the congested veins and to perfuse it oxygenated into an arterial tree would seem to combine the advantages of venesection and adequate oxygenation. The mechanism would have a particular advantage in preparing grossly cyanotic sufferers from mitral stenosis for a valvulotomy which they could not otherwise sustain. Dr. Melrose proceeded to devise an apparatus which might ultimately be applied to these and similar purposes and to test that apparatus in the form of an assisted circulation in parallel with the natural circulation of an experimental animal. The results of his whole-time activity in this field over a period of four years are recorded in the two papers which follow. His data indicate that his machine might now be justifiably and humanely applied to certain varieties of anoxia and circulatory failure in man.

Artificial Maintenance of Life

The possibility of maintaining life artificially in animals or parts of animals has excited the interest of many over the last century. Numerous attempts have been made to create an apparatus capable of this, but until the last decade these efforts have been sporadic and isolated in character. The publication by J. H. Gibbon, jun., in 1937 of a paper on artificial maintenance of the circulation during experimental occlusion of the pulmonary artery heralded the beginning of a new approach to the problem. His object was to short-circuit the heart and lungs of animals to allow intracardiac surgery in a bloodless field. Coinciding as it did with the emergence of a new era in surgery—that of replacement rather than ablation—his work stimulated a great new interest and challenge which was taken up in many surgical centres. This challenging problem was, and still remains, to produce an extracorporeal apparatus capable of maintaining temporarily the entire circulation of the human body.

Contributions in this field from Crafoord and Björk in Sweden, Kolff and Jongbloed in Holland, André Thomas in France, Dennis and Leland Clark in America, and many others have resulted in continuous progress, and now the stage is almost set for clinical use.

*Part of the work was done under the tenure of a Studentship in Research from the Medical Research Council.

Stimulated by these advances, it was decided in 1949 that a machine should be constructed to serve three purposes. First, to carry temporarily a part of the cardio-respiratory function when the function of either or both of these organs is impaired; secondly, to provide a complete diversion of the blood stream away from the chambers of the heart in order to permit intracardiac surgery; and, thirdly, to perfuse isolated organs.

FIG. 1.—The pump-oxygenator with controlling apparatus.

The Pump-oxygenator

Initially, considerable experience was gained with a pump-oxygenator essentially similar to that described by Björk (1948a) and Crafoord (1949). In this machine a number of disks dip into a trough of blood; when rotated they pick up on their surfaces thin films of blood, exposing them to an atmosphere of oxygen, and in this way create a large gas-liquid interface. After this method had been investigated and compared with other methods, particularly those involving the dispersal of oxygen as minute bubbles, it was decided to embody the principle of rotating disks in a machine of novel design capable of circulating adequately oxygenated blood at a rate of at least 3 litres a minute (Fig. 1).

Two factors are important in the utilization of this principle. The first is that the optimum period for which each red blood cell should be exposed to oxygen is in the region of half a second. The second factor is that the exposure time must be achieved without increasing above a definite maximum the speed of rotation of the disks. Rotational speeds in excess of 80 r.p.m., corresponding to a velocity at the periphery of the disks of 52 cm. a second, frequently lead to foaming of the blood.

These two factors are of major importance, since they limit the effective area for oxygenation of each plate, but there are others which also affect the efficiency of such machines. These are: (1) The area of interface between blood and gas should not vary with changes in the volume of blood in the machine. (2) Mixing of blood should be very complete, for no provision can be made to replace completely each saturated film, after exposure to oxygen, with one as yet unsaturated. (3) The measures to ensure mixing should not impede smooth and rapid flow through the machine. (4) Foaming if it occurs should not greatly interfere with the efficient operation of the artificial lung. (5) The

apparatus should be capable of being completely dismantled for effective cleaning. (6) Adequate surface area should be obtained without the use of a very large volume of blood to charge the machine. (7) For ease in maintaining sterility, a system as nearly closed as possible is desirable.

From these considerations it is seen that enlarging the diameter of the disks makes a longer exposure of time inevitable if a safe rotational speed is to be maintained, while increasing the number of disks greatly increases the blood volume required for charging. Neither of these measures, nor a combination of them, confers any improvement in the mixing, or in ensuring the maintenance of an optimum blood film size.

The Oxygenator

The new oxygenator (Fig. 2) is essentially a rotating cylinder set at a slight angle to the horizontal. Within it blood travels under the influence of gravity from one end to the other, and is spread over a multitude of surfaces formed by blades projecting into the lumen of the cylinder. These blades are arranged within the cylinder somewhat in the form of an Archimedean helix. This analogy is not strictly accurate, but may clarify the method of construction, which is in fact carried out in the following manner. A cylinder of "perspex" 30 in. (76 cm.) in length and of internal diameter of 8 in. (20.3 cm.) is free to rotate in its long axis while supported on rollers at 20 degrees to the horizontal. It is closed at each end by an end-plate of "tufnol" 10 in. (25.4 cm.) in diameter. In this cylinder

FIG. 2.—The oxygenator.

annular plates of $\frac{1}{24}$ in. (1 mm.) perspex, with an overall diameter of $7\frac{7}{8}$ in. (20 cm.) and an internal diameter of $4\frac{3}{8}$ in. (11 cm.), are separated one from the other by annular washers. These washers have the same external diameter as the plates but an internal diameter of $5\frac{1}{8}$ in. (13.6 cm.).

FIG. 3.—Sectional drawing of the oxygenator, indicating form of construction.

They are constructed from perspex 5/16 in. (0.8 cm.) thick. The holes in the plates are concentric, but the holes in the washers are cut eccentrically, the centres of the hole being 1 in. (2.54 cm.) from the true centre of the washers.

These plates and washers are mounted in the cylinder on chromium-plated rods and are freely detachable for cleaning. A maximum of 76 plates and their separating washers may be used to provide a surface area of approximately 1 square metre.

The plates (Fig. 3) are fitted so that their central holes form a cylindrical cavity, coaxial with the enclosing cylinder.

FIG. 4.—A portion of the oxygenator in sagittal section.

der, but the arrangement of the washers between each plate varies with each group of five. Each group of five washers are mounted so that the holes in them form a cylinder eccentric to the cylinder formed by the holes in the plates, and, as the internal diameter of the washers is larger by 1 in. (2.54 cm.) than that of the plates, these plates project as crescents within the cavity formed by the holes in the washers. Each group of five washers and plates is arranged relative to the next group so that crescentic protrusions of the plates are at opposite poles. In sagittal section, therefore, the cylinder presents a series of alternating platforms and troughs, these troughs being subdivided into five compartments by the five plates (Fig. 4).

Passage of the Blood

In describing the path of the blood passing by gravity along the tilted oxygenator, it is helpful to consider first the hypothetical case in which both washers and plates are fitted so that the holes in them are concentric. A pool of blood would lie along the base of the cylinder formed by the holes in the washers to the depth of the level of the holes in the plates. Rotation of the oxygenator would not affect the position of the pool in space. Where the holes in the washers are eccentric to the holes in the plates and the circumferences of these holes touch at one point in any group of five, a different action is apparent. When the point of touching is at top centre a pool of blood will be formed as in the previous case. During any revolution, as the point of touching approaches bottom centre this pool will be emptied by the action of the washers, which in effect rise to the level of the holes in the plates. The alternation of phase between adjacent groups ensures that as a pool in one group empties so it is re-created in the next group.

Further, during any revolution the crescentic projections of the plates are alternately fully covered by the pool of blood, and also, together with adherent films of blood, fully exposed to the gas contained in the oxygenator.

The stream of blood passing down the oxygenator is alternately raised and lowered by the action of the washers. This action avoids the necessity of the edges of the plates cutting across the streamline in order that they may become

covered with blood. In fact, blood is poured over the plates and into the pool formed between them, then raised up and carried down the oxygenator before being poured over the succeeding group of plates.

This arrangement permits a thorough mixing of the blood with a smooth and rapid passage along the machine. It also ensures that all the available surface area is utilized without reference to the volume of blood contained in the machine. Furthermore, it enables the machine to work with a relatively small charge of blood.

The cylinder is rotated by a friction wheel, driven by an electric motor. Adjustment of the position of the driving wheel in relation to the diameter of the end-plate allows cylinder speeds between 24 and 120 revolutions a minute to be obtained. At maximum speed the surface area exposed per minute is in the region of 120 square metres, though it is preferable to use only as great a rotational speed as is necessary to obtain efficient oxygenation.

The inlet and outlet tubes for the blood, together with the inlet for oxygen and a fourth tube used to estimate the level of the blood in the collecting chamber, enter at one end. These tubes are carried by a perspex rod which passes through a simple felt seal in one end-plate down the centre line of the oxygenator to a chromium-plated bearing on the other end-plate. No attempt has been made to make a gas-tight seal, though this is undoubtedly possible.

Experience has shown that the distance between the plates can be reduced from 5/16 to 3/16 in. (0.8 to 0.48 cm.) without increasing the danger of foaming. It is expected that a moulding in nylon or teflon (polytetrafluorethylene) will be produced combining both washer and plate, and this will enable the number of parts to be halved. It will also enable autoclaving to be used for sterilization. This latter point is undoubtedly of great importance, and might be considered essential in a machine for use in human beings. Further, during the moulding process a large number of corrugations can be pressed into the surface of the plates and the surface area of each greatly increased.

The Pumping Circuit

Since the oxygenator is not part of a closed circuit, it is essential that two pumps be used; one to act as the right side of the heart to supply the artificial lung and the other as the left side of the heart to return oxygenated blood to the recipient.

The pumping circuit must not only be capable of circulating blood at physiological rates and pressures without destroying its constituents, but must also be capable of exact regulation. For this purpose the most satisfactory pump is a rotary type without valves in which an elastic tube, through which the blood passes, is squeezed within the pump by the action of a roller. The decision to employ this type of pump was made after comparison with many forms of the Dale-Schuster pump. It has obvious advantages over the latter in simplicity and ease of sterilization, but had a disadvantage in that the action of the roller in squeezing the tube sometimes caused greater haemolysis. Investigation revealed that this was largely due to the scrubbing of one wall of the tube on the other, an effect produced by the severe shearing forces set up by the roller in squeezing the tube. Re-design of the pump tested, a model based on the design of Henry and Jouvet, resulted in a considerable reduction of this effect, which in turn reduced the rate of haemolysis by as much as 70% in some comparisons.

The pump (Figs. 5, 6, 7, and 8) now used consists of a circular housing in which there is clamped a loop of tube. Within this loop is a large single roller free to rotate on ball-bearings which are mounted on a compound driving shaft. This shaft is made up in the following way. A central driving spindle has fixed to it a cam, the centre of which is $\frac{1}{4}$ in. (0.6 cm.) from the true centre. On this cam is carried an outer cam the centre of which is also $\frac{1}{4}$ in.

from the true centre. This outer cam is free to rotate about the inner one, but may be locked in any position. Adjustment of the outer cam on the inner one provides any degree of throw from 0 to $\frac{1}{2}$ in. (1.3 cm.). In operation the outer cam is adjusted so that the roller carried on the shaft just closes the lumen of the tube in the pump. Then rotation of the driving spindle carries the roller around the

arterial pump is working, and one measures the pressure at which the venous pump is working. The signals from the three manometers are used to form a pattern of servo-loops which result in automatic adjustment of the pumping speed.

The most important function of the control is to ensure that the volume of blood in the machine is constant within narrow limits under varying conditions. As no two pumps of the type employed can be relied upon to deliver constantly matched outputs, it is essential that the rates should be controlled from the blood volume within the artificial circulation, as this will give a true indication of any imbalance between the two. For this reason the signal from the volume indicator manometer is used to induce the two pumps to work at such speeds that they will deliver the same blood flow.

The signal derived from the arterial pressure manometer can be used either to adjust automatically the rate of circulation to such a value that it maintains the blood pressure at a fixed level or as a limiting device to slow down or stop the circulation in the event of the pressure rising above a limit. Since the rate of circulation ultimately depends on the venous return available and as any excessive suction from the venous pump merely collapses the veins, it is clear that the suction pressure must play an important part in the controlling system. The signal from the venous suction manometer performs the same function in the control of the venous pump as the arterial pressure signal does for the arterial pump—that is, it tends to reduce the speed of pumping as the suction pressure rises, and will bring the circulation to a stop when a limit is reached. Hence a compromise is established

—the two pumps working to maintain the blood volume and the arterial pressure, and to prevent the venous suction becoming too great.

This control gear, with its indicating instruments, warning lights, and switches, is housed in a separate trolley. On the control panel of this trolley are arranged the manual and automatic speed-control selectors with meters indicating the rate and pressure at which each pump is working. Another meter indicates the volume of fluid in the oxygenator. Warning lights are placed above the pressure gauges and the fluid volume meter to draw attention to any variations that exceed chosen limits.

Working of the Pump-oxygenator

On the main trolley the pumps and oxygenator are arranged so that easy access to them is possible. The oxygenator lies on rollers in a recess in the top of the trolley and is covered by a transparent hood. Within this recess a constant temperature can be maintained by warm air, heated by infra-red lamps and circulated by a fan under thermostatic control. Oxygen and carbon dioxide are delivered to the oxygenator through rotameters which allow adjustments of flow rate to maintain the appropriate carbon dioxide tension. The gases pass through a warming, filtering, and humidifying system and then traverse the oxygenator in the opposite direction to that of the blood. A device

FIG. 5

FIG. 6

FIG. 5.—One of the pumps unassembled, with the roller in a concentric position within the circular housing. FIG. 6.—The roller moved to a fully eccentric position.

FIG. 7

FIG. 8

FIG. 7.—A loop of tubing in position between the pump housing and the roller, which is in the concentric position. FIG. 8.—The pump fully assembled. The tube is clamped within the pump housing and is compressed by the roller, which has been adjusted to a degree of eccentricity sufficient to close the lumen of the tube.

inner wall of the pump housing, causing it to squeeze the tube in such a manner that a pumping action is obtained. Adjustment is simple and rapid, no blood comes into direct contact with any moving parts, and a pulsatile flow is obtained.

Control Mechanism

The stroke volume depends only on the bore of the tube, and in the pump used this may be varied from $\frac{1}{8}$ to $\frac{1}{2}$ in. (0.3 to 1.3 cm.). Both pumps are driven by fractional horsepower D.C. shunt wound motors through reduction gearing and belt drive. These motors are of special type and are so designed that their speed may be controlled over a 10:1 range by magnetic amplifiers (English Electric $\frac{1}{4}$ h.p. "magamp"). These amplifiers have no thermionic valves in them and are almost mechanical in operation, gaining thereby in reliability and stability. They can be controlled by hand, and such adjustment will enable the circulation to be operated under direct control. However, a rather more sensitive control is achieved by an electronic apparatus designed to replace the delicate balancing mechanism of the animal. This control gear takes as its guide three quantities measured continuously by electrical manometers. These are of the differential-transformer type and are particularly stable. One measures the volume of blood in the artificial circuit from the depth of blood in the collecting chamber of the oxygenator, one measures the pressure at which the

for automatically controlling the gas mixture from the pH of the blood in the machine is under construction, but is not at present fitted to the machine.

All the tubes enter and leave the oxygenator at one end of the trolley in close proximity to the two pumps (Fig. 9). Connecting blocks at this end of the trolley facilitate the various junctions in the tubing. The arterial block acts both as a coarse filter and as a bubble trap from which bubbles may be returned to the oxygenator through a by-pass tube, which passes through a clamp controlled by a solenoid operated from the control panel. This clamp also acts as a relief valve in the event of a sudden blockage in the arterial line. Connexion to the arterial manometer is also made from the arterial connecting-block.

FIG. 9.—The general arrangement of the oxygenator, pumps, manometers, and tubing connexions. 1, Gas entry. 2, Oxygenator level manometer. 3, Oxygenator. 4, Venous pump. 5, Solenoid clamps. 6, Venous manometer. 7, By-pass tube. 8, Arterial filter block and manometer. 9, Arterial pump. 10, "Artery." 11, "Vein."

From the venous connecting-block a tube goes to a reservoir from which the machine is charged and from which transfusions may be made. The tube passes through another electrically operated clamp controlled from the control panel. A second tube connects the venous line to the venous suction manometer.

The tubing in the circuit is of "portex" P.V.C. (polyvinyl-chloride), and is used but once. Before use it is treated with Dow Corning "dimethyl-dichlorosilane" to ensure that the surface is completely non-water-wettable. The arterial catheters are fashioned from this tubing, which is softened by immersion in boiling glycerin and drawn out into a convenient taper, after which rapid cooling fixes the tube in this shape. It can then be hardened by immersion in ethyl ether for 24 hours.

"Polythene" or portex is equally suitable for the venous catheter. Numerous side holes punched in this tube provide efficient collection of blood from the thin-walled veins by gentle suction.

Results of Tests

The machine is efficient, reliable, and comparatively easy to operate. *In vitro* tests confirm that it adequately fulfils the design criteria.

The volume of blood required to charge the machine has been varied between a minimum of 0.5 litre and a maximum of 2 litres without affecting the efficiency of operation.

When stored blood was used it was possible to introduce approximately 100 volumes of oxygen into each litre of blood flowing through it. At 2 litres a minute an average figure is 205 c.cm. of oxygen uptake into stored blood. With

fresh blood circulating at 900 ml. a minute, the difference in oxygen saturation between blood entering and leaving the oxygenator was 174 volumes per litre, indicating an uptake of 156 c.cm. of oxygen a minute. At this time the oxygenator was rotating at 60 revolutions a minute.

The average amount of haemolysis in 500 ml. of stored blood circulated for two hours at rates of up to 2 litres a minute was 0.1 g. Hb per 100 ml. When fresh blood was circulated at a rate of approximately 1 litre a minute the haemolysis was rarely detectable to the naked eye and did not exceed 0.05 g. Hb per 100 ml. an hour.

Foaming is not a serious problem, though it will occur if many air bubbles are carried into the oxygenator in the ingoing blood. This foam does not affect the rate of oxygenation, but may increase the rate of haemolysis. It can be prevented by ensuring that the tubes on the suction side are air-tight. If a film of Dow Corning "antifoam A" is deposited over the interior of the machine no foaming occurs whatsoever in spite of injecting foam into the oxygenator.

The servo mechanism controlling the operation of the pump will maintain the blood volume within 50 ml. over a speed range between 300 ml. and 3 litres a minute. It will adjust the pumping rate in response to the pressure manometers, and when one pump is suddenly stopped will bring the other to a standstill within a volume change of 150 ml. The electronic mechanism possesses remarkable stability, and during trials extending over several days, when the apparatus was called upon to repeat frequently, without adjustment, an elaborate series of tests, it proved faultless. The method of construction allows various forms of servo control to be employed by altering the interconnexions between the various units.

Cleaning is effectively carried out after dismantling the apparatus, including all the washers and plates within the oxygenator. It has been found more economic to replace the tubing after each perfusion, though this is not absolutely necessary. As the entire surface with which the blood is in contact is of inert plastic material and the "intima" non-water-wettable, very little deposition of fibrin and other proteins occurs. This property also reduces the amount of anticoagulant required to maintain the fluidity of the blood. At present the oxygenator must be sterilized with chemical agents, for perspex deforms at 100° C. However, this material is being replaced by nylon or teflon, when it will be possible to autoclave in one unit everything in contact with blood.

An extensive series of animal experiments has amply confirmed that the performance under these conditions is in no wise different from that *in vivo*, and justifies the further development of this prototype heart-lung machine.

I am indebted to Professor C. Crafoord and Dr. V. O. Björk for their kindness during my visit to Stockholm in 1949, when I greatly benefited from their experiences in this field. Also I wish to acknowledge the important contributions of the many who have helped in the design and construction of this machine, particularly Mr. R. L. G. Rainbow, Mr. C. Lordon, and Mr. J. Neal, of the Postgraduate Medical School, Mr. Schranz, of the Genito-Urinary Co., Ltd., and Mr. Kellerman and Mr. Brown, of New Electronic Products Ltd. The design of the control gear was carried out by Mr. Brown, who will describe the mechanism in detail elsewhere. Patent applications have been filed and assigned to the National Research Development Corporation.

BIBLIOGRAPHY

- Björk, V. O. (1948a). *Lancet*, 2, 491.
 — (1948b). *Acta chir. scand.*, Suppl. 137.
 Clark, L. C., jun., Gollan, F., and Gupta, V. B. (1950). *Science*, 111, 85.
 — Hooven, F., and Gollan, F. (1952). *Rev. sci. Instrum.*, 23, 748.
 Crafoord, C. (1949). *Surg. Gynec. Obstet.*, 89, 629.
 De Bakey, M. E. (1934). *New Orleans med. surg. J.*, 87, 386.
 Dennis, C., Karlson, K. E., Eder, W. P., Nelson, R. M., Eddy, F. D., and Sanderson, D. (1951). *Surgery*, 29, 697.
 — Spreng, D. S., jun., Nelson, G. E., Karlson, K. E., Nelson, R. M., Thomas, J. V., Eder, W. P., and Varco, R. L. (1951). *Ann. Surg.*, 134, 709.
 Dodrill, F. D., Hill, E., and Geisch, R. (1952). *J. thorac. Surg.*, 24, 134.
 Fischer, H. W., Albert, H., Riker, W. L., and Potts, W. J. (1952). *Ann. Surg.*, 136, 475.
 Gerst, G. R., Grossman, J., and Kantrowitz, A. (1951). *Science*, 114, 258.

- Gibbon, J. H., jun. (1937). *Arch. Surg., Chicago*, 34, 1105.
 — (1939). *Surg. Gynec. Obstet.*, 69, 602.
 Helmsworth, J. L., Clark, L. C., jun., Kaplan, S., Sherman, R. T., and Largent, T. (1952a). *J. thorac. Surg.*, 24, 117.
 — (1952b). *J. Amer. med. Ass.*, 150, 451.
 Jongbloed, J. (1949). *Surg. Gynec. Obstet.*, 89, 684.
 — (1950). *Ned. T. Geneesk.*, 94, 161.
 — (1951). *S. Afr. med. J.*, 25, 840.
 Kantrowitz, A., Hurwitz, E., and Kantrowitz, A. (1951). *Arch. Surg., Chicago*, 63, 604.
 — and Kantrowitz, A. (1950). *Proc. Soc. exp. Biol., N.Y.*, 74, 193.
 Karlson, K. E., Dennis, C., Westover, D., and Sanderson, D. (1951). *Surgery*, 29, 678.
 Kolff, W. J. (1946). *The Artificial Kidney*. Kok, Kampen, Holland.
 — and Dubbelman, C. P. (1949). *Geneesk. Gids.*, 27, 241.
 Lorincz, A. L., Jacoby, J. J., and Livingstone, H. M. (1948). *Anesthesiology*, 9, 162.
 Miller, B. J., Gibbon, J. H., jun., and Gibbon, M. H. (1951). *Ann. Surg.*, 134, 694.
 Mustard, W. T., and Chute, A. L. (1951). *Surgery*, 30, 684.
 Peirce, E. C., and Southworth, J. L. (1953). *Arch. Surg., Chicago*, 66, 218.
 Potts, W. J., Ricker, W. L., DeBord, R., and Andrews, C. E. (1952). *Surgery*, 31, 161.
 Rossi, R. (1950). *Minerva med., Torino*, 41, 835.
 Sewell, W. H., jun., and Glenn, W. W. L. (1950). *Surgery*, 28, 474.
 Southworth, J. L., Peirce, E. C., Tyson, T., and Bowman, R. L. (1953). *Arch. Surg., Chicago*, 66, 53.
 Stokes, T. L., and Flick, J. B., jun. (1950). *Proc. Soc. exp. Biol., N.Y.*, 73, 528.
 — and Gibbon, J. H., jun. (1950). *Surg. Gynec. Obstet.*, 91, 138.
 Thomas, J. A., and Beaudouin, P. (1951). *J. Physiol., Paris*, 43, 311.
 Tosatti, E. (1949). *G. Ital. Anest.*, 15, 120.
 Wesolowski, S. A., Miller, H. H., Halkett, J. A. E., and Welch, C. S. (1950). *Bull. New Engl. med. Cent.*, 12, 41.

EXPERIMENTAL PHYSIOLOGY OF A HEART-LUNG MACHINE IN PARALLEL WITH NORMAL CIRCULATION*

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Recognizing the magnitude of the task involved in the mechanical maintenance of the entire circulation in humans, we turned our attention to the lesser problem of assisting the natural circulation by shunting part of it from vein to artery through the heart-lung machine. To assess the value of the heart-lung machine in resuscitation and in the treatment of a failing circulation, and to acquire information to be applied later to the problem of complete artificial maintenance of the circulation in man, a series of experiments was carried out in dogs.

In the planning and execution of these experiments we were helped by experience gained from a large number of studies carried out with earlier machines. These studies included the perfusion of isolated organs, which proved particularly valuable in providing the basis of the present technique, and enabled controlled conditions to be established.

The Experiments

Thirty greyhound dogs, each weighing between 20 and 30 kg. and apparently healthy, were used. They were dewormed and given intramuscular penicillin (300,000 units 8-hourly) for two days prior to the day of an experiment, when each dog was anaesthetized with intra-

peritoneal pentobarbitone sodium (30 mg. per kg. body weight). A cuffed tube was introduced into the trachea and inflated, and was then connected to a Benedict-Roth spirometer containing oxygen. Electrocardiograph leads were fixed to the two fore-limbs and to the right hind-limb.

A small incision was made in the neck, and an external jugular vein and a carotid artery were exposed. A radio-opaque catheter was then introduced into each of these vessels and was guided under fluoroscopic control into the right auricle and the ascending aorta or left ventricle respectively. The catheters were connected to continuously recording manometers and arranged so that blood samples could be obtained from them. The patency of each was maintained by flushing with saline which contained heparin (2 mg. per 100 ml.).

One or other deep femoral artery was exposed together with one of the major branches of a femoral vein. A tapered plastic catheter was then introduced into the deep femoral artery and connected to the "arterial" tube of the heart-lung machine. A plastic catheter 25 cm. in length, with numerous side openings, was passed into the femoral vein and advanced until its tip was judged to rest within the thoracic portion of the inferior vena cava. Connexion was made between this catheter and the "venous" tube of the heart-lung machine.

Concurrently with the preparation of the animal the machine was being assembled and made ready for perfusion (see Appendix I). In the first five experiments it was charged with approximately 750 ml. of donor dog's blood, but subsequently, because facilities for sterile blood collection proved inadequate and no definitive method could be achieved for cross-matching mongrel and greyhound blood, a solution of 6% dextran in normal saline was used.

The animals were now given heparin or dextran sulphate (Walton, 1951), 1-2 mg. per kg. being used. Primary readings were then made of the arterial and central venous pressures, the oxygen consumption, and the electrocardiograph, which were then recorded con-

Fig. 1.—Red-cell count, haematocrit, and plasma haemoglobin in 10 dogs before, during, and immediately after one hour's perfusion.

*This work was carried out at the Buckston Browne Research Farm, Kent, by permission of the President and Council of the Royal College of Surgeons of England.