Developmental Rationale:

A Nonpulsatile Artificial Heart

ABSTRACT

The concept of a nonpulsatile artificial heart is discussed. The approach used in this investigation embodies the principle of a constrained force vortex to propel the blood. This novel approach eliminates the need for the direct contact of blood with impellers and also alleviates the constrictive and mechanical forces on blood cells resulting from high levels of shear and turbulence as well as direct impaction. By eliminating the direct contact of blood with these mechanical forces and with the use of atraumatic and nonthrombogenic low temperature isotropic carbon as the material of construction, a largely atraumatic pump is made possible. The use of force vortices enables the investigators to retain the high efficiency of a centrifugal impeller pump without encountering the traumatic effects of direct impeller contact. The result is a highly efficient, atraumatic, nonpulsatile blood pump. The possible effects of nonpulsatile flow in the body are reviewed. Pump optimization is discussed. Finally, implantation of the artificial heart pump into animals on an experimental basis is discussed briefly.

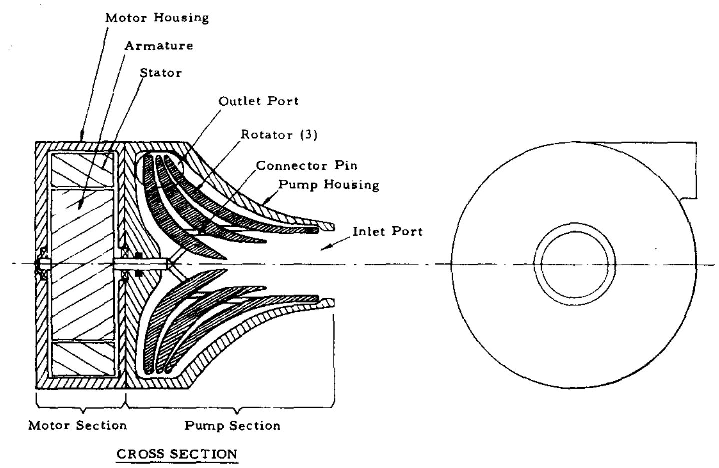


Figure 1. Cross Section of Optimized Force Vortex Artificial Heart

By
Edson H. Rafferty, B.S.M.E.,
Harold D. Kletschka, M.D.
and
Douglas A. Olsen, Ph.D.
Bio-Medicus, Inc.
15308 Minnetonka Industrial Road
Minnetonka Industrial Park
Minnetonka, Minnesota 55343

Part Two

DEVELOPMENT OF NONPULSATILE HEART PUMPS

Nonpulsatile pumps of various designs have had wide industrial and commercial application because of their high operating efficiency. These devices are the most efficient pumps available in terms of converting available energy to useful work. One such pump which has been evaluated for use as an artificial heart[28] was discarded because of the destructive effects on blood produced by the high shear rates, turbulence and cavitation. These observations were based on results obtained using a pump intended for commercial application.

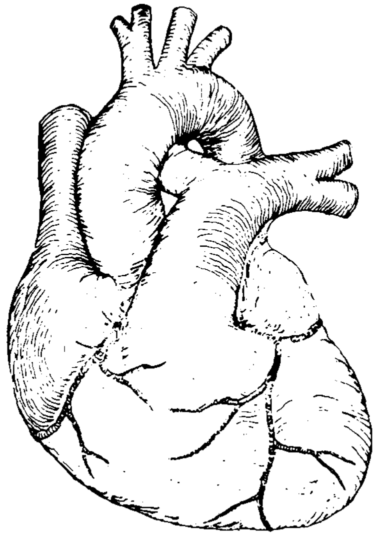
Another research group has recently reported[29] engineering and biological testing of several centrifugal-type pumps. However, they have been most successful with impeller-type pumps which have the inherent disadvantages previously mentioned with regard to shear and turbulence.

Rafferty, Kletschka, et al., [30-36] by careful selection of design criteria based on operating in a biological environment, have successfully developed atraumatic nonpulsatile pumps. These pumps exhibit performance analogous to the mammalian heart with respect to the very important parameters of peripheral auto-regulation of the blood flow which has been described [28-36] as one of the most important features of nonpulsatile artificial hearts.

One such pump, operating on what is termed a constrained force vortex principle,[30] has been studied in detail by Rafferty and Kletschka.[30,31] Details of the constrained force vortex pump are shown in Figure 1. The operating principles of the pump are as follows: Fluid enters centrally and is accelerated by contact with rotating disk shaped plates. The fluid, having been accelerated by these rotators, imparts its acceleration to adjacent fluid with a minimum of shear and turbulence. Thus, the fluid, in a sense, becomes its own prime mover. This has been found to be an atraumatic method for handling blood. It is also an extremely efficient method of pumping.[30]

Much work has been done to assess the physiological implications of nonpulsatile versus pulsatile flow [28, 30-34, 37-70] This subject is discussed in more detail below. At this point, however, it can be noted that although some studies indicate there may be significant differences, there is no conclusive evidence to date that pulsatile flow has physiological advantages over nonpulsatile flow. Should nonpulsatile flow be demonstrated to be physiologically acceptable, it will be possible to take advantage of the high efficiencies of pumps producing this type of flow.

If, however, pulsatile flow is demonstrated to be advantageous or necessary, it would be relatively easy to convert a constrained force vortex pump to produce pulsatile flow while retaining the efficient nonpulsatile character of the basic pumping unit to do the bulk of the pumping work. Such a pumping system would offer the additional advantage of requiring no valves.[42]



Artificial Heart (cont.)

PULSATILE VERSUS NONPULSATILE FLOW

A number of investigations have been conducted to evaluate the physiological effects of nonpulsatile versus pulsatile flow. Early work by Hooker[51] and Gesell[52] indicated that nonpulsatile flow had various adverse physiological consequences. This work, however, was done with somewhat artificial physiological experimental models.

In recent years, Goodyear and Glenn[53] conducted experiments in which they removed the pulse from the arterial supply to the kidney and found no change in mean arterial pressure. The excretion of water and of electrolytes, and renal clearance of inulin and para-aminohippuric acid remained essentially unchanged.

Selkurt[54] and Ritter[55] carried out studies which showed no significant change in renal blood flow or renal function when the kidney was perfused at normal mean pressures using flows with reduced amplitude of pulsation.

Parsons and MacMaster[56, 57] conducted experiments in which they concluded that the arterial pulse played a role in the dynamics of fluid circulation. However, this was done by noting the rapidity of spread of a dye injected into the tissues of the rabbit ear. As pointed out by those authors, the spread of the dye is not necessarily related to lymph flow. Experiments performed by Rafferty and Kletschka indicate that when the intact hind limbs of dogs were perfused with a nonpulsatile flow, there was no evidence of lymphostasis.[58]

Perhaps some of the best experiments assessing the implications of nonpulsatile flow were those carried out by Wesolowski, et al. [59-64] which culminated in the statement that "the presence of the pulse may represent nothing more in physiological terms than an expression of the limitations of organic heart design."[64] They found that perfusion of the lesser circulation with a nonpulsatile flow had no deleterious effects.

Wesolowski carried out experiments in which he maintained the systemic circulation with nonpulsatile flow for periods up to six hours. From the results of this investigation it could be said that the entire circulation appeared to be maintained equally well by the use of either pulsatile or nonpulsatile flow for this period of perfusion.[64]

During the nonpulsatile perfusion experiments, observations were made regarding the maintenance of central nervous system functions as evidenced by progressive lightening of the depth of anesthesia. Repeated doses of nembutal were required to keep the animal anesthetized to the bypass needs. Nembutal requirements were retained throughout the six-hour experiment. In the nonpulsatile perfusion experiments, most of the hearts continued to beat in a coordinated manner.

Results of the tests of phenosulfonphthalein excretion suggested there was no disturbance in renal function during the nonpulsatile experiments. Wesolowski, et al. noted that the recovery of the animals was remarkable and that on recovery from anesthesia the animals could walk and eat well by the first post-operative day.

They also studied the vascular reactivity by noting the reaction to injection of intraarterial neosynephrine and acetylcholine. Neosynephrine led to a generalized rise in pressure and administration of acetylcholine led to hypotension. They concluded from these findings that the total peripheral resistance remains unaltered by nonpulsatile flow. The experiments of Wesolowski and his group entailed the removal of the heart and lungs of the animal being perfused so that it seemed to be "ultimate" in terms of studying the effects of nonpulsatile flow on the total animal.

Indirect evidence also points to the acceptability of nonpulsatile flow in the intact animal. For example, so-called pulseless disease is known to be compatible with long-term survival.[71] Likewise, coarctation of the aorta may result in a markedly diminished pulse to the lower half of the body. However, even this has been found to be compatible with a long life span.[71-72]

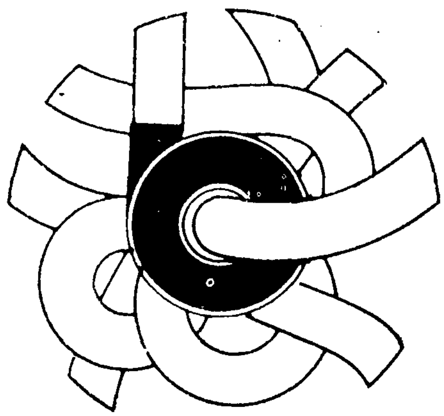
Severe aortic stenosis with a markedly diminished pulse pressure is also compatible with long-term survival.[71] Chronic obliterative arterial diseases of various extremities have been recorded on numerous occasions in which there is no pulse pressure distal to the obstruction. This situation has been proven compatible with tissue and organ survival.[71] Any difficulties that may result appear to be related to deficient circulation rather than to the nonpulsatile characteristics of the flow.[73]

Hall[66] and Liotta[67] have stated that they were able to maintain nonpulsatile circulation to the lungs for periods up to 30 days without observable adverse effects. Kletschka and Rafferty have been involved in several experiments in which all flow distal to the aortic arch was depulsed.[58] No alterations were detected which signified adverse consequences of this type flow as measured by urine output, urine specific gravity, albumin in the urine, glucose in the urine, lymphostasis in the hind limbs or in the abdominal viscera, and by the general appearance of the animal.

Kolff has stated that he knows of no evidence which shows that nonpulsatile flow is physiologically incompatible with life.[65] The validity of any of the experimental studies that have been done to date can be questioned on the basis of their experimental design.

It should also be noted that certain types of extracorporeal pumps used for open heart operations generally produce an essentially nonpulsatile flow. Total heart bypass using this type pump is compatible with recovery.

Finally, it is of interest that the body normally converts the pulsatile flow of the primary portion of the arterial tree into a nonpulsatile type flow as it perfuses the capillaries where metabolic exchanges principally take place.



Artificial Heart (cont.)

PUMP OPTIMIZATION

In the absence of a coherent theory which can predict to a reasonable degree the mechanical and biological performance of a pump which incorporates the primary design parameters given below, we have undertaken an experimental program aimed at systematically evaluating the overall performance characteristics of impellerless Rafferty-Kletschka blood pumps of various designs. Our approach is presented in the sections to follow.

For any cardiac replacement or assist device, we can define three fundamental performance or operating constraints which must be satisfied:

1) flow rate
2) mean pressure
3) blood destruction rate

Typically, a normal human heart supplies blood to the systemic circulatory system at a rate of approximately six liters/minute and at a mean pressure of about 100 mm Hg. Blood pressure and flow requirements in the case of nonpulsatile delivery cannot be assessed until long-term in vivo tests have been conducted and evaluated. In the absence of this information, we can assume that the average values of flow and pressure given above will be adequate to satisfy normal physiological demands.

Ideally, a blood pump should be designed so that it can operate without inducing blood damage. In practice, it has been found that animals undergoing prolonged periods of mechanical circulatory support demonstrate post-perfusion anemia and decreased red blood cell life span which appear to be related to the degree of mechanical trauma to which the blood is subjected.[74]

In these instances, erythrocyte hemolysis, as determined from plasma hemoglobin concentration, was taken as a measure of mechanical trauma. One may define an acceptable rate of hemolysis as that which occurs at a rate not exceeding the body's ability to absorb or remove plasma hemoglobin.

Bernstein, et al,[74] have experimentally established tolerance levels to chronic hemolysis in dogs by continuous intravenous infusion of free hemoglobin solution. It appears that dogs are able to tolerate indefinitely the chronic infusion or release of 0.1 gm Hb per 100 liters of blood pumped. This index of hemolysis (grams of hemoglobin released per 100 liters of blood pumped) of 0.1 may serve as a maximum acceptable rate of hemolysis in man.

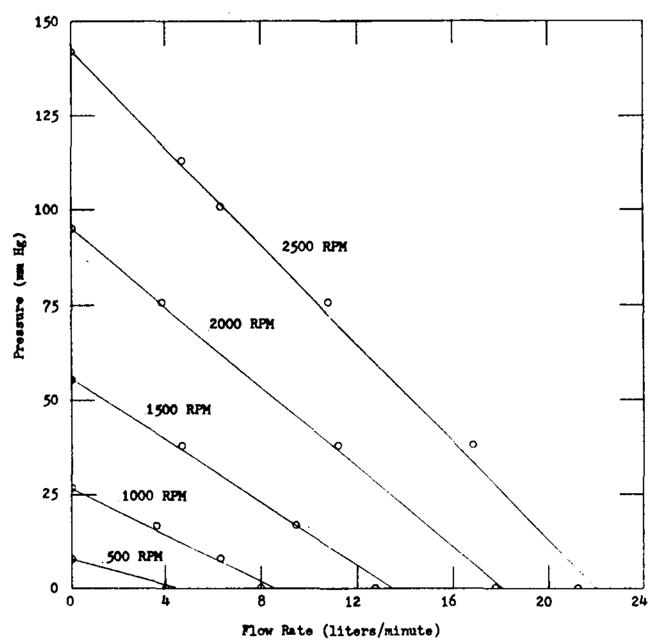
Simply stated, therefore, we have attempted to optimize the design of an impellerless pump capable of delivering six liters of blood/minute at a pressure of 100 mm Hg and with an index of hemolysis (I.H.) not exceeding 0.1.

The Rafferty-Kletschka constrained force-vortex artificial heart consists essentially of rotating plates (rotators) within a housing as illustrated previously in Figure 1. Blood is pumped as a result of the development of centrifugal forces created by the rotary motion.

This device utilizes the viscous properties of real fluids and imparts an acceleration to the blood through shear forces resulting from the relative velocity difference between the rotors of the pump and the fluid. Hence, the shear force is a function of the fluid viscosity and shear rate. After the blood enters the pump, it is pressurized and flows toward the periphery of the pump where it is discharged. Typical performance curves using a glycerin-water solution as a blood simulant are shown in Figure 2.

The performance characteristics of the blood pump will depend primarily upon the following design parameters or variables:

1) rotator shape (curvature)
2) number of rotators
3) spacing between rotators
4) angular velocity of rotators
5) inlet and outlet radius
6) cross-sectional area between rotators at the various radii
7) surface properties of rotators
8) housing and diffuser configuration



Flow Rate (1/min)

Figure 2. Pressure versus Flow Rate for 2½ inch diameter three rotator Constrained Force-Vortex Pump with a 42% by volume solution of glycerine in water.

As a first step toward optimizing the pump, the relative importance of each of these parameters and their interdependence in determining the overall performance of the pump must be accurately assessed. It was expected, for example, that decreasing the spacing between the rotators would improve the transfer of

Continued on Page 30

Artificial Heart (cont.)

kinetic energy from the rotators to the blood and might also decrease the rate of hemolysis as a result of lower shear rates to which the blood would be subjected. On the other hand, for a given flow rate, narrower spacing would tend to increase the frictional losses associated with the radial flow through the pump.

Shear force pumps which use parallel flat disks as rotators have been investigated theoretically and experimentally.[75-79] In these studies emphasis has been placed on the relationship between flow rate and profile, pressure head and pump efficiency for various disk radii (inlet and outlet), spacing, angular velocity and fluid kinematic viscosity.

Hasinger and Kehrt[78] calculated that, for laminar flow, the optimum rotor efficiency occurred when the ratio of radial flow velocity to disk velocity evaluated at the periphery of the pump was 0.02 and when the distance between the disks, d, angular velocity of the disks, ω, and the kinematic viscosity of the fluid, ν, satisfied the relation:

$$\frac{d}{2}\sqrt{\frac{\omega}{\nu}} = \frac{\pi}{2}$$

This analysis predicts rotor efficiencies greater than 60% for outer to inner disk radius ratios of 2 to 5. Experimental results were also reported and showed an efficiency of 54% for rotors with a radius ratio of 4.

The present authors have found that to design an efficient shear-force pump, the following dimensionless parameters should be made as small as practical:

$$\alpha_0 = \frac{Qd}{\nu r_0^2}$$

$$\varphi_0 = \text{(Flow Coefficient)} = \frac{Q}{2\pi d \omega r_0^2} = \frac{\alpha_0 \nu}{2\pi d^2 \omega}$$

$$\frac{r_i}{r_0} = \text{(inner/outer radius ratio)}$$

where Q is the pump volumetric flow rate, r_0 and r_i are the outer and inner disk radii respectively, and the remaining notation is as above.

Variations in r_0 and Q will alter α_0 and φ_0 to the same degree while a change in disk spacing will increase one of these dimensionless parameters while having the effect of decreasing the other. The angular velocity is essentially a free variable in the sense that only the flow coefficient is affected by changes in ω.

It is possible to reduce α_0 by lowering the disk spacing without changing the flow coefficient if the disk speed is increased by increasing the number of disks to the point at which the additional transfer and frictional losses outweigh the resultant increase in pump head. These results are based on the assumption that fully developed boundary layer flow exists between the disks. That is, both the radial and tangential flow profiles are considered parabolic.

Breiter and Pohlhausen[79] indicate that the validity of this assumption may be in doubt when the parameter $\frac{d}{2}\sqrt{\frac{\omega}{\nu}}$ is larger than about 4. Considering a spacing of ¾ mm, angular velocity of 3000 rpm and a kinematic viscosity of 4 centipoises, this parameter is approximately 3.2. It may be that the boundary layer assumption is fully justified in this case.

It should be emphasized here that optimum mechanical performance (e.g., efficiency) of a pump will not necessarily be compatible with biological constraints (e.g., blood trauma) imposed on the operating characteristics of the pump. Obviously, trade-offs among the design variables and perhaps significant sacrifices in overall mechanical performance of an impellerless blood pump must be made to ensure that excessive trauma or damage to the blood does not occur.

To be concluded.

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