# MAGNETIC BEARING SYSTEM FOR AN ARTIFICIAL HEART

**Michael J. Baloh,[1] Paul E. Allaire,[2] Edgar F. Hilton,[2] Naihong Wei,[2] Eric H. Maslen,[2] Daniel Baun,[2] Ronald D. Flack,[2] Don B. Olson,[3] Gill B. Bearnson,[3] Pratap S. Khanwilkar[4]**

**Abstract**

This paper describes a prototype continuous flow blood pump (CFVAD3) utilizing magnetic bearings. These bearings support a thin pancake impeller whose shape allows for a very compact pump with total axial length below 5cm and with a diameter of about 10cm. This gives a total volume of about 275cc. The impeller itself has 4 vanes with a designed operating point of 6 *l/min* at 100 *mmHg* of differential pressure and 2000 rpm. The advantages of magnetic bearings, such as large clearance spaces and zero mechanical wear, are elaborated upon. Bearing parameters such as load capacity and current gains will also be described. Another purpose of this article will be to present the results of bearing performance measurements in a simulated adult human circulation system.

## 1 Introduction

The development of artificial hearts has preceded through several stages. Initially, these devices attempted to mimic the natural pulsatile operation of the heart. This was accomplished by the use of flexible membranes and pneumatic pumps. However, the pumping diaphragm eventually failed because of small cracks that formed on its surface. These failure points then acted as seeding points for clot formation, and ultimately death.

In recent years, researchers have focused on rotating pumps because of their inherent simplicity. Usually, rotating pumps are classified into two distinct families, axial and centrifugal types. Each method has both benefits and drawbacks. Axial flow pumps are typically tubular in shape and are much smaller then a centrifugal pump (Butler, 1992). Unfortunately, these devices have encountered thrombi formation about the blood bearing or many failures of the blood lubricated bearings (Allaire et al., 1996;Jarvik, 1995;Belenger and Knight, 1983). Centrifugal flow pumps are more energy efficient. In addition, they operate at much lower speed and may be more blood friendly. Even with these advantages, conventional centrifugal pumps still rely on mechanical bearings lubricated by blood (Akamatsu and Itoh, 1992).

Therein lies the design challenge: how to support a rotating pump while avoiding the difficulties associated with ordinary bearings. CFVAD3 approaches this problem by employing magnetic bearings. By this method, the impeller remains suspended within the pumping housing without any point-to-point

[1]Electrical Engineering Department, University of Virginia, Charlottesville, VA, USA, e-mail: mjb5s@virginia.EDU.
[2]Mechanical, Aerospace and Nuclear Engineering Department, University of Virginia, Charlottesville, VA, USA, e-mail: pea@virginia.edu, efh4v@virginia.EDU, nw4n@virginia.edu, ehm7s@virginia.edu, dan@romac1.mech.virginia. edu, respectively.
[3]Artificial Heart Laboratory, University of Utah, Salt Lake City, UT, USA, e-mail: Don.B.Olsen@m.cc.utah.edu and bearnson@ee.utah.edu, respectively.
[4]MedQuest Products, Inc., Salt Lake City, UT, USA, e-mail: Pratap.K@m.cc.utah.edu.

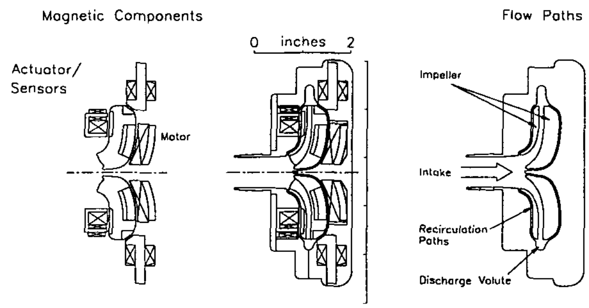


Figure 1: Profile of CFVAD3 prototype

contact. Large clearances between the rotating impeller and stationary housing allow for continuous blood washing of the interior surfaces which prevents thrombi formation. In this fashion, both mechanical wear and blood trauma associated with bearings are eliminated. As a consequence, total cost of implementation reduces because of expected lifetimes of 10-15 years (Allaire et al., 1996; Hogness and Van Antwerp, 1991).

### 1.1 Technical Challenges

The principal technical challenges in the construction of these devices include:

1. Keeping the power consumption low enough so that blood flow through the pump can easily expel excess heat and permitting mobile operation from a battery with suitable charge–to–charge cycle time;

2. Providing the required flow and pressure in a package which is sufficiently compact to permit implantation;

3. Designing the internal flow to prevent excessive blood recirculation, stagnation, and mechanical trauma which lead to cell death and/or clotting;

4. Designing a flow control system which ensures that the patient's blood pressure and flow rate is appropriate to his or her level of activity;

5. Finding motoring and impeller suspension mechanisms which are reliable enough to provide at least five years of zero maintenance service;

6. Finding materials for the internal surfaces of the device which are both compatible with the mechanical function and not chemically perceptible to the body so as to avoid triggering immune responses.

## 2 General Structure of CFVAD3

The CFVAD3 consists of several subsystems: pump, magnetic bearing, motor, hemocompatible surfaces, and electronic controls. All but the last component are internal to the body. Figure 1 illustrates the pump prototype. Common commercial pumps have a long shaft with an impeller directly mounted to it, since this simplifies the magnetic actuator design. However, CFVAD 2 (Allaire et al., 1996), which was based on this principle, suffered from thrombosis due to long and complicated blood recirculation paths surrounding the impeller. To avoid these issues CFVAD3 needed short smooth blood recirculation paths to prevent blood trauma (O'Hara and Nose, 1994). In response to these needs, the shaft was discarded,

Table 1: CFVAD 2 and 3 Design Specifications

| Specification | CF II | CF III | % Reduction |
|---|---|---|---|
| Length(cm) | 11.7 | 4 | 66 |
| Outside diameter(cm) | 17.5 | 10.4 | 61 |
| Total weight-pump(gm) | 3500 | 1200 | 66 |
| Volume(cc) | 852 | 275 | 68 |
| Rotor length(cm) | 10.5 | 1.8 | 83 |
| Rotor weight(gm) | 365.5 | 180 | 51 |
| Rotor speed(rpm) | 2400 | 2000 | NA |
| Inlet bearing actuators | 2 | 1 | 50 |
| Radial bearing actuators | 2 | 1 | 50 |
| Position sensors | 9 | 0 | 100 |
| Number of wires | 52 | 24 | 55 |

making for a very compact unit. Overall CFVAD3 is quite small with an axial length of about 50mm (2inches) and diameter of about 100mm (4inches) The pancake impeller which resides center most within the pump housing has an axial length of about 19mm (0.75 inches) and diameter of 70mm (2.75inches). See Table 1 for further details.

## 2.1 Magnetic Components

The leftmost image of Figure 1 shows the magnetic actuators which are responsible for maintaining the position of the pumping impeller relative to the pump housing. The impeller has six degrees of freedom – three translations along x,y, and z directions and three rotations about those axes. Thus, to control the impellers motion the magnetic system must be able to completely control each degree of freedom. A circular array of horseshoe electromagnets, Fig 2, directed at the inlet face of the impeller provides a combination of positive axial force and pitching moments, $F_z$, $M_x$, and $M_y$. An eight pole radial stator, Fig 3, with tapered faces directed at the tapered discharge edge of the impeller provides an assignable combination of negative axial force, pitching moments, and both positive and negative radial forces along the radial axes, $F_x$, $F_y$, $F_z$,$M_x$, and $M_y$. Lastly, a motor behind the impeller governs rotation. This motor comprises an array of permanent magnets embedded in the rear of the impeller and an associated three phase stator embedded in the housing.

## 2.2 Flow Paths

As stated earlier, previous work on CFVAD 2 indicated the need for an efficient pumping geometry. This meant short, smooth, and uncomplicated blood recirculation paths that minimize blood trauma through constant blood washing of pump surfaces. Furthermore, consideration was given to the length of recirculation clearances in order to reduce hemolysis caused by the excessive shear forces induced by pumping. The resulting design makes for a more blood friendly environment (O'Hara and Nose, 1994). Figure 1 illustrates the flow paths in the rightmost drawing. As shown, the blood enters an intake cannula where it is accelerated by an impeller. The impeller has a front and rear shroud that improves efficiency and which is required to contain various magnetic materials such as the motor's permanent magnets and the bearing iron targets. Sandwiched between the inlet shroud and back shroud (hub) are four vanes whose design increases the hydraulic efficiency of the pumping action, while minimizing shear stress to blood cells. Having been accelerated by the impeller, blood accumulates in a vaneless diffuser as it exits the rotating impeller via a single discharge cannula.

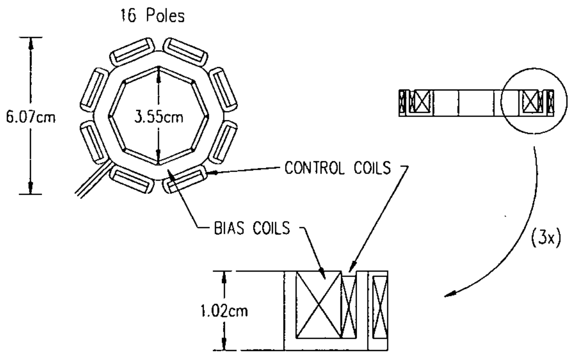


Figure 2: The geometry of inlet actuator (located in front of the impeller)

# 3 Magnetic Actuators

Testing of an earlier prototype (CFVAD 2) not only provided information about blood related concerns, but also about electrical ones. Primarily, the power consumption of former prototype was unacceptably high, previously reported to be 50W. Several factors contributed to these losses: poor integration of bearing actuators resulting in excessive coil resistances; high magnetic flux densities requiring large currents to maintain; and linear current amplifiers inherently power inefficient (Allaire et al., 1996).

Based on this evaluation, the design of CFVAD3 emphasized reducing power consumed by the actuators, and maximizing recycled energy so as to limit demand on power sources. This section explains some of the methods employed in CFVAD3 to reduce bearing power consumption. Since magnetic actuators and power electronics are the primary sources of energy loss in a magnetic bearing, this discussion will confine itself to those subjects. Magnetic actuators are nothing more than copper wire coils wrapped around iron cores with a carefully chosen shape. These coils are non-ideal, and therefore have an effective resistance associated with them. It is this resistance that accounts for the greatest power dissipation in a magnetic actuator. Evidently, to reduce power dissipated in the coils both the resistance and current requirements of the actuator must be minimized

To this end, large amount of effort went into limiting the redundancy of the magnetic bearing system. Clearly, each actuator requires a lossy coil – therefore more actuators equal higher losses. Figure 2 shows the design of the inlet actuator of CFVAD3. The final design of the inlet actuator has four moment coils and one bias coil. The bias coil performs the majority of work, correcting the axial motion of the impeller. The moment coils correct for slight and transient tilting of the impeller. As a result, the moment coils demand very slight amounts of current, while an efficient bias coil requires 1A. The efficiency of the bias coil is due to using heavy gauge wire, as well as a configuration that does not have end-winding. For this reason the total length of wire is reduced by about two-thirds.

Similarly, we have the combined functions of both radial and discharge moment actuators into a single conical actuator near the discharge. This design improvement further reduces power loss by limiting the number of coils, and therefore the amount of wire used. Like the inlet actuator, the largest gauge wire possible was used to reduce $I^2R$ losses. Figure 3 shows the final design. As stated earlier, CFVAD 2 actuators consumed about 50W. A large reason for this was the very high gap flux densities used, about 0.4T. However, at the time CFVAD 2 was designed, the pumping forces where unknown and the impeller more substantial. As a result, the actuator was designed toward operating at high flux densities to achieve greater force capacity. Unfortunately, the losses are a function of the square of the current. So, if the current demand reduces by half, the losses drop by a fourth. Here is another way in which CFVAD3 improves upon its predecessor. With knowledge of pumping forces and a smaller impeller the actuators

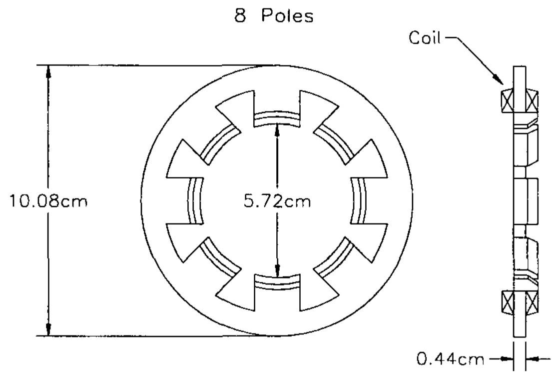


Figure 3: The geometry of conical actuator (located behind of the impeller)

Table 2: Actuator Design Specifications

| | |
|---|---|
| Conical Actuator | |
| Power consumed | 6.4W |
| Max Radial Force | 2 lb-f |
| Slew Rate | 17 klb-f/s |
| Inlet Actuator | |
| Power consumed | 1.8W |
| Max Axial Force | 3.1 lb-f |
| Max Torque | 0.06 lb-f-ft |
| Slew Rate | 8.2 klb-f/s |
| Total Power Consumption | 8.2W |

operate at a bias flux of about 0.1T, or one fourth of the flux density in CVFAD II. Theoretically, this reduces losses by a factor of one sixteenth that of CFVAD 2.

Considering that the total amount of current circulating within the CFVAD3 presently amounts to about nine amps for all actuators, something must be one to allow portability (batteries). For this reason CFVAD3 utilizes custom high efficiency switching amplifiers, with typical on resistances of about 0.02Ω, which pushes efficiency to near 95%.

Table 2 highlights a few of the design specifications of this prototype. Based on measurements of actuator currents the estimated power consumed by the actuators was calculated. The inlet actuators consume the least power, about 1.8W, while the conical actuator consumes about 6.4W. The total wattage of 8.2W will improve once more efficient levitation algorithms are developed.

Other specifications listed are maximum forces and slew rate. Although the forces may seem small, they are more then adequate to control the motion of the rotating impeller up to 8 $g$ force in the axial direction. The slew rates are a measure of the responsiveness of the actuators to force changes.

# 4 Feedback Control

The controller uses sensor feedback information to regulate the actuator current so that the rotor maintains its position within the pump cavity (Hilton et al., 1998). The device must be able to stabilize the

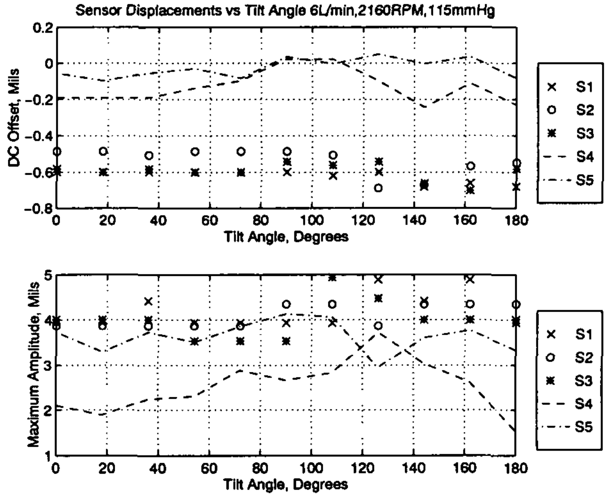


Figure 4: Rotor position as function of orientation

impeller motion given its characteristics and various disturbances. On the face of it, position control of the impeller appears to be relatively simple because the rotor (impeller) is very rigid and should be reasonably well damped by the fluid surrounding it. However, complications arise from two sources. At the nominal operating speed of 2000 RPM, the disk shaped impeller will exhibit some degree of gyroscopic coupling. The second, and more important, complicating influence is that the fluid will produce substantial stiffness, inertia and cross-coupling effects whose magnitude (and sign) will depend on impeller position, rotational rate, and flow rate (Jery et al., 1985). Rather than attempt an extended modeling effort to establish the range of fluid coupling effects, the impeller has been levitated using high gain PID loops along each of the five decoupled modal axes (Hilton et al., 1998). The unmodeled couplings are presently being determined. With this more precise model a far more power efficient coordinated centralized controller becomes possible.

# 5 CFVAD3 Performance

CFVAD3 has been operated in water and blood to test the performance of the magnetic bearing system. In each of these tests, a modally based PID controller regulates five of the six axes of impeller motion. The sixth axis – impeller spin – is controlled by a brushless DC motor. Initially, the impeller occupies a space at the bottom-most part of the impeller cavity. Once the bearing system becomes active, it takes only a fraction of a second for the impeller to approach the center of pump cavity. In this position there are no contact points to cause wear. The impeller will remain in the nominal position until the system is shutdown. Figure 4 shows a typical plot of the measured pump impeller position while operating in water and attached to a mock circulation system. This plot indicates the robustness of the controller to steady flows and pump orientations. For these tests, operation of the pump inlet cannula started in the upward vertical position $0^o$ and progressed until the inlet cannula faces downward $180^o$, while pumping $6\,l/min$ at a differential pressure of $115\,mmHg$ and $36\,Hz$ (2160 rpm). Figure 4 shows the DC offset (steady state position under impeller weight loading) of the impeller relative to the centered position in axis five with a peak value of -0.6 mils, compared to a clearance around the impeller of 30 mils. Here S1 and S2 are the radial impeller motions, S3, S4, and S5 are the axial and tilting motions. The ratio of maximum impeller DC displacement to clearance of 1 to 50. It can be seen that the impeller is very well centered under static loading and various pumping orientations.

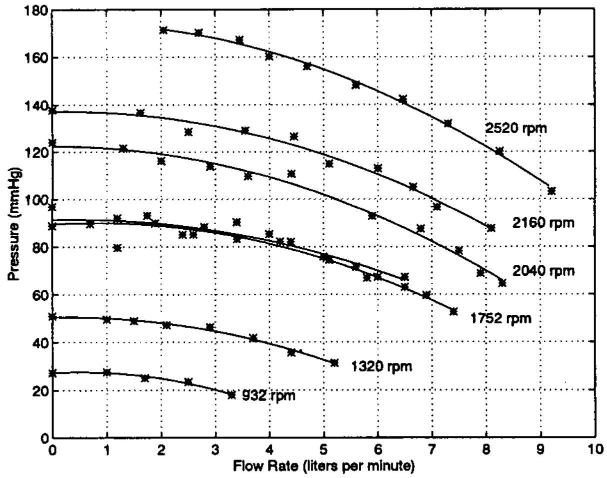


Figure 5: Measured pump characterization curves for several speeds while attached to mock circulatory system

Under pump dynamic loading conditions, the impeller is somewhat displaced due to fluid forces acting upon it. The magnetic bearing keeps the impeller nearly centered in the clearance space. The typical measured displacements in different pump orientations, shown in Fig 4, have a maximum value in the axial direction (S3) of 5 mils compared to the impeller clearance of over 30 mils for one orientation. This indicates a ratio of maximum amplitude of impeller motion to clearance of about 1/6. Clearly, the magnetic bearing keeps the impeller very near the center of the pumping cavity while in numerous orientations and under dynamic load fluid loads.

The pump was tested for pressure increase and efficiency with the CFVAD 3 unit connected to a mock circulation system pumping water. The water was circulated with a physiological level of pump output and differential pressure. Pump inlet pressure (left arterial pressure), pump outlet pressure (aortic pressure), and volume flow rate were measured. Figure 5 shows observed differential pressure head verses the flow rate at various impeller speeds. A typical flow of $6\,l/min$ was achieved against a differential pressure of $120\,mmHg$ at $36\,Hz$ (2160 rpm). A range of pressure heads versus flow rate characteristics may be obtained from this pump with small variations in the impeller rotation speed.

The top left graph of Figure 6 shows pump pressure gain as function of impeller speed. The remaining graphs show the relationship between voltage and power of motor with respect to impeller speed. Under nominal pump conditions of $6\,l/m$ at $115\,mmHg$ and $36\,Hz$ (2160rpm) the motor voltage and power are approximately 11V and 9 watts. The strong linearity of these parameters to pressure and impeller speed indicate that a physiological controller will be implementable with the present design.

# 6 Conclusion

In conclusion, this prototype improves upon its predecessor CFVAD 2 in every way. It has a compact shape more suitable for implantation. The reduced flow paths and impeller design lessen the likelihood of blood trauma. Furthermore, the use of magnetic bearings are more efficient and compact. Even so, this device will require further development. Present research focuses on better system characterizations of both the plant and controller – information that will also aid in the development of later prototypes.

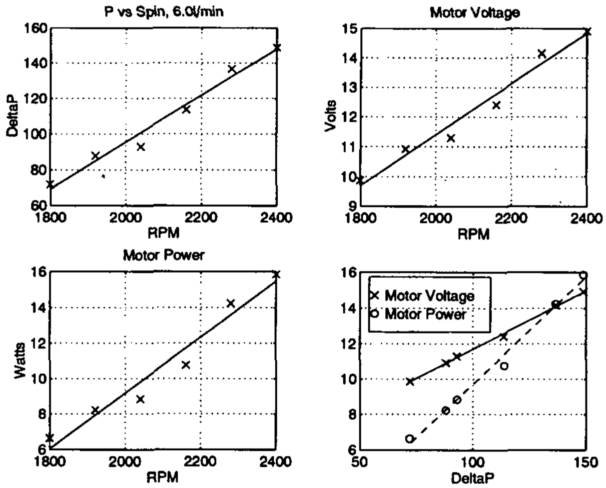


Figure 6: Pressure gain at 6 $l/m$, motor voltage, motor power, and motor voltage and power as function of impeller rpm

# 7 Acknowdgements

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