A BEARINGLESS MOTOR FOR A LEFT VENTRICULAR ASSIST DEVICE (LVAD)

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ABSTRACT

A magnetically levitated motor for a long-term implantable centrifugal LVAD (HeartMate III) has been developed and tested. It is based on so-called "bearingless motor" technology that combines drive, magnetic bearing, and pump rotor functions in a single unit. Several prototypes and animal studies demonstrate that this is a simple, compact, robust, and highly efficient solution that fulfills the requirements of an implantable long-term centrifugal blood pump.

BACKGROUND AND MOTIVATION

In the United States alone almost 200,000 patients die annually from congestive heart failure and about the same figure is estimated for Europe and Japan combined. (Accurate worldwide figures are not available.) 70,000-120,000 of these patients could benefit from a heart transplant, although worldwide only about 5,000 patients are actually transplanted each year. A desperate paucity of donor hearts and tremendous costs ($250,000-400,000 per patient) are limiting factors to heart transplants. Since the left ventricle accounts for 80% of all heart failures, implantable left ventricular assist devices (LVAD) are seen as a promising alternative to many heart transplants. Diaphragm-type LVADs are already routinely used as "bridge to transplant" devices at more than 160 clinical centers around the world. More than 4000 LVAD's worldwide have been implanted to date [1]. HeartMate I (from the market leader Thermo Cardiosystems Inc.) alone has accounted for over 2000 of these implants. Clinical trials to demonstrate that LVADs are also an "alternative to transplant" are currently underway. However, two main problems remain unsolved with diaphragm-type LVADs: large size and inevitable mechanical wear. The size of today's devices limits their application to patients whose weight exceeds about 40 kg, and wear limits device life to 2-3 years. In order to address the size problem, several second-generation devices which are based on the principle of small axial flow pumps (HeartMate II, DeBakey VAD, and Jarvik 2000) are currently in clinical trials [2]. However, these pumps will still eventually wear out due to their mechanical bearings. Third-generation heart-assist systems featuring magnetic bearing technology are therefore being developed. The design and test results of a magnetically levitated motor for a long-term implantable centrifugal LVAD (HeartMate III) is presented in this paper.

PRINCIPLE OF THE PUMP

A Simple Motor

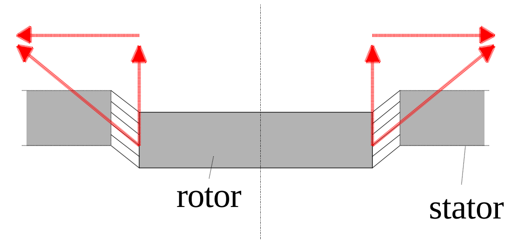
The LVAD is based on bearingless motor technology and combines drive, magnetic bearing, and pump rotor functions in a single unit. Three spatial degrees of freedom are stabilized passively and the remaining three by active control. Figure 1 shows the functional principle of such a bearingless slice motor. Active control of the rotation and the radial position of the rotor is assured by the principle of the bearingless motor. Figure 1(a) shows an axial displacement of the rotor. Axial displacement results in attractive magnetic forces, which act in the opposite direction to the displacement and therefore stabilize the axial position of the rotor. Figure 1(b) shows tilting of the rotor. This leads to stabilizing magnetic forces as well.

A Simple Pump

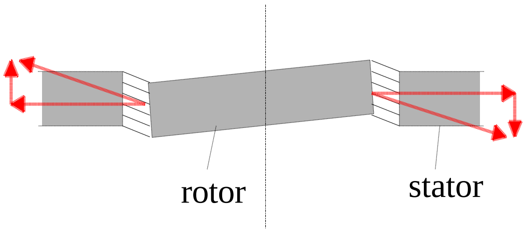
With this concept, a simple and compact solution for a centrifugal pump becomes feasible. Figure 2 shows the basic arrangement of such a bearingless slice motor pump. The rotor-slice can be directly integrated in the impeller. The pump consists of very few parts: the impeller with the integrated motor-rotor and an outer two piece shell. In order to maximize freedom for the pump design, a so called temple motor arrangement was chosen. This special type of motor has no protruding turn windings. Therefore the pump can be directly mounted on the motor. The basic geometry of this type of motor is shown in Figure 3. It is described with more detail in [3].

Simple Electronics and Sensors

Due to passive stabilization of 3 spatial degrees of freedom, only one motor phase and two control phases (which makes a total of three power amplifiers) are needed as a minimum requirement for operation of a bearingless slice motor. With an additional motor phase, a vector-controlled drive becomes feasible [4]. Additionally, only two position sensors are required. This leads to a straightforward electronic control scheme and a simple motor assembly. As a result of this simplicity, it is possible to build a fault tolerant system without increasing the size of the motor and control electronics.



a) axial displacement of the rotor



b) angular displacement of the rotor

FIGURE 1: Passive stabilization of the axial and angular displacement of the slice rotor

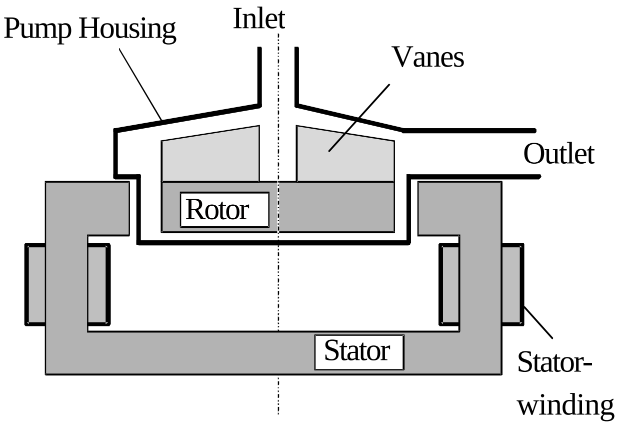


FIGURE 2: Basic arrangement of a bearingless centrifugal pump

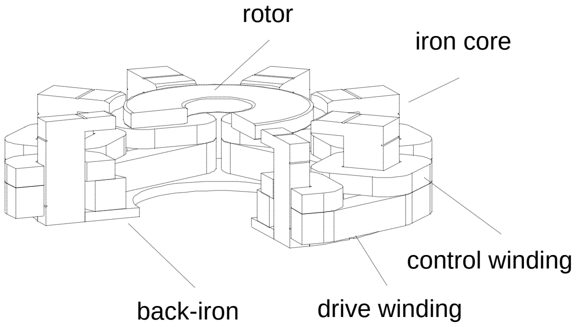


FIGURE 3: Basic arrangement of a temple motor

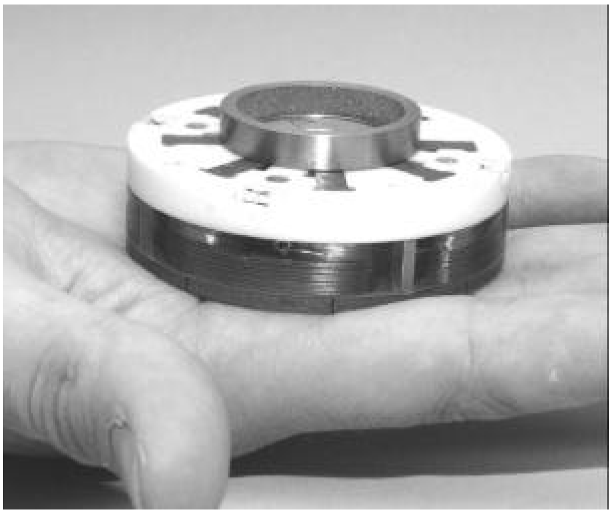


FIGURE 4: Prototype motor

THE PROTOTYPE SYSTEM

The Motor

Through several steps, a highly efficient and compact bearingless motor was developed to prove feasibility through *in vivo* studies. It measures only 65 mm in diameter and 20 mm in height. A relatively large radial magnetic gap of 1.5 mm gives a high degree of freedom in the design of the pump. The motor utilizes large blood gaps to reduce trauma to blood cells and the application of TCI's unique textured blood contacting titanium surface, which discourages thromboembolism. Figure 4 shows an example of this motor with a titanium can insert.

The Pump

A simple centrifugal pump housing was constructed in two pieces – an upper housing containing inflow and outflow channels, and a lower housing containing the motor – enclosing an isolated, shrouded, titanium impeller with blades of a number, size, and shape found to be most hydrodynamically efficient near the design point. Attachment of the pump inflow to the left ventricle and of the pump outflow to the aorta was accomplished by adapting established HeartMate I and II techniques. A flexible segment was placed upstream of the pump inflow to improve anatomical fit, to decouple the pump from cardiac motion, and to provide a point for clamping and ligation for pump explant.

Figure 5 shows a front-view and a section-view of the pump. Figure 6 shows the motor with the lower part of the pump housing and the impeller. All stationary, wetted surfaces were textured in order to promote the growth of a stable, adherent pseudoneointima, thereby reducing the complexity of construction and risk of embolus. The overall dimensions of the motor and pump are just 69 mm in diameter and 30 mm in height.

MEASUREMENTS OF THE PUMP AND MOTOR

The simple design enabled relatively fast and inexpensive fabrication and assembly of several motors and pumps. In vitro testing has demonstrated the desired steady-state flow and pressure characteristics (7 lpm at 135 mmHg achieved at 4800 rpm) and hydrodynamic efficiency (30% at the design point in water). The total power consumption of the motor and bearing for this operating point was on average 8.1 W. In addition, the magnetic bearings can handle forces of up to 10.8 N radially and 7.9 N axially. With a rotor weight of only 34.9 g, a very robust system results.

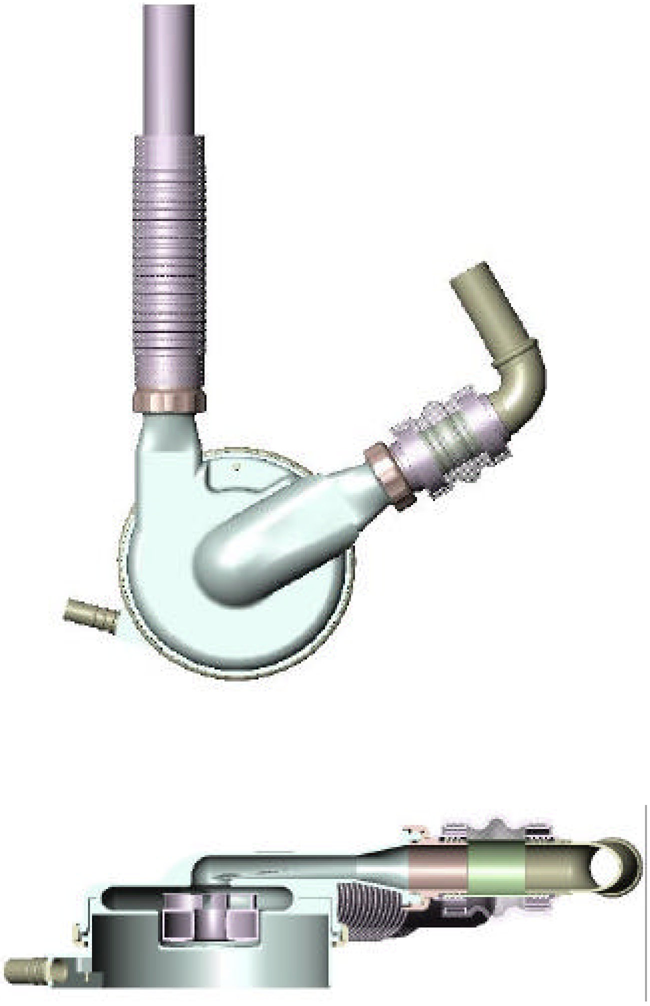


FIGURE 5: Front-view (top) and a section-view (bottom) of the pump

FIGURE 6: Motor with lower part of the pump housing and impeller

IN VIVO TESTING

Four prototype systems have been built for animal tests. The pumps were implanted in five calves (≤ 75 kg at implant) that were anticoagulated with Coumadin (2.5 ≤ INR ≤ 4.0) throughout the chronic studies. Four studies were electively terminated (at 40, 27, 59, and 42 days), while a final study was terminated after the development of severe pneumonia and lung atelectasis (at 27 days). Mean pump flows ranged 3-6 lpm (5.3 ± 0.6, 4.3 ± 1.1, 4.0 ± 0.5, 4.0 ± 0.6, and 5.7 ± 1.1 lpm), except for brief periods of exercise at 6-9 lpm. Plasma free hemoglobin ranged 4 -10 mg/dl (7.8 ± 1.6, 5.8 ± 1.3, 5.6 ± 2.4, 5.3 ± 2.4, 5.2 ± 4.7 mg/dl). All measured biochemical indicators of end organ function remained within normal range.

FIGURE 7: Pump system, ready for implant

FIGURE 8: Explanted pump system

A FAULT TOLERANT SYSTEM

A Fault Tolerant Motor

In parallel with these studies, a fault tolerant motor system has been designed and tested *in vitro*. It combines 4 control phases and 2 drive phases into a 6-phase system. This is the same as competing systems require just for the magnetic bearings, without fault tolerance [5].

For the drive, a simple 2-phase winding was chosen consisting of only 2 coils per phase. If one phase breaks, the motor can still operate as a single phase motor. The minimum requirement for the control winding is two operating phases. In order to be compatible with the 2-phase motor winding, a 4-phase control winding was chosen. With this configuration, any single control phase or two non-related control phases can fail. Figures 9 and 10 show two different winding schemes for the control phases, both of which have been tested. The winding shown in Figure 9 consists of only 8 concentrated coils. Normally, in an 8-slot/2-phase temple motor winding, all gray coils in the left part of the figure (coils 1, 3, 5, 7) form one phase and the gray coils in the right part of the figure (coils 2, 4, 6, 8) form the second phase. In the fault tolerant winding configuration, both phases are subdivided into two (marked as light gray and dark gray in Figure 9). It has been shown experimentally that the bearingless motor can still be operated with only half of the winding, either the light gray coils or the dark gray coils. In a second winding scheme that was tested, two coil sets are wound concentrically on the same core (see Figure 10). In order to balance differences in resistance and induction of the inner and outer coils, each phase consists of 2 inner and 2 outer coils.

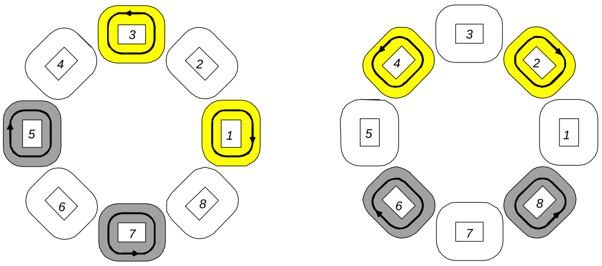


FIGURE 9: Fault tolerant winding with 8 concentrated coils and 4 phases

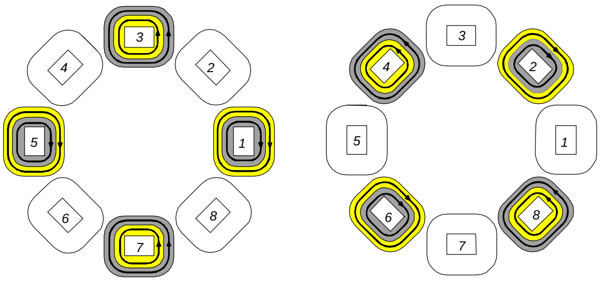


FIGURE 10: Fault tolerant winding with concentrically wound coils and 4 phases

A Fault-Tolerant Electronics

For this special motor type, an electronic system was designed also to be single-fault-tolerant (see Figure11). It comprises two identical subsystems that, during normal operation, work together in "hot redundancy". This means that all 6 motor phases continuously conduct current. The utilization factor of the motor is therefore the same as of a non fault-tolerant type. However, it is essential that both part systems are precisely synchronized in order to avoid coupling effects in the windings. In case of a motor or electronics fault, the functions of sensing, control, and drive revert to the intact part system without interruption. To monitor the system and to handle error modes, several supervisory modules are included.

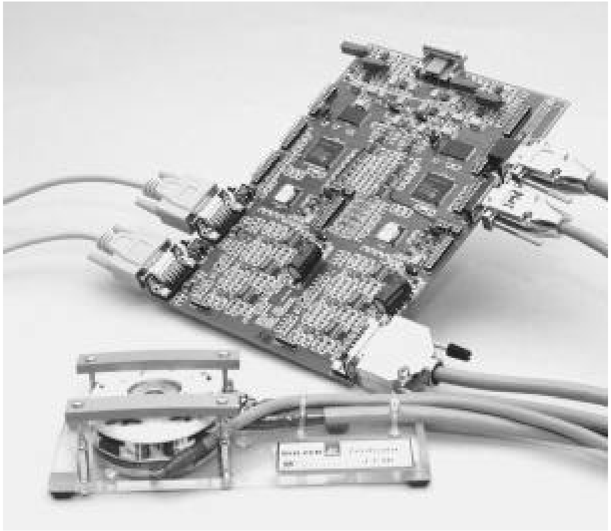


FIGURE 11: Experimental setup for development and test of fault tolerance schemes

Test Results

The motor allows single short or open circuits in all phases of the winding system. Under extreme conditions, the motor can be operated with only half of the motor and bearing coils. Figures 12 through 14 show the behavior of the motor under the 3 most relevant fault conditions. Figure 12 shows a break of two control winding phases (loss of one systems half) and Figure 13 shows short circuit of two control winding phases. The controller can handle both conditions without changing the control parameters. This allows enough time for the processors to detect the fault and to adapt to the fault condition. A very safe fault monitoring and an extremely robust system results. Figure 14 shows a short circuit, followed by an interruption of one drive winding phase. During short circuit, a current is induced in the defective phase. In reaction, the current in the remaining phase increases, since it has to provide the whole torque.

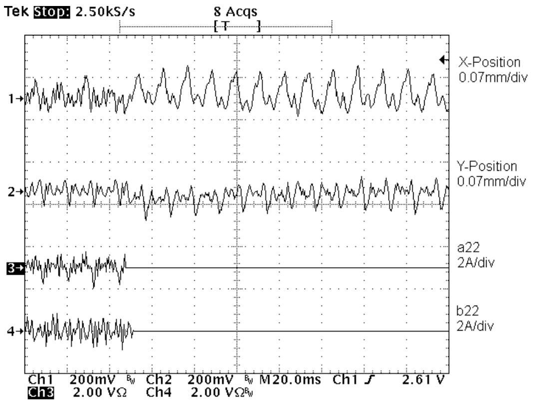


FIGURE 12: Rotor position (channels 1 and 2) and phase currents (channels 3 and 4) during break of two control winding phases

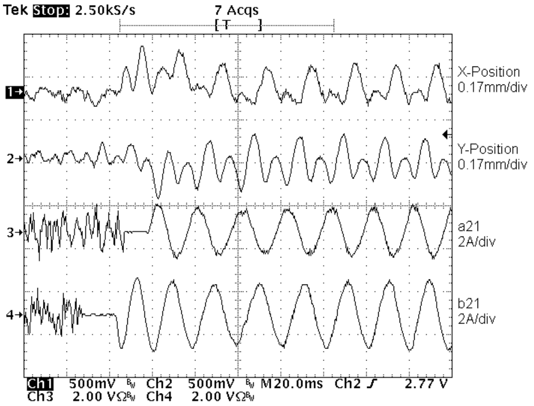


FIGURE 13: Rotor position (channels 1 and 2) and phase currents (channels 3 and 4) during short circuit of two control winding phases

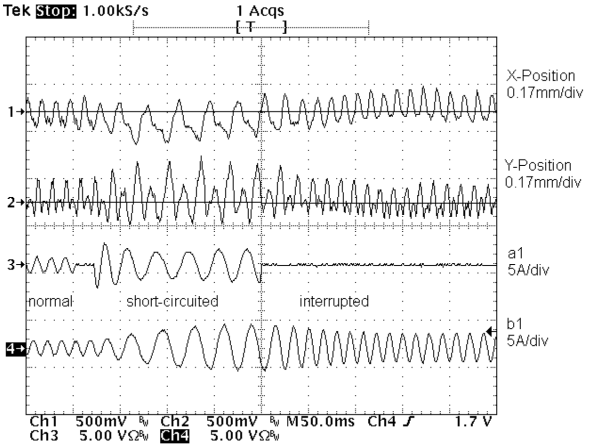


FIGURE 14: Rotor position (channels 1 and 2) and drive phase currents (channels 3 and 4) during a short circuit, followed by an interruption of one drive winding phase.

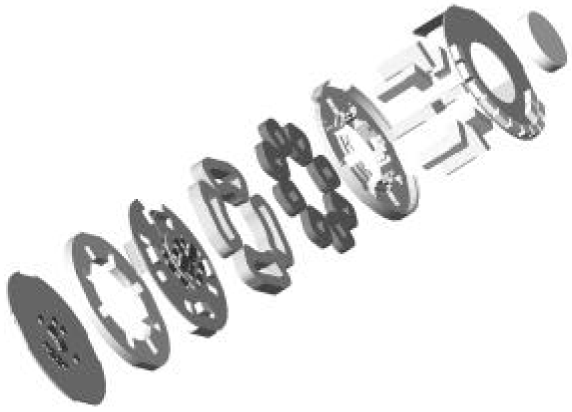


FIGURE 15: Exploded view of the bearingless motor with integrated electronics

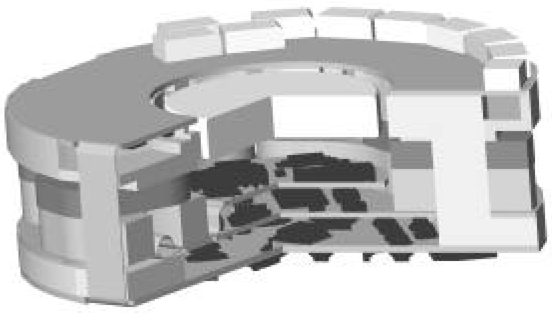


FIGURE 16: Cross-section of the bearingless motor with integrated electronics

CONCLUSION AND OUTLOOK

With several prototypes and animal tests it was demonstrated that our present bearingless motor is a simple, compact, robust, and highly efficient solution that fulfills the requirements of an implantable long-term centrifugal blood pump. In a parallel effort, it was successfully demonstrated that this motor can be made fault-tolerant. With this feature it should be possible to fulfill FDA safety requirements for Class 3 medical devices. The next step will be a maglev centrifugal LVAD pump with fully integrated electronics. An embodiment of the motor-electronics unit is shown in Figures 15 and 16. Overall measures of motor and pump will be 69 mm diameter and 35 mm height.

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