Implantable physiologic controller for left ventricular assist devices with telemetry capability

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Objective: Rotary type left ventricular assist devices have mitigated the problem of durability associated with earlier pulsatile pumps and demonstrated improved survival. However, the compromise is the loss of pulsatility due to continuous flow and retained percutaneous driveline leading to increased mortality and morbidity. Lack of pulsatility is implicated in increased gastrointestinal bleeding, aortic incompetence, and diastolic hypertension. We present a novel, wirelessly powered, ultra-compact, implantable physiologic controller capable of running a left ventricular assist device in a pulsatile mode with wireless power delivery.

Methods: The schematic of our system was laid out on a circuit board to wirelessly receive power and run a left ventricular assist device with required safety and backup measures. We have embedded an antenna and wireless network for telemetry. Multiple signal processing steps and controlling algorithm were incorporated. The controller was tested in in vitro and in vivo experiments.

Results: The controller drove left ventricular assist devices continuously for 2 weeks in an in vitro setup and in vivo without any failure. Our controller is more power efficient than the current Food and Drug Administration–approved left ventricular assist device controllers. When used with electrocardiography synchronization, the controller allowed on-demand customization of operation with instantaneous flow and revolutions per minute changes, resulting in a pulsatile flow with adjustable pulse pressure.

Conclusions: Our test results prove the system to be remarkably safe, accurate, and efficient. The unique combination of wireless powering and small footprint makes this system an ideal totally implantable physiologic left ventricular assist device system. (J Thorac Cardiovasc Surg 2014;147:192-202)

By 2030, it is anticipated that 10 million people in the United States will live with heart failure (1/33 people).^{1,2} However, the available donor hearts remain stagnant at approximately 2000 per year.³ In recent years, implantable left ventricular assist devices (LVADs) have offered another option to this ill population. Although showing a survival advantage, the earlier pneumatically driven positive displacement pumps showed high mechanical failure due to wear and tear associated with multiple moving parts and associated friction. The technologic improvement of rotary pumps with a single moving part has led to increased durability and patient survival compared with their earlier generations both as a bridge to transplant and as destination therapy. However, the improved durability of rotary blood pumps comes at the compromise of having a pulseless continuous blood flow.

Despite the technologic advances to the pump body, the requirement of a transcutaneous driveline to conduct power (previously needed for shuttling air for pulsatile devices), controller algorithms, and data exchange between the pump and the extracorporeal controller remains unchanged. Despite the benefits of this technology, driveline-associated infections are a common and devastating complication^{4,5} that causes a significant negative impact on a patient's quality of life and increased medical cost.⁶ The presence of nonphysiologic pulseless blood flow in patients with long-term LVAD support has been implicated in increased gastrointestinal bleeding,⁷⁻⁹ limited cardiac unloading,^{10,11} vascular malformations,^{12,13} and aortic incompetence.^{14,15} Moreover, because successful LVAD explantation occurs less often with continuous-flow pumps, concerns have been raised regarding their use in patients with a potential for myocardial recovery.

We have previously demonstrated the efficacy of a Free-Range Resonant Electrical Energy Delivery (FREE-D) system that uses high-quality factor resonant coupling technology to wirelessly transfer energy to power an LVAD.¹⁶ The FREE-D system uses magnetically coupled resonators to seamlessly supply energy without compromising mobility or requiring direct contact between the patient and the energy source,¹⁷ thus freeing the patient from a driveline. Application of such technology also requires a small implantable controller with wireless communication.

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Abbreviations and Acronyms	
ECG	= electrocardiography
EMF	= electromotive force
FREE-D	= Free-Range Resonant Electrical Energy Delivery
LVAD	= left ventricular assist device
MCL	= mock circulation loop
rpm	= revolutions per minute
Δ RPM	= difference between systolic and diastolic pump speeds
UMC-Physio	= ultra-compact implantable physiologic controller

In the current study, we present an ultra-compact implantable physiologic controller (UMC-Physio) to entirely untether patients from the LVAD driveline, increase patients’ quality of life, and significantly reduce complications associated with nonphysiologic continuous flow.

MATERIALS AND METHODS

Left Ventricular Assist Device Motor

Current-generation LVADs are based on brushless direct current motors, also known as electronically commutated motors that are powered by a direct current electric source (Appendix 1). An LVAD has 3 sets of coils positioned around its impeller. Once a coil is energized, it produces a magnetic field that causes the permanent magnet on the rotor to align with that magnetic field, which in turn rotates the impeller. However, the rotating action stops when the rotor is aligned with the magnetic field created by the energized coil. To keep the impeller in motion, the coils are energized in a particular sequence (120° out-of-phase from one another). The process of activating and deactivating the coils to keep the impeller spinning is termed “commutating.”

Pump Speed

Because an LVAD does not have any sensor, the speed of the pump is measured from back electromotive force (EMF) generated by the motor’s rotation. The amplitude of the back EMF is proportional to the angular velocity of the LVAD, but its shape will not change with speed and only depends on the pump characteristic. Our controller uses attenuation and filtering processing to sense this signal. The attenuation is required to bring the signal down to an allowable common mode range of the sensing circuit, and the low pass filtering is necessary for smoothing the high switching frequency noise.

When to Commutate and Power Efficiency

The challenge in controlling a brushless direct current motor is knowing when to commutate, because commutation must occur at precise points during rotation for the motor to have maximum torque and smooth operation. Because the implanted system must comply with the highest efficiency, we have incorporated a control algorithm that spins the motor more efficiently to reduce the overall power consumption by calculating when to commutate on the basis of the back EMF of the motor. To further reduce the power, we have introduced a sleep mode for our system that wakes the controller up at only certain point for a short amount of time (5 ms/s) to capture the sensor data and monitor the device operation.

Mode of Operation (Mimicking Physiologic Flow)

The purpose of an LVAD physiologic controller is to match the aortic pressure to a certain value derived from the patient activity. The UMC-Physio has 2 different modes of operation to mimic physiologic flow: continuous and pulsatile modes. In the continuous-flow mode, the user can set 1 speed. However, for the pulsatile mode, the user is required to define separate speeds during systole and diastole. Moreover, in case of a patient with potential myocardial recovery, the controller will have a higher speed during diastole and a lower speed during systole to decrease the preload of the heart. This allows the left ventricle to pump less blood during systole, which means less work for the left ventricle.¹⁸ In the event when a higher pulse pressure is necessary, such as patients with gastrointestinal bleeding, the system uses a higher speed during systole and a lower speed during diastole to create a larger pulse. The difference between systolic and diastolic pump speeds is called “ Δ RPM.”

Signal Processing

The pulsatile algorithm in our controller needs to swiftly modulate the pump speed to be able to gate it with an electrocardiography (ECG) signal for co-pulsation, counterpulsation, or fixed operation modes. Thus, the response time of the system is of a unique interest to us, because heart rates up to 120 beats/min with 30% systolic duration would limit us with a window of 0.15 second for the period of systole. We have used amplifiers and signal processing and filtering methods to further analyze an ECG signal.

Communication

Many medically implanted devices use wired or wireless methods to communicate with their external circuitry. Furthermore, we have integrated an antenna, a transceiver unit (transmitter and receiver), and a wireless network algorithm to enable our UMC-Physio to establish a reliable telemetry communication with an extracorporeal platform, such as a smart phone, tablet, or personal computer. In our wireless networking, the amplitude and width of the data pulses are kept constant in the system. The position of each pulse is varied by instantaneous sampled value of the modulating wave. This method is called “pulse position modulation.” Our controller puts out a stream of pulses called “sync pulses,” which the external receiver recognizes with a circuitry called “phase locked loop.” To create synchronization, an oscillator in the phase locked loop generates pulses at the same frequency as the controller.

Backup and Safety

Because the system will be powered by using our FREE-D emitted electromagnetic fields, we had to consider the noise potential might influence the proper function of our electronic implant. To significantly reduce the EMF susceptibility of our controller, we have combined multiple filtering mechanisms, ground isolation, and ground decoupling in our design. Backup motor driver, overvoltage, and overcurrent protection circuits, thermal shutdown circuit, and independent pulse width modulation generator are among other safety issues that we have incorporated in our design. In addition, a suction event detection algorithm also was incorporated in the system for additional safety by detecting power consumption peaks that are abnormal.

Design and Development

The schematic of our system was laid out on a compact 4-layer printed circuit board using Altium Designer software (Altium, Australia) to run a rotary blood pump with required safety and backup measures. The control algorithms and graphical user interface were further developed in IAR System and LabVIEW software (both National Instruments, Austin, Tex) respectively. The final system was coupled with our alternating current to direct current converter and a receiver coil to wirelessly receive power (Figure 1, A).

FIGURE 1. A, The developed UMC-Physio coupled with the FREE-D receiver coil to wirelessly obtain power and run an LVAD pump. B, The external graphical user interface embedded in a smartphone. This platform displays and controls an LVAD pump by wireless communication or Internet access.

In Vitro Experiments

The UMC-Physio initially was tested in a mock circulation loop (MCL) with the HeartMate II (Thoratec Co, Pleasanton, Calif) and HeartWare Ventricular Assist Device (HeartWare Inc, Framingham, Mass) pumps (Figure 2). For the continuous mode, the UMC-Physio ran the HeartMate II axial pump for 2 weeks. The pump speed was gradually increased from 6000 up to 13,000 revolutions per minute (rpms) to ensure functionality. Next, the UMC-Physio drove the centrifugal pump for a 2-week period. The speed of the LVAD was changed from 1800 to 4000 rpms by 100 rpm steps.

For the pulsatile mode, we tested both axial and centrifugal pumps in the MCL. The fraction of systole was gradually increased from 20% to 80%, and the mean LVAD flow, produced by our controller, was measured at various heart rates. In the second test, systolic, diastolic, mean aortic, and pulse pressure were measured. Last, we investigated how our UMC-Physio automatically increases the mean LVAD flow with increase in heart rates of 40 to 120 beats/min. These were then verified by changes in Δ RPMS.

Moreover, the pump's outlet flow was estimated by recording the changes in power consumption and speed of the controlled LVAD in the MCL. The estimated flow was compared by the actual flow measured by the Transonic Flow Sensor (Transonic System Inc, Ithaca, NY) to ensure its accuracy.

In Vivo Experiments

The protocol was approved by the Institutional Animal Care and Use Committee and complied with the "Guide for the Care and Use of Laboratory Animals." For the in vivo testing, the combination of our physiologic controller and HeartWare Ventricular Assist Device pumps was implanted in 2 large female pigs. We performed proof of concept in vivo experiments to investigate the feasibility and implantability of our controller, to study its interference with biological tissue and temperature increase, to examine the effect of tissue impedance on our telemetry communication, and to prove our system capability of introducing pulsatile flow in the animals. The experiments were performed without any manipulation in the animal's vasomotor system (during the

FIGURE 2. In vitro test setup where an LVAD pump is placed in an MCL. The pressure and flow are measured by the corresponding sensors. HMII, HeartMate II; HVAD, HeartWare Ventricular Assist Device; LVAD, left ventricular assist device.

pulsatile mode, the preload, afterload, and contractility were not manipulated with inotropic infusions and were kept in steady state before initiating pulsatile mode).

The surgical approach briefly consisted of induction by intravenously administered anesthesia. The animals were then intubated and ventilated. The animals were maintained on isoflurane anesthesia throughout the surgery. External ECG was set up via needle electrodes in the extremities. A dwell-time silicone catheter was placed in the right internal carotid and external jugular vein for pressure monitoring. A 16g catheter was placed in the left external jugular vein for fluid maintenance and monitoring of the central pressure in the heart. In addition to the central line, an intravenous catheter was placed in the cephalic or saphenous veins. Once the animals were instrumented, systolic and diastolic pressure, central venous pressure, heart rate, rectal temperature, oxygen saturation, and end-tidal carbon dioxide were monitored continuously and whole blood sample was collected for immediate hematocrit, coagulation profile, and blood gases for regular assessment.

The chest of the animal was opened via a left lateral thoracotomy at the fifth/sixth intercostal space, and the small portion of the fifth rib was removed at the costochondral junction. Surgery was performed without the use of cardiopulmonary bypass. The descending aorta was exposed, and the azygous vein was separated.

A dose of heparin was administered to obtain an activated clotting time greater than 400 seconds in preparation for anastomosis of the outflow graft to the descending thoracic aorta. The graft was placed transdiaphragmatically, positioned dorsally to prevent atelectasis of the lung, and anastomosed end-to-side to the aorta. Antiarrhythmics were administered in preparation for insertion of the inlet of the pump into the left ventricle of the heart. Teflon pledgeted sutures were placed radially around the left ventricular apex, and a 10- to 12-mm core of the apex was removed using a circular trocar. The HeartWare Ventricular Assist Device inlet cannula was placed in the ventricle. The inlet of the pump was secured by suturing of the cannula sewing-ring to the apical pledgets.

The animal experiment was designed to test for the following parameters: bio-interference, thermal changes, tissue impedance, telemetry communication, continuous mode operation, and pulsatile mode operation.

Statistical Analyses

The correlation among the variables (pump flow, heart rate, diastolic pressure, systolic pressure, and pulse pressure) was tested by Pearson correlation for all observations, in which a 2-tailed *P* value less than .05 shows the existence of statistically significant correlation between 2 variables. Furthermore, Pearson correlation measures the strength and direction of the linear relationship between the 2 variables. The correlation coefficient can range from -1 to $+1$, with -1 indicating a perfect negative correlation, $+1$ indicating a perfect positive correlation, and 0 indicating no correlation at all. In addition, linear, logarithmic, and cubic regressions were used where appropriate for examining relations between correlated variables.

RESULTS

In Vitro

The UMC-Physio tolerated dynamic power fluctuations introduced by wireless power delivery. During each 2-week in vitro testing with axial and centrifugal pumps, there was no telemetry disconnection, thermal shutdown event, sudden speed changes, power dropout, or pump failure. Moreover, the graphical user interface embedded in an iPhone (Microsoft Corp, Redmond, Wash), iPad (Microsoft Corp), and laptop enabled us to wirelessly observe the recorded data and modify the LVAD configuration (Figure 1, B).

Response to Physiologic Demands

When used with ECG gating, the UMC-Physio allowed on-demand customization of operation with instantaneous flow and rpm changes, resulting in a pulsatile flow with adjustable pulse pressure. The controller responds to increases and decreases in heart rate and automatically adjusts pump flow and pulsatility accordingly.

Axial Left Ventricular Assist Device (HeartMate II)

Figure 3, A, demonstrates that our controller increases the mean flow of an axial pump by an average of 0.313 L/min per 10 beats/min change in heart rate. Likewise, 1000 Δ RPM increases the mean flow in this pump by an average of 0.24 L/min. The linear regression equation with R^2 of 0.981 is used; therefore, approximately 98% of the mean pump flow (L/min) is explained by heart rate and Δ RPM. This model is expressed as follows:

Mean LVAD Flow (L/min) = $0.031333 \cdot \text{Heart Rate (beats/min)} + 0.00024089 \cdot \Delta\text{RPM (rpms)} + 2.38056$.

Figure 3, B, shows that a 10% increase in fraction of systole leads to an increase of 0.29 L/min more mean flow in our system. The R^2 for this linear model is 0.975. The relationship (linear regression) is expressed as follows:

Mean LVAD Flow (L/min) = $0.02486 \cdot \text{Heart Rate (beats/min)} + 0.02934 \cdot \text{Fraction of Systole (\%)} + 1.8883$.

Figure 3, C to F, show the UMC-Physio increases the systolic and diastolic pressures by increased in the heart rate. The linear regression is expressed as follows:

Systolic Pressure (mm Hg) = $0.44014 \cdot \text{Heart Rate (beats/min)} + 0.01041 \cdot \Delta\text{RPM (rpms)} + 2.02963$ and Diastolic Pressure (mm Hg) = $0.41806 \cdot \text{Heart Rate (beats/min)} + 0.00042593 \cdot \Delta\text{RPM (rpms)} + 4.85185$.

Figure 3, G, shows that increase in Δ RPM creates more pulse pressure. Figure 3, H, proves that our controller consumes less power compared with the HeartMate II controller to achieve the flow and speeds for the HeartMate II pump, being more efficient by a mean of 17.54%. The relation between efficiency and flow (independent variable) is estimated by curve approximation. As a result, the cubic model has been used because of its highest R^2 (0.999). The efficiency is estimated by the following:

Efficiency (%) = $0.08309 \cdot \text{Flow}^3 + 1.897 \cdot \text{Flow}^2 + 13.907 \cdot \text{Flow (L/min)} - 15.6778$.

Centrifugal Left Ventricular Assist Device (HeartWare Inc)

Figure 4, A, reveals that our controller can increase the mean flow of a centrifugal pump by 0.22 L/min when the heart rate increases by 10 beats/min. Likewise, 500 Δ RPM increases the mean flow in this pump by an average of 0.319 L/min. The linear regression was used because of an R^2 value of 0.885. Thus, approximately 88% of the data in the LVAD mean flow are explained by

FIGURE 3. The result of our in vitro testing with the HeartMate II (Thoratec Co, Pleasanton, Calif) pump. Most of these equations are related to a specific MCL. A, Our controller adapts to changes in heart rate to create the corresponding mean LVAD flow. The degree of pulsatility also changes the mean flow. The linear regression for this model is as follows: Mean LVAD Flow (L/min) = $0.031333 \cdot \text{Heart Rate (beats/min)} + 0.00024089 \cdot \Delta\text{RPM (rpms)} + 2.38056$. B, A 10% increase in fraction of systole increases the mean flow in our system by 0.29 L/min; therefore, more flexibility is added to adapt to the patient's everyday life. This relationship is modeled as follows: Mean LVAD Flow (L/min) = $0.02486 \cdot \text{Heart Rate (beats/min)} + 0.02934 \cdot \text{Fraction of Systole (\%)} + 1.8883$. C to F, The UMC-Physio changes the systolic and diastolic pressures by increases or decreases in the heart rate. The following equations represent the estimated systolic and diastolic pressures: Systolic Pressure (mm Hg) = $0.44014 \cdot \text{Heart Rate (beats/min)} + 0.01041 \cdot \Delta\text{RPM (rpms)} + 2.02963$ Diastolic Pressure (mm Hg) = $0.41806 \cdot \text{Heart Rate (beats/min)} + 0.00042593 \cdot \Delta\text{RPM (rpms)} + 4.85185$. G, Increments in ΔRPM generate more pulsatility. H, For 2 watts of power, the UMC-Physio produces 5.5 L/min of flow, whereas the HeartMate II controller can create only 3.8 L/min; therefore, our controller is on average more efficient by 17.54%. This result is good evidence for how pump controller optimization and precise control can contribute to the efficiency of an LVAD. The efficiency is estimated by the following: Efficiency (%) = $0.08309 \cdot \text{Flow}^3 + 1.897 \cdot \text{Flow}^2 + 13.907 \cdot \text{Flow (L/min)} - 15.6778$. bpm, Beats/min; LVAD, left ventricular assist device; ΔRPM , difference between systolic and diastolic pump speeds; UMC-Physio, ultra-compact implantable physiologic controller.

heart rate and ΔRPM . The regression model for this graph is as follows:

Mean LVAD Flow (L/min) = $0.0226 \cdot \text{Heart Rate (beats/min)} + 0.000638222 \cdot \Delta\text{RPM (rpms)} + 3.7245$.

Figure 4, B, confirms that an increase in fraction of systole increases the mean by 0.2 L/min. The R^2 for this linear regression is 0.998; thus, approximately 99.8% of

the data in LVAD mean flow are explained by heart rate and fraction of systole. The regression equation seems to be useful for making predictions because the value of R^2 is close to 1. This formula explains this linear regression:

Mean LVAD Flow (L/min) = $0.0046 \cdot \text{Heart Rate (beats/min)} + 0.0205 \cdot \text{Fraction of Systole (\%)} + 4.09312$.

FIGURE 4. The results of the in vitro testing with the HeartWare Ventricular Assist Device (HeartWare Inc, Framingham, Mass) pump. The results are related to our MCL. A, The UMC-Physio increases the mean flow of the centrifugal pump by 0.22 L/min with 10 beats/min in the heart rate. The linear regression model for this graph is Mean LVAD Flow (L/min) = $0.0226 \cdot \text{Heart Rate (beats/min)} + 0.000638222 \cdot \Delta\text{RPM (rpms)} + 3.7245$. B, Confirms that 10% increase in fraction of systole linearly increases the mean LVAD flow by 0.2 L/min. This formula explains this linear regression: Mean LVAD Flow (L/min) = $0.0046 \cdot \text{Heart Rate (beats/min)} + 0.0205 \cdot \text{Fraction of Systole (\%)} + 4.09312$. C to F, Heart rates up to 80 beats/min will increase the systolic and diastolic pressures produced by our controller. The systolic and diastolic pressures can be estimated by the following formulas: Systolic Pressure (mm Hg) = $0.50067 \cdot \text{Heart Rate (beats/min)} + 0.02364 \cdot \Delta\text{RPM (rpms)} + 31.9855$ Diastolic Pressure (mm Hg) = $0.26817 \cdot \text{Heart Rate (beats/min)} + 0.00188 \cdot \Delta\text{RPM (rpms)} + 32.085$. G, Pulse pressure can be easily controlled by our system; the linear regression is modeled as Pulse Pressure (mm Hg) = $0.2325 \cdot \text{Heart Rate (beats/min)} + 0.02176 \cdot \Delta\text{RPM (rpms)} - 0.14$. H, Proves that our controller is also more efficient than the HeartWare Ventricular Assist Device controller. The descriptive statistics indicate that our controller is more efficient by a mean of 35.49%. Efficiency is modeled as follows: Efficiency (%) = $2.648 \cdot \text{Flow}^2 - 32.630 \cdot \text{Flow (L/min)} + 132.964$. bpm, Beats/min; LVAD, left ventricular assist device; ΔRPM, difference between systolic and diastolic pump speeds; UMC-Physio, ultra-compact implantable physiologic controller.

Figure 4, C to F, display the heart rates up to 80 beats/min that will increase the systolic and diastolic pressures produced by our controller and the LVAD. The systolic and diastolic pressures can be estimated by the following formulas:

Systolic Pressure (mm Hg) = $0.50067 \cdot \text{Heart Rate (beats/min)} + 0.02364 \cdot \Delta\text{RPM (rpms)} + 31.98556$ and Diastolic Pressure (mm Hg) = $0.26817 \cdot \text{Heart Rate (beats/min)} + 0.00188 \cdot \Delta\text{RPM (rpms)} + 32.08556$.

Figure 4, G, shows that pulse pressure can be easily controlled by our system. The linear regression is as follows:

Pulse Pressure (mm Hg) = $0.2325 \cdot \text{Heart Rate (beats/min)} + 0.02176 \cdot \Delta\text{RPM (rpms)} - 0.1$.

Figure 4, H, shows that our controller is more efficient than the HeartWare Ventricular Assist Device controller. The relation between efficiency and flow (independent variable) is estimated by cubic estimation because it has the

FIGURE 5. Representative demonstration of the power consumption and speed of a HeartWare Ventricular Assist Device pump in one of our in vivo experiments. The consistent and smooth pump speed indicates how our control adapts to the sudden changes of our wireless power delivery. In addition, it displays the EMF noise-free telemetry data. The temperature of the implanted unit also is graphed to illustrate there was no temperature increase in the unit that might affect the surrounding tissues. *HVAD*, HeartWare Ventricular Assist Device; *RPM*, revolutions per minute; *UMC-Physio*, ultra-compact implantable physiologic controller.

highest R^2 compared with other models. The descriptive statistics indicate that our controller is more efficient by mean of 35.49%. Efficiency is modeled as follows:

Efficiency (%) = $2.648 \cdot \text{Flow}^2 - 32.630 \cdot \text{Flow (L/min)} + 132.964$ (these results are based on the specific MCL that we used and can vary in a biological system).

In Vivo

The controller successfully ran an untethered HeartWare Ventricular Assist Device pump for 3 hours in each animal with continuous and pulsatile modes (cumulatively 6 hours of experimental data). We initially tested the ability of the controller to run the pump in the continuous mode to assess power consumption in comparison with HeartWare Ventricular Assist Device controller. Then we tested efficiency as a baseline measure of overall controller performance. Our study showed that there was no biological tissue interference, temperature increase, or network mismatching in our system. The tissue impedance decreased our communication distances. Figure 5 also shows that

our controller automatically adapts to the dynamic changes in the wireless power to create a steady pump speed.

DISCUSSION

The human heart and circulation have developed over millennia—starting from the primitive coelom that was a passive receptacle for gas exchange, feeding, and sexual reproduction lined by endoderm to subsequent lining of the coelomic cavity by mesodermal contractile cells, which became a “gastrovascular organ.” Further differentiation into a peristaltic heart in Chordata and vertebrates leads to the sophisticated differentiation from a primitive tubular hearts into a looped, unidirectional circulation with a highly specialized conductive and pacemaker cells. The subsequent parallel pulmonary circulation was a further evolutionary step. The evidence to suggest the primordial gastrovascular structure to today’s highly specialized cardiac structure is demonstrable by the common nerve supply to heart and gut by the central nervous system and the pulsatility that has been the hallmark of mammalian circulation for millennia.¹⁹

Rotary pumps use controllers that run an LVAD pump at a constant speed.²⁰ However, the continuous flow is mostly acceptable for short-term use of an LVAD (eg, bridge to transplant), but as we witness more widespread use and long-term support such as destination therapy, several unintended effects of continuous flow are becoming apparent.^{8,9,12-15,21} After all, the continuous flow associated with current LVAD devices was a result of replacing earlier pulsatile devices with a more durable pumping system at the cost of pulsed operation. Our technologic advance allows leveraging both the durability of rotary pumps and restoring the pulsed operation simultaneously.

Several researchers have tried to mimic nature by designing physiologic LVAD controllers by using proportional-integral^{22,23} and fuzzy logic^{24,25} control algorithms to automatically modulate the constant speed of the pump. However, our system combines adaptive optimal proportional feedback control algorithm, ECG gating, and head pressure and flow estimation to perform its physiologic modulation during systole and diastole. Previous studies^{26,27} produced acceptable results for the estimation of the pulsatile flow and head pressure, but they did not investigate the stability of the transient response of the pump flow. The transient response has been demonstrated.²⁸⁻³⁰ Moreover, the head pressure estimation observed in other studies³¹ suffers from a time delay because of the use of excessive filtering. Our system uses a simple low-pass filter to estimate the transient and steady-state head pressure and flow. This method produces an instant result with no delay. Although the real-time estimation process requires extra computation, it is a simple way to substitute sensor implementation. All of our flow and head pressure estimations were performed in our MCL.

Prior attempts include magnets, capacitive communication, Bluetooth, Wi-Fi, ultrasound, and other radio frequencies to wirelessly communicate with an implantable medical device. The capacitive intra-body communication requires electrodes to be placed on or near the skin. Of these, ultrasound and magnets have low transmission efficiency and range through the biological tissue and are challenging to miniaturize. Radiofrequencies, such as a medical implant communication service band or Wireless Medical Telemetry Service, require bulky antennas that restrict how small the controller can be. We have used industrial, scientific, and medical radio bands of 2.4 GHz with a highly automated communication that has a short wavelength and therefore a small size.

Data modulation has been accomplished by amplitude shift key, phase shift keying, continuous phase modulation, and other methods to translate data. Our pulse position modulation method has the advantage of pulses with constant amplitude and width, but the disadvantage of requiring synchronization between transmitter and receiver. The obtained telemetric data during our experiments were

noise-free from sources such as the EMF and pulse width modulation switching. Our graphical user interface enables a physician to monitor and control the patient's LVAD even when not in wireless range of the implanted device by using the Internet to access the patient's external receiver that in turn relays the information to and from the implanted controller. The corresponding data can be placed in the database for future review and analysis.

The UMC-Physio rapidly synchronizes the pump to the heart rate to mimic physiologic flow to adapt to the patient's everyday physiologic demands. Our controller can produce the same pump speed and flow rate at a lower power than the HeartWare Ventricular Assist Device and HeartMate II controllers, making it more efficient. The power consumption of the controller is extremely low because of the accurate commutation and sleep mode. The UMC-Physio enables an LVAD to be controlled digitally, bringing precise control, higher efficiency, and faster response time for this type of pump. The UMC-Physio can control both axial and centrifugal blood pumps.

Future Research and Study Limitations

The scope of the current article is limited to proving the feasibility of the bench-top testing of such a system in a physiologic MCL and initial in vivo proof of concept. In our future research, we will study the performance of the controller in heart failure models in an acute and chronic setting. In these studies, we will test how our technology creates pulsatile flow that is responsive to physiologic demand in the complex milieu of a neurologic, baroreflex, and hormonal changes. Theoretic conditions that may have bearing on the clinical application will include extremes of physiologic states, such as extreme hypertension, severe hypotension, severe bradycardia, and tachyarrhythmias. However, these conditions are usually controlled in the population with LVADs by pharmacologic means and use of appropriate pacing and tachyarrhythmia therapies; thus, they should not pose problems for the workings of the UMC-Physio controller.

CONCLUSIONS

The novel combination of wireless powering, a user-friendly interface, and a small footprint makes this an ideal totally implantable LVAD system. We have demonstrated the feasibility of a controller that can create versatile physiologic flow with a conventional continuous-flow pump.

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References

1. Heidenreich PA, Albert NM, Allen LA, Blumke DA, Butler J, Fonarow GC, et al; American Heart Association Advocacy Coordinating Committee; Council on Arteriosclerosis, Thrombosis and Vascular Biology; Council on

- Cardiovascular Radiology and Intervention; Council on Clinical Cardiology; Council on Epidemiology and Prevention; Stroke Council. Forecasting the impact of heart failure in the United States: a policy statement from the American Heart Association. *Circ Heart Fail*. 2013;6:606-19.
2. Roger VL, Go AS, Lloyd-Jones DM, Benjamin EJ, Berry JD, Borden WB, et al. Heart disease and stroke statistics—2012 update: a report from the American Heart Association. *Circulation*. 2012;125:e2-220.
 3. Stehlik J, Edwards LB, Kucheryavaya AY, Aurora P, Christie JD, Kirk R, et al. The Registry of the International Society for Heart and Lung Transplantation: Twenty-Seventh Official Adult Heart Transplant Report—2010. *J Heart Lung Transplant*. 2010;29:1089-103.
 4. Sivaratnam K, Duggan JM. Left ventricular assist device infections: three case reports and a review of the literature. *ASAIO J*. 2002;48:2-7.
 5. Simon D, Fischer S, Grossman A, Downer C, Hota B, Heroux A, et al. Left ventricular assist device-related infection: treatment and outcome. *Clin Infect Dis*. 2005;40:1108-15.
 6. Slaughter MS, Rogers JG, Milano CA, Russell SD, Conte JV, Feldman D, et al. Advanced heart failure treated with continuous-flow left ventricular assist device. *N Engl J Med*. 2009;361:2241-51.
 7. Pagani FD, Miller LW, Russell SD, Aaronson KD, John R, Boyle AJ, et al. Extended mechanical circulatory support with a continuous-flow. *J Am Coll Cardiol*. 2009;54:312-21.
 8. Demirozu ZT, Radovancevic R, Hochman LF, Gregoric ID, Letsou GV, Kar B, et al. Arteriovenous malformation and gastrointestinal bleeding in patients with the HeartMate II left ventricular assist device. *J Heart Lung Transplant*. 2011;30:849-53.
 9. Crow S, John R, Boyle A, Shumway S, Liao K, Colvin-Adams M, et al. Gastrointestinal bleeding rates in recipients of nonpulsatile and pulsatile left ventricular assist devices. *J Thorac Cardiovasc Surg*. 2009;137:208-15.
 10. Birks EJ, Tansley PD, Hardy J, George RS, Bowles CT, Burke M, et al. Left ventricular assist device and drug therapy for the reversal of heart failure. *N Engl J Med*. 2006;355:1873-84.
 11. Li YY, Feng Y, McTiernan CF, Pei W, Moravec CS, Wang P, et al. Downregulation of matrix metalloproteinases and reduction in collagen damage in the failing human heart after support with left ventricular assist devices. *Circulation*. 2001;104:1147-52.
 12. Amir O, Radovancevic B, Delgado RM III, Kar B, Radovancevic R, Henderson M, et al. Peripheral vascular reactivity in patients with pulsatile vs axial flow left ventricular assist device support. *J Heart Lung Transplant*. 2006;25:391-4.
 13. Westaby S, Bertoni GB, Clelland C, Nishinaka T, Frazier OH. Circulatory support with attenuated pulse pressure alters human aortic wall morphology. *J Thorac Cardiovasc Surg*. 2007;133:575-6.
 14. Pirbodaghi T, Asgari S, Cotter C, Bourque K. Physiologic and hematologic concerns of rotary blood pumps: what needs to be improved? *Heart Fail Rev* 2013 [Epub ahead of print].
 15. Cowger J, Pagani FD, Haft JW, Romano MA, Aaronson KD, Kolias TJ. The development of aortic insufficiency in left ventricular assist device-supported patients/clinical perspective. *Circ Heart Fail*. 2010;3:668-74.
 16. Waters B, Sample A, Smith J, Bonde P. Powering a ventricular assist device (VAD) with the Free-Range Resonant Electrical Energy Delivery (FREE-D) System. *Proc IEEE*. 2012;100:138-49.
 17. Waters B, Sample A, Smith J, Bonde P. Toward total implantability using free-range resonant electrical energy delivery system: achieving untethered ventricular assist device operation over large distances. *Cardiol Clin*. 2011;29: 609-25.
 18. Pirbodaghi T, Weber A, Axiak S, Carrel T, Vandenbergh S. Asymmetric speed modulation of a rotary blood pump affects ventricular unloading. *Eur J Cardiothorac Surg*. 2013;43:383-8.
 19. Bishopric NH. Evolution of the heart from bacteria to man. *Ann N Y Acad Sci*. 2005;1047:13-29.
 20. Akimoto T, Yamazaki K, Litwak P, Litwak KN, Tagusari O, Mori T, et al. Rotary blood pump flow spontaneously increases during exercise under constant pump speed: results of a chronic study. *Artif Organs*. 1999;23:797-801.
 21. Frazier OH. Unforeseen consequences of therapy with continuous-flow pumps. *Circ Heart Fail*. 2010;3:647-9.
 22. Parnis SM, Conger JL, Fuqua JM Jr, Jarvik RK, Inman RW, Frazier OH, et al. Progress in the development of a transcutaneously powered axial flow blood pump ventricular assist system. *ASAIO J*. 1997;43:M576-80.
 23. Waters T, Allaire P, Tao G, Adams M, Bearnson G, Wei N, et al. Motor feedback physiological control for a continuous flow ventricular assist device. *Artif Organs*. 1999;23:480-6.
 24. Choi S, Antaki JF, Boston JR, Thomas D. A sensorless approach to control of a turbodynamic left ventricular assist system, Control Systems Technology. *IEEE Trans*. 2001;9:473-82.
 25. Boston JR, Simaan MA, Antaki JF, Yu Yih-Choung. Control issues in rotary heart assist devices. American Control Conference, 2000. *Proc 2000*. 2000;5: 3473-7.
 26. Yoshizawa M, Sato T, Tanaka A, Abe K, Takeda H, Yambe T, et al. Sensorless estimation of pressure head and flow of a continuous flow artificial heart based on input power and rotational speed. *ASAIO J*. 2002;48:443-8.
 27. Ayre PJ, Vidakovic SS, Tansley GD, Watterson PA, Lovell NH. Sensorless flow and head estimation in the VentrAssist rotary blood pump. *Artif Organs*. 2000;24: 585-8.
 28. Tsukiya T, Akamatsu T, Nishimura K, Yamada T, Nakazeki T. Use of motor current in flow rate measurement for the magnetically suspended centrifugal blood pump. *Artif Organs*. 1997;21:396-401.
 29. Yoshizawa M, Sato T, Tanaka A, Abe K, Takeda H, Yambe T, et al. Sensorless estimation of pressure head and flow of a continuous flow artificial. *ASAIO J*. 2002;48:443-8.
 30. Karantonis DM, Cloherty SL, Mason DG, Ayre PJ, Lovell NH. Noninvasive pulsatile flow estimation for an implantable rotary blood pump. *Conf Proc IEEE Eng Med Biol Soc*. 2007;2007:1018-21.
 31. Giridharan GA, Sklar M. Physiological control of blood pumps using intrinsic pump parameters: a computer simulation study. *Artif Organs*. 2006;30:301-7.

Discussion

Dr Vivek Rao (Toronto, Canada). You mentioned at the outset that this is a physiologic controller system, but it seems to me you are still setting your speed and algorithms for inducing pulse. Does your controller allow you to modify it on the basis of the native heart rate? If your patient is exercising and the native heart rate increases, will this controller automatically change your ventricular assist device setting?

Dr Asgari. Yes, the system is synchronized with ECG that is measured from ECG sensors, so the LVAD will respond to increases in heart rate associated with day-to-day physiologic stresses, such as exercise, climbing stairs, or Valsalva maneuver. The decreases in heart rate associated with diurnal variation (eg, sleeping or resting) will be detected, and the LVAD will respond accordingly. One can define different percentages of systole and diastole and automatically vary their duration on the basis of the heart rate. The controller can run the LVAD in a co-pulsation, counterpulsation, or fixed mode of operation. This way, it is truly a pump that responds to physiologic demands.

APPENDIX: DESIGN OF A LEFT VENTRICULAR ASSIST DEVICE

The stationary part in a typical rotary LVAD is the electromagnet, called the “stator.” The rotating part contains a permanent magnet, called the “rotor.” Permanent magnets are made of magnetic materials, such as neodymium, iron, and boron. Each magnet has 2 poles: North and South. The North pole of a magnet is the pole that, when the magnet is freely suspended, points toward the Earth’s North magnetic pole in the Arctic. An electromagnet is made from a coil of wire that acts as a magnet when an electric current passes through it, but stops being a magnet when the current stops. The coil often is wrapped around a core of “soft” ferromagnetic material, such as steel, which greatly enhances the magnetic field produced by the coil. By changing the direction of the current that goes through the electromagnet, the magnetic poles will change.

When 2 magnets are brought close to each other, similar poles repel each other and opposite poles attract. For example, in [Appendix Figure 1, A](#), the stator’s electromagnets are magnetized. Therefore, the opposite poles of the electromagnets and permanent magnets attract each other.

In [Appendix Figure 1, B](#), the direction of the current in the coils is reversed, and therefore opposite poles are formed. Now the rotor’s poles are repelled by the electromagnet’s poles and a rotational movement is generated. In [Appendix Figure 1, C](#), the opposite poles are aligned again. With this method, one can easily control the rotation of a rotor.

Current rotary blood pumps are made of a 3-phase brushless direct current motor, that is, they have 3 sets of electromagnetic coils. Some LVADs (eg, the HeartMate II, Thoratec Co, Pleasanton, Calif) have impellers made of permanent magnets with 2 poles, whereas some have 4 poles (eg, the HeartWare Ventricular Assist Device, HeartWare Inc, Framingham, Mass). [Appendix Figure 1, D](#), displays a 3-phase brushless direct current motor with 2-pole permanent magnets. The repulsion and attraction forces of the stator poles and rotor poles will generate the rotational movement needed to propel the blood.

[Appendix Figure 1, E](#), shows a simplified illustration of brushless direct current motor construction. To control the movement of the impeller, we can use 6 switches to control the direction of the currents that go through electromagnets and therefore choose when an electromagnet is a north or south pole.

APPENDIX FIGURE 1. A-C, The step by step movement of a rotor in a simple 2-phase brushless motor. When current passes through the electromagnets (wire winding around an iron core), they will be magnetized and their poles will attract the opposite poles of the permanent magnet (rotor). Once the direction of the current is reversed, the poles of the electromagnets will be reversed and, therefore, like poles of the electromagnet and rotor will be aligned. This would create a force that causes the rotor to move. The rotor moves until its poles are facing the opposite poles of the electromagnet. By alternating the direction of the current, we can therefore rotate the rotor. D, A simplified version of a 3-phase brushless direct current motor that is a basis of a left ventricular assist device. E, The schematic of the electronic switches that control the direction of the current that passes through the electromagnets of a left ventricular assist device. The schematic view of how electromagnets are wound is shown on the *right side*.