Ventricular assist devices now clinically used for treatment of end-stage heart failure require responsive and reliable hemodynamic control to accommodate the continually changing demands of the body. This is an essential ingredient to maintaining a high quality of life. To satisfy this need, a control algorithm involving a trade-off between optimal perfusion and avoidance of ventricular collapse has been developed. An optimal control strategy has been implemented in vitro that combines two competing indices: representing venous return and prevalence of suction. The former is derived from the first derivative of diastolic flow with speed, and the latter derived from the harmonic spectra of the flow signal. The responsiveness of the controller to change in preload and afterload were evaluated in a mock circulatory simulator using a HeartQuest centrifugal blood pump (CF4b, MedQuest Products, Salt Lake City, UT). To avoid the need for flow sensors, a state estimator was used, based on the back-EMF of the actuator. The multiobjective algorithm has demonstrated more robust performance as compared with controllers relying on individual indices. ASAIO Journal 2005; 51:000–000.

Congestive heart failure remains a leading cause of morbidity and mortality around the world, and heart transplantation, the only accepted method to treat severe cases of the disease, cannot meet the vast demand. Consequently, considerable effort has been put into development of implantable prosthetic heart substitutes and ventricular assist devices (VADs).

Because the loading conditions are not static, these devices typically require feedback control. As patients are rehabilitated and return to a normal, active lifestyle, the demands for flow capacity and responsiveness can challenge conventional control. As the importance of the physiologic (demand-based) control for rotary blood pumps has now been recognized, modern control concepts have started to be evaluated for these applications.1–11

Unlike pulsatile VADs that fill passively, the second generation of VADs are based on turbodynamic pump technology and do not intrinsically respond to the availability of incoming blood. Furthermore, their characteristic response to alterations in hydrodynamic load is precisely opposite to the requirement of the body: their pressure generation decreases with flow, whereas the body demands increased flow for increased pressure. Therefore, the speed of the pump must dynamically respond to changes in both preload and afterload to meet the demand for perfusion while avoiding the risk of outstripping the available blood flow capacity. The consequences of both underpumping it and overpumping it will limit the quality of life and can pose serious risks to the health of the patient. Underpumping, for example, may cause pulmonary congestion, and overpumping may cause the heart and vessels to collapse.

Therefore, the basic objective of a controller for a pump operating as a LVAD (LVAD) is to operate within a range of flow (hence speed) that matches the blood flow capacity of the right ventricle. This, in turn, requires the pump to operate within proximity of suction but without actually achieving suction. In the absence of sensors to directly detect the immensity of suction or the hemodynamic load, we have introduced a variety of indices, including a pulsatility index, a venous return index, and harmonic spectral index as surrogates.3,5,11

In this article, the results of in vitro studies are reported that evaluate these indices, both independently and in combination. Experiments were conducted using a HeartQuest centrifugal blood pump (MedQuest Products, Salt Lake City, UT) in an apically cannulated LVAD configuration within a mock circulatory simulator. The response to changes in preload and afterload are presented.

### Materials and Methods

#### Robust Multiobjective Control

Previous in vivo studies with rotary LVADs have revealed correspondences between several features of the flow waveform and the appearance of suction.10 Those observations suggest that these features may be used to provide closed feedback control algorithms. A harmonic suction index (HSI) has been presented based on the observation of altered morphology of the flow waveform near suction. Specifically, higher-order harmonic content is amplified as suction occurs. Because of the continued beating of the native heart, the flow waveform is observed to vary periodically with the cardiac
cycle. However, as suction approaches, it becomes increasingly asymmetric.\textsuperscript{5,10} The HSI is, therefore, defined as the ratio of the power of harmonics above the fundamental to the total power:

\[
\text{HSI} = \frac{\sum_{i=2} a_i}{\sum_{j=1} a_j}
\]

where \(a_i\) is the coefficient of the Fourier power spectrum of the flow waveform.

A venous return index (VRI) has been defined as the first derivative of pump flow with respect to pump speed. As pump speed increases, the mean flow rate increases at an incrementally decreasing rate. When venous return is matched by the pump, the rate of change is approximately zero. Ventricular suction occurs within close proximity of this point.

Although VRI is particularly sensitive for flows below the suction point, it is relatively insensitive above this point. Conversely, HSI is more sensitive above the suction point. Therefore, it is advantageous to combine the two indices to achieve added robustness to the control.

Furthermore, clinicians typically may wish to satisfy additional, sometimes competing objectives, such as:

- Left atrial pressure should be maintained below approximately 10 to 15 mm Hg to avoid pulmonary edema.
- Systolic arterial pressure should be maintained between patient-specific limits to avoid consequences of hypertension (and hypotension).
- However, VRI and HSI do not explicitly consider atrial or arterial pressure. If afterload is high, excessive arterial pressure can be developed when the pump produces maximum flow.\textsuperscript{5} This motivates the use of multiple indices or the multiobjective control approach.\textsuperscript{1,2,5}
- Our group has studied various approaches to combining indices. Baloa \textit{et al.}\textsuperscript{5} proposed statistical methods, and Boston, Antaki \textit{et al.}\textsuperscript{1} introduced penalty-based convolution of indices. However, as a preliminary evaluation of feasibility, we implement a simple weighted sum of two indices (HSI and VRI).

\textit{In Vitro Experimental Setup}

\textbf{Mock Circulatory System.} Figure 1 shows a diagram of the \textit{in vitro} experimental setup. A flexible sac within a pressurized contained rigid chamber represents the left ventricle. The compressed air source is controlled by pneumatic driver to create the pumping action of the sac. Two accumulators are used to represent the pulmonary and systemic compliances. A needle valve between accumulators is used to impose the systemic resistance (afterload). To mimic the ventricular suction phenomenon, a collapsible “suction element” is inserted between the left ventricle and the LVAD (Figure 2). Preload is varied by adjusting the valve between the ventricle and the suction element.

A transit time flowmeter (Transonic, Ithaca, NY) is used to measure flow through the LVAD and a differential pressure transducer is used to measure the pressure rise across the LVAD, a magnetically levitated centrifugal pump (HeartQuest CF4b, MedQuest Products). Data I/O is achieved at 200 Hz using an A/D D/A board (Quanser MultiQ). The acquisition and control algorithms are programmed using WinCon real-time software and MATLAB/Simulink. The output of the controller is a reference voltage representing the speed setpoint, and the inputs include the electrical current from the motor and the output of the motor drive tachometer.

\textbf{Suction Element.} To simulate the anatomic collapse of the ventricle caused by negative pressure, a “suction element” was fashioned by inserting a collapsible tube within a rigid cylinder (Figure 2). The space there between was regulated with various levels of vacuum to simulate corresponding degrees of suction susceptibility.

\textbf{Flow Estimator}

Based on the uncertain reliability of indwelling sensors over several years as required by a long-term VAD use, sensorless approaches, using model-based estimators, have been pursued...
by our group and by others.9–12 The use of brushless-DC motors for the LVAD actuator provides an opportunity to relate back-EMF to load, and thereby to estimate pressure and flow from current and speed.11,12 Briefly, the form of the state estimator used in these studies is represented by a set of differential equations:

\[ \frac{d\Omega}{dt} + K_\Omega T_f + \frac{1}{\eta} H Q = K_I I \]  

(1)

\[ H = f(Q, \Omega) \]  

(2)

\[ \eta = f(H, Q, \Omega) \]  

(3)

where \( J \) is the net moment of inertia (motor and load), \( \Omega \) is the angular speed, \( K_t \) is the torque constant of the motor, \( T_f \) is the friction torque, \( H \) is the pressure head of the pump, \( I \) is the motor current, and \( \eta \) is the hydraulic efficiency. Equations 2 and 3 are two empirical relationships used for the hydrodynamics of the pump, both in nondimensional form according to widely accepted formulas.13 In other efforts,11,14 Equations 2 and 3 have been approximated by reduced-order polynomials resulting in the following form for the estimator:

\[ \frac{dQ}{dt} = \left( c_0 + c_1 \frac{\Omega}{\Omega} + c_2 \Omega + c_3 \frac{d\Omega}{dt} + c_4 \frac{d\Omega}{dt} + c_5 Q \right) \]  

(4)

\[ \frac{dI}{dt} = \left( c_6 + c_7 \frac{I}{I} + c_8 I \right) \]  

(5)

Because the calculation of VRI is based on derivative of flow, \( Q \), the low-order polynomial coefficients expressed in Equation 5 were regressed to experimental data; but these resulted in unsatisfactory fit. By directed trial-and-error, we obtained an alternative realization of the estimator using a lookup table with cubic interpolation as follows:

At constant speed \( \left( \frac{d\Omega}{dt} = 0 \right) \), Equation 1 becomes

\[ K_\Omega T_f + \frac{1}{\eta(H, Q, \Omega)} H Q = \frac{dH}{dt} Q = \frac{d\Omega}{dt} I = \left( \frac{\Omega}{K_I} \right) \frac{T_{net}}{K_I} \]  

(6)

where \( T_{net} \) is the net torque. Equation 6 implies that flow can be obtained from an interpolation of measured data using quasi-static speed and current information only. To account for the dynamics ignored in the quasi-static assumption, the net torque in Equation 6 is amended with a dynamic term as in Equation 7:

\[ T_{net} = K_I I - \frac{d\Omega}{dt} \]  

(7)

Hence, the alternate form of the flow estimator can be expressed as:

\[ Q = \frac{d\Omega}{dt} I \left( \Omega, I - \frac{\Omega}{K_I} \right) \]  

(8)

The lookup table for cubic interpolation of flow versus current and speed is shown in Figure 3.

Figure 4 shows a representative sample of the flow estimator performance for MedQuest CF4b pump operating at 1,500 rpm in water. The estimator was found to track the mean, maximum, and minimum peaks over the specified range of operating speeds in which flow is positive. This enabled the feedback control algorithm to use the estimated flow rather than the measured flow. (Note, however, that the performance of the flow estimator is not guaranteed for regurgitant (reverse) flow; accordingly, the lookup table in Figure 3 is not defined in this region.)

Implementation and Validation of Control Algorithm

The calculation of VRI is defined as the first derivative of flow with respect to speed. This is based on the relationship of flow to speed, as depicted in Figure 5. This is typically a

The lookup table for cubic interpolation of flow versus current and speed is shown in Figure 3.
unimodal relationship, in which the extremum of flow is found in the neighborhood of the suction zone. The slope of the relationship may be found through the chain rule:

$$\frac{dQ}{dN} = \frac{dQ}{dt} \frac{dt}{dN}$$  \hspace{1cm} (9)$$

where $N$ is the pump speed. However, at constant pump speed

$$\frac{dt}{dN} \rightarrow \infty$$  \hspace{1cm} (10)$$

Therefore, real-time calculation of VRI requires a continual change (modulation) of speed. Hence we applied sinusoidal dither signal with small amplitude ($\pm$ 50 rpm) and excitation frequency at 1 rad/s. The excitation frequency of the dither signal was set to be slow; almost one decade below the heart rate (11.07 rad/s assuming a heart rate of 70) to uncouple from the fundamental cardiac modulation of flow, and to average the response over several cardiac cycles.

As depicted in Figure 5, at low operating speed $N_{low}$ the dither signal creates large fluctuation in the flow $Q$, but small fluctuation at high operating speed $N_{high}$. Assuming dither signal amplitude is small enough to guarantee the linearity of the slope, the amplitude of the flow fluctuation is directly proportional to the slope at the operating point. Hence, without using numerical differentiation, the flow fluctuation following dither excitation can be used to calculate VRI. The flow fluctuation signal is passed through the signal conditioning loop depicted in Figure 6 to calculate DC level VRI.

The pass band for band-pass filter was set at approximately 0.16 Hz (approximately 1 rad/s) to sense variations caused by dither signal only. The low-pass filter cutoff frequency was set at 0.015 Hz to compute the DC baseline of the slope.

The second index, HSI (defined above), was approximated by the power of the second harmonic to the fundamental. Figure 7 shows the calculation process of the harmonics component. The first and second harmonics of the flow signal is obtained by passing the estimated flow signal through a low-pass filter first. A notch filter was then used to eliminate the first harmonic at approximately 1.16 Hz (assuming a heart rate of 70). The cutoff frequency of the low-pass filter for DC leveling was set to approximately 0.042 Hz. This approach was adopted for convenience, and is applicable only to this controlled in vitro experiment wherein the fundamental frequency is known in advance. When implemented in vivo, additional signal processing would be required to adapt the filters based on the heart fundamental frequency.

Controllers using these two indexes were implemented using proportional-integral (PI)-type control loops. The integrator in the PI controller acted as the primary pump speed set point. The initial condition for the integrator was set to the pump startup speed. The controller was implemented to generate an immediate pump speed change based on the error between the measured speed and the set point via the proportional gain. The integral gain causes the integrator speed set point to slew in the direction to correct the error. When the pump speed reaches a new set point, minimizing the error, the proportional term ceases to effect a change in pump speed while the integrator locks the new equilibrium speed set point. To prevent integrator windup when using the combined indices, saturation limits were imposed on the integrators. (See Figure 8 for the block diagram of the controller.)

Mock-Circulatory Loop Experiments

In vitro studies were performed using a HeartQuest centrifugal pump in a LVAD configuration within a mock circulatory loop (Figure 1). The inflow cannula to the pump was modified with a collapsible section (suction element, Figure 2) to simulate the hemodynamics of ventricular collapse.

Figure 9 presents a typical set of hemodynamic data for open-loop control of the pump through its full operating range: from 0 rpm (demonstrating regurgitant flow) through maximum speed (demonstrating suction). The online calculation of control signals, VRI and HSI, is also presented. Here it is important to note that VRI is most sensitive to underpumping conditions.
at low speed, whereas HSI is most sensitive to overpumping (suction) conditions at high speed.

We evaluated the response of the closed-loop controller under two disturbances: (1) step changes in preload caused by closing the venous return valve, and (2) change in afterload caused by closing the corresponding systemic resistance valve.

Figures 9–13 illustrate the system response under closed-loop control described previously. Figure 10 demonstrates feedback control based on VRI alone, using a set-point of 0.0013. In this experiment, afterload was suddenly increased, causing an initial transient decrease in flow. The controller compensated by increasing speed to normalize flow and return VRI to normal. This is to be contrasted with open-loop response, where an increase in SVR, caused by, for example, sympathetic response, would cause a drop in flow and elevation in inflow pressure (thus LAP)—which is precisely the opposite response as would be needed. Figure 11 shows the open loop response to step decrease in preload. Pump flow and inlet pressure drops suddenly and suction insues as a result. HSI is shown to increase due to the suction and is maintained high because pump speed is maintained high.

Figure 12 demonstrates feedback control based on HSI alone. In this experiment, the preload was reduced, and the controller responded by dramatically reducing speed (rpm) to

![Figure 8. Saturating PI controllers showing equal weighting for the HSI and VRI.](image-url)

![Figure 9. Open-loop response to ramp speed increase demonstrating behavior of control indices: HSI and VRI.](image-url)

![Figure 10. Closed-loop response of VRI control to step increase in afterload, in vitro.](image-url)
mitigate ventricular suction. However, even at reduced speed, there is still evidence of suction, indicated by downward spikes in the flow signal, and negative inflow pressure. Nevertheless, the controller behaved rapidly and stably to reduce speed.

This observed settling error is suspected to be related to the sensitivity of the HSI used in this experiment. Future development of this controller will require additional investigation of the sensitivity of different forms of HSI to this phenomenon, and consideration of the set point for eliminating suction. In this experiment, the set point for HSI was 0.09. Ideally, it should be set to near-zero.

Note that there may be circumstances when complete elimination of suction may result in excessive reduction of speed, thereby reducing flow and perfusion pressure dangerously low. This is our motivation for adding pressure and/or flow to the overall control objective. We aim to explore this in the future. For the purpose of the current study, we explored the benefit of combining VRI to HSI to counteract the tendency to reduce speed excessively.

**Figure 13** demonstrates the performance of the feedback controller based upon the linear combination of two separate PI-type controllers for HSI and VRI control, respectively. Here, two controllers are equally weighted. We observed avoidance of suction while still maintaining perfusion flow.

In this case, at the start of the data record, the VRI loop is commanding full speed, because the VR index has not yet reached its set point. Because there is no suction, the HSI loop has reached equilibrium, with a speed set point of 1,700 rpm. The linear combination of the two set points gives the initial speed of 2,200 rpm. When the pump preload decreased, the HSI index rose because of the onset of suction, which caused the HSI loop to decrease motor speed and decrease the HSI pump speed set point. The system reached its new equilibrium speed when the HSI loop saturated at 0 speed command, leaving the VRI loop controlling the pump speed. The VRI loop prevented the HSI loop from excessively reducing pump speed.

The weighting coefficient for the linear combination of the two separate indices can be tuned for a better performance because the sensitivity of each controller is different depending on the operating condition. Saturation limits and controller gains are also candidates for further tuning. Fuzzy logic could be considered to mix indices that have validity for different regions.

**Conclusions**

In vitro studies were performed to validate the hemodynamic control indices VRI and HSI for suction avoidance in the operation of a rotary LVAD. Cubic spline interpolation was implemented for flow estimation and, accordingly, the calculation of the control indices. Calculation of VRI was obtained by dither modulation of the speed, which obviated numerical differentiation. An in vitro suction element made of a collapsible balloon was found to be effective to create the suction phenomenon in vitro and to validate the suction-avoidance of the physiologic controller.
A physiologic controller based on the VRI and HSI indices showed a synergistic benefit as compared with control based on a single index. It was shown that those indices have different sensitivities for different operating conditions (pump speed) so that their combination could provide additional controllability of the system. The success of this preliminary implementation of multiobjective control demonstrates the validity of this approach for achieving a trade-off between optimal LV unloading and avoidance of suction.

Ongoing research is seeking to implement more robust means of identifying the dynamically changing indices, combining these indices, and driving the system to its set point. It is hoped that this approach will provide additional robustness without introducing unnecessary complexity.

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References