# Minimally Invasive Estimation of Cardiac Function for Patients with Rotary VAD Support

Yih-Choung Yu[1] and Pujitha Weerakoon[2]
[1]Department of Electrical and Computer Engineering, Lafayette College, Easton, PA 18042 USA
[2]Department of Biomedical Engineering, Yale University, New Haven, CT 06520, USA

***Abstract*** - Maximum ventricular elastance, $E_{MAX}$, is a reliable quantitative index representing the contractial status of a patient's heart. However evaluating $E_{MAX}$ usually requires invasive pressure and flow sensors, which only can be performed under certain clinical facility. If an indirect index of $E_{MAX}$ can be identified from the measurements of a ventricular assist device (VAD) without any indwelling sensor, this would facilitate an effective way to monitor the healthy condition of the patient's heart while the patients are under VAD support. This index can also be used to determine the control strategy of VAD operation and gradually wean the patient from the mechanical circulatory support.

In this paper, two possible indices, pump flow pulsatility and arterial pulse pressure, were evaluated as alternative representations of $E_{MAX}$ using data from a computer simulation model of cardiovascular system with a HeartMat II left ventricular assist system (LVAS). Pump flow pulsatility showed a strong correlation to $E_{MAX}$ regardless of pump speed changes, and thus can be used as an index for $E_{MAX}$.

## I. Introduction

Heart disease is a major health problem in the United States and throughout the world. Although heart transplant is an acceptable method to treat patients with severe heart failure, the demand for heart transplants exceeds the supply. For many patients, a VAD could offer a satisfactory alternative as a "bridge" to heart transplant.

Recent success at several centers in rehabilitating patients under VAD support has encouraged efforts to provide support systems that allow patients to return to normal lifestyle. A few patients have experienced sufficient improvement in their cardiac functions during VAD support that the devices were removed and their native hearts returned to normal function [1]. Explantation represents the ideal outcome for cardiovascular patient because the problems of managing a permanently implanted device or a transplanted organ are avoided. Improving this ideal outcome of VAD explantation requires a proper timing to wean the patient from VAD support [2].

Transesopheageal echocardiograph (TEE) is usually used to determine the patient's cardiac function, while the patient is assisted by VAD. This test usually needs to intervene with the device operation to reduce the level of mechanical support and observe the native heart response to the load changes, which may increase the risk of thrombus formation [2]. Although $E_{MAX}$, defined by [2],

$$E_{MAX} = \frac{LVP_{ES}}{LVV_{ES} - V_0}, \tag{1}$$

where $LVP_{ES}$ and $LVV_{ES}$ are end-systolic left ventricular (LV) pressure and volume, and $V_0$ is the unstressed LV volume, is a consistent index to represent ventricular contractility, it is difficult to measure under most clinical environments due to the need of invasive sensors for pressure and flow measurements.

Mandarino et al. [3] estimated $E_{MAX}$ by using end-systolic arterial pressure and ventricular chamber area (from TEE) measurements to substitute $LVP_{ES}$ and $LVV_{ES}$ in (1). However, femoral arterial pressure measurement can't be obtained non-invasively. Also, this approach is only applicable under certain clinical facility due to the need of TEE. Nitta et al. [2] varied the pump operation mode as the changes of ventricular load conditions and characterized $E_{MAX}$ using pump inflow and end-systolic aortic pressure measurements. Although this technique can be performed without interrupting VAD operation, the use of indwelling sensors was not completely avoided. Kikugawa [4] and Takahashi et al. [5] observed the correlation between motor current waveform and the amplitude of the left ventricular pressure to determine $E_{MAX}$. Since, left ventricular pressure amplitude is heavily dependant to the pump speed of the VAD, the consistency of using LV pressure to represent cardiac function while the pump speed is changing is still uncertain. Kinichi et al. [6] observed the correlation between the derivative of the left ventricular pressure and the external work provided by the actuator of the pump to determine $E_{MAX}$. However, this method requires that the pump speed to be regulated to 0 L/min at the diastolic phase, which interrupts the normal VAD operation. This could be potentially life threatening to patients.

This paper evaluates two possible indices: pump flow amplitude and arterial pressure pulses to represent $E_{MAX}$ using data from a computer simulation of a cardiovascular system with HeartMate II axial-flow LVAS (Thoratec Co., Pleasanton, CA). Since, the pump flow from the HeartMate II can be estimated using pump speed and current signals from the VAD control console [7] and arterial pressure pulse can be obtained from cuff pressure measurement [8], these $E_{MAX}$ indices could be obtained non-invasively. Sensitivity of these indices to $E_{MAX}$ changes was evaluated using computer-generated data simulating various pump operations as well as different cardiac functions.

## II. System Description

The HeartMate II LVAS, shown in Fig. 1, is an axial flow pump that has a spinning rotor as its only moving part. It has a left ventricular apical inflow cannula with a sintered titanium blood-contacting surface. The bladed impeller spins on a bearing and is powered by an electromagnetic motor. No compliance chamber or valves are necessary, and a single driveline exits the right lower quadrant of the abdomen. The inlet cannula is placed in the ventricular appendage, and the pump is placed either intraperitoneally or extraperitoneally. The outflow cannula is connected to a Dacron graft, which is then anatomized to the ascending aorta. Pump operation is controlled by setting pump rotational speed from 6,000 to 15,000 rpm to deliver as much as 10 L/min of cardiac output.

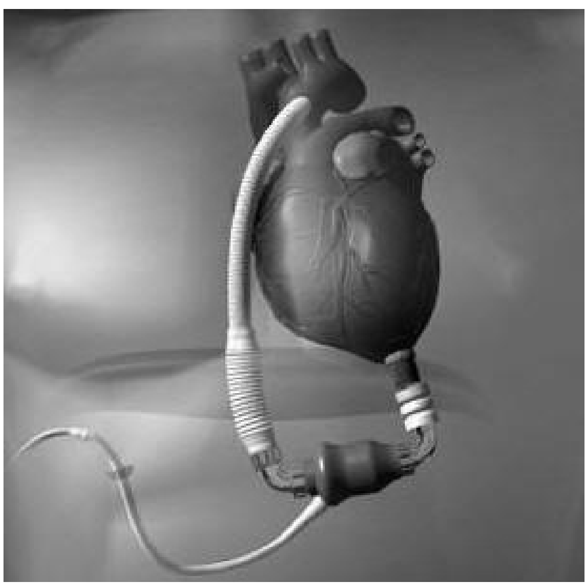

Figure 1, HeartMate II LVAS

An electrical analogue of the cardiovascular system along with the HeartMate II LVAS is shown in Fig. 2. The LVAS model was adopted from [7], which characterized the relationship of pressure difference between ventricular pressure and arterial pressure, H, the blood flow, Q, through the pump, and the pump speed ω as

$$\frac{dQ}{dt} = -\left(\frac{A_P + A_C}{B_P}\right)\cdot Q - \frac{A}{B_p}\cdot\omega^2 + \frac{1}{B_P}\cdot H , \qquad (2)$$

where $A_P$, $B_P$, $A_C$ and A are pump parameters. The cardiovascular model was modified from [9] to include the baroreceptor of arterial pressure regulation and septum compliance. In human bodies, heart rate and systemic vascular resistance ($R_{A3}$ in Fig. 2) are controlled by "sensors" (a.k.a. baroreceptors) in the circulation that provide a feedback mechanism to regulate mean arterial pressure (mean $P_{A3}$ in Fig. 2). Mathematical model of the baroreceptors was adopted from [10]. Anatomically, the pressure of one ventricle is affected by the pressure in other ventricle due to the shift of the septum between both ventricles. This interaction was modeled by the septum compliance, $E_S$, in Figure 2 [11]. Performance of the model to simulate the hemodynamics of the cardiovascular system with or without the LVAS in response to various physiologic conditions was validated by comparing the simulation data with the data described in literatures. The pressure–volume relationships of the ventricles in simulation were consistent to $E_{MAX}$ settings with strong linearity, regardless of the preload and afterload changes [12]. Stroke work produced by the left ventricle was linear to its end-diastolic ventricular volume. The slop of this linear relationship was sensitive to $E_{MAX}$ settings and insensitive to afterload ($R_{A3}$) changes [13].

## III. Identification methods

Due to the valve-less design of the rotary VAD, the pulsatilities of blood pressures and flows from a rotary VAD patient would be affected by the strength of the native heart. A stronger heart would introduce more significant pulsatilities to the hemodynamics at the same level of VAD support, while increasing VAD speed reduces the load of the patient's heart and thus the hemodynamic pulsatilities due to the beating heart would be lower. Hence, the pulsatilities of some hemodynamic variables could be used as indices representing $E_{MAX}$.

Two hemodynamic signals, pump flow and arterial pressure, were used to evaluate their sensitivities to $E_{MAX}$ changes. These signals were chosen because they are close to the left ventricle, and thus, are more sensitive to the disturbance produced by the beating heart. In addition, these two signals could be obtained non-invasively. Pump flow can be estimated by using pump current and speed signals [7], while arterial pressure is available indirectly by calibrating the measurement from finger pulses [8].

The pulsatility was measured by using windows to segment the collected data. Figure 3 shows how the segmentation of the data was determined. Since the predominant frequency in the collected data is the heart rate, each window corresponds to one cardiac cycle with frequency $W_{HR.}$ Pulsatility of a signal was defined by,

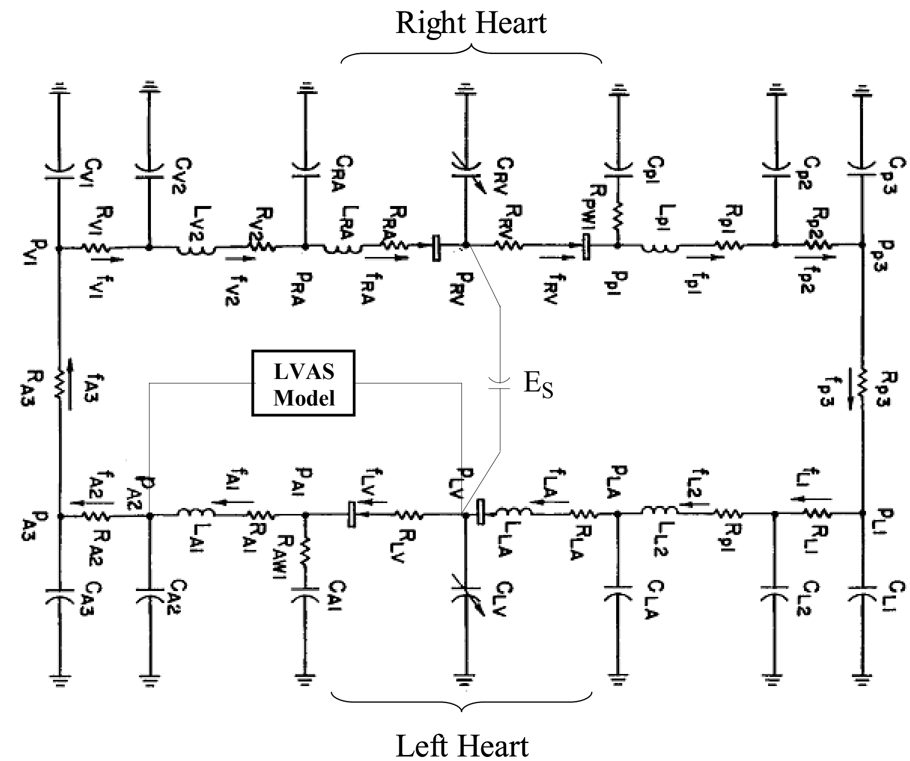


Figure 2, Electrical analogue of the cardiovascular system and the HeartMate II

$$\text{Index(mean)} = \left[\sum_{k=1}^{N} \frac{\Delta S(k)}{\overline{S}(k)}\right] / N, \qquad (3)$$

where $\Delta S(k)$ is the difference between the maximum and the minimum of the signal (pump flow and arterial pressure) in the $k^{th}$ segment (or cardiac cycle) and $\overline{S}(k)$ is the mean value of the signal in the $k^{th}$ segment. N is the total number of segments used for the index calculation.

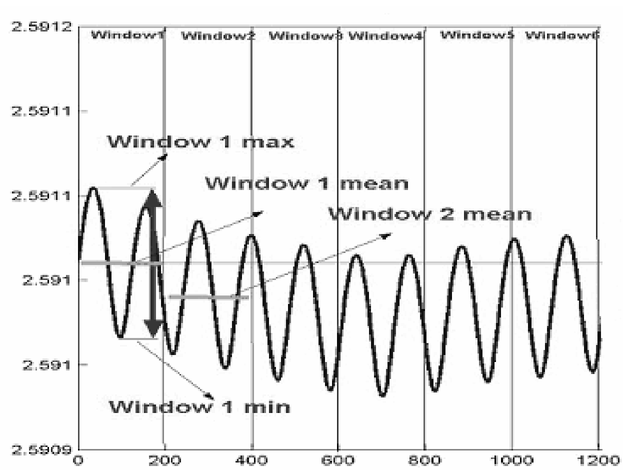


Figure 3, Segment of data to measure pulsatility

Determination of the pulsatility index as defined in (3) requires human intervention, which can only be implemented in post-processing. In order to detect the pulsatility in real-time, a peak-detection algorithm [15] was carried out to determine the amplitudes of pump flow and arterial pressure signals. The amplitudes were then correlated with $E_{MAX}$ changes. Block diagram of the peak detection algorithm is shown in Figure 4. Since the dominant radian frequency $W_{HR}$ of the test signal (pump flow or arterial pressure) is related to the heart rate, $W_{HR}=2\pi \cdot HR/60$ (HR is the heart rate in beats/min), the test signal, $f(t)$, could be approximated as a sinusoidal signal,

$$f(t) = A \cdot \cos(W_{HR} \cdot t) + B, \qquad (4)$$

where A is the amplitude and B is the dc bias of the signal. Passing the signal through a high-pass filter eliminates the dc bias, B. Hence the input signal at the rectifier in Figure 4 becomes,

$$A \cdot \cos(W_{HR} \cdot t). \qquad (5)$$

The signal is subjected to half wave rectification through a diode, which is essentially equivalent to multiplying the input at the diode with a square waveform of amplitude 1 and frequency of $W_{HR}$. Hence, the output from the diode, VR, is

$$VR = \frac{A}{\pi} + \frac{A}{2} \cdot \cos(W_{HR} t) + \text{higher frequency terms}. \qquad (6)$$

Passing VR through a low-pass filter with a cutoff frequency lower value than $W_{HR}$ eliminates all frequencies components in (6) except the dc term $A/\pi$. Passing the output from the low-pass filter through an amplifier with a gain of $\pi$ results in the output of the block diagram to be A, which is the amplitude of the signal.

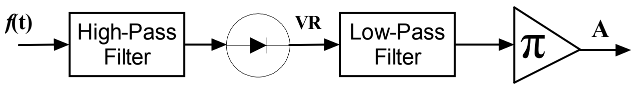


Figure 4, Block diagram of the amplitude detection algorithm

## IV. TEST RESULTS

The model shown in Fig.2 was implemented in SIMULINK to generate data for testing the $E_{MAX}$ indices as defined in (3) as well as the amplitude detection described in Eqs. (4), (5), and (6). The Runge-Kutta method with a variable step size using Dormand-Prince pair was applied for numerical integration. The value of $E_{MAX}$ was changed to 20%, 35%, 50%, 100%, and 150% of its nominal value (2.8 mmHg/ml) to simulate the severe failure, medium failure, recovered, healthy, and strong heart. A nominal value of 2.8 mmHg/ml was chosen for $E_{MAX}$ because it yielded the closest hymodynamics for a healthy human as described in [14]. Heart support was varied by setting the VAD speeds to 10000, 9200, and 8000 rpm to simulate high, normal, and low VAD support. Pump speed selected within this range would cause neither regurgitated flow nor ventricular collapse for all the chosen values of $E_{MAX}$.

Pump flow and arterial pressure from simulation under different $E_{MAX}$ and VAD speed settings were used to calculate their pulsatility indices defined in (3). The resulting indices are shown in Figure 5. Both indices were elevated when $E_{MAX}$ increased or pump speed decreased. Pulsatility indices from both signals showed significant increases in the initial phase of heart recovery.

The amplitudes of arterial pressure and pump flow extracted by the amplitude detection scheme in Figure 4 were recorded for the same test conditions. The high and low pass filters were implemented using the third-order Butterworth filters with cutoff frequencies at 0.5 Hz and 0.25 Hz respectively. The Butterworth filters were chosen due to the advantage of their flat pass-bands. The resulting amplitudes for pump flow are shown in Figure 6, and the amplitudes corresponding to arterial pressure is shown in Figure 7. Both amplitudes were elevated when $E_{MAX}$ increased or pump speed decreased. They demonstrated significant increases in the initial phases of heart recovery, similar to the indices obtained from equation (3).

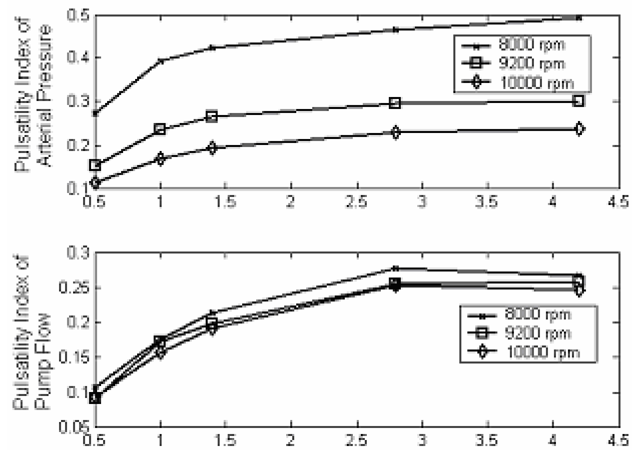


Figure 5, Pulsatility of arterial pressure and pump flow with different $E_{MAX}$ and pump speed settings

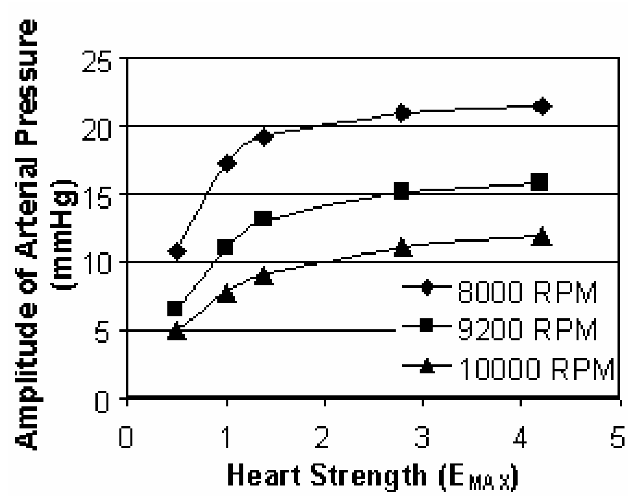


Figure 6, Amplitude of arterial pressure from peak detection with different $E_{MAX}$ and pump speed settings

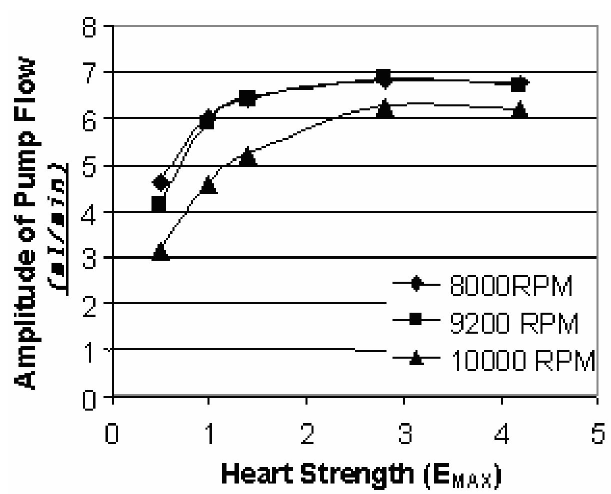


Figure 7, Amplitude of pump flow from peak detection with different $E_{MAX}$ and pump speed settings

## V. Discussion and Conclusion

Two pulsatiliy indices using arterial pressure and pump flow signals to estimate the left ventricular function of a patient with VAD support without any invasive sensor were evaluated. The possibility of using amplitude detection to estimate the cardiac function in real-time was also investigated. A computer simulation model of cardiovascular system with a HeartMateII LVAS was implemented to generate data for algorithm testing.

The index derived from pump flow was more consistent at the same $E_{MAX}$ regardless of pump speed changes as shown in Figure 5. In addition, the index reached its maximum value when $E_{MAX}$ was at its nominal value, which could be useful to provide the optimal timing to wean the patient from VAD support. Therefore, the pump flow pulsatility is a more appropriate index to represent patient's cardiac function. Since the pump flow can be estimated using pump speed and current signals from the VAD control consoles, there is no need for any invasive sensor to estimate the patient's cardiac function using this identification method. Moreover, performing this identification procedure does not require interrupting the normal VAD operation. This would improve patient's safety as well as reduce the cost of healthcare.

For VAD control purpose, the amplitude of pump flow can be used to act as an index of cardiac function. Since the amplitude detection can be carried out in real-time using a rectifier amplitude detector circuit, the cardiac function index estimator can be integrated with a physiologic VAD controller to gradually wean a patient from VAD support if the patient's native heart recovered.

In conclusion, this research lays a foundation for further investigations to identify a VAD patient's cardiac function in a non-invasive way. First, the results presented herein were based upon the hemodynamic data from a computer simulation. Further verification of the algorithms using clinical data would be an important step in the near future. Second, since the motor current is the energy source to drive the VAD, the turbulent produced by the native heart could cause fluctuation on the motor current. Therefore, it might worth further study to find the correlation of current pulsatility with $E_{MAX}$. Finally, although the algorithms to detect $E_{MAX}$ were developed based upon the HeartMate II LVAS, this identification scheme could be applicable to other rotary VADs, where the pumps are operated by speed control and the pump flow signal is available either from measurement or from estimation.

## References


[1] Thoratec's Heartbeat, D.J. Farrar, Ed., 10.2, August, 1996, pp.1.

[2] Nitta, S., et. al., "A Less Invasive Emax Estimation Method for Weaning from Cardiac Assistance," *IEEE Trans. On Biomed. Eng.,* 1995, pp. 1165-1172.

[3] Mandarino, W. A., J. Gorcsan III, T. A. Gasior, S. Pham, B. Griffith, and R. L. Kormos (1995). Estimation of Left Ventricular Function in Patients with a Left Ventricular Assist Device, *ASAIO J.*, *41*, M544-547.

[4] Kikugawa, D, "Motor Current Waveforms as an Index for Evaluation of Native Cardiac Function During Left Ventricular Support with a Centrifugal Blood Pump", Artificial Organs, 2001, pp. 703-708.

[5] Takahashi, K. et al., "Estimation of Left Ventricular Recovery Level Based on the Motor Current Waveform Analysis on Circulatory Support with Centrifugal Blood Pump", Artificial Organs, 2001, pp. 713-717.

[6] Kinichi, N. et al., "The Estimation of Cardiac Function from the Rotary Blood Pump", Artificial Organs, 2001, pp. 709-712.

[7] Choi, S., et al., "Modeling and identification of an axial flow blood pump," Proceedings of the American Control Conference, 1997, pp 3714-3715.

[8] Yu, Y., et al., "Minimally Invasive Estimation of Systemic Vascular Parameters," Annuls of Biomedical Engineering, 2001, pp. 595-606.

[9] V.C. Rideout, Mathematical and Computer Modeling of Physiological System, Prentice Hall Co., 1991.

[10] Xu, L., et al., "Computer Modeling of Interactions of an Electric Motor, Circulatory System, and Rotary Blood Pump," ASAIO Journal, 2000, pp. 604-611.

[11] Santamore, W.P. et. al., "Hemodynamic consequences of ventricular interaction as assessed by model analysis," American Journal of Physiology, 1991, H146-157.

[12] Sagawa, K. et al., "End-Systolic Pressure-Volume Ratio: a new index of ventricular contractility", The American Journal of Cardiology, 1977, pp. 748-753.

[13] Glower, D. et al., "Linearity of the Frank-Stirling relationship in the intact heart - the concept of preload recruitable stroke work", Circulation, vol. 71, 1985, pp. 994-1009.

[14] A. Guyton and J. Hall, Textbook of Medical Physiology, 9th Ed. W. B. Saunders Co., 1996.

[15] Lathi, B, "Modern Digital and Analog Communication Systems", Oxford University Press, 1998, pp.166-168.