Minimally Invasive Identification of Ventricular Recovery Index for Weaning Patient from Artificial Heart Support [§]

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Abstract - Maximum ventricular elastance, E_{MAX}, is a reliable quantitative index of the contractial state of the ventricle. It is a strong candidate to determine the healthy status of the patient's heart. However, estimating 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 using measurements from 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 patient is under VAD support. This index can also be used to control the VAD to gradually wean the patient from the mechanical circulatory support.

In this paper, three possible indices, systemic vascular resistance, maximum VAD inflow acceleration rate, and the maximum VAD inflow acceleration rate during heart ejection, were evaluated as a representation of E_{MAX} using Novacor VAD volume and mean arterial pressure measurements from a computer simulation. The maximum VAD inflow acceleration rate during heart systole showed a strong correlation to the E_{MAX} regardless the variation of native heart rate, and thus can be used as an E_{MAX} index.

1. Introduction

Heart disease is a major health problem in the United States and throughout the world. Although heart transplantation is an accepted method to treat severe cases of the disease, the demand for heart transplants exceeds the supply. For many patients, a left ventricular assist device could provide a satisfactory alternative to transplantation.

Recent successes at several centers in rehabilitating patients under VAD support have encouraged efforts to provide support systems that allow patients to return to normal lifestyle. A few patients have been experienced sufficient improvement in their cardiac function during ventricular 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 requires a proper timing to wean the patients 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 needs to turn on and off of the device, which would reduce the mechanical support and increase the risk of thrombus formation [2]. Although E_{MAX}, defined by

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

where LVP_{ES} and LVV_{ES} are end-systolic left ventricular pressure and volume, is a good representation of ventricular contractility, it is difficult to perform under most clinical environment due to the need of invasive pressure and flow sensors. 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 is not completely avoided.

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This paper evaluated three indices: systemic vascular resistance, maximum VAD inflow acceleration rate ($\frac{dQ}{dt}|_{MAX}$), and the maximum VAD inflow acceleration rate during heart ejection (systolic $\frac{dQ}{dt}|_{MAX}$), to represent E_{MAX} using Novacor VAD (Novacor Division, Baxter Healthcare, CA) volume and mean arterial pressure measurements. Since VAD volume signal is always available from VAD control console and mean arterial pressure can be measured from a cuff pressure, these E_{MAX} indices can be obtained non-invasively. The relationship of these indices to E_{MAX} was evaluated using data from a computer simulation.

2. System Description

The Novacor VAD is a spring-decoupled dual pusher-plate, sac-type blood pump driven by a pulsed-solenoid energy converter. During pump filling, the pump solenoid is unlatched, resulting in a low pump pressure that allows blood from the left ventricle to flow into the pump sac. During pump ejection, the solenoid is latched rapidly, deflecting the beam springs through the pump pusher plates and exerting a balanced force on the top and bottom surfaces of the blood in the pump sac. This action ejects the blood from the pump sac into the descending thoracic aorta. An electric analog of the Novacor LVAD with a cardiovascular model is shown in Figure 1 [4].

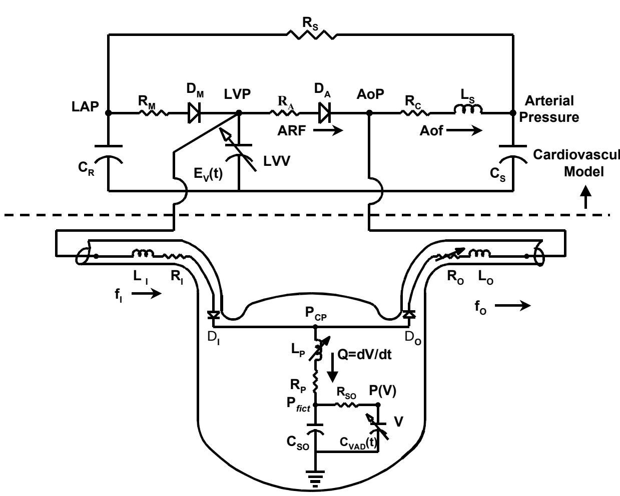

Figure 1. Electric analog of the Novacor VAD model with the cardiovascular system model

By operating the VAD in counter-pulsation mode, the pump receives blood from the left ventricle (LV) during LV systole, in which the pump pressure is significantly lower than aortic pressure. Since the patient's LV contractility is significantly lower than the normal heart, most blood flows into the pump sac and the blood flow through the aortic valve is negligible. During LV filling, the VAD ejects blood from the pump sac to the aorta. The VAD control console receives the signal of solenoid gap measurement and converts it to the corresponding pump volume.

3. Identification Methods

3.1 Systemic vascular resistance (R_S) estimation:

When the heart failure patient is completely supported by the VAD, the patient's ventricular stroke volume (SV) and cardiac output (CO) can be approximated by pump stroke volume and pump output (PO). If pump volume and mean arterial pressure (P_S) are measurable, the systemic resistance can be approximated by [4]

$$R_S \equiv \frac{MAP}{CO} \approx \frac{MAP}{(V_{MAX} - V_{min}) \cdot PR/60}, \tag{2}$$

where MAP is the mean arterial pressure, V_{MAX} and V_{MIN} are the maximum and minimum pump volume in one pumping cycle, and PR is the pump rate in beats per minute. When patient's LV function is improved, part of the blood volume from the LV passes through the aortic valve. This would cause the pump output to be lower than total cardiac output, resulting in a long-term increase in the estimated value of R_S.

3.2 $\frac{dQ}{dt}|_{MAX}$ estimation:

When the VAD is operated in a counter-pulsation mode, the pump receives blood from the LV during heart systole. The LV pressure produces a force to accelerate blood flow into the pump chamber. It can be shown by (1) that LV pressure would be higher if E_{MAX} is higher by keeping the LV volume constant. This higher pressure would cause blood flow into the pump chamber with a higher acceleration rate. The acceleration rate was obtained by calculating the second time derivative of the pump volume signal using a five point Lagrange approximation [5]

$$\dot{V}(t_k) = [V(t_{k-2}) - 8 \cdot V(t_{k-1}) + 8 \cdot V(t_{k+1}) - V(t_{k+2})] \cdot \frac{f_S}{12}, \tag{3}$$

where $V(t_k)$ is the k^{th} volume measurement and f_S is the sampling frequency. The calculated time derivative signal was then filtered by a 3^{rd} order Butterworth lowpass filter with 5 Hz cutoff frequency to reduce the noise from differentiation.

3.3 Systolic $\frac{dQ}{dt}|_{MAX}$ estimation:

When the VAD is operated in a synchronous counter-pulsation mode, the pump usually starts receiving blood from patient's left ventricle when the heart is in the isovolumic contraction period. The maximum VAD inflow acceleration rate may occur at this period causing by the fluid transition created by the pump solenoid unlatching. By definition, the maximum ventricular contraction occurs near the end of heart ejection. Detecting the maximum VAD inflow acceleration rate during heart ejection might be related to the heart contractility.

4. Test Results

The E_{MAX} indices, described in section 3, were evaluated using data from a computer simulation. The electric analog of the model including a cardiovascular model and a Novacor VAD, shown in Figure 1, was implemented in MATLAB to generate data. The value of E_{MAX}, the maximum value of $1/C_V(t)$ in Figure 1, was changed to 25%, 50%, 75%, and 100% of its nominal value to simulate the severe failure, medium failure, recovered, and the healthy heart. At each E_{MAX} setting, heart rate (HR) was changed to 60, 80, and 100 beats/min in the simulation. The value of the systemic vascular resistance (R_S) was adjusted to maintain the mean arterial pressure at the normal blood pressure range of 90 and 100 mmHg. VAD was operated under synchronized counter-pulsation by setting the pump operation thresholds: end of ejection (EOE), end of filling (EOF), and ejection delay (E_Delay) in simulation [6].

The pump volume and arterial pressure data were used to test the identification methods. The left ventricular pressure was used to determine the heart ejection period so that the maximum VAD acceleration rate during heart ejection, as described in section 3.3, can be identified. Table 1 summarizes the test conditions and the identification results.

The R_S estimates were almost constant at each heart rate regardless the changes of E_{MAX}. This implies that the R_S estimates are not sensitive to E_{MAX} change. Therefore, the R_S estimate is not an appropriate candidate as an E_{MAX} index.

The index, $\frac{dQ}{dt}|_{MAX}$, was about 2800 to 3600 ml/sec^2, which was not very sensitive to E_{MAX} as shown in Figure 2. This maximum acceleration point is usually occurred at the beginning of pump filling, as shown in Figure 3. The net force pushing blood from the left ventricle into the pump sac is the pressure difference between the left ventricular pressure (P_V) and the pump chamber pressure (P_{CP}). When the VAD started filling, P_V was still low and the VAD created strong suction ($P_{CP}<0$) due to solenoid unlatching. Therefore, the pressure difference, which forced blood into the pump chamber, was dominated by P_{CP} rather than P_V, and thus, resulting in an insignificant relation between $\frac{dQ}{dt}|_{MAX}$ and E_{MAX}.

Systolic $\frac{dQ}{dt}|_{MAX}$, defined by the maximum VAD inflow acceleration rate during heart ejection, showed a very strong relationship with E_{MAX} in Figure 2. Figure 3 shows the waveforms of $\frac{dQ}{dt}$ from simulation at heart rate of 100 beats/min with 25% (severe) and 75% (recovered) of nominal E_{MAX}, while LV systole was indicated by left ventricular pressure. With a severe failing heart, this index was about 1000 ml/sec^2 at the heart rate of 60 beats/min and was about 1300 ml/sec^2 at the heart rate of 100 beats/min. As the heart recovered to 75% of the nominal E_{MAX}, the index increased to 2500 ml/sec^2 or higher. This significant increment suggested that this index could be a useful candidate of ventricular recovery.

5. Conclusion

Three identification methods to estimate the VAD patient's left ventricular function without any invasive sensor were evaluated using data from a computer simulation of a Novacor pump with a cardiovascular model. The R_S estimates, calculated by mean arterial pressure to pump output ratio, were sensitive to the heart rate variations and insensitive to E_{MAX} changes. The index, $\frac{dQ}{dt}|_{MAX}$, obtained by calculating the 2nd time derivative of VAD volume signal during VAD filling, was almost constant in simulation regardless the E_{MAX} changes. It is not suitable to detect VAD patient's ventricular contractility using these two indices. However, $\frac{dQ}{dt}|_{MAX}$ during heart ejection was very sensitive to the changes of ventricular contractility, represented by E_{MAX}. This index was doubled as E_{MAX} increased from 25% to 75% of its nominal value.

Heart ejection interval was determined by using left ventricular pressure waveform from simulation. In real application, this ejection period can be detected by using electrocardiogram (ECG) [7]. Since the pump volume signal can be obtained through the pump control console and the ECG can be obtained without indwelling sensor, this identification method does not need any invasive sensor to estimate patient's ventricular contractility. In addition, performing this identification procedure doesn't need to interrupt VAD support. This would improve patient's safety and reduce the cost of healthcare as well. For VAD control purpose, this identification algorithm can be implemented in the VAD controller to gradually wean the patient from the dependence of the assist device.

Although the algorithm was designed using Novacor VAD model, this identification method could be applicable to most pulsatile VADs, such as Thoratec and TCI Heartmat, for which the pumps are typically operated under synchronized counter-pulsation to provide circulatory support and the pump volume signals are measurable.

Table 1. Summary of the Simulation Conditions and Test Results

Case #	E_{MAX}	HR	EOE	EOF	E_Delay	MAP	R_S	R_S Estimate	$dQ/dt\|_{MAX}$	Systolic $dQ/dt\|_{MAX}$
1	25%	60	100	60	70	90.12	1.25	1.32	3.19E+03	1.04E+03
2	50%	60	100	30	120	90.94	1.2	1.4	3.21E+03	2.00E+03
3	75%	60	100	30	120	92.91	1.2	1.37	3.17E+03	2.45E+03
4	100%	60	100	30	120	93.35	1.15	1.38	3.16E+03	2.68E+03
5	25%	80	130	30	0	94.54	1	1.08	3.19E+03	1.18E+03
6	50%	80	100	30	20	94.51	0.95	1.03	3.06E+03	2.19E+03
7	75%	80	100	30	60	94.91	0.92	1.05	3.26E+03	2.84E+03
8	100%	80	100	30	80	93.35	0.85	1.03	3.34E+03	3.33E+03
9	25%	100	150	30	0	92.5	0.9	0.96	2.80E+03	1.31E+03
10	50%	100	130	30	0	99.89	0.86	0.93	3.15E+03	2.37E+03
11	75%	100	130	30	20	98.75	0.8	0.9	3.18E+03	2.97E+03
12	100%	100	100	30	20	97.33	0.75	0.87	3.67E+03	3.67E+03

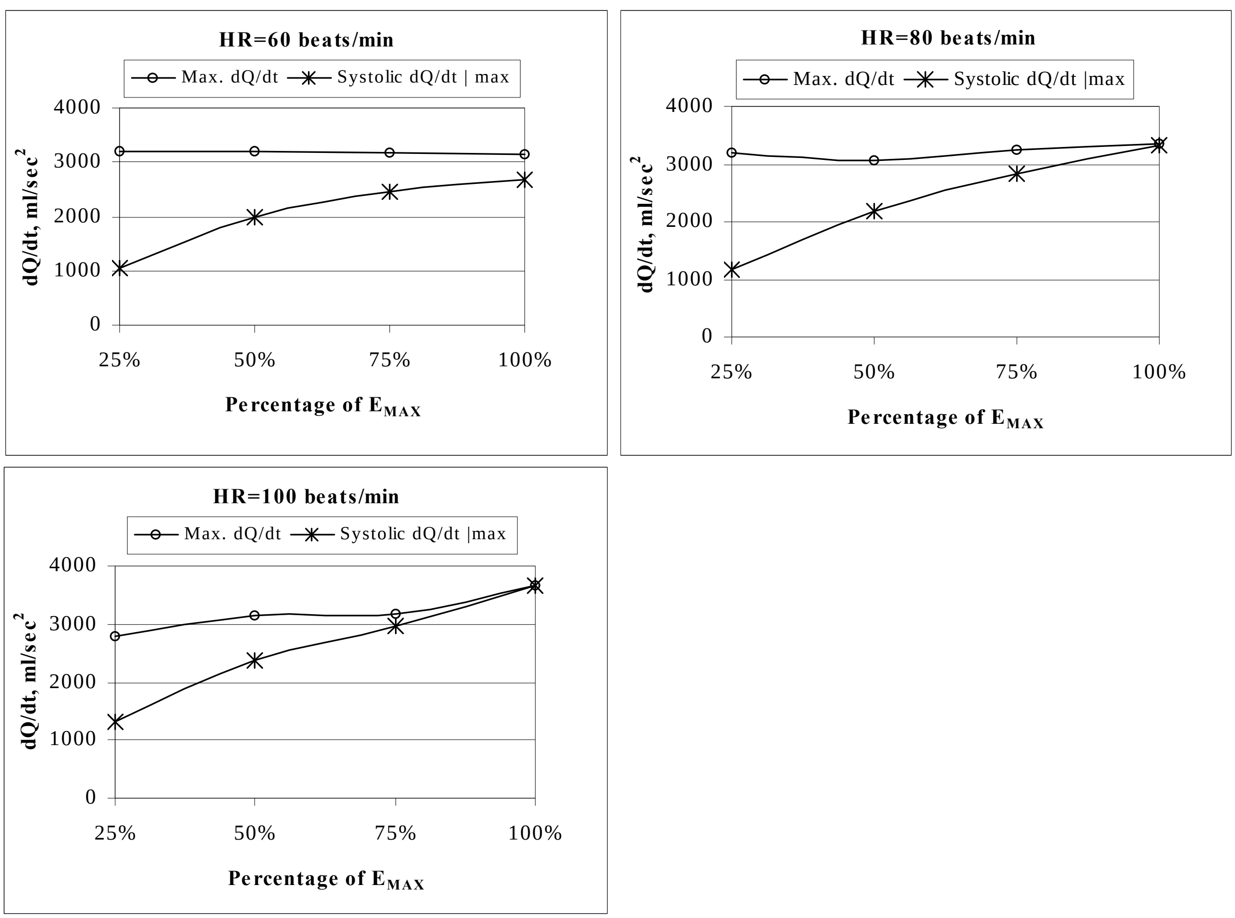

Figure 2. Maximum dQ/dt vs. E_{MAX} at different heart rate

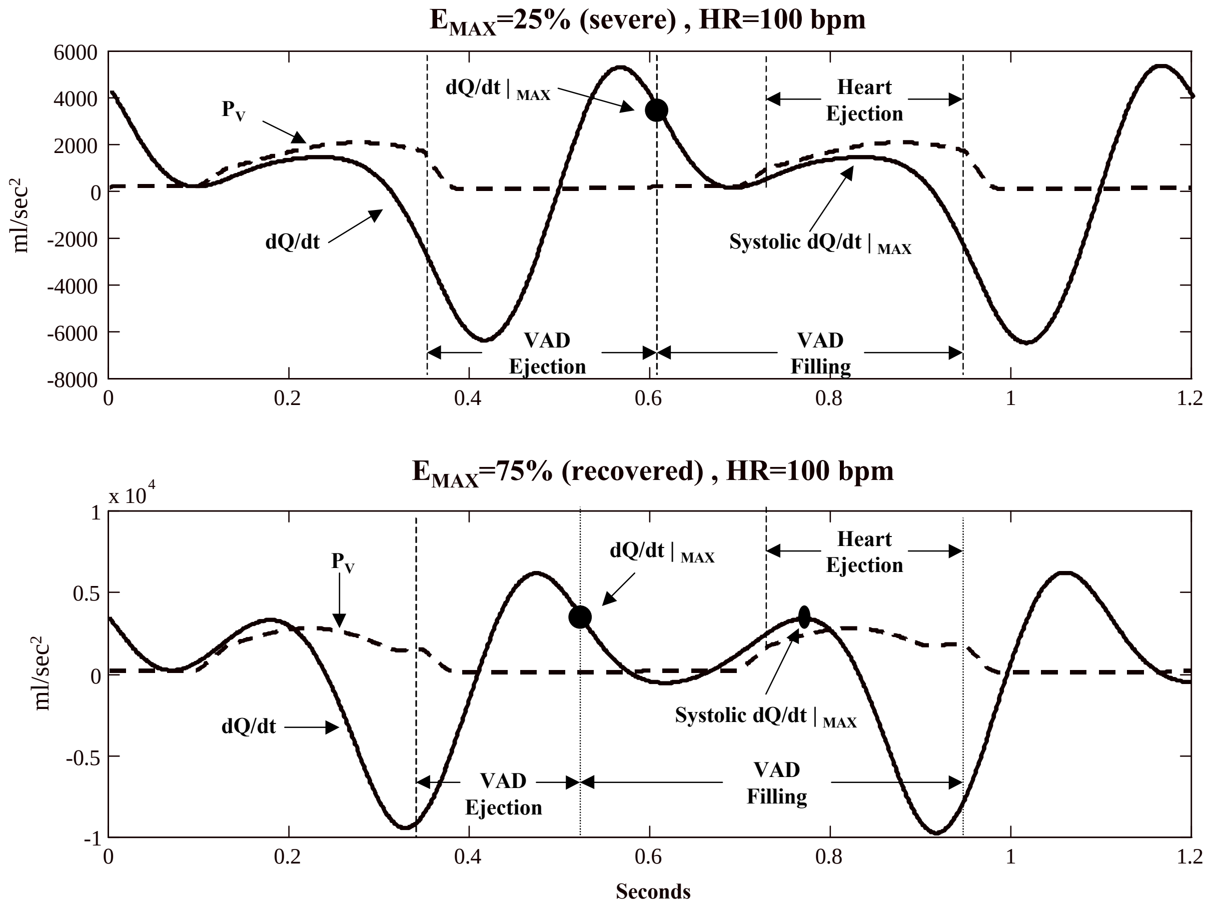

Figure 3. dQ/dt of sever and recovered E_{MAX} with heart rate of 100 beats/min

6. References

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