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A New Current-Based Control Model of the Combined Cardiovascular and Rotary Left Ventricular Assist Device

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Abstract— **Rotary Left Ventricular Assist Devices (LVAD) are mechanical pumps implanted in patients with congestive heart failure to assist their heart in pumping the required amount of blood in the circulatory system. Until recently, the combined mathematical model of the LVAD coupled with the left ventricle has assumed the availability of the rotational speed of the pump as the independent control variable. In reality, however, the device is controlled by the pump motor current which, in turn, produces the desired rotational speed of the pump motor. Therefore, the actual implementation of any desired speed controller for the device requires the solution of an inverse problem in order to determine the corresponding motor current that yields the desired pump speed. Recently, it has been observed from in-vivo experiments that an LVAD that is controlled by a motor current with a given profile (constant or ramp-like) has yielded a corresponding pump speed that exhibits a superposition of an oscillatory component which is synchronized with the pulsatility of the heart hemodynamic variables. Because of this, it has become evident that the solution of this inverse problem is extremely difficulty to accomplish. In this paper, we reformulate the existing combined LVAD and left ventricle model in such a way so as to introduce the pump motor current instead of the pump speed as the control variable, hence avoiding the inverse problem altogether. This new model is not only a more realistic representation of the LVAD control variable but also is much more practical in that it allows for the derivation of a controller directly in terms of the pump motor current rather than indirectly in terms of its rotational speed. Validation of this model and the challenges involved in using it when designing a feedback controller for the LVAD are also discussed.**

I. Introduction

THE Rotary Left Ventricular Assist Device (LVAD) is a continuous flow mechanical pump implanted as a bridge between the left ventricle of the heart and the aorta in patients with congestive heart failure who are awaiting heart transplantation. The purpose of the pump is to assist the native heart in providing the needed cardiac output (CO) and mean arterial pressure (MAP) [1][2] to sustain the life of the patient until a donor heart becomes available [3]. In recent years, it has been observed that the LVAD can also be used to help the sick heart recover and as a result avoid the transplant option. This rotary type of ventricular assist devices is much smaller, more efficient, quieter and more reliable than its predecessor the pulsatile type. But unlike the pulsatile type, the rotary LVAD continuously pumps blood in the circulatory system and hence suffers from two problems. If the pump is rotated at a speed too low to maintain the needed blood perfusion, regurgitation (backflow) might occur. If the pump is rotated at a high speed attempting to draw more blood than available in the ventricle, ventricular suction might occur. Both of these extreme conditions are undesirable and need to be avoided by insuring that the pump is always rotated at speeds that remain between these two extremes.

In order for the LVAD to operate properly, a feedback controller is needed to automatically adjust the pump speed so as to meet the body's needs for perfusion for different levels of patient activity. Since implanting sensors in the human heart comes with its own set of problems from being a liability on the battery energy to its vulnerability to thrombus formation over the sensing diaphragm [4], it is imperative that a design of a controller for the pump speed be accomplished while avoiding the use of feedback from the patient's hemodynamic variables. Several successful attempts have recently been made to design feedback controllers for the device using the pump flow signal as the feedback variable [5-7].

One of the main difficulties encountered with these approaches is that the pump speed is not directly accessible as a control variable. In reality, the device is controlled by the pump motor current and hence an inverse problem needs to be solved to determine the motor current that corresponds to a desired feedback pump speed. This inverse problem is extremely difficult to solve because of the highly nonlinear relationship that exists between the pump speed and the pump motor current and because of the systemic coupling that exists between the left ventricle and the LVAD. Most notably, it has been recently observed from in-vivo experiments that a constant pump motor current always yields a pump speed that exhibits a superposition of an oscillating component synchronized with the heart pulsatlity [8]. This coupling effect basically means that a constant pump speed cannot be simply produced from a constant pump motor current.

In this paper we report on a new model for the LVAD that uses the pump motor current instead of the pump speed as the control variable. The proposed model consists of the standard 5^{th} order model of the cardiovascular system coupled with a first order model of the LVAD. The LVAD, however, is modeled by establishing a relationship between the pressure difference across the pump and the pump motor current. The combined 6^{th} order model is now controlled by the LVAD motor current. Simulation results presented in this paper show that this model is a more accurate representation of the real system and is much more useful for exploring the

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application of modern control methods for optimally controlling the LVAD.

II. Cardiovascular Model

The heart is a very complex dynamic system that is very difficult to model mathematically. Although various complete heart models including both left and right ventricles and pulmonary circulation already exist, in this paper we are interested in a much simpler approach that involves only the left ventricle. We assume that the right ventricle and pulmonary circulation are healthy and normal and as a result their effect on the LVAD, which is connected from the left ventricle to the ascending aorta, can be neglected.

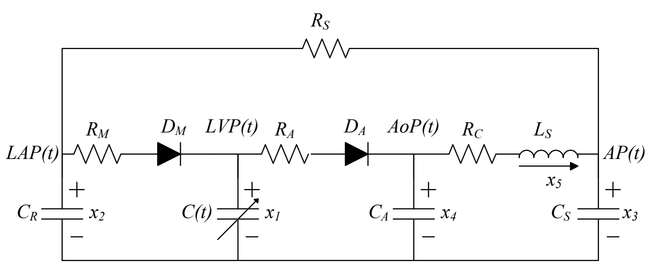

Figure 1: Cardiovascular circuit model

A 5[th] order lumped parameter circuit model of the left ventricle is shown in Figure 1. This model reproduces the left ventricle hemodynamics of the heart and can be described by the differential equations [5-7]:

$$\begin{bmatrix} \dot{x}_1 \\ \dot{x}_2 \\ \dot{x}_3 \\ \dot{x}_4 \\ \dot{x}_5 \end{bmatrix} = \begin{bmatrix} \frac{-\dot{C}(t)}{C(t)} & 0 & 0 & 0 & 0 \\ 0 & \frac{-1}{R_S C_R} & \frac{1}{R_S C_R} & 0 & 0 \\ 0 & \frac{1}{R_S C_S} & \frac{-1}{R_S C_S} & 0 & \frac{1}{C_S} \\ 0 & 0 & 0 & 0 & \frac{-1}{C_A} \\ 0 & 0 & \frac{-1}{L_S} & \frac{1}{L_S} & \frac{-R_C}{L_S} \end{bmatrix} \cdot \begin{bmatrix} x_1 \\ x_2 \\ x_3 \\ x_4 \\ x_5 \end{bmatrix} + \begin{bmatrix} \frac{1}{C(t)} & \frac{-1}{C(t)} \\ \frac{-1}{C_R} & 0 \\ 0 & 0 \\ 0 & \frac{1}{C_A} \\ 0 & 0 \end{bmatrix} \begin{bmatrix} \frac{1}{R_M} r(x_2 - x_1) \\ \frac{1}{R_A} r(x_1 - x_4) \end{bmatrix} \quad (1)$$

where $x_1(t)$ is the Left Ventricular Pressure $LVP(t)$, $x_2(t)$ is the Left Atrial Pressure $LAP(t)$, $x_3(t)$ is the Arterial Pressure $AP(t)$, $x_4(t)$ is the Aortic Pressure $AoP(t)$, all in $mmHg$, and $x_5(t)$ is the blood flow in the aorta in ml/s.

In this model, the behavior of the left ventricle is modeled by means of a time varying capacitance (or compliance) $C(t) = 1/E(t)$ where $E(t)$ is the elastance of the left ventricle. The elastance $E(t)$ describes the relationship between the ventricle's pressure and volume [9] according to an expression of the form:

$$E(t) = LVP(t) / \left(LVV(t) - V_0\right) \quad (2)$$

where $LVV(t)$ is the left ventricular volume and V_0 is a reference volume, which corresponds to the theoretical volume in the ventricle at zero pressure. Several mathematical expressions have been derived to approximate the elastance function $E(t)$. In our work, we use the expression [10]:

$$E(t) = (E_{\max} - E_{\min}) E_n(t_n) + E_{\min} \quad (3)$$

where

$$E_n(t_n) = 1.55 \cdot \left[\frac{\left(\frac{t_n}{0.7}\right)^{1.9}}{1 + \left(\frac{t_n}{0.7}\right)^{1.9}} \right] \cdot \left[\frac{1}{1 + \left(\frac{t_n}{1.17}\right)^{21.9}} \right] \quad (4)$$

and where $t_n = t / T_{\max}$, $T_{\max} = 0.2 + 0.15 t_c$ and t_c is the cardiac cycle, i.e., $t_c = 60 / HR$, where HR is the heart-rate. The constants $E_{\max}$ and $E_{\min}$ are related to the end-systolic pressure volume relationship (ESPVR) and the end-diastolic pressure volume relationship (EDPVR) respectively.

Typical values in (3) for a normal healthy heart with a heart rate of 60 beats per minute (bpm) are $E_{\max} = 2.0$ and $E_{\min} = 0.06$ ($mmHg / ml$). In the model from which (1) was derived, preload and pulmonary circulations are represented by the capacitance C_R; the aortic compliance is represented by the capacitance C_A, and afterload is represented by the four-element Windkessel circuit model [11] comprising R_C, L_S, C_S, and R_S. The left ventricle's mitral and aortic valves are represented by two non-ideal switches (or diodes) consisting of a resistance R_M and ideal diode D_M for the mitral valve, and resistance R_A and ideal diode D_A for the aortic valve. The expression $r(\xi)$ in (1) is defined by:

$$r(\xi) = \begin{cases} \xi & \text{if} \quad \xi \geq 0 \\ 0 & \text{if} \quad \xi < 0 \end{cases} \quad (5)$$

In the representation given in (1), we have kept the number of model parameters at a minimum while maintaining enough complexity in the model so that it can reproduce the hemodynamics of the left ventricle. The various model parameters appearing in (1) and their typical associated values can be found in [6] and [12].

We note that the model in expression (1) is autonomous. Its solution is oscillatory due to the cyclic nature of the terms $\dot{C}(t)$ and $1/C(t)$ in the matrices in (1). Within each cycle (called the cardiac cycle) there are three different phases of operation which occur over four different time intervals. The three phases are summarized in Table I. Clearly, every phase within the cardiac cycle is modeled by a different set of linear time-varying differential equations resulting from (1).

Table I: Phases within each cardiac cycle

Valves		**Phases**
Mitral	**Aortic**	
Closed	Closed	Isovolumic Relaxation (*I*)
Open	Closed	Filling (*F*)
Closed	Closed	Isovolumic Contraction (*I*)
Closed	Open	Ejection (*E*)
Open	Open	Not feasible

III. A Model for the LVAD controlled by its Motor Current

The LVAD considered in this paper is a rotary mechanical pump connected with two cannulae between the left ventricle and the aorta. The LVAD pumps blood continuously from the left ventricle into the aorta. The pressure difference between the left ventricle and the aorta is characterized by the following relationship:

$$\begin{aligned} LVP(t) - AoP(t) = R_i Q + L_i \frac{dQ}{dt} \\ + R_o Q + L_o \frac{dQ}{dt} \\ + R_p Q + L_p \frac{dQ}{dt} - H_p \\ + R_{su} Q \end{aligned} \qquad (6)$$

In the above expression, H_p is the pressure (head) gain across the pump and Q is the blood flow rate through the pump. The parameters R_i, R_o, and R_p represent the flow resistances and L_i, L_o, and L_p represent the flow inertances of the cannulae and pump respectively. The nonlinear time-varying resistance R_{su} has the form:

$$R_{su} = \begin{cases} 0 & if \quad LVP(t) > \bar{x}_1 \\ \alpha(LVP(t) - \bar{x}_1) & if \quad LVP(t) \le \bar{x}_1 \end{cases} \qquad (7)$$

It is included in the model to characterize the phenomenon of suction. Clearly, R_{su} is zero when the pump is operating normally and is activated when $LVP(t)$ (x_1) becomes less than a predetermined small threshold $\bar{x}_1$, a condition that represents suction. The value of R_{su} when suction occurs increases linearly as a function of the difference between $LVP(t)$ and $\bar{x}_1$. The parameter α is a cannula dependent scaling factor. The values used for the suction parameters are $\alpha = -3.5\, s/ml$ and $\bar{x}_1 = 1\, mmHg$ [5-7].

The pressure gain across the pump H_p is modeled using the direct relation between the electric power supplied to the pump motor P_e and the hydrodynamic power generated by the pump P_p scaled by the pump efficiency η as:

$$P_p = \eta P_e \qquad (8)$$

Furthermore, the electric power may be written in terms of the supplied voltage V and the supplied current $i(t)$ to the pump motor while the hydrodynamic power may be written in terms of the pump head or pressure gain H_p and the pump flow Q as:

$$\begin{aligned} P_e &= V \cdot i(t) \\ P_p &= \rho g H_p Q \end{aligned} \qquad (9)$$

where ρ is the density of the reference fluid and g is the acceleration of gravity ($\rho_{Hg} = 13{,}600\, kg/m^3, g = 9.8 m/s^2$).

Using the expressions obtained in (9) and substituting in (8) yields:

$$\rho\, g\, H_P\, Q = \eta\, V\, i(t) \qquad (10)$$

Solving for the pump pressure gain H_P,

$$H_P = \frac{\eta V}{\rho g} \cdot \frac{i(t)}{Q} \qquad (11)$$

or

$$H_p = \gamma \frac{i(t)}{Q} \qquad (12)$$

where $\gamma = \frac{\eta V}{\rho g}$. For a typical LVAD, after applying the appropriate conversion factors and assuming a pump motor supplied voltage $V = 12\, volts$ as well as a pump efficiency of 100% (assuming that most losses are accounted for by the pressure losses induced by R_p and L_p), the constant γ can be computed to be $\gamma = 89{,}944\, mmHg \cdot ml/s \cdot amp$.

Substituting (12) in (6) we obtain the nonlinear state equation governing the behavior of the LVAD as:

$$LVP(t) - AoP(t) = R^* Q + L^* \frac{dQ}{dt} - \gamma \frac{i(t)}{Q} \qquad (13)$$

where $R^* = R_i + R_o + R_p + R_{su}$ and $L^* = L_i + L_o + L_p$. Note that it is important to validate the numerical solution when expression (13) is used by ensuring that the system does not allow for operation at zero pump flow $Q(t)$ at any point during the cardiac cycle since equation (13) exhibits its nonlinearity with the pump flow $Q(t)$ in the denominator.

When combined with the model of the left ventricle (1), the LVAD state equation model in (13) will yield a model that is controlled by the pump motor current $i(t)$ as desired. Furthermore, using the relation between the pump pressure H_p and the pump speed $\omega(t)$ [12-15]:

$$H_p = \beta\omega^2(t) \tag{14}$$

and comparing with (12), an expression for the pump speed in terms of the pump motor current can be derived as follows:

$$\omega(t) = \sqrt{\frac{\gamma\, i(t)}{\beta Q(t)}} \tag{15}$$

where $\beta = 9.9025 \cdot 10^{-7}\, mmHg/(rpm)^2$. Note that it is now clear how the heart hemodynamics through $Q(t)$ influence directly, in a highly nonlinear manner, the pump speed $\omega(t)$.

IV. The Combined Model with Pump Motor Current as The Control variable

The addition of the LVAD to the left ventricle model (1) will yield the 6th order system shown in Figure 2 and described by the differential equations:

$$\begin{bmatrix}\dot{x}_1\\ \dot{x}_2\\ \dot{x}_3\\ \dot{x}_4\\ \dot{x}_5\\ \dot{x}_6\end{bmatrix} = \begin{bmatrix} \frac{-\dot{C}(t)}{C(t)} & 0 & 0 & 0 & 0 & \frac{-1}{C(t)}\\ 0 & \frac{-1}{R_S C_R} & \frac{1}{R_S C_R} & 0 & 0 & 0\\ 0 & \frac{1}{R_S C_S} & \frac{-1}{R_S C_S} & 0 & \frac{1}{C_S} & 0\\ 0 & 0 & 0 & 0 & \frac{-1}{C_A} & \frac{1}{C_A}\\ 0 & 0 & \frac{-1}{L_S} & \frac{1}{L_S} & \frac{-R_C}{L_S} & 0\\ \frac{1}{L^*} & 0 & 0 & \frac{-1}{L^*} & 0 & \frac{-R^*}{L^*} \end{bmatrix} \cdot \begin{bmatrix}x_1\\ x_2\\ x_3\\ x_4\\ x_5\\ x_6\end{bmatrix}$$
$$+ \begin{bmatrix} \frac{1}{C(t)} & \frac{-1}{C(t)}\\ \frac{-1}{C_R} & 0\\ 0 & 0\\ 0 & \frac{1}{C_A}\\ 0 & 0\\ 0 & 0 \end{bmatrix} \begin{bmatrix} \frac{1}{R_M} r(x_2 - x_1)\\ \frac{1}{R_A} r(x_1 - x_4) \end{bmatrix} + \begin{bmatrix} 0\\ 0\\ 0\\ 0\\ 0\\ \frac{\gamma}{L^* x_6} \end{bmatrix} i(t) \tag{16}$$

Notice that the additional state variable $x_6 = Q$ represents the blood flow through the pump. Eight other passive variables: $R_i, R_p, R_o, L_i, L_p, L_o$ and γ have also been added. The combined model (16) is now a forced system where the control variable is the supplied current to the pump motor $i(t)$.

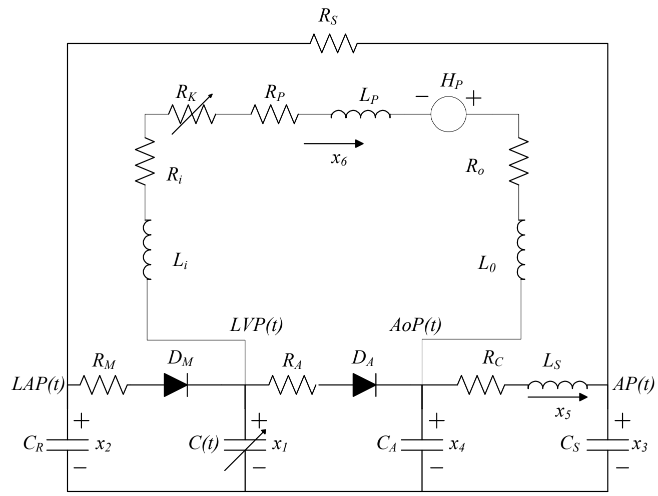

Figure 2: Combined cardiovascular and LVAD model

This model can be expressed in the standard state-space form:

$$\dot{\underset{\sim}{x}}(t) = \underset{\sim}{A}_k(t)\underset{\sim}{x}(t) + \underset{\sim}{b}\, i(t), \quad k = I, F, E \tag{17}$$

where the subscripts *I, F,* and *K* denote (see Table I) the Isovolumic Relaxation and Contraction phases (*k=I*), the Filling phase (*k=F*), and the Ejection phase (*k=E*) [5]. In addition, the vector $\underset{\sim}{b}$ multiplying the control variable $i(t)$ is a function of one of the state variables, specifically $x_6(t)$ which appears in the denominator. Therefore, special care must be taken in the numerical solution of the system in (17) as the non-linearity can significantly affect the stability of the solution algorithm. In this case, a simple time-lagging combined with a sub-level iteration scheme was devised to control the stability of the numerical solution process.

V. Challenges in the Development of a Feedback Controller Based on the Pump Motor Current

Clearly that the model derived above allows for the LVAD to be controlled by its pump motor $i(t)$ instead of its pump speed $\omega(t)$. In this model, the motor current must be adjusted to meet the patient needs for cardiac output while at the same time keeping the speed in the safe region to avoid backflow and suction.

It is important at this stage to validate the model by examining how the pump speed is affected when using the pump motor current $i(t)$ as the control variable. Figure 3 shows a plot of the pump speed when the model, with a heart rate $HR = 60\, bpm$, is exited with a linearly increasing pump current starting at $i(0) = 0.09578\, amp$ and increasing with a slope $m = 0.00934364\, amp/s$. There are several important observations that can be made from this figure. First, note that the resulting pump speed $\omega(t)$ does not also increase linearly. Instead, it increases nonlinearly with a decreasing rate of increase. Second, the pump speed has a superposed oscillatory component that has the same pulsatility as the heart rate of 60 *bpm*. This is a very interesting and extremely important new phenomenon that has recently been observed in in-vivo data obtained through clinical studies of intensive care patients implanted with LVADs [8]. This is the first time that such a phenomenon has been reproduced from a combined cardiovascular and LVAD model and represents a breakthrough in accurately modeling this complex bio-mechanical system.

A third observation that can be made from Figure 3 is that the amplitude of the oscillatory component in the pump speed signal $\omega(t)$ seems to decrease in time up to a point when a breakdown occurs and the amplitude exhibits a sudden increase when the pump current is increased beyond this point. In Figure 3, this breakdown occurs at $t = 40\,s$ which corresponds to a pump speed of $15,500\,rpm$. Clearly this value of pump speed corresponds to the onset of suction as was demonstrated in [5-7].

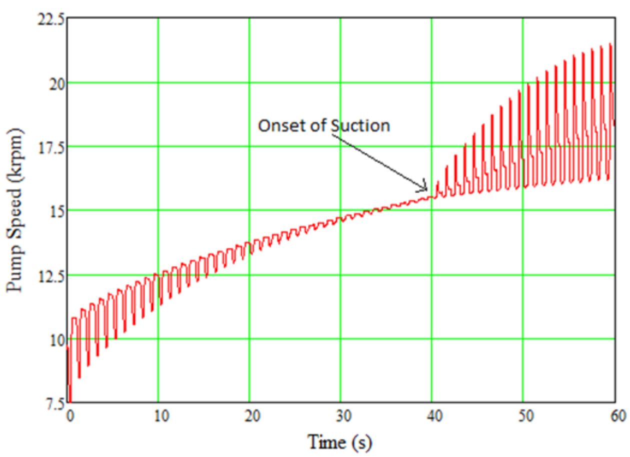

Figure 3: Pump Speed signal as a function of time derived from our model when the pump motor current is increased linearly.

If we take a closer look at the pump speed signal $\omega(t)$ at different constant values of pump motor current $i(t)$, it can be clearly seen that the waveform signature of the pump speed signal drastically changes from when the pump is operating normally to when the pump operates in suction. Figures 4 and 5 show plots of the pump speed when $i = 0.18\,amp$ and $i = 0.5\,amp$ respectively. In Figure 4 the pump is operating in the normal range while in Figure 5 the pump is in suction. This drastic change in the signature and pattern of the oscillatory component of the pump speed signal can be used in the determination of the onset of suction using pattern recognition methods. Such work is in progress and will be reported in future publications.

A fourth interesting phenomenon is observed when the model is excited with a constant pump motor current and the maximum elastance value of the left ventricle $E_{\max}$ is progressively reduced representing a heart with a worsening degrees of heart disease. This causes the availability of pressure in the left ventricle to significantly diminish hence causing the pulsatility of the pump to be substantially reduced. Figures 6 and 7 show this phenomenon when the maximum elastance value of the left ventricle is gradually reduced from $E_{\max} = 1.0$ (representing a sick heart) to $E_{\max} = 0.25$ (representing a critically sick heart) and the pump motor current is kept constant at $i = 0.18\,amp$. Figure 6 shows the left ventricular pressure revealing that the availability range goes from $8 - 45\,mmHg$ (from diastole to systole) to $8 - 13\,mmHg$ (from diastole to systole) when the heart is critically sick. Figure 7 shows the corresponding pump speed in *rpm* revealing that the pump pulsatility decreases from over 10% to less than 2.5% of the mean speed. This phenomenon, which was not observed in models that use pump speed as the control variable, now clearly indicates that it should be taken into consideration in the design of an effective feedback controller.

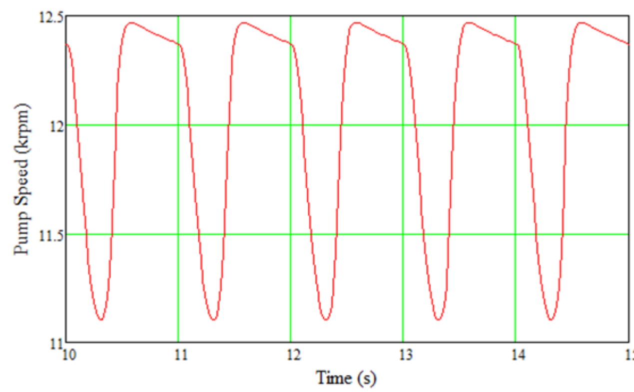

Figure 4: Pump Speed signal as a function of time when the motor current $i = 0.18\,amp$

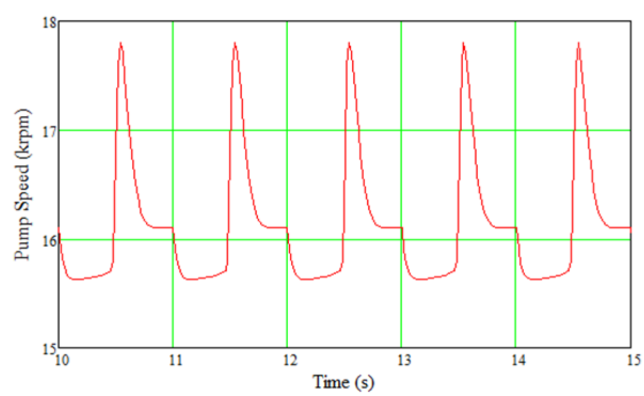

Figure 5: Pump Speed signal as a function of time when the motor current $i = 0.5\,amp$.

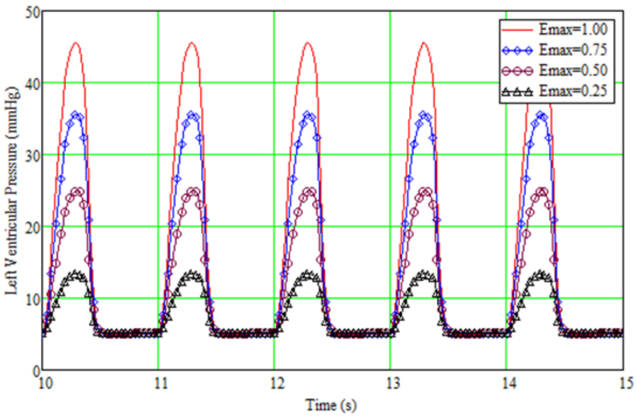

Figure 6: Left Ventricular Pressure as a function of time for different values of the maximum left ventricular elastance and motor current $i = 0.18\,amp$ **.**

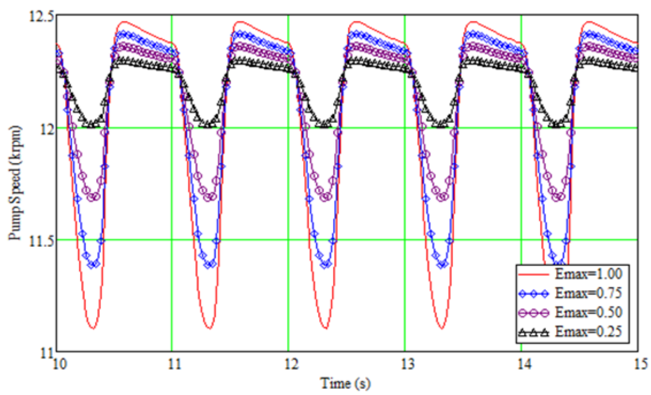

Figure 7: Pump Speed signal as a function of time for different values of the maximum left ventricular elastance and motor current $i = 0.18\,amp$

VI. CONCLUSION

In this paper, a new 6th order state-space model of a rotary Left Ventricular Assist Device connected to a cardiovascular system is presented. The control variable in this model is the pump motor current instead of the pump speed which so far has been used as the only control variable in the currently existing models. This model is much more useful for optimally controlling the LVAD since it avoids solving the inverse problem for determining the pump motor current that produces an already determined optimized pump speed. The challenges in using this model to design a feedback controller for the LVAD motor current are discussed. The characteristics of the pump speed signal, which is one of the variables that can be directly and accurately measured, when the pump is operating normally with no suction and when it is operating in suction, are also described based on results obtained from the model. Possible approaches for exploiting these characteristics in the development of LVAD control algorithms are also discussed.

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