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Feedback Control of Rotary Blood Pump for Preventing Left Ventricular Suction*

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Abstract— **Left ventricular assist devices (LVAD) have been used as a treatment option for end-stage heart failure patients, which can assist an ailing heart to pump blood into human body to meet body's circulatory demand. For long-term use an LVAD as a destination therapy, the device must be able to automatically adjust its pump speed to meet the cardiac demands at different levels of activity without inducing suction. Suction happens when an LVAD seeks to pump out more blood from the left ventricle than the available blood, which can collapse the heart and result in death. In this work, a new control system was developed, which involves two consecutive steps, i.e., the calculation of a pulsatility control index and the adjustment of pump speed to meet the blood flow requirement at different physiological conditions. The control strategy can prevent suction, while maintaining a desired cardiac output. The performance of the feedback controller has been tested with computer simulations, which demonstrates the feasibility and the efficiency of the proposed control algorithm.**

I. INTRODUCTION

Cardiovascular diseases are the main cause of death worldwide, and approximately 630,000 Americans die from heart diseases every year [1]. Heart transplantation is a recognized solution for treating severe cardiovascular diseases. However, the number of available donor hearts is very limited [2]. Left ventricular assist device (LVAD) has been recently accepted as a desired treatment option for heart failure patients, as a bridge to transplantation or as a destination therapy [3]. The purpose of the LVAD is to assist the native heart to provide the desired cardiac outputs in order to meet physiological demands and to maintain adequate perfusion of the patient's body [4].

The main challenge for using LVADs as a treatment option is to develop a robust control system. Note that most of the commercial LVADs can only be operated at a constant speed, which cannot satisfy the blood perfusion when the activity level of patients changes. The controller should appropriately adjust the pump speed to provide adequate blood flow, while avoiding the left ventricular suction [5]. Suction happens when LVADs seek to pump out more blood from the ventricle than the available blood in the ventricle. This can cause serious heart conditions such as myocardial damages, ventricular collapse, and ventricular arrhythmias that can potentially induce adverse events or sudden death [6]. The prevention of suction and the automatic adjustment of the perfusion over a wide range of physiologic conditions remain a major concern for the control of an LVAD.

Several methods were previously reported in academia for suction detection, including threshold-based techniques [7], support vector machine [8], discriminant analysis [9], and Gaussian mixture model [10]. Although these methods can successfully detect suction after its onset, they fail to prevent suction. Different control algorithms have been previously developed, which can meet the demands of physiologic flow, while maintaining the LVAD operating at a level that suction can be avoided [11, 12, 13]. However, these methods either cannot consistently provide an adequate perfusion over a wide range of physical activities or require extra information such as blood pressure to achieve a better control performance. For example, a fuzzy set logic controller was developed to satisfy the blood demands without inducing suction [14]. However, a linear relationship between the pump flow and the heart rate was used, which cannot account for the nonlinearity between pump flow and the patient variability. Using multiple sources of information of LVAD, control strategies were developed to provide desired blood flow and to detect suction, but these algorithms can be complicated and require measuring multiple parameters, e.g., ventricular pressure, that could be difficult to measure [15, 16].

Currently, commercial LVAD has been successfully used as a bridge to transplantation therapy with a constant pump speed. The long-term use of an LVAD, e.g., as a destination therapy, can greatly benefit patients with severe heart diseases. In this case, it is essential to automatically adjust the pump speed with respect to different patient activities. For example, patients may sleep, rest, and do mild exercise [17]. These activities generally have different demands on the cardiac outputs. In addition, with the support of LVAD, the ventricles of patients may get better or worse [18]. Thus, an effective and reliable control system of an LVAD should be able to adjust the pump speed adaptively to meet different blood demands for the long-term use for end-stage heart failure patients.

To address these aforementioned challenges, we developed a new control strategy using a previously developed model of LVADs by Simaan et al [11]. The control algorithm integrates a suction prevention unit and a feedback controller in order to achieve a better control performance of LVAD. Specifically, the control strategy can prevent suctions and optimally adjust the LVAD pump speed to meet the physiological demands. The rest of this paper is organized as follows. The nonlinear model of the cardiovascular system involving an LVAD is briefly discussed in Section II. Note that more information about the model can be found in [11]. The control design is

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presented in Section III, which is followed by results in Section IV and brief conclusions in Section V, respectively.

II. CARDIOVASCULAR-PUMP MODEL

Different cardiovascular models with varying complexity have been previously developed [11, 19]. Following the work in [11], we focus on the left ventricle in this work, and assume that the pulmonary circulation and right ventricle are healthy. Thus, their effect on the LVAD can be neglected. To study the hemodynamics of the left ventricle of the heart and to evaluate the performance of an LVAD, a sixth-order nonlinear time-varying lumped parameters circuit model is used in this work [11]. A schematic of the nonlinear cardiovascular-pump model can be found in [11], which can be described as [11]:

$$\dot{\boldsymbol{x}} = \boldsymbol{A}(t)\boldsymbol{x} + \boldsymbol{P}(t)p(\boldsymbol{x}) + bu(t) \tag{1}$$

$$y = [0 \quad 0 \quad 0 \quad 0 \quad 0 \quad 1]\boldsymbol{x} \tag{2}$$

where $\boldsymbol{x}$ is the states used in the cardiovascular-LVAD model, and y is the measured pump flow. There are six state variables in (1), i.e., x_1 (left ventricular pressure), x_2 (left atrial pressure), x_3 (arterial pressure), x_4 (aortic pressure), x_5 (total blood flow), and x_6 (blood flow via an LVAD). Note that the units of x_1, x_2, x_3, and x_4 are *mm Hg*, while the units for blood flow (x_5 and x_6) are *ml/s.* Details about each state can be found in [11]. In (1), $\boldsymbol{A}(t)$ and $\boldsymbol{P}(t)$ are 6×6 and 6×2 time-varying matrices, b is a 6×1 constant vector, and $p(\boldsymbol{x})$ represents the nonlinear behavior of mitral and aortic values. In addition, $u(t)=\omega^2(t)$ in (1) is the control variable, i.e., pump speed $\omega(t)$, that can be adjusted in order to meet different physiological demands [11].

Note that the model in [11] is used because it was validated with clinical data of patients. In this model, a compliance C_R is used to represent the pulmonary and preload circulations. A resistor R_M and a diode D_M are used to describe the mitral valve, while a resistor R_A and a diode D_A are used for the aortic valve. The left ventricle compliance $C(t)$ is approximated with a time-varying compliance and C_A is the aortic compliance. A four-element Windkessel model (i.e., R_c, L_s, C_s, and R_s) is used to approximate the afterload, whereas resistor R_s is the systemic vascular resistance (SVR) [11, 14]. Physiologically, R_s varies according to the level of activity, e.g., SVR decreases when patients are exercising and has a larger value when patients are resting. Note that R_s and SVR will be used in this current work alternatively. The dynamic behavior of the left ventricle can be defined by an elastance function $E(t)=1/C(t)$ as [11, 20]:

$$E(t) = (E_{max} - E_{min})E^*(t^*) + E_{min} \tag{3}$$

where $E^*(t^*)$ is the normalized elastance, and E_{max} and E_{min} are the maximum and the minimum value of $E(t)$, respectively. For a failing heart, a smaller value of E_{max} is often used in model. Such a value represents the pumping strength of a native heart is weaker than a healthy heart [11].

Resistance R_i and inductor L_i in the model were used to represent the inlet cannula, which is a plastic rigid tube that connects the left ventricle and pump [11]. The outlet cannula is described by resistance R_o and inductor L_o, which connects the pump to the aorta. A semiempirical model is used to describe the relation between the pressure difference H across the pump and pump speed as well as pump flow as [11]:

$$H = \alpha_0 x_6 + \alpha_1 \frac{dx_6}{dt} + \alpha_2 u^2 \tag{4}$$

where α_0, α_1, and α_2 are LVAD-dependent model parameters estimated from experiments [11]. Note that resistance R_k is a time varying and pressure related parameter in this model that defines the phenomenon of suction, which is described with the following equation [11]:

$$R_k = \begin{cases} 0 & \text{if } x_1(t) > x_1^T \\ \varepsilon\,(x_1(t) - x_1^T) & \text{if } x_1(t) \le x_1^T \end{cases} \tag{5}$$

where ε and x_1^T are the weight and the threshold of pressure used in the model [11].

III. FEEDBACK CONTROL DESIGN

The objective of LVAD control is to automatically adjust the pump speed to provide optimal cardiac outputs without resulting in suction in the left ventricle. Pump flow of LVAD will be used in this work for the control design, since it can be measured without invasive sensors [11, 19]. For example, the pump flow can be recorded with ultrasonic flow transducers that can be clamped on the pump cannula or can be measured with contrast echocardiography [21]. If measurements of pump flow are not available, estimations can be used [22].

A. *Pulsatility Control Index*

For end-stage heart failure patients, their native hearts can provide a partial and inconsistent contractility, which may not be sufficient to meet the physiological demands. In this case, hemodynamics such as the aortic pressure can exhibit varying degrees of pulsatiltiy. Consequently, the pump flow will also be pulsatile. As previously reported, the pulsatility in pump flow will decreases as the pump speed increases during left ventricular unload [14, 23, 24]. In addition, it was found that the pulsatility in pump flow will reach a minimum when suction occurs and increase immediately after the onset of suction (see Fig. 2, first row). Following previous works [14, 24, 11], this phenomenon will be capitalized by defining a pulsatility control index in this work, which is further used as a control inference (or control variable) to automatically adjust pump speed.

The pulsatility control index is calculated from the pump flow signal. Note that the set point of the control variable, i.e., the pulsatility control index, can be determined such that the pulsatility in the pump flow approaches the minimum where the suction occurs, but can still leave sufficient space to avoid the onsite of suction. The calculation of the pulsatility control index and the optimal selection of its set point includes two major steps, i.e., *offline calibration* and *online calculation*.

Offline Calibration

Step i: *Excitation of open-loop cardiovascular-pump system*

As explained above, the SVR (or R_s) in the cardiovascular-pump model is used to represent the continuous change of a patient's physical activity. The pump speed should be adjusted to maintain a desired cardiac output by comparing a dynamic pulsatility control index to its set point. To determine an optimal set point (i.e., *a priori*) for the feedback controller, a linearly increasing ramp pump speed, as shown in Fig. 1, is first used to excite the open-loop cardiovascular-pump system with different R_s values, representing different levels of physical activity of a patient.

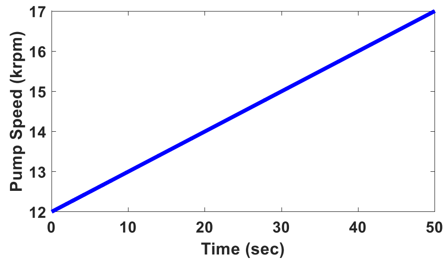

Figure 1. Pump speed profile for exciting an open-loop cardiovascular system without a feedback controller

When the pump speed increases linearly, the pump flows will change according. Fig. 2 (the first row) shows the resulting pump flows for three R_s values, where each value represents a specific level of the physiologic activity. For example, R_s=0.5 mm Hg/ml/s in Fig. 2 (a) means that the patient is doing mild exercise such as walking stairs, while R_s=2 mm Hg/ml/s means that patient is resting in bed. Note that the vertical line (red dash line) in Fig. 2 shows the time when the suction occurs. When suction is observed, the corresponding time is referred to the pump speed profile (see Fig. 1) to find the speed that can induce suction. These suction speeds are given in Fig. 2 (see first row). For example, suction occurs if the pump speed is increased to 16480 rpm, when patient is doing mild exercise (i.e., R_s=0.5 mm Hg/ml/s).

Step ii: Selection of set point of pulsatility control index

Using the pump flow signals, the next step is to extract the pulsatility control index. First, a high-pass filter will be used to eliminate baseline drift in pump flow measurements [14, 24]. The resulting signal will then be converted into absolute value by taking the magnitude and by using a low-pass filter to remove the amplitude of the pulsatiltiy. The high-pass and low-pass filters used in this work are the third-order Butterworth filters [25]. The calculation of the pulsatility control index from pump flow with the Butterworth filter was validated with sinusoidal signals over a wide range of frequency [14, 24]. For brevity, it is not shown in this work due to limited space.

Based on the open-loop cardiovascular-pump system, the pulsatility control index is studied over a full range of pump speeds. As an example, the second row in Fig. 2 shows the results for three different levels of activity. As seen, when the pump speed increases, the pulsatility control index is observed to decrease as the pump speed approaches the suction speed and then increases as the pulsatility in pump flow increase. The pulsatility control indices for different levels of activity have a large range and show a well-defined minimum. For example, all the pulsatility indices are found to be smaller than 70 ml/s with a minimum which is approximately 5 ml/s. It is important to note that only noise free simulations are shown Fig. 2 for clarity. However, it was found that the effect of measurement noise from the sensor on the calculation of pulsatiltiy index is negligible, which is not shown for brevity.

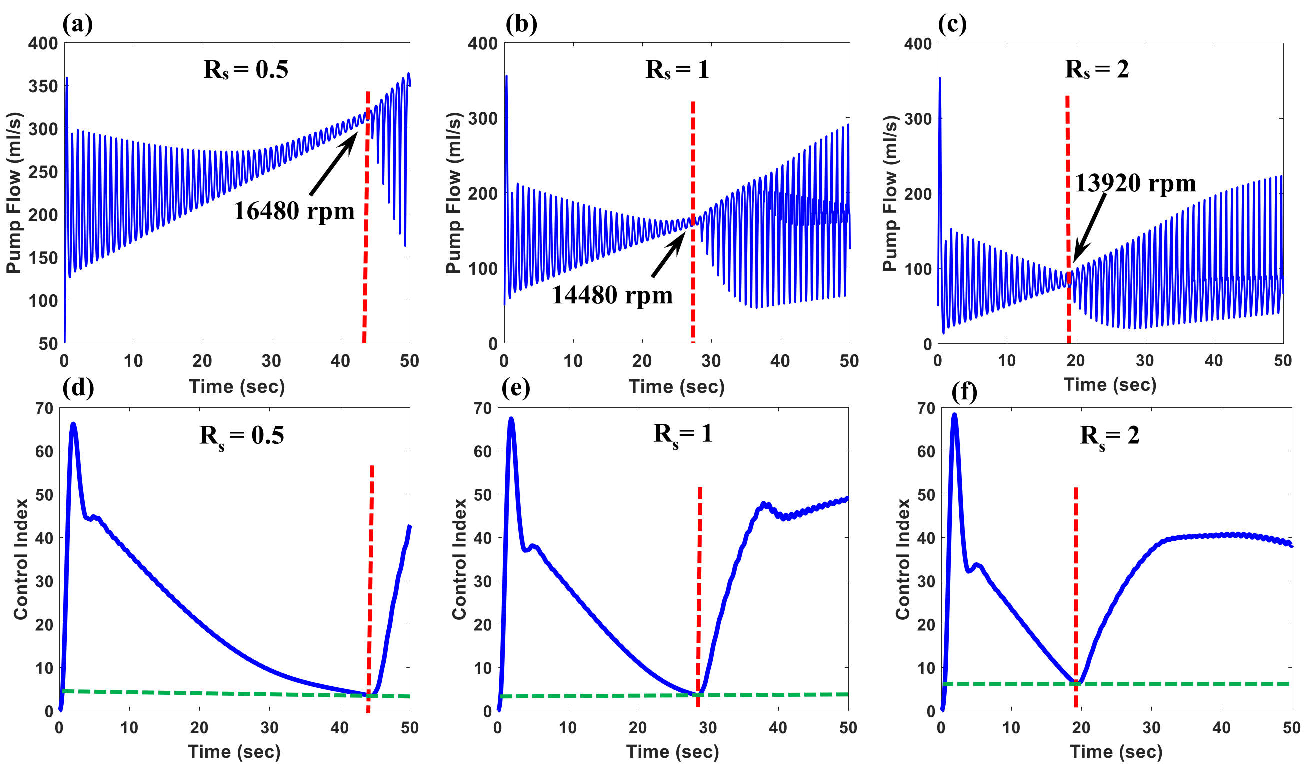

Figure 2. Calculation of control index from pump flow for different levels of activity (R_s), where the vertical line (red) in each sub-plot represents the onsite of suction and the horizontal line (green) represents the minimum value of pulsatility for each R_s when suction occurs

As discussed, the pulsatility control index of pump flow is an indicator of suction, which will be used as a control variable to adjust the pump speed and to avoid suction. As seen in Fig. 3, a value which is larger than the minimum pulsatility control index will be used as a set point for the feedback controller. The set point is a heuristic value which can prevent suction and ensure a desired cardiac output for a specific level of activity. Using the set point, it is possible to define an operating range as shown in Fig. 3. This can separate the control of pump speed into two steps, i.e., a *coarse tuning* and a *fine tuning* of pump speed, which will be discussed in the next section.

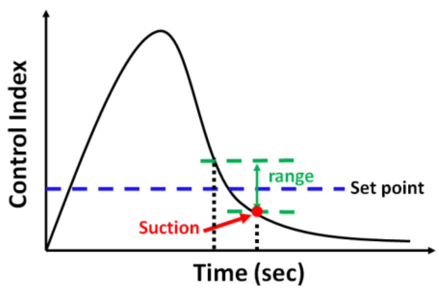

Figure 3. Selection of set point for pulsatility control index

Online application

Step iii: The offline calibration in previous steps provides an optimal set point used for the feedback control algorithm to adjust pump speed. To maintain the cardiac output at a desired level, the dynamic pulsatility control index of pump flow needs be calculated in real time and compared with the set point. In this work, a moving time window is used to calculate the pulsatility control index. A schematic of the moving window is shown in Fig. 4, where L is the size of the moving window and M is the moving rate, i.e., L determines a total number of measurements and M decides the overlap between windows. A larger window size (L) can better capture hemodynamics, while a smaller one can induce oscillation in the pump speed. The moving rate decides the number of measurements removed or appended to a window. For example, when M is 1, the first measurement in L will be removed and a new measurement will be appended to L. It is worth mentioning that a larger M may lead to a poor estimation of the pulsatility control index, whereas a smaller M means that the calculation of pulsatility control index needs to be executed frequently, thus increasing the computational load. The last value of the pulsatility control index in each moving time window, corresponding to the latest pump flow measurement, will be used and compared with the set point to adjust the pump speed with a feedback controller.

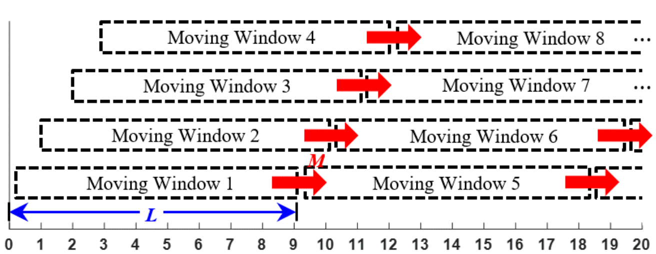

Figure 4. Schematic of moving window to calculate pulsatility control index for feedback control

B. *Design of Feedback Controller*

Based on the pulsatility control index and its set point, an integral (I) feedback controller is developed in this work, and Fig. 5 shows the general block of the feedback control strategy. The update criterion of the pump speed is defined as:

$$\omega(t+1) = \omega(t) + K_I[c_p(t) - c_{set\text{-}point}] \qquad (6)$$

where K_I is the controller parameter, $c_{set\text{-}point}$ is the set point of the pulsatility control index selected in *Step ii* in previous section, $c_p(t)$ is the dynamic pulsatility control index calculated from a moving window of pump flow at time t, and $\omega(\cdot)$ is the pump speed. The controller parameter K_I is patient specific and has two different values, which divides the adjustment of pump speed into two consecutive steps, a *coarse tuning* of pump speed and a *fine tuning* of pump speed. Using the set point of control variable, a range can be defined as seen in Fig 3. When the pulsatility control index is found to be outside the range, a relative larger controller gain will be used to quickly adjust the pump speed to meet the physiologic demands. However, the *coarse tuning* step must be followed by a *fine tuning* procedure. When the pulsatility control index is found to be within the range, which means that the pump is operated in the vicinity of suction, a smaller controller gain will be used. This will result in a more cautious control law to avoid aggressive adjustment in pump speed, which may potentially induce suction.

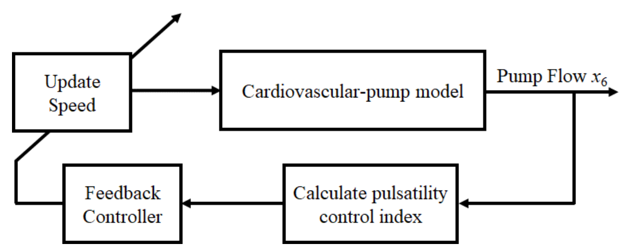

Figure 5. Schematic of the gain-scheduling feedback control of LVADs

It is important to note that the size of moving window for estimating the pulsatility control index and different values of K_p for pump speed control would be chosen by a clinician and can be adjusted depending on the physiological conditions of the patient and the quality of pump flow measurements. The size of the moving window is set to 5 cardiac cycles in this work, and the controller parameters are proportional to the slew rate of the pump speed.

IV. RESULTS

The objective of the feedback controller is to adjust pump speed to meet the physiological demands at different levels of activity, while remaining the pump speed of an LVAD below a level at which suction happens. To evaluate the efficiency, two case scenarios were investigated in this work, which are described as follows.

A. *Constant Systemic Vascular Resistance*

In the first case study, a constant R_s (SVR) value is used where the activity level of the patient remains unchanged for a long period of time. In this study, R_s is set to 1 mm Hg/ml/s for which the suction can occur when the pump speed $\omega(t)$ reaches approximately 1.48×10^4 rpm. Fig. 6 shows the pump speed, the corresponding pump flow, and the pulsatility control index. As seen in Fig. 6, in order to meet the physiological demands, the controller can increase and maintain the pump speed without inducing a suction. Due to the limited space, only the noise-free simulations are given in Fig. 6. The heart rate in this case study is 75 bpm, and the contractility strength of the native heart (E_{max}) is set to 2 mmHg/ml. The set point of the control variable, i.e., pulsatility control index, is set to 15 ml/s.

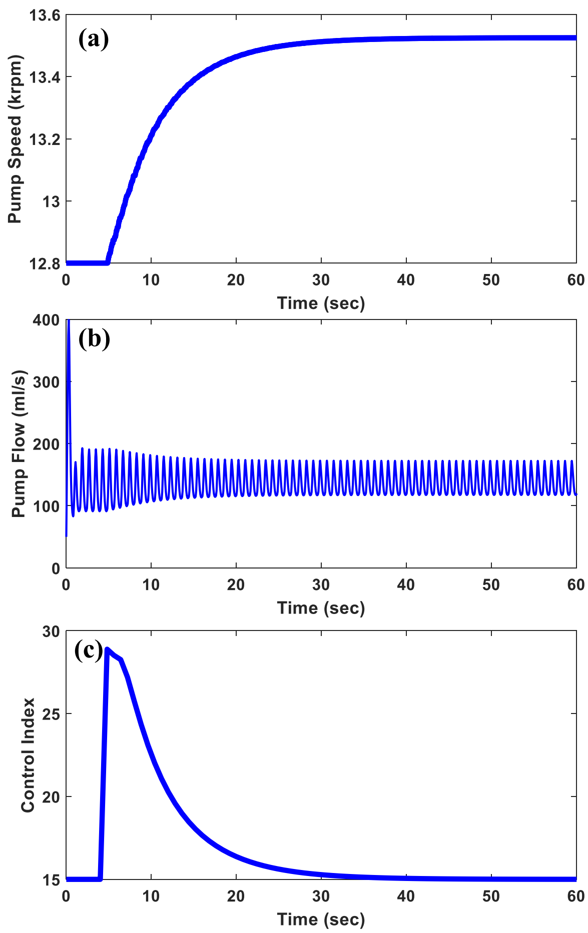

Figure 6. Simulated hemodynamic waveforms and the results of the feedback control for a constant R_s (a) pump speed, (b) pump flow, and (c) pulsatility control index

As seen in Fig. 6, the pump speed and the pulsatility control index remain constant before 5 s, since the calculation of the dynamic pulsatility has not been executed and the control system is collecting the measurements of pump flow for the first moving window. At 5 s, the closed loop system is initiated and the pulsatility control index can be calculated. As seen, the control index converges to a value (~28 ml/s), which is much higher than the set point. Thus, the controller begins to increase the pump speed in order to reduce the pulsatility control index. After approximately 27 s, the set point is reached and the pump speed settles to about 1.35×10^4 rpm.

B. Time-varying Systemic Vascular Resistance

In the second case study, R_s is set to different values, which simulates the changes of patient's activity level over time. The dynamic changes of R_s is shown in Fig. 7 (a).

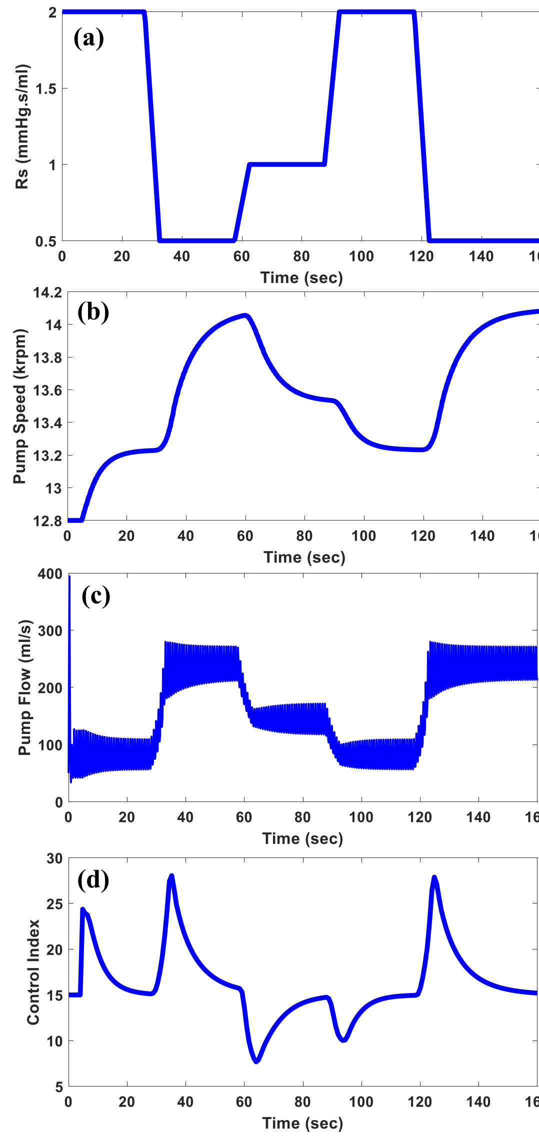

Figure 7. Simulation resuts of a time-varying R_s which represents a sequence of changes in the level of activity. (a) time-vaying activities (b) controlled pump speed, (c) pump flow, and (d) the pulsatility control index

As seen in Fig. 7 (a), R_s was initially kept at 2 mm Hg/ml/s, which was decreased to 0.5 mm Hg/ml/s and remained for a short period of time. After that, R_s was then gradually increased to a medium value, i.e., 1 mm Hg/ml/s, from which it was further increased to 2 mm Hg/ml/s, followed by a decrease to 0.5 mm Hg/ml/s again. A larger value of R_s indicates a lower intensity of physical activity, and vice versa. For example, a smaller value of R_s (0.5 mm Hg/ml/s) can be used to represent that the patient did mild exercise such as walking, while a larger value of R_s means that the patient was resting. The results of a feedback controller are shown in Figs. 7 (b) to (c).

As seen in Fig. 7 (b), the pump speed can be appropriately adjusted according to these dynamic changes in different levels of activity. Such an adjustment can ensure the cardiac outputs, while preventing suction. Fig. 7 (c) shows the time-varying pump flow, which was used to calculate the pulsatility control index (see Fig. 7 (d)). The set point of the pulsatility control index was set to 15 ml/s, and the range used to ensure a safe operation of the pump was set to [5 ml/s, 25 ml/s]. These results imply that the controller can successfully adjust pump speed according to patient's physiological activities without suction.

V. CONCLUSION

The control system for the left ventricular assist device must be safe and adaptable. Safety means that the system should be able to prevent suction and protect the heart from ischemia, while adaptability means that the control system should be able to automatically adjust the pump speed to meet the body's demands at different levels of physical activity. In this current work, a pulsatility control index-based feedback controller is developed. This system involves two consecutive steps, i.e., the calculation of the pulsatility control index and the autonomous adjustment of the pump speed to meet the desired cardiac output. The performance of the controller is assessed with computer simulations for different scenarios. The proposed controller can maintain the cardiac outputs within an acceptable range and prevent suction, which demonstrates its feasibility for animal experiments.

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