A Sensorless Robust Tracking Control of an Implantable Rotary Blood Pump for Heart Failure Patients

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Abstract—Physiological control of a left ventricle assist device (LVAD) is generally a complicated task due to diverse operating environments and patient variability. In this work, a tracking control algorithm based on sliding mode and feed forward control for a class of discrete-time single input single output (SISO) nonlinear uncertain systems is presented. The controller was developed to track the reference trajectory to a set operating point without inducing suction in the ventricle. The controller regulates the estimated mean pulsatile flow $\overline{Q_p}$ and mean pulsatility index of pump rotational speed $\overline{PI_\omega}$ that was generated from a model of the assist device. We recall the principle of the sliding mode control theory then we combine the feed-forward control design with the sliding mode control technique to follow the reference trajectory. The uncertainty is replaced by its upper and lower boundary. The controller was tested in a computer simulation covering two scenarios (preload and ventricular contractility). The simulation results prove the effectiveness and the robustness of the proposed controller.

Keywords—robust control system, discrete-sliding mode, left ventricularle assist devicse, pulsatility index.

I. INTRODUCTION

Heart failure (HF) is the final stage of heart disease and a major cause of mortality worldwide. This disease may arise for a number of reasons, including factors internal to the heart which include valvular and coronary heart disease or external factors which place excessive demands on the heart muscle such as; hypertension, and excessive volume load [1]. In Australia, cardiovascular disease is the leading cause of death. In 2007, over a third of deaths ($46,623$ Australians) were caused by cardiovascular disease (CVD). 50% of these deaths were due to coronary heart disease, and 18% to stroke. Over 78% of the CVD deaths were of people aged 75 years and over, and more than half were female 52.7%. (AIHW 2010) [2].

Today, different therapies for HF have been developed, which include surgery, pharmacological treatment and mechanical cardiac assist devices. Since left ventricle failure accounts for a large percentage of heart failures, an implantable rotary blood pump (IRBP) type of left ventricular assist device (LVAD) emerges as a viable long-term treatment option for HF patients. The purpose of the LVAD is to provide an adequate cardiac output at a sufficient pressure to allow the tissues of the body to auto-regulate their flow. Sensorless control of an IRBP is one of the most important design gaols in providing long-term alternative treatment for HF. The implantation of an additional sensor is not desirable as it can result in thrombus formation, a reduction in system reliability, an increase in cost and the need for regular calibration to correct measurement drifts [3].

One of the main goals required to improve the clinical application of IRBP technology includes the development of a control strategy, which automatically adjusts the pump speed to cater for cardiovascular system perturbations and the changing metabolic demand. In a healthy individual, the frank-starling mechanism ensures that the stroke volume of left ventricle (LV) is adjusted to compensate for changes in LV end-diastolic pressure such that the LV ejects whatever volume of blood it receives from the right ventricle [1]. Salamonsen et el [4] found that the responses of an IRBP when maintained at a fixed speed to changes in preload and after load are very different from the natural heart, in so far that they have insufficient preload sensitivity to inherently sense the amount of blood with which they are supplied and are affected by variations in left ventricular afterload. Therefore, it is imperative that a pump control strategy maintains a safe operating range where pump outflow matches right heart output.

A number of control systems which have been designed to provide physiological control of IRBP output regulate the control variable to a given value or "set point" which often becomes inappropriate when circulatory circumstances change [5]. The goal of the controller should be to achieve a balance between the venous return from the pulmonary circulation and pump output. To attain this outcome, various physiological control strategies have been proposed. These can be classified according to the controlled variable as: pump differential pressure control [6] [7], afterload control [8] [9], flow control [10] [3], preload control [11], control of pulsatility ratio [12] and control of pulsatility gradient [13].

In this paper, we aim to design a robust controller for

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solving trajectory-tracking problems of set points. We use a sensorless linear time variant (LTV) dynamical model of pulsatile flow to describe their behavior with bounded disturbances in system dynamics [14]. Recently our research group has developed and validated an approach that used pulsatility measures derived for IRBPs to be used as surrogates for LV stroke work [15]. This enables the development of a Starling-like controller to apply physiological control. The proposed strategy of the SMC is a special discontinuous control technique applicable to various practical systems [16]. The main advantages of using a SMC include fast, good transient response and robustness with respect to system uncertainties and external disturbances. Therefore, it is attractive for many highly nonlinear uncertain systems [17].

This paper is organized as follows. In Section II, the dynamical model of sensorless pulsatile flow is presented. An analytical representation of the system, some basis of the quasi-sliding mode and the developing of control strategy are described in Section III. A computer simulation results using the proposed technique are presented in Section IV. In section V a discussion has taken place. Future work is illustrated in VI. Finally, section VII concludes the paper. The advantages of the presented algorithm are investigated by numerical simulation using Matlab-Simulink (The Math-Works Inc., Natick, MA, USA).

II. THE DYNAMIC MODEL

An empirical model of the LVAD has been described previously [14]. This model was developed to estimate the mean pulsatile flow $\overline{Qp}$ and the mean pulsatility index of pump rotational speed $\overline{PI_\omega}$ using mean pulse-width modulation $\overline{PWM}$ as control input. In this model different cardiovascular states are simulated by using different values of hemodynamic parameters including; total blood volume V_{total}, systemic vascular resistance SVR and parameters representing the left ventricular contractility El_v. Since LVADs are implemented inside patients whose hearts are abnormal, the diversity of such parameters may represent different degrees of heart failure severity. Pump rotational speed was varied from 1500 to 2950 rpm in stepwise increments of 100 rpm with each step lasting 30 seconds. The dynamic equations of the model in the state space can be represented as:

$$\begin{aligned} x(k+1) &= Ax(k) + \delta Ax(k) + Bu(k) + \omega(k) \\ y(k) &= Cx(k) \end{aligned} \quad (1)$$

where, $A(k) = \begin{bmatrix} -a & 0 \\ d & -c \end{bmatrix}$, $B(k) = \begin{bmatrix} b \\ 0 \end{bmatrix}$, $C(k) = \begin{bmatrix} 0 & 1 \end{bmatrix}$ here, $x(k) = \begin{bmatrix} x_1(k) & x_2(k) \end{bmatrix}^T$, $x_1(k)$ is the $\overline{PI\omega}$, $x_2(k)$ is the $\overline{Qp}$, $u(k)$ is the $\overline{PWM}$ which is the real pump control input, δA is system parameter variations, $\omega(k)$ is the system noise, $y(k)$ is the system output, a, b, c and d are the system parameters estimated using system identification algorithm.

III. CONTROLLER DESIGN

The control problem in this paper is to force the state vector $x(k)$ to track a specific vector $r(k)$ in presence of uncertainties and disturbance, where $r(k)$ is a known vector. The design objective is to construct a state feedback control law $u(k)$ based on a sliding mode control strategy in combination with feed-forward controller to guarantee tracking of operating points in a stable manner.

1) Feedback controller design : This design was achieved based on discrete-time sliding mode, which performs measurements and control signal applications at regular intervals of time and keeps the control signal constant between intervals. Discrete-time sliding mode offers invariance to uncertain parameters, compensating for uncertainties that exist in real dynamic applications, thus making it a good choice for the error trajectory tracking problem. Designing of discrete sliding mode control is known to consist of the following steps:

- reaching law: In this step, the choice of the reaching law that guarantees strong reachability is illustrated. In [18] Gao et el proposed quasi-sliding mode (QSM) approach that describes the reaching law conditions to guarantee the ideal sliding motion. The equivalent discrete-time reaching law that satisfies our model and design is:

$$s(k+1) = (1 - qT)s(k) - \epsilon T sign(s(k)) \quad (2)$$

where , $T > 0$ is the sampling period, $\epsilon > 0$ $q \geq 0$ such that $0 < (1 - qT) < 1$.

- control law: In this step the control law is synthesized from the reaching law in conjunction with a known model of the plant and known perturbations. A commonly used switching function in discrete time can be defined as:

$$s(k) = c^T x(k) = c_1 x_1(k) + c_2 x_2(k) \quad (3)$$

where c^T is a constant vector, stated with $[1 * 2]$ matrix chosen to ensure that $x(k)$ is asymptotically stable on $S(k) = 0$ [19]. To satisfy the reaching law in eq (3), we obtain:

$$s(x + 1) = c^T x(k + 1) \quad (4)$$

from eq (1) and eq (4) we have:

$$s(k+1) = c^T(Ax(k) + \delta Ax(k) + Bu(k) + \omega(k)) \quad (5)$$

the equlization of eq (2) and eq (5) gives:

$$\begin{aligned} c^T Ax(k) + c^T Bu(k) &+ c^T \delta Ax(k) + c^T \omega \\ &= (1 - qT)s(k) - \epsilon T sign(s(k)) \end{aligned} \quad (6)$$

solving the above equation for the command signal $u(k)$ yields:

$$\begin{aligned} u(k) = &-(c^T B)^{-1}(c^T A - (1 - qT)c^T x(k)) \\ &-(c^T B)^{-1}\epsilon T sign(s(k)) \\ &-(c^T B)^{-1}[c^T \delta Ax(k) + c^T \omega(k)]) \end{aligned} \quad (7)$$

where c^T is chosen such that $c^T B$ is non-singular to simplify notation, let: $\alpha = (c^T B)^{-1}(c^T A - (1 - qT)c^T)$ and $\beta = (c^T B)^{-1}\epsilon T$.

It is reasonable assume that the upper and lower bounds of $[c^T \delta Ax(k) + c^T \omega(k)]$ are known as:

$$-kT < [c^T \delta Ax(k) + c^T \omega(k)] < kT \quad (8)$$

the control law in eq(7) can be expressed in the following compact form:

$$u(k) = -\alpha x(k) - \beta sign(s(k)) - kT sign(s(k)) \quad (9)$$

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$$u(k) = -\alpha x(k) - (\beta + kT)sign(s(k)) \qquad (10)$$

It is pointed out by Bartoszewicz [20] that the control law in (10) only guarantee a discrete sliding mode control if:

$$kT < \frac{qT\epsilon T}{2(1 - qT)} \qquad (11)$$

the sliding mode trajectory will move within the QSM band as:

$$|s(k)| < \frac{\epsilon T}{(1 - qT)} \qquad (12)$$

Now the closed loop system can be found by substituting (10) into (1), which results in:

$$\begin{aligned} x(k+1) = \quad & Ax(k) + B(-\alpha x(k) \\ & -(\beta + kT)sign(s(k))) \\ & +\delta Ax(k) + \omega(k) \end{aligned} \qquad (13)$$

$$\begin{aligned} x(k+1) = \quad & (A - B\alpha)x(k) - B(\beta + kT)sign(s(k)) \\ & +\delta Ax(k) + \omega(k) \end{aligned}$$
$$(14)$$

to be more convenience with the discussion below we define:

$$M = A - B\alpha \qquad (15)$$

The closed loop system (14) is an autonomous system, whose dynamics are determined by its system matrix M, in spite of a non-linear part. For the stability of the system, all eigenvalues of M must lie inside the unit circle and these eigenvalues are calculated by the choice of the switching function c^T in (3) and the time factor in the reaching law (2).

2) Feed-forward controller design : This design aims at zero tracking error and it is based on the inverse of a closed loop transfer function of the system [21]. The feed-forward path is then used as a pre-filter for the reference input signal to compensate for the phase lag of the feedback system and consequently achieve a good tracking [22]. To achieve this aim the reference signal has to be known in advance for at least a few sampling intervals, meaning the inverse of the model is non-causal. The design method can be described as follow: convert the system model in (1) into controllable canonical form in state space, yielding the state space model matrices A, B and C as:

$$A(k) = \begin{bmatrix} -a_1 & -a_2 \\ 1 & 0 \end{bmatrix}, B(k) = \begin{bmatrix} 1 \\ 0 \end{bmatrix}, C(k) = \begin{bmatrix} 0 & c_1 \end{bmatrix}$$

the closed loop system matrix in (15) equals:

$$M = A - B\alpha = A - B(c^T B)^{-1}(c^T A - (1 - qT)c^T) \quad (16)$$

In QSM band the discontinuous part of the plant states will change its sign in each successive step, so that the average control action during QSM band is a simple state feedback controller, with an average closed loop transfer function. From (16) the average of closed loop transfer function is:

$$D(z) = C(zI - M)^{-1}B \qquad (17)$$

IV. SIMULATION RESULTS

Computer simulations were run to demonstrate that the controller responded appropriately to cardiovascular changes in accordance with the control law and feed forward controller that have been established. The simulation is based on a dynamic model defined by equation (1) using the proposed method. The design parameters of the switching function in equation (3) are $c_1 = 1$, $c_2 = -3.5$ and those of the control law in equation (7) are $qT = 0.025$, $\epsilon T = 0.001$ and $kT = 0.5$. The controller was evaluated using data obtained from a lumped parameter-optimized model of the cardiovascular system over wide range of operating conditions, including variations in preload (V_{total}) and left ventricular contractility (LVC) [23]. Fig. 1(a) to Fig. 1(f) show the desired trajectory with interrupted red line and with a continuous blue line the resulted trajectory during the response of the proposed controller to variation in preload across wide range of three scenarios (law, medium and high. Fig. 2(a) to Fig. 2(f) show the desired trajectory with interrupted red line and with a continuous blue line the resulted trajectory during the response of the proposed controller to variation in left ventricle contractility across wide range of three scenarios (law, medium and high). The results showed that the designed controller was able to track the changes in states of the system x_1 and x_2 during different physiological conditions with a stable and short transient response.

V. DISCUSSION

This paper proposes, for the first time, a new control scheme that combines a sliding mode dynamic controller and a feed-forward control strategy under uncertain and external disturbance for an IRBP to track desired trajectory. According to the simulation results, the proposed algorithm demonstrates a robust tracking performance in comparison with traditional controller. In this context, different controls methods have proposed to track set points by automatically regulate the pump speed to ensure enough perfusion of the patients. Most recently, Gawk el el (2010), [24] successfully validated a new algorithm using an extremum seeking control strategy to drive an LVAD. Also Karantonis et elet al. (2010), [36], suggested a new control algorithm for a centrifugal IRBP based on a sensorless indicator of the implant recipients activity level. In a similar fashion, Wu et elet al. (2007), [9], used a state-space model of the circulatory system together with measurements of the pump differential pressure and developed a control algorithm to regulate and control the aortic pressure. In addition, Giridharan and Skliar (2006), [25], have proposed a dynamic model to estimate the differential pressure across the pump and then used it as the input to a control system for an implantable of rotary blood pumpIRBP. Others, Bullister et elet al. (2002), [7], designed a feedback control system based on hierarchical approaches to control the speed of the pump from pressure sensors measuring pump inlet and outlet. Also, Zhou et al. (1999), [26], used a non-linear static model of the pump to simulate the hemodynamic responses of the assisted circulatory system as a function of different speeds of the pump. Waters et elet al. (1999), [27] designed a proportional-
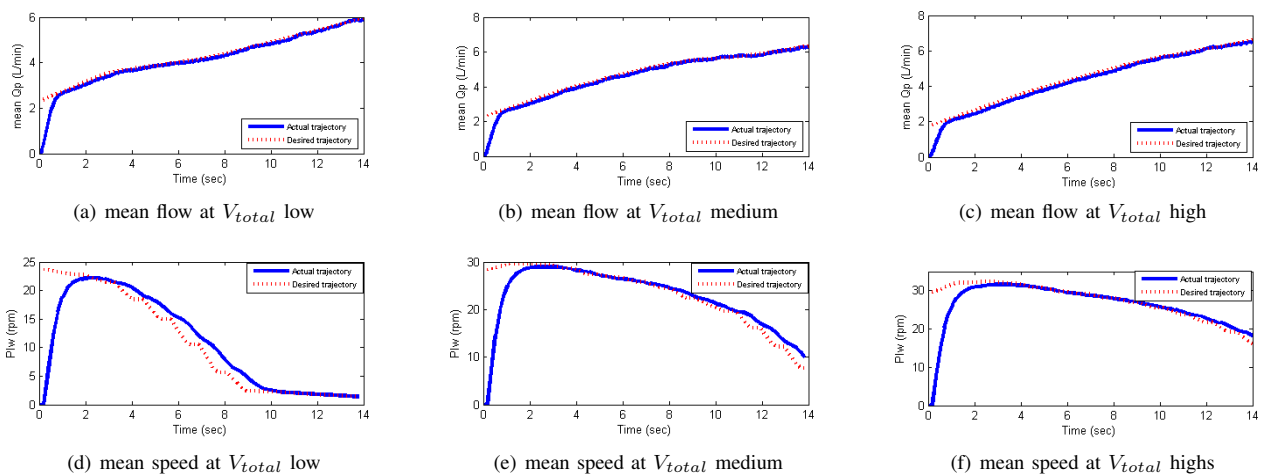
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Fig. 1.   Preload scenario (a) mean flow at V_{total} low , (b) mean flow at V_{total} medium, (c) mean flow at V_{total} high, (d) mean speed at V_{total} low, (e) mean speed at V_{total} medium, (f) mean speed at V_{total} high.
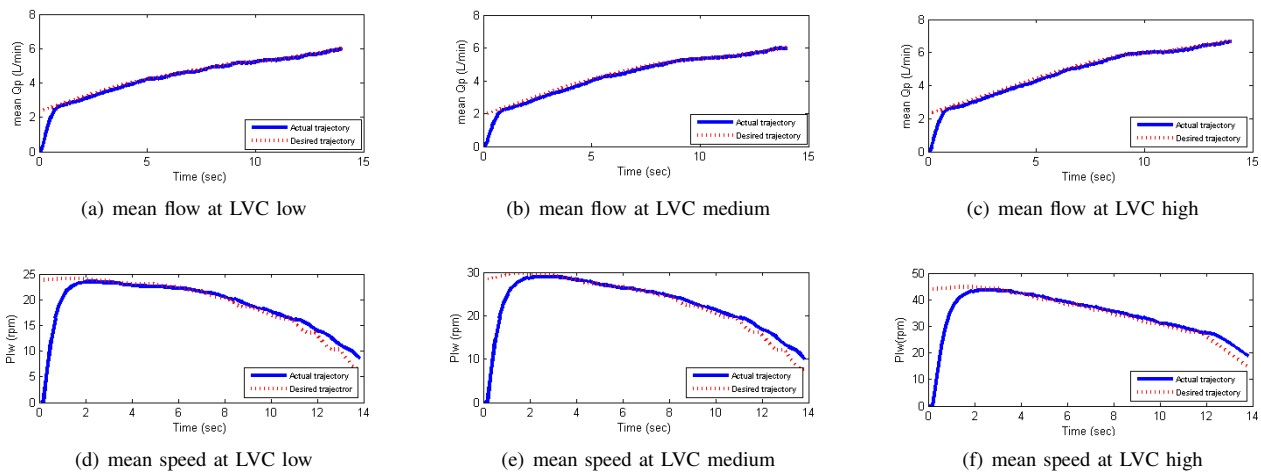

Fig. 2.   LVC scenario (a) mean flow at LVC low , (b) mean flow at LVC medium, (c) mean flow at LVC high, (d) mean speed at LVC low, (e) mean speed at LVC medium, (f) mean speed at LVC high.

integral algorithm to adjust the motor speed and maintain the system reference differential pressure. There are two main limitations of previously designed control algorithms. The first is the need to estimate differential pressure and flow using steady-state models without data relating to the transient response of the pump. The second is that other systems measure pressure and/or flow and thus require the implantation of additional sensors to provide measurements of parameters used as inputs to their control algorithm.

VI. FUTURE WORK

The first priority is to complete validation studies of model and controller responses to changes in physiological parameters associated with varying levels of simulated HF using cardiovascular system model. Also, future work will cover the evaluation of model and controller using a mock circulatory loop $in-vitro$ and in $in-vivo$ animal experiments.

VII. CONCLUSION

In this paper, we examined the performance of a feedforward/feedback controller in the presence of model uncertainty, with varying degrees of heart failure. The new control strategy for pulsatile flow of an LVAD and its potential advantages and requirements were evaluated using a numerical simulation. Results showed that the controller was able to track the reference input with minimal error, by using sensorless measurements. The controller also smooth and prompt transient response and good disturbance rejection ability. These merits represent a tool for a controlled ventricular training that may potentially, and hopefully, contribute to an increase in the number of patients showing ventricular recovery.

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