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Application of multiobjective neural predictive control to biventricular assistance using dual rotary blood pumps

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ABSTRACT

Rotary blood pumps are used to provide mechanical circulatory support to the failing heart in patients who are ineligible or waiting for a transplant. One of the major challenges when implementing two rotary blood pumps for biventricular support is the difficulty in maintaining pulmonary and systemic circulatory volume balance. In this study, a novel multiobjective neural predictive controller (MONPC) hybridized with a preload-based Frank-Starling-like controller (PFS) has been proposed for a dual rotary blood pump biventricular assist device in two different configurations: PFS_L-$MONPC_R$ and $MONPC_L$-PFS_R. The flow rate of one pump is regulated by PFS as a function of preload, while the other pump is controlled by MONPC, which is intended to meet cardiac demand, avoid pulmonary congestion and ventricular suction. A comparative assessment was performed between the proposed controllers and a Dual Independent Frank-Starling-like control system (DI-FS) as well as a constant speed controller. The numerical simulation results showed that $MONPC_L$-PFS_R helped unload the congested left ventricle while maintaining high cardiac output during exercise. In contrast, improper flow regulation by DI-FS has resulted in pulmonary congestion. During blood loss, PFS_L-$MONPC_R$ delivered the lowest suction risk, as compared to the constant speed mode, which produced negative right ventricular preload. When sensor noise and time delays were introduced in the flow and end-diastolic pressure signals, the proposed controllers were able to respond with adequate robustness during the transition from rest to exercise. This study demonstrated that the proposed controllers are superior in matching the pump flow with the cardiac demand without causing hemodynamic instabilities.

1. Introduction

Development of right ventricular failure occurs in approximately 10–40% of patients supported by a left ventricular assist device (LVAD) [1–3]. If these LVAD recipients become unresponsive to pharmacologic treatment for right ventricular dysfunction, a secondary procedure is required to implant a right ventricular assist device (RVAD) [4]. However, implementation of dual implantable rotary blood pumps for biventricular support may lead to difficulty in balancing the pulmonary and systemic circulatory blood volume [5]. This issue is attributed to the low preload and high afterload sensitivity of the continuous flow devices in comparison to the native heart [6].

An active physiological LVAD controller has been proposed by Salamonsen et al. [7] to reproduce the Frank-Starling mechanism of the natural heart. Similar control scheme has been extended to a dual LVAD biventricular assist device (BiVAD) using a Dual Independent Frank-Starling-like controller (DI-FS) [8]. Nevertheless, as reported by Stevens et al. [9], DI-FS exhibited pulmonary congestion when a transition from rest to exercise took place. This drawback is attributed to the independent control systems that adjust the left and right pump speeds irrespective of the preload in the opposite ventricle. In order to address this issue, Stevens et al.

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[9] proposed a Master/Slave controller, with the master controller acting as a Frank-Starling-like controller, which sets flow rate as a function of preload, while the slave controller maintains a linear correlation between the right and left ventricular end diastolic pressure. This control scheme has been proven to produce fewer suction events and did not cause pulmonary congestion [9,10]. However, its stability is very much dependent on the sensitivity of the linearly correlated ventricular end diastolic pressures. In addition, it remains an open issue if a constant ratio between these two variables appropriately reflects the actual physiological scenario of a native heart.

One of the key issues of the control design is to satisfy multiple clinical requirements, which includes perfusion demand, avoidance of pulmonary congestion and minimization of suction risk [11]. Multi-objective control is an elegant way to adjust pump flow to strike a balance between these sometimes contrasting requirements [11,12]. However, limited investigations associated with this control technique for a BiVAD have been performed, despite the possible benefits.

The development of a multiobjective control system often requires the incorporation of a high-fidelity predictive model that can well represent the underlying nonlinear dynamics of the BiVAD assisted cardiovascular system (CVS). Research on a linear model based controller for a uni-ventricular support system has been reported in the literature [13,14]. However, developing a linear model with satisfactory predictions of multiple system outputs is a challenging task, as the real CVS is highly nonlinear and there exists nonlinear flow characteristic, valve systems and complex interconnection among the various intrinsic subsystems that mutually affect one another.

Artificial neural networks (ANN) have been well known for their capability as a universal approximator to fit a nonlinear system without prior physical knowledge [15]. To date, ANN model has been reported to deliver satisfactory prediction performance in various biomedical applications [16]. Incorporation of an ANN as the predictive model in a controller helps to optimize the control signal at each time instant by delivering plant output prediction as well as process knowledge to the controller.

In this study, the framework of the aforementioned DI-FS has been extended through the incorporation of a multiobjective neural network based predictive controller (MONPC). The main motivation of the control design is to minimize the risk of suction and pulmonary congestion, while maintaining a Frank-Starling mechanism of the VADs in response to different physiological states of the patients (rest, exercise and blood loss). In the following section, the numerical model of a BiVAD-assisted CVS adopted for the evaluation of the control schemes is first introduced. Then, the working principle of the proposed ANN-based control schemes is presented. Subsequently, comparative assessment between the proposed controller and two other control counterparts, namely the DI-FS and the constant speed mode, is performed. In addition, the robustness of the proposed control schemes is further evaluated in the presence of sensor noise and delays. Finally, in depth discussion with regard to the simulation results, limitations of the study as well as future works is presented.

2. Methods

2.1. Model description

As shown in Fig. 1, a numerical model of the human CVS developed in previous studies [17,18] has been adopted to investigate the implementation of the proposed controller. In brief, this model was derived based on first principles, which describes the dynamics of the left and right hearts, as well as the systemic and pulmonary circulations. Additionally, two models of a rotary blood pump with inlet and outlet cannulae attached to the ventricles and arterial segments respectively were integrated with the CVS model in a BiVAD configuration.

Table 1
Published clinical data and simulated hemodynamics for heart failure subjects. P_{sa}: systemic arterial pressure; P_{la}: left arterial pressure; P_{ra}: right arterial pressure; P_{pa}: pulmonary arterial pressure.

Variable	Heart failure (NYHA III)	
	Clinical range	Model
P_{sa} (*mmHg*)	69–78 [24–26]	73.8
P_{la} (*mmHg*)	19–29.5 [25,27–29]	21.1
P_{ra} (*mmHg*)	12–20 [25,27–29]	17.5
P_{pa} (*mmHg*)	30–37.9 [25–29]	31.6
Cardiac output(*L/min*)	2.6–4.5 [25,29]	3.0
Pulmonary vascular resistance ($dyn\ s\ cm^{-5}$)	80–380 [22,29]	280
Systemic vascular resistance ($dyn\ s\ cm^{-5}$)	600–3500 [30]	1500

The model for both LVAD and RVAD was identified based on the VentrAssist™ centrifugal pump [19–21]. In addition to the cannula resistance, a banding resistance, was included at the outflow cannula of the RVAD to represent the banded graft downstream of the RVAD. A banding diameter of 6 mm was chosen, and the flow dependent banding resistance was modelled based on the experimental data provided in [22].

It is noteworthy to mention that the VentrAssist™ pump was designed with hydrodynamic levitation, and thus a minimum pump speed (≈1500 rpm) is required to prevent impeller touch-down. As distinguished from VentrAssist™, a magnetically levitated pump can achieve a much lower minimum pump speed through active control of the rotor position [23]. In this study, the speed range of the pump model has been extrapolated to 900–3300 rpm, allowing the controller to be tested under extreme conditions such as blood loss and intensive exercise that requires a much wider operating speed range as compared to that achieved by a rotary pump with hydraulic levitated impeller.

The CVS-BiVAD interaction model has been validated previously using data collected from in vivo animal experiments, in vitro mock-loop as well as other published data. Detailed description of this model can be found in [18,20,21].

Initially, the hemodynamics of a biventricular heart failure scenario (New York Heart Association (NYHA) class III) at rest was reproduced by optimizing the model parameters associated with heart failure. As depicted in Table 1, it can be observed that the resultant simulated hemodynamic variables for the biventricular heart failure scenario were in agreement with the published clinical data.

Three different patient scenarios were simulated to assess the control scheme performance: rest, exercise and blood loss. In the rest scenario with BiVAD support, it was assumed that the systemic and pulmonary vascular resistance (SVR and PVR) were restored back to the normal range of a healthy human. The other two test scenarios were realized through variation of the peripheral unstressed volume V_0, SVR, PVR, heart rate and blood volume. Quantitative settings for each test scenario simulated on the MCL are given in Table 2. The values of these parameters have been determined based on previous studies by our research group that investigated the barorelfex mechanism using the same CVS model [31,32]. Transitions from the resting condition to exercise and blood loss were realized by changing the values of the respective variables following a ramp trajectory within a time frame of 10 s.

In order to evaluate the impact of unmodelled and unfiltered sensor noise on the ANN based control system, uniformly distributed white noise signal has been added to the blood flow and end diastolic pressures signals, with amplitude intervals of [−0.3 0.3] and [−0.1 0.1], respectively. The amplitudes of the white noise