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# Minimizing left ventricular stroke work with iterative learning flow profile control of rotary blood pumps

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ABSTRACT

Rotary blood pumps are gaining importance in the successful treatment of advanced heart failure. However, the application of fixed pump speeds is discussed controversially. Since the natural heart delivers pulsatile flow, many physicians presume that pulsatile pumping provides therapeutical advantages. To address this, we combine the technical advantages of continuous flow devices with the supposed physiological advantages of pulsatile flow. We present an iterative learning control (ILC) strategy for continuous flow ventricular assist devices that minimizes the left ventricular stroke work (LVSW). For that, a comprehensive nonlinear model for rotary blood pumps that is used for simulation and controller design is introduced. The controller is tested using a hardware-in-the-loop cardiovascular system simulator with a Medos deltastream DP1 blood pump. The tracking performance of the proposed ILC approach is compared to a benchmark controller that uses additional sensor information, both controllers significantly reduce the residual LVSW compared to the fixed speed case. In addition to decreasing ventricular load, the proposed ILC strategy can be used as an inner control loop to any physiological controller that sets reference flow profiles. The introduced controller might be useful for the investigation of effects of various pulsatile flow patterns independent from the type of VAD in future in vivo studies. The targeted manipulation of physiological quantities such as the residual cardiac work has the potential to considerably improve ventricular assist device therapy.



## 1. Introduction

Patients with advanced heart failure are unable to perform any physical activity without discomfort [1]. Approximately 1–2% of the adult population in developed countries suffers from heart failure [2]. In case of an unfavorable progression of the disease, heart transplantation remains the gold standard therapy. Due to the limited availability of donor organs and the rapid development of ventricular assist device (VAD) technology, the number of patients bridged to transplant or receiving destination therapy by implantation of a VAD increased significantly [3]. This trend is confirmed by the latest Interagency Registry for Mechanically Assisted Circulatory Support (INTERMACS) report, which includes clinical information from 15,745 adult patients that received primary prospective implants between June 2006 and December 2014. The vast majority of these patients (84.4%) were treated with a left ventricular assist device (LVAD) [4].

The purpose of an LVAD is to relieve the native heart and to increase the cardiac output by pumping blood from the left ventricle into the aorta. In modern LVAD therapy, displacement pumps are virtually replaced by rotary blood pumps, as the latter are associated with dramatically lower overall adverse event rates [4]. These pumps are normally operated at a constant rotational speed; this leads to a reduced vascular pulsatility compared to physiological flow [5]. There is evidence that fixed speed support using continuous flow devices can cause aortic valve insufficiency [6,7], gastrointestinal bleeding [8], and thromboembolic events [9]. It is not yet known whether these complications are induced by the pump type or the lack of pulsatile flow. A systematic overview of clinical studies comparing pulsatile and continuous flow support and their conflicted findings is given in [10]. These studies cannot separate pump design from flow effects as they are limited to the comparison of previous pulsatile technology to recent continuous

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flow devices. This issue may be addressed by pulsatile flow control of rotary blood pumps.

A control strategy that restores pulsatility might be the solution for combining the benefits of smaller and more reliable rotary blood pumps with the positive hemodynamic effects of pulsatile devices [11]. A recent review of studies investigating cardiac cycle-synchronized speed modulation of continuous flow devices [12] confirmed that such strategies can increase pulsatility in the systemic arterial circulation and enhance the unloading of the left ventricle. These studies focused on the modulation of pump speed as this is the control variable in standard VAD therapy and hence clinically available. However, the quantity that directly affects hemodynamics is the pump flow, which depends not only on the pump speed, but also on the differential pressure across the pump. Hence, it is preferable to directly control the pump flow [12]. It enhances the comparability of studies conducted with different VADs and heterogeneous groups of subjects. Ising et al. [13] demonstrated in silico that the arterial pulsatility and ventricular work can significantly be affected by the modulation of the pump flow. The specific control of VADs might eventually lead to higher myocardial recovery rates [14].

In this paper, we present an iteratively learning flow controller for rotary blood pumps. It enables good reference tracking of numerically optimal flow profiles that minimize the left ventricular stroke work (LVSW). Moreover, we show that pump flow modulation outperforms the fixed speed control operation case in terms of LVSW reduction, which is consistent with the literature. Therefore, we present a comprehensive nonlinear model for continuous-flow VADs. It is used to design and test the iterative learning control (ILC) strategy. ILC was chosen because it exploits the repetitive nature of the changing differential pressure across the pump to improve the flow tracking performance. Besides, it was already successfully applied in clinical trials in other areas of biomedical engineering [15,16]. The improvement of the flow tracking performance is achieved by generating a control input trajectory that incorporates error information of previous heartbeats. The LVSW-optimal reference trajectories, which are solutions of a nonlinear program, allow a meaningful comparison of different control strategies. We tested two strategies: ILC combined with a PID feedback controller and a classical PID controller with PD disturbance feedforward control. Their performance was assessed in a hybrid mock circulatory loop with a cardiovascular system (CVS) model simulating severe heart failure.

This paper is structured as follows. Section 2 presents our approach and the test setup. In Section 3, in vitro results are analyzed and discussed. Our conclusions are drawn in the final section.

## 2. Materials and methods

This section begins by introducing a generic nonlinear model for rotary blood pumps. It is used to design and test the iteratively learning pump flow controller proposed in Section 2.2. The tracking performance of the ILC is tested with numerical optimal flow profiles described in Section 2.3 in a hybrid mock circulatory loop which is presented in Section 2.4. The last section gives a brief overview of the experimental procedure.

### 2.1. Rotary blood pump model

The functional principle of a continuous-flow VAD is identical to that of conventional centrifugal pumps. As VADs are in direct contact with blood, their designs are optimized for hemocompatibility. Continuous flow devices include an impeller in the flow path that accelerates the blood stream. Different pump types can be distinguished by impeller design and position of the flow outlet. The rotation axis and outlet are aligned in axial-flow pumps, whereas they are tangential in radial-flow pumps. Fig. 1 shows the rotary blood pump used in this study (deltastream DP1, Medos AG, Stolberg, Germany). It can be classified as a diagonal pump, because the blades are mounted diagonally on the impeller. The VAD is designed for extracorporeal operation, in which tubes connect the inlet of the pump to the apex of the left ventricle and the outlet of the pump to the ascending aorta.


**Fig. 1.** Medos deltastream DP1 rotary blood pump with a diagonally streamed impeller.

We developed a nonlinear dynamical model of the blood pump in order to design and test controllers in silico. For this purpose, the standard electro-hydraulic modeling approach was applied [17]. The overall model of the system is depicted in Fig. 2.

The electrical subsystem consisting of an electronically commutated motor is modeled by applying Kirchhoff's voltage law to the armature

$$L_{\mathrm{arm}} \frac{\mathrm{d}i_{\mathrm{arm}}}{\mathrm{d}t} = v_{\mathrm{arm}} - K_{\mathrm{e}}\omega - R_{\mathrm{arm}} i_{\mathrm{arm}} \tag{1}$$

with $L_{\mathrm{arm}}$ armature inductance [mH], $K_{\mathrm{e}}$ back-EMF constant [$\mathrm{V\,s\,rad^{-1}}$], $R_{\mathrm{arm}}$ armature resistance [m$\Omega$], $i_{\mathrm{arm}}$ armature current [A], $v_{\mathrm{arm}}$ armature voltage [V], $\omega$ angular velocity of the rotor [$\mathrm{rad\,s^{-1}}$].

According to Newton's third law, the torque generated by the electrical drive is transmitted over a motor shaft that is directly connected to the impeller. The resulting rotary motion of the mechanical subsystem can be described by

$$J_{\mathrm{rot}} \frac{\mathrm{d}\omega}{\mathrm{d}t} = K_{\mathrm{t}} i_{\mathrm{arm}} - K_{\mathrm{f}}\omega - \tau(\omega, q_{\mathrm{pump}}) \tag{2}$$

with $J_{\mathrm{rot}}$ moment of inertia of the rotor and impeller [$\mathrm{kg\,m^2}$], $K_{\mathrm{t}}$ motor torque constant [$\mathrm{N\,m\,rad\,A^{-1}}$], $K_{\mathrm{f}}$ viscous friction constant [N m s], $q_{\mathrm{pump}}$ pump flow [$\mathrm{L\,min^{-1}}$], $\tau$ load torque as a function of $\omega$ and $q_{\mathrm{pump}}$ [N m rad].

Besides inertia and load torque, Eq. (2) only contains viscous friction as a complete stop of the impeller is prevented in normal operation in order to reduce the risk of blood clotting in the pump.

Eqs. (1) and (2) model a conventional brushless DC electric motor which is very similar in all pump designs. In contrast, the hydraulic subsystem is different for every individual pump design. We determined a two-dimensional characteristic map of the pump in static differential pressure and flow measurements. From this measurement data, we derived a model for the differential pressure

$$\Delta p_{\mathrm{pump}} = A\omega^2 - R_{\mathrm{turb}} q_{\mathrm{pump}}^2 \operatorname{sign}(q_{\mathrm{pump}}) - R_{\mathrm{lam}} q_{\mathrm{pump}} \tag{3}$$

with $A$ pump constant [$\mathrm{mmHg\,s^2\,rad^{-2}}$], $R_{\mathrm{turb}}$ turbulent flow resistance [$\mathrm{mmHg\,min^2\,L^{-2}}$], $R_{\mathrm{lam}}$ laminar flow resistance [$\mathrm{mmHg\,min\,L^{-1}}$], $\Delta p_{\mathrm{pump}}$ differential pressure generated by the pump [mmHg].

Note that [mmHg] is a manometric unit of pressure that is still widely used in the medical community. The pump constant $A$ describes the relationship between rotational speed of the impeller and generated differential pressure $\Delta p_{\mathrm{pump}}$. Two hydraulic resistances in series model losses in the flow path due to laminar and turbulent flow. Although the rotational speed $\omega$ is strictly positive, the direction of flow can reverse at high negative differential