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Emulation of Ventricular Suction in a Hybrid Mock Circulation

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Abstract— **Ventricular suction occurs when a turbodynamic ventricular assist device (VAD) tries to pump more blood from the ventricle than is available. Suction causes a stagnation of the blood flow and can damage the heart muscle and must therefore be avoided in clinical practice. A hybrid mock circulation is a test bench for VADs, which is based on a hardware-in-the-loop concept. It consists of a numerical model of the human blood circulation and a numerical-hydraulic interface, which allows interaction between the numerical model and a VAD to be tested. This paper shows how ventricular suction is implemented in the hybrid mock circulation to allow testing of suction detection algorithms and physiological VAD controllers. Experimental results show that suction can be emulated as desired.**

I. Introduction

Mechanical circulatory support (MCS) denotes the use of mechanical pumps to support the blood circulation of patients suffering from heart failure. In the last decade, MCS has gained much significance for the treatment of heart failure. This increase yields from a shortage in donor hearts and from improved outcomes of MCS. The most important class of MCS devices are turbodynamic ventricular assist devices (VADs) [1]. Turbodynamic VADs, also known as rotary VADs or continuous-flow VADs, are pumps with a spinning rotor exerting a pressure on the blood. VADs are implanted between the left ventricle (LV) or atrium and the aorta, or between the right ventricle or atrium and the pulmonary artery, i.e., they pump in parallel to the failing heart. Nowadays, clinically used VADs are operated at a constant pump speed, which is chosen by the physician or the patient. This mode of operation lacks an adaptation of the pumping function to the demand of the patient and therefore reduces the quality of life of VAD recipients. To overcome this problem, control algorithms for a physiological adaptation of VADs have been developed [2], [3].

The task of a physiological VAD controller is the adaptation of the pump speed to meet the patients blood perfusion demand in a safe manner. The requirements which need to be met by the controller are manifold, but the hard constraints, which limit the allowed pump speed, are easily explained in the following. When the pump speed is chosen too low, the blood flows backwards through the pump, which potentially increases blood damage (hemolysis, platelet activation) and leads to a congestion in the pulmonary veins, potentially causing lung edema. When the pump speed is chosen too high, the ventricle is emptied and the ventricular wall gets sucked onto the inflow cannula of the VAD. This event is called suction and is dangerous because the blood flow is disturbed and the heart muscle can be injured. It is therefore necessary for any VAD controller that suction is avoided.

VAD controllers are tested in different types of models, namely numerical models, mock circulation models, and animal models. These models have their advantages and disadvantages and are all used during the development of VAD controllers. In order to properly test VAD controllers, these models must be able to emulate suction. Animal models inherently offer this feature. For classic mock circulations and numerical models, approaches to emulate ventricular suction have been presented in the literature [4], [5].

In a previous publication [6], we have presented a hybrid mock circulation, which integrates a numerical model and a classic mock circulation, combining the advantages of both models. So far, this test bench has not been able to emulate suction. In this paper we present a method to implement ventricular suction in a hybrid mock circulation. Experimental results are provided, which show that the reference pressures during suction cannot be tracked accurately, but the overall behavior of suction can be emulated sufficiently well.

II. Materials & Methods

A. Hybrid mock circulation

1) Overview: The hybrid mock circulation developed in our group consists of a numerical model of the human blood circulation running in real-time and a numerical-hydraulic interface to allow an interaction between this numerical model and a real VAD. Whereas the numerical circulation model was developed based on existing work [7], the interface was developed specifically for the purpose of the hybrid mock circulation. Fig. 1 shows a picture of this interface. The fluid circuit of the interface consists of two pressure-controlled reservoirs (a), (b), the blood pump (c), and the backflow pump (d). The blood pump is the device to be tested. In this paper, a non-implantable, mixed-flow blood pump equipped with an encoder and an industrial motor-control unit is used. The blood-pump controller is described in more detail in Section II-A.2. The backflow pump (d) is installed in the circuit to keep the fluid-level in the two reservoirs (a), (b) constant despite a flow through the blood pump. The input to the backflow pump controller is the fluid-level in the reservoirs measured by infrared range finders (e). The pressure in the reservoirs is controlled to track the reference signals from the numerical circulation model. The pressure in each reservoir is controlled independently using pressurized air. The pressure control system consists of one proportional

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solenoid inlet valve and two proportional solenoid outlet valves (f) per reservoir, a vacuum pump (g) and a vacuum receiver (h), one air-pressure sensor (i) per reservoir, and one fluid-pressure sensor per reservoir (not visible in picture). The pressure control algorithm is described in more detail in Section II-A.3.

Depending on the pump speed and the pressure difference across the pump, a flow rate results, which is measured using clamp-on ultrasonic flow probes (k). The measured flow rate signal from one probe is fed back to the numerical circulation model, where it is used as an outflow from the LV and an inflow to the aorta. This setup allows a real-time interaction between the simulated blood circulation and the real blood pump. The overall delay from a reference pressure signal to a measured flow signal was identified to be approximately 10 ms [6]. The test bench is equipped with a PC running MATLAB Real-Time Windows Target, which executes five tasks: simulation of the numerical circulation model, data logging, blood-pump control, backflow pump control, and pressure control.

The numerical circulation model from [7] consists of a systemic and a pulmonary circulation, each with arterial and a venous part. The arterial trees are modelled by five-element Windkessel models; the venous parts are modelled by three-element Windkessel models [8]. The pressure in both arterial trees is regulated by a baroreflex, which adapts the respective arterial resistance. The cardiac output (CO) is controlled by an autoregulation mechanism, which adapts the unstressed volume of the systemic veins. The heart consists of two atria and two ventricles, all of which are contracting actively triggered by a time-varying elastance model. Fig. 2(a) shows measured pressure-volume (PV) loops for the circulation model in different conditions obtained at the hybrid mock circulation. The time-varying elastance model is implemented by a time-varying function $F_{\text{iso}}(t) \in [0, 1]$ (see Fig. 2(b)), which indicates the contractile state of the muscle fibres and which is used to interpolate between the dashed lines in Fig. 2(a). The upper dashed line is not touched by the PV loop due to the modelled visco-elasticity of the LV during systole.

2) Blood pump controller: The blood pump is equipped with an industrial motor-control unit. In the original setup of the test bench, the motor-control unit was operated in speed-control mode, i.e., an analog signal from the test bench PC was used as the reference pump speed. Because many VAD controllers and suction detection algorithms make use of the pump motor current signal, a measurement of this signal is required in the setup. The measurement of the current in the three motor phases rendered impractical and the digital output of the motor-control unit could not be used because the sampling rate is too low.

In order to get an estimate of the motor current, the speed control of the blood pump is implemented as a cascaded control system, where the outer speed-control loop is running on the test bench PC and the inner current-control loop is running on the motor-control unit. The motor-control unit is therefore operated in current-control mode. i.e. the analog signal from the PC is a reference current signal. Because the current control has a bandwidth of 600 Hz, the reference current available on the PC is used as an estimate of the actual motor current.

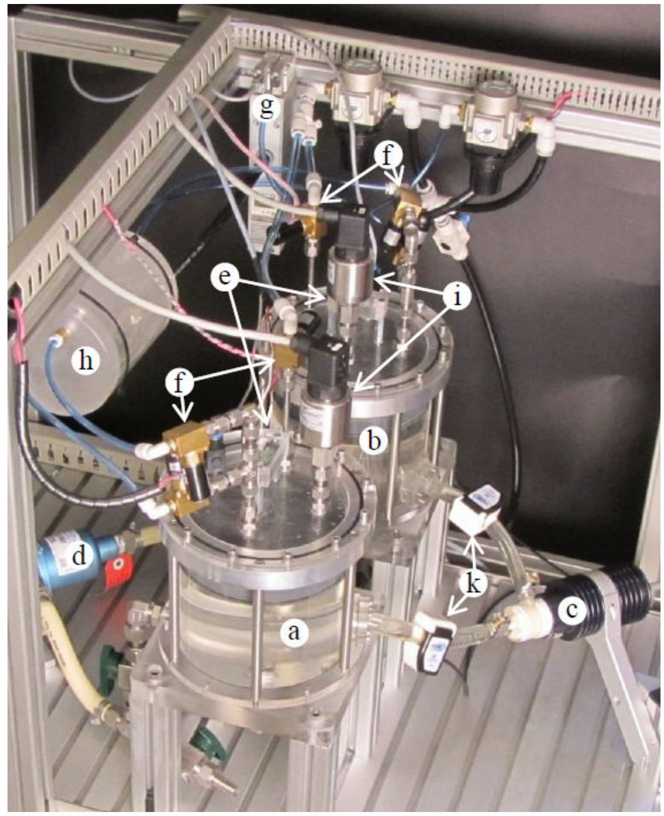

Fig. 1. Picture of the hybrid mock circulation consisting of two pressure-controlled reservoirs (a), (b), a blood pump (c), a backflow pump (d), two fluid-level sensors (e), one proportional solenoid inlet valve and two proportional solenoid outlet valves (f) per reservoir, a vacuum pump (g), a vacuum receiver (h), two air-pressure sensors (i), and and two ultrasonic flow probes (k). The two fluid-pressure sensors are not visible from this angle.

The speed-control loop implemented on the PC is a proportional-integral controller. The transfer function is given by [9]

$$C_s(s) = k_{p,s} \cdot \left(1 + \frac{1}{T_{i,s} \cdot s}\right), \tag{1}$$

where s is the Laplace variable, $k_{p,s}$ =0.005 A/rpm is the proportional gain and $T_{i,s}$ = 0.2 s is the integrator time constant. The measured pump speed signal is low-pass filtered by a second order IIR filter. The transfer function of this filter is given by

$$F_s(s) = \left(\frac{1}{\tau_s \cdot s + 1}\right)^2, \tag{2}$$

where τ_s = 0.0045 s is the filter time constant.

3) Pressure controller: The most challenging task of the numerical-hydraulic interface is the pressure control of the two reservoirs. Two single-input single-output controllers are implemented for this task. The input to these controllers is the deviation of the measured from the reference pressure signal; the output is the current sent to the proportional

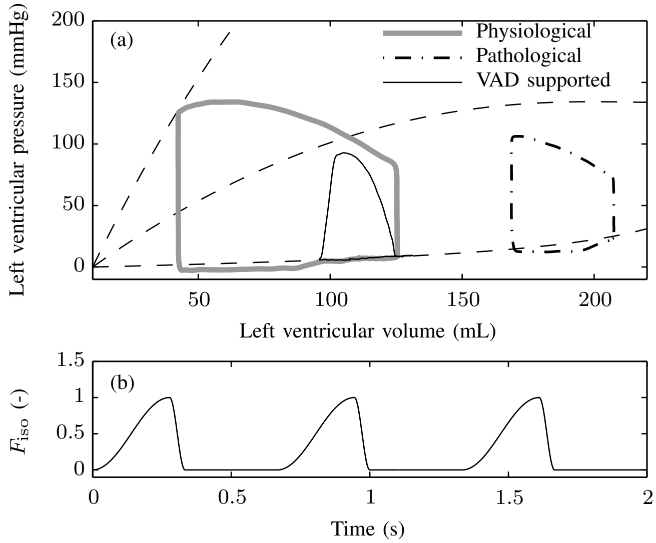

Fig. 2. Panel (a) shows measured PV loops from the hybrid mock circulation. The loops labelled "Physiological" and "Pathological" were obtained with a clamped pump cannula; the loop labelled "VAD supported" was obtained with the pathological circulation and a pump speed of 4180 rpm. The dashed lines are used for the calculation of the time-varying elastance. Panel (b) shows a 3 s excerpt of the trigger function for the time-varying elastance. This function is used to interpolated between the dashed lines in panel (a).

inlet and outlet air valves. Proportional-integral-derivative controllers are used; their transfer function is given by [9]

$$C_p(s) = k_{p,p} \cdot \left(1 + \frac{1}{T_{i,p} \cdot s} + T_{d,p} \cdot s\right), \tag{3}$$

where $k_{p,p}$ is the proportional gain, $T_{i,p}$ is the integrator time constant, and $T_{d,p}$ is the derivative time constant. The two pressure controllers have different specifications and parameters. The specifications include zero steady-state error, a small step-response overshoot, and a high bandwidth. Table I lists the parameters for the pump inlet and the pump outlet pressure controllers. The resulting control systems have a bandwidth of 25.5 Hz and 27.2 Hz, respectively.

TABLE I
PRESSURE CONTROLLER PARAMETERS

Pump inlet / LV		**Pump outlet / aorta**	
Parameter	**Value**	**Parameter**	**Value**
$k_{p,p}$	$4 \cdot 10^{-5}\,\mathrm{Pa}^{-1}$	$k_{p,p}$	$3 \cdot 10^{-5}\,\mathrm{Pa}^{-1}$
$T_{i,p}$	0.1 s	$T_{i,p}$	0.1 s
$T_{d,p}$	0.005 s	$T_{d,p}$	0.01 s

B. Suction Model

Suction describes the phenomenon when a VAD tries to pump more blood from the ventricle than is available. Since the test bench is a hybrid mock circulation, there exist two possibilities to implement suction. The first option is to use a hardware device on the inflow cannula of the VAD, which can block the flow when suction occurs. A possible hardware approach is presented in [4]. The second option

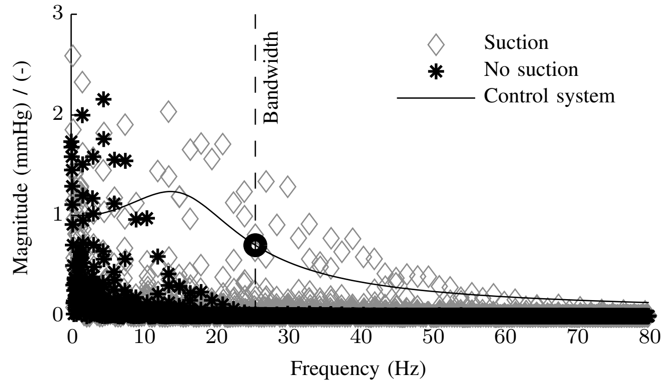

Fig. 3. Fourier analysis of the reference pressure signals at the pump inlet. The solid line represents the magnitude of the closed-loop transfer function of the pressure control system; the vertical line shows where this magnitude drops below -3 dB. Clearly, the frequency content of the reference signal without suction is below the bandwidth whereas this is not the case for the signal with suction. Accurate reference tracking can thus only be ensured when no suction occurs. The axes are cropped for a better visibility of the relevant features.

is to implement a suction model in software and to use the simulated pump inlet pressure as the reference signal instead of the LV pressure. A possible approach to model suction in a numerical environment is presented in [5]. The major advantage of the hybrid mock circulation over classic mock circulations is the flexibility, which yields from the possibility to change most parameters in the numerical model. Therefore, we decided to pursue the software approach.

The implementation of suction in the numerical model is realized as follows. The pressure used as the reference signal is not anymore the LV pressure $p_{\mathrm{lv}}(t)$ but the pump inlet pressure $p_{\mathrm{pi}}(t)$. These pressures coincide as long as no suction occurs. When suction occurs, the inflow of the cannula is blocked, which is modelled by a suction resistance $R_{\mathrm{suc}}(t)$, and the two pressures differ. The suction resistance can change depending on the suction state and it is computed by

$$R_{\mathrm{suc}}(t) = \begin{cases} 0 & p_{\mathrm{lv}}(t) \geq p_{\mathrm{th}} \\ -r \cdot (p_{\mathrm{lv}}(t) - p_{\mathrm{th}}) & p_{\mathrm{lv}}(t) < p_{\mathrm{th}}, \end{cases} \tag{4}$$

where $r = 3.5\,\mathrm{s/ml}$ is a gain, and $p_{\mathrm{th}} = 0\,\mathrm{mmHg}$ is the suction pressure threshold. The pump inlet pressure $p_{\mathrm{pi}}(t)$ downstream of this resistance is computed by

$$p_{\mathrm{pi}}(t) = p_{\mathrm{lv}}(t) - R_{\mathrm{suc}}(t) \cdot q_{\mathrm{vad}}(t), \tag{5}$$

where $q_{\mathrm{vad}}(t)$ is the VAD flow rate.

C. Reference signal frequency analysis

The main problem for emulating suction in the hybrid mock circulation is the higher frequency content of the pump inlet pressure reference signal during suction. Fig. 3 depicts the frequency spectrum of $p_{\mathrm{pi}}(t)$ with suction and without suction together with the closed-loop transfer function of the pressure controller. Clearly, the bandwidth of the control system is high enough to track the reference signal without suction, but will not be able to accurately track the reference signal with suction. Due to delays in the control system, a significant increase of the bandwidth is not possible without sacrificing the robust behavior of the controllers.

D. Experiments

Two experiments are conducted at the test bench for this paper. Both experiments are inspired by in-vivo suction measurements. The goal of the experiments is an analysis of the reference tracking during suction and an analysis of the overall suction emulation. For both experiments, the numerical circulation model is set to a pathological state by reducing the LV contractility to 34% of the physiological value and by setting the heart rate to 90 bpm [7].

1) Pump speed ramp: The first experiment inspired by [10] consists of a slow pump speed ramp from 3000 rpm to 7000 rpm. This experiment is conducted to analyze the reference tracking behavior of the controller and to analyze the overall behavior of suction.

2) Suction indices: In the second experiment inspired by [11], the pump speed is increased from 3800 rpm to 5800 rpm in 200 rpm steps. At every step, the speed is held constant for 30 s of which the last 10 s are used to calculate two suction indices. The pulsatility index (PI) is a time-domain index and is calculated by

$$\mathrm{PI} = \frac{\max\left(x_{\mathrm{vad}}(t)\right) - \min\left(x_{\mathrm{vad}}(t)\right)}{2}, \tag{6}$$

where $x_{\mathrm{vad}}(t)$ is either the pump flow rate $q_{\mathrm{vad}}(t)$, the pump motor current $I_{\mathrm{vad}}(t)$, or the pump pressure head $h_{\mathrm{vad}}(t)$. This index describes the amplitude of the respective signal. The harmonic index (HI) is a frequency-domain index and is calculated by

$$\mathrm{HI} = \frac{\int_{f_0-\delta f}^{f_0+\delta f} X_{\mathrm{vad}}(f)\mathrm{d}f}{\int_{f_0+\delta f}^{f_{\max}} X_{\mathrm{vad}}(f)\mathrm{d}f}, \tag{7}$$

where f is the frequency variable, f_0 is the fundamental frequency (heart rate), $\delta f = 0.1\,\mathrm{Hz}$ is a frequency window constant, $f_{\max} = 80\,\mathrm{Hz}$ is the maximum frequency, and $X_{\mathrm{vad}}(f)$ is the magnitude of the discrete fourier transform of either the pump flow rate, the pump motor current, or the pump pressure head. This index describes the ratio between the integral over the fundamental frequency and the integral over the higher harmonics. The second experiment is conducted to make a quantitative assessment of the suction emulation.

III. RESULTS

A. Pump speed ramp

Fig. 4 shows the experimental results of the pump speed ramp from 3000 rpm to 7000 rpm. Fig. 4(a) shows the reference and the measured pump speed. The small oscillations visible in the measured pump speed signal with the same frequency as the heart rate are caused by the changing hydraulic load from the beating heart. Fig 4(b) shows the reference and the measured pump inlet pressure and the simulated LV pressure. The reduction of the pressure in the ventricle is caused by the increasing flow rate through the pump, which reduces the blood volume in the ventricle. After 23 s suction occurs, which causes the pump inlet pressure to become increasingly negative while the LV pressure stays around 0 mmHg. From the agreement between the reference and the measured signal, one can see that the reference tracking is accurate before suction but not anymore during suction. Fig. 4(c) shows the measured pump flow rate. This measurement shows the typical reduction of the flow pulsatility towards suction and the distorted flow rate signal during suction. Fig. 4(d) shows the pump motor current. Similar to the flow rate signal, the motor current gets less

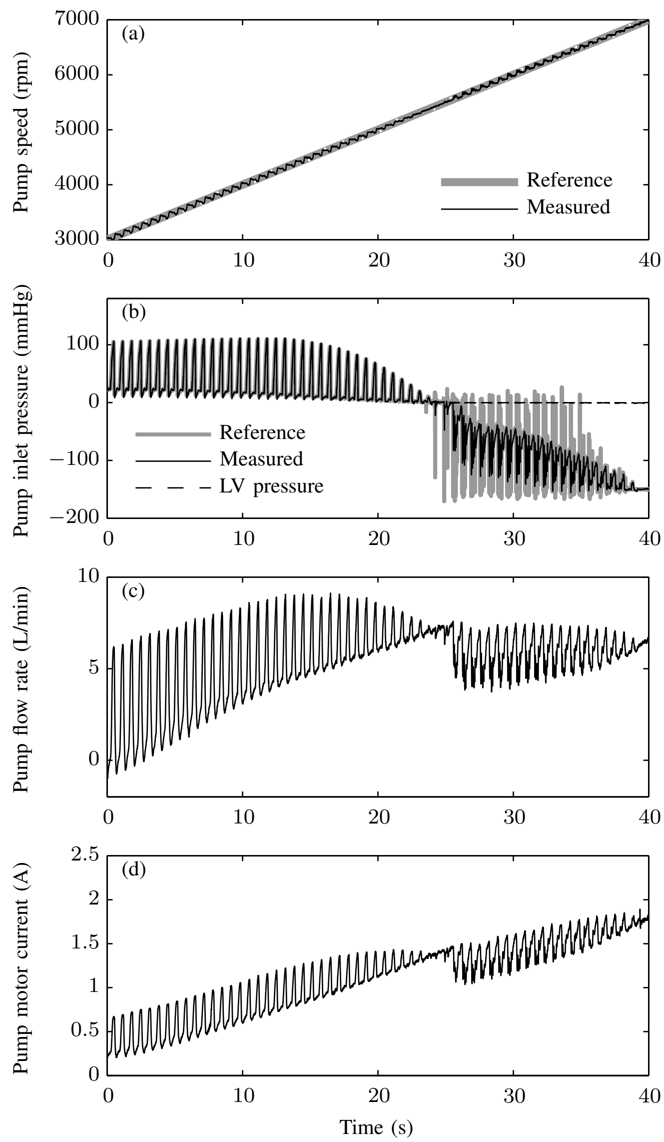

Fig. 4. Suction emulation experimental results. Panel (a) shows the time course of the reference and the measured pump speed. Panel (b) shows the time course of the reference and the measured pump inlet pressure. The reference tracking is accurate before suction but not anymore during suction. The simulated LV pressure (dotted line) coincides with the pump inlet pressure before suction but does not become negative. Panel (c) shows the resulting pump flow rate. Panel (d) shows the required motor current for the pump.

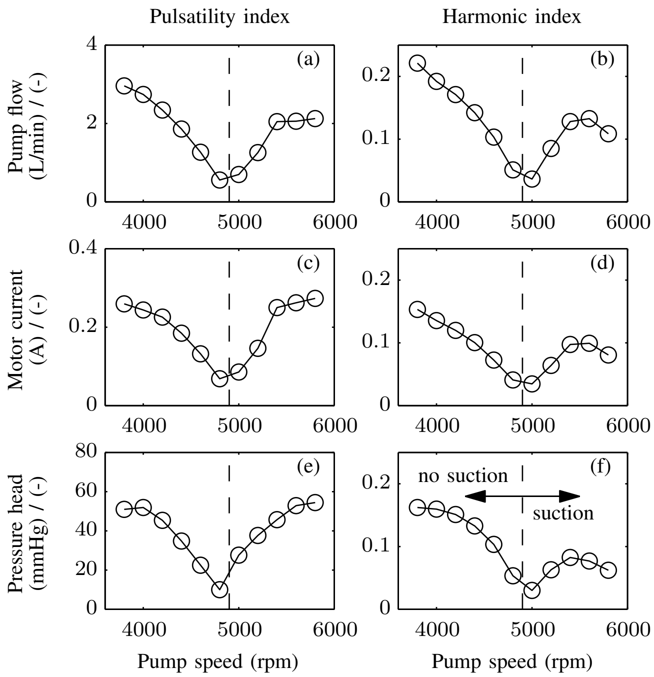

Fig. 5. Analysis of two suction detection indices. The pump speed is increased from 3800 rpm to 5800 rpm in steps of 200 rpm and held constant for 30 s. The last 10 s of every step are taken to compute the indices using (6) and (7). The panels on the left hand side represent the PI in the respective unit; the panels on the right hand side represent the HI, which has no unit. Panels (a) and (b) show the indices for the pump flow rate; panels (c) and (d) show the indices for the pump motor current; panels (e) and (f) show the indices for the pump pressure head. All pump speed values above 4800 rpm yield ventricular suction.

pulsatile towards suction and is distorted with high-frequency oscillations during suction.

B. Suction indices

Fig. 5 shows the results of the analysis of the suction indices. The indices are computed using (6) and (7) on the measured pump flow rate, the pump motor current, and the pump pressure head. All six indices show a clear minimum around the onset of suction.

IV. Discussion

This paper presents an approach to emulate ventricular suction in a hybrid mock circulation. The mock circulation is a highly flexible, physiological test bench for VADs. It has been developed as a testing environment for VAD controllers and pressure or flow observers [6] and the results from this paper show that the test bench is also able to emulate ventricular suction. With this improvement, the hybrid mock circulation is becoming an alternative method for preliminary animal experiments.

Fig. 4 shows that the pressure controller is not able to accurately track the reference signal. This behavior was predicted by an analysis of the frequency content of the reference signals with and without suction (see Fig. 3). Nevertheless, the overall behavior of suction matches well with in-vivo experimental results [10]. This observation can also be explained by the analysis in Fig. 3. The bandwidth of the pump inlet pressure controller lies at 25.5 Hz, but the amplitude spectra of $p_{\mathrm{pi}}(t)$ and $p_{\mathrm{lv}}(t)$ already differ significantly below the bandwidth when suction occurs. Additionally, the reference tracking at e.g. 30 Hz is not accurate, but the output still has more than half the amplitude of the reference signal. Fig. 5 shows a quantitative analysis of the suction emulation. Even though the numbers do not exactly match numbers from in-vivo experiments [11], it is assumed that the variability between different in-vivo experiments (different animal, different pump, different positioning of the cannula) is in the same range. All six indices show a clear minimum around the onset of suction. Fig. 4 and Fig. 5 show a difference in the pump speed when suction occurs. The reason for this difference is the relatively fast pump speed ramp in the first experiment compared to the second one, i.e., the first experiment includes the dynamics of the circulation model, whereas the second experiment shows a steady-state analysis.

V. Conclusion

In addition to the ability to test VAD controllers, the hybrid mock circulation can now also be used to test suction detection algorithms.

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