ACTIVE PNEUMATIC PULSATION DAMPER FOR PERISTALTIC PUMP FLOW LOOPS

Matthias Liermann
Fluid-Mechatronics Lab
Department of Mechanical Engineering
American University of Beirut
Lebanon 1107-2020
Email: matthias.liermann@aub.edu.lb

ABSTRACT

A novel concept of an active pulsation damper is described that cancels parasitic flow pulsatility of peristaltic pumps and is able to inject desired pulsatility signatures such as physiological heart beat. Peristaltic pumps avoid contact between the moving parts of a pump and the operating fluid. They are used for clean or sterile fluids as well as for highly aggressive fluids, whenever it is important to isolate the fluid from the environment. The background application for the proposed active pulsation damper is the simulation of hemodynamic flow.

The paper presents a novel pulse damper concept that allows the use of roller or peristaltic pumps as primary pumps for hemodynamic flow loops. The problem with peristaltic pumps is that they exhibit a high parasitic pulsatility that needs to be canceled before a desired pulsatility can be injected. The active pulsation damper does this and is also used to superpose a desired flow pattern that resembles measured heart flow rate profiles. The non-linear dynamic equations of a test system with active pulsation dampers are established and linearized to allow a first analysis of the achievable bandwidth. Simulation results of the closed loop system are presented based on the non-linear equations.

1 Introduction

According to the World Health Organization WHO, cardiovascular diseases are the main cause of death worldwide [1]. A large scientific community focuses on understanding the causes and developing possible treatments for these ailments. Pumps that replicate hemodynamic flow and pressure waveforms are central to this type of research. They have been developed since the mid 20th century and are today an integral part of laboratory research. Another class of systems simulating hemodynamic flow is used for clinical treatment. They are used on patients, for example, as mechanical circulatory support systems or for dialysis. It has been found that natural pulsatility enhances perfusion [2, 3]. It is important for those systems to be extremely reliable and to have low-cost disposables. They can be based on roller pumps, bulb pumps [4] or more recently impeller pumps. This paper gives a brief overview over both types of systems but focuses on those that are used in laboratory research for testing of implants and other experimental research such as flow visualization studies. The presented novel concept for a hemodynamic flow loop uses a peristaltic pump and an actively controlled compliance chamber that cancels parasitic pulsatility and injects a desired pulsatility profile. It is intended for precise laboratory setups, where an accurate reproduction of a desired flow waveform is important. An application is, for example, to reproduce flow and pressure signatures in various flow phantoms of the cardiovascular system based on apriori measured flow profiles from real patients. The requirements for such a device include: accurate and reproducible volume flow waveform, wide range of flow rates (even reversible flow), ease of programming, capability of producing continuous flow, prevention of entrainment of gas bubbles and cavitation, and low shear forces.
1.1 Hemodynamic Flow Devices in Literature

The literature provides a rich palette of existing devices that address these requirements. Fig. 1 can serve as a guide to categorize each device presented in the following literature review. Three categories can be distinguished based on how the pulsatile flow is created. In the first category the pulsation is created just like in the real heart: A volume contracts and expands, while the displaced fluid is delivered through an arrangement of (check) valves. The variable volume can be realized as an elastic bulb that is pressurized from outside or by a motion-controlled piston/cylinder. In the second category of devices, a pump delivers a constant flow, while a second device is used (either valve or pump) that modulates the flow to inject the desired pulsatility. In the third category a single pump is used, and through means of motion control of the pump, both, the average baseline flow and the pulsatility are realized.

<table>
<thead>
<tr>
<th>Category A</th>
<th>Category B</th>
<th>Category C</th>
</tr>
</thead>
<tbody>
<tr>
<td>Reciprocating pump &amp; valves</td>
<td>Continuous pump + flow modulator</td>
<td>Motion controlled pump</td>
</tr>
</tbody>
</table>

FIGURE 1: Categories of pulsatile pump systems

Each system has at its core a pump that can be categorized by its pumping action principle, drive system, and its mode of control to achieve the desired pulsatility. Fig. 2 lists various embodiments found for these functions.

Category A A large area of need for pulsatile blood pumps is for testing of heart implants. The design of such a device has to make a compromise between the need to accurately simulate the hemodynamic flow characteristics and being practicable for routine laboratory use. Systems for the characterization of heart valves have to simulate the contraction of the heart muscle. This motion has been emulated in many setups with the bulb pump principle, where the bulb resembles the heart chamber and is caused to contract through application of outside pressure, see left column of Fig. 1. In the Yoganathan-FDA system, pressurized air was used while the Aachen pulse duplicator used pressurized liquid. Other setups have simulated the heart contraction directly through piston pumping action (Sheffield pulse duplicator) [5]. Similar to the Aachen pulse duplicator, the commercial ViVitro Pulse Duplicator system developed by ViVitro Labs Inc. in Canada also uses pressurized liquid applied to the outside of a model heart to cause its pumping action [6]. It goes back to a device developed in 1979 by Scotten et al. to assess mitral valve prostheses [7]. A commercial version is available since 2007 and is today being used in many research projects. Other examples for commercial pulsatile blood pumps are the Harvard Apparatus Pulsatile Blood Pump or the BDC Laboratories HDT-500 Pulse Duplicator. Nevertheless many research groups still build their own test benches on the same principle with small variations, for example [8,9].

In many cases, such as in flow visualization studies in the peripheral arterial system, it is not necessary to simulate the contracting nature of the heart ventricle. The above described systems that use the bulb pump principle always require artificial valves (mitral and aortic). A model heart is expensive to replace. In addition, even though the pumping action is mechanically similar to the natural heart, the downstream pulsatility is not necessarily accurate, since it is also dependent on the compliance of the downstream system. This compliance needs to be tuned with devices such as the Vivitro device. When performing research on parts of the cardiovascular system downstream of the aortic valve, other flow loops are simpler that reproduce the pulsatility not by a true mechanical representation but by controlled operation of one or more pumps.

Category B Many systems employ two pumps, where one is used to provide a baseline flow, while the second is used to superpose a pulsatile flow, see middle column of Fig. 1. In [10] a roller pump is used to provide a baseline flow, while the pulsatile flow is simulated by a piston pump. Roller pumps or peristaltic pumps are positive displacement pumps that cause minimum damage to blood cells and can easily be cleaned, because the fluid is confined by a tube. However, they are very pulsatile themselves, which is undesired for the purpose of mimicking heart flow. In section 3 of this paper, a measurement of such pulsatile flow is presented. They have nevertheless been used. Law et al. [11] use a modified roller pump with controlled stepper motor. The implementation is limited to a specific output flow rate, and despite modifications to the pump, pulsations from the pumping action could not be eliminated. A gear pump is employed to generate the baseline flow in [12]. Gear pumps cause significant noise in the flow and pressure signal and should not be used for studies with blood cells. In [13–15] a centrifugal pump is used to provide the baseline flow because of its low noise signature compared to a roller pump. A piston pump superposes the pulsatile flow. Various patents have been published that define features of pulsatile pumps [16]. In [17] a pulsatile membrane pump device is described that is actuated by predefined cam mechanisms that are driven by a constant speed motor. Such a non-controlled system
is possible when the baseline pump is of the displacement type and therefore the flow pulsations are only affecting the downstream.

**Category C** Some systems try to emulate the pulsatile flow with only a single pump which is electronically or mechanically controlled in a way to regenerate the complete flow pattern, see right column of Fig. 1. In [18] a pump is described that uses a progressive cavity pump with a motion control system. Such a pump exhibits a low noise signature compared to other displacement pumps, such as roller pumps, but is also more complex and less simple to clean and sterilize. Servo-driven piston pumps can combine the advantage of simple components that can easily be cleaned, minimum pump noise and accurate controllability. A double piston system is described in [19] that provides non-interrupted pulsatile flow. It consists of two pistons that are connected to a common shaft. While the first one extends and delivers fluid to the system in a controlled manner, the other one is retracted and charged with fluid from a reservoir. The piston motion is controlled to realize a desired pulsatility. Piston pumps do not cause shear load on blood particles and allow good reproduction of the pulsation profile.

In more recent studies, there have been attempts to achieve pulsatile flow by means of speed modulation of an impeller pump. However, it has been shown experimentally and theoretically that at near-healthy arterial pressure pulsation patterns, strongly regurgitant flow occurs in the pump with several detrimental effects. It causes the impeller to operate in regions of inferior efficiency, which increases energy consumption; furthermore, high shear levels that result from the impeller working in regurgitant flow causes blood cell damage. To avoid regurgitation with the impeller speed modulation method, the pulsation profiles have to be properly planned and controlled [20, 21].

### 1.2 Active Compliance Chamber

Very few studies, such as [10] (top middle in Fig. 1) use a roller pump as a baseline pump because roller pumps exhibit high parasitic pulsatility. This pulsatility has to be canceled first using compliance chambers before a desired pulsatility can be superposed by another pump. The system presented in this paper uses a roller or peristaltic pump with a special compliance chamber that is used to cancel parasitic and inject desired pulsatility at the same time, see Fig. 3. This has not been done in previous studies.

The compliance chamber is a flow-through type, also referred to as inline type pulsation damper. The setup is very simple, easy to clean, and the amount of fluid in the loop is small. The pressure in the compliance chamber is modulated to accelerate and decelerate the downstream flow. A pneumatic control valve is connected to the compliance chamber. The valve is controlled by a controller that realizes the flow and pressure pulsatility based on an internal reference and measured signals of flow and pressure in the downstream tube.

In the following, the paper describes the mathematical model of a flow loop with active compliance chambers. The frequency response of the system allows preliminary assessment about the achievable bandwidth of the pulse control based on a linearized model. Simulation results are presented that are obtained with a nonlinear simulation model and a simple proportional controller.
2 Mathematical Model

The flow loop with two active compliance chambers is illustrated in Fig. 4. The system consists of five main components: compliance chambers A&B, pneumatic high response control valve, flow loop test section and the pump. For the purpose of analysing the dynamics of the active compliance chambers, and their ability to inject desired oscillating flow patterns, it is not important to include the pump in the model as long as it can be assumed that it is a displacement pump with high input and output impedance. That means that the output flow rate is little affected by pressure pulsations in the system. The peristaltic pump is therefore excluded from the following model, but included later in the analysis. The pump delivers the same amount of fluid that it consumes and introduces parasitic pulsatility into the system. This pulsatility can be treated as a disturbance and can later be superposed to the flow produced by the active flow chambers.

The air volume of the compliance chamber can be modeled with the assumption that no heat exchange takes place during fast changes in pressure. With this assumption the change of internal energy \( U \) is equal to the enthalphy influx \( H \) minus the work done by the air volume \( W \).

\[
U = H - W
\]  

(1)

The work done by the fluid is

\[
W = p_A V_A,
\]

(3)

and the enthalpy influx is

\[
H = \begin{cases} 
\dot{m}_A c_p T_A & \text{if } \dot{m}_A < 0, \\
\dot{m}_A c_p T_{\text{in}} & \text{if } \dot{m}_A > 0.
\end{cases}
\]

(4)

where \( T_A \) is the chamber temperature and \( T_{\text{in}} \) is the temperature of the incoming gas from the valve. Assuming adiabatic flow through the valve, the incoming temperature is

\[
T_{\text{in}} = T_S \left( \frac{p_A}{p_S} \right)^{\frac{T_A}{T_S}-1}
\]

(5)

In the following \( T_{\text{in}} \) is used as the temperature of the fluid entering or leaving the chamber. Combining Eqs.(2-4) and isolating for the pressure gradient gives

\[
\dot{p}_A = \gamma \dot{m}_A R T_{\text{in}} - \gamma p_A V_A, 
\]

(6)

where the change of chamber volume is the difference between out- and inflowing volumetric flow rates

\[
\dot{V}_A = Q_L - Q_p. 
\]

(7)

Chamber B is modeled accordingly.

The control valve modulates the mass flow in and out of the chambers. It is nonlinearly dependent on the upstream and downstream pressures, upstream temperature and valve opening. Several ways exist to model this flow. Applying the model described in the standard ISO 6358 [22] is useful when the respective model parameters are given in the valve data sheet, namely the sonic conductance \( c \) and the critical pressure ratio \( b \). Using the density of air at standard reference conditions \( \rho_0 \) and \( T_0 = 293^\circ K \), the mass flow rate equation can be written as
The equation depends on the flow direction given by the valve opening $u$ and the ratio between downstream and upstream pressure. The mass flow rate of chamber B is obtained similarly.

The flow loop test section is usually not very long ($< 1 \text{ m}$) and can be modeled with a hydraulic resistance and inductance. It is assumed that the capacitance of the line is insignificant compared to the capacitance of the compliance chambers and is therefore not modeled. The combination of resistance and inductance can be modeled as a dynamic system of first order. The pressure drop $p_A - p_L$ caused by resistance to the flow $Q_L$ depends on whether it is laminar or turbulent. It can be modeled with the Darcy-Weisbach equation [23, 24] as:

$$p_A - p_L = \left\{ \begin{array}{ll}
Q_L \frac{12\mu h}{A} \left( \frac{1}{2} \right) &, \text{Re} < 2300 \\
Q_L \frac{0.3164 \cdot 8 \cdot \rho_l}{L} \left( \frac{1}{Re^{0.25}} \right) &, \text{Re} \geq 2300
\end{array} \right. \quad (9)$$

Eq. 9 is a coarse approximation. It does not attempt to describe laminar- turbulent flow transition and is assuming smooth pipes and stationary flow conditions. For the purpose of this study this simplification is sufficient.

The pressure drop $p_L - p_B$ due to acceleration of the fluid is modeled by a simplified inductance term as

$$p_L - p_B = \frac{L}{A} Q_L \quad (10)$$

The parameters of the model are listed in Table 1.

### 3 Performance Analysis and Simulation results

The above described model is of at least 5th order. The state variables are: the pressures $p_{A,B}$ and liquid volumes $V_{A,B}$ of each chamber, and the volumetric flow rate in the test section $Q_L$. All other variables are algebraic variables. The valve spool dynamics were neglected because the desired closed loop bandwidth is much lower than the valve natural frequency, which is estimated $\omega_v = 400 \text{ Hz}$. The system can be linearized using the parameters listed in Table 1. Because of the symmetry of the system it seems appropriate to cancel the pressure in chamber B as a state. In the linearized model the change of pressure in chamber B is always equal in value but opposite in sign to the change of pressure in A. The simplified linear model $G_p$ can be represented as a transfer function with valve signal $u / u_{max}$ as input and $Q_L$ as output.

$$G_p = \frac{1.619 \cdot 10^5 \text{ m}^3}{s^4 + 3519 s^3 + 6.316 \cdot 10^5 s^2 + 2.218 \cdot 10^6 s + 3.536 \cdot 10^8} \quad (11)$$

Represented as frequency response plot, the model is depicted in Fig. 5. The plot shows only the relevant frequency range up to 15 Hz = 100 rad/s. In this range the system appears simply as a low damped second order system with a high resonance at around $1.2 \text{ Hz} = 7.5 \text{ rad/s}$. The damping ratio is $\zeta = 0.02$. The simplicity of the model is a pleasant surprise.

The control loop is closed by taking the measurement of the pulsatile flow $Q_L$ and comparing it to the desired pulsatile flow $Q_{ref}$. The difference $e = Q_{ref} - Q_L$ is amplified by the controller and applied as input $u$ to the valve. A proportional-derivative control is suitable for such a system. To prove the concept of the system, a simple controller has been determined as

$$G_c = \frac{U(s)}{E(s)} = 3.018 \cdot 10^{-4} + 1789 s \quad (12)$$

Fig. 6 shows the closed loop frequency response that shows a very promising bandwidth of around $10 \text{ Hz} = 63 \text{ rad/s}$. The controller is tested in simulation with a nonlinear implementation of the model in Dymola using the equation based modeling language Modelica. This model includes, beyond the men-
mentioned equations in the previous section, a 2nd order dynamics model for the motion of the valve spool ($\omega_V = 400\text{Hz}, \zeta_V = 0.7$), a limitation of the valve spool velocity, and the saturation of the valve spool opening at 100%. A graphical representation of this model is shown in Fig. 7.

The reference flow rate is a typical aortic flow at a heart rate of 85 bpm [25]. The average flow rate is $5.3 \cdot 10^{-5} \text{m}^3/\text{s} = 3.18 \text{L/min}$. The peristaltic pump is set to deliver this average flow rate. The flowrate pulsatility of a peristaltic pump was measured by PIV in [26,27] and is given as input to the simulation. The pulsatility of the peristaltic pump is on a much higher frequency band and can therefore be absorbed in the compliance chamber, while the desired pulsatility can be injected by the control loop. Fig. 8 shows the two input signals used in the simulation.

### 3.1 Simulation Results

Two simulation results are shown in the following. The first simulation result shows the damping capability of the compli-
rance chamber, when the control is switched off, see Fig. 9. The top diagram shows the pulsatile flow delivered from the peristaltic pump \( Q_P \). The second diagram shows the flow in the test section \( Q_L \). It starts from zero initial conditions and is accelerated to the average flow rate delivered by the peristaltic pump. The passive compliance of the damping chambers smoothes out the flow ripples effectively. After 3.5 s steady state flow conditions can be observed. The 3rd and 4th diagram show the development of pressures in the chambers and the height of the water line. In both signals the small perturbations caused by the flow ripples of the peristaltic pump can clearly be observed. At steady state conditions there is a certain pressure drop that corresponds to the resistance of the test section.

![FIGURE 9: Simulation results showing flows, pressures and heights of liquid levels in chambers while control is switched off.](image)

The second simulation result, where the control is switched on, is seen in Fig. 10. The plots show larger detail in the temporal resolution. The simulation time is chosen between 3.5 – 5 s. At this time, the large scale transients that could be seen in Fig. 9 have vanished. The top diagram compares the reference flow \( Q_{ref} \) with the actual flow in the test section \( Q_L \). The pump flow \( Q_P \) is also shown. The top plot shows a good reproduction of the desired flow pattern with some phase lag and overshoot. The second diagram shows the valve input signal. The valve cannot open more than 100%, therefore it is evident that the control demands higher flow gains than the valve can produce. The size of the valve corresponds to an available valve in our lab (Festo MPYE-5-M5-010-B), which is planned to be used for experimental testing in future work. It is the smallest valve of its product range, so the saturation effect that is seen in the simulation results does not represent a general limitation of the working principle. Even with the saturation, the match between desired and achieved pulsatility is very promising. The 3rd and 4th diagram show the pressure and waterline height variations in the chambers. It can be seen that the average value of pressure in chamber A is still above the average pressure of chamber B. The valve, however, modulates the instantaneous pressure difference back and forth in order to superpose the desired pulsatility. The heights of the waterlines in chambers A and B shown in the bottom diagram are slightly offset compared to Fig. 9. This is caused by a mismatch between the average flows delivered by the peristaltic pump and the reference flow profile. It can also be caused by the control error. The level of the compliance chambers should be monitored by a sensor and a super-ordinate control loop should ensure that neither is depleted.

4 Conclusion

This paper presents a proof-of-concept for a novel active pneumatic damper concept that allows the use of peristaltic pumps for the simulation of heart flow patterns in laboratory setups. The peristaltic pump is the pump of choice for testing with sterile or aggressive fluids because no pump components are in contact with the fluid. However, due to its high pulsatility it is rarely used and the literature shows that preference is often given to other pumps that are much more difficult to clean and sterilize. For hemodynamic flow simulation, a peristaltic pump has to be coupled with a compliance chamber that can cancel the flow
pulsations and another device that modulates a desired pulsatility. Examples for such systems have been found in the literature. It was the research question of this study, whether the latter two functions, the elimination of parasitic pulsatility and the generation of a desired flow pattern, could be combined in a single device. A simulation study was used to answer this question.

An active pulsation damper is described. It has a characteristic passive compliance that is effective in cancelling the parasitic pulsatility of the pump. At the same time it is connected to a controlled pressure supply by which its internal pressure can be adjusted. A control loop is used to modulate the internal pressure based on measurement of the actual flow in the test section. The paper presents the nonlinear system equations and the linearized model that is the basis of a proportional-derivative control design. The control is implemented in simulation with the nonlinear model. The passive pulsation cancelling properties of the system are demonstrated, as well as the ability to follow the heart flow rate pattern for a typical 85 bpm pulse. It is seen that the valve is slightly undersized compared to the sizes of the chambers and geometry of the test section. But it is also evident that the bandwidth of the control loop is high enough to reproduce the desired pulsatility.

The work is ongoing to implement the proposed concept. The main anticipated challenge will be the measurement of the actual flow in the test section. A significant lag added in the measurement is going to decrease the achievable bandwidth of the closed loop control. It will be tested as well, whether the flow pulsations can simply be added by feed-forward control with pre-generated valve signals and a superordinate control of the levels of the compliance chambers.

5 Acknowledgements

The author would like to thank Dr. Ghanem Oweis for the measurement of the peristaltic pump pulsatility.

Funding The author gratefully acknowledges the funding by the American University of Beirut, University Research Board for its support to conduct this research.

REFERENCES


