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Fluidic Operational Amplifier for Mock Circulatory Systems – Simulation and Experimental Results

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***Abstract*— For the development of cardiovascular devices and the study of the dynamics of blood flow through the cardiovascular system, hardware fluidic models are commonly used to minimize animal experiments and clinical trials. These systems, called "mock circulatory systems" (MCS) are also critical for the development of ventricular assist devices (VAD). The passive and active elements in these systems are frequently "hard-plumbed" and are difficult to modify in experimental studies. Therefore we propose a novel fluidic operational amplifier comprised of a high-gain feedback-controlled gear pump. With pressure being the analog of voltage, design with the fluidic op-amp is analogous to electrical op-amp design. Initial computer and hardware simulation results demonstrate that device may be programmed for use in mock circulatory systems to emulate the function of the energy sources (the ventricle) or passive networks (hemodynamic loads).**

I. Introduction

Mock circulatory systems (MCS's) are valuable laboratory instruments for the understanding of blood flow dynamics, or "hemodynamics," in animal and human cardiovascular systems. Our particular interest is motivated by the development and *in vitro* evaluation of ventricular assist devices (VAD's). As various types of VAD's and related control systems are developed, new features must be evaluated before animal testing and subsequent use in humans. This need is driven by the obvious ethical and economic requirements, and also the scientific need to evaluate VAD's over a wide range of operating conditions that would not be possible with a single *in vivo* trial. Conventional MCS's are not flexible in modifying their characteristics because they are made of "hard-plumbed" (approximately) linear fluidic elements for simulating resistances (R), inertances (L), compliances (C) of the cardiovascular system, and pumps for the heart (the R, L, C and source analogs to electric circuits). Hence, the MCS setup needs to be physically modified and recalibrated each time a change is desired to its parameters and/or features. This is inefficient, time consuming and costly. Moreover, it has proven very difficult to simulate the precise nonlinear physics observed in the body with static fluidic elements.

Motivated by this need, we have proposed an innovative fluidic operational amplifier. Both passive and active elements of a conventional mock loop are replaced by a programmable servo-controlled gear pump. Pumps of this class behave analogous to controller current sources. Thus, they may be programmed to provide a prescribed dynamic relationship between pressure and flow depending on the sensing element and feedback control. A general configuration includes a pair of integrated pressure sensors.

An issue in the design of any operational amplifier (pneumatic, mechanical, electronic or fluidic) is the gain-bandwidth product. A high-impedance load for a fluidic op-amp is simply a short obstructed flow path. To obtain high-bandwidth pressure response, therefore, the output stage of the op-amp requires high torque, low-inertia in the motor, and low inductance in the motor coils. Given that cardiovascular simulation requires bandwidths on the order of 10 times the heart rate (i.e. 10-20 Hz), this is a reasonable requirement for such a fluidic element.

A complete mock circulatory system can comprised of three fluidic op-amps, one for left ventricle simulation, a second for combined right ventricle source and left atrium compliance simulation, and the third for combined aortic compliance, systemic resistance and inertance.

In this paper, we present the simulation results of the fluidic op-amp based MCS to evaluate the performance of the MCS in terms of physiological requirements. We also report the initial experimental results from the prototype fluidic op-amp that was built from the off-the-shelf components.

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II. METHODS

A. Design of Mock Circulatory System

Mock circulatory systems have been used as test platforms for *in vitro* evaluation of VADs and their feedback controllers [1-12]. The purpose of a MCS is to reproduce the *in vivo* hemodynamic responses of the native circulatory system to VAD operation over a wide range of cardiac (dys)function [11].

The native human body cardiovascular circulatory system can be simply modeled and implemented in MCS using linear lumped elements as in Fig. 1. This model is adopted and modified from Loh et al. [10] and is the target system of the proposed op-amp based MCS.

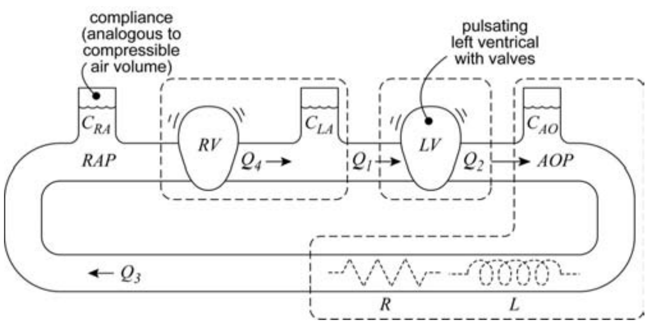

Figure 1. Simplified cardiovascular circulatory system schematic.

In the figure, LV and RV represents left and right ventricle respectively. The time-varying flows Q_1 and Q_2 are the flows into and out of left ventricle, Q_3 is the flow through the systemic circulation, and Q_4 represents the flow out of the right ventricle. The compliances C_{RA}, C_{LA}, and C_{AO} denote the compliance of the right atrium, left atrium and aorta respectively and, are represented schematically by compressible air volumes above the blood circulating in the system. The impedances R and L represent the systemic resistance and inertance. LAP, AOP and RAP represent left atrial pressure, aortic pressure and right atrial pressure.

Simulating the native ventricle's pumping function is a key ingredient for the physiologically meaningful MCS. The native left ventricle's pressure and volume is influenced by the venous return to the ventrical (preload) and the arterial dynamics seen during ejection (afterload). In rough terms, the ventricle is very elastic during filling (diastole) and accepts essentially all the blood supplied by the venous return. During systole, the ventricles become inelastic and contract to a volume which is roughly independent of the arterial pressure (pulmonary or aortic) against which they are pumping. This rough behavior of pumping all blood that is returned to the heart is called Frank-Starling's law [13]. A more detailed model of the heart incorporates nonzero elastance during diastole, and finite elastance during systole so that some effects of pre-load and afterload appear in simulation. A commonly used model of the ventricles describes the ventricular behavior as linear, elastic, and time-varying so that the left ventricular pressure (LVP) is related to the left ventricular volume (LVV) by

$$LVP = E(t) \times LVV \qquad (1)$$

where *E(t)* is modeled, for example, as a raised cosine pulse. Defining E_{min} and E_{max} as the minimum and maximum values of *E(t)* over a cardiac cycle, one can capture the P-V loop between lines with corresponding slopes as shown in Figure 2. The line $LVP = E_{\max} \times LVV$ is called the end-systolic pressure volume relationship (ESPVR) and the line corresponding to $E_{\min}$ is called the end-diastolic pressure-volume relationship (EDPVR).

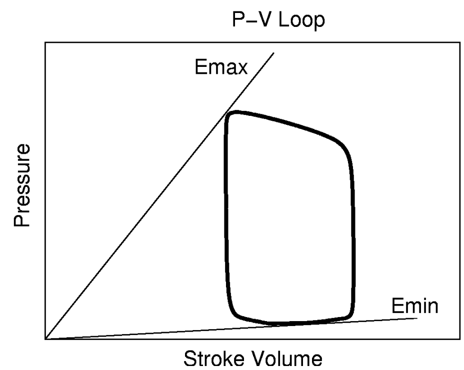

Figure 2. Pressure-volume relationship (E(t)) of the left ventricle

In developing and evaluating a MCS, controlling the load conditions seen by the ventricle is critical since different load conditions create different hemodynamic responses (pressure and flow) due to the time-varying elastance model. Formal definitions of "preload" and "afterload" vary, but for our purposes we define preload as the venous return *flow* into the left ventricle and afterload as aortic *pressure*. Changes in these loads occur with exercise, sympathetic and parasympathetic stimuli, and VAD intervention etc. In MCS's, a valve is frequently used for the mock systemic resistance and is manually adjusted for the afterload variation.

Cardiac output (the average flow $\overline{Q}_2$ in Fig. 1) is known to be dependent on the *RAP* and can be modeled empirically as [10]:

$$\overline{Q}_2 = K_{co}\left[1 - \exp\left(-\frac{RAP + 0.5}{3}\right)\right] \qquad (2)$$

The gain K_{co} is determined from the level of activity and/or sympathetic stimulation, thereby controlling the cardiac output. By Frank-Starling's law, the venous return flow is assumed to be the same as the cardiac output. Hence, (2) can be used for the preload control (Q_4).

The mathematical model of the MCS in Figure 1 can be developed as follows using the electrical analogy:

Left Ventricle

$$LVP = E(t) \times LVV \tag{3}$$

$$LVV = \int (Q_1 - Q_2) \tag{4}$$

Mitral Valve

$$Q_1 = \begin{cases} \dfrac{LAP - LVP}{R_{mtr}} & LAP \geq LVP \\ 0 & otherwise \end{cases} \tag{5}$$

Aortic Valve

$$Q_2 = \begin{cases} \dfrac{LVP - AOP}{R_{ao}} & LVP \geq AOP \\ 0 & otherwise \end{cases} \tag{6}$$

Aortic Compliance

$$A\dot{O}P = \frac{1}{C_{AO}}(Q_2 - Q_3) \tag{7}$$

Vascular Inertance and Resistance

$$L\dot{Q}_3 + RQ_3 = AOP - RAP \tag{8}$$

Right Atrial Compliance

$$R\dot{A}P = \frac{1}{C_{RA}}(Q_3 - Q_4) \tag{9}$$

Simplified Right Heart Model

$$Q_4 = f(RAP) = K_{co}\left[1 - \exp\left(-\frac{RAP + 0.5}{3}\right)\right] \tag{10}$$

Left Atrial Compliance

$$L\dot{A}P = \frac{1}{C_{LA}}(Q_4 - Q_1) \tag{11}$$

R_{mtr} and R_{ao} represents the mitral and aortic valve resistance.

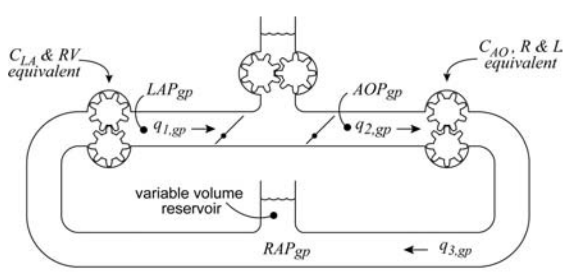

Figure 3 Fluidic op-amp based MCS

B. Gear Pump Element

Conventional MCS's are built with hard-plumbed approximately linear lumped passive elements. Therefore, it is difficult to modify the existing MCS to vary parameters or create nonlinear elements etc. In this section we present an innovative fluidic operational amplifier based on a gear-pump output stage.

As shown in the Fig. 1, the whole MCS is divided into 3 subsystems partitioned with dotted lines. Our approach is to simulate the overall function of each subsystem with discrete fluidic op-amps shown in Fig. 3.

A more comprehensive model of the fluidic op-amp, including the instrumentation and control components is shown in Fig. 4.

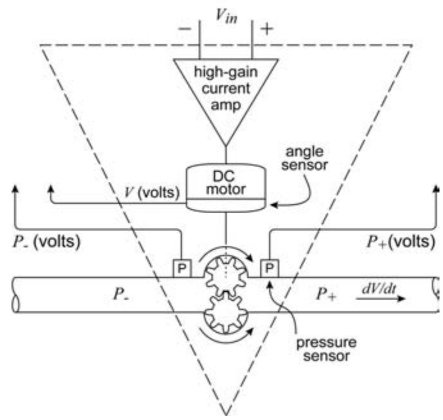

Figure 4 Fluidic Op-Amp

The central element of the op amp is a gear pump, a positive displacement pumps in the sense that the displace volume, V, is proportional to the pump rotation, θ

$$V = \alpha\theta \tag{12}$$

where α is the pump constant (volume per unit rotation). Differentiating we have the flow relationship

$$Q = \frac{dV}{dt} = \alpha\dot{\theta} \tag{13}$$

Assuming a frictionless pump, conservation of energy dictates that the input torque, τ, is related to the pressure across the output ports by the same constant α.

$$\tau = \alpha(P_+ - P_-) \tag{14}$$

where P_+ and P_- are the outlet and inlet pressures respectively.

A servo-position controlled motor will control the flow, and velocity-control will regulate volumetric flow rate. If we add dynamic terms for the pump and motor inertia, J, and combine the gain of the current amplifier and the motor torque constant into the single relationship $\tau = KV_{in}$ the fluidic op-amp can be modeled as

$$J\dot{\omega} - \alpha(P_+ - P_-) = KV_{in} \tag{15}$$

$$Q = \alpha\omega \tag{16}$$

where J is the combined inertia of the motor and the gear pump and V_{in} is the motor control input.

C. Experimental Set-Up

In order to verify the concept of the fluidic op-amp, we have assembled a prototype with the off-the-shelf components. This prototype consists of a magnetically-coupled gear pump (Model TXS79EEEV3WN; Tuthill Corp., Concord, CA) and a brushless servomotor (Model A0100-104-3; Applied Motion Products, Montville, NJ). The servomotor is driven by a servo amplifier (Model B25A20FACQ; Advanced Motion Controls, Camarillo, CA). The inlet and outlet pressures of the gear pump are measured by the combination of the disposable pressure transducers (Model PX260; Edward Life Sciences, Irvine, CA) and the patient monitors (Model 78532A; Hewlet-Packard, Palo Alto, CA). An electromagnetic flow probe (Model 300A; Carolina Medical, King, NC) measures the flow rate either at the inlet or the outlet of the pump. The mechanical hardware was combined into a feedback loop system that either controls the flow or the pressure difference at the pump. The block diagram shown in Fig. 5 controls the flow the pump.

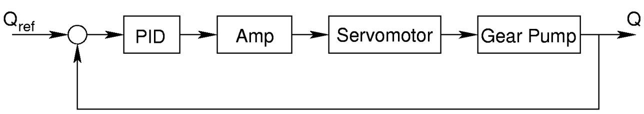

Figure 5 Block diagram of the flow control loop.

MatLab/Simulink (MathWorks Inc. Natick, Ma) was used as the software platform interfaced to the gear pump with a PCI interface board (Quanser Inc., Markham, Ontario). Fig. 6 shows the picture of the experimental setup.

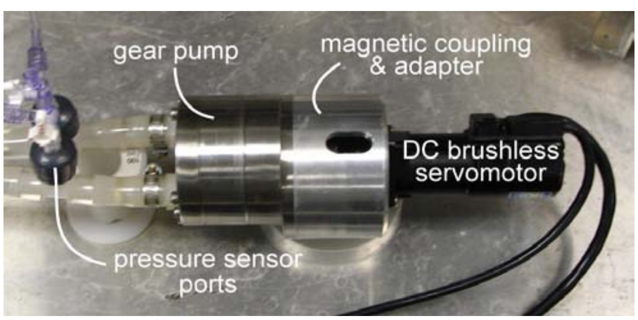

Figure 6. Preliminary prototype assembled from off-the-shelf components. The displacement of the pump is 7.9cc and the gears are made of PEEK. The servomotor has the optical encoder which has the 2000 counts per revolution.

III. Results

A. Simulation of Cardiac Dynamics

As stated before, a set of three fluidic op-amps can replace the full mock circulatory system. We designed a nonlinear controller for the gear pump [15] and carried out a computer simulation of the fluidic op-amp based MCS to investigate the hemodynamic response of the proposed MCS to changes in load as would be experienced in the body. For any MCS to be physiologically valid, the following performance objectives need to be satisfied to the load changes:

- Obey Frank-Starling law—linear relation between stroke work and stroke volume (responsiveness to preload change)
- Stroke work done by left ventricle ($W = \int PdV$) should remain unchanged for afterload variation (responsiveness to afterload change)
- Consistency of Emax (ESPVR) regardless of preload and afterload changes

Before validating above performance objectives to the load changes, waveforms of key hemodynamic variables of the fluidic op-amp MCS simulator need to be examined to confirm the physiologic validity. Fig. 7 shows the wave forms generated by the fluidic op-amp MCS simulator, corresponding to Fig. 3 with nominal values of preload and afterload (Kco = 65, R = 2.1). The hemodynamics appear to be physiologically reasonable as they closely match the nominal values established by literature [13]. They also look reasonable in a qualitative sense: the ventricular pressure resembles the elastance waveform, with the aortic pressure corresponding to LVP through systole, and decaying according to a typical windkessel model. LAP likewise rises during systole, and drops to correspond to LVP during diastole.

As hemodynamic waveforms are verified to be physiologically reasonable, performance objectives are now

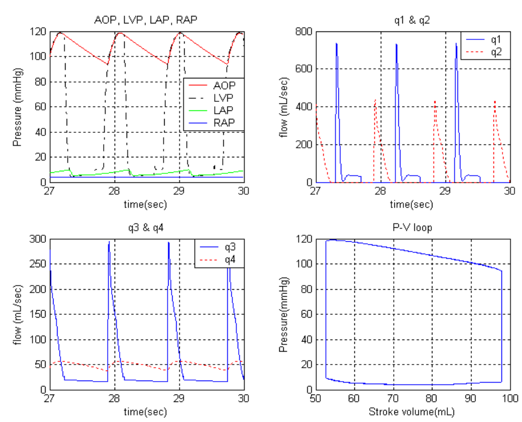

Figure 7 Waveforms of hemodynamic variables in the proposed MCS

to be verified with respect to load changes. The subsequent preload and afterload responsiveness test were performed using the same nominal values.

Step changes of preload were introduced by varying value of Kco – as defined in (2). The resulting left ventricular pressure-volume (P-V) loops for three cases (Kco = 65, 100, 150) are shown in Fig. 8. Here, it is observed that increased preload results in an increased end-diastolic volume (EDV), as expected physiologically. Furthermore, the increased EDV causes an increase in the developed pressure, maintaining Emax (ESPVR) constant indicated by the straight line. The slope of this line is very close to the prescribed Emax (2.24) of the elastance function, thereby verifying the consistency of Emax with respect to preload changes. The relationship of stroke work (the integral within the PV curve) and stroke volume was observed to follow a straight line, which is another indication of the Frank-Starling law.

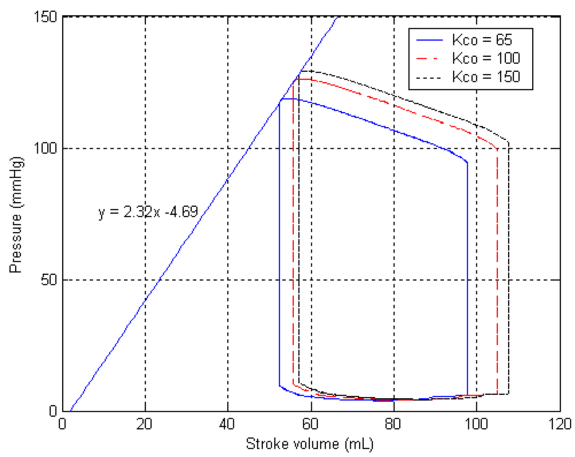

Figure 8 P-V loops for preload variation.

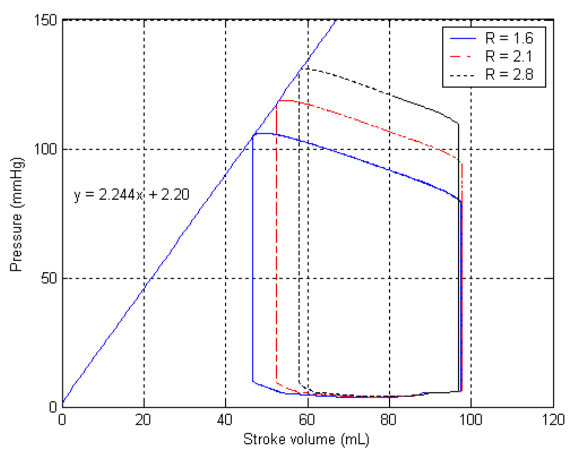

Figure 9 P-V loops and consistency of Emax (ESPVR) for afterload variations

Afterload changes were introduced by varying the systemic resistance value (R = 1.6, 2.1, 2.8). Corresponding P-V loops are shown in Fig. 9. As is the case in preload changes, ESPVR is maintained constant for the afterload changes as well and it is found to be 2.24, which is very close to Emax of the elastance function. It is verified that increased afterload reduces the returning volume, thereby reducing the stroke volume while maintaining end-diastolic volume as is observed *in vivo*. The work done by the ventricle was maintained unchanged, as observed in the body.

B. *Experimental Results*

The prototype fluidic op-amp was used to simulate either the left ventricle or the systemic resistance. In the case of the simulation for the left ventricle, the flow output of the pump must match the physiologically relevant flow characteristics, while, for the simulation of the systemic resistance, the pressure profile must follow the hemodynamic data. Fig. 10 shows the results of the experiments, where the top graph compares the outlet pressure of the pump with the pressure waveforms measured in the human aorta [15]. The bottom graph contains the results of flow control experiment, where the pump flow is forced to follow the flow rate in the human left ventricle [15].

Although the experiments confirm the feasibility of the fluidic op-amp concept, the results reveal several limitations of the system. Our prototype a) does not have tachometer feedback and hence operates at limited bandwidth, b) suffers from driveline compliance due to the magnetic coupling, c) has relatively high friction due to high design pressures (tight tolerances), d) has friction ripple due to the tight tolerances of the gear teeth and bearing eccentricity ("runout") [17]. These limitations may only be overcome by the custom-made gear pumps specifically designed for the fluidic op-amp.

IV. Conclusion

A new and innovative mock circulatory system using feedback-controlled gear pumps has been proposed. The gear pumps are controlled to mimic the impedance of its corresponding active/passive fluidic elements of the conventional MCS, thereby guaranteeing the corresponding hemodynamic characteristics. Computer simulations and initial experimental results show the feasibility of the fluidic op-amp and its application to the mock circulation systems. Experimental results also reveal the limitations of the existing gear pumps. In order to obtain an adequate performance, we would need gear pumps specifically designed for fluidic op-amps that has minimal pressure ripples and high control bandwidth.

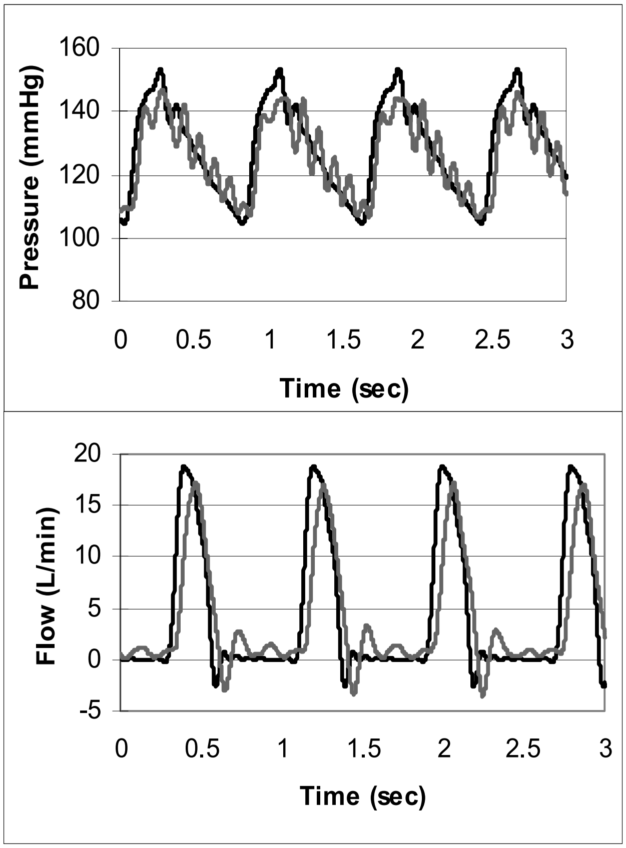

Figure 10. Pressure and flow waveforms synthesized with the prototype (with pressure and flow feedback respectively). The black lines represent the human data found in [16] and the grey lines show the experimental results. Due to the design feature of the gear pump, the results exhibit ripples [17].

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