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An Unsteady Flowmeter for Pulsatile Flow

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Abstract—This paper outlines the design of an unsteady flowmeter for ex-vivo heart perfusion applications. In order to determine the performance of a given heart, the flowrate in the system must be recorded. The proposed design is based on an orifice plate flowmeter comprising three pressure transducers and in which the flow measurement is obtained by numerically solving a modified version of the unsteady Bernoulli equation. As an intrinsic characteristic, pulsatile flows have acceleration and deceleration phases, so the flowmeter must account for these unsteady effects to produce accurate measurements. In addition to measuring laminar and turbulent bi-directional flows, the operational context also imposes several restrictions on the design, such as being disposable, low cost, easy handling and small in size. Three prototypes were experimentally tested by applying sinusoidal flows using a peristaltic pump at a frequency range of 15 - 30 Hz and then verified by applying physiological-like flow with a ventricular assist device (VAD) at a frequency of 1 Hz. The device was experimentally calibrated by means of a gravimetric procedure. The results showed that oscillatory flows can be measured with an accuracy of approximately 1% for the range from 1 to 2.5 L/min, while for lower flow rates (< 0.5 L/min) the accuracy was around 5%. It was found that the average discharge coefficient among the three manufactured flowmeters was C_d = 0.7119.

Keywords-component; Flowmeter; Design & Manufacturing; EVHP

I. INTRODUCTION

The ex-vivo heart perfusion (EVHP) system is a transplant-assisting device that keeps a human donor heart beating outside the body during the time prior to transplant surgery. This allows for transplant windows to be extended by days compared to the traditional method of static cold storage (SCS) [1]. By allowing more time for the heart transplant, it would be feasible to transport organs longer distances across the country which would lead to fewer wasted organs. This is especially important considering that in Europe and North America, 10-12% of patients on the heart transplant waitlist die before receiving a heart [2]. In addition, the EVHP system is also designed to allow the transplant team to quantitatively assess the organ performance prior to transplantation. For this to happen, accurate readings of the heart's inflow and outflow rates must be taken. Since the heart works in a pulsatile regime, such measurements pose a challenge due to the unsteady effects of the acceleration and deceleration phases of the flow, which cannot be negligible. Additional design constraints, such as the need for the flowmeter to be disposable due to biological hazards from blood contact, make the use of off-the-shelf flowmeters unfeasible, imposing the development of a new device.

Flow measurement by pressure drop across an orifice plate is well known and has the potential to meet the above criteria. The orifice principle first needed to be adapted to the unsteady flow conditions. Thus, the Bernoulli equation for the flowmeter proposed here had to be slightly modified due to the current use of three pressure sensors, as proposed in [3].

This paper covers the design and manufacturing of the flowmeter to meet the design requirements imposed by the EVHP system. In addition, the development of an analytical model to obtain flow from differential pressure measurements is discussed, as well as experimental validation of the prototype.

II. FLOWMETER DESIGN

A. Design Requirements

For the flowmeter to work in the EVHP system, the following specifications must be met:

- Inexpensive to manufacture;
- Small in size;
- Easy to install and remove;
- Accurate and repeatable for unsteady flow;

Since the flowmeter will come in contact with blood, it must be disposed of whenever a new heart is placed in the EVHP system. Thus, to reduce the overall cost, the flowmeter must be cheap and quick to manufacture. Also, it must be easy to install and remove from the system, since they will have to be changed out for every new heart, and be sized according to the hydraulic circuit of the EVHP system, which is 12.7 mm (1/2") inner diameter. And in addition to being able to work in a specific frequency range, the flowmeter must be suitable for measuring both laminar and turbulent bi-directional unsteady flows.

B. Flowmeter Prototype

Flow measurement through an orifice plate is well established and has the potential to meet the above criteria. By applying the principle of differential pressure measurement with the unsteady Bernoulli equation, it is possible to infer the flow rates of partially developed flows in laminar and turbulent regimes, whether quasi-stationary or oscillatory, with satisfactory accuracy. The design parameters beta ratio (β), bevel angle (α) and orifice thickness (e) can be carefully determined to optimize the pressure drop, minimizing unrecoverable friction losses. An image and a schematic drawing of the prototype can be found in Figure 1 and Figure 2, respectively.

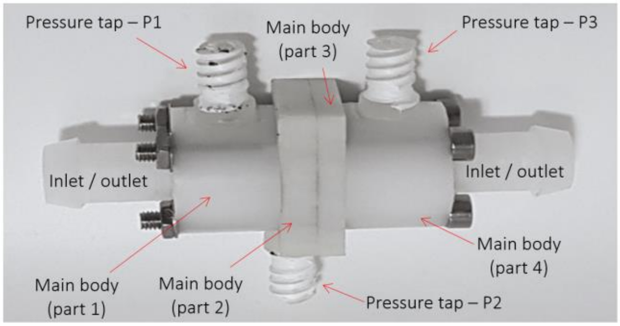


Figure 1. Image of the flowmeter prototype.

The main body of the flowmeter is made up of four parts, as can be seen in Figure 2. All parts were machined in-house using a 5-axis CNC milling machine (V2-50CHB, PocketNC Inc.). The material selected for manufacture was Delrin for its low cost, easy machinability and low surface roughness. The device also comprises three pressure taps that were printed on an SLA printer (Form 3, FormLabs) and glued to its body, allowing easy connection to Luer-lock fittings of pressure transducers (TruWave Pressure Transducers, Edwards Lifesciences Corporation). In this way, the pressure difference between the inlet and the throat, regardless of the flow direction, can be measured. Finally, four screws were used to compress and lock the entire body and three o-rings were placed between each face to seal the flowmeter.

III. MODELING

The unsteady Bernoulli equation for an irrotational, constant density flow while neglecting gravitation effects, as proposed in [3], is:

$$\frac{\partial \varphi(x,r,t)}{\partial t} + \frac{1}{2}u^2(x,r,t) + \frac{P(x,r,t)}{\rho} = C \qquad (1)$$

where $\varphi(x,r,t)$ is a velocity potential, $u(x,r,t)$ is the time varying velocity field, $P(x,r,t)$ is the pressure field and C is a constant for a given streamline.

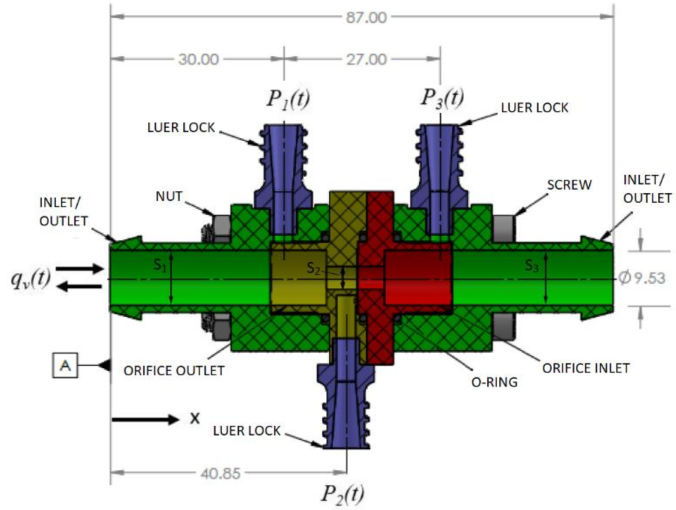


Figure 2. A cross section view of the flowmeter assembly. All units are in millimeters.

As showed in [3], when assuming a uniform 1D velocity profile, the solution to the unsteady Bernoulli equation becomes:

$$A\frac{dq_v(t)}{dt} + B|q_v(t)|q_v(t) = \frac{\Delta P(t)}{\rho} \qquad (2)$$

where $q_v(t)$ is the volumetric flowrate which can flow in either direction, $P(t)$ is the pressure at each pressure transducer, and S is the cross-sectional area at each section, ρ is the fluid density and the constants A and B are parameters based on the flowmeter geometry defined as:

$$A = \int_{x_1}^{x_2} \frac{dx}{S(x)} \qquad (3)$$

$$B = \frac{1}{2}\left(\frac{1}{{S_2}^2} - \frac{1}{{S_1}^2}\right) = \frac{1}{2}\left(\frac{1}{{S_2}^2} - \frac{1}{{S_3}^2}\right) \qquad (4)$$

Here, S is the cross-section area at a given axial position. Finally, the change in pressure, ΔP, is arbitrarily chosen to be positive when the flow is in the positive x-direction and negative when reversed. It is important to note that in equation (2) the ${q_v}^2(t)$ term is replaced by $|q_v(t)|q_v(t)$ so that the solution becomes sensitive to the flow direction. Once the volumetric flowrate is calculated (${q_v}^c$), a discharge coefficient, C_v, must be defined to account for all unrecoverable losses in the system by the correlation:

$${q_v}^c = C_v {q_v}^m \qquad (5)$$

where ${q_v}^m$ is the measured flowrate in the device calibration process. In order to solve equation (2), the Runge-Kutta 4th order method was employed with an initial guess of zero.

IV. Procedure and Equipment

A schematic representation of the experimental setup used for flowmeter validation is shown in Figure 3. A peristaltic pump (Masterflex L/S Standard Digital, Cole-Parmer Canada Company) was used to generate a range of pulsatile flows, and a pulsation damper was installed upstream of the flowmeter. Pressure transducer data was collected by a data acquisition system (DAQ). Each pressure transducer was individually calibrated against a range of water column head. The flowmeter, in turn, was calibrated using a gravimetric procedure due to its simplicity and reliability. As can be observed in Figure 3, a mass scale (GSE 550, Advanced Industrial Technologies, Inc.) was placed on the output reservoir and connected to the computer. As a complementary test, a ventricular assist device (VAD) was used to evaluate the flowmeter performance under a quasi-physiological flow regime. As the VAD is a medical grade diaphragm pump used to replace the function of a failing heart, it produces a flow profile analogous to that generated by a healthy human heart. However, as the VAD applied here is only capable of delivering one flow rate, the test becomes restricted to only one data point. In this test scenario, the pulsation dampener was not included in the experimental setup.

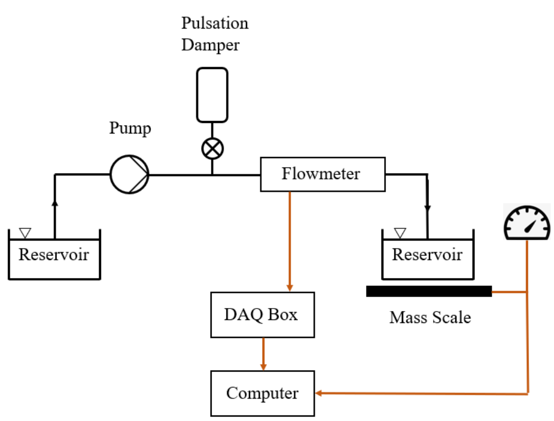


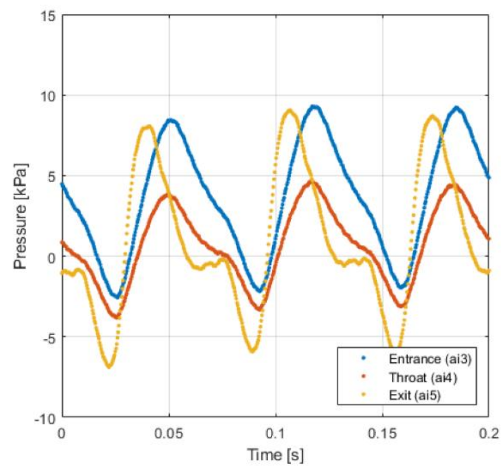
Figure 3. A schematic of the experimental setup.

The flowmeter was first tested to determine its discharge coefficient in two configurations, with and without a pulsation damper. Flow rates ranged from 0.5 L/min to 2.5 L/min in 0.5 L/min steps. A MATLAB script was written to process the pressure transducers and mass scale data as described in section 3. At each flow rate, data was acquired for 60 seconds. Although the effect of the manufacturing process on the flowmeter performance is not the scope of the present work, three similar models were tested and compared to each other to verify the consistency of the results obtained. The mass scale data was differentiated and converted to an average flow rate and compared with the flow rate calculated from the pressure data to determine the discharge coefficient of the flowmeter.

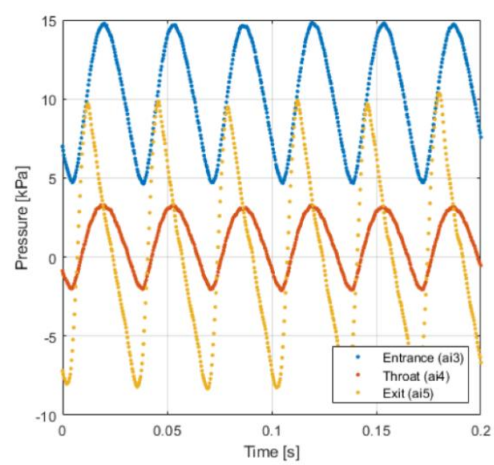
V. Results and Discussion

A. Data Processing

Figure 4 shows the raw data collected from the flowmeter pressure transducers at flow rates of approximately 1 and 2 L/min. Although a wider range of flow rates have been tested (0.5 – 2.5 L/min), only two bulk set points are being presented here for better comparison purposes. Since the flow delivered by the peristaltic pump is directly proportional to the pumping frequency, the pressure transducers sampling rate was set at 3 kHz, ensuring sufficient temporal resolution at all range of flow rates. This can be seen in Figure 4(a) and Figure 4(b), where the pump frequency at a flow rate of ~2 L/min is twice the frequency related to a flow rate of ~1 L/min. Figure 5 shows the difference between the inlet and throat pressure signal readings varying in a time interval of 0.2 s. Figure 6 presents the flow rate calculated by solving the unsteady Bernoulli equation as described in section 3. In addition, the average flow measured from the mass scale is represented by a solid line. It can also be noted that the model's solution method converges quickly after the initial estimate of zero.



(a)



(b)

Figure 4. Pressure transducer data collected using the peristaltic pump flowrate at (a) 1 L/min and (b) 2 L/min.

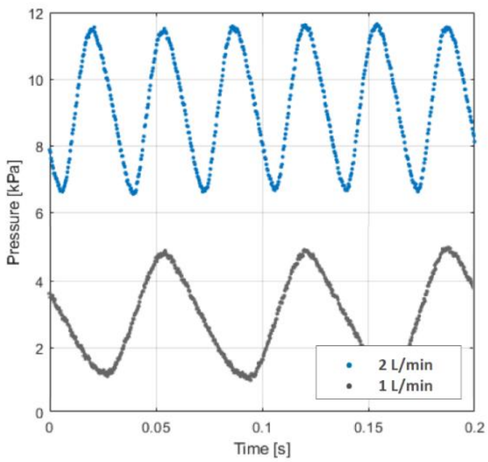


Figure 5. Difference in pressure recorded across the flowmeter using the peristaltic pump flowrate at approximately 1 L/min and 2 L/min.

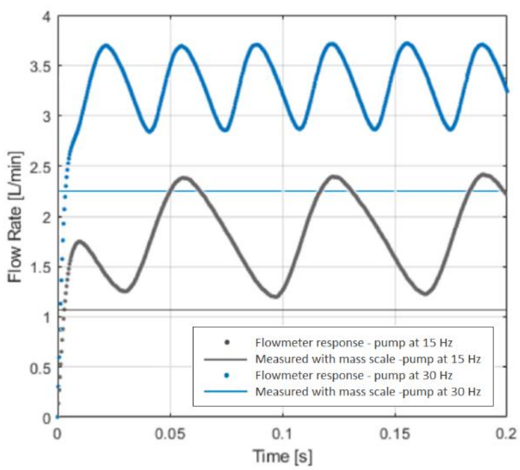


Figure 6. Calculated flowrate from the flowmeter using the peristaltic pump flowrate at approximately 1 L/min and 2 L/min.

B. Damped vs Undamped Cases

In this test scenario, the effect of the pulsation dampener on the flowmeter performance was evaluated. Following the same procedure discussed above, five different flow rates were tested ranging from 0.5 L/min to 2.5 L/min in 0.5 L/min steps. Figure 7 shows the average flow derived from the pressure data compared to the average flow measured with the mass scale. It is observed that both cases have a linear trend with the same gradient. That is, in both tests the flowmeter showed the same discharge coefficient. However, it can also be seen that both data sets are horizontally offset from each other. The case in which the pulsation dampener was present is in better agreement with the theory, since its origin is closer to the origin point of the coordinate system. This difference can be attributed to the unsteady effects of the flow in the surroundings of the flowmeter, which appear to be more prominent in the case tested without the pulsation dampener.

In the single test performed with the VAD at a nominal flow rate of approximately 2.25 L/min, the result showed agreement with the data from the tests carried out with the peristaltic pump and pulsation damper installed. Given that the VAD works at a pumping frequency of approximately 1 Hz—an order of magnitude lower than the peristaltic pump (15 - 30 Hz), such agreement between the two tests indicates that since the low pumping frequency of the VAD also induces less flow instabilities around the flowmeter. The test result with the VAD is also plotted in Figure 7.

To assess the consistency of the results obtained, three similar flowmeters (named A, B and C) were tested under the same experimental conditions, i.e., with the peristaltic pump and pulsation dampener in place. Figure 8 shows the comparative results of the average flow derived from the pressure data in relation to the average flow measured with the mass scale from the three flowmeters A, B and C. As can be seen, the three devices presented the same discharge coefficients, $C_d = 0.7119$.

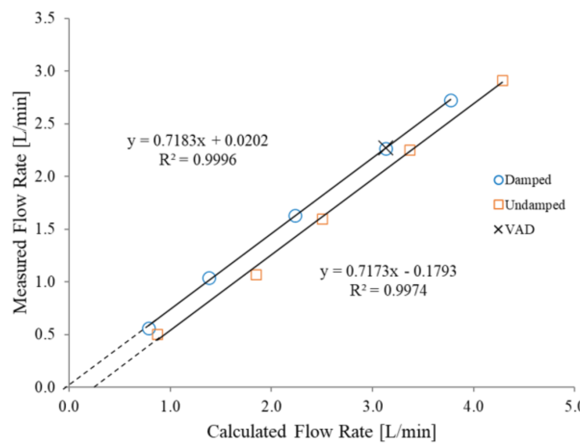


Figure 7. Calculated flowrate from the flowmeter *vs* the measured flowrate from the mass scale with and without a pulsation damper in series.

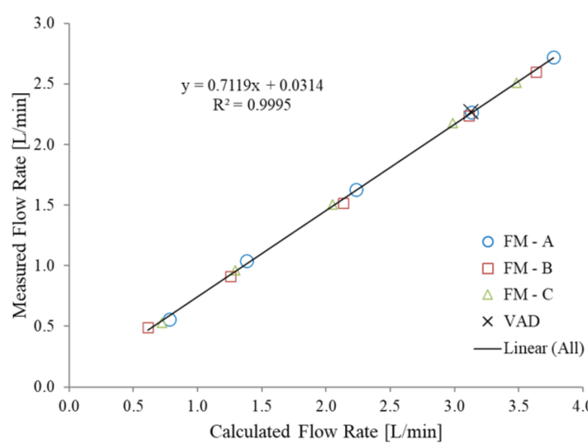


Figure 8. Calculated flowrate from the flowmeter *vs* the measured flowrate from the mass scale for three similar prototypes.

Table 1 shows the error between the flow rate obtained from the flowmeter already corrected with the discharge coefficient and the flow rate obtained with the mass scale. The results show that at low flow rates (0.5 L/min) the errors vary from 2 to 5%, while at higher flow rates, from 1.5 L/min, the errors decrease and reach values smaller than 1%, indicating an optimal operating range.

Table 1. Percent errors of the flowmeter prototypes A, B and C by using a discharge coefficient of $C_d = 0.7119$

Flowmeter	Pump Flow Rate [L/min]	Error (%)
A	0.5	5.08
	1	2.18
	1.5	0.57
	2	0.22
	2.5	0.30
B	0.5	4.96
	1	1.42
	1.5	2.15
	2	0.56
	2.5	0.66
C	0.5	2.49
	1	1.42
	1.5	0.99
	2	0.92
	2.5	0.02

VI. CONCLUSION & RECOMMENDATION

A new design of an orifice flowmeter and a method to obtain flow rate from differential pressure with an unsteady Bernoulli equation has been presented. The device was tested with water under a pulsatile flow regime at frequencies ranging from 15 to 30 Hz (peristaltic pump) and at 1 Hz (VAD), varying flow rate from 0.5 L/min to 2.5 L/min. The discharge coefficient was determined from a correlation between the flow rate measured with mass scale and the flow rate inferred from the flowmeter, showing good agreement between the three evaluated prototypes. The flowmeter was also tested with the VAD to verify its performance under a physiological-like flow. The result obtained was consistent with those of the setup containing peristaltic pump and pulsation damper. The measurement errors of the flow meters were within 5%, prevailing mainly around 1%.

Future considerations include a design study on the flowmeter's intake length. It is hypothesized that the longer entrance length can reduce the unsteady effects inherent in pulsatile flows leading to a better outcome. By the same token, the use of a flow straightener is an alternative that is worth investigating. Further tests with a blood-like solution, i.e., a non-Newtonian shear thinning fluid, will be carried out for a proper assessment of the applicability of the flowmeter.

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