Design of a novel pump for bio-applications

Sheng-Li Chang\textsuperscript{a}, Kuo-Chi Chiu\textsuperscript{a,1}, Fu-Yuan Hsu\textsuperscript{b,2}, Jhewn-Kuang Chen\textsuperscript{c}
\textsuperscript{a} Industrial Technology Research Institute, Electronics and Optoelectronics Research Laboratories, 195, Section 4, Chung Hsing Road, Hsinchu, Taiwan (R.O.C.)
\textsuperscript{b} Department of Materials Science and Engineering, National United University, No.2, Lien-Da, Miaoli City, Miaoli County 36063, Taiwan (R.O.C.)
\textsuperscript{c} Institute of Materials Science and Engineering, National Taipei University of Technology, 1, Section 3, Zhong-Xiao East Road, Taipei 10608, Taiwan (R.O.C.)

ABSTRACT

Fluid driven devices have been widely used in many applications, such as pumping, circulating, and cooling systems in handling liquid. Their driving conditions are highly dependent on the operation purposes. Some of them work with high pressure and high flow rate without the need of flow stability. On the other hand, the steady flow with low pressure and flow rate is required for bio-applications. In a perfusion system for culturing cells, a suitable shear stress from a cultivated fluid is one of key factors to reproduce the fluid conditions of cells in a living organism. A special pump is needed to provide a steady flow rate and stress in such system. In this study, a novel design of the pump constituted by a housing and a screw-type rotor with micro-channels was proposed. To understand the flow phenomena in this design, both computational modeling and real experiment are utilized. In the experiment, a minimum rotational speed is needed to drive the fluid flow. In the modeling, the steady state with low pulsation was achieved within a short period of time. A perfusion system with 7.8\% variation in flow rate could be obtained in comparison with traditional peristaltic pump with up to 29\% variation in flow rate. Steady fluid flow for a perfusion system then could be obtained in this screw-type pump.

Keywords: fluid flow, steady flow, perfusion system, cell culturing

1 INTRODUCTION

In tradition, cells are cultured in a culture dish which provides a proper liquid nutrient environment. In the culture dish, the liquid medium is still and unable to flow. There is no flow shear stress induced. However, the flow shear stress is the main factor to stimulate cells to have some physiological functions. Using this concept, a three-dimensional circulating perfusion system was developed to replicate the similar environmental conditions of the human body \cite{1-3}. This system supplies cells not only nutrition but also a suitable flow shear stress.

Figure 1 shows the perfusion system for creating an environment, which is an incubator having conditions of 37\(^\circ\)C and 5\% of CO\(_2\), similar to the living body; the fluid medium and nutrition are pumped from the reservoir through Tube 1 by the driving device. After the driving device, the compressed fluid flows into the culture device through Tube 2. This fluid medium provides the nutrition for cells and moves away the waste material discharged from it. The medium flows back to the reservoir through Tube 3 to complete the circulation in the system.

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1 KuoChi@itri.org.tw; phone: +886-3-5912639; fax:+886-3-5829781
2 Willyhsu@nuu.edu.tw; phone:+886-37-382243; fax:+886-37-382247
Figure 1. The layout of a perfusion system simulating cell culturing conditions in the living body.

For example, the fragile liver cell is not easy to be cultured in the perfusion system. For cultivating this cell to survive and having its physiological response, the same condition in human body with a low fluid pressure (i.e., less than 15 psi), a low volumetric flow rate (i.e., \(10^{10} \sim 100 \mu\text{l/min}\)) and a steady shear stress is needed in the system. If the circulated fluid medium were unsteady, a big change of fluid shear stress would be induced in the cell culture device. A high stress flow pushes the cell to crack and deform its shape, as a result it would die. Therefore a unique pumping device is required to control the flow continuously and steady.

Many pump devices such as gear pump, cam pump, peristaltic pump, and syringe pump, are the compressed type. For instance, a peristaltic pump (see Fig. 2a), containing a series of rollers to squeeze the fluid forward within a plastic tube, is widely used in bio-applications. Figure 2b displays the volumetric flow rate as function of arbitrary time frame in a peristaltic pump. The shear stress derived from the large fluctuation of flow rate with variations as large as \(\pm 29\%\) can damage cultured cells or flush them away in the cultured chamber.

In order to obtain steady condition of flow rate, two or more multiple sets of pumps are assembled [4-5]. However, the multiple pumps are expensive and not easy to reduce its size. Therefore, the purpose of this study is to design a compact pumping device to drive the flow medium which will provide the steady condition suitable for the perfusion system.

Figure 2. (a) A setup of a peristaltic pump. (b) The history log of flow rate in the pump. (The unit of abscissa is arbitrary time while ordinate is \(\mu\text{l/min}\)).

2 METHOD

Figure 3 shows the novel pump design, including a cover, a rotor, a housing and a seal, in this study. A hydrophobic material is selected for the housing, and a hydrophilic rotor is used. This is to reduce the resistance on the housing surface during the rotor’s rotation. Also, it is good for the liquid medium to be sucked into the gap between the rotor and the housing by capillary forces derived from these two surface conditions.
In the rotor, there are three sections as shown in Fig. 3. In Section 1, an inlet is located in the region of the large cavity to avoid the fluid directly impacting on the rotor. In Section 2, it includes a screw shape of micro-channel. In this channel between the rotor and the housing, the micro-fluidic flow is transformed. In addition to the capillary forces obtained from two different surface conditions, the rotational forces by rotating the rotor drives the fluid forward. In Section 3, an outlet is located next to the cavity formed from the gap in the range of 100 μm ~ 3 mm. In this cavity, there is a sufficient space for collecting the fluid from the micro-channel and keeps flowing fluid during the rotation.

![Image of the micro-pump](image-url)

Figure 3. (a) Exploded and (b) assembled drawings of the micro-pump.

In this study, both the computational modeling and the actual experiment were conducted. In modeling, the detail of flow behavior in the pump could be observed. Also, the improvement of the pump design could be suggested by the result of the modeling. In the experiment, the real condition of operating the pump was tested. Further modification for the pump could be found in this experiment.

### 2.1 EXPERIMENT

The experimental setup for this novel pump design is illustrated in Fig. 4. To rotate the rotor, a motor is connected to the rotor by a coupling. Various rotating speeds (i.e., revolutions per minute, rpm) of the rotor therefore could be adjusted by the controller. Water in the tank was conducted into the inlet of the pump by the 1/16 inch outer diameter, 1 mm inner diameter plastic tube. Another tube with the same dimensions is connected to the end of the outlet for gauging the water coming out of the pump.

![Image of the experimental setup](image-url)

Figure 4. The layout of the experimental setup for the novel pump test.
2.2 COMPUTATIONAL MODELING

Two geometries of the pump were modeled. One is the original design as shown in Fig. 5(a) and the other is the improved design using a different baffle geometry next to the outlet cavity as shown in Fig. 5(b). A computational fluid dynamics (CFD) code, Flow-3D™, has been used for these investigations. Since the flow phenomena in micro-channel during the rotation of the rotor were mainly considered here, one fluid (i.e. liquid medium) with moving obstacle (i.e., the rotating rotor) algorithm was employed [6].

Various rotating speeds of the rotor along a single axis were tested. The flux of the flow in the inlet and the outlet were monitored by invisible sectional surfaces which did not interact with the fluid flow. Newtonian fluid viscosity with turbulence model of two-equation (k-ε) model was applied. The boundary conditions of the inlet and outlet in the pump are the continuous boundary. Cubic grids with the size of 0.2mm×0.2mm×0.2mm are meshed. The total numbers of grids are around 2,187,000.

3 RESULTS AND DISCUSSIONS

Figure 6 is the experiential results of the water droplet coming out from the plastic tube in the outlet at the rotating speed of 1000 rpm. This result demonstrates the feasibility of such novel pump design characterized by changing the surface properties of the rotating rotor and the housing. However, the minimum rotating speed of 650 rpm is necessary to drive the water out of the pump.

In this pump, the energy losses mainly obtained from a high friction between the seal and the rotating shaft (Fig. 3). Also, in the pump there are some empty regions in the beginning of the rotation. Additional revolutions are needed to pump the air out and to form a sufficient vacuum state to start the outlet flow. Therefore, the cavity of the pump and the tube should be filled with fluids prior to pumping.

Figure 5. (a) The original and (b) new designs of outlet geometries for the pump.

Figure 6. The water droplet flowing out from the outlet tube at different time.
Figure 7 shows the modeling result of the volumetric flow rate at the inlet and outlet regions of the pump as function of time in the original pump design. At rotation speed of 64 rpm, the flow is steady after 0.15s. At higher speed of 162 rpm, the time in the steady state is around 0.32s, which is nearly double the time than that at 64 rpm. The averaged flow rates for these two rotation speeds are listed in Table 1.

Moreover, the velocity contour of the flow in the cross sectional plane at the outlet region is shown in Fig. 8. As shown in this figure, the velocity magnitudes of the flow in the outlet tube are various from time to time. It implies that the fluctuation speed happens in the beginning of the rotation (Fig. 7). That is because the flow transforming from unsteady to steady states needs some time. The higher the rotation speed, the longer time is required to stabilize the flow.

In Fig. 8, Vortex flows found next to the outlet tube. It is believed that the vortex results in the energy losses of the pump during rotation. Furthermore, this could develop more variation of the flow rates in the outlet area. Therefore, some modification had been done in the new geometry design in the outlet area (Fig. 5b). Figure 9 shows the modeling result of the new design. Vortex is greatly reduced at the outlet tube and the variation of the flow velocity is also reduced. Figure 10 demonstrates that the variation of flow rates in the new design is much smaller than that in the original design. Furthermore, the outlet flow rate increases from 562.5 to 658.5 μl/min by 17% using the new design (Table 1).

Table 1. The averaged flow rate for two designs of the pump at varied rotation speeds.

<table>
<thead>
<tr>
<th>Rotation speed (rpm)</th>
<th>The averaged flow rate (μl/min)</th>
<th>Changes of the flow rate (μl/min) (= outlet-inlet)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Inlet (variation)</td>
<td>Outlet (variation)</td>
</tr>
<tr>
<td>64 (original design)</td>
<td>529.8 (±65)</td>
<td>562.5 (±87)</td>
</tr>
<tr>
<td>162 (original design)</td>
<td>1427.7 (±370)</td>
<td>1483.6 (±320)</td>
</tr>
<tr>
<td>64 (new design)</td>
<td>536.1 (±32)</td>
<td>658.5 (±52)</td>
</tr>
</tbody>
</table>

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Figure 8. The magnitude and vector of the velocity contour of the flow at the cross sectional plane next to the outlet of the pump at various time frames in the original design (Fig. 5a).

Figure 9. The magnitude and vector of the velocity contour of the flow at the cross sectional plane next to the outlet of the pump at various time frames in the new design (Fig. 5b).

Figure 10. History data of the volumetric flow rate for the inlet and outlet tubes at various rotation speeds of 64 rpm in two different designs.
The time for the flow transforming from unsteady to steady states in the new design is around 0.15s, which is similar to that in the original design. Thus, this transformation is dependent on the rotation speed. As shown in Table 1, the averaged flow rate in the new design is much bigger than that in the original one. In new design, the change of the flow rate between the outlet and the inlet tubes is 122.4 μl/min, which is approximately 4 times the value (i.e., 32.7 μl/min) in the original design at the rotation speed of 64 rpm. This means that the pump can operate at a lower rotation speed to generate the same amount outlet flow.

Moreover, in the new design, the variation of averaged flow rate is reduced by a half from 15.5% in the original design to 7.8% in the new design indicating that very small fluctuation is observed in these pumps. This 7.8% variation in the flow rate is also much smaller than 29% in peristaltic pump (Fig. 2b). It means that the energy losses in the fluids of the newly designed pump are much smaller and the flow is much steadier. The geometry could be optimized further for the future design of the pump.

4 CONCLUSIONS

1. In the experiment, a minimum rotation speed (i.e., 650 rpm) is required to drive the water coming out the outlet tube due to the large friction between the seal and the rotating shaft.
2. It is demonstrated that the outlet geometry modifications for this novel pump design is capable of reducing energy losses of the fluids and thus stabilizing the flow-rate variations.
3. In the modeling for the optimized design, the pulse variation percentage of the flow rate of as small as 7.8% is achieved in comparison with 29% variation observed in peristaltic pump.
4. The variation of the flow rate increases with increasing rotation speed. And, the time for the flow to stabilize also increases with rotation speed in the pump.

REFERENCES