Centrifugal Force Based Magnetic Micro-Pump Driven by Rotating Magnetic Fields

S H Kim1, S Hashi1 and K Ishiyama1
1 Research Institute of Electrical Communication, Tohoku University, 2-1-1 Katahira, Aoba-ku, Sendai 980-8577, Japan
E-mail: kshoon@riec.tohoku.ac.jp

Abstract. This paper presents a centrifugal force based magnetic micro-pump for the pumping of blood. Most blood pumps are driven by an electrical motor with wired control. To develop a wireless and battery-free blood pump, the proposed pump is controlled by external rotating magnetic fields with a synchronized impeller. Synchronization occurs because the rotor is divided into multi-stage impeller parts and NdFeB permanent magnet. Finally, liquid is discharged by the centrifugal force of multi-stage impeller. The proposed pump length is 30 mm long and 19 mm in diameter which much smaller than currently pumps; however, its pumping ability satisfies the requirement for a blood pump. The maximum pressure is 120 mmHg and the maximum flow rate is 5000ml/min at 100 Hz. The advantage of the proposed pump is that the general mechanical problems of a normal blood pump are eliminated by the proposed driving mechanism.

1. Introduction

Numerous studies have been conducted on micro-pumps for various applications. Moreover, the developments in materials and micro/nano fabrication techniques have led to the development of various types of micro-pumps. However, most of the micro-pumps are controlled by wires or batteries. Recently, magnetic micro-pumps with external magnetic field controls have been introduced in order to overcome these problems [1]. Micro-pumps have been receiving attention because of their important role in drug delivery and uTAS [2-3]. But, macro size pumps have been used as blood pumps. Pumps for medical applications can be classified into three types according to their actuating mechanisms: centrifugal pumps, axial pumps, and vibrating flow pumps. Moreover, blood pumps can be classified into two types: pulsating flow pumps and continuous flow pumps (rotary type). In general, the pulsating flow pumps are valve type, expensive, very heavy, low efficiency, complex controls, high power consumption, and low productivity. To solve these problems, continuous flow pumps have been developed for blood pumps [4-6]. Their features are very simple structure, small size, easy control, low power consumption and high productivity. In this paper, we focus on blood pumps used in the medical field. The main features of proposed pump system are that it is wireless, battery-free, and valve-less. It generates no heat and delivers continuous flow by centrifugal force and rotating magnetic fields. General centrifugal pumps consist of a shaft, a bearing, and an electrical motor. However, the proposed pump system does not require these mechanical components.
2. Principle of operating mechanism

The magnetic micro-pump is driven by a rotating magnetic field. The original energy source is magnetic torque. An NdFeB permanent magnet on the rotor is synchronized with a rotating magnetic field. The rotating speed is dependent on the operating frequency. Figure 1 illustrates the principles of the rotating magnetic field and synchronized state within the rotating magnetic field. To generate uniform rotating magnetic field, the angle of cross-point between coil 1 and coil 2 is 90° and the difference of phase of input current signal is 90°. In this state, the rotating magnetic field is produced as a sum of the vector as shown in Fig. 1 (c) and (d). Also, the magnetic torque between the rotating magnetic field and the magnetic moment on the NdFeB permanent magnet can be express as

$$ T = MH \sin \theta \quad [\text{Nm}] $$

(1)

where $M$ is the magnetic moment of the magnet, $H$ is the rotating magnetic field, and $\theta$ is the angle between m and H.

3. Basic theory of centrifugal pump and impeller

A centrifugal pump is analyzed by the angular momentum theory and the principle of moment of momentum. The centrifugal pump’s main idea is the energy conversion of kinetic energy to pressure energy. The amount of energy given to the liquid is proportional to the velocity at the edge or blade tip on the impeller. Figure 2 illustrates velocity at the edge or blade on the single impeller. Characteristics of the impeller or pump are dependent on the blade shape which is decided by blade-angle $\beta$. According to angular momentum theory, torque and power can be express as

$$ T_{\text{torque}} = \rho Q (r_i v_i \cos \alpha_i - r_j v_j \cos \alpha_j) \quad (2) $$

$$ P_{\text{power}} = T_{\text{torque}} \times \omega = \rho g Q H_p $$

(3)

Where $\rho$ is fluid density, $Q$ is volume of flowing liquid, $H_p$ is pump head, $g$ is gravity, and $\omega$ is angular velocity. To obtain real pump head, we assume the angle ($\alpha_i = 90^\circ$). In this case, the pump head ($H_p$) can be express as

$$ H_p = \frac{1}{g} u_2 v_2 \cos \alpha_2 $$

(4)
To analyze influence of the blade-angle ($\beta$), equation 4 is converted in term of $\beta$, then, $\alpha$ is converted by $v_{z_0}$, as shown in Figure 2. The pump head ($H_p$) can be rewritten as

$$H_p = \sqrt{\frac{1}{g}(u_z^2 - u_z v_{z_0} \cot \beta)}$$

The pump head is dependent on the angle ($\beta$) according to three kinds of conditions (at constant rpm).

1. $\beta > 90^\circ$: $\cot \beta < 0$ and $v_{z_0} \cot \beta < 0$, Head is increased according to decrease of flow rate.
2. $\beta = 90^\circ$: $\cot \beta = 0$ and $v_{z_0} \cot \beta = 0$, Head is a constant value without relation to flow rate.
3. $\beta < 90^\circ$: $\cot \beta > 0$ and $v_{z_0} \cot \beta > 0$, Head is decreased according to increase of flow rate.

Figure 2. Velocity diagram between input and output of impeller: $w$ is the relative velocity of fluid particle, $v$ is the absolute velocity of fluid particle, $u$ is the peripheral velocity, $r$ is the radius, $\alpha$ is the angle between $u$ and $v$, and $\beta$ is the angle of the blade.

4. Pump design and basic property

To realize a wireless pump and to overcome various problems, the proposed pump consists of a multi-stage impeller and an NdFeB permanent magnet (diameter: 18.8 mm and thickness: 4 mm). The pump does not require a shaft and bearing because its impeller is of a floating type. Therefore, general mechanical problems are overcome. The magnetic pump has various advantages for medical application. It is wireless, battery-free, of simple structure with no mechanical problems, and it generates no heat. The basic properties of the pump are dependent on the magnetic field and operating frequency. The cross-point of two coils and the magnetic field density decide the distance between the pump and the driving coil. Then, the frequency can control the discharge pressure. In the case of a vibrating flow pump, it is dependent on the resonant frequency. However, a centrifugal pump is proportional to the operating frequency up to saturating point. The proposed pump is possibly bi-directional in rotation according to rotating direction (counter-clockwise or clockwise) of the rotating magnetic field. In this case, flow rate and pressure is decided by equation (5). Fabricated blade-angle ($\beta$) on the impeller is less than 90° when the rotating direction is counter-clockwise. But when the rotating direction is clockwise, blade-angel ($\beta$) is more than 90°. Figure 3 shows a 3D model, fabricated impeller, and pump case. Each measurement of the single impeller is 1mm and the gap between the rotor and inner wall of the pump case is 0.2 mm. This size is an important factor in order to decide the starting torque because this rotor is of a floating type. The output port diameter decides flow rate and dynamic pressure. For example, a small diameter can increase the dynamic pressure but flow rate is decreased at a constant rpm. Figure 4 illustrates changes of pressure by discharge part on the pump case. Through the flow simulation, we can expect optimized output port size and design.
Figure 3. Design and fabrication of magnetic micro-pump: (a) 3D model of impeller and blades: it produces pressure energy and (b) configuration of the rotor (multi-stage impeller with disc type NdFeB permanent magnet): rotation of the rotor brings about the centrifugal force in the pump, and (c) fully implemented magnetic micro-pump

Figure 4. Flow dynamics simulation of output port: (a) pressure distribution of diameter of discharge part (6 mm and 3 mm) and (b) results of pressures (dynamic, static, and total: discharge part 6mm)

5. Experimental results

We constructed a magnetic driving centrifugal pump system. Its configuration consists of a pump, driving coil pair, and power supply. As mentioned before, the distance between the pump and the driving coil is decided by the angle of cross-point. In this paper, all experiments are carried out at 90°. Also, the difference of phase of two current signals is fixed at 90°. To drive the magnetic rotor, the operating frequency is 10 Hz up to 100 Hz (rpm: up to 6000 rpm). In this experiment, we used two kinds of pump case (output diameters of $\phi$ 3 mm and $\phi$ 6 mm) and tubes each of $\phi$ 6 mm, $\phi$ 8 mm, and $\phi$ 10 mm. The magnet properties (size, magnetic moment) on the rotor are an important factor within the rotating magnetic field because it produces torque. Figure 5 illustrates these properties. First, the relation between flow rate and pressure is in inverse proportion. Second, the flow rate and pressure is proportional to the operating frequency (rpm). However, impedance on a coil is dependent on the frequency. After all, increasing the operating frequency causes a decrease in the driving current. Finally, output port size decides flow rate and pressure with the operating frequency. We compared discharge parts ($\phi$ 3 mm and $\phi$ 6 mm) for flow rate and pressure. In this case, the difference in pressure is 200 Pa and flow rate is 500 ml/min at 70 Hz when the diameter of the output tube is $\phi$ 6 mm and $\phi$ 3mm as shown in figure 5 (b). However, a discharge part of $\phi$ 6 mm and an output tube of $\phi$ 10 mm produces a high flow rate (3200 ml/min at 70 Hz and 4800 ml/min at 100 Hz), but variation of the pressure is a maximum of 400 Pa at 70 Hz as shown in figure 5 (b) and (c).
Figure 5. (a) Test-bed for circulation (b) the comparison of discharge parts $\phi$ 3mm and $\phi$ 6 mm, (c) the pump performance of discharge part $\phi$ 6 mm and output tube $\phi$ 10 mm: this result satisfies with the general blood pump: the red box is operating zone of the general artificial heart pump and the blue box is operating zone of the proposed pump, and (d) the change of flow rate at increasing frequency.

7. Conclusion

This proposed pump system was developed by mathematical analysis, flow simulation and experimentation. The developed pump is very small size but it produced flow rate and pressure sufficient to apply to a blood pump. Moreover, it provides a wireless and battery-free function which has important advantage for medical application. However, an innovated coil system is required to generate constant magnetic field at the fixed condition because heating problem on the coil decrease the magnetic field.

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