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Enhancement of Hydrodynamic Properties of Blood Pump Using Multiple Pumps Based on Synchronous Magnetic Radial Coupling

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Multiple pump systems are generally able to enhance pressure and flow rates through parallel and serial connections. However, the use of multiple pumps has various drawbacks, including their high power consumption, complex configuration, and large volumes. These problems make these systems unsuitable as blood pumps. To avoid these issues, we propose a new method of multiple-pump system for use with VADs (Ventricular Assist Devices) and BiVADs (Bi Ventricular Assist Devices). In particular, we focus on enhanced hydrodynamic performance through the use of a magnetic pump. The system employs synchronous magnetic radial coupling, thereby controlling up to four pumps through a single pump driver. Through serial and parallel configurations, the multiple pumps are able to extend the operating range of pressure and flow rate for use in blood pumps. The enhanced hydrodynamic performance was verified through various experimental analyses.

Keywords : magnetic multiple pumps, artificial heart assist devices, magnetic radial coupling, wireless control

1. Introduction

The properties required of an artificial heart-assist blood pumps are that it should be sufficiently hydrodynamic, of small size, simple in configuration, biocompatible, durable, and safe to use [1-3]. Blood pumps are classified according to their mechanical mechanism. Pumps are either pulsatile or continuous flow, and the latter are generally either centrifugal pump, or axial pumps [4-8]. A magnetic wireless pump is driven by magnetic torque and force, using a coil system and an external magnet with an electrical motor. We previously developed a magnetic wireless ventricular assist device (VAD) based on a magnetic wireless pump [9, 10]. One advantage of the magnetic wireless pump is that a multiple pump system can be achieved by using synchronous magnetic radial coupling, following the principle of magnetic gears. In this study, we introduce a system that utilizes multiple pumps. Typically, the purpose of the multiple-pump system is to increase flow rate and pressure by making use of serial or parallel connections. However, if normal pumps are used for a multiple-pump system, the total configuration becomes too complex and the scale of the system becomes too large due to the requirements of the individual pumps and their control systems. To avoid these issues, we propose a new system for a blood pump, which uses a multiple-pump mechanism based on magnetic radial coupling. This achieves a simple configuration with an enhanced hydrodynamic performance. The proposed method provides various advantages, such as low power consumption, small volume, wireless operation, and a multiple pump operation. We verified the performance advantages through experimental analysis.

2. Principle of Magnetic Multiple Pump and Configuration

Figure 1 shows a synchronous magnetic radial coupling for driving multiple (two) pumps using a single driver. Through the magnetic radial coupling, the rotational speed of the magnetic impellers is synchronized by the magnetic driver (driving magnet). When the driving magnet rotates clockwise (CW), the two magnetic impellers rotate counter-clockwise (CCW) due to magnetic attraction and torque. In the proposed method, a magnetic pump can operate a maximum of four pumps while using

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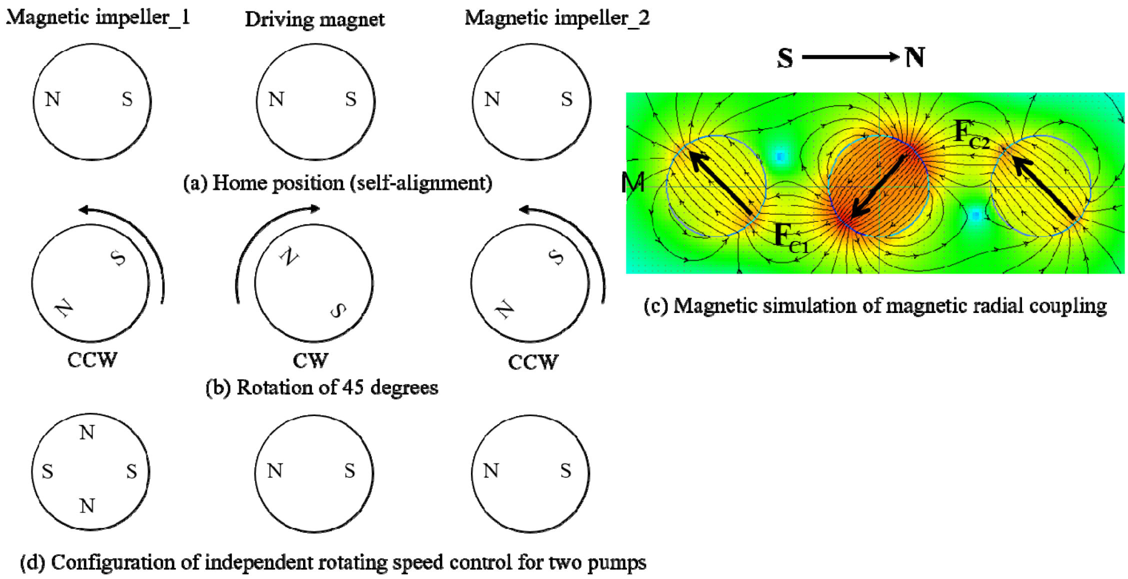

Fig. 1. (Color online) Principle of multiple pumps using synchronous magnetic radial coupling to drive multiple pumps: (a) Magnetic radial coupling (self-alignment) between the driving magnet and the magnetic impellers; (b) Control of the magnetic impellers using the driving magnet: rotation angle is 45 degrees (each magnet has two-poles); (c) Simulation of magnetic radial coupling: black arrow denotes the direction of magnetization. FC1 and FC2 are the magnetic coupling forces between the driving magnet and the magnetic impellers; and (d) Principle of different rotation speeds of multiple magnetic impellers using a single driver.

a single driving magnet. The radial coupling distance determines the driving torque and the step-out point (driving range). In our study, we conducted magnetic simulation with experiments and used magnetic analysis to design a multiple pump system.

Figure 1(a) shows the driving magnet and the FMI (fully magnetic impeller) with the same number of poles. In this case, the magnetic impeller_1 and the impeller_2 both generate the same rotational speed because of the synchronous magnetic radial coupling between the driving magnet and these two magnetic impellers. The magnetic impeller was fabricated by injection molding method using 50 wt% Sr-ferrite and NdFeB magnetic powders with a polyamide type A (PA12) resin. The sample had a magnetic moment of 4.812×10^{-3} Am^2 (4.812 emu) and a coercive force of 0.521×10^6 A/m. Figure 1(d) utilizes a two-pole driving magnet and a differing number of poles in the two FMIs: FMI_1 has four poles and FMI_2 has two poles. In this configuration, the rotational speed of impeller_1 is half that of impeller_2. This principle is the equivalent of a magnetic gear. The synchronous speed is expressed as follows:

$$V_{synch} = f \times (2/p) \times 60 \tag{1}$$

Where, f is the frequency and p is the number of poles. Therefore, the rotational speeds of multiple pumps can be controlled independently by changing the number of poles in the magnetic impeller. Using the control method, the magnetic coupling force is written as [11]:

$$\begin{aligned} \boldsymbol{F} &= \sum_{V} \boldsymbol{F}(\boldsymbol{x}, \boldsymbol{y}, \boldsymbol{z}) = \sum_{V} \nabla[\boldsymbol{m}\mu_0(\boldsymbol{x}, \boldsymbol{y}, \boldsymbol{z}) \cdot \boldsymbol{H}_{ext}] \\ &= \int_V \nabla[\boldsymbol{M}\mu_0(\boldsymbol{x}, \boldsymbol{y}, \boldsymbol{z}) \cdot \boldsymbol{H}_{ext}]\boldsymbol{dV} \end{aligned} \tag{2}$$

where μ_0 is the permeability of free space, $\boldsymbol{V}$ is the volume of the magnetic impeller, $\boldsymbol{M}$ is the magnetization of the impeller, $\boldsymbol{H}_{est}$ is magnetic field strength, and $\boldsymbol{m} = \boldsymbol{MdV}$. If we consider the position of the magnetic impellers with the driving magnet on the x-axis, the resulting force is described by:

$$\boldsymbol{dF} = \nabla(\mu_0 \boldsymbol{m} \cdot \boldsymbol{H}_{ext}) = \partial \frac{\mu_0 \boldsymbol{m}_x \boldsymbol{H}_{ext\_x}}{\partial \boldsymbol{x}} = \mu_0 \boldsymbol{M} \mathrm{d}A \mathrm{d}x \frac{\partial \boldsymbol{H}_{ext\_x}}{\partial \boldsymbol{x}}$$

$$\boldsymbol{F}_x = \mu_0 \boldsymbol{M} \int \mathrm{d}A \frac{\partial \boldsymbol{H}_{ext\_x}}{\partial \boldsymbol{x}} \mathrm{d}x \tag{3}$$

where $\boldsymbol{H}_{est\_x}$ is the magnetic field strength in X direction, $\boldsymbol{d}$ is the thickness of the magnet, $\boldsymbol{dF}$ is the infinitesimal magnetic force between the two magnets, A is the polar surface, and $\boldsymbol{dAdx}$ is the infinitesimal volume of the magnetic impeller and the driving magnet.

3. Experimental Results and Discussion

Figure 2 shows our previous wireless magnetic pump

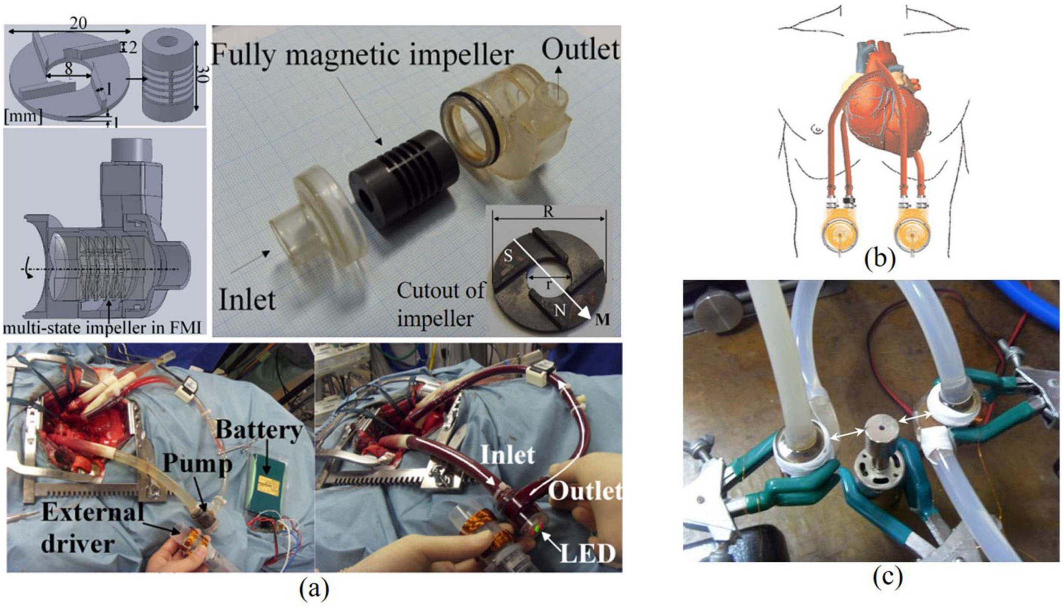

Fig. 2. (Color online) (a) The structure of a single pump in a previous study for wireless VADs using a magnetic pump based on a single pump; (b) A typical configuration of BiVADs; (c) The configuration of a magnetic multiple (two) pump operation: the pump system will be applied to a BiVAD system. The white arrows denote the coupling distance between the driving magnet and the magnetic impellers.

for VAD (ventricular assist device) and the proposed multiple-pump for the BiVAD system. The magnetic pump system consists of an FMI [12] and a control magnet with a motor. The magnetic pump utilizes a hydrodynamic bearing, thereby avoiding heating and abrasion. The driving magnet is a cylindrical NdFeB, and the FMIs are made of ferrite compounds (HM-1219) using an injection molding method. The coercive force of the FMI is 183 kA/m. A single pump has a total volume of 20 cc and weighs 52 g. The diameters of the inlet and outlet are 13 mm and 9.5 mm, respectively. The length of the pump housing is 35 mm. The fabricated magnetic impeller includes backward curvature blades on the multistage impeller and its VAD application is as shown in Fig. 2(a). The impeller structure generates a centrifugal flow. In the total configuration, the distance between the FMI and the driving magnet is a maximum of 40 mm. Figure 2(b) shows general configuration of BiVAD system. The BiVAD system generally requires two independent pumps and drivers, whereas the proposed method cam use just one motor to drive two pumps via magnetic radial coupling for the driving two pumps, as shown in Fig. 2(c). The magnetic pump is designed to consist of two FMIs and a single driving magnet (cylindrical NdFeB permanent magnet) with a DC motor. In general, RVDA and LVAD require each different hydrodynamic performance. To satisfy this requirement, the proposed method meets the requirements of hydrodynamic performance of both the RVAD and LVAD by adjusting the rotational speed of the two pump using different number of poles of the two magnetic impellers.

Figure 3 shows the hydrodynamic properties of the normal multiple pump configurations. An in-serial connection of the two pumps produces double the pump head at a constant fluid flow, whereas a parallel connection of the two pumps produces double the fluid flow at the fixed pump head, as shown in Fig. 3. In a typical multiple

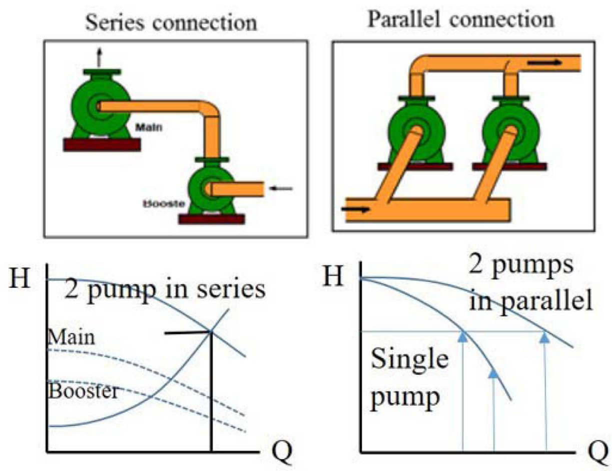

Fig. 3. (Color online) General multiple pump connections (with series and parallel connections) and their hydrodynamic properties: H is the pump head and Q is the flow rate.

pump system, each pump requires an independent motor. Thus, a typical multiple pump configuration is not suitable for VADs because of its complex configuration, large volume, and high power consumption. In contrast, a system employing a magnetic multiple pump has a simple configuration. The pump uses a single driver to act on two magnetic impellers via magnetic radial coupling. Thus, the pump achieves a smaller size and lower power consumption. In particular, in order to increase the flow rate and the blood pressure, the rotational speed of the impeller must be increased. However, the rotational speed of the increased impeller can cause damage to the blood cell. Therefore, the proposed method can improve the hydrodynamic performance without increasing the rotational speed. With a wireless magnetic operation, we observed the magnetic coupling force varying according to changes in the distance between the driving magnet and the magnetic impeller, as shown in Fig. 4(a). In order to measure the magnetic coupling force and the magnetic flux density, we used a force gauge and a gauss meter (DS2 digital force gauge-IMADA and BELL 5150-F.W BELL). Greater distances between the driving magnet and the magnetic impeller resulted in a weaker attractive force. The attractive force corresponded to variations in driving torque, as shown in Fig. 4(b). To measure the driving torque, we installed two pick up coils to observe the phase differences at the pump and at the driving magnet.

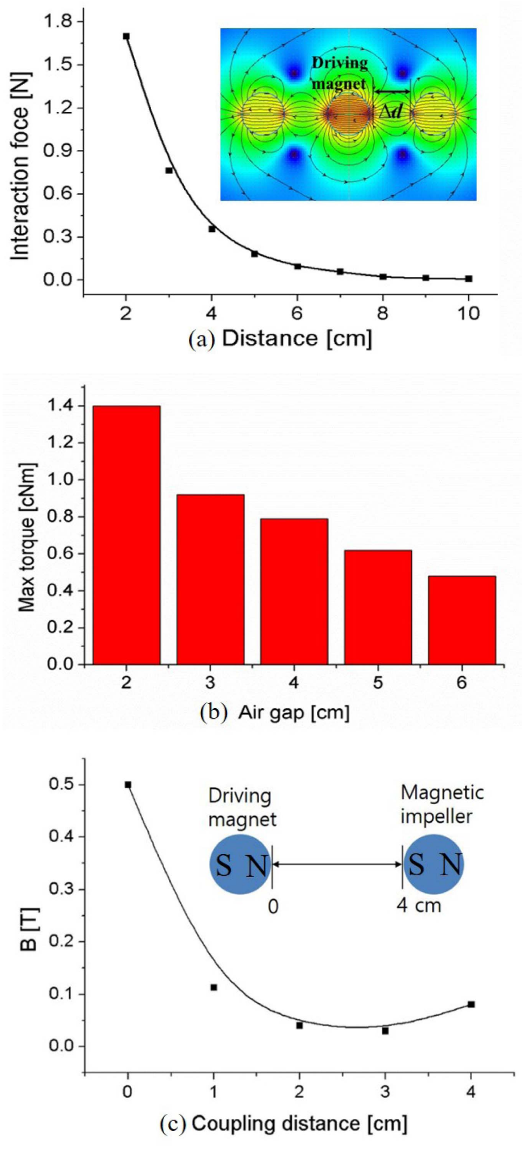

Fig. 4. (Color online) (a) Multiple pump connections (serial and parallel connections) and their hydrodynamic properties; (b) The driving torque according to variations in the coupling distance; (c) The maximum magnetic flux density between the magnetic impeller and the control magnet according to variations in the radial coupling air-gap.

Using these phase differences, we estimated driving torque because driving torque corresponds to magnetic torque [13]. The measured phase difference was converted to an angle between the external magnetic field and a direction of magnetic moment in magnetic torque equation. A distance of 2 cm between the driving magnet and the magnetic impeller generated an attractive force of 1.68 N and a torque of 1.38 cNm. Thus, the distance can determine the rotational speed of the pump and its power consumption. The distance of 2 cm achieved a maximum rotation of 6800 rpm, compared with 1500 rpm at 10 cm distance. Figure 4(c) shows the measurement for magnetic flux density between the FMI and the control magnet vary according to changes in the coupling distance. In this result, the magnetic flux density of the magnetic impeller is important because the impeller is installed in the human body. At a coupling air-gap of 2 cm, the FMI produced a maximum flux density of 0.04 T. The driving magnet and the magnetic impeller produced maximum flux densities of 0.5 T and 0.08 T at 0 cm and 4 cm, respectively. The flux density will not influence the heart or other organs because the driving magnet (the strong field) is located outside of the body. Figure 5 shows a hydrodynamic in vitro performance test and the power consumption of both serial and parallel configurations using the magnetic VADs. We observed the flow rate and the pump head of a water at the outlet. The flow rate was observed by a flowmeter (FDM-5L-KEYENCE), and the output pump head was measured by a pressure-meter (KDM30-Krone) according to changes in the rotating speed of the impeller.

In the case of the parallel connection of two magnetic VADs, the system produced 11.9 kPa and 36 kPa, at rotational speeds of 3000 and 4800 rpm, respectively. At equivalent rotational speeds, the single pump produced

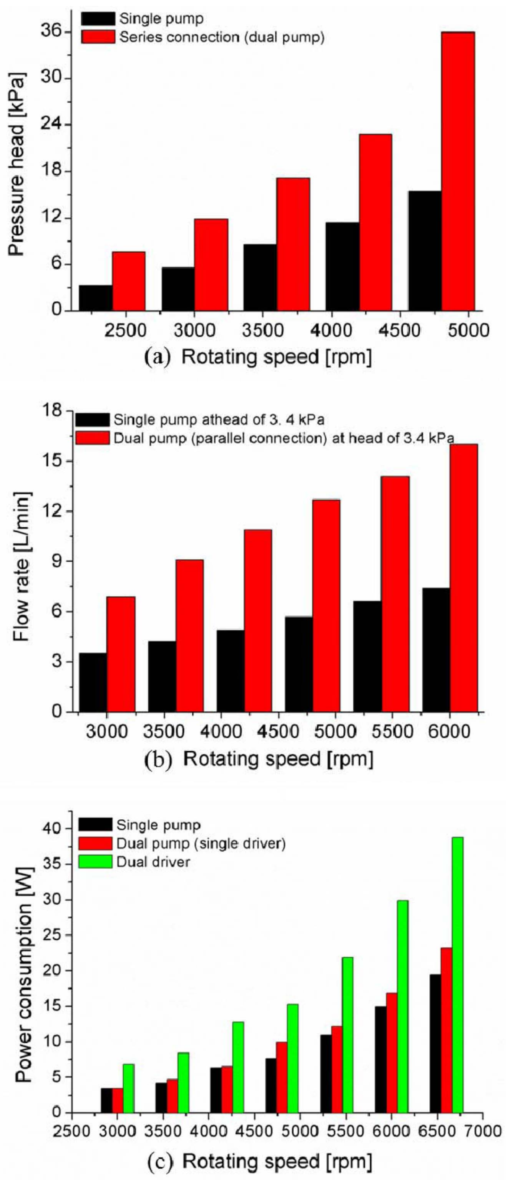

Fig. 5. (Color online) Results of experiment and proposed method: (a) comparison of pump head pressure using a single pump and a dual pump with parallel connection; (b) Comparison of flow rate between a single pump and a dual pump with serial connection (c) electrical power consumption: a single pump with a dual driver, a dual pumps with a single driver, and a dual pump with a dual drivers.

only 5.6 kPa and 15.5 kPa as shown in Fig. 5(a). Figure 5(b) shows the flow rate of a serial connection at the fixed pump head of 3.4 kPa. In this experiment, we compared the flow rates of a single pump and a dual pump with a parallel connection. The single pump generated 3.5 L/min and 5.7 L/min at 3000 rpm and 4800 rpm, respectively, whereas the parallel connection produced 6.9 L/min and 12.7 L/min, respectively. Through these experiments, we were able to verify that the synchronous magnetic radial coupling achieved lower power consumption, as shown in Fig. 5(c). When using two drivers to control two-pumps, the two drivers together consumed 6.8 and 15.3 W at 3000 rpm and 4800 rpm, respectively; whereas the proposed method, utilizing a single driver for a two-pump operation, consumed only 3.4 and 9.97 W at the equivalent revolutions. It thus appears that the proposed pump system can produce a wide hydrodynamic range of pressure and flow rate which maintaining low power consumption. These properties are all necessary for VADs.

Although we did not conduct an animal test with BiVAD in this study, we did conduct an animal test using a single pump for blood circulation. In the animal test, the single pump was applied to the LVAD (Left Ventricular Assist Device) and the RVAD (Right Ventricular Assist Device), respectively, as a preliminary validation for the later application of BiVAD. In these experiments, we confirmed the hydrodynamic performance of both applications. In particular, we are considering pediatric use of the VAD system using the magnetic pump because of the small volume of the pump. Typically, a pediatric blood pump is required to pump an average of 1.7 to 2.0 L/min with a pressure rise of 50 to 60 mmHg. Observing these requirements, our previous study used the magnetic pump to produce a flow rate of 1.3 to 2.5 L/min at 3600 rmp. The rotating speed of the impeller determined the flow rate and the pump head. With a driving speed ranging from 1200 to 3600 rpm, the RVAD generated flow rates of 1 to 3.18 L/min, and the driving speed ranged from 4140 to 5400 rpm for the LVAD, generating flow rates of 3.94 to 5.3 L/min in the animal test using a single pump. These results satisfy the requirements of both the pediatric and adult pumps. Another factor to consider is that high-speed rotation can cause damage to blood cells. Thus, a low-speed rotation of the impeller is better than in order to reduce blood damage. Using a single magnetic pump, for 6 hours we checked the separated ratio of blood cells using the hematocrit test. During that time, the blood damage was checked every hour. The separation rate was approximately 25 %, which is categorized as normal.

Because a centrifugal flow pump produces a higher flow rate than an axial flow pump at an equivalent rotation speed, the proposed pump adopts a centrifugal flow mechanism. If a higher flow rate or a higher pressure can be achieved without the driving speed control, we can use multiple pumps (with either serial or parallel connections) with the proposed method at a fixed rotation speed of the impeller. In the case of a parallel connection, the rotation speed of 4800 rpm produced 12.7 L/min, whereas the single pump produced a maximum of 8 L/min at 4800 rpm. Using the connecting method, the proposed multiple pump mechanism will easily be able to control hydrodynamic performance. The centrifugal pump has a relatively simpler impeller mechanism than that of the axial pump [14], whereas the volume of the axial pump is more compact [15]. State-of-the-art blood pumps have now reached their third generation, and their mechanical features include continuous flow, a centrifugal design, noncontact bearing design, and hydrodynamic levitation [16]. The mechanical structure of the proposed pump is simpler than that of third generation of blood pumps, and the pump is driven wirelessly.

In order to apply the pump system to BiVAD, the RVAD (right ventricular assist device) must generate lower blood pressure and flow rate than LVAD (left ventricular assist device). Therefore, when the driving magnet has two poles, we are able to use magnetic impellers with four poles for RVAD and two poles for LVAD. The magnetic impeller with four poles generates half of hydrodynamic performances, involving blood pressure and blood flow rate. In this study, we focused on the verification of the proposed mechanism. At this stage, we did not consider various biocompatibilities, such as influence on blood cell, inflammatory response and oxidative stress organism, the surrounding tissues, and the whole body. These studies are necessary and therefore we intend to investigate these areas in our next study.

4. Conclusion

In this study, we proposed a new mechanism, involving synchronous magnetic radial coupling, for enhanced hydrodynamic performance of a blood pump. In particular, the proposed method realized low power consumption in driving the pump driving. Because multiple pumps require multiple-pump drivers, the electrical power consumption is proportional to the number of drivers. Synchronized magnetic radial coupling makes it difficult to control the rotational speed of each pump when using a single driver. In the present study, this was achieved by using magnetic impellers with differing numbers of poles. Parallel and series connections between magnetic pumps using synchronous magnetic radial coupling resulted in extended pressure and flow rate, respectively, while still maintaining low power consumption. In addition, the use of different numbers of magnetic poles allowed independent control over the hydrodynamic performance of each pump. In other words, the proposed multiple pumps can be used in both general VADs and in BiVADs, as the smallest centrifugal blood pump available. Through simulation with in-vitro testing, we were able to enhance the hydrodynamic properties. We plan to perform animal testing with the BiVAD system and will conduct a plasma free hemoglobin test using multiple pumps for a quantitative analysis. In this study, we considered the influence of magnetic flux density on the surrounding tissues and the organism. The magnetic impeller utilized produced a maximum flux density of 80.32 mT, which is less than of the magnetic manipulation system of a magnetic medical robot. Therefore, in the present study, we ignored any influence on surrounding tissues and the organism. However, our long-terms plans include a more detailed investigation of this issue. The proposed method is suitable for application to increase the hydraulic pressure and flow rate in a narrow space because it can drive four impellers with one actuator. An electric vehicle cooling system is an example.

Acknowledgments

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