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Fluid Dynamic Design for Low Hemolysis in a Hydrodynamically Levitated Centrifugal Blood Pump

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Abstract— **We have developed a hydrodynamically levitated centrifugal blood pump for extracorporeal circulatory support as a bridge to decision pump. The impeller is levitated using hydrodynamic bearings without any complicated control circuit or displacement sensor. However, the effect of the outer circumferential velocity and the bearing area on the hemolytic property has not been clarified, even if the bearing gap is same size. The purpose of this study is to experimentally evaluate the effect of the outer circumferential velocity and the bearing area in the bearing gaps on the hemolytic property in a hydrodynamically levitated centrifugal blood pump. We prepared three models for testing. These models have the same bearing gap size by adjusting the impeller levitation position. However, the outer circumferential velocity of the impeller and the bearing area in the minimum bearing gaps are different. The outer circumferential velocity of the impeller and the bearing area were assumed to be related to the maximum shear rate and the exposure time. For the evaluation, we conducted an impeller levitation performance test and an in vitro hemolysis test. As a result, the normalized index of hemolysis (NIH) was reduced from 0.084 g/100L to 0.040 g/100L corresponding to a reduction in the outer circumferential velocity and a reduction in the bearing area, even if the minimum bearing gaps were same size. We confirmed that, even if the bearing gap was same size under the stably levitated condition, the outer circumferential velocity and the bearing area should be decreased in order to improve the hemolytic property.**

I. Introduction

Heart transplantation has been performed for serious heart failure patients as an effective treatment. However, the number of heart donors is quite limited all over the world. Therefore, an implantable left ventricular assist device (LVAD) has been developed and applied to serious heart failure patients as a bridge to transplantation [1]. However, it is difficult to apply an LVAD immediately to heart failure patients who suffer from multi-organ failure or are of unclear neurological status. At present, an extracorporeal blood pump as a bridge to decision pump is required for patients with serious heart failure. A bridge to decision pump is used before deciding whether an LVAD should be used or whether a heart transplantation should be undertaken [2].

We have developed a hydrodynamically levitated centrifugal blood pump for extracorporeal circulatory support as a bridge to decision pump. This pump uses a hydrodynamic bearing as a non-contact bearing with a long-term durability. The mechanism of the hydrodynamic bearing is that grooves on the bearing surface generate localized pressure by blood flowing into the narrow bearing gap, and the hydrodynamic bearing supports a rotating impeller by acting as a nonlinear spring. However, since the bearing stiffness of the hydrodynamic bearing is small, the bearing gap tends to become narrow, and high hemolysis has been observed. Therefore, in order to obtain the suitable bearing gap size for the hydrodynamic bearing, the relationship between the bearing gap and hemolytic property has been investigated in some studies [3-4]. However, the effect of the outer circumferential velocity and the bearing area on the hemolytic property has not been clarified, even if the bearing gap is same size.

The purpose of this study is to experimentally evaluate the effect of the outer circumferential velocity and the bearing area in the bearing gaps on the hemolytic property in a hydrodynamically levitated centrifugal blood pump.

II. Materials and Methods

A. Hydrodynamically levitated centrifugal blood pump

We have developed a hydrodynamically levitated centrifugal blood pump for extracorporeal circulatory support as a bridge to decision pump, as shown in Fig. 1. The developed pump is composed of three parts: a top housing, a bottom housing and an impeller. The diameter of the pump head is 72.2 mm, and its height is 19.1 mm. The diameter of the impeller is 57.7 mm, and its height is 57.7 mm. The impeller is levitated by using two thrust bearings with spiral groove and a radial bearing with herringbone groove. The developed pump uses the motor unit for the commercial extracorporeal centrifugal blood pump, HCF-MP23 (Senko Medical Instrument Mfg. Co., Ltd., Tokyo, Japan). The permanent magnets in the motor are mounted on the motor shaft, and arranged inner side of the pump head. The impeller is rotated by means of the magnet coupling between the permanent magnets in the impeller and the permanent magnets in the motor. This pump achieves a pressure of 200 mmHg and a flow rate of 4.0 L/min at approximately 2,800 rpm.

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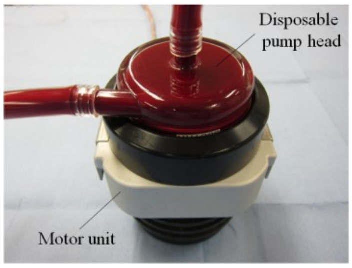
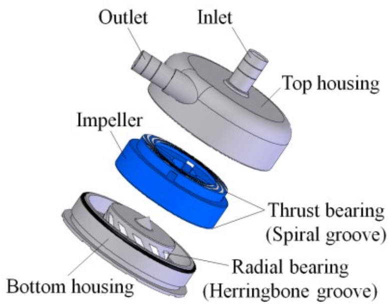
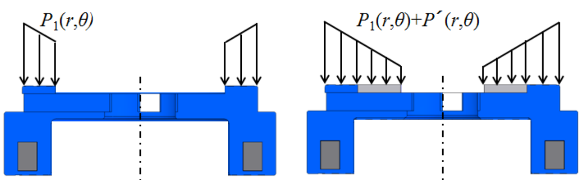


(a) Photograph of the developed hydrodynamically levitated centrifugal blood pump



(b) Components of the developed pump

Figure 1. Developed hydrodynamically levitated centrifugal blood pump.



(a) Default shroud area to generate the hydrodynamic force acting on the top surface of the impeller

(b) Shroud area added to change the force acting on the top surface of the impeller

Figure 2. Method of adjusting the impeller levitation position.

B. Principle of impeller position control

In order to adjust the impeller levitation position in the hydrodynamically levitated centrifugal blood pump, the balance of the force acting on the impeller was controlled by changing the top shroud area of the impeller. The hydrodynamic forces acting on the top and bottom surfaces of the impeller were obtained by integrating the pressure distribution on the top and bottom surfaces of the impeller. The hydrodynamic force, F_t, acting on the top surface of the impeller is expressed as:

$$F_t = \int_0^{2\pi}\int_{r_1}^{r_2} P_1(r,\theta)\cdot drd\theta \quad (1)$$

where r_1 is the inner radius of the impeller, r_2 is the outer radius of the impeller, $P_1(r,\theta)$ is the pressure distribution on the top surface of the impeller, r is the radial direction and θ is the angular direction.

Similarly, the hydrodynamic force, F_b, acting on the bottom surface of the impeller is expressed as:

$$F_b = \int_0^{2\pi}\int_{r_1}^{r_2} P_2(r,\theta)\cdot drd\theta \quad (2)$$

where $P_2(r,\theta)$ is the pressure distribution on the bottom surface of the impeller. Then, the difference of the

(a) Top gap model (b) Outer-bottom gap model (c) Inner-bottom gap model

Figure 3. Geometries of three tested impellers, the top shroud area of which are 1,326 mm^2, 1,504 mm^2, and 1,717 mm^2.

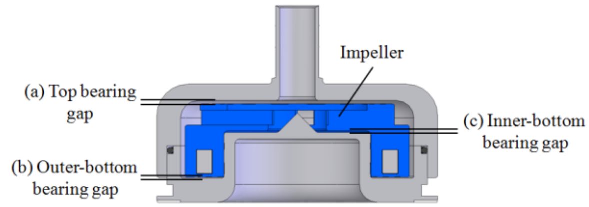


Figure 4. Positions of the bearing gaps of the tested pump.

hydrodynamic force, F_s, between F_t and F_b is obtained as follows:

$$F_s = F_t - F_b \quad (3)$$

As shown in Fig. 2, the pressure distribution on the top surface of the impeller increases from $P_1(r,\theta)$ to $P_1(r,\theta)+P'(r,\theta)$ by increasing the top shroud area of the impeller. The hydrodynamic force acting on the top surface of the impeller, which is the sum of the pressure distribution, can also be increased. Therefore, in order to adjust the impeller levitation position, the balance between the hydrodynamic forces acting on the top and bottom surface of the impeller are changed by adjusting the top shroud area of the impeller.

C. Tested pump models

Tested pump models having different top shroud areas were prepared. The top shroud geometries are shown in Fig. 3, and the positions of bearing gaps of the tested pumps are shown in Fig. 4. In the pump models, we focused on the outer circumferential velocity and the bearing area. In the top gap model, the top shroud area was set at 1,326mm^2, and the impeller levitation position was adjusted to the top gap side of the bearing. Therefore, the outer circumferential velocity of the impeller and the bearing gap area in the minimum bearing gap were 7.3 m/s and 1,326 mm^2, respectively, at the rated rotational speed. In the outer-bottom gap model, the top shroud area was set at 1,504mm^2, and the impeller levitation position was adjusted to the outer-bottom gap side of the bearing. Therefore, the outer circumferential velocity of the impeller and the bearing gap area in the minimum bearing gap were 8.5 m/s and 1,504 mm^2, respectively, at the rated rotational speed. Furthermore, the inner-bottom gap model was prepared by applying an additional spacer to the thrust bearing on the inner bottom side of the impeller. The top shroud area of this model was set at 1,717mm^2. The impeller levitation position of the inner-bottom gap model was adjusted to the inner-bottom gap side of the bearing. Therefore, the outer circumferential velocity of the impeller and the bearing gap area in the minimum bearing gap were 5.4 m/s and 943 mm^2, respectively, at the rated rotational speed.

D. Impeller levitation performance test

In order to evaluate the impeller levitation performance of the tested pump models, an impeller levitation performance test was conducted using a laser focus displacement meter (LT-8110, Keyence Corp., Osaka, Japan). The mock circulation loop is shown in Fig. 5. This mock circulation loop consisted of the tested pump model, a polyvinyl chloride blood reservoir (special order product, Senko Medical Instrument Mfg. Co., Ltd.), a polyvinyl chloride tube (MERA Exceline-H, Senko Medical Instrument Mfg. Co., Ltd.), and a resister. A glycerol aqueous solution with the same viscosity as blood at 37°C was used as a working fluid. The laser focus displacement meter was positioned at the top of the pump housing. The laser was focused at the surface of the titanic ring inside the impeller. The total thrust bearing gap of the pump models was 300 ± 2 µm. The sampling frequency was set to 1 kHz and the sampling time was set to 10 s. The impeller levitation position was measured as the rotational speed of the tested pump model was increased from 0 to 3,500 rpm. The top and bottom bearing gaps were then calculated from the measured data. This test was repeated five times for each pump model.

E. In vitro hemolysis test

In order to evaluate the hemolytic property of the tested pump models, an in vitro hemolysis test was performed using a mock circulation loop, as shown in Fig. 6. The mock circulation loop consisted of tested pump model, a polyvinyl chloride blood reservoir (special order product, Senko Medical Instrument Mfg. Co., Ltd.), a polyvinyl chloride tube(Senko Medical Instrument Mfg. Co., Ltd.), and an adjustable resister. Purchased bovine blood (Funakoshi Co., Ltd., Tokyo, Japan) was used as a working fluid. The tested pump model was operated at a pressure of 200 mmHg and a flow rate of 4.0 L/min. The operating time was 2 h. The temperature of the blood was maintained at 37°C using a water bath. A commercial extracorporeal centrifugal blood pump, BPX-80 (Medtronic Inc., Minneapolis, MN, USA) was used as a control pump. Plasma-free hemoglobin in the plasma of blood sampled from the mock circulation loop was determined using the tetramethylbenzidine method. Then, the normalized index of hemolysis (NIH) [5] was calculated from the concentration of plasma-free hemoglobin and was compared to that obtained from three tested pump models. This test was repeated three times for each pump model.

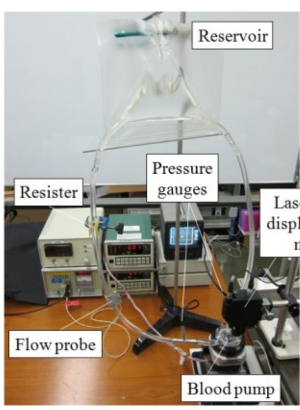


Figure 5. Photograph of impeller lavitation performance test.

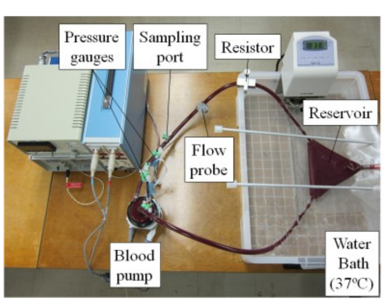


Figure 6. Photograph of the in vitro hemolysis test.

III. Results

Fig. 7 shows the results of the impeller levitation performance test. At the rated rotational speed, the impeller of the top gap model levitated by 103 ± 12 µm from the top housing. The bottom gap became 197 ± 12 µm. The impeller of the outer-bottom gap model levitated by 195 ± 7 µm from the top housing. The outer-bottom gap became 106 ± 7 µm. The impeller of the inner-bottom gap model levitated by 191 ± 13 µm from the top housing. The inner-bottom gap became 107 ± 13 µm.

Fig. 8 shows the results of the in vitro hemolysis test. The NIH value of BPX-80 was 0.036 ± 0.009 g/100L. The NIH values of the top gap model, the outer-bottom gap model, and the inner-bottom gap model were 0.040 ± 0.010 g/100L, 0.084 ± 0.027 g/100L, and 0.145 ± 0.074 g/100L, respectively. The relative NIH ratios in comparison with the NIH value of BPX-80 were 1.1, 2.3, and 4.0, respectively.

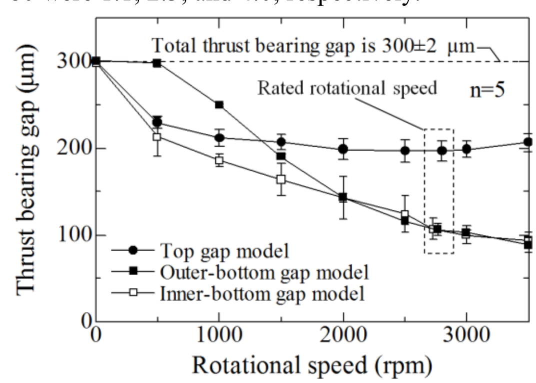


Figure 7. Results of the impeller levitation performance test.

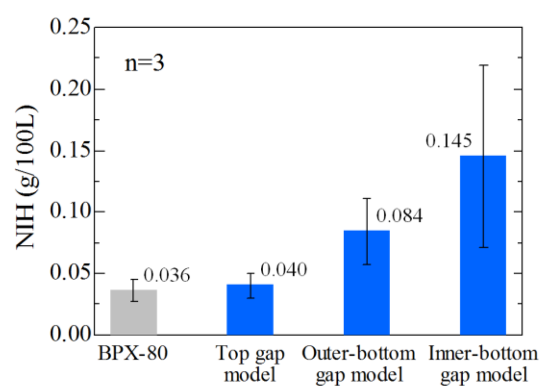


Figure 8. NIH values of tested pump models obtained by an in vitro hemolysis test at 200 mmHg and 4.0 L/min.

IV. DISCUSSION

Table 1 shows the characteristics of three tested pump models obtained in this study. The top shroud area was changed to adjust the impeller levitation position. In the impeller levitation performance test, the impeller levitation position of three tested pump models were adjusted to the top gap side, the outer-bottom gap side and the inner-bottom gap side, respectively. The minimum bearing gap of the pump models was able to be adjusted to the almost same as 100 μm.

In the tested pump models, we focused on the outer circumferential velocities and the bearing area. The circumferential velocity and the bearing area were assumed to be related to the maximum shear rate and the exposure time. In order to calculate the maximum shear rate in the minimum bearing gap, we assumed that the flow in the thrust bearing gap was a Couette flow of a Newtonian fluid. The estimated maximum shear rate in the minimum bearing gap was obtained as follows:

$$D_{\max} = \frac{\partial U_{\max}}{\partial h_{\min}} \tag{4}$$

where $D_{\max}$ is the estimated maximum shear rate in the minimum bearing gap, $U_{\max}$ is the outer circumferential velocity of the impeller in the minimum bearing gap, and $h_{\min}$ is the minimum bearing gap. In the top gap model, the estimated maximum shear rate was calculated as 71,000 s^{-1} at the outer circumference of the top bearing. In the outer-bottom gap model, the estimated maximum shear rate was calculated as 80,000 s^{-1} at the outer circumference of the outer-bottom bearing. In the inner-bottom gap model, the estimated maximum shear rate was calculated as 51,000 s^{-1} at the outer circumference of the inner-bottom bearing. Therefore, the estimated maximum shear rate of the pump models were different even if the minimum bearing gaps were same size. The bearing area in the minimum bearing gap of the top gap model, the outer-bottom gap model, and the inner-bottom gap model were 1,326 mm^2, 1,504 mm^2, and 943 mm^2, respectively. This bearing area was assumed to be related to the exposure time. Therefore, the exposure time of the tested models were considered to be different, even if the minimum bearing gaps were same size. In the in vitro hemolysis test, the NIH values of the pump models were also different, even if the minimum bearing gaps were same size. Giersiepen et al. reported that the hemolytic property is correlated with the shear stress and the exposure time [6]. From Table 1, the NIH value of the top gap model was smaller than that of the outer-bottom gap model. The estimated maximum shear rate and the bearing area of the outer-bottom gap model were larger than these of the top gap model. Therefore, the hemolysis in the outer-bottom gap model is considered to be caused by the estimated maximum shear rate and the bearing area, which is related to the exposure time. In contrast, although the inner-bottom gap model had the smallest estimated maximum shear rate and bearing area among the tested pump models, the hemolytic property was the worst. In the inner-bottom gap model, since the outer circumferential velocity of the impeller in the inner bottom bearing was small, the bearing stiffness generated by the inner-bottom bearing and the moment of force to recover the tilt of the impeller were small. Therefore, the impeller was considered to have a tendency to tilt, and the bottom gap of the thrust bearing and the radial gap of the radial bearing became locally narrow. As such, high hemolysis was caused. In the in vitro hemolysis test, since the standard deviation of the NIH value in the inner-bottom gap model was the largest among the tested pump models, the impeller of this model was considered to levitate under the unstable levitated condition.

These results demonstrate that the hemolytic property was influenced by the estimated maximum shear rate and the bearing area, which is related to the exposure time in the hydrodynamically levitated centrifugal blood pump under the stable levitated condition, even if the bearing gap is same size. Therefore, in order to improve the hemolytic property of hydrodynamically levitated centrifugal blood pump, it is important to evaluate not only the bearing gap size but also the outer circumferential velocity and the bearing area.

TABLE I. CHARACTERISTICS OF TESTED PUMP MODELS

	Top gap model	**Outer-bottom gap model**	**Inner-bottom gap model**
Top shroud area	1,326 mm^2	1,568 mm^2	1,717 mm^2
Top gap	<u>103 μm</u>	195 μm	191 μm
Bottom gap	197 μm	<u>106 μm</u>	<u>107 μm</u>
Outer circumferential velocity	7.3 m/s	8.5 m/s	5.4 m/s
Estimated maximum shear rate	71,000 s^{-1}	80,000 s^{-1}	51,000 s^{-1}
Bearing area	1,326 mm^2	1,504 mm^2	943 mm^2
NIH	0.040 g/100L	0.084 g/100L	0.145 g/100L
NIH ratio (vs BPX-80)	1.1	2.3	4.0

Underlines indicate the minimum bearing gap.

V. CONCLUSION

In this study, we confirmed that the outer circumferential velocity corresponding to the maximum shear rate and the bearing area corresponding to the exposure time should be decreased under the stably levitated condition in order to improve the hemolytic property in the hydrodynamically levitated centrifugal blood pump, even if the bearing gap was same size.

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