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Improvement of Hemocompatibility for Hydrodynamic Levitation Centrifugal Pump by Optimizing Step Bearings

Ryo Kosaka, Toru Yada, Masahiro Nishida, Osamu Maruyama, Takashi Yamane

Abstract— **We have developed a hydrodynamic levitation centrifugal blood pump with a semi-open impeller for a mechanically circulatory assist. The impeller levitated with original hydrodynamic bearings without any complicated control and sensors. However, narrow bearing gap has the potential for causing hemolysis. The purpose of the study is to investigate the geometric configuration of the hydrodynamic step bearing to minimize hemolysis by expansion of the bearing gap. Firstly, we performed the numerical analysis of the step bearing based on Reynolds equation, and measured the actual hydrodynamic force of the step bearing. Secondly, the bearing gap measurement test and the hemolysis test were performed to the blood pumps, whose step length were 0 %, 33 % and 67 % of the vane length respectively. As a result, in the numerical analysis, the hydrodynamic force was the largest, when the step bearing was around 70 %. In the actual evaluation tests, the blood pump having step 67 % obtained the maximum bearing gap, and was able to improve the hemolysis, compared to those having step 0% and 33%. We confirmed that the numerical analysis of the step bearing worked effectively, and the blood pump having step 67 % was suitable configuration to minimize hemolysis, because it realized the largest bearing gap.**

I. Introduction

An implantable left ventricular assist system (LVAS) has been developed and used for medical treatments for curing a serious heart disease patients as a bridge to heart transplantation and as a permanent implantation. The inlet cannula of the LVAS is connected to an apex of a left ventricle, and the outlet cannula of the LVAS is connected to an aorta. By delivering blood flow from the left ventricle to the aorta, the LVAS assists the diseased cardiac function of the left ventricle. Currently, as a long-term artificial heart for destination therapy, a rotary blood pump with non-contact bearings such as a hydrodynamic bearing and a magnetic bearing has been developed [1]. The mechanism of the hydrodynamic bearing is that grooves located on the bearing surface generate localized high pressure and support a rotating impeller by acting as a nonlinear spring. Compared to a conventional contact bearing, the hydrodynamic bearing has some advantages. Since the hydrodynamic bearing is non-contact bearing, the potential for wear and thrombus formation around the pivot bearing has been eliminated. And compared to a magnetic bearing, since the hydrodynamic bearing is passive bearing, the blood pump does not need to incorporate any additional displacement-sensing module and additional complex control circuits. Therefore, this results in a simple design and high reliability. However, the hydrodynamic bearing has some disadvantages, such as the destruction of red blood cells and the thrombus formation caused by high shear stress and flow stagnation in a small gap of the hydrodynamic bearing.

We previously developed the old model of the hydrodynamic levitated centrifugal blood pump. However, since the bearing gap of the blood pump was so small that high hemolysis was observed [2]. Therefore, we focused on the improvement of the impeller levitation properties to expand the hydrodynamic bearing gap.

The purpose of the study is to investigate the geometric configuration of the hydrodynamic step bearing to minimize hemolysis by expansion of bearing gap.

II. Materials and Methods

A. Hydrodynamic Levitation Centrifugal Pump

We have developed a hydrodynamically levitated centrifugal blood pump with a semi-open impeller (HH100 series). A photograph of the developed blood pump is shown in Fig. 1, and the internal structure of the pump is shown in Fig. 2. This pump is mainly composed of three parts: the top casing, the bottom casing, and the impeller. The diameter of the pump casing is 74 mm, and its height is 38 mm. The diameter of the impeller is 37 mm, and its height is 25 mm. The upper side of the impeller forms a spiral-groove bearing, and the lower side of the impeller forms the semi-open impeller. The impeller levitates by using the thrust bearings of an original spiral bearing, and the journal bearing of a herringbone bearing. The permanent magnets are arranged inside the rotor, and the rotor is rotated within a magnetic field of stator coils. The pump attains a pressure head of 100 mmHg with a flow rate of 5 L/min at 3000 rpm. Because the hydrodynamic bearing was not formed at the bottom of impeller, we applied the step bearing to the bottom of the impeller.

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Ryo Kosaka, Masahiro Nishida, Osamu Maruyama, Takashi Yamane are with the National Institute of Advanced Industrial Science and Technology (AIST), 1-2-1, Namiki, Tsukuba, Ibaraki, 305-8564, Japan (e-mail: ryo.kosaka@aist.go.jp).

Toru Yada was with the Tokyo University of Science (TUS), Noda, Chiba 278-8510, Japan

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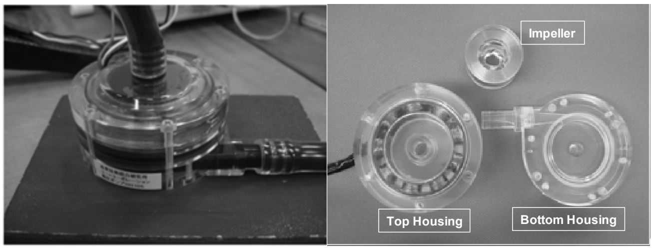

Fig. 1 Photographs of hydrodynamic levitated centrifugal blood pump, HH100

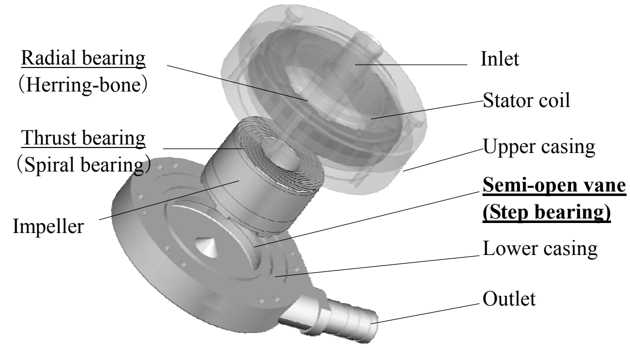

Fig. 2 Hydrodynamic levitated centrifugal blood pump, HH100

B. Numerical Analysis

In order to optimize the geometric configuration of the step bearings, the numerical analysis was performed. In this analysis, assuming that the pressure inside the bearing gap only varied in horizontal coordinate, Reynolds equation based on Lubrication theory was applied to analyze the pressure distribution in the bearing as follows;

$$\frac{\partial}{\partial r}\left(h^3 r\frac{\partial P}{\partial r}\right)+\frac{1}{r}\frac{\partial}{\partial \theta}\left(h^3\frac{\partial P}{\partial \theta}\right)=6r\omega\mu\frac{\partial h}{\partial \theta} \tag{1}$$

where h is gap, p is pressure, r is radius, θ is angle, ω is angular velocity and μ is viscosity. This equation is converted to the discrete model by using the central difference method with gap h and pressure p, as follows;

$$p_{i,j}=\frac{\frac{1}{\Delta r^2}\left\{h_E^3(r+0.5\Delta r)p_{i+1,j}+h_W^3(r-0.5\Delta r)p_{i-1,j}\right\}+\frac{1}{r\Delta\theta^2}\left\{h_N^3 p_{i,j+1}+h_S^3 p_{i,j-1}\right\}-\frac{6\mu\omega}{\Delta\theta}(h_N-h_S)}{\frac{1}{\Delta r^2}\left\{h_E^3(r+0.5\Delta r)+h_W^3(r-0.5\Delta r)\right\}+\frac{1}{r\Delta\theta^2}\left\{h_N^3+h_S^3\right\}} \tag{2}$$

where $p_{i,j}$ is surrounded by gap (h_N, h_S, h_W, h_E) and by pressure ($p_{i,j+1}$, $p_{i,j-1}$, $p_{i+1,j}$, $p_{i-1,j}$).

The analysis model of the step bearing is shown in Fig.3. The height of vane was 1.5 mm and the depth of step bearing was 100 µm. The working fluid was assumed to be blood with a specific gravity of 1.05, viscosity of 3.0 cP.

In the numerical analysis, three kinds of analyses were performed to investigate the optimal geometric configuration of the hydrodynamic step bearing. Firstly, in order to analyze the optimal step length that generates the maximum hydrodynamic force, the step length against the vane length was increased every 10 % from 0 % to 100 %, and the bearing gap was set at 50 µm, 100 µm and 150 µm. Based on the LVAS driving condition, the outer pressure was 100 mmHg, the inner pressure was 0 mmHg and the rotational speed was 3000 rpm. Secondly, the hydrodynamic force generated by the step bearing was calculated using three kinds of impellers having step length of 0 % (step 0%), 33 % (step 33%) and 67 % (step 67%) against the vane length. The outer pressure was 100 mmHg, the inner pressure was 0 mmHg and the rotational speed was 3000 rpm. Finally, to evaluate the accuracy of the numerical analysis, the calculated hydrodynamic force was compared to the measured hydrodynamic force using the three kinds of impellers having step 0 %, 33 % and 67 %.

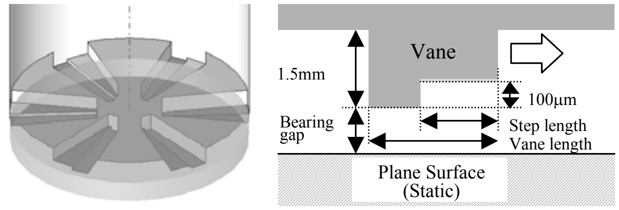

Fig. 3 Sectional view of analysis area of hydrodynamic step bearing

In the numerical analysis, the pressure distribution in the step bearing was calculated, and the hydrodynamic force that was sum of the pressure distribution was determined. Then, by subtracting the pressure distribution of step 0 %, the hydrodynamic force generated only by the step bearing was determined.

C. Measurement of hydrodynamic force in step bearings

In order to evaluate the accuracy of the numerical analysis, the actual measurement experiments of the hydrodynamic force generated by the step bearing were performed. The measurement system of the hydrodynamic force was shown in Fig. 4. The impeller was placed into a cylindrical container that filled with a 38 wt% aqueous glycerin solution (Glycerol, Wako Pure Chemical Industries, Ltd., Osaka, Japan) that had the same viscosity as 37 °C blood. The vortex baffle was placed inside the cylindrical container. The impeller was connected with a milling machine (FM100, Kotobuki Boeki co., Tokyo, Japan) using a shaft coupling, and was driven by the motor of the milling machine. The hydrodynamic force was measured by using an electronic balance (GX-3000, A&D Company, Ltd, Tokyo, Japan). The bearing gap between the step bearing and the bottom casing was measured by using a laser focus displacement meter (LT8110, Keyence Co., Osaka, Japan) which had ±0.2 µm resolution.

In the measurement test, three kinds of impellers having step 0 %, 33 % and 67 % were applied to the experiment. The impeller models were shown in Fig.5. The hydrodynamic force was measured at 1500 rpm, because the oscillating motion was observed over 2000 rpm. The bearing gap was increased from 80 µm to 300 µm. The sampling frequency was 1 kHz and the sampling time was 20 s. After measurement, the average value of the measured data was calculated. Then, by subtracting the pressure distribution of step 0%, the hydrodynamic force generated only by the step bearing was obtained, and was compared to that obtained by the numerical analysis.

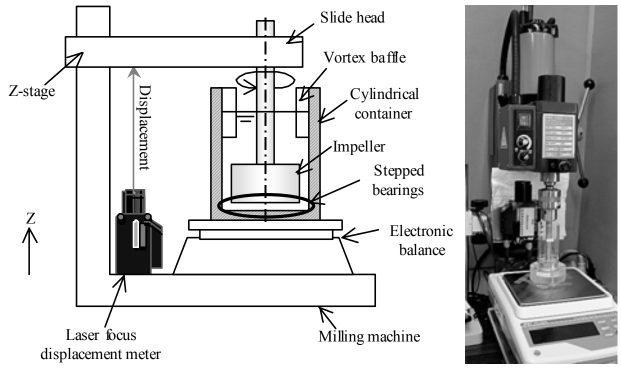

Fig. 4 Experimental system for hydrodynamic force measurement

Fig. 5 Three kinds of tested impellers whose step length was 0 %, 33 % and 67 % of the vane length

D. *Levitation performance test*

In order to evaluate the levitation performance of the blood pump with step bearing, a levitation performance test was performed using a mock circulation loop. The mock circulation loop consisted of the blood pump, an adjustable resistor and a soft reservoir bag (Special designed product, Senko Medical Instrument Mfg. Co., Ltd., Tokyo, Japan). Each component was connected with Tygon tubing (MERA Exceline-h, Senko Medical Instrument). A 38 wt% aqueous glycerin water solution was employed as the working fluid. As measured data, the bearing gap between the step bearing and the bottom casing was measured by using a laser focus displacement meter as shown in Fig. 6. The focus of the laser was set at the bottom impeller.

In the levitation performance test, three kinds of impellers having step 0%, 33% and 67% were used. The bearing gap was measured at the rotational speed of 3000 rpm. The sampling frequency was 1 kHz and the sampling time was 30 s. Then, the average value of the measured data was calculated, and was compared to that obtained by each model.

E. *In vitro hemolysis test*

In order to evaluate hemolysis performance of the blood pump with the step bearing, an in-vitro hemolysis test was performed using a mock circulation loop in accordance with ASTM [3]. The mock circulation loop consisted of the developed blood pump, an adjustable resistor and a soft reservoir bag, as shown in Fig. 7. Purchased bovine blood (Funakoshi Co., Ltd., Tokyo, Japan) containing sodium citrate solution was used as the working fluid. The driving conditions were a pressure head of 100 mmHg and a flow rate of 5 L/min. The operating time was 4 h. The temperature of the blood was maintained at 37ºC using a water bath.

In the hemolysis test, blood was sampled from the circulation loop. Plasma-free hemoglobin in the circulating blood was obtained by the tetramethylbenzidine method. A normalized index of hemolysis (NIH) was calculated from the plasma-free hemoglobin. The commercial conventional blood pump, BPX-80 (Medtronic Inc., Minneapolis, MN, USA), was used as a control pump to compare to the tested blood pump. BPX-80 has enough hemolysis property to apply the clinical use during a operation. However, BPX-80 is indicated for usage not to exceed 6 hours, because the contact-bearing with a shaft coupling was adopted.

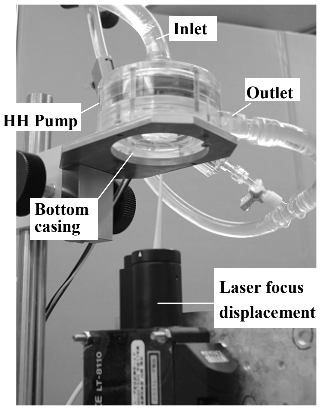

Fig. 6 Photograph of levitation performance test.

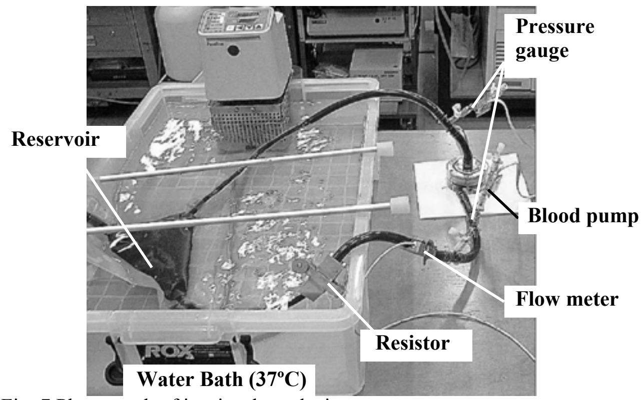

Fig. 7 Photograph of in-vitro hemolysis test

III. Results and Discussion

In the numerical analysis of the step bearing, the hydrodynamic force generated by the step bearing was calculated as shown in Figures 8, 9. Figure 8 shows the hydrodynamic force generated only by the step bearing corresponding to the step length, when the bearing gaps were set at 50μm, 100μm and 150μm. It was found that the maximum hydrodynamic force was obtained around step length of 70 %, and the hydrodynamic force was increased corresponding to the reduction of the bearing gap. Figure 9 shows the hydrodynamic force generated only by the step bearing corresponding to the bearing gap. It was found that the hydrodynamic force increased corresponding to the reduction of the bearing gap. And the hydrodynamic force of the impeller having step 67% was the largest among the tested impellers.

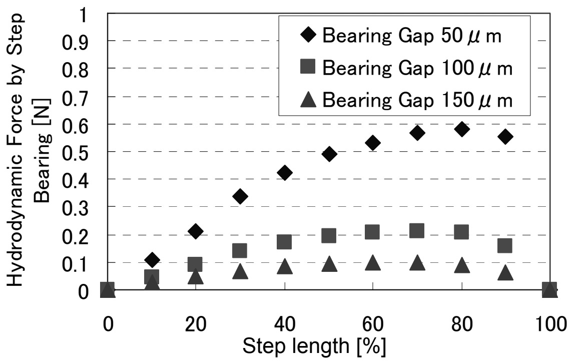

Fig. 8 Hydrodynamic force generated by step bearing in numerical analysis

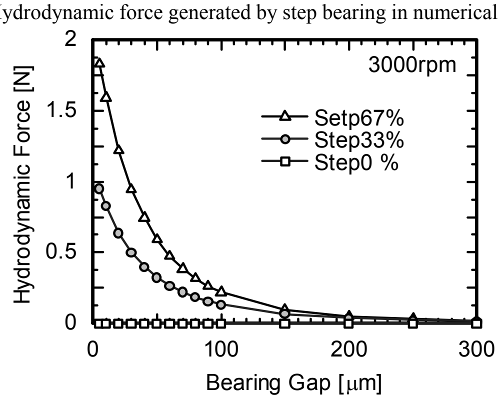

Fig. 9 Hydrodynamic force generated by step bearing in numerical analysis

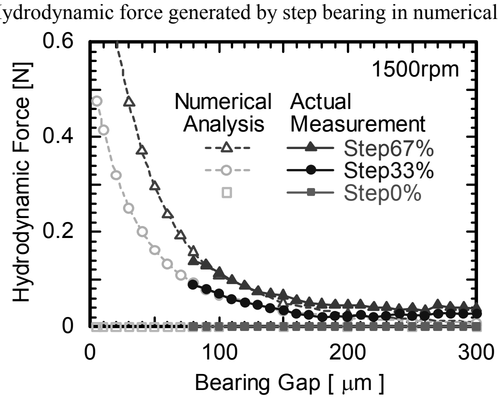

Fig. 10 Comparison between hydrodynamic force obtained by the numerical analysis and by the actual measurement

Figure 10 shows the evaluation of the accuracy of the numerical analysis. Under the gap of 80 μm, it is difficult to measure the actual hydrodynamic force, because there is a possibility that the impeller makes contact with the bottom casing. Above the gap of 80 μm, the hydrodynamic force calculated by the numerical analysis was almost equivalent to that obtained in the actual measurements. Based on Eq. 1, the hydrodynamic force of the step bearing has the relationship with the rotational speed. Therefore, this result shows high reliability of the numerical analysis based on Reynolds equation under all rotational speed.

Figure 11 shows the results of the impeller levitation test. The blood pumps having step 0 %, 33 % and 67 % had the gaps of 0.0 μm, 7.5 μm and 15.9 μm respectively. Therefore, it was found that the blood pump having step 67%, which generated the largest hydrodynamic force, had the largest bearing gap as compared with the blood pumps having step 0% and step 33%.

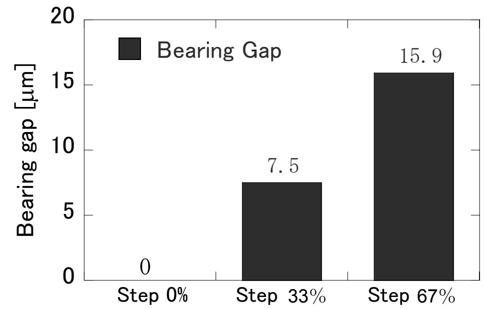

Fig. 11 Results of measurement tests of bearing gap at 100 mmHg, 5 L/min

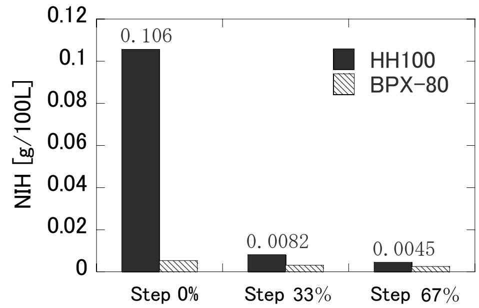

Fig. 12 Results of in-vitro hemolysis tests at 100 mmHg, 5 L/min

Figure 12 shows the results of the hemolysis test. The NIH values of the blood pumps having step 0 %, 33 % and 67 % became 0.106 g/100L, 0.0082 g/100L and 0.0045 g/100L respectively, and the relative NIH ratios in comparison with BPX-80 were 19.9, 2.7 and 1.7 respectively. Therefore, it was found that the hemolysis level was the lowest, when the blood pump having step l67% among these models.

These evaluation tests demonstrated that the hemolysis was improved corresponding to the expansion of the bearing gap by increasing the hydrodynamic force.

IV. Conclusion

We have developed the centrifugal blood pump with hydrodynamic bearings. The numerical analysis worked effectively to analyze the performance of the hydrodynamic bearings. The blood pump having the step 67 % was suitable configuration to minimize hemolysis, because it realized the largest bearing gap.

Acknowledgment

The present study was supported in part by the Industrial Technology Research Grant Program in 2009 from the New Energy and Industrial Technology Development Organization (NEDO) of Japan, and by a Grant-in-Aid for Scientific Research (KAKENHI).

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