High-sensitivity pulsation estimation and application of heartbeat synchronization control in magnetically-levitated ventricular assist device

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Abstract
In recent years, the number of heart failure patients has been increasing. However, there is a shortage of donors for treatment. As a result, ventricular assist devices (VAD) are being used as an alternative, and long-term use has become common. However, there are issues such as aortic valve dysfunction caused by VAD's continuous flow support. As a solution to this problem, heartbeat synchronization control, which adjusts the motor speed to match the pulsation of the diseased heart, is being explored. Conventional studies have used electrocardiograms and flow meters to estimate the pulsations necessary for heartbeat synchronization control. However, it is difficult for patients to wear these devices continuously. Therefore, our research group has developed a technology that estimates pulsations using only sensors from a magnetic levitation system of an impeller, eliminating the need for additional ECG sensors and applying it to heartbeat synchronization control. However, one of the issues of this method is that it was difficult to estimate weak pulsations and hence, it is required to improve the signal-to-noise (S/N) ratio. To improve the S/N ratio, we established a control system that allows for high-sensitivity pulsation estimation while maintaining levitation stability by lowering the stiffness of the support impeller for pulsations (0.7-2 Hz) and increasing the stiffness for unbalanced forces (10-50 Hz) using the loop shaping method based on H_∞ control theory. We successfully applied the designed control system to a magnetically-levitated VAD and the estimated flow's S/N ratio was doubled.

Keywords : Magnetically levitated ventricular assist device, Heartbeat synchronization control, Estimated fluid force, Loop shaping, Stiffness of magnetic bearing, H_∞ control theory.

1. Introduction

The number of cardiovascular diseases is increasing due to lifestyle-related diseases and the aging of the population (Groenewegen et al., 2020) . Although cardiovascular disease is the leading cause death since more than 40 years ago, doners are in short supply (Cameli et al., 2022). Therefore, ventricular assist devices (VADs) are used for the blood circulation. Especially, a magnetically-levitated (MagLev) VAD which can levitate an impeller becomes main stream of this field because they can be used for longer term (Tunuguntla et al., 2020). We have developed thrombus detection and prevention techniques using magnetic bearings to improve the blood compatibility of MagLev VADs (Hatakenaka et al., 2023). Another issue is dysfunction of the aortic valve due to the continuous flow generated by the VAD (Kim et al., 2024). One of the solutions for this problem is heartbeat synchronization control of a motor in VAD (Hirohashi et al., 2016) . In the previous studies, flow rate measurement for heartbeat synchronization control has been performed using electrocardiograms (Ando et al., 2011) and flow meters (Gaddun et al., 2014). However, it is not practical to always wear an electrocardiogram (Cordeiroa et al., 2020). Therefore, our research group proposed a method for estimating the heartbeat by magnetic bearings without the use of additional sensors (Tanaka et al., 2021). Figure 1 shows the configuration of the self-sensing heartbeat synchronization control. First, the fluid force acting on the impeller, is estimated by using disturbance observer. Next, the flow rate through the blood pump is estimated from the estimated fluid force (Pai et al., 2010). Furthermore, using the estimated flow rate, the cardiac function of the diseased heart is evaluated by machine learning method, and the target flow rate that the VAD aims to assist is determined. Then, by changing the motor speed using a feedback controller based on the difference between the target flow rate and the estimated flow rate, heartbeat synchronization control is achieved. This paper focuses on the fluid forces estimation.

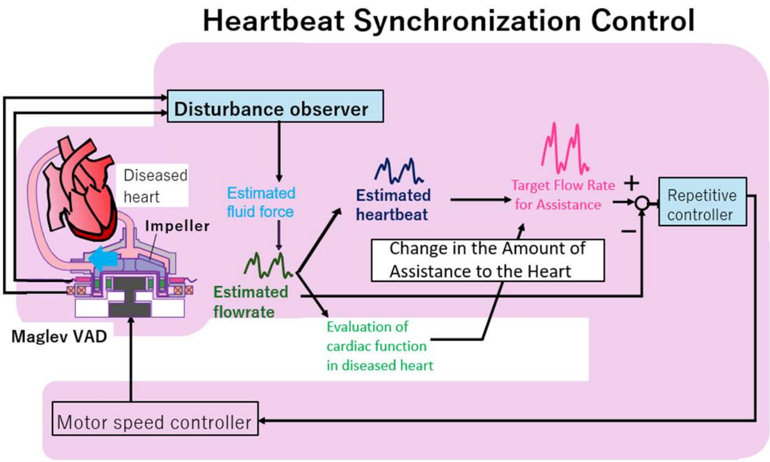


Fig. 1 Configuration of the proposed heartbeat synchronization control and pulsation estimation methods using magnetic bearing. The patient's pulsation acts on the Maglev impeller as fluid force. The fluid force is estimated by a disturbance observer installed in the control system of the magnetic bearing.

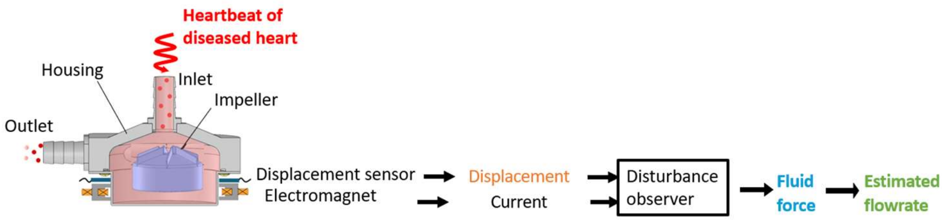


Fig. 2 The method for estimating fluid force. The fluid force is estimated by the disturbance observer using the displacement of the impeller and the current of the electromagnet for the magnetic bearing.

However, since this fluid force is small, it must be estimated with high sensitivity by improving the signal-to-noise (S/N) ratio. In order that, loop shaping by H_∞ control is used in this study.

2. Principle of Magnetic Levitation

The transfer function P_1 from the voltage applied to the electrical circuit to the current flowing through the electromagnet for the magnetic bearing is as follows. Kirchhoff's Equations for the electrical circuit is as follows:

$$\frac{I}{V} = \frac{1}{Ls + R} = P_1 \tag{1}$$

Where, V is the voltage of the electromagnet circuit, I is the current flowing through the electromagnet for the magnetic bearing, R is the total resistance of the circuit, and L is the total inductance of the electromagnet.

Following, it is important to consider the dynamics of an impeller levitated by electromagnetic force. The transfer function P_2 from the current flowing through the electromagnet for magnetic bearings to the displacement of the impeller is derived from the motion equation of the impeller as follows,

$$\frac{X}{I} = \frac{k_i}{ms^2 + cs + k_x} = P_2 \tag{2}$$

Where, x is the displacement of the impeller, m is the mass of the impeller, c is the viscous coefficient due to the influence of blood viscosity, k_x is the proportional constant for displacement, and k_i is the proportional constant for current.

3. Control system of magnetic bearing to improve S/N ratio

For highly sensitive estimation of the fluid force due to pulsation, the stiffness of the magnetic bearing should be small, and the impeller should be displaced significantly by the pulsation. On the other hand, it is desirable to maintain high stiffness against unbalance forces caused by rotation. One of the solutions is that the stiffness is designed so that it becomes low at the frequency of heartbeat component (0.7-2 Hz), and it becomes high at the frequency of unbalance force generated by the impeller rotation (10-50 Hz). The loop shaping based on H_∞ theory is adopted for compensating both stability and shaping desired stiffness characteristics. The generalized plant was designed to become a modified mixed sensitivity problem as shown in Fig. 3.

The transfer function from disturbance d_1 to control quantity z_1 is as follows:

$$\frac{z_1}{d} = \frac{P_2}{1 + P_1 P_2 K} W_1 = P_2 S W_1 \tag{3}$$

Where, S is the sensitivity function. This is an indicator of the suppression performance against disturbances. The smaller the magnitude of S, the greater the stiffness. In this study, to reduce stiffness in the frequency range of 0.7-2 Hz, a weighting function W_1 is designed to increase the magnitude of S in that frequency range. Therefore, W_1 is designed to have a peak in the frequency range of 0.7-2 Hz.

The transfer function from disturbance d_1 to control quantity z_2 is as follows:

$$\frac{z_2}{d} = \frac{P_1 P_2 K}{1 + P_1 P_2 K} W_2 = T W_2 \tag{4}$$

Where, T is the complementary sensitivity function. This is an indicator of noise removal performance. The smaller the magnitude of T, the better the noise removal performance. Therefore, T should be reduced in the high-frequency range by increasing the weighting function W_2. The constant weight ϵ of the direct term d_2 was introduced to satisfy the assumptions of the standard H_∞ problem. In this study, the value of ϵ was set to 7.0×10^{-7}.

The formulas for the designed weighting functions are given by Eqs. (5) and (6) respectively, and Fig. 4 shows these functions.

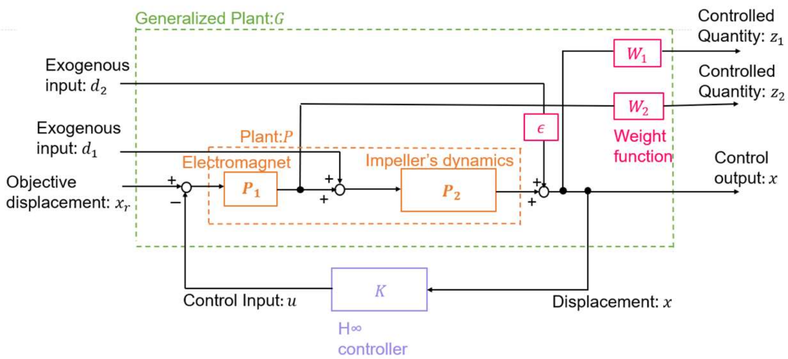


Fig. 3 The design is intended to address the modified mixed sensitivity problem. By appropriately designing the weighting functions W_1 and W_2, the objectives of this study can be achieved. The constant weight ϵ is used to satisfy the assumptions of standard H_∞ control theory.

$$W_1 = \frac{0.037 \times (s^2 + (2\pi \cdot 1.2)s/2 + (2\pi \cdot 1.2)^2)}{(\,s/(2\pi \cdot 0.3) + 1)(\,s/(2\pi \cdot 4) + 1)^2} \tag{5}$$

$$W_2 = \frac{0.5s^2}{s^2 + 50s + 100} \tag{6}$$

Figure 5 shows the stiffness characteristics in the conventional controller (pole placement controller) and designed controller using loop shaping. The shaped stiffness characteristics reduce in the pulsation component band (0.7-2 Hz) compared to the conventional ones. Conversely, in the unbalance component band (10-50 Hz), the stiffness characteristics are increased compared to conventional ones.

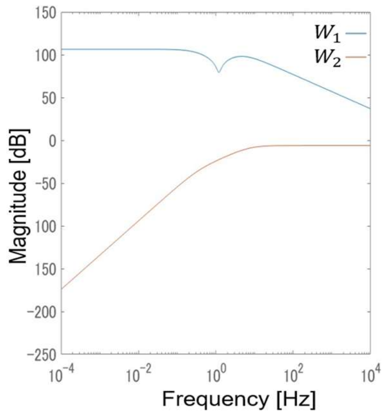


Fig. 4 Designed weighting function. The design of the weighting function W_1 related to disturbance suppression is a key factor. To reduce stiffness against disturbances in the 0.7-2 Hz range, the sensitivity function is increased in that frequency range. To achieve this, W_1 is designed to decrease in that frequency range. The weighting function W_2, which relates to noise removal, is increased in the high-frequency range to enhance its performance.

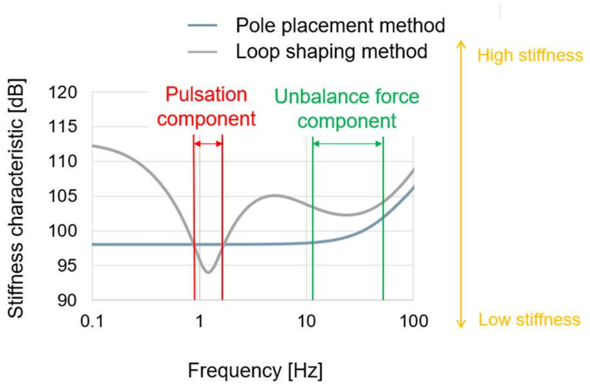
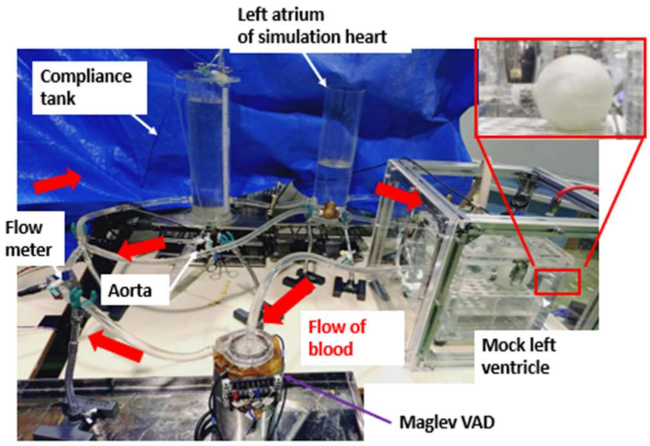


Fig. 5 Loop shaping of stiffness characteristics, which is the frequency response of the disturbances force to impeller's displacement. By shaping the stiffness characteristics to down peak in the frequency range of 0.7-2 Hz (pulse component), it is possible to reduce stiffness compared to conventional control systems. Additionally, in the frequency range of 10-50 Hz (unbalance force component), the stiffness characteristics are increased to enhance it compared to conventional control systems.

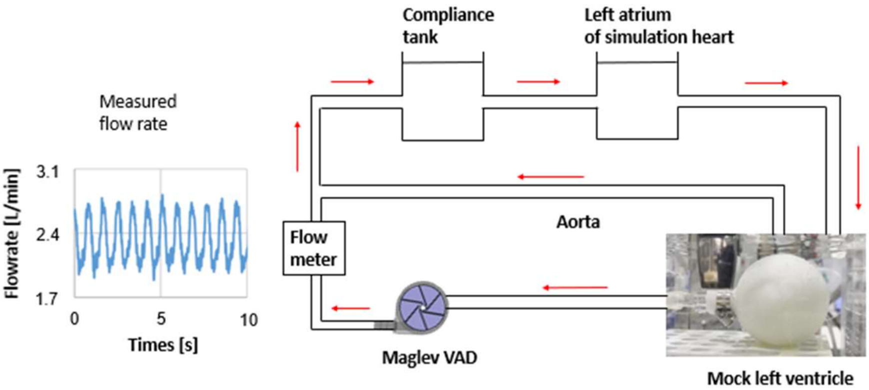
4. Experiment of fluid force estimation

4.1 Equipment of experiment

The experimental setup composed of a mock circulatory loop and Maglev VAD is shown in Fig. 6. The mock left ventricle is made of silicone and mimics the volume (107 mL) and myocardial thickness (10 mm) of a human left ventricle (Clay et al., 2016). Mock left ventricle made of silicone with the same Young's modulus and thickness as human myocardium is driven by the compressed air. With this ventricle, the pulsation was applied to the Maglev impeller. In this experiment, the fluid forces acting on the impeller were estimated and S/N ration was evaluated. The flow rate flowing through the VAD was measured with a flow meter. The average flow rate was 2.4 L/min, and the amplitude was 0.8 L/min.



(a) The photo of the mock circulation loop used in the experiment.



(b) The schematic diagram of the mock circulation loop used in the experiment.

Fig. 6 (a) Configuration of the experiment equipment mock circulation loop. The adjustment of the pulsation magnitude is performed using a vacuum regulator (for air absorption) and an electric air regulator (for air injection). (b) The mock circulation loop used in the experiment is shown in the schematic diagram.

4.2 Results of frequency analysis of impeller displacement

Figure 7 shows the results of frequency analysis of the impeller displacement when a weak pulsation with a heart rate of 70 bpm is applied, comparing the conventional control system (pole placement controller) with the loop shaping method in this study.

Through loop shaping method, it was confirmed that the pulsation component increased by 6.2 times compared to the conventional control system, resulting in lower stiffness. On the other hand, the unbalance force component was confirmed to decrease to 0.75 times, resulting in higher stiffness. Therefore, based on this result, the support stiffness of the impeller decreases in response to pulsation, while it can be improved in relation to unbalanced forces. In other words, the control system design aligned with the objectives of this study can be achieved using the loop shaping method. In this study, components other than pulsation, including unbalanced force components, are defined as noise.

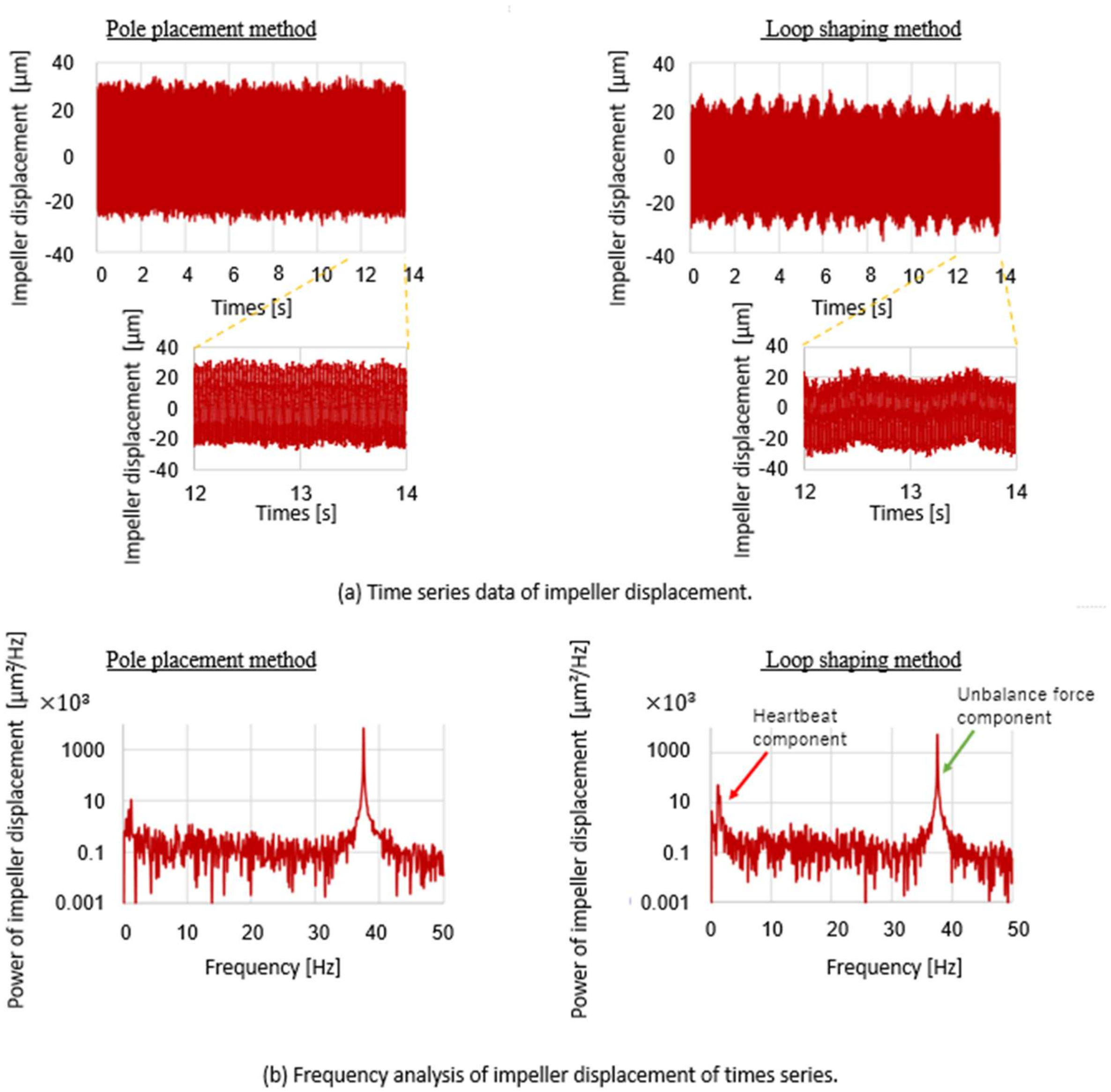


Fig. 7 (a) Time series data of impeller displacement. (b) Frequency analysis of impeller displacement. Both the left side represents the control system of the conventional pole placement method, while the right side represents the control system of the loop shaping method from this study.

4.3 Results of fluid force estimation

The fluid forces estimated by each method are shown in Fig. 8. The loop shaping method doubled the S/N ratio compared to that of the conventional method in case of heartrate was 70 bpm.

It can be considered that the estimated S/N ratio improved because the displacement of the impeller was increased in response to noise owing to the control system of this research. If possible, I would like to improve the S/N ratio a bit more. One method is to enhance the peak of the sensitivity function weights up to the limit where the stability of magnetic levitation for impeller can be maintained. In this study, we are allowing some margin for the stability of magnetic levitation. Furthermore, if the weight function determined that can also reduce the unbalanced force components of the stiffness characteristic in the future, further improvement in the S/N ratio can be expected. The S/N ratio of estimated fluid force was doubled with weak pulsations, but as a result of experiments conducted with strong pulsations, we have also achieved better effects.

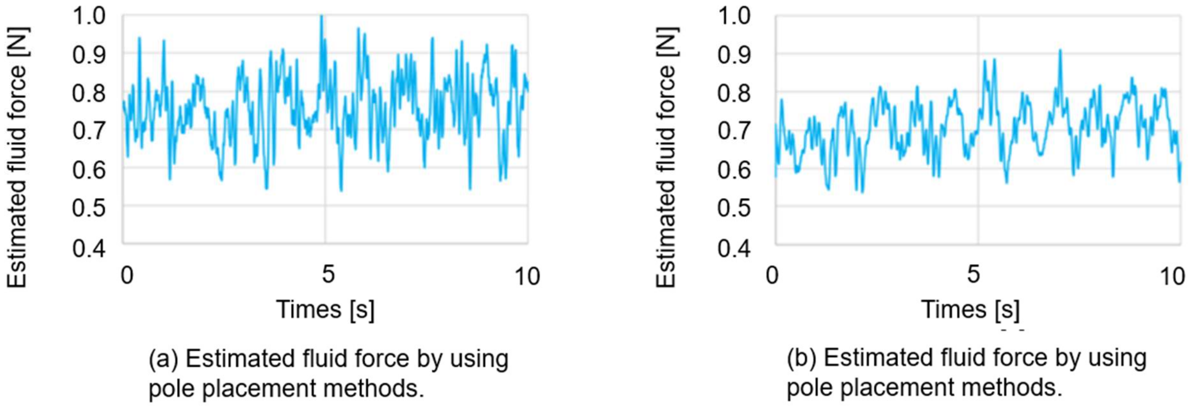


Fig. 8 Estimated fluid forces. (a) The fluid force estimated using the conventional control system. The pulsation component is buried in noise and is not clear. (b) The fluid force estimated using the control system incorporating the loop shaping method, which is the approach of this study.

5. Conclusion

Previous research has conducted heartbeat synchronization control in Maglev VAD. However, there was an issue with the low S/N ratio of the estimated fluid force, making it difficult to estimate weak pulsations. Therefore, in this study, we introduced a loop shaping method based on H_∞ control theory to construct a control system that enhances the pulsation. With this control system, we are able to double the S/N ratio of the estimated fluid force compared to conventional methods. This achievement raises precise heartbeat synchronization control in the future.

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