Dynamic Characteristics of Large Gap Magnetic Driving blood pump
during start-up period

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In order to obtain operational dynamic law driven by large gap magnetic force, and achieve optimal control of pulse frequency during transient period, dynamic characteristics of blood pump were studied.

First, driving torque of the large gap magnetic drive system was solved. Second, the dynamic equilibrium equation, the continuity equation of water flow, and the net head-flow equation of the pump during start-up period were deduced. Then the mathematical model of dynamic characteristics of the pump was adduced.

Taking the start-up process under the coupling distance of 60 mm between the driving electromagnet and the driven permanent magnet as an example, the mathematical model was solved using MATLAB, and the relation between various important parameters such as speed, flow with regard to time were obtained. Calculated results showed that the blood pump is accelerated rapidly from zero speed to its maximum speed at 3450 r/min and then established at the steady-state speed, and the time required to reach steady state is 1.348 s , while the flow increases first and then decreases, and the maximum flow is $3.18 \times 10^{-5}\text{ m}^3/\text{s}$, and the time required to reach maximum flow is 0.821 s .

Then the optimal driver was designed according to the discretization of the acceleration curve, and used for blood pump experimental research. By comparing the curves resulted from the numerical solution of the dynamic mathematical model with the experimental characteristics curves, the correctness of the model was demonstrated. The experimental results also confirmed that, compared to the driver based on the linear acceleration curve, the optimal driver can significantly shorten the acceleration time, and efficiency of energy transfer of the driving system is comparatively higher.

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ABSTRACT

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Key Words: magnetic drive; dynamics; blood pump; start-up period; large gap

1. INTRODUCTION

Avoiding mechanical wear and tear of traditional mechanical transmission, non-contact magnetic driving has become a hotspot of both academical and industrial interests. To realize the external magnetic driving of wireless endoscope, Li et al. (2008) propose a method to design coils of external magnetic field using three pairs of orthogonal coils; Chao et al. (2007) come up with a method to generate an external rotating magnetic field. Moreover, Karel (2007) applies rotary magnetic field to drive fluid in square vessel, and establishes a magnetic force computation model. Working on the axial flow blood pump, Tan et al. (2008, 2009) present a large gap magnetic drive method. When frequency of current pulse of coil being changed, electromagnet will generate a traveling wave magnetic field to drive permanent magnet. As such, rotation of blood pump impeller etc. connected with the permanent magnet can then be realized.

Stepping mode is chosen for the blood pump driven by large gap magnetic force; its speed is determined by pulse frequency of current. Controllers of the stepping motor are usually based on linear, exponential and parabolic profiles of experimental results. However, their real-time tracking performances need to be improved (Wang, 2006). Wang (2000) designs a better controller by linearly interpolating the

torque-frequency characteristic curve of the step motor, solving dynamic equation during start-up process, and applying the result for speed up control. In order to obtain a smooth and fast start-up and speed regulation process for the blood pump, it is necessary to find out its dynamic characteristics and operation rules. Based on the studies (Komori and Uchino, 2004; MASATSUGU, 2006) on driving torque of magnetic transmission system, Xu et al.(2009) apply equivalent current method theory, and establish a mathematical model for driving torque through numerical solution deduction. Yan (1989) and Ge et al. (2006) study the dynamic characteristics of large axial-flow pump during start-up period, providing a method to calculate and deduct load torque of blood pump.

In this paper, a mathematical model of dynamic characteristics of blood pump was established by solving the driving torque of traveling wave magnetic field. In addition, we deduced the dynamic equilibrium equation, the continuity equation of water flow, and the net head - flow equation of the blood pump during start-up period. We applied MATLAB to solve the model and obtain acceleration and flow curves. From acceleration curve, we obtained speed adjustable control parameters, which were used for water pumping experiment

2. ESTABLISHMENT OF BLOOD PUMP DYNAMIC MODEL DURING START-UP PERIOD

2.1 Establishment of Blood Pump Dynamic Equilibrium Equation

The large gap magnetic driving system of blood pump is illustrated in Figure 1, and the structure of the pump rotor is given as shown in Figure 2. The dynamic equilibrium equation during pump's start-up period is constituted by system's driving torque (T_M), water pumping resistance torque (T_B), inertia torque (T_I) and additional inertia torque (ΔT_I) as follows:

$$T_M = T_B + T_I + \Delta T_I \quad (1)$$

Figure 1. SCHEMATIC OF THE DRIVE SYSTEM

Figure 2. SCHEMATIC OF THE BLOOD PUMP ROTOR'S STRUCTURE

1. Bearing 2. Shaft 3. Impeller 4. Permanent Magnet

2.1.1 Driving Torque of the Large Gap Magnetic Drive System

The mathematical model for the driving torque of the large gap magnetic drive system is: (Xu et al, 2009)

$$T_M = \frac{2M}{\cos \theta} \cdot \int_{-\frac{l_1}{2}}^{\frac{l_1}{2}} \int_{r_2 \cos \theta}^{r_1 \cos \theta} y \cdot (B_y + B_z \tan \theta) dy dx \quad (2)$$

Where r_1 , r_2 , l_1 and M are the outer radius, the inner radius, the axial length, and the magnetization of the permanent magnet respectively. θ is the angle between the magnetic pole of the permanent magnet line and the y-axis.

B_y and B_z are the spatial magnetic field components in the y and z directions at the point (x, y, z) respectively.

2.1.2 Water Pumping Resistance Torque of the Blood Pump

Assuming that the blood pump's power loss can be ignored, i.e., the pump's efficiency equals to 1, and according to the pump's operating characteristics, its water pumping resistance torque can be described by:

$$T_B = \frac{\rho g Q H}{\omega} \quad (3)$$

Where Q is the discharge, H is the head, and ω is the instantaneous angular velocity of the pump.

2.1.3 Inertia Torque and the Additional Inertia Torque

The inertia torque of the blood pump can be described by:

$$T_I = J \frac{d\omega}{dt} \quad (4)$$

Where J is the blood pump rotor's rotary inertia.

During the blood pump's normal starting, water inside the pump chamber rotates driven by the pump blade, which leads to the additional inertia torque (Yan, 1990) which can be described by:

$$\Delta T_I = \Omega_J D^5 \frac{d\omega}{dt} - \Omega_M D^2 \frac{dQ}{dt} \quad (5)$$

Where D is the impeller diameter of the pump, Ω_J and Ω_M are the rotational inertia constant and the flowing inertia constant of the water flow respectively, which can be calculated according to the references in the axial flow blood pump.

Substituting formula (2), (3), (4), (5) into formula (1) gives rise to:

$$T_M = \frac{\rho g Q H}{\omega} + J \frac{d\omega}{dt} + \Omega_J D^5 \frac{d\omega}{dt} - \Omega_M D^2 \frac{dQ}{dt} \quad (6)$$

Based on the related literature (Chang, 2005) and a preliminary calculation, we can conclude that the additional inertia torque of the pump can be ignored for $\Delta T_I \ll T_M$, thus,

$$T_M = \frac{\rho g Q H}{\omega} + J \frac{d\omega}{dt} \quad (7)$$

That is,

$$\frac{d\omega}{dt} = \frac{1}{J} (T_M - \frac{\rho g Q H}{\omega}) \quad (8)$$

2.2 Continuity Equation of Water Flow during Pump's Start-Up Period

The water level difference between the inside and outside the blood pump device continued to increase during start-up process. The flow passage was supposed to be a rigid system, and the compressibility of water wasn't considered, then we can apply water-mass continuity to derive the following:

$$\frac{dH_a}{dt} = Q / S \quad (9)$$

Where H_a is the water level difference between inside and outside the blood pump device, namely, net head, S is the cross-sectional area of the pipe.

2.3 Net Head – Flow Equation

The hydraulic characteristics of the blood pump and its device was analyzed to get the pump's net head – flow

equation, and then $\frac{dQ}{dt}$ can be deduced.

For a certain pump, its head (set H) and flow (set Q) of any working point at any speed have a similarity relation with a given working point (Q_0, H_0) under the rated speed within a certain range during the stable transition process. The relation is described by (Yan 1989; Ge, 2004):

$$H = A Q^2 + B (\omega / \omega_0) Q + C (\omega / \omega_0)^2 \quad (10)$$

Where ω_0 is the pump's rated speed, A, B and C are coefficients of the head – flow equation under the rated speed.

The pump system's start-up transition is an unstable transition, when there is a hysteretic effect of the flow inside the pump and the flow passage, which causes the parameters of the liquid flow velocity field between unstable movement at any time and stable movement at the very previous time to be similar. Moreover, sufficient accuracy for model calculation on unstable transition can be provided only by using characteristics on stable transition. Thus, formula (10) can be taken as a dynamic parameters relation formula approximately.

Besides, the pump's head under an arbitrary speed during start-up period can be described by:

$$H = H_a + H_f + H_l \quad (11)$$

Where H_f and H_l are the loss head (including the linear loss and the local loss) and the inertial head inside the pump flow passage. Using equations (10) and (11), the following relation can be obtained:

$$A Q^2 + B (\omega / \omega_0) Q + C (\omega / \omega_0)^2 = H_a + H_f + H_l \quad (12)$$

So the net head – flow equation can be rewritten as follows:

$$H_a = A Q^2 + B (\omega / \omega_0) Q + C (\omega / \omega_0)^2 - H_f - H_l \quad (13)$$

Where H_f and H_l are described by (Ge, 2004):

$$\begin{cases} H_f = \lambda Q^2 \\ H_l = \frac{M}{g} \frac{dQ}{dt} \end{cases} \quad (14)$$

Where λ is the hydraulic friction coefficient inside the pump flow passage (Guan, 1995), and M is the flow passage's inertia constant, which can be expressed by the general formula for an equal diameter straight pipe:

$$M = \frac{l}{S} \quad (15)$$

Consequently, formula (13) can be rewritten as follows:

$$H_a = (A - \lambda) Q^2 + B (\omega / \omega_0) Q + C (\omega / \omega_0)^2 - \frac{l}{gS} \frac{dQ}{dt} \quad (16)$$

That is,

$$\frac{dQ}{dt} = \frac{gS}{l} [(A - \lambda) Q^2 + B (\omega / \omega_0) Q + C (\omega / \omega_0)^2 - H_a] \quad (17)$$

2.4 Establishment of Blood Pump Dynamic Model during Start-Up Period

A set of differential equations can be obtained by substituting formula (10) into formula (8) and then synthesizing formula (9) and formula (17), namely dynamic model of the blood pump during start-up period as follows:

$$\begin{cases} \frac{d\omega}{dt} = \frac{1}{J} [T_M - \frac{\rho g Q}{\omega} (A Q^2 + B (\omega / \omega_0) Q + C (\omega / \omega_0)^2)] \\ \frac{dQ}{dt} = \frac{gS}{l} [(A - \lambda) Q^2 + B (\omega / \omega_0) Q + C (\omega / \omega_0)^2 - H_a] \\ \frac{dH_a}{dt} = \frac{Q}{S} \end{cases} \quad (18)$$

3. NUMERICAL SOLUTION OF THE DYNAMIC MATHEMATICAL MODEL OF THE BLOOD PUMP DURING START-UP PROCESS

By invoking the fourth-order Runge - Kutta function in MATLAB, the dynamic mathematical model was solved.

3.1 Calculation Conditions

(1) driving torque

When the parameters of the electromagnet and the permanent magnet were fixed, the driving torque only determined by the driving current and the position coordinates (x, y, z) of the permanent magnet in the driving system. The position coordinates were adjusted to $x = 5\text{mm}$, $y = 0\text{mm}$, $z = 60\text{mm}$ (the distance along the z direction is also the coupling distance), and the coil current was adjusted too, so that the magnetic induction intensity at the position of the center of the permanent magnet will be $0.5 \times 10^{-3} T$. Other parameters were consistent with the literature (Xu et al, 2009).

By the calculation of equation (2), the drive system's torque-angle characteristic was obtained. For the convenience of research, the average driving torque was defined that it did the same work with the actual driving torque during each circle of the blood pump's rotation. By analyzing the torque-angle characteristic of the drive system, the driving torque was calculated out to be $0.201 N \cdot \text{mm}$.

(2) related parameters of the blood pump and it's device

$$\begin{aligned} J &= 0.279 \times 10^{-6} N \cdot \text{mm}^2, & D &= 0.0158 \text{m}, \\ \omega_0 &= 39000 \text{deg/s}, & \lambda &= 6.806 \times 10^7 \text{s}^2/\text{m}^5, \\ S &= 1.77 \times 10^{-4} \text{m}^2, & l &= 1.75 \text{m}, & A &= -0.0245, \\ B &= 0.0155, & C &= 0.951 \end{aligned}$$

3.2 Numerical Calculation and Results Analysis

The calculated results of the dynamic mathematical model were shown in Figure 3.

It is can be seen that:

(1) The blood pump is accelerated rapidly from zero speed to its maximum speed at 3450 r/min , and then established at the steady-state speed, and the time required to reach

steady-state is 1.348 s .

(2) The flow of blood pump increases first and then decreases, and the maximum flow is $3.18 \times 10^{-5} \text{ m}^3/\text{s}$, and the time required to reach maximum flow is 0.821 s .

Figure 3. THE CALCULATED RESULTS OF THE DYNAMIC MATHEMATICAL MODEL

4 EXPERIMENT RESEARCH

4.1 Experimental Principle

According to calculated results of the dynamic mathematical model, there is a corresponding acceleration curve, or pulse frequency curve, under certain coupling distances. Taking pulse frequency curve under 60mm as an example, optimal driver was designed according to the discretization of the acceleration curve, and water-pumping experiment was done.

The start-up process of the blood pump was photographed, and then the pictures of the experiment was processed and analyzed. The speed was gained from the digital display tube of the SCM control system, and the time for the pump to accelerate to the specified speed was measured. While the net head was gained from the glass tube with scale, then the difference of the net head between tiny time intervals t and $t + \delta t$ (here, $\delta t = 0.167 \text{ s}$, is the time for high-speed shooting intervals of the camera) was obtained. Further the instantaneous flow could be calculated

out by:

$$Q = \frac{dHa}{dt} \cdot S = \frac{Ha_2 - Ha_1}{\delta t} \cdot S \quad (19)$$

Where Ha_1 is the net head at time t , and Ha_2 is the net head at time $t + \delta t$.

4.2 Experimental Conditions

Experimental device including DC Power Supply, SCM Control System, Experimental Table, and Digital Camera was shown as Figure 4.

Figure 4. EXPERIMENTAL DEVICE

1. DC Power Supply 2. SCM Control System
3. Water Reservoir 4. Driving Electromagnet 5. Glass Tube
6. Switching Ball Valve 7. Experimental Table 8. Blood Pump

4.3 Experimental Procedure

- (1) Set the pulse frequency difference $\Delta f = 5\text{Hz}$, and discretized the frequency dispersion curve, then obtained the required running time under each frequency class. The discretization of pulse frequency curve was shown in Figure 5.
- (2) Adjusted the magnet position and coil current, and photographed the start-up process of the blood pump.
- (3) Analyzed the experimental records of pictures, and measured the time for the pump to accelerate to the

specified speed. Took δt as the time interval, and read the net head at each time, then calculated the corresponding flow.

Figure 5. SCHEMATIC OF DISCRETIZATION OF PULSE FREQUENCY CURVE

4.4 Experimental Data

- (1) The relation between speed, net head, flow etc. with regard to time of the pump during start-up process were shown in Figure 6, 7, 8, where Exp was short for the Experimental data, while Sim was short for the simulated data.
- (2) The time for the pump to accelerate to the specified speed was shown in Figure 9, where broken line 1 was in stand of the acceleration process controlled by the driver based on the acceleration curve from dynamics research, while broken line 2 was in stand of the acceleration process controlled by the driver based on the linear profile.

4.5 Experimental Result Analysis

- (1) It can be seen from Figure 6 that the difference between the transient speed curve from experimental research and the one from numerical calculation is very tiny, and the former one is always slightly above the later one due to the neglect of the blood pump power loss during the modeling process of the dynamic characteristics.
- (2) Comparing the transient net head resulted from experimental research and the one from numerical

Figure 6. SPEED CURVE OF THE BLOOD PUMP DURING START-UP PERIOD

Figure 7. NET-HEAD CURVE OF THE BLOOD PUMP DURING START-UP PERIOD

Figure 8. FLOW CURVE OF THE BLOOD PUMP DURING START-UP PERIOD

Figure 9. TIME FOR THE BLOOD PUMP ACCELERATE TO SPECIFIED SPEED

calculation shown in Figure 7, it is obviously that they are not quite consistent. Moreover, with the progressing of the start-up process, their difference will be larger until the end of the start-up process. The error, generated by the discretization of speed curve during the design of the driver, leads to a longer Δt , which slows the start-up process and causes the differences between net head resulted from experimental research and the one resulted from numerical calculation. After the starting process, the difference between them will be stabilized.

- (3) It is inferred from Figure 8 that the transient flow resulted from experimental research and the one from numerical calculation are pretty much the same. What's more, with the progressing of the start-up process, they will be more consistent. From the calculation method of the experimental flow shown as equation (19), it is known that the transient flow is determined by difference between the net head at time $(t + \delta t)$ and

time t . That means the difference between the net head resulted from experimental research and the one from numerical calculation will cause the difference between the transient flow resulted from experimental research and the one from numerical calculation, while the difference won't be larger with the progressing of start-up process. After the start-up period, the difference between them will disappear.

- (4) From Figure 9, it is exciting that the time for the pump to accelerate to the specified speed is significantly shortened by the adoption of the driver based on the acceleration curve from numerical calculation of dynamic model. Acceleration curve resulted from the numerical calculation of the dynamic mathematical model meets the dynamic equations of the blood pump during start-up period, which guarantees that the torque required by acceleration can take full advantage of the torque provided by the drive system, and therefore the obtained pulse frequency control curve is optimal. However, the slope of the linear acceleration curve, on which the former driver was based, should be very low to ensure not miss steps, which will lead to a much longer acceleration time.

5. CONCLUSIONS

This paper studied dynamic characteristics of large gap magnetic driving blood pump during start-up period. The main research work was summarized as follows:

- (1) The mathematical model of dynamic characteristics of the blood pump was established by solving the driving torque of the traveling wave magnetic field, deducing the dynamic equilibrium equation of the pump during start-up period, and analyzing the hydraulic characteristics of the blood pump.
- (2) The mathematical model under the coupling distance of 60 mm was solved using MATLAB. Calculated results gave the relation between various important parameters such as speed, flow with regard to time.
- (3) Parameters for speed regulating, Δf and Δt , were obtained by discretization of the speed curve of the blood pump during start-up period, and were used for blood pump experimental research. Experimental results

confirmed that compared to the driver based on linear acceleration curve, the driver based on the study results can significantly shorten the acceleration time, and efficiency of energy transfer of the driving system is much higher.

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