Research on synchronisation control method of frequency and phase of pulsating blood pump

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Abstract: At present, there exists a problem that the frequency and phase are not synchronised in the blood pump pulsation flow control. After analysing the time sequence of the blood pump-heart coupling model, this study has designed a synchronisation control method based on heart rate prediction and delay control. By predicting the next moment of cardiac action, the change in blood pump speed was delayed to match the frequency and phase of the cardiopulmonary system. The new prediction method increases the linear weight and state variables, and solves the problem that the traditional method has poor real-time performance, and still has better detection accuracy. Through the synchronous tracking experiments of the blood pump, the feasibility of the method has been proved, and the technical accumulation of the fully synchronised blood pump pulsation flow control method could be provided in the future.

1 Introduction

As one of the most important methods to solve the problems of heart failure and heart disease, ventricular assist device – blood pump – has drawn more and more attention and research [1]. At present, many mature products have been put into clinical practice, such as HeartWare and Jarvik2000 and so on, which have achieved good application results. These blood pumps mainly provide stable and constant flow, and relevant control theories and methods are relatively mature. However, with the progress of relevant studies, it has been found that steady-state bleeding pump can lead to impaired renal function [2], haemorrhagic stroke [3], oxidative stress [4], increased aortic stiffness [5], multiple organ failure [6] and many serious problems.

At present, there have been many researches on pulsation control system of blood pump. Petrou [7] developed a novel multi-objective physiological control system that relied on the pump inlet pressure. The pump flow is adapted in a physiological manner, following the preload changes, while the aortic pulse pressure yielded a threefold increase compared to a constant-speed operation. Amacher et al. [8] used electrocardiogram (ECG) signals as input signals; blood pump control can be synchronised with the heart circulation, and can monitor irregular heartbeat, but it does not take into account the delay problems caused by the actual synchronisation control process. According to the relevant papers, only strict frequency and phase synchronisation can achieve better pulse performance and energy utilisation.

The frequency and phase require the blood pump to provide a higher flow rate during cardiac contraction and a lower steady flow rate during cardiac diastole. Synchronisation control has following difficulties: as for signal acquisition, the traditional ECG instrument is used to detect the electrical signal of the heart. However, relevant studies have proved that devices equipped with blood pump can interfere with the collection of corresponding signals by sensors [9]. Therefore, photoplethysmography (PPG) signal can only be used. The PPG signal detects pulse pressure and has a certain delay relative to the cardiac action time. Due to the limited volume power consumption, the blood pump cannot achieve a high driving torque, and its acceleration performance is limited, which will also cause delay in dynamic response.

In order to solve the problem that the blood pump pulsation frequency and phase do not match the best requirements of the real human body, this paper designed a synchronous control strategy based on heart rate prediction and delay control method. It can stably match the heart rate characteristics of the human body, so as to achieve the state that the blood pump working frequency and phase meet the best needs of the human body.

2 Method

2.1 Synchronisation control strategy

The timing sequence diagram (Fig. 1) of human blood circulation system can be established through the research and experiment of human blood system [10].

The curve (a) refers to the ECG signal, and curve (b) refers to the left ventricular pressure signal. Curve (c) refers to the pulse volume signal, and curve (d) refers to the filtered and calculated output signal. Cardiomyocytes start to act after the peak of the ECG signal (time t₀). If the left ventricular volume contracts, then the ventricular blood pressure increases, and the blood pressure peak is reached at time t₁. The pressure wave is transmitted in the blood vessel with the human blood circulation system, and the pulse volume peak signal is received at time t₂. The sensor actually detects the pulse volume peak value at the time t₃ after operations such as filtering and decoupling.

The traditional synchronous control detects the occurrence of the relevant signal to drive the motion unit to carry out corresponding actions. However, due to the weak dynamic performance of the blood pump, the time consumption of the signal acquisition and the transmission delay of the PPG signal itself, combined with the short time of the cardiac cycle, simple follow-up control strategy will cause a large following hysteresis.
Seriously, even counterpulsation appears, which means the lag about half cardiac cycle. At this time, the blood pump not only cannot increase the pulse for the blood system, but also will inhibit the fluctuation performance.

Based on the above considerations, this paper designed a delay action control strategy. Through the method of heart rate prediction, the next heart action time ($t_4$) is estimated. The blood pump delays the action to match the next cardiac action moment, thereby achieving the goal of synchronous follow-up.

The calculation equation of active delay time can be obtained from the sequence diagram. Take the last pulse pressure peak signal as the initial time

$$T_d = |p(s)| = |p(s)| - |p(s)| = T_p - T_1 - T_c$$  \hspace{1cm} (1)

In (1), $T_d$ refers to the time of the motor to delay the action, which corresponds to the time from $t_1$ to $t_4$ in Fig. 1. $T_p$ refers to the cardiac cycle, in Fig. 1 corresponding to the time from $t_1$ to $t_4$. $T_c$ refers to the pulse wave transit time, in Fig. 1 corresponding to the time from $t_0$ to $t_2$. $T_r$ refers to the computation time taken by the processor to filter and decouple the signal, in Fig. 1 corresponding to the time from $t_2$ to $t_3$. $T_r$ is mainly affected by blood pressure, but does not change much for an individual [11], so it can be considered as a constant value. $T_c$ is primarily affected by controller performance and is essentially unchanged. The main time variable is $T_p$, so the prediction of $T_p$ is the key of the synchronisation control method.

### 2.2 Synchronisation control strategy

Heart rate signal is a complex signal with both periodicity and randomness. According to the statistics of heart rate of normal people and patients with heart failure, most of them are between 0.5 and 2 Hz, and the specific value depends on individual differences and exercise status. The overall frequency has a significant regularity, so it is possible to predict the heart motion moment of the next cycle within a certain error range.

The current mainstream heart rate prediction methods exhibit some problems. Zhang et al. [12] formalised a multi-step prediction scheme for heart rate during running using the Bayesian combined predictor. They constructed a neural network predictor and a linear regression predictor as the basic prediction models by parameter learning with training data. Then a Bayesian combined predictor was constructed by training the weights of the basic predictors to give the multi-step prediction process of heart rate during running. This method has good accuracy, but requires a large amount of calculation, and is difficult to be applied to real-time prediction and control of heart rate.

The traditional prediction method uses the average heart rate for direct prediction, so the prediction accuracy is low and the fluctuation is large. This paper added weights and judgment variables to the simple average prediction, which can greatly improve the overall prediction accuracy without affecting the real-time performance

$$T_p = \frac{1}{N_p}\sum_{k=1}^{N_p}(kT_{N-k} + kT_{N-k} + \cdots + kT_{N-k} + \cdots + kT_{N-k})$$

$$+ T_L$$  \hspace{1cm} (2)

$T_p$ can be calculated by (2). The value of the weight $k$ has a great influence on the prediction performance. The most common values include the equivalent weight, the linear weight and the square weight. The specific equation is shown in Table 1. $T_L$ refers to the judgment variable, whose value depends on the changing trend of the previous heart rate cycles. If the previous heart rate cycles are continuously increasing or decreasing, that is, the heartbeat is accelerated or slowed down, then

$$T_L = \frac{1}{n} \sum_{i=0}^{n}(T_{N-i+1} - T_{N-i})$$  \hspace{1cm} (3)

Otherwise

$$T_L = 0$$  \hspace{1cm} (4)

### 2.3 Delay time calculation

The ECG signal is collected in real time by the ECG sensor, and the PPG sensor detects the volume change signal at the pulse in real time. The time difference between the two modes is the transmission time $T_t$ of the pulse wave. After calculating the time difference between the waveform of the real-time sampling and the waveform after filtering and decoupling, the calculation delay time $T_c$ can be obtained (Fig. 2).

### 3 Result and discussion

In this paper, six groups of 360 s (129,612 data points, 426 cardiac cycles) in the professional MIT-BIH database were used for statistical analysis. It is the data under the steady state of the human body, that is, the heart rate of the human body is generally stable. The values in (1) and (2) are: $n = 10$, $m = 5$. The program code was written and the three methods were compared and verified. The relevant comparison is shown in Table 2.

For the prediction under the stable heart rate state, the difference between the three methods is very small, and the errors are basically within 30 ms. We also used the pulse sensor to collect heart rate data from rest to motion for 100 s, that is, the heart rate has changed significantly. The specific results are shown in Table 3.

According to the comprehensive analysis, the linear weight method can most effectively fit the heart rate in multiple states. The heart rate prediction model of the new calculation method has higher prediction accuracy and is easier to calculate. This method is used in subsequent control systems to predict $T_p$.

ECG sensor BMD101 and PPG sensor PulseSensor were used (shown in Fig. 3). After the relevant data is transmitted to the acquisition device, the waveform is compared and judged. After counting the ECG waveform, PPG waveform and calculation data in 100 s, we obtained the final result as $T_c = 0.344 s$, $T_c = 0.133 s$.

The data collected by PulseSensor was taken as the input signal. The axial flow blood pump developed by ourselves was used as the control object, and the experiment was carried out by the traditional tracking control and the pulse synchronous control under heart rate prediction. The specific results are shown in Figs. 4 and 5.

In the traditional control method, it can be seen that the target speed of the blood pump starts to rise after the peak of the pulse volume is detected. Due to the performance of the motor and various detection delays, the difference between the peak moment of the real heart motion and the peak speed of the blood pump is

| Table 1 | Coefficient weight expression |
|---|---|
| Weight coefficient | Coefficient expression |
| equivalent weighting | $k = k_0 = c$ |
| linear weighting | $k_i - k_{i+1} = c$ |
| square weighting | $k_i/k_{i+1} = f(t - 1)^2$ |

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about 0.4 s. This situation will not increase the pulsation performance, but will inhibit the pulsation.

According to the previous statistics, the peak value of the PPG signal lags behind the true peak of the ECG signal by about 0.344 s, so the speed rise time of the blood pump should be about 0.344 s before the peak of the PPG signal. The delay control strategy is reduced to an error of 0.1 s, which is basically synchronised with the detection signal, and the synchronisation performance is greatly improved compared with the conventional method.

4 Conclusion

In this paper, a delay pulsation control method based on heart rate prediction is designed. By predicting the next cardiac moment to delay the action, the purpose of synchronisation between blood pump pulsation and heart pulsation is achieved. The experiment

| Table 2 | Comparison of prediction methods in stable state |
|---------|--------------------------------------------------|
|          | Predicted number | Prediction error, ms | Variance |
| equivalent weighting | 238 | 27.10 | 1255 |
| linear weighting | 238 | 26.60 | 1181 |
| square weighting | 238 | 27.18 | 1194 |

| Table 3 | Comparison of prediction methods in motion state |
|---------|--------------------------------------------------|
|          | Predicted number | Prediction error, ms | Variance |
| equivalent weighting | 102 | 12.17 | 356 |
| linear weighting | 102 | 10.73 | 313 |
| square weighting | 102 | 9.90 | 296 |
proves that compared with the traditional tracking control method the new method has better results.

The new heart rate prediction method uses weighting coefficients and state judgment variables, and has higher prediction accuracy. Compared with the traditional prediction method, the calculation is relatively simple, and can meet the requirements of real-time and accuracy at the same time.

In the future, the synchronous control of frequency and phase and the feedback of flow pressure will be combined to dynamically adjust the rotating speed and establish a complete human-blood pump synchronous control system, so that the blood pump can better serve patients.

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6 References

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