Heart Simulator: A Periodic Pump to Simulate the Cardiac Motion in an Aortic Test-rig

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Abstract

A periodic pump that simulates blood ejection from human heart to aorta is the core element for building an aortic robotics test-rig. This paper is to describe the design of such a prototype human heart simulator and its performance under different working status, such as simulating the physiological states of a healthy adult and/or a child in sleep, relax and physical exercise. By balancing the cost and performance, this prototype has these specifications: (1) Using ordinary plumbing components and water to simulate the cardiac motion and blood flow. (2) Simulating the volume change of human heart chamber by controlling movement of a mechanical piston. (3) Performing a friendly user interface and delicate control via a MCU system with high reliability. (4) Simulated physiological output parameters such as volume per stroke, heart beat rate and waveform can be easily adjusted and monitored in real-time.

Keywords: Heart simulator, cardiac motion, volume per stroke, heart beat rate, aortic test-rig

1. Introduction

When developing surgical robotics for cardiovascular interventions, conducting experiment on human body is no doubt dangerous as any defect or failure of such robots can be fatal. In this case, experiments need to be conducted on an aortic test-rigs which designed as human physiology structure and built in physical dimension before any test on human body. However, some time a test-rig with physiological shape and dimension is not enough, especially for development of surgical robot aiming for cardiovascular interventions. Because thrust of blood flow near human heart is strong and considered as a factor that may affect the performance of surgical robot. Therefore, it is necessary to design and develop a heart simulator which is able to generate simulated liquid flow with physical thrust and pattern inside test-rigs. It is worth mentioning that currently there are no many researches aiming on this area. Therefore, literatures in close areas are also used for references.

[CardioSmart, 2008], [Down's Heart Group, 2012]and [IVLine, 2010]introduced how a real human heart works in details, including how each parts of the heart work, how they cause effect with each other. Meanwhile, AVRA Medical Robotics Inc has listed many products of artificial robot systems all about Robotic Surgery, such as Pacemakers and Implantable Cardioverter Defibrillators(ICDs), The Left Ventricular Assist Device(LVAD). These works have discussed the general idea of how the human heart works and the basic functionality andprinciple of current robotic products.

[Amoore, et al., 1985] has mentioned the change of human heart volume (VOL) can be calculated by voltage of RS signal and the wall thickness (TH). The relationship can be described as formula 1. It is worth to mention that the RS signal is a certain period of time on ECG, which is shown in Fig.1.

$$RSvoltage(mV) = 1.34 + 0.016\,TH - 0.002\,VOL \tag{1}$$

All of [Liu et al., 2006], [Wu et al., 2004], [Gregoryet al., 2010] and [Roche et al., 2014] have introduced the development of test-rigs targeting different cardiac devices by building a mock cardiovascular circulation system. [Liu et al., 2006], [Wu et al. 2004] and [Gregory et al. 2010] were targeting building test platforms for left ventricular assist devices (LVADs), while [Roche et al., 2014] presented a development of an elastomeric cardiac model for test of all cardiac devices, by using 3D printing technology to restructure the entire human heart and deploying control system. Although the test-rig that [Roche et al., 2014] presented is more advanced than others, it is too complex and time consuming. Considering the short time frame and resource wehave, the other models that targeting specific surgical robotics seem with less complexity and more feasible.

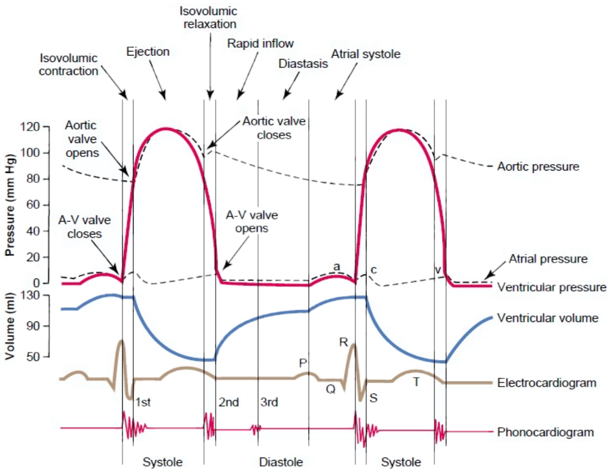

Figure1 Cardiac cycle ([IVLine, 2010])

[Hinteregger et al., 2016], [Nejadrajabali et al., 2016] focused on the development of the flow pump. [Hintergger et al., 2016] designed a pneumatical pump to drive blood, while [Nejadrajabali et al., 2016] concentrated on investigating the effect of blade angle on the performance of regenerative pump. Meanwhile, [Magneticbearing 2012]introduced recent development on ventricular assist device. It presented the mechanical structure of a device called Life-Flow, which is also an implanted mini pump system that enhances human blood flows. Unlike other designs, this device drives blood flow with an internal duct rotor.

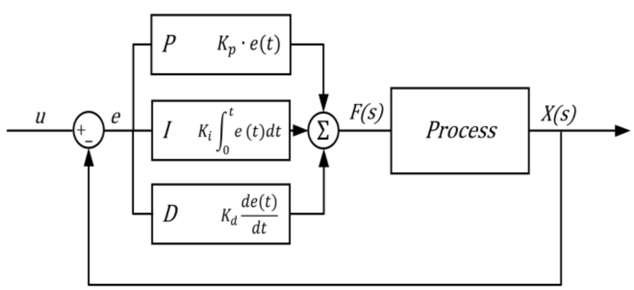

Figure2 PID Controller

As for control algorithm, there are two types of control algorithms may suitable for this project. The traditional PID controller, or fuzzy logic control system. [Astrom,2002] and [Astrom and Murray, 2008] introduced principle and implementation of PID controller. While [Martinez, 2015] gave a detail example about how to implement PID controller on Matlab, which is very useful as functions it introduced can be directly used in firmware development. Compare to PID controller which has already been widely used for decades, fuzzy logic controller is more powerful and fancy. However, according to [Lee, 1990] and [Passino andYurkovich, 1998], deploying fuzzy controller requires a well-designed mathematic model. Additionally, fuzzy controller requires much more computational resources than PID controller, in this case, normal MCU system may not be able to work in real time. Considering the performance limitation of low-cost MCU system, choosingtraditional PID controller seems a much more realistic option.

2. Design of the heart simulator

2.1 Aim

The aim of this workis to develop a device to output water flow in a pre-set physiological pattern as human blood flow. The output is adjustable within a physiological range, which is heart beat rate (HBR) from 60 to 100 per minute and volume per stroke (VPS) from 60 to 80mL. This device shall take 200 measurements within each cardiac cycle and report it as feedback, in order to monitor the piston movement.

In order to simulate the blood ejection flow, the most direct way is by using a piston to imitate the volume change of heart chamber. Under ideal condition, there shall be a linear relationship between the velocity of output flow and movement of the piston. Therefore, it is feasible to simulate the volume and velocity of output flow by controlling the displacement of piston. Meanwhile, in order to minimize the effect caused by factors such as liquid compression and fluid resistance, the diameter of tubes shall be as close as possible.

2.2 Mechanical design

The 3D model of mechanical design is showed in Fig.3.

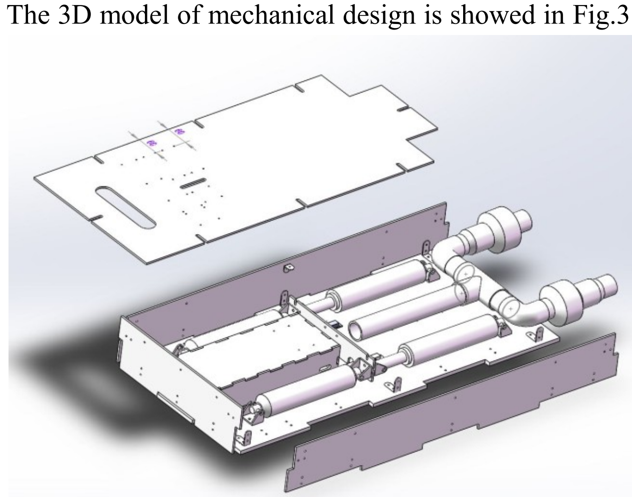

Figure3 3D model of the system

Ventricular stroke volume of a healthy human heart beat rate is around 70mL ([Klabunde, 2015]). Therefore, a cylinder piston that has slight larger displacement is required as it is easier to control with linear actuators. Considering the fastest linear actuator that available in the market can only perform a 270mm/s speed, which requires the cylinder piston the bigger inner diameter the better. Therefore, a 200mL syringe with an internal diameter of 45.63mm is selected as the body of the piston system. According to calculation, the relation of piston travel and displacement is showed in Table 1. Considering when the system is fully loaded with water, it requires more thrust to perform high travel speed. In this case, 4 linear actuators are used to setup 2 sets of push-and-pull system, which shall be able to output 400N thrust in total theoretically. 2 linear potentiometers are used to measure travel of piston. Meanwhile, in order to simulate the function of heart valves, 2 one-way valves are installed at the two ends of water in and out.

Table 1 Relation of piston travel and displacement

Displacement	Piston travel	Potentiometer Voltage	AD Data
60mL	36.71mm	1.52V	156
70mL	42.83mm	1.78V	182
80mL	48.94mm	2.03V	208

2.3 Electronic design

The block diagram of electronic system is showed in Fig. 4. As previously mentioned, linear potentiometer is used to measure piston travel. Considering the 2 sets of linear actuator might have different output thrust, 2 linear potentiometers with 60mm travel are connected in parallel and installed closed to linear actuators in order to balance the reading. Because the actuators are the major sources of electro-magnetic interference (EMI) which might damage the entire electronics system, a serial of methods are deployed to improve the electromagnetic compatibility (EMC). For example, the driving power of actuators is isolated from the digital power with opto-couplers, so that EMI will not be conducted via power lines. Meanwhile, a high precision reference circuit (Fig.5) is used to provide 2.5V reference voltage (Vref) for ADC and potentiometers. Given that this device does not have to perform a fancy UI and the shortest PID sampling period is up to 3ms (HBR on 100/minutes), in order to lower the cost, there is no need to used high performance embedded systems. An Arduino-Nano MCU system is good enough for all the tasks. Moreover, an integrated UART-control LCD touch-screen is used to perform the UI.

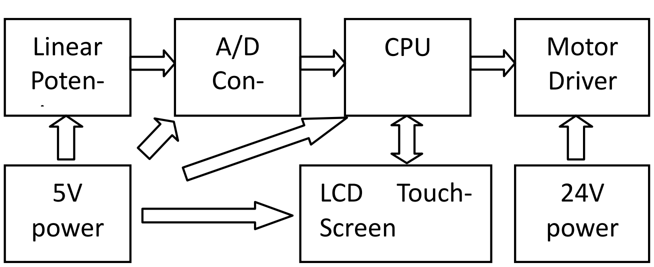

Figure4 Block diagram of electronic system

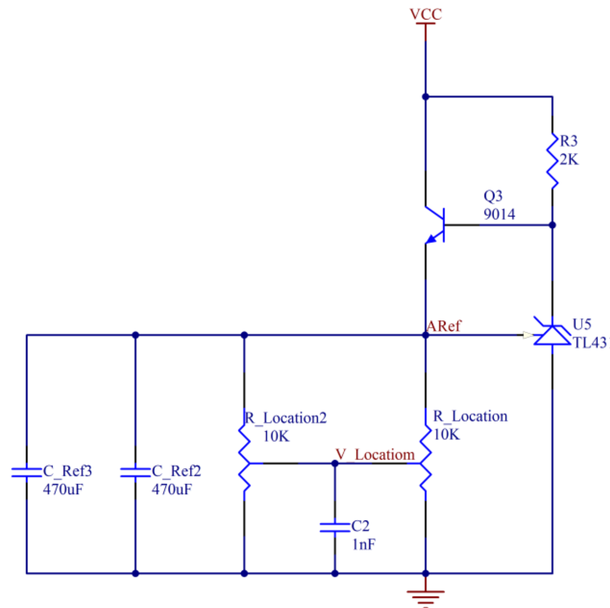

Figure5 High Precision Reference circuit

2.4 Firmware Design

Relations between piston travel and displacement, output voltage and AD data are shown in Eq (2), (3) and(4) respectively. Since the piston travel is digitalized into 8-bit binary number, it is easy to control by digital system. PID algorithm is one of the most common-use micro-controller. It calculates output value based on the difference between current state, desired state and its preset parameters. Then the output value is directly used in a PWM generator to drive the actuators. By digitalizing the desired curve and adjusting the PID parameters, the piston is traveling as the desired curve indicates. However, due to the limitation of PID principle, the curve of desired piston position and actual piston position can never 100% match (unless the system stops moving), there are always defects like overcharge, delay and oscillation on the actual position curve. Considering the water flow can also resist small amount of oscillation due to inertia, as long as the pattern of actual position curve generally matches the desired position curve. Those small defects are considered not affecting the output.

$$\frac{\text{Displacement}}{\text{Travel}} = \pi r^2 \qquad (2)$$

$$\frac{Travel}{Trave\,l_{Max}} = \frac{Vo}{Vref} \qquad (3)$$

$$\frac{\text{Travel}}{\text{Travel}_{\text{Max}}} = \frac{\text{Data}_{\text{AD}}}{256} \qquad (4)$$

Since the piston is moving exactly as the pre-set curve indicates, it is easy to output desired VPS, by enlarging or shirking the curve. Given that the data from piston sensors have to be moving inside an 8-bit binary range, an offset value is involved to shift the piston to a proper starting-point. Eq (5) shows the relation between desired VPS, data of the original set-point curve and processed curve. Due to the hysteresis effect of water along with the defects of PID controller, it is no doubt that there is error in actual VPS. Even though, the error can be easily managed with interpolation compensation methods later.

$$Setpoint_{new} = \frac{Setpoint_{ori} \times VPS \times 2.16}{255} + Offset \quad (5)$$

Timer1 of Arduino system is used to generate output flow as given HBR frequency. Considering there is a requirement for sampling rate of 200/ cardiac cycle, combining this sampling rate with PID controller is no doubt an efficient way to use computing resource. Meanwhile, considering PID controller is a lag controller that needs time for actuation in each cycle, and the timer can generate interrupt once PWM counter is fully loaded, so that the binding PWM frequency with PID controller is the most straight forward way of implementation. To be more specifically, the PID interval can be set certain times longer than PWM period easily by counting interrupt activation of PWM-counter-full for certain times. That is to say, once the new output of PID is calculated, it will always have a certain number of PWM periods to actuate. Considering the UI is performed via UART which requires about 10 bytes data transmission to draw a dot on LCD screen, in order to balance performance and UI, the certain number is set to 10 and the MCU saves all points in each cardiac cycle but only draws one dot in each 5 PID cycles. Relation between PWM period and PID period is shown in Eq(6). Relation of PWM period and HBR is shown is Eq(7). Relation between PWM period and the update interval of UI is shown in Eq (8).

$$Period_{PWM} = \frac{Perio\, d_{PID}}{10} \quad (6)$$

$$Period_{PWM}(uS) = \frac{30000}{HBR} \quad (7)$$

$$Period_{PWM} = \frac{Interva\, l_{UI}}{50} \quad (8)$$

2.5 Prototype

Fig.6 shows the prototype of the heart simulator. Although it slightly differs from the original 3D module due to availabilities of components, they have same structure and follow same working principle.

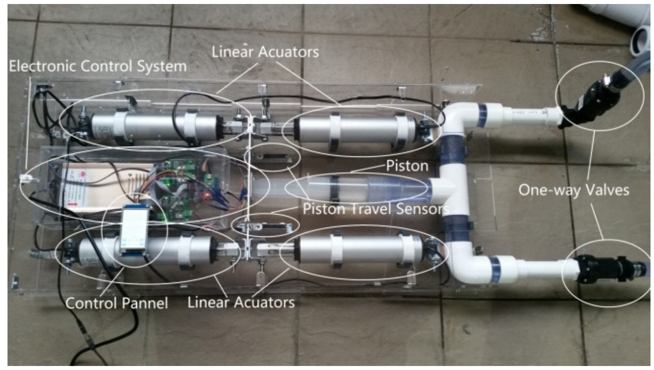

Figure6 Structure of Prototype

The entire system is installed in clear acrylic parts, so that user can closely monitor its working status. The critical parts of electronic systemareisolatedfor water proof consideration.

As shown in Fig.7, the UI is quite straight forward. There are only 4 buttons controlling the setting of VPS and HBR. Each value will increase by 1 for each click. There are 2 curves showing the data of piston travel. The red curve represents the current set-points of piston, while the blue curve shows the current actual position of piston. It is worth mentioned that user can switch the working status between “Go” and “On hold” by clicking on the curving area. Once the heart simulator is on a “On hold” mode, the piston will stop moving until it is switched back to “Go” mode. Therefore, user can modify and setup experiment environment via this function.

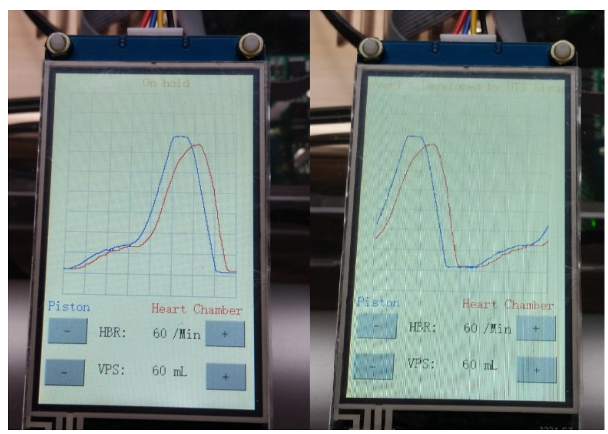

Figure7 User Interface

3. Experiments

3.1 Test method

This device is tested under three settings of VPS (60, 70, 80mL) and five settings of HBR (60, 70, 80, 90, 100/Minutes). During the test, the output pattern is evaluated by feedback of the piston-location sensors,the actual HBR is measured by an oscilloscope and the actual VPS is evaluated by travel of piston and the amount of outputted water is measured by a 1L-counting cup.

3.2 Test of output flow pattern

1000 continuous sample points of experiments under the setting of 60/Minute HBR and 60mL VPS is shown in Fig.8. As show in the figure, although there are some defects on the curve such as delay and some overcharge and oscillation on the actual position, the piston generally moved as the set-points indicated. As previously predicted, these defects can be compensated via later calibration or some of them are considered no affecting the system such as the phase shifting. Therefore, the output pattern meets the requirement.

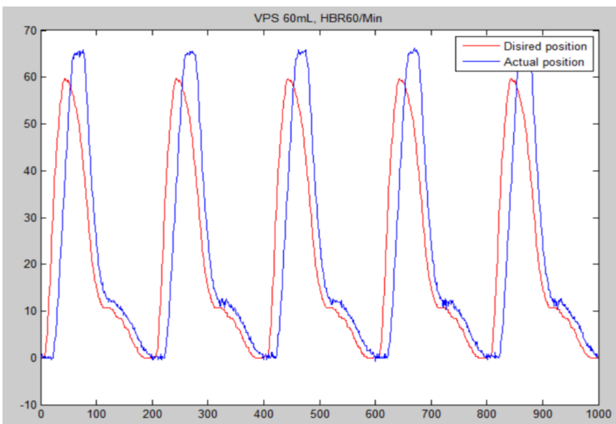

Figure8 Comparison of set-point and actual piston travel (1000 sample points)

3.3 Test of VPS

It is worth mentioning that the actual output of VPS were different under varies setting of HBR. Fig.9 shows the comparison of piston movement under setting of 60mL VPS at HBR of 60, 70, 80, 90 and 100/Minute.

It is obvious that when setting on higher HBR, the device outputs lower VPS.

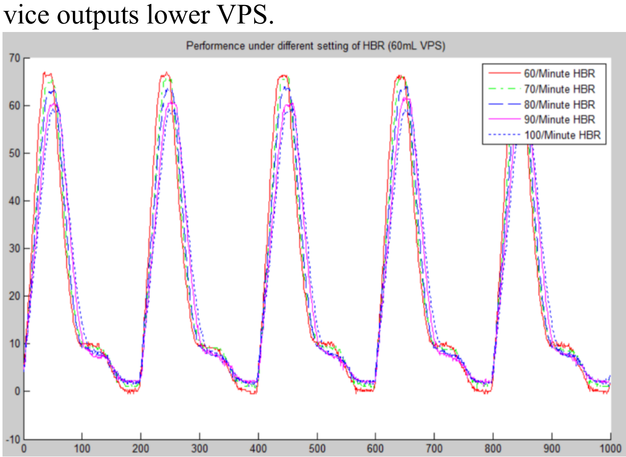

Figure9 Actual VPS on each HBR setting (1000 sample points)

By collecting output water of 10 cardiac cycles and measuring with counting cup, average output VPS was obtained and shown in Table 2. Although the amount of output water slightly differed from the settings, it matched the volume that calculated by data obtained from piston travel sensors. Therefore, as long as the system is calibrated, it is obvious that the actual VPS output will match the setting.

Table2 Average VPS output at different HBR & VPS settings (Center the tables)

VPS\HBR	60/Minute	70/Minute	80/Minute	90/Minute	100/Minute
60mL	67	65	62	60	58
70mL	77	75	70	66	65
80mL	85	82	76	72	69

3.4 Test of HBR

Movement of the pistons can be measured by testing the voltage output of piston-location-sensor with a digital oscilloscope. In this case, the HBR in Hz can be directly read with frequency measurement function as shown in Fig.10. Although the frequency reading is in Hz which is not the unit for HBR, it can be easily transformed to HBR in times/minute by multiplying by 60. Reading of frequency and actual HBR is shown in table3. According to test result, no matter how VPS was given, frequency reading reminded on a certain number. That is to say, actual HBR is not affected by VPS setting.

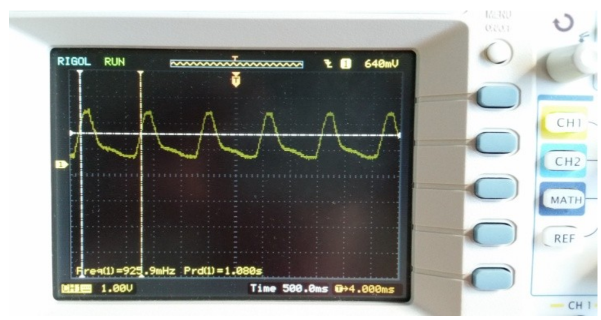

Figure10 Measuring HBR with oscilloscope

Table3AverageHBR output (times/Minute) at different HBR & VPS settings

HBR\VPS	60mL	70mL	80mL
60/Minute	55.56	55.56	55.56
70/Minute	70.56	70.56	70.56
80/Minute	81	81	81
90/Minute	90.6	90.6	90.6
100/Minute	97.8	97.8	97.8

3.5 Calibration

Considering the micro-controller only has limited computing speed and it has been working on its limit, the calibration algorithm has to be simple and effective in order to avoid affecting system performance. The calibration will focus on VPS as the error of HBR is already acceptable (lower than 10%)

Given that the actual VPS shows linear relationships with current VPS setting on certain HBR setting, a set of linear calibration formulas can be involved to adjust the actual VPS settings.

Given that the errors of VPS show different relations on varies setting of HBR, separating and calibrating them on different range of HBR is no doubt an easy way. Eq (9) will be used when VPS setting is below 65;Eq (10) will be used when VPS setting is in a range of 65 to 75;Eq(11) will be used when VPS setting is over 75.

$$VPS_{ACT} = VPS + \frac{3}{20}HBR - 15 \tag{9}$$

$$VPS_{ACT} = VPS + \frac{11}{40}HBR - 23 \tag{10}$$

$$VPS_{ACT} = VPS + \frac{21}{40}HBR - 38 \tag{11}$$

By comparing the test result of before and after applying calibration method, which is shown in Fig.11, the system error of VPS is decreased to 4.5% from the original 14%. Therefore, the calibration method has largely increased the output accuracy of VPS.

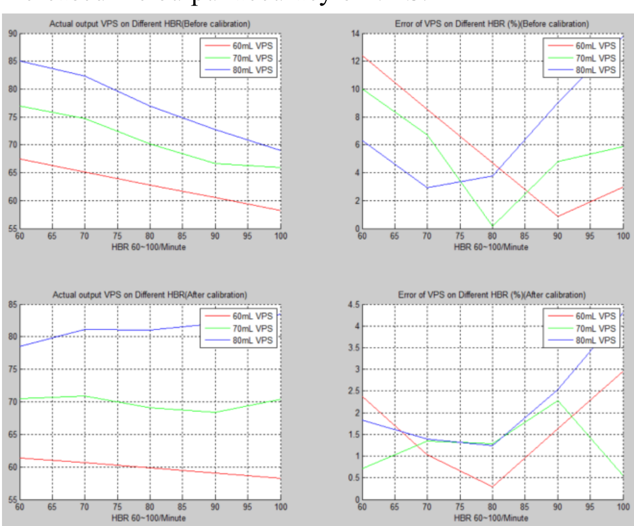

Figure11 Result of calibration (100 cardiac cycles)

3.5 Error analysis and future development

Error on output VPS may due to varies reasons. According to test result, the over-charge of piston occurs under every HBR setting, and it wasmuch severer on low HBR setting than high HBR setting.Given that PID parameters Kp, Ki and Kd(Proportion Integration and Differentiation, which are used to adjust the moving curve of the system)need to be adjusted according to specific system, it is possible that they are needed to be adjusted according to different HBR setting. Meanwhile, it is also feasible to compensate VPS loss by deploying interpolation compensation method. Additionally, replacing PID controller with fuzzy controller is also a possible solution.

According to test result, error on HBR is due to delay of UI control. When the update of UI was disabled, HBR accuracy was largely improved. A possible explanation could be that the communication between MCU and touch-screen was affecting the piston control. Because the touch-screen is controlled via UART port, it needs 9~11 byte of data to draw a dot in the drawing area. In this case, even it is setting on the highest baud rate (115200bps), it still takes at least 781~955us to draw each dot (depending on its certain value). However, the interval of piston control was in a range between 300~500us. It is no doubt that these strenuous UART communication were affecting the process of piston control, even Arduino system has multiplexing function. In this case, replacing the Arduino system with a faster MCU or FPGA system can be considered to solve this problem.

4. Conclusion

According to data collected from experiments, this device basically meets the functionality requirements. It is now able to output water flow in a desired physiological interval and pattern, but it still requires further development to minimize VPS error and compensate VPS loss on different HBR setting. Meanwhile it is worthy to mention that bubbles inside the system can massively affect the performance of this device, especially the output of VPS. Therefore, it is necessary to fully load the system with water or simulated blood before conducting any further experiments.

Acknowledgements

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