Abstract

Our Mechanical Heart Model (MHM) is a device that allows for the simulation of human heart function without biologic restraints. Currently, students in BIME 492 are forced to use themselves as subjects when experimenting with heart function. A MHM will allow these students to physically study the heart under different conditions. This allows for the students to gain a more solid foundation in understanding basic heart function. Current MHM devices are too expensive and complex for use by these students.

Introduction

The goals of this project are to explore currently marketed MHM devices. Specifically, we will attempt to better cost efficiency and operational simplicity, as well as a design that clearly displays heart subsystem function to the user. The expected result is a practical prototype that can model various blood flows of a human heart, with the eventual goal of being reproducible and used in RIT BIME labs. This prototype design must adhere to current RIT BIME lab resources and needs.

Solutions to our project currently exist on the commercial market, but are unfortunately too complex & expensive for what the RIT BIME department is looking for. These models are capable of producing an adjustable, pulsatile pressure & flow like that of the human heart. To aid in our design of our MHM, various specifications of these models have been used to compare them to our own expected design.

Design Process

The design of this heart model must use the hydraulic and pneumatic systems available in RIT labs and display clear mathematical models of heart function under various physiological conditions. The device must be safe for use, meaning the use of a non-hazardous fluid, precautions for consistent use, and a size feasible for use on a lab bench are imperative. LabVIEW is currently used by the target students in their current lab courses, which requires this model to also be controlled by LabVIEW. Most importantly, the heart model must properly interface with the systemic circulatory system designed by project group P16081. A budget of $700 was assigned to this project with a few exceptions being made for certain unexpected expenses, specifically the need for new pressure sensors.

Other than the choice of a membrane to replicate the heart, the logic behind the system design
remained relatively consistent throughout the design process. In the end, a rolling diaphragm was chosen to replicate the pumping action of the heart due to its ability to be easily controlled. A spring and piston plate are used to provide the force behind the pumping action, while a two sided chamber is used to separate the pneumatic and hydraulic components of the model. The device has various control mechanisms, including a pressure regulator, 3-way valve, and controlled release valve. A LabVIEW VI is used as the interface for the user to operate these controllers, allowing the system to operate as they choose.

Figure 1: Heart Model System Architecture

**Pump Design**

Extensive benchmarking was carried out to assess the best design that could behave similarly to a human heart while meeting two major customer requirements, pneumatic power and hydraulic circulation. Commercial models employed in industry commonly utilized electric piston pumps or portable, pulsatile, pneumatic pumps paired with left ventricular assist devices (LVAD). Additionally, we looked at more mainstream piston/diaphragm systems commonly employed in truck air-braking systems, which were designed to handle two different pressures on either side of the diaphragm. We then looked past commercial models to experimental models. One model employed a polyurethane ventricle/atrium paired with a commercial device that used wall air as an input and output pulsatile airflow. The other model utilized a 3D molded silicone ventricle paired with a pressurizable chamber to induce hydraulic flow. Based on designs and morph charts generated by the team, the most common issues with many of the designs was difficulty of modeling expected outputs. The LVAD, 3D mold, and polyurethane materials proved difficult to relate and fit into fluids calculations due to their nonlinear properties, whereas a piston/diaphragm paired with a spring proved simple to model due to its linearity. Therefore, the final design was based on an airbrake, as it could drive hydraulics via pneumatics with a simple piston/diaphragm/spring combo.
**Fluids Calculations**

A large portion of dimensioning and part selection were dependent on the fluid mechanics calculations derived using the engineering requirements and desired system outputs. The two main components of the fluids calculations were 1. The **required pump head** & associated head loss, and 2. The **hydraulic to pneumatic relationship**, where the required air pressure could be calculated based off of the desired output fluid pressure.

**REQUIRED PUMP HEAD:**

The primary formula used to calculate the required increase in pump head was Bernoulli’s equation:

\[
\frac{P_3}{\rho g} + \frac{V_3^2}{2g} + z_3 = \frac{P_2}{\rho g} + \frac{V_2^2}{2g} + z_2 - h_f + h_p
\]

*Figure 3: Example Bernoulli’s equation used to calculate \( h_p \) (pump head), where \( P \) = pressure, \( V \) = velocity, \( z \) = height, and \( \rho \) & \( g \) = density and gravity, respectively.*

Using this equation, along with the given maximums / minimums that our system was required to reach, and the given inputs expected from P16081, an estimated required hydraulic pressure increase, \( h_p \), was calculated.

**HYDRAULIC-PNEUMATIC RELATIONSHIP:**

The next step in verifying the feasibility of the design was to calculate / estimate the required pneumatic air pressure in order to step up the fluid pressure to the value previously calculated (\( h_p \)), while still meeting requirements of flow rates and frequencies. In order to do so, a simple free body diagram (FBD) was constructed to provide a simplified model of the system:
This FBD simplified the design to have 3 forces acting on it: Pressure force of the pneumatic air, spring force of the compression spring, and \( P_h \), the hydraulic pressure force increase that would be required to take place within the fluid. Using the simple relationship that Force = Pressure * Area, where the area in this equation was simply the area of the diaphragm (since that is the area all pressures would be directly acting upon), the 3 forces could be balanced in the \( Y \) direction. Solving for \( P_{air} \) would give the equation used to calculate the expected required pneumatic air pressure.

\[
P_{air} = \frac{V_{work} K}{(\pi r^2)} + h_p
\]

This value was critical in ensuring all parts, such as the pressure regulator, end caps, 3 way valve, acrylic tubing, etc, were designed to properly work with our expected air pressures. This equation was also critical in determining the spring constant. In order to spec out what spring would be best for the design, different \( k \) values could be input to the equation, and the required \( P_{air} \) value could be calculated and checked for feasibility. For example, if a \( k \) value was put into the equation and a \( P_{air} \) value of 150 PSI was calculated, then it would be too large of a spring constant, as the system would have to be highly pressurized to compressed the spring at that \( k \) value.

**Diaphragm Selection**

After developing the model for the pump, the next step was to find an appropriate diaphragm that could reliably act as an interface between the hydraulics and pneumatics. First, air brake diaphragms were tested to determine their effectiveness, but they were found to be too thick and inflexible. After further research, rolling diaphragms were found to be the best option. These diaphragms were designed to remain in contact with both the piston and the chamber walls while under pressure, ensuring a predictable travel path and subsequent volume change. Specifically, top hat designs were designed to work with pistons with long travel distances and large volumes. Two 4-450-275-FCJ fabric reinforced, top hat rolling diaphragms were ordered from Bellofram Corp. The dimensions were 4” piston bore and 4.5” cylinder bore with a 1.97” half stroke. This diaphragm acted as the interface between the pneumatic and hydraulic sections of the chamber.

**Chamber Design**

The chamber was designed based on the requirement that the system be responsive to changes in the circulatory system. As such, the flexible diaphragm/piston/spring design was chosen because it would output the same pressure regardless of system resistance. Based on the membrane, available we designed a two-sectioned chamber. One end would supply the pneumatic force to drive the pumping action and the other half would hold the working fluid.

**Diaphragm Assembly**
The diaphragm assembly consisted of the diaphragm, a 3D printed piston plate, and spring to help linearize the motion of the diaphragm. The piston plate had a 4.5” diameter and a support for the spring. The spring itself had a k value of 150 PSI/in and had a length of 1.75”.

![Image](image1.jpg)

*Figure 5: This drawing shows the diaphragm, spring, and plate assembly.*

**Hydraulic Assembly**

The piston and spring were housed in the hydraulic side of the chamber with the spring sitting in a retainer cup in the hydraulic end cap. A prototype of the end cap was 3D printed but was determined to be too flexible and too porous. New aluminum end caps were designed to compensate for the high pressures and sealing issues. The hydraulic cap has two through holes which hold two uniseal fittings. These fittings are used to connect to one-way valves, one output which feeds pressurized water to the circulatory system and one input which returns working fluid from the circulatory system.

![Image](image2.jpg)

*Figure 6: The drawing on the left shows the original design for the hydraulic interface, and the photo on the right shows the realized design.*

**Pneumatic Assembly**

The pneumatic end cap housed the IR sensor, the pneumatic intake valve fed by the pressure regulator and 3 way valve, and the emergency pressure relief valve. The relief valve ensured that the pressure in the chamber would not exceed 25 psi. The Pneumatic end cap did not experience the same pressures and did not have the same sealing issues, this cap will use the 3D printed material from the prototype parts.

**Material Selection and Hardware**

The chamber walls are made of acrylic for visibility as well as ability to hold the required pressures. The acrylic tubing was sealed to the end cap with ethyl-butyl tape and had been tested to be both watertight and airtight. The two sides of the chamber are held together with 4 threaded rods through both end caps.
These rods provide the alignment of the chamber as well as the clamping force to keep the diaphragm in place and sealed tight.

**Pressure Regulator**

The customer asked that the device utilize wall pressure to drive the heart. A pressure regulator was required to limit and control the wall pressure to the pressures described in the hydraulics calculations. An electronic pressure regulator was necessary to completely control the system via LabView, as per the engineering requirements. Even a single electronic pressure regulator proved to be expensive and would have consumed a large percentage of the budget, but we had a decommissioned Conoflow GT2108ED Pressure Regulator donated to us by FMS. This pressure regulator required an input current of 4-20mA to get an output of 3-15 PSI. It was rated to handle a 100 PSI supply, and used 1/4” pneumatic inputs and outputs.

**3 Way Valve**

The three way valve was key to the operation of the pump, as it was the device that pressurized and depressurized the pneumatic chamber. It was important to have a valve that could be electronically controlled via LabView to automate the pumping action. Initially, we purchased a 12VDC valve with 1/8” ports, but after testing, the valve was found to be too slow in both pressurizing and depressurizing the chamber. As a result, a decommissioned valve rated for much higher pressures was donated to us from FMS. The new valve, the Johnson Controls V11HAA-100 Air Valve, required 120VAC at 60HZ to operate, and the valve was rated at a maximum pressure of 30 PSI. Safety concerns were brought up due to the high power requirements for the valve. An RIB PSH75A Transformer was acquired to address this issue, as it was a grounded 120V power supply with a breaker. It also provided a convenient set of 24V terminals for powering a heavy duty relay box.

**Pressure Sensor**

Since the heart pump worked in conjunction with P16081’s circulatory system, the pressure inputs and outputs had to be constantly monitored to ensure the simulated systole/diastole equalled approximately 120/80 mmHg. The customer originally asked to use PASCO PS-2181 Dual Pressure Sensors, but they were incompatible with LabView, another customer requirement. The customer preferred using LabView over the PASCO sensors, so we researched alternative sensors. After researching 5V pressure sensors, we found that we needed liquid compatible, gauge sensors to measure the hydraulic pressures, and the Honeywell TruStability HSCMRNT005PGA5 fulfilled those requirements. These sensors were connected to an NI USB-6215 DAQ for both data input and 5V supply. The sensors were configured to measure four locations in P16081’s circulatory system.

**Flow Meter**

A flow meter was originally purchased as a means to track flow and measure its profile over time, as per the customer’s requirement. These requirements reflected a desired overall systemic responsivity of less than 30s and a desired “real time” responsivity. By extension, these engineering requirements demand an accuracy that can return the desired output values expressed by the customer. We purchased an FS-8800H Futurlec Flow Sensor, an impeller based flow meter with a hall effect sensor.

After testing, however, we found that the sensor was only able to read an average flow over time. The hall effect sensor could not resolve instantaneous flow, which was necessary for determining a flow profile during a single “heartbeat”. The flow sensor was still incorporated into the system as a metric to compare to instantaneous flow.
**IR Sensor**

After researching several ways of measuring instantaneous flow in the chamber, we decided on an infrared sensor to monitor the distance that the diaphragm traveled in the chamber. Knowing the surface area and distance traveled, the volume displaced per stroke could be determined. One could then find the flow based on a given frequency. We chose the Sharp GP2Y0A41SK0F Short Proximity IR Sensor for its ability to resolve a low minimum distance. The IR sensor was placed on the inside of the pneumatic cap and connected to the NI USB-6215 DAQ for both input and 5V supply. The response time of the IR sensor fell well within the engineering requirement of 100ms.

**Relays**

Relays were necessary to switch the 3 way valve on and off for pressurizing/depressurizing the pneumatic section of the heart chamber. They also enable control of the valve via the LabView interface. We purchased a 2 Channel 10A 30VDC Relay Board to operate the valve, but after selecting the Johnson Controls 3 Way Valve, we realized that the power requirements for the valve would overload the relay. As a result, we acquired a second relay rated to handle the valve’s power requirements.

Donated from FMS, the RIB RIBU1C Relay Box could handle 10A 120VDC and was able to supply the necessary amount of power to the valve. The RIB relay, however, required a higher signal voltage (10-30VDC) that could not be powered by the NI DAQ. Therefore, the smaller 2 channel relay paired with 24V from the transformer was used to connect the NI USB-6215’s 5V signal output to the RIB relay. The response time of the relays + valve fell well within the engineering requirement of 100ms.

**Voltage to Current Converter**

The pressure regulator required a 4-20 mA signal current to operate correctly, and there was no means to produce a signal current directly from the NI DAQ. Therefore, the only controllable output from the DAQ, 0 to 5 volts, had to be converted to a current. We used a Burr Brown OPA705PA op-amp to convert the voltage to a current. The device employed an inverting op-amp circuit where the variable voltage signal was applied to the input of the op-amp. A 100 ohm resistor was connected to ground and the negative lead of the op-amp, and the output was connected to the load resistance (the pressure regulator) and ground as well. This layout would allow for a current output proportional to the voltage input, and the current output would not be influenced by the load.

**LabView**

From our customer requirement our system had to be controlled by LabVIEW in terms of determining the outputs of the controls and build a simple interface. Therefore, the front page of the LabVIEW program has a manual and automatic control set-up. In the manual set-up the user is allowed direct control of the pressure being applied to the system and control of the three way valve. In automatic control the user has control of pressure being applied and the frequency of the system in beats per minute.

There was also an attempt at creating a while loop within the Labview code to read the input voltages of P16081 pressure sensors; but lack of time did not allow us to have it run correctly and have accurate readings from the systemic system. Ultimately had we had more time we would have found a way to incorporate their pressure sensor readings in our Labview and have a single control and measuring Labview interface. For further understanding and an example of our working Labview model refer to Appendix A.

**Interfacing with P16081**

The system was physically interfaced with P16081 using ½” Tygon tubing and barbed fittings. The
heart pump has permanently attached barb fittings that are press fit into the ends of the one way valves. One of the biggest issues created with this interface is that there exists no good way to prime the heart pump. In order to prime the pump, the pneumatic side of the chamber needs to be pressurized to keep the membrane from blowing out into the pneumatic side if the hydraulic pressure becomes too high. Once the system is pressurized, the circulatory system had to be lifted above the heart pump to release any air in the system and to completely fill the hydraulic chamber with water. Another aspect of the interface between P16080 and P16081 is the pressure transducers that read pressures in the tygon tubing before and after the circulatory system. The pressure transducers are read by the same DAQ as the rest of our sensors and controls and sent into a combined Labview program.

Results and Discussion

The main goals of this project included meeting the customer’s requirements of simulating and measuring physiologically relevant flow rates, pressure profiles, and frequencies. The final product produced physiologically relevant pressure values for the aortic and venous parts of the circulatory system. Initial flow testing showed that the system could not output the required flow, but this test was not done with the complete circulatory system. Estimates show that the system would come close to meeting flow requirements if atrial preloading was simulated in the circulatory model. Increasing the frequencies resulted in depressurization issues that limited stroke volume. This issue could be easily solved by installing a quick release valve.

Figure 7: The chart on the left shows the pressure profile generated from the combined heart pump and circulatory system. The chart on the right shows the desired volume output at each stage of the heart beat cycle.

The customer required that P16081’s circulatory system be hydraulically driven by the heart pump, and the pump had to be powered by compressed air. The final product met these requirements, as seen by the chamber function. The system inputs and outputs were also required to be controlled by LabVIEW. The final product met these requirements, as shown by the GUI. The budget had to stay under $700. The final product met these requirements, as shown by the BOM. The device setup, use, and tear down also had to fall within a three hour lab period. The final product was not formally tested for this requirement, but initial testing showed that setup, operation, and tear down would fall well within the three hour requirement.

Conclusion and Recommendations

Unfortunately, the pump was not capable of reaching certain ranges of numbers, such as stroke volume, frequency, and flow rate. However, with slight modifications it is reasonable to believe reaching those numbers is possible. One issue the pump experienced was not exhausting pneumatic air fast enough in
order to fully expand the hydraulic chamber to reach maximum stroke volume. A possible solution is hooking up the chamber to a vacuum source to exhaust the air faster which would create larger stroke volumes and higher potential flow rates. Another possible modification the pump could use is installing a tab of some type on the inside of the pneumatic chamber for the diaphragm to rest on when the air side is not pressurized. This would stop the diaphragm from blowing out when the system is filled with water and not pressurized on the pneumatic side. This was a consistent issue experienced during the last couple’s weeks of build & test. Ultimately the team was able to build a functional mechanical heart pump that can be used with P16081 to model some mathematical models of the circulatory system and demonstrate specific heart conditions in a laboratory setting.

References

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Appendix A

Figure 8: LabVIEW Interface
Figure 9: Manual Control LabVIEW Block Diagram

Figure 10: Automatic Control LabVIEW Block Diagram
Figure 11: Data Acquisition LabVIEW Block