Ultrasound-Compatible Cardiac Simulator

by

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ES 100 Senior Capstone Engineering Project Report submitted to the School of Engineering and Applied Sciences in partial fulfillment of the requirements for the degree of Bachelor of Science (S.B.) in Mechanical Engineering at Harvard University Spring 2015

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Abstract

Devices are being developed so that robots can perform cardiac surgery. These robots will steer catheters through a beating heart using image guidance from ultrasound imaging. An appropriate ultrasound-compatible cardiac simulator has been developed in order to test these systems. This simulator mimics the geometry and the complex motion of a beating human heart. The left ventricle phantom is comprised of a soft echogenic material that mimics heart tissue in an ultrasound image. The cardiac simulator will be a useful tool for research and industry applications where testing on an ultrasound compatible phantom is desirable.
1. Introduction

1.1 Motivation

Figure 1. Catheters can be introduced to the heart to perform procedures in any of the four chambers.

Every year, approximately 5 million people undergo cardiac surgeries [1]. Many of these clinical procedures involve stopping the heart and rerouting the blood with a cardiopulmonary bypass machine, which is then pumped to the rest of the body. This is critical for performing life-saving cardiac surgeries, but also introduces additional health risks to the patient. Cardiac catheters provide a minimally invasive treatment alternative to certain procedures. Cardiac catheters are beneficial because simple
procedures can be done while the heart is still beating. However, catheters are limited to simple tasks that do not require dexterity. To improve cardiac catheter technology for a wider range of cardiac procedures, cardiac catheter robots are being developed to perform beating-heart surgeries. These robots are designed to compensate for the motion of the heart as it beats throughout the procedure. The continued validation and improvement of these robots and the controls that steer them is necessary in order to make them reliable for a clinical setting. Having a reliable environment in which to test these devices is needed for their development.

“Beating-heart” phantoms are not readily available and commercial models have significant limitations. There is a need in the medical device community for test subjects that accurately replicate the interior geometry of the heart as well as the motion experienced during the heartbeat. In addition, a simulator that can be easily imaged with ultrasound will allow researchers to refine the algorithms necessary for steering robotic catheters. This report describes the design, prototyping, and testing of a cardiac-motion simulator that will be a pivotal test rig for robotics researchers to improve cardiac catheter technology.

1.2 Prior Art

Some existing cardiac-motion simulators have been previously developed; however, there remains room for improvement. The phantom specifications needed by the BioRobotics Laboratory do not currently exist in commercial or in research prototype forms. Some solutions include models that mimic fluid flow or the electrical signals of the heart but neglect the displacement of the cardiac walls. There has been some research and development into simulators that mimic that motion of
the heart, but they have significant limitations that make them unsuited for developing beating-heart surgery robots.

The commercially available Ultrasound and CT-Compatible Multi-Modality Heart Phantom by Shelly Medical Imaging Technologies is the only commercially available cardiac motion simulator [2]. This simulator actuates a combined left and right ventricle phantom by compressing and twisting the phantom with an actuator attached at the apex. The phantom is made of a polyvinyl alcohol hydrogel that mimics cardiac tissue when imaged via ultrasound or CT. While this simulator has one of the most realistic phantoms available, there are other significant limitations that prevent this simulator from being used to develop surgery robots. Since this simulator is designed for only the ventricle phantom provided, it is not possible to perform test procedures in either of the atria. Also, the interior of the phantom in this simulator is accessed by a small port, which prohibits any visual inspection of the interaction between the phantom and cardiac-intervention tools. The simulator presented in this paper will be designed such that cardiac tools can easily be introduced to the inside of the cardiac phantom, which will be a great benefit to researchers who are in the early phases of developing full motion compensation catheter robots.
Another novel approach to simulating the complex twisting and compression motion of the heart is “A Bioinspired Soft Actuated Material” by Roche et al [3]. This device is intended to be a ventricular assist device that uses soft actuators to compress the heart. These soft pneumatic actuators are imbedded in a rubber matrix at an angle of -60° from the horizontal which corresponds to the angle at which the muscle fibers are oriented in cardiac tissue. This device does not address many of the functional requirements that are needed in order to properly develop and verify cardiac catheter robots. This system only allows one degree of freedom, which does not make it possible to create the motion of other parts of the heart. The difference in density between the actuator material and the matrix in which they are embedded makes this device unable to simulate cardiac tissue when imaged via ultrasound.
Figure 3. A Bioinspired Soft Actuated Material [3]
A third device that simulates the motion of the heart is the heart simulator developed at the University of Leeds [4]. This simulator uses metal leaf springs inside a soft cardiac phantom to create the motion of the exterior of the heart. There are many factors that make this simulator ill-suited for the application described previously. One of these aspects includes not being echogenic due to the metal part inside that phantom and the fact that the simulator cannot be submersed in a liquid, which is necessary for ultrasound waves to propagate.

![University of Leeds heart simulator](image)

**Figure 4. University of Leeds heart simulator [4].**

The chart below lists these and several other prior existing cardiac simulators and their features and limitations.
Table 1. Cardiac Simulator Prior Art

<table>
<thead>
<tr>
<th>Product</th>
<th>Part of Heart</th>
<th>What is Simulated</th>
<th>Target Market</th>
<th>Limitations</th>
</tr>
</thead>
</table>
| Low-Cost, Take-Home, Beating Heart Simulator for Health-Care Education [5] | Full heart    | Outer motion is simulated using solenoid actuation inside the heart model.        | Medical students interested in studying arrhythmia | • Electrical components inside heart, therefore can’t be put in water tank  
• Can’t introduce a catheter inside the phantom |
| Ramphal Cardiac Surgery Simulator (RCSS) [6]                            | Full heart    | Outer motion is simulated by inflating balloons in each ventricle.                | cardiac surgeons                     | • Uses a porcupine heart. Balloons in the ventricles would interfere with the catheter.          |
| Harvey the Cardiopulmonary Patient Simulator/Patient Simulator K [7]    | Human phantom | The patient's body is simulated with sounds and possibly pulse. Some models also have capability for producing EKG signals | doctors, nurses                      | • No physical simulation of heart motion or geometry  
• Directed towards patient diagnostics. |
| Design of an anatomically accurate, multi-material, patient-specific cardiac simulator with sensing and controls [8] | Full heart    | Phantom is connected to a pump, simulating the flow rate and the motion of the heart. | research                            | • Very complex method to build the heart phantom.  
• Not made of echogenic material  |
| 4D-Analysis of Left Ventricular Heart Cycle Using Procrustes Motion Analysis [9] | Left ventricle | Computer modeling of shape and displacement                                      | Researchers studying healthy left ventricle motion | • Not a physical simulator |
| Hardware-in-loop-simulation of the cardiovascular system, with assist device testing application [10] | Left ventricular assist devices | The motion of one band around the atria and computer feedback                     | research                            | • No phantom included  
• Only a heart assist device |
| Ultrasound & CT Compatible Multi-Modality Heart Motion Phantom [2]       | left and right ventricle | Ventricular motion and some fluid flow is simulated.                | anyone doing testing                  | • Limited to the geometry/motion of the combined left/right ventricle phantom |
| A Bioinspired Soft Actuated Material [3]                               | twisting and contracting motion of the heart | twisting and contracting motion of the heart | research                            | • Doesn’t use echogenic material |
2. Design Goals

In order to create the most effective simulator for developing and validating cardiac catheter robots, it is necessary to define specific requirements that must be met. These requirements largely fall into two categories—laboratory requirements and anatomical specifications. By combining these two categories, the best possible simulator was developed.

2.1 Anatomical Requirements

Since this simulator is an analog for the human heart, it must meet certain specifications to accurately represent the heart in a research setting. Most importantly, the heart phantom must mimic the geometry and motion of the human heart. Since most cardiac procedures only focus on a single heart chamber during a given procedure, the simulator was designed to have interchangeable phantoms for each of the heart chambers. The left ventricle has the largest displacements and velocities during a heartbeat, so the choice to design the simulator to left ventricle specifications means that the simulator will be able to accommodate the motion of the other heart chambers. Cardiac tissue also has a characteristic appearance when imaged via ultrasound—which is the method of imaging used to guide certain cardiac catheter robots—and the phantom material must replicate this feature.

2.2 Laboratory Requirements

Many of the requirements for this simulator come from the laboratory setting that it will be used in. Ultrasound imagers propagate sound waves through fluids such as water and blood. Therefore, in order to properly ultrasound image the phantom while it is moving, it will be necessary for it to be submerged in a water bath. This
introduces the need to isolate any electronics from the water bath. Another important sensing tool used in conjunction with the catheter robots is electromagnetic tracking. To prevent interference with the electromagnetic tracking, there must not be any metal components in the area that is being observed. Therefore, any components of the simulator that are near the phantom must be made of nonmetallic material. The development stage of any complex system requires many hours of testing. Consequently, the simulator must be robust enough to last through many hours of testing.
3. Design Approach

The design approach of this project involved three main segments. First, collect and analyze data pertaining to heart geometry and motion. Second, create an anatomically accurate, echogenic phantom that can be actuated in a way that mimics the cardiac motion. And finally, build an actuation system and structure that will move said phantom within a water bath while safely isolating any electronics.

3.1 Cardiac Motion

The information below describes normal cardiac motion that was used to inform the design of the simulator. The success of the simulator relies on how well the simulator motion and geometry matches that of the human heart. The left ventricle is the largest chamber in the heart and as such has the largest displacements during the heartbeat. By meeting the demanding specifications of simulating the left ventricle, it follows that other heart chambers can be accurately represented, since they will have smaller displacements and slower velocities. There has been extensive research into the motion of the heart in healthy and diseased patients. This data along with a data set provided by Cavusoglu et al. (Case Western Reserve University, USA) of a measured calf heart beat was used to recreate the motion of the heart beat in a simulator.\[11\] This data has allowed for the proper recreation of the displacements, velocities, and accelerations found in the heart beat. A simulator that can recreate these important parameters of the heartbeat is a suitable test platform for developing robotic cardiac intervention tools.
Figure 5. Plot of calf heartbeat displacement trajectory.

Figure 6. Displacement v. Time for calf heart data.
Table 2. Characteristic Motion of the Left Ventricle

<table>
<thead>
<tr>
<th>Source</th>
<th>Displacement (mm)</th>
<th>Velocity (mm/sec)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Assessment of myocardial velocities in healthy... [12]</td>
<td></td>
<td>98</td>
</tr>
<tr>
<td>Cavusoglu [11]</td>
<td>10.3</td>
<td>100</td>
</tr>
<tr>
<td>Caliskan 2012 [14]</td>
<td></td>
<td>96</td>
</tr>
<tr>
<td>Kapusta 2000 [16]</td>
<td></td>
<td>80-100</td>
</tr>
<tr>
<td>Mori 2000 [17]</td>
<td></td>
<td>40-90</td>
</tr>
<tr>
<td>Schechter 2006 [18]</td>
<td>4-26</td>
<td>30-130</td>
</tr>
<tr>
<td>Yamada 1998 [19]</td>
<td></td>
<td>50-70</td>
</tr>
</tbody>
</table>

The data taken from this literature refers to the left ventricle and can be applied directly to the design of this simulator. From the available literature on cardiac motion, two things are apparent: (1) there is a wide range of displacements and velocities experienced in the heartbeat and that (2) most of these velocities are similar to those calculated from the Cavusoglu heart data. Due to their similarities, it makes sense to use the Cavusoglu data set to model the simulator.

By analyzing this data it was possible to find characteristic measurements for cardiac velocities as well as the size of heart chambers. Caliskan measured the left ventricle wall motions of healthy patients and found that these values fell between 73 and 96 mm/second. This is consistent with Kapusta who sited left ventricle wall velocities between 80 and 100 mm/second.

3.2 Phantom Mold Geometry

The phantom used in this simulator will replicate the geometry of the left ventricle. This phantom will be created by fabricating a mold that has an anatomically
realistic interior and then casting an echogenic material into this mold. Any materials of differing densities will be isolated away from the interior of the phantom chamber in order to prevent any unrealistic scatter when imaged via ultrasound.

Since interchangeable heart-chamber phantoms are necessary, the technique used to mold each phantom needed to be easy to translate across multiple complex geometries. Molds utilized in many applications use multiple parts allowing for easy assembly. Then the material to be cast is poured into the mold and once cured, the mold can be disassembled and the phantom released. The phantom used in this simulator was made by this method. This allows for repeatable results when making multiple phantoms. Using three-dimensional printing technology (Dimension 1200es, Stratasys, USA) to fabricate the mold for the cardiac phantom made creating complex geometries possible. This technique also allowed for the introduction of multiple materials into the phantom so that some portions were rigid and others compliant.

3.3 Actuation Method

The phantom described previously will then be mounted to a supported structure and connected to an actuation system to create the motion of the heart. The top of the simulated heart chamber must be kept clear of actuation equipment (cables and pulleys) in order for cardiac catheter robots to be properly introduced. Therefore the actuation mechanism and the support structure must be designed to increase the workable area above the chamber. This structure supports the phantom while keeping it submerged in a water bath—which is the medium used for ultrasound imaging—while also supporting the other main components of the system.
The proper actuation method needed to be chosen. Many potential methods of creating the complex twisting and compression motion of the heart were evaluated. These included cables, pneumatics, and shape memory alloys. Pneumatics and cable-driven approaches were favorable methods due to the relatively large displacements necessary and their compatibility with water.

![Figure 7. Sketches of potential actuation designs.](image)

Eventually, a pneumatic system was ruled out due the need to introduce a fluid that would have a different density then the surrounding phantom. This would cause inaccuracies when imaged via ultrasound. Cable-driven actuation remained as the best candidate. This approach had many advantages, including the wide range of actuators that could be used as well the simplicity of transmitting power in many directions using pulleys. The downsides of cable-driven actuation include maintaining
proper cable tension, friction in pulleys, and assembly effort. These issues are addressed during the design of the actuation system. Since the motion of the heart is characterized by two major motions—lift and twist—the actuation system is comprised of these two degrees of freedom. Actuation is executed by electric motors, which must be isolated from the portion of the simulator that is submerged in the water bath.
4. **Design Evolution**

Due to the highly integrated nature of the two major components of this project, the geometry of the phantom and the mode of actuation were iteratively improved in parallel. Once the proper phantom geometry and mode of actuation were settled upon, it was then possible to incorporate these components in to a full system architecture that met the prescribed requirements.

4.1 **Echogenic Materials**

An important requirement for the simulator is that it is made of a material that mimics heart tissue when imaged via ultrasound. Materials of this nature are often referred to as echogenic materials and have been studied previously. In “A Review of Tissue Substitutes for Ultrasound Imaging,” Culjat et al. describes many of the potential materials that can be used to replicate human tissue [20]. This research is further broken down into specific tissues. Several materials are listed as being suitable substitutes for cardiac tissue. Three of these materials—gelatin, silicone rubber, and a polyvinyl alcohol hydrogel—are the best potential materials for the simulator due to their ability to hold their shape after being molded into a complex geometry.

Gelatin mixed with sugar-free fiber supplement is one of the commonly used materials used to simulate cardiac tissue. This gelatin material is easy to create and uses supplies that are readily available on grocery-store shelves. The addition of the fiber supplement creates a slight variation in density throughout the material providing the properties necessary to image it via ultrasound. The problem with this material is that it is not shelf-stable once mixed and molded into the desired shape.
Prolonged exposure to ambient air will dry the gelatin out while long periods in humid environment will promote microbial infestations that will cause rotting or mold growth.

Silicone rubber is a material commonly used in making molded parts. Silicone comes in a variety of densities, which makes it easy to match the elasticity to that of the tissue being mimicked. By adding glass microspheres, some researchers have been able to simulate cardiac tissue. This process, compared with the gelatin, is much more difficult to reproduce, due to issues in creating an even distribution of the microspheres while the rubber cures.

A hydrogel made from polyvinyl alcohol is another promising material. By creating an aqueous solution of 10% polyvinyl alcohol, Lesniak-Plewinska et al. was able to mimic the elasticity and sound propagation properties of cardiac tissue [21]. The solution is heated to 80°C while mixing then cooled to room temperature before pouring into the desired mold. To cross-link the polymer, the solution is then frozen at -25°C for 24 hours and then thawed at room temperature for 24 hours. These freeze-thaw cycles determine the elasticity and echogenic properties of the phantom material. The main drawback of this material is the complex process and the relatively lengthy procedure required to make a single phantom.

The best material is one that will last for thousands for cycles while also exhibiting the proper imaging qualities. Below is a comparison of various phantom materials.
Table 3. Comparison of Cardiac Phantom Materials

<table>
<thead>
<tr>
<th>Material</th>
<th>Echogenicity</th>
<th>Durability</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gelatin w/ Fiber Supplement</td>
<td>+1</td>
<td>0</td>
</tr>
<tr>
<td>Silicone rubber</td>
<td>-1</td>
<td>+1</td>
</tr>
<tr>
<td>Polyvinyl alcohol hydrogel</td>
<td>+1</td>
<td>+1</td>
</tr>
</tbody>
</table>

As previously mentioned, many materials have been developed to mimic cardiac tissue when imaged via ultrasound. For this simulator the best materials include gelatin with fiber supplement, silicone rubber, and polyvinyl alcohol hydrogel. Since getting the proper scattering properties with silicone rubber is difficult, silicone was only used as a prototyping material. The Young’s Modulus for cardiac tissue as well as the phantom materials is compared in the chart below.

Table 4. Young's Moduli for Cardiac Phantom Materials.

<table>
<thead>
<tr>
<th>Material</th>
<th>Young’s Modulus</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cardiac Tissue [21]</td>
<td>117-230 kPa</td>
</tr>
<tr>
<td>Gelatin [22]</td>
<td>36-150 kPa</td>
</tr>
<tr>
<td>Polyvinyl alcohol (PVA) hydrogel [21]</td>
<td>56-138 kPa</td>
</tr>
<tr>
<td>Smooth-On Ecoflex 00-30 silicone rubber [23]</td>
<td>125 kPa</td>
</tr>
</tbody>
</table>

Since the Ecoflex 00-30 silicone rubber is within the ranges of Young’s moduli for cardiac tissue as well as the PVA hydrogel, it is a suitable prototyping material because similar forces will be needed to actuate all of these materials. These three different materials were used throughout the design process since each material has slightly favorable properties for different applications. Gelatin and PVA are both have
excellent echogenic properties. Gelatin, however, is not shelf-stable and can harden or mold over time. PVA requires a complex mixing process and then must be frozen and thawed in the mold in order to cross-link the polymer chains that mimic cardiac tissue fibers. If the PVA hydrogel is not defrosted under the correct circumstances, it will not hold its shape. Additionally, the Young’s modulus for gelatin is dependent on its concentration by weight so any variances in concentration will either stiff or soften the phantom. The Young’s modulus for PVA is dependent on the freeze and thaw crosslinking cycles. A colder or longer freeze will produce a firmer phantom. For these reasons, a different material is needed for prototyping. Silicone rubber is very durable but lacks the small differences in density necessary to scatter ultrasound waves the way cardiac tissue does. For the initial refinement stages of the motion control, using a silicone rubber ensured that the phantom would not be damaged by the actuation system. By using a rubber with a similar Young’s modulus to that of the phantom materials, it was possible to verify that the actuators had sufficient force to displace the phantom.

4.2 Prototype Evolution

4.2.1 Phantom Geometry

Initially, the shape of the left ventricle was approximated as a cylindrical tube. As such, it was easy to create initial prototypes by making simple molds using coffee cups with cylinders inside to create the hollow interior.
Figure 8. Coffee cup mold used to make preliminary prototype.

Then Ecoflex 00-30 silicone rubber was cast into this mold. This was a sufficient design for initial prototypes, but as the system improved so did the geometry of the phantom. This design was then improved by approximating the ventricle as half of an ellipsoid.

Figure 9. Ellipsoid phantom (Ecoflex 00-30) and yellow 3D printed mold.
To fabricate this more complex shape, a two-piece mold was 3D printed. Again Ecoflex 00-30 silicone rubber was cast into the mold. Since this geometry was similar to what the final geometry would be, the forces that were needed to actuate the phantom could be measured in order to determine the necessary actuators to be selected later. Further iteration was necessary in order to accurately represent the anatomical features of the human heart. A tool used to evaluate the geometry of the human heart is a three-dimensional computer drawing of the heart produced by Zygote Media Group [24].

![Figure 10. Zygote Media Group CAD model of human heart.](image)

Five cross-sections spaced 20 mm apart were taken through the left ventricle portion of this drawing. A sketch of each section was subsequently drawn in SolidWorks and then drafted together in order to create an anatomical shape for the mold’s central core.
Figure 11. Cross sections cut from CAD heart model.

Figure 12. Drawings of each cross section in SolidWorks.

Figure 13. Anatomical core for molding the left ventricle.

The two, papillary muscles that are in the left ventricle were created by cutting two cone-shaped cavities into this core. An outer shell was then made using two identical, cylinder halves. This shell securely held the mold core in the proper
position in the center of the mold. The shell halves were then wrapped in plastic-wrap to prevent any leaks at their seams.

Ecoflex 00-30 silicone rubber was again cast into the mold for prototyping. After confirming the proper geometry, the gelatin and polyvinyl alcohol hydrogels were also cast using this mold.

Figure 14. Fully assembled left ventricle mold.

Figure 15. Left ventricle phantom made of Ecoflex 00-30. Red markers added for visualization of twist motion.
Casting PVA involves a more complex process than silicone rubber or gelatin. The aqueous solution of PVA is cooled to room temperature and then poured into the mold. The mold is then placed in a freezer for 24 hours to solidify the PVA. After this freezing period, the phantom is removed from the freezer and left to thaw at room temperature for another 24 hours. During this thawing process, the material must be exposed to ambient air. The thawing/drying process not only cross-links the polymer but also creates a thin skin on the outside of the phantom. This skin produces bright spots when imaged via ultrasound. Thawing the phantom in a sealed container was tested to prevent this unfavorable skin. The sealed thawing process prevented the formation of the skin, however it also prevented the hydrogel from properly gelling. This meant that when the mold was disassembled, the phantom material would not support itself and would instead melt into a puddle.

4.2.2 Actuation Technique

Creating the complex twisting and compression motion of the heart could be completed in many ways. For the many advantages explained previously, the prototyping of actuation methods began with a cable driven approach. This method proved to be effective throughout the prototyping stages, and as such it is the method used in the final prototype. By embedding plastic tubing in the walls of the cylindrical tube phantom, cables could be routed through the tubes and used to create a change in the interior volume of the phantom.
While this motion was promising because a decreased volume of the cavity could be created, the differing density of material in the phantom due to the tubing would make accurate ultrasound imaging nearly impossible. To combat this issue a design was needed that isolated the attachment points for actuation away from the phantom cavity. To do this a second method of actuation was devised. By embedding an acrylic ring in the top 15 mm of the phantom, a rigid actuation point was introduced away from the main ultrasound imaging region. By twisting and compressing the phantom from this point, the motion of the heart could be replicated.
Figure 17. Concentric cylinder phantom with embedded acrylic ring.

This initial prototype could only be actuated in a downward motion due to the lack of any attachment points in the lower portion of the phantom. As the phantom shape transitioned into an ellipsoidal shape, a second acrylic ring was added.

Figure 18. Ellipsoid phantom in the actuated configuration.
Having a ring in the upper and lower extremities of the phantom allowed the lower ring to be pulled toward the upper ring to simulate cardiac motion. These two rings were incorporated into the anatomically accurate ventricle phantom. Indentations were added to the outer case of the mold so that the actuation rings were be in same position every time a phantom was made.

![Mounting and actuation rings inside phantom mold.](image)

**Figure 19. Mounting and actuation rings inside phantom mold.**

This design also allowed for the separation of the cardiac motion into two degrees of freedom: twist and lift. In doing such, the simulator would be able to accommodate phantoms for each chamber of the heart by being able to adjust the amount of lift or twist necessary.
4.2.3 Actuator Selection

Motors were chosen to actuate the simulator since this is an effective method for actuating cables. In order to properly specify and command motors that could actuate the cables quickly and powerfully enough to recreate cardiac motion, velocity and force specifications were measured. Using the calf heart introduced earlier, it was clear that most of the velocities that occur during the heart beat fall within ±80 mm/second.

![Velocity Histogram](image)

**Figure 20. Velocity histogram of calf heart data [11].**

The specification was increased to ±100 mm/second in order assure that any velocity that occurs during the heartbeat could be easily replicated. This value is consistent with the velocity values from the literature described previously.

The force needed to actuate the phantom was determined experimentally. The ellipsoid phantom was used since it was a close approximation of the final phantom. A test apparatus was built using 20 mm 8020 aluminum extrusion. This
frame provided a sturdy support for an acrylic plate that holds the phantom. Pulleys were also added to this frame in order to visualize how actuation cables could be incorporated into a full system. The cables attached to the phantom were actuated using a spring scale and the force measured to create a full displacement of the phantom was 20 N. Many pulleys would be used to route the cables along the desired path, which provides some friction and increases the force required to actuate the system. Additionally, if the phantom material has a higher elasticity than expected due to differences in the batches of material, the phantom will resist displacement more than the measured values. Therefore, a factor of safety of 2 was introduced and the motors were to be selected to pull with a force of 40 N at 100 mm/sec.

Initially, stepper motors controlled by an Arduino Uno microcontroller and Adafruit Motor Shield were tested as a potential solution. These motors were inexpensive and utilized the simplicity of the Arduino environment for control. The cables used for translating rotational motion into linear motion were attached directly to the motor shaft to increase the amount of force that could be applied without over-loading the motors. The diameter of the motor shaft is 6.35 mm so in this configuration the motors will experience a maximum torque of 0.127 Nm. Rated at 0.88 Nm of holding torque, these motors appeared to meet the necessary specifications. The following calculations show the maximum power calculations for the motors as well as the maximum power output from the motor shield.

\[
P_{\text{motor}} = \tau \times \omega = 0.127 Nm \times 31.5 \frac{\text{radians}}{\text{second}} = 4.00 W
\]

\[
P_{\text{maximum}} = I \times V = 1.2 A \times 12 V = 14.4 W
\]
The expected power is less than 30% of the ideal power output of the motor shield. However, after testing the reliability of these motors, some issues were uncovered. By lifting a 4 kg mass, the force of 40 N was simulated.

By running the motors forward five full rotations to lift the mass and then and backward five rotations to return to the original position, it was apparent that the motors were skipping steps. The inaccuracies were occurring at the direction changes during sharp decelerations and accelerations. While the motors were being operated at less than one third of the maximum holding torque, this operating torque was much higher than the torque the motor could produce while moving between steps when rapidly accelerating or decelerating. Once the motors skipped a step, then it was no longer possible to reliably control the position using open loop control. This means that it would have been necessary to install
additional sensors on the phantom, increasing the cost and complexity of operating the simulator. Additionally, the ±100 mm/sec velocity specification was outside the operating range of the stepper motors since moving two motors simultaneously required sending alternating commands to each motor and this slowed down the motion.

To overcome the limitations of the stepper motors, a more robust actuation system was needed. Motors with position encoders, gearboxes, and motor controllers from Maxon Motor are commonly used in robotic systems. These motors can be easily controlled from a PC via USB. Since the motors included encoders it is possible to make sure the motors can return to the same position on every cycle of the simulator. In consulting with the sales representative at Maxon, it was determined that there was not a tradeoff in terms of price between a faster motor with less torque or a slower motor with more torque. A 6 cm diameter motor pulley was selected based on the size of the other components in the system.

\[
\tau = F \times r = 40N \times 6cm = 1.2Nm
\]

\[
\omega = 60 \times \left( \frac{v}{2\pi r} \right) = 60\frac{\text{seconds}}{\text{minute}} \times \left( \frac{100 \text{ mm}}{\text{second}} \times \frac{1}{2\pi \times 30\text{mm}} \right) = 31.8 \text{ RPM}
\]

In order to achieve the proper force and speed specifications, this meant a package capable of achieving 1.2 Nm of torque and 32 revolutions per minute. These specifications were easily met with in stock components and were quickly supplied by Maxon. The motors included encoders that have 2048 counts per revolution. The gearboxes added to the motor are 128:1. With this gearbox and encoder combination, each encoder count or ‘qc’ is equivalent to \(1/262144\)th of
one rotation of the output shaft. Using the prescribed 6 cm diameter motor pulley, a move of one qe pulls the actuation cable 0.000719 mm. This is much finer position control than can be measured using optical tracking which will be discussed later.

4.2.4 Architecture Design

In order to combine all of the necessary components, a supporting structure was designed. The structure built from the aluminum extrusions pointed out important considerations that lead to a better final design. The mounting brackets used to support the pulleys were “L” shaped and could not provide enough support to actuate the phantom without breaking. The designs of the pulleys were also adjusted at this stage. A wider channel is beneficial for keeping the cables from slipping out of the channel. It was also determined that the twist pulleys could not be placed level the actuation ring in the phantom. This is the case because when the phantom is pulled by the lift cables, the twist pulleys would then have extra forces applied to them. This structure allows for the phantom to be supported by a submersible stage and safely isolate the electronic components above the water bath.
Figure 22. Design sketch of full system architecture.

The support structure also provides attachment points for the various pulleys needed to route the cables from the phantom to the motors. The structure is made entirely of nonmetallic components. As discussed previously, this is important so that there is not interference with electromagnetic tracking and so that the structure is not affected by the water bath. By minimizing the bulk of the structure, there is ample space around the cardiac phantom to interact with it using interventional or imaging tools. The horizontal platforms are laser cut from 6.35 mm sheets of acrylic. This manufacturing technique allowed for clearance holes for bolts and cables to be precisely positioned and cleanly cut. The vertical supports along with the angle brackets that attach the platforms to the vertical
supports are cut from clear, 38 mm wide PVC 90° angle. Jigs were made to reliably cut angle brackets to size as well as ensure proper hole spacing.

![Image of angle bracket and hole positioning jig.](image)

**Figure 23. Angle bracket and hole positioning jig.**

The orientation of these angle brackets also helps to provide maximum support for all portions of the structure. Since the forces on the phantom would push the center stage upward, the angle braces were placed above the stage so that the applied forces were distributed across the bracket rather than being applied only to the screw head. All components of the structure were assembled using nylon screws. While these are not as strong as metal ones, they provide more than enough strength for this application while meeting the important specification of being nonmetallic. The shear stress in the fasteners was calculated to ensure that they would not yield. All of the fasteners are \( \frac{1}{4}'' \)-20 or 10-24 nylon machine screws. The worst case scenario is that a 10-24 screw has to support the entire 40N force applied to the system so this is what is shown.

\[
\tau = \frac{F}{A} = \frac{40N}{\pi (3.5mm)^2} = 1.04 \text{ MPa}
\]
The shear strength of the nylon used in these screws is 66.2 MPa so these screws should not fail under normal operation of the simulator [25].

The left ventricle beats with most of the twisting and compression motion centered about its central axis. To keep the phantom from being pulled off-axis, it was necessary to provide both lifting and twisting forces on opposing sides of the phantom. To get the appropriate displacements to both lift cables and both twist cables, the two lift cables had to be connected to the motor responsible for lift and the same with the cables and motor that produce the twisting motion. Since this simulator was developed to be a development platform for cardiac catheter robots, optimizing the amount of workable space around the phantom was a necessity. Routing the actuation cables into the corners of the structure increased the workable space. Accomplishing the main objectives of moving the cables to the outer region of the structure and routing two cables to each motor meant devising a system of pulleys. To reduce the friction caused by these pulleys, ball bearings were a necessary component. The smallest readily available bearings that were entirely nonmetallic had an outside diameter of 12.7 mm. In order to give ample room to mount these within the pulleys, the diameter of the pulleys were chosen to 25.4 mm. Due to limitation in the geometry, two pulleys needed to be made 38.1 mm in diameter. These pulleys are mounted in a horizontal position on the top platform. They also utilize the same 12.7 mm diameter ball bearing used in the 25.4 mm pulleys. Two other pulleys had to be much smaller to properly route the cables. These pulleys have 6.35 mm diameter and rest directly on the 4.76 mm wear-resistant nylon shafts used to support all of the pulleys. All of the pulleys are
made from three custom cut layers of acrylic. The outer layers are 3.175 mm thick and the center layer is 6.35 mm thick. The pulleys are supported by acrylic brackets that are cut from 3.175 mm acrylic and then heated and bent into the needed “U” shape. Bending the brackets out of a single piece gave them more strength than cutting three individual pieces and gluing them together. This process was simplified by making a jig that help to bend the brackets to the proper widths. In addition to the “U” shaped brackets, special angled brackets were necessary to create the twisting motion. These brackets were bent so that the cables attached to the phantom run tangentially away from the phantom to maximize the angle that the phantom can be twisted. It is important to note that these cables must be routed in the proper directions to produce a counterclockwise rotation of the phantom when viewed from the apex since this is the normal motion of the human heart. All brackets are then affixed to the complete structure and cables are routed from the ventricle phantom to the appropriate motors.
Figure 24. Full CAD model of simulator. Lift components in pink. Twist components in green.

The entire system measures 500 mm tall by 300 mm wide and 300 mm deep. This size fits easily with a plastic water tank used for ultrasound imaging.
From the calf heart data provided by Cavusoglu the displacement of the tagged point on the left ventricle was calculated and plotted. This displacement profile was then simplified into a curve composed of seven linear segments. This simplification
allows for an accurate representation of the cyclical displacements without trying to produce motions that are too short for the motor’s PID controller to settle on.

Table 5. Displacement Segments of Calf Heartbeat

<table>
<thead>
<tr>
<th>Time (sec)</th>
<th>Start/end position (mm)</th>
<th>Elapsed time (ms)</th>
<th>Displacement (mm)</th>
<th>Move Index</th>
</tr>
</thead>
<tbody>
<tr>
<td>19.02-19.41</td>
<td>0.968-7.490</td>
<td>390</td>
<td>6.522</td>
<td>A</td>
</tr>
<tr>
<td>19.41-19.45</td>
<td>7.490-6.555</td>
<td>40</td>
<td>-0.935</td>
<td>B</td>
</tr>
<tr>
<td>19.97-20.08</td>
<td>9.661-1.176</td>
<td>130</td>
<td>-8.485</td>
<td>F</td>
</tr>
<tr>
<td>20.08-20.12</td>
<td>1.176-0.968</td>
<td>40</td>
<td>-0.206</td>
<td>G</td>
</tr>
</tbody>
</table>

In order to separate this displacement into components that could independently be actuated by the twist and lift motors, the geometry of the system was applied. From cardiac research, we know that the muscle fibers in the ventricles are oriented at -60° from the basal plane [3]. From this we can construct the images shown below and make the appropriate calculations to find the horizontal and vertical portions of the left ventricle displacement.
Figure 26. Diagram of displacement in terms of lift and twist.

\[ \text{twist} = \left(\frac{1}{2}\right) \times \text{absolute displacement} \]

\[ \text{lift} = \left(\frac{\sqrt{3}}{2}\right) \times \text{absolute displacement} \]

These values were then converted into encoder counts by using the size of the motor pulley, the motor gear ratio, and the number of encoder counts per motor revolution.

\[ q_{c_{\text{lift}}} = \frac{\text{lift}}{\text{linear displacement of } 1qc} = \frac{\text{lift}}{0.00719\text{mm}} \]

\[ q_{c_{\text{twist}}} = \frac{\text{twist}}{\text{linear displacement of } 1qc} = \frac{\text{twist}}{0.00719\text{mm}} \]

These calculations were performed using Matlab in order to output a list of displacements and velocities that could be input into the motor control software. The
ratio of lift to twist can be adjusted to reflect the specific motions of each of the other cardiac chambers.

Creating a repeatable heart motion was accomplished by repetitively sending the commands for one heart beat to the motor controllers. Since two motors needed to be run cyclically and simultaneously, Microsoft Visual Studio was used to send the proper commands to the EPOS motor controllers. By setting the displacement and velocity for specific segments of the heartbeat, the desired displacement profile was created.
5. Data Collection

5.1 Ultrasound Imaging

To evaluate the echogenic properties of the phantom material, it was necessary to image the phantom. A portable ultrasound machine was used to collect the videos of the beating heart which were subsequently saved to a computer using the Epiphan screen capture system. Many different views were taken so that individuals familiar with cardiac ultrasound images could later evaluate the geometry and ultrasound properties of the phantom. These views included horizontal and vertical cross sections with and without cardiac catheters interacting with the phantom.

Figure 27. Ultrasound image of cardiac phantom.
5.2 Data for Motion Refinement

Throughout the refinement of the motor controls it was necessary to measure the displacement of the phantom in order to check that the motors were reaching their intended destinations. A Claron Micron Tracker optical tracking system was used to measure the time and displacement during test cycles. The Claron measures the $x$, $y$, and $z$ positions of specific checkerboard markers with 0.25 mm accuracy at 22 Hz.
Figure 29. Claron Micron Tracker and cardiac phantom with optical tracking marker.

The checkerboard marker was placed halfway between the upper and lower edge of the phantom on the outer surface. This position was used in order to best match the position of the sonomicrometer crystal used to in the data collected by Cavusoglu.

Figure 30. Calf heart with sonomicrometer crystals sutured in place [11].
As changes in the controller code were made, data was collected to determine if any further changes were necessary. This data was then analyzed using Matlab in order to refine the motion of the simulator.

5.3 Data for Motion Reliability

Once the simulator was operating reliably, it was necessary to measure the accuracy over an extended period of use. The simulator was set to beat ten times per test and then position was measured for ten consecutive tests for a total of 100 measured heartbeats. The checkerboard marker was kept in the same position as previous tests. None of the components of the system showed signs of fatigue during these tests.

5.4 Matlab

The data that was collected was then imported into Matlab for plotting and analysis. The absolute displacement of the marker was calculated and then plotted. Using the ginput command, different sets of data points could easily be collected from each test by manually selecting the data point with the computer cursor. The data collected included the duration of each beat and the height of each of the three major displacement peaks in each heartbeat. The height of the peaks was determined by taking the difference between the position of the marker at the start the heartbeat and the maximum value at each peak.
Figure 31. Displacement peaks during one simulator beat.

The beat length was taken by collecting the time at which the marker returned to its origin.
Figure 32. Start and end points for one simulator beat.

The analogous data points were taken from the calf heart data so that comparisons could be made. Theses sets of data were saved into matrices that could be analyzed in aggregate.
6. Data Analysis

6.1 Phantom Material

Ultrasound images of the cardiac phantom were evaluated by members of the BioRobotics lab who were experienced in reading ultrasound images. The researchers agreed that the gelatin phantom material used in the final simulator was an accurate analog for cardiac tissue. Additionally, images taken of a PVA hydrogel phantom also produced favorable results. Since either of these materials can function in the simulator as well as be imaged properly, it is possible to use the material that is better suited for the tests to be performed. Gelatin is not as durable as PVA so if numerous, lengthy test are to be performed, molding a PVA phantom would be favorable. However, if only a few minor tests are needed, a gelatin phantom will suffice.

6.2 Simulator Performance

The measured phantom position data was analyzed to determine the reliability of the simulator. The raw position measurements were transformed into absolute displacements in the same way as the calf heart data. This was done by calculating the difference in the $x$, $y$, and $z$ coordinates between subsequent data points. The average length of the heartbeat as well as the average height for each of the three main peaks was calculated for the simulator data and the calf heart data. The chart below lists the results from this analysis.
Table 6. Comparison of Calf Heart and Simulator

<table>
<thead>
<tr>
<th></th>
<th>Calf Heart</th>
<th>Simulator</th>
<th>Percent Error</th>
</tr>
</thead>
<tbody>
<tr>
<td>First peak</td>
<td>6.1 mm (std=0.39 mm)</td>
<td>6.3 mm (std=0.11 mm)</td>
<td>2.9%</td>
</tr>
<tr>
<td>height</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Second peak</td>
<td>9.3 mm (std=0.88 mm)</td>
<td>9.4 mm (std=0.09 mm)</td>
<td>1.9%</td>
</tr>
<tr>
<td>height</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Third peak</td>
<td>8.2 mm (std=0.23 mm)</td>
<td>8.6 mm (std=0.15 mm)</td>
<td>5.4%</td>
</tr>
<tr>
<td>height</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Length of one</td>
<td>1.06 sec (std=0.03 sec)</td>
<td>1.47 sec (std=0.04 sec)</td>
<td>38.7%</td>
</tr>
<tr>
<td>heartbeat</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

The simulator displacement peaks closely match those of the calf heart data. This was the primary specification that needed to be met for simulating the heartbeat. The limitation of this simulator is its ability to replicate a fast heart rate. The difference of 410 milliseconds between the simulator heartbeat length and the calf heartbeat length is due to several factors. When the commands are sent to the motor controllers, there is some latency due to the USB interface. This is likely on the order of tens of milliseconds which is non-negligible when the duration of some of the displacement segments are as short as 40 milliseconds. Another source of this timing error is the limitations of the motor controller. While the motors have the proper specifications for reaching the desired velocities and accelerations, the PID controllers cannot produce ideal acceleration profiles. Any delay introduced because of the motors not getting to full velocity fast enough increases the total beat length. It would be possible to create a heartbeat that had the proper duration with this system, however it would mean sacrificing the fine details of the complex displacement pattern. In order to guarantee that the motion profile was achieved, the duration of the commands for
each of the peaks was increased, resulting in a longer heartbeat time. The primary focus of this simulator is the recreate the motion of the heart so it was decided that a slower heartbeat with a more realistic motion was favorable to a faster beat with a more simplistic motion.
7. Conclusions/Future Work

7.1 Conclusions

From the data that was collected, it can be concluded that all of the specifications of the simulator were addressed. By using motors to actuate cables attached to a cardiac phantom, a physical development platform was realized. The utilization of an echogenic material that contains isolated attachment points for actuation allows for the proper motion of the heart to be mimicked. The components of the simulator are fabricated from non-metallic materials and any metal parts of the motors and controllers are isolated from the area immediately around the cardiac phantom. This configuration allows for ultrasound imaging and electromagnetic tracking to be utilized without interference. The simulator uses two degrees of freedom to actuate the cardiac phantom allowing for phantoms of any heart chamber to be introduced into the simulator. The relationship between the two degrees of freedom can be adjusted depending on the anatomical motions of the chamber being simulated. The primary phantom used in the simulator was the left ventricle since the displacements for this chamber are largest during the heartbeat cycle, and therefore the simulator was designed for the most demanding set of motion specifications. The displacements of this simulator closely mimic calf-heart data that is used as an analog for the human heart. The combination of these favorable features makes this simulator an excellent tool for the continued development and validation of cardiac catheter robots.

7.2 Future Work

With further refinement, this device will be a versatile cardiac simulator. Molds to make cardiac phantoms of heart chambers other than the left ventricle need to be
made so that procedures can be tested on various portions of the heart. These phantoms can incorporate similar attachment and actuation rings so that they easily interface with the current actuation system.

The current heart rate of the simulator is limited to approximately 40 beats per minute. Simulating faster heart rates would be useful for testing cardiac catheter robots. The interface between the computer and motor controllers needs to be improved in order to create the proper velocities and accelerations associated with a faster heart rate. This can be done by sending control commands to the motors through a protocol with lower latency, by tuning the PID controller to reach velocity profiles more quickly, or by purchasing more powerful motors that are capable of higher accelerations. With these improvements, the simulator will be a full platform for recreating all portions of the heart and its motion.
8. Acknowledgements

The author would like to express his deepest gratitude to all of those who have provided support and guidance with this project. Special thanks to Professor Rob Howe and Paul Loschak who advised and supported this project from start to finish. Thank you to the ES100hf teaching staff for all your time and advising as well as the incredible support in the Undergraduate Teaching Labs. Last but not least, thank you to the Harvard College Class of 2015 engineers.
9. References


10. Appendices

10.1 Budget

Table 7. List of Materials and Prices

<table>
<thead>
<tr>
<th>Item</th>
<th>Quantity</th>
<th>Total Cost</th>
<th>Purchasing Lab</th>
</tr>
</thead>
<tbody>
<tr>
<td>Clear acrylic, 3.175mm, 300x600mm</td>
<td>1</td>
<td>$15.76</td>
<td>Undergraduate Teaching Labs</td>
</tr>
<tr>
<td>Clear acrylic, 6.35mm thick, 300x600mm</td>
<td>3</td>
<td>$84.24</td>
<td>Undergraduate Teaching Labs</td>
</tr>
<tr>
<td>PVC 90° angle</td>
<td>2.25 m</td>
<td>$35.87</td>
<td>Undergraduate Teaching Labs</td>
</tr>
<tr>
<td>¼”-20 nylon machine screws</td>
<td>44</td>
<td>$3.45</td>
<td>Undergraduate Teaching Labs</td>
</tr>
<tr>
<td>10-24 nylon screws</td>
<td>20</td>
<td>$1.25</td>
<td>Undergraduate Teaching Labs</td>
</tr>
<tr>
<td>¼”-20 nylon nuts</td>
<td>44</td>
<td>$3.14</td>
<td>Undergraduate Teaching Labs</td>
</tr>
<tr>
<td>10-24 nylon nuts</td>
<td>20</td>
<td>$1.19</td>
<td>Undergraduate Teaching Labs</td>
</tr>
<tr>
<td>Ball bearings, acetyl with glass balls</td>
<td>10</td>
<td>$76.00</td>
<td>Undergraduate Teaching Labs</td>
</tr>
<tr>
<td>Nylon 4-40 set screws</td>
<td>24</td>
<td>$1.50</td>
<td>Undergraduate Teaching Labs</td>
</tr>
<tr>
<td>4.76 mm wear-resistant nylon shaft</td>
<td>750 mm</td>
<td>$0.75</td>
<td>Undergraduate Teaching Labs</td>
</tr>
<tr>
<td>Maxon motor and gearbox</td>
<td>2</td>
<td>$726.60</td>
<td>Biorobotics Lab</td>
</tr>
<tr>
<td>EPOS controller for Maxon motor</td>
<td>2</td>
<td>$676.10</td>
<td>Biorobotics Lab</td>
</tr>
<tr>
<td>Cyanoacrylate adhesive</td>
<td>-</td>
<td>&lt;$10.00</td>
<td>Biorobotics Lab</td>
</tr>
<tr>
<td>PVA (56 grams per phantom)</td>
<td>1</td>
<td>$10.00</td>
<td>Undergraduate Teaching Labs</td>
</tr>
<tr>
<td>TOTAL:</td>
<td></td>
<td>$1645.85</td>
<td></td>
</tr>
</tbody>
</table>
%% Cardiac Simulator Converter
% Script for converting cardiac displacements into lift and twist qc values.
% Demetrio Anaya
% March 20, 2015
%

%% Clear
clear all;
close all;

%% Read in Displacements and Elapsed Times
% Input displacement and elapsed time for any heart data here.
% Use these displacements for more increments during the heartbeat.
% displacements = [.0543;4.291;0.912;1.263;-0.935;0.938;3.777;-0.87;0.47;0;
% -3.109;1.715;-0.398;0.583;-8.6913]; % mm
% elapsedtime = [30;180;90;90;40;80;90;30;60;110;60;30;30;60;110]; %msec
% Use these displacements for the three main peaks in the heartbeat.
displacements = [6.5218; -0.935; 4.715; -3.479; 2.5; -9.1; -0.206]; % mm
elapsedtime = [390; 40; 170; 140; 110; 130; 40]; %msec

n=size(displacements,1);
displacements = displacements.*2; % Double displacements to account for difference in measurement and actuation position.

%% Constants
circumference = 2*pi*30; % mm
mmtoqc = circumference/(2048*128); % mm/qc
scalingfactor = 2; % This is used to tune the cable tension in the system

%% Convert Values
% This is based on the knowledge that the cardiac tissue fibers are
% oriented at -60 degrees in the left ventricle. If using another area, the proper geometry will need to be used to decompose the displacement into lift and twist.

\[
\text{liftqc} = (-0.5.*\text{displacements})./\text{mmtoqc}; \quad \text{The minus sign is due to the positioning of the motor.}
\]

\[
\text{twistqc} = (\sqrt{3}/2).*\text{displacements}./\text{mmtoqc};
\]

%%% Round to Nearest Integer

\[
\text{twistqcint} = \text{scalingfactor}.*\text{round}(\text{twistqc}); \quad \text{% THIS IS NODE ONE}
\]

\[
\text{liftqcint} = \text{scalingfactor}.*\text{round}(\text{liftqc}); \quad \text{% THIS IS NODE TWO}
\]

% Calculate the absolute positions for each step. This is used so that absolute positioning commands can be used rather than relative ones. The advantage being that an error in one segment can be corrected in the next segment.

\[
\text{twistqcintabs} = \text{zeros}(n,1);
\]

\[
\text{liftqcintabs} = \text{zeros}(n,1);
\]

\[
\text{twistqcintabs}(1) = \text{twistqcint}(1);
\]

\[
\text{liftqcintabs}(1) = \text{liftqcint}(1);
\]

\[
\text{for} \; i=2:n
\]

\[
\text{twistqcintabs}(i) = \text{twistqcintabs}(i-1) + \text{twistqcint}(i);
\]

\[
\text{liftqcintabs}(i) = \text{liftqcintabs}(i-1) + \text{liftqcint}(i);
\]

\[
\text{end}
\]

%%% Calculate Velocities

\[
\text{times} = \text{elapsedtime}.1000; \quad \text{% sec}
\]

\[
\text{twistvelocities} = 0.5.*\text{displacements}.\text{times}; \quad \text{% mm/sec}
\]

\[
\text{liftvelocities} = (\sqrt{3}/2).*\text{displacements}.\text{times}; \quad \text{% mm/sec}
\]

\[
\text{twistvelocitiesrpm} = \text{twistvelocities}.*60./\text{circumference}.*128;
\]

\[
\text{twistvelocitiesrpmint} = \text{scalingfactor}.*\text{abs}(%\text{round}(\text{twistvelocitiesrpm}));
\]

\[
\text{% Take integer value and absolute value}
\]

\[
\text{liftvelocitiesrpm} = \text{liftvelocities}.*60./\text{circumference}.*128;
\]

\[
\text{liftvelocitiesrpmint} = \text{scalingfactor}.*\text{abs}(%\text{round}(\text{liftvelocitiesrpm}));
\]

\[
\text{% Take integer value and absolute value}
\]

\[
\text{sleptimes} = \text{round}(\text{times}.*1.0*1000); \quad \text{% Add 10\% and convert to ms}
\]

\[
\text{totalbeattime} = \text{sum}(\text{sleptimes})/1000;
\]
bpm = 60/totalbeattime;

%% Check that the qc return to start position
% Since 1 qc translates a linear displacement of <0.001mm as long as these
% values are less than 10 there is no need to make any adjustments.
twistsum = sum(twistqcint);
liftsum = sum(liftqcint);