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# CPR Assistive Device

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# **CPR Assistive Device**

A Major Qualifying Project  
Submitted to the Faculty of  
Worcester Polytechnic Institute  
in partial fulfillment of the requirements for the degree  
of a Bachelor of Science

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## *Abstract*

The American Heart Association states that more than 350,000 cardiac arrests occur outside of hospitals each year, and about 90% of those are fatal. In order to increase a cardiac arrest patient's chance for survival, adequate compressions must be performed at the center of the chest down to 2-2.4 inches at a rate of 100-120 beats per minute, however it is estimated that only 1 in 6 Americans actually know how to perform CPR. Currently, existing CPR devices that give real-time feedback to bystander rescuers, do not satisfy all criteria in being ergonomic, intuitive, cost-effective, portable and lightweight, and giving both audio and visual feedback. Our final CPR device satisfies all of those qualities. It gives both audio and visual feedback with a 110-bpm metronome from a buzzer and a series of LEDs that correspond to compressions that are "Too Shallow", "Good", and "Too Deep" from acceleration data collected from a 9DOF accelerometer inputted into a microcontroller. It is also compact, uses an interchangeable 9V battery, lightweight, and can withstand a safety factor of 3 times the maximum 120 pound-force needed to compress the chest. A spring box apparatus was designed to test compressions with the device between 0-3 inches and verified with one hundred 2-minute compression tests, the CPR device proved to work 73% of the time. IRB approval was obtained to test the device with 27 college-aged subjects and 7 CPR-certified subjects, which showed that the device improved CPR techniques in 81.4% of subjects, and had a 96% increased user confidence level in performing CPR.

# *Table of Contents*

<i>Abstract</i>	<b>1</b>
<i>List of Figures</i>	<b>4</b>
<i>List of Tables</i>	<b>7</b>
<i>List of Appendices</i>	<b>8</b>
<i>Acknowledgments</i>	<b>9</b>
<i>Disclaimer</i>	<b>10</b>
<i>Chapter 1. Background</i>	<b>11</b>
1.1. Cardiac Arrest Background	<b>11</b>
1.2. CPR Guidelines & Common Errors	<b>12</b>
1.3. Technological Need	<b>14</b>
<i>Chapter 2. Device Specifications</i>	<b>21</b>
2.1. Design Criteria	<b>21</b>
2.2. Specifications based on Existing Devices	<b>21</b>
2.3. Specifications based on Biological Reasoning	<b>22</b>
<i>Chapter 3. Electrical Components</i>	<b>26</b>
3.1. General Design Approach	<b>26</b>
3.2. Specific Module Design	<b>28</b>
<i>Chapter 4. Prototyping</i>	<b>48</b>
4.1. Factor of Safety	<b>48</b>
4.2. Stress Tests	<b>49</b>
4.3. Material Selection	<b>50</b>
4.4. Component Modification 1	<b>74</b>
4.5. Component Modification 2	<b>79</b>
<i>Chapter 5. Testing Apparatus Design</i>	<b>83</b>
5.1. First Design Iteration	<b>83</b>
5.2. Second Design Iteration	<b>85</b>
5.3. Third Design Iteration	<b>87</b>
5.4. Fourth Design Iteration	<b>88</b>
5.5 Final Design	<b>90</b>
<i>Chapter 6. Results</i>	<b>91</b>

<b>6.1. Calibration Results</b>	<b>91</b>
<b>6.2. Subject Results</b>	<b>92</b>
<i>Chapter 7. Ethics</i>	<b>93</b>
<i>Chapter 8. Conclusion</i>	<b>94</b>
<i>Chapter 9. Future Plans</i>	<b>94</b>
<i>References</i>	<b>96</b>
<i>Appendix</i>	<b>102</b>

## *List of Figures*

Figure 1. Hypoxia Timeline	12
Figure 2. Correct CPR Positioning (EMS Safety, 2015)	13
Figure 3. PocketCPR	14
Figure 4. CPR Ezy	15
Figure 5. CPR-Plus	17
Figure 6. Laerdal CPR Meter	18
Figure 7. CPRGlove	19
Figure 8. Anthropometric Data: Hand Breadth	23
Figure 9. Anthropometric Data: Hand Length	24
Figure 10. Anthropometric Data: Sternum Width	24
Figure 11. 9V Battery	30
Figure 12. AA Battery	30
Figure 13. 3V Disk	31
Figure 14. Slide Switch	33
Figure 15. Cutler-Hammer	33
Figure 16. Red Rocker	34
Figure 17. ADXL345 Accelerometer	37
Figure 18. MPU6050 Accelerometer	37
Figure 19. MMA8452	38
Figure 20. Arduino Uno	40
Figure 21. Sparkfun Pro Micro	40
Figure 22. Arduino Mini Nano	41
Figure 23. Single Tone Buzzer	43
Figure 24. Mini Speaker	43
Figure 25. 16 Tone Buzzer	44
Figure 26. 10mm Red LED	46
Figure 27. RGB Clear LED	46
Figure 28. RGB Addressable LED	47
Figure 29. 25 pack 5mm red LED	47
Figure 30. Image of stress test showing force placement	49
Figure 31. Image of stress test showing location of fixation	49
Figure 32. Image of device on a flat surface	50
Figure 33. Isometric view of Design 1	51
Figure 34. Side view of bottom half of device showing chamfer	52
Figure 35. Stress test on Design 1 for average female palm size	52
	4

<b>Figure 36. Stress test on design 1 for average male palm size</b>	<b>52</b>
<b>Figure 37. Configuration of Components with 2:1 Length-to-Width Ratio and Minimum Height</b>	<b>53</b>
<b>Figure 38. Initial Prototype to House Components with Wall Thickness of 10mm</b>	<b>54</b>
<b>Figure 39. Adjusted Prototype with Smaller Button and Wall Thickness of 4mm</b>	<b>54</b>
<b>Figure 40. Initial Printed Prototype for Configuration of Electrical Components (Top)</b>	<b>55</b>
<b>Figure 41. Initial Printed Prototype for Configuration of Electrical Components (Bottom)</b>	<b>55</b>
<b>Figure 42. Image of design 3 where bosses, thinning of walls, and ribbing can be seen</b>	<b>57</b>
<b>Figure 43. Image of design 3 filleting at bottom of boss</b>	<b>57</b>
<b>Figure 44. Image of Design 3 displaying LED speaker, button holes, and hand blocking bar</b>	<b>58</b>
<b>Figure 45. Stress test on design 3 for average female palm size</b>	<b>58</b>
<b>Figure 46. Stress test on design 3 for average male palm size</b>	<b>58</b>
<b>Figure 47. Design 4 ribbing to hold each electrical component in place</b>	<b>59</b>
<b>Figure 48. Design 4 ribbing purely for structural support</b>	<b>60</b>
<b>Figure 49. Stress Tests on Design 4 for average female and male palm size respectively</b>	<b>60</b>
<b>Figure 50. Stress test on design 4 for average female palm size</b>	<b>61</b>
<b>Figure 51. Stress test on design 4 for average male palm size</b>	<b>61</b>
<b>Figure 52. Assembly View of Design 5</b>	<b>62</b>
<b>Figure 53. Section View inside design 5a to see 1mm gap in ribbing</b>	<b>63</b>
<b>Figure 54. Section view inside design 5b to see no gap in ribbing</b>	<b>63</b>
<b>Figure 55. Stress test on design 5a for average female palm size</b>	<b>63</b>
<b>Figure 56. Stress test on design 5a for average male palm size</b>	<b>64</b>
<b>Figure 57. Stress test on design 5b for average female palm size</b>	<b>64</b>
<b>Figure 58. Stress test on design 5b for average male palm size</b>	<b>65</b>
<b>Figure 59. Electronic Component Test Plate</b>	<b>66</b>
<b>Figure 60. Bottom and top half of design 5 showing electronic component placement</b>	<b>67</b>
<b>Figure 61. Port cut into side of device for testing</b>	<b>67</b>
<b>Figure 62. Image of upside down "T" s in middle ribbing of design 6</b>	<b>68</b>
<b>Figure 63. Redesigned "T" for holding wires</b>	<b>68</b>
<b>Figure 64. Image of Device Bottom without pins and a porthole for easier wiring</b>	<b>69</b>
<b>Figure 65. Top half of device with moved speaker and button location to make room for new components</b>	<b>69</b>
<b>Figure 66. The battery slot on the back of the device</b>	<b>70</b>
<b>Figure 67. Battery Cover</b>	<b>70</b>
<b>Figure 68. Isometric view of the device</b>	<b>71</b>
<b>Figure 69. Inside bottom of device where middle bar has been removed and PCB is located</b>	<b>71</b>
<b>Figure 70. Inside top of device where new bar separating battery from rest of components</b>	<b>72</b>
<b>Figure 71. Stress test on final design for average female palm size</b>	<b>73</b>

<b>Figure 72. Stress test on final design for average male palm size</b>	<b>73</b>
<b><i>Figure 73. UlincoS U16A Pushbutton</i></b>	<b>74</b>
<b>Figure 74. CO-RODE Tactile Switch</b>	<b>75</b>
<b>Figure 75. The BNO055 9DOF Accelerometer</b>	<b>76</b>
<b>Figure 76. Pin Layout of the Sparkfun Pro Micro</b>	<b>77</b>
<b>Figure 77. Power Regulation Circuitry</b>	<b>80</b>
<b>Figure 78. 8 Ohm Speaker</b>	<b>81</b>
<b>Figure 79. Audio Amplifier</b>	<b>81</b>
<b>Figure 80. First Design of the Printed Circuit Board</b>	<b>82</b>
<b>Figure 81. Second Design of the Printed Circuit Board</b>	<b>82</b>
<b>Figure 82. Solidworks Model of the Preliminary Testing Apparatus</b>	<b>84</b>
<b>Figure 83. Prototype of First Design</b>	<b>85</b>
<b>Figure 84. Second Design of Test Apparatus</b>	<b>85</b>
<b>Figure 85. Third Design of Testing Apparatus</b>	<b>87</b>
<b>Figure 86. Final Design</b>	<b>90</b>



## *List of Tables*

<b>Table 1. Comparative List of CPR Devices</b>	<b>20</b>
<b>Table 2. Battery Weight Assignments</b>	<b>29</b>
<b>Table 3. Battery Value Analysis</b>	<b>31</b>
<b>Table 4. On/Off Switch Weight Assignments</b>	<b>32</b>
<b>Table 5. On/Off Value Analysis</b>	<b>34</b>
<b>Table 6. Accelerometer Weight Assignments</b>	<b>36</b>
<b>Table 7. Accelerometer Value Analysis</b>	<b>38</b>
<b>Table 8. Microcontroller Weight Analysis</b>	<b>39</b>
<b>Table 9. Microcontroller Value Analysis</b>	<b>41</b>
<b>Table 10. Buzzer Weight Analysis</b>	<b>42</b>
<b>Table 11. Buzzer Value Analysis</b>	<b>44</b>
<b>Table 12. LED Weight Analysis</b>	<b>45</b>
<b>Table 13. LED Value Analysis</b>	<b>47</b>

## *List of Appendices*

<b>Appendix 1. Testing Procedure</b>	<b>102</b>
<b>Appendix 2. Code for MPU6050</b>	<b>108</b>
<b>Appendix 3. Code for BNO055</b>	<b>111</b>
<b>Appendix 4. IRB Approved Subject Testing Guidelines</b>	<b>115</b>
<b>Appendix 5. PCB Diagram</b>	<b>120</b>
<b>Appendix 6. Biomedical Engineering Society Code of Ethics</b>	<b>121</b>

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## ***Disclaimer***

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# Chapter 1. Background

## 1.1. Cardiac Arrest Background

The Center for Disease Control and Prevention (CDC) stated that in 2016, the number one most common cause of death in the U.S. was from Heart Disease, which claimed approximately 600,000 lives or about 23.4% of deaths [7]. Heart Disease is an umbrella term for many different pathological issues with the heart, such as Hypertension, Congenital Heart Diseases, and even stress [7]. Though the etiology, or cause of the issue, is varying and sometimes unknown, most cases exhibit the same outcome, death.

All life depends on the cardiovascular system to deliver oxygen and nutrients to tissues throughout the entire body, in order for normal cell metabolism and functions to occur. The cardiovascular system, at its core, works as a closed tubular system that contains a pump (the heart), pipes (vasculature), and fluid (blood); all of which are necessary for perfusion to tissues and return to the heart.

Death occurs when one or more parts of the cardiovascular triad critically fail; this situation is known as shock. There are four major types of shock: Hypovolemic shock, Distributive shock, Cardiogenic shock, and Obstructive shock. Hypovolemic shock occurs when there is a failure in the fluid aspect of the cardiovascular triad. Typically, due to trauma, if there is a loss of at least 20%, or 0.3 gallons, of the blood from the system, this is considered a life-threatening state [27]. Distributive shock occurs when the piping vasculature dilate too wide and severely decrease the blood pressure of the system [1]. This situation can be caused by a neurological problem, like with spinal cord injuries, or by a chemical release of neurotransmitter, Histamine, in an allergic anaphylactic shock [45]. Cardiogenic shock occurs when the heart fails to pump blood autonomously [1]. This condition is typically due to myocardial infarctions or electrical issues in the heart where it either hyperacts in ventricular spasms called fibrillation, or doesn't act at all, in asystole [35]. Obstructive shock occurs when the heart fails to pump blood due to an outside force inhibiting it [1]. This event can be caused by a blood clot inhibiting blood flow to the tissues of the heart causing the tissue to die or from a buildup of blood around the heart due to trauma that inhibits the heart to physically contract and expand [1].

However, it happens, the cardiovascular system eventually fails and results in Cardiac Arrest. When the heart is not doing its normal function of pumping oxygen and nutrient-rich blood to the body, cells start to die, particularly in the brain.

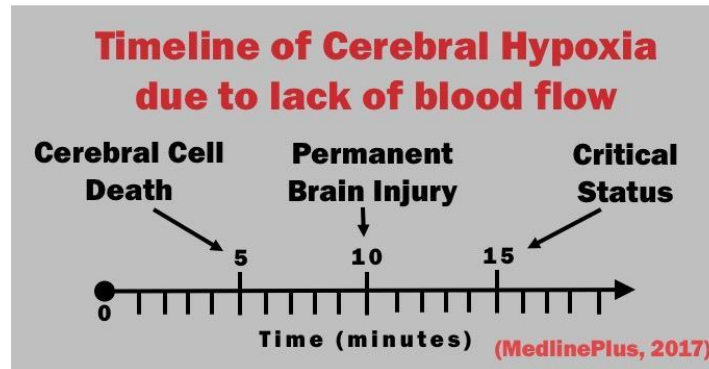


Figure 1. Hypoxia Timeline

After only five minutes without oxygen, brain cells begin to die, and within the next five minutes, the patient will suffer from permanent brain injury and eventually be in a lethal state, as shown in Figure 1.

The current solution to Cardiac Arrest includes Cardiopulmonary Resuscitation (CPR), in tandem with the use of Automated External Defibrillator (AED) to spark the electrical circuit of the heart and induce a normalized heartbeat.

## 1.2. CPR Guidelines & Common Errors

Cardiopulmonary Resuscitation (CPR) is performed by a rescuer on a patient undergoing Cardiac Arrest and ultimately attempts to manually pump blood throughout the body in order to preserve as much nervous and body tissue as possible. This is done by a bystander or trained medical personnel by placing the palm of the rescuer's dominant hand in the center of the patient's chest, interlocking the other hand on top of the dominant hand, locking arms, and pressing down into the patient's chest and then releasing [2]. Figure 2 illustrates proper hand placement for a left-handed First Responder during CPR.



Figure 2. Correct CPR Positioning (EMS Safety, 2015)

The American Heart Association mandates that the compression rate is between 100 and 120 beats per minute (BPM) and must be at least 2.0 to 2.4 inches deep into the chest, with upwards recoil, in order for the CPR to be successful. Though rescue breaths have been recommended in the past, the American Heart Association is focusing their efforts on the effectiveness of bystander compression-only, i.e. hands-only CPR [2], due to the fact that only 4% of oxygen is used per breathing cycle and that the patient's blood supply already carries ample oxygen if circulated properly. In fact, the AHA in a 2013 study found that patients could still survive past 38 minutes of proper CPR with good brain function [34]. Since the national average EMS response time is about 15 minutes [5], it is critical that present bystanders know how to perform compressive CPR properly in order to increase their chances of survival by three times [16]. However, every year 350,000 cardiac arrests occur outside of a hospital, and 90% of these occurrences are fatal [2]. This high statistic could be due to the severe lack of public knowledge in adequate CPR practices, in that only about 16% of Americans know how to perform CPR [37], and even if one knew the proper technique, they may not be able to tell how effectively they are performing in-field.

However, even with proper CPR, there is still a risk for death or permanent brain damage of the patient as well as other common consequences. The most common side-effect to the patient during hands-only CPR or CPR with ventilation is vomiting due to pushed air into the stomach, which can further cause threatening aspiration [12]. Other side-effects include broken ribs that may lacerate the lung [12]. CPR also poses as a risk to the rescuer and standard personal protective

equipment is always recommended, such as gloves, goggles, or a one-way valve respiratory mask if rescue breaths are given [2].

### 1.3. Technological Need

Based on the statistics provided by the American Heart Association (AHA) on the inadequate execution of CPR, several devices have been created in an attempt to mitigate the situation. Our background research has illuminated five devices that currently exist. Four of these devices are currently on the market, while the fifth was a student designed device that has yet to be marketed. We researched the available products or prototypes to determine their strengths and weaknesses. Our design would be targeted at addressing the weaknesses and augmenting the strengths of the existing units. The five existing devices researched are the Zoll Pocket CPR, the CPR Ezy, the CPR-Plus, the Laerdal CPR Meter, and the CPRGlove.

#### 1.3.1. Zoll Pocket CPR



Figure 3. PocketCPR

The Zoll Pocket CPR device is a pocket-sized CPR coaching device that uses both audio and visual feedback, seen in Figure 3. It coaches the user through the correct steps to perform CPR, while also measuring the compression depth of the chest that it is placed on. Based on the compression depth, this device can respond to the user by saying either “Push Harder” or “Good compressions”. It does not however respond with any feedback if the chest compressions are too deep into the chest. This device uses an accelerometer, which is a small electromechanical device



that will measure acceleration forces acted upon it. The accelerometer within this device measures the depth of each compression made onto the chest. This device uses a metronome system in order to coach the user to compress the chest at the rate of 100 beats per minute (bpm), which is the recommended rate provided by the AHA. It also uses flashing lights to assist the user in proper usage of the device. This device is powered by a replaceable battery. The Zoll Pocket CPR device costs approximately \$220 [48].

There have been two studies to date that have examined the Zoll Pocket CPR device. In one of these studies it was found that there were no significant changes in CPR quality parameters. In the other study, it was found that there was improved compression depth and compression rate. Throughout both studies, it was found that the fatigue of the users did not increase subjectively or objectively [24].

The major limitation of the Zoll Pocket CPR device is that its frequent prompts may confuse users. If the device is not turned off properly after each use, the battery may die and leave the device possibly unreliable for use in case of an emergency. The constant prompts of “Push Harder” and “Good compressions” could possibly confuse users based on when the prompts are recognized. The inability to inform the user that they are compressing too deep into the chest is also a problem with this device.

The major advantages of the Zoll Pocket CPR device are that it is portable, lightweight, easy to use, reusable, and it measures compression depth [24].

### 1.3.2. CPR Ezy



Figure 4. CPR Ezy

Figure 4 above is the CPR Ezy is a CPR assistance device that uses both audio and visual feedback. It coaches the user through the correct steps to perform CPR, while also measuring the compression depth of the chest that it is placed on. In order to determine that the CPR being performed while using this device is proper under the AHA's guidelines, this device uses an LED pressure sensor. LED pressure sensors measure the force acting upon them, rather than measuring the relative depth of the device as an accelerometer would. It returns audio feedback through the use of a metronome system that beats at a rate of 100 bpm in order to direct the user to compress at the suitable rate provided by the AHA. It returns visual feedback through a row of LED lights that inform the user of the applied force and whether that force is suitable for specific body types designated by size and weight. This device is powered by a battery. The CPR Ezy device costs approximately \$130 [22].

The major limitations of the CPR Ezy are that it uses a pressure sensor and was found to cause wrist discomfort due to its slippery plastic exterior [24]. If the device is not turned off properly after each use, the battery may die and leave the device possibly unreliable for use in case of an emergency. Although pressure sensors do properly measure the amount of force applied onto them, they do not measure depth in any way. Based on AHA guidelines, the amount of force required to compress a person's chest during CPR varies for each individual. However, the depth that a proper compression must reach does not vary. A compression must reach a depth of anywhere between 2 and 2.4 inches for an average adult as described by the American Heart Association. This device's incapability of measuring compression depth, makes it possibly dangerous to use, because it only measures force applied. In some instances, 120 pounds of force may not be enough to compress a person's chest, whereas in others, it may be too much force and could cause further harm to the person. The CPR Ezy also has a plastic outer case that gets slippery and leads to wrist discomfort for the user, making it uncomfortable [24].

The major advantages of the CPR Ezy device are that it is portable, lightweight, easy to use, and reusable [24].

### 1.3.3. CPR-Plus



Figure 5. CPR-Plus

The CPR-plus is a handheld pressure-sensing CPR assistance device that uses both audio and visual feedback. It uses a force plate and an analog display in order to show the compression force acting on the person's chest rather than compression depth of the chest. It uses a metronome system consisting of a speaker and an LED light to direct the user to compress at the suitable rate provided by the AHA. This device has been taken off the market [19].

The major limitations of the CPR-plus are that it measures compression force rather than compression depth, and it is not ergonomic. This device's inability to measure compression depth makes it possibly dangerous to use. This is because based on AHA guidelines, the amount of force required to compress a person's chest during CPR varies for each individual. However, the depth that a proper compression must reach does not vary. A compression must reach a depth of anywhere between 2 and 2.4 inches for any individual. The device's plastic case is also uncomfortable leading to wrist discomfort during use.

The major advantages of the CPR-plus device are that it is easy to use and it does not rely on a battery for power [19].

### 1.3.4. Laerdal CPR Meter



Figure 6. Laerdal CPR Meter

The Laerdal CPR Meter is a portable CPR assistance device that provides visual feedback. It uses an accelerometer as well as an extra pressure sensor in order to most accurately measure the chest compression depth during use. This device returns specific symbols on its screen that contains feedback for compression depth, compression rate, and chest decompression. It also records CPR data while being used that can be reviewed later. This device is powered by one replaceable battery. The Laerdal CPR Meter costs approximately \$900 [11].

The major limitations of the Laerdal CPR Meter are that it relies on a battery and that one must have specific training in order to understand how to properly use it. If the device is not turned off properly after each use, the battery may die and leave the device possibly unreliable for use in case of an emergency. This device's only feedback to the user comes in specific symbols on a small display screen. These symbols are unrecognizable for someone who has not had training in the use of this device, and therefore makes it less effective to an untrained person [24].

The major advantages of the Laerdal CPR Meter device are that it is reusable, portable, lightweight, it saves and records data during its use that can be reviewed later on, and that it measures both compression force and depth in order to most properly provide feedback for the user [11].

### 1.3.5. CPRGlove



Figure 7. CPRGlove

The CPRGlove is a CPR assistance device designed by two students from McMaster University that uses both audio and visual feedback. The glove uses an electrocardiograph to measure the person's heart rate which can then be assessed to determine if he/she requires CPR. It uses an accelerometer to measure the chest compression depth. Piezoelectric sensors are used to provide visual feedback based on whether the compressions are deep enough into the chest. A metronome provides audio feedback in order to direct the user to keep compressions at a proper rate. The glove itself is made up of a nylon-spandex composite. The CPRGlove is currently a prototype [43].

The major limitations of the CPRGlove are that it is not portable, it is has not been marketed, it is expensive, and the materials that were used to make the glove are not suitable for repeatable use. The glove itself is very large which makes it harder to carry around when compared to the other devices listed previously. It also is too large to fit inside a normal first aid kit. The device has yet to be marketed 11 years after its development, and even though the designers believe that it could be mass produced for as little as \$60, there is no evidence to support that claim. The development cost of the existing glove was \$2500. The materials chosen to make the glove portion of the device were also inexpensive and subject to failure after many uses [43].

The major advantages of the CPRGlove are that it is comfortable and ergonomic, it is reusable, it measures compression depth, and it is easy to use. Table 1 summarizes the CPR assistance devices.

Table 1. Comparative List of CPR Devices

Device	Measuring Mechanism(s)	Audio Feedback	Visual Feedback	Cost	Pros	Cons
Zoll Pocket CPR	Accelerometer - measures compression depth	Yes	Yes	\$220	Portable, Lightweight, Easy to use, Reusable, Measures compression depth	Relies on a battery, Frequent prompts may confuse users
CPR-Ezy	Pressure Sensor – measures compression force	Yes	Yes	\$130	Portable, Lightweight, Easy to use, Reusable	Relies on a battery, does not measure compression depth, Not ergonomic
CPR-Plus	Force Plate – measures compression force	Yes	Yes	Not on Market	Easy to use, No battery needed	Does not measure compression depth, Not ergonomic
Laerdal CPR Meter	Accelerometer and Pressure Sensor – measures both compression force and depth	No	Yes	\$900	Reusable, Portable, Lightweight, Saves and Records data, Measures compression force and depth	Relies on a battery, requires specific training in order to properly understand its output

CPRGlove	Accelerometer - measures compression depth	Yes	Yes	\$2500	Ergonomic, Reusable, Easy to Use, Measures compression depth	Not portable, Expensive, Not suited for repeatable use
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## Chapter 2. Device Specifications

### 2.1. Design Criteria

The first step is customer requirements, this step comes from our clients need, and it specifies what they expect from the device, and it is general guidelines to follow. From there we develop our own product requirements, which are more specific requirements than the customers so that we can process through what parts we might need to create this device. The final step to determining what we need to make this device is turning our product requirements into specifications, where we determine the general parts needed to bring the device from concept to prototyping.

### 2.2. Specifications based on Existing Devices

Based on the strengths and weaknesses of the existing devices shown in Table 1, we have created a list of specifications that our device must follow in order to be most effective. This list includes being:

- Cost efficient
- Portable
- Lightweight
- Easy to use/understand
- Must measure compression depth
- Provide both audio and visual feedback
- Ergonomic

Our device must be cost efficient in order to market it as a part of any normal first aid kit. Compared to the prices of existing devices, we intend to design our device to cost no more than \$100 dollars, but keep the price as low as possible. It must be portable in order to fit within an average first aid kit and be easily accessible case of emergency. The average size of a first aid kit can range anywhere between 10" x 5" to 12" x 7". The device must also be lightweight in order to make it easy to use in any instance by any person. It must also be ergonomic and comfortable for the user so that during the application of CPR using the device, there are no complications with wrist discomfort or possible injury. The device must be easy to use and understand so that in case of emergency, anyone can quickly turn it on and begin CPR. The faster the device is applied the better for the person in danger, since it takes only minutes for someone that has gone into cardiac arrest to experience severe damage. As previously mentioned, the use of pressure sensors and force plates are a great way to measure the applied force to someone's chest during CPR. However, according to the AHA, the amount of force required to compress someone's chest during CPR varies per person, whereas depth does not. Therefore, our CPR device focuses on compression depth measurements rather than force. In order to measure compression depth, the device will use a tri-axial accelerometer rather than a pressure sensor or force plate. Our device must also provide both audio and visual feedback to the user in order to be most effective in aiding the CPR being performed. Based on the existing devices, audio feedback should return a metronome beat of 100 bpm so that the user can keep the correct pace during compressions, and visual feedback should be focused on whether the compressions being applied are not deep enough, too deep, or sufficient [21].

### **2.3. Specifications based on Biological Reasoning**

The actual size and design of the device is also influenced by a multitude of specifications that include:

- The placement of visual feedback
- The type of visual feedback
- The placement of audio feedback
- The type of audio feedback
- The placement of the power switch



- The size of the average human hand
- The size of the electrical component
- The size of the average human sternum
- The material used to make it
- Contact with the Patient’s Chest

The general size of the device is based on a combination of the size of the average human hand breadth, the size of the average human sternum, and the size of the electrical components that will sit within the device. The device should be split into two sections, a user interface and a physical interface due to the necessity of space for the electrical components as well as the necessity for a large enough area for the user to compress on. The user interface will contain most of the electrical components and will not experience any applied forces because the hands of the user will not be placed on it. The physical interface will be the section of the device where the user places his palm and compresses during CPR. The average width and palm length of the human hand in combination with the average size of the human sternum gives us the necessary information to determine what the size of the physical interface of our device should be. Based on anthropometric data of the human hand in Figure 8, the average female hand spans 3.13 inches wide spans while the average male hand spans 3.56 inches wide [3].

Hand Breadth					
FEMALE N = 2208			MALE N = 1774		
Centimeters		Inches	Centimeters		Inches
7.94	Mean	3.13	9.04	Mean	3.56
.38	Std Dev	.15	.42	Std Dev	.17
9.80	Maximum	3.86	10.60	Maximum	4.17
6.60	Minimum	2.60	7.70	Minimum	3.03
Percentiles			Percentiles		
7.09	1 <sup>st</sup>	2.79	8.07	1 <sup>st</sup>	3.18
7.19	2 <sup>nd</sup>	2.83	8.19	2 <sup>nd</sup>	3.22
7.25	3 <sup>rd</sup>	2.86	8.27	3 <sup>rd</sup>	3.25
7.34	5 <sup>th</sup>	2.89	8.36	5 <sup>th</sup>	3.29
7.47	10 <sup>th</sup>	2.94	8.51	10 <sup>th</sup>	3.35
7.56	15 <sup>th</sup>	2.98	8.61	15 <sup>th</sup>	3.39
8.63	20 <sup>th</sup>	3.00	8.69	20 <sup>th</sup>	3.42
7.69	25 <sup>th</sup>	3.03	8.75	25 <sup>th</sup>	3.45
7.74	30 <sup>th</sup>	3.05	8.82	30 <sup>th</sup>	3.47
7.79	35 <sup>th</sup>	3.07	8.87	35 <sup>th</sup>	3.49
7.84	40 <sup>th</sup>	3.09	8.93	40 <sup>th</sup>	3.51
7.89	45 <sup>th</sup>	3.11	8.98	45 <sup>th</sup>	3.54
7.93	50 <sup>th</sup>	3.12	9.03	50 <sup>th</sup>	3.56
7.98	55 <sup>th</sup>	3.14	9.09	55 <sup>th</sup>	3.58
8.03	60 <sup>th</sup>	3.16	9.14	60 <sup>th</sup>	3.60
8.08	65 <sup>th</sup>	3.18	9.20	65 <sup>th</sup>	3.62
8.13	70 <sup>th</sup>	3.20	9.26	70 <sup>th</sup>	3.64
8.18	75 <sup>th</sup>	3.22	9.32	75 <sup>th</sup>	3.67
8.25	80 <sup>th</sup>	3.25	9.40	80 <sup>th</sup>	3.70
8.32	85 <sup>th</sup>	3.28	9.48	85 <sup>th</sup>	3.73
8.42	90 <sup>th</sup>	3.31	9.59	90 <sup>th</sup>	3.78
8.56	95 <sup>th</sup>	3.37	9.76	95 <sup>th</sup>	3.84
8.66	97 <sup>th</sup>	3.41	9.86	97 <sup>th</sup>	3.88
8.74	98 <sup>th</sup>	3.44	9.93	98 <sup>th</sup>	3.91
8.86	99 <sup>th</sup>	3.49	10.04	99 <sup>th</sup>	3.95

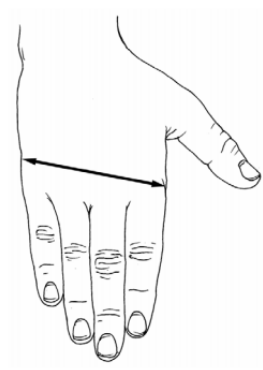


Figure 8. Anthropometric Data: Hand Breadth

The average female hand spans 7.1 inches long while the average male hand spans 7.63 inches long, shown in Figure 9 [26].

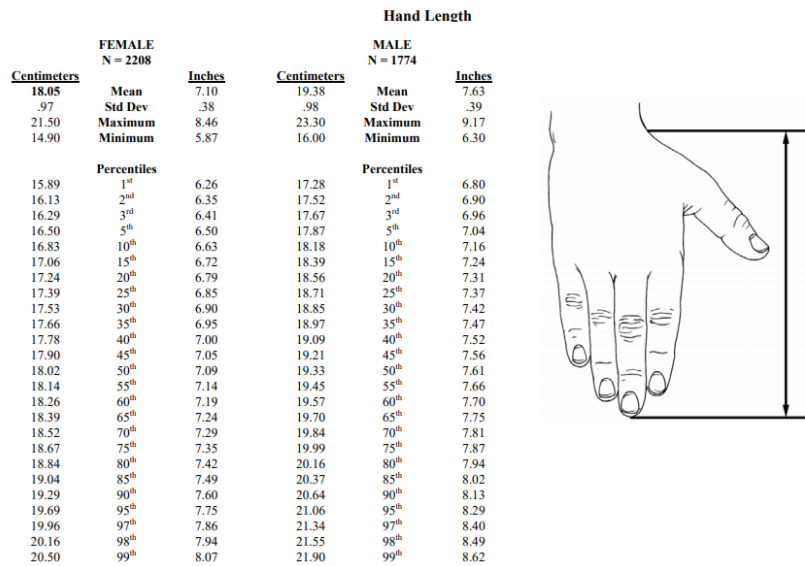


Figure 9. Anthropometric Data: Hand Length

The average sternum width for a female is 1.45 inches while the average sternum width for a male is 1.58 inches, shown in Figure 10 [38].

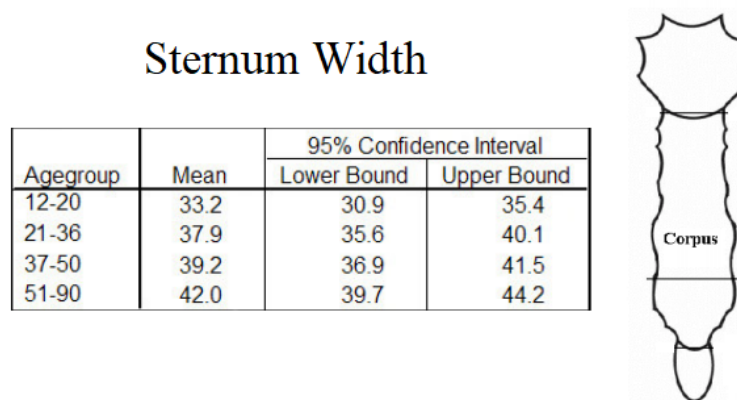


Figure 10. Anthropometric Data: Sternum Width

In order for any person to be able to correctly use our device, the physical interface section of it should be approximately 3.5 inches wide and 7.5 inches long. At this size it will accommodate most users, and rest steadily on the chest of the patient. The size of the user interface must be large enough to fit most of the electrical components.

The type and placement of visual feedback, type and placement of audio feedback, and type and placement of the power switch are all needed to properly design the user interface section of the device and also to maximize the device's effectiveness. In order for the user to properly use the device during CPR, the audio and visual feedback must be easily recognizable. Based on existing devices, LEDs will provide good illumination on all lighting situations and not draw a significant amount of energy for the display information to the user regarding chest compression depth. This is because LEDs are bright, easy to incorporate, and will help the user easily understand when a compression is either too deep, not deep enough, or suitable. The placement of these LEDs within the device is very important. In order to be most effective in providing visual feedback, the LEDs should be placed on the top of the device facing the user. In this position, the user will be looking directly down at the LEDs which will ensure they can see them, but it will also protect the LEDs from being damaged during compressions as they do not contact the user or the receiver of the CPR.

Based on existing devices, a speaker can effectively provide audio feedback to the user. This speaker has to be loud enough for the user to hear in any situation so that they can keep pace at 100 bpm. The speaker should be placed on the top of the device aimed directly towards the user to ensure that they will hear it.

The placement of a power switch is an extremely important design component. In order to avoid the device being shut off accidentally during an emergency situation, the power button or switch should be relatively small, but also be located on the top of the device facing the user. If the user can properly see the button, they can avoid interfering with it. Also, with it being located on the top of the device, it will not come in contact with the patient's chest, therefore eliminating the chance of the device being turned off through accidental contact.

The material of the device must also be considered. The device has to be able to withstand 120 lbs of force so the material itself must be strong. Compressive strength, young's modulus, and fatigue are also important in determining whether or not the material could support the applied forces, and also if the material is able to hold its shape for long periods of time. Another important factor in material selection are the material's thermodynamic properties. Since there are electrical components within the device, there could be an elevated temperature inside, so the material must be able to withstand the varying temperatures. An important factor when determining the material is how it breaks. In this case, we want the material to break into larger, dull chunks, rather than

sharp and dangerous parts, as this could injure either the rescuer or the patient. Another important factor in material selection is degradability. We want the material to be capable of functioning even in contact with various biological fluids.

The device has the potential to be mass produced using injection molding, consequently, the design requirements for injection molding as given by the “Injection Molding Design Guidelines” from Stratasys Direct, Inc [13] were followed. The wall thickness of the device will be held between 2-4mm. Ribs and gussets will be used to create support throughout the device to withstand required loads and to help partition many of the internal components. We will place all bosses away from each corner and wall in order to secure the device. The bottom of the device should have rounded edges and a chamfer to remove any sharp edges and also allow the device to fit naturally on the chest of the patient. For initial prototyping, the top of the device will remain flat and blank with rounded edges, until the exact locations of each electrical component are determined.

## **Chapter 3. Electrical Components**

After identifying the product specifications, we worked to identify the exact components that would be used in the product. In order to do so, value analysis was performed on individual components using a list of weighted value criteria and metrics determined by the team. The following section outlines this process.

### **3.1. General Design Approach**

From the product specifications we defined, we can divide the electrical component of the device into four basic subsystems: power, measurement, processing, and feedback. By doing so, we can identify individual priorities which we can use to develop value criteria for the components themselves. Because the device is meant to be relatively small and self-contained, this means that all subsystems must facilitate that. In addition, one of the other main priorities is to keep the cost low enough to allow for accessibility and therefore cost will be a factor in all components.

### **3.1.1. Power Subsystem**

The main priority of this subsystem is to ensure that the device will have power even if left untouched for long periods of times. When the device is in usage, it should have enough charge to last through variable length sessions of CPR whether it be for only a couple minutes or an entire hour. Because we want to keep the cost low and do not want the device to require any lengthy maintenance, we have excluded the option of a self-powered device or rechargeable battery. It is therefore best to simply use a readily available, inexpensive, and replaceable battery to both supply and buffer the power usage. In order to ensure that the device only uses up power when it is in use, the device should also include a switch to turn it on and off.

### **3.1.2. Depth Measuring Subsystem**

In order for the device to be able to tell the responder when s/he is providing proper CPR, it must be able to determine if the chest compressions are at the correct depth. As listed in the customer requirements, the ideal depth for compression is in the 2 - 2.4-inch range. This could be accomplished using either an accelerometer or a force plate. The amount of force required to achieve this level of compression could vary by factors such as gender or build and as such, a force sensor would not be as consistent in providing this data. As such, we are planning to use an accelerometer to read this information.

### **3.1.3. Information Processing Subsystem**

It is unlikely that we would be able to design the circuits to process information from the circuits and provide feedback within the scope of this project. As such, we will be using a microcontroller to interpret data from the sensor and control the response system to alert the user when they are performing CPR properly. One of the goals is to also provide feedback regarding compression rate or frequency; the use of a microcontroller eliminates the necessity for an external timer to accomplish this goal.

### **3.1.4. Feedback Subsystem**

In order for the device to be useful, it must be able to not only capture information about the depth of compressions but also relay it back to the user in real time without distracting them from the task at hand. It was determined that in order to accommodate for users with disabilities, this feedback would be given both audibly and visually. In previous iterations, complex signals

such as vocal commands and Liquid Crystal Display (LCD) prompts were utilized. However, this resulted in an overload of information to the user which confused them at the time of operation. As such, we decided that it would be more beneficial to use simpler cues such as buzzers and LEDs.

### **3.2. Specific Module Design**

Having determined the priorities of each subsystems and the components that would form them, we created more specific criteria for each component. This starts by first identifying the requirements of each component and the different value criteria that would affect its operation in the device. These criteria were discussed by the team and each individual member compared every criterion in relation to every other criterion, declaring which was more important to them. This data was then compiled to form our final weight assignments. Potential options for components were identified and, using the weight assignments, value analysis was performed to determine the part that would be used in the prototype. Once again, cost and size are persisting factors in each component for the same reasons listed previously and are therefore not discussed in this section.

#### **3.2.1. Battery**

The battery provides the power for the entire system.

**Value Criteria:**

- Cost
- Size
- Ease of Integration and Usage
- Safety and Durability
- Life Expectancy

One of the main concerns with regard to the battery is whether or not it will be easy to replace. If we want this device to be easily accessible, it cannot use a niche battery that you can only buy online or at specific stores. The batteries should be easily available and take very little effort to replace. Users should not require a technician to do it for them.

Because the battery will be within the device, it will also be experiencing a significant amount of movement over a relatively long period of time. It is important that the battery does not malfunction or leak during operation.

We do not expect the device to be operated or even handled often. For the convenience of the user, we must ensure that the battery has a relatively long life-expectancy so they do not need to replace it relatively often. Any more than once a year is excessive. Table 2 shows the assignments of weight.

Table 2. Battery Weight Assignments

Ben	Cost	Size	Ease of Integration	Safety and Durability	Life Expectancy	Total
Cost	0	0	0	0	0	0
Size	1	0	0.5	0	1	2.5
Ease of Integration	1	0.5	0	0.5	1	3
Safety and Durability	1	1	0.5	0	1	3.5
Life Expectancy	1	0	0.5	0	0	1.5
Eric	Cost	Size	Ease of Integration	Safety and Durability	Life Expectancy	Total
Cost	0	0	0	0	0.5	0.5
Size	1	0	0.5	0	1	2.5
Ease of Integration	1	0.5	0	1	1	3.5
Safety and Durability	1	1	0	0	1	3
Life Expectancy	0.5	0	0	0	0	0.5
Zoey	Cost	Size	Ease of Integration	Safety and Durability	Life Expectancy	Total
Cost	0	0	0	0	0	0
Size	1	0	0	0	0	1
Ease of Integration	1	1	0	0	0	2
Safety and Durability	1	1	1	0	0.5	3.5
Life Expectancy	1	1	1	0.5	0	3.5
John	Cost	Size	Ease of Integration	Safety and Durability	Life Expectancy	Total
Cost	0	0	0	0	1	1
Size	1	0	0	0	1	2
Ease of Integration	1	1	0	0.5	1	3.5
Safety and Durability	1	1	0.5	0	1	3.5
Life Expectancy	0	0	0	0	0	0
<b>Totals of Totals</b>	1.5	8	12	13.5	5.5	-

The team put large emphasis on the safety and durability of the battery to ensure that in using the device, it would not cause additional harm to the user and patient. In addition, it was determined that for both prototyping purposes and usage by the customer, the battery is easy to both wire and replace. Because batteries don't differ greatly by price, cost was not very important to us.

Here are the options we evaluated:

1) 9V



Figure 11. 9V Battery

The 9V battery stood out because of how easy it is to use. Being 9V, we require only one to power the microcontroller. Because both positive and negative leads are on the side side, it is easy to wire such that it can be replaced easily, giving it the highest score in terms of ease of integration [20].

2) AA



Figure 12. AA Battery

Although a single AA battery is relatively small in size, each battery has a voltage of 1.5V. As such, we would need four AA batteries in series to power a 5V microcontroller. This makes AA batteries very large and difficult to both wire and replace [19].



### 3) 3V Disk



Figure 13. 3V Disk

The 3V disk batteries are by far the smallest available batteries for our application. In order to achieve the required 5V, two batteries would have to be stacked but because of the nature of the battery, this is fairly simple. The only concern is that when building the chassis, we would have to ensure that there are leads on both the top and bottom off the case to attach to the positive and negative terminals of the battery. This causes an increase in complexity in the chassis and potentially difficulties in preliminary testing. At the end of the day, it is a trade between size and ease of integration [9].

Here are the results of the value analysis:

Table 3. Battery Value Analysis

Product	Cost	Size	Ease of Integration	Safety and Durability	Life Expectancy	Total
9V	2	2	3	3	1	101
AA	3	1	1	3	3	81.5
3V	2	3	2	3	1	97

As can be seen, it was decided that the 9V battery would be best for usage in a prototype. A deciding factor in this decision was its ability to be wired and replaced easily. One 9V battery is enough to drive the controller, is commonly used in similar applications, and can be wired with access to only the top of the battery. Although it is slightly bulkier than the alternative of using 3V batteries, its simple to use battery clips make it the best choice.

### 3.2.2. On/Off Switch

The on/off switch ensures that the device does not consume power when not in use.

**Value Criteria:**

- Cost
- Size
- Visibility

The main concern when choosing an on and off switch is whether or not it is quickly accessible when the device is needed. Most of this is handled in the design of the device chassis, but it would also be beneficial to have a switch that stands out.

Table 4. On/Off Switch Weight Assignments

<b>Ben</b>	<b>Cost</b>	<b>Size</b>	<b>Visibility</b>	<b>Total</b>
Cost		0	0	0
Size	1		0.5	1.5
Visibility	1	0.5		1.5
<b>Eric</b>	<b>Cost</b>	<b>Size</b>	<b>Visibility</b>	<b>Total</b>
Cost		0	0	0
Size	1		0	1
Visibility	1	1		2
<b>Zoey</b>	<b>Cost</b>	<b>Size</b>	<b>Visibility</b>	<b>Total</b>
Cost		0	0	0
Size	1		0	1
Visibility	1	1		2
<b>John</b>	<b>Cost</b>	<b>Size</b>	<b>Visibility</b>	<b>Total</b>
Cost		1	1	2
Size	0		0.5	0.5
Visibility	0	0.5		0.5
	<b>Cost</b>	<b>Size</b>	<b>Visibility</b>	<b>-</b>
<b>Totals of Totals</b>	2	4	6	

Here are the options we evaluated.

1) Slide Switch



Figure 14. Slide Switch

The slide switch costs \$2.00 and comes with 2 switches. These switches are relatively small and harder to access on devices than other switches. Because of their size, they are also harder to see or find on a device as well [39].

2) Cutler-Hammer



Figure 15. Cutler-Hammer

The Cutler-Hammer is a non-illuminated pushbutton that costs \$6.53 per unit. These buttons are much larger than the on-off switches above as they are 1.17 inches wide and 22.5mm in diameter. Its bright red color allows it to be much more visible on a device, and its larger size allows easier access for the user [17].

### 3) Red Rocker



Figure 16. Red Rocker

The Red Rocker is a rocker switch that is .6875 inches in width, making it larger than the slide switch and a bit smaller than the cutler-hammer pushbutton. This unit costs \$2.00 making it less expensive than the Cutler-Hammer pushbutton. Its bright red color coupled with its distinct on and off settings provide the user with good access and visibility [15].

Table 5. On/Off Value Analysis

Product	Cost	Size	Visibility	Total
Slide Switch	2	1	1	14
Cutler-Hammer	1	3	3	32
Red Rocker	2	2	3	30

After performing value analysis on the three different on/off switch options above, the team decided to use the cutler-hammer pushbutton in order to provide the best method of turning the device on and off. Despite cost, this pushbutton provides the user of the device the best possible accessibility and visibility when compared to the slide and rocker switches. The main reason why the group chose to use the pushbutton however is for safety. With switches such as the slide switch and rocker switch, it is possible that during use, the user could easily hit the switch and turn the device off. If the device is turned off in an emergency situation while being used, it could be detrimental to the patient's outcome. With the pushbutton, it would be extremely difficult to turn the device off while using it, as the user would have to push it with enough force perpendicular to the direction of the force being applied to the patient to deactivate it, rather than just hitting it by accident.

### 3.2.3. Accelerometer

The accelerometer is used to measure data that can then be used to determine compression depth

**Value Criteria:**

- Cost
- Size
- Range
- Frequency
- Requires Filtering/Amplification

CPR is fast paced and is delivered through high impulse chest compressions. It is important that accelerometer has a large enough range to capture this information without clipping the largest impulses.

In order to ensure that the device functions properly, we must be able to derive data about the position of the data from the device. The device must operate consistently in reporting information at the peak of the impulses. Under the assumption that the accelerometer will record accurate data at that magnitude, it must also be ensured that data samples are taken frequently enough to capture these near instantaneous peaks.

Depending on how the sensor outputs this information, it may need filtering or amplification. Both of these operations would require additional electronic circuitry which would contribute to the size of the device.

Here are the results of the weight assignments:

Table 6. Accelerometer Weight Assignments

Ben	Cost	Size	Range	Frequency	Filtering/Amplification	Total
Cost		0	0	0	0	0
Size	1		0.5	0	0	1.5
Range	1	0.5		0	0	1.5
Frequency	1	1	1		0.5	3.5
Filtering/Amplification	1	1	1	0.5		3.5
Eric	Cost	Size	Range	Frequency	Filtering/Amplification	Total
Cost		1	0	0	0	1
Size	0		0	0	0	0
Range	1	1		1	0	3
Frequency	1	1	0		0	2
Filtering/Amplification	1	1	1	1		4
Zoey	Cost	Size	Range	Frequency	Filtering/Amplification	Total
Cost		1	0	0	0	1
Size	0		0	0	0	0
Range	1	1		1	1	4
Frequency	1	1	0		1	3
Filtering/Amplification	1	1	0	0		2
John	Cost	Size	Range	Frequency	Filtering/Amplification	Total
Cost		0	0	0	0	0
Size	1		1	1	1	4
Range	1	0		1	1	3
Frequency	1	0	0		1	2
Filtering/Amplification	1	0	0	0		1
	Cost	Size	Range	Frequency	Filtering/Amplification	-
<b>Totals of Totals</b>	2	5.5	11.5	10.5	10.5	

The team put large emphasis on the range, frequency, and need of filtering or amplification for the accelerometers. This was used to ensure that the data acquired would not be subject to clipping and to minimize inaccuracy. This is primarily due to the nature of the product as a biomedical instrument. However, the size and cost of the accelerometer was still important in ensuring that the device as a whole would be a competitive alternative.

Here are the options we evaluated.

1) ADXL345

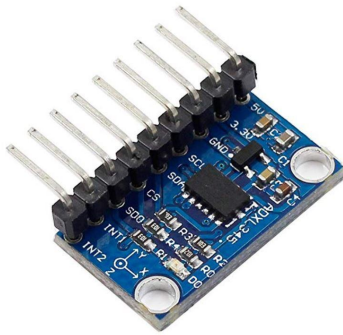


Figure 17. ADXL345 Accelerometer

The ADXL345 is an accelerometer with 13-bit resolution which can measure up to +/- 16g. Its available range far exceeds anything we would expect to see in our application which makes it a good choice in ensuring that the data will not be cut off due to excessive force. However, it is expensive at \$17.52 [41].

2) MPU6050

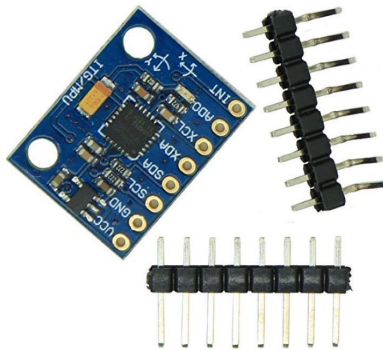


Figure 18. MPU6050 Accelerometer

The MPU6050 rivals the ADXL345 in range and sensitivity with a 16-bit resolution that also measures up to +/- 16g. Having the same SCL clock frequency and approximate size, the MPU6050 is simply superior to the ADXL345 for our purposes because of the decreased cost at \$5.88 (GY-521).

### 3) MMA8452

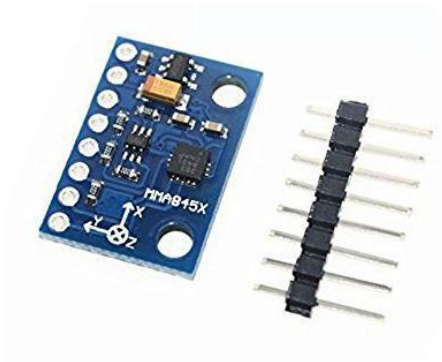


Figure 19. MMA8452

The MMA8452 accelerometer is an accelerometer that is similar to the MPU6050. It is slightly more expensive, but also has a smaller range and resolution, making it an inferior alternative [47].

Here are the results of our value analysis:

Table 7. Accelerometer Value Analysis

Product	Cost	Size	Range	Frequency	Filtering/Amplification	Total
ADXL345	1	3	3	3	3	116
MPU6050	3	3	3	3	3	120
MMA8452Q	3	3	2	3	2	98

The accelerometers were very similar in operation, with the first two having the capability of measuring up to 16g in acceleration with a standard 400kHz response rate using the I2C serial protocol. Being all relatively similar in size, the deciding factor between the ADXL345 and MPU6050 ended up being the price. Costing only \$5.88, the MPU6050 will be used in the prototype.

#### 3.2.4. Microcontroller

The microcontroller processes the information from the accelerometer to find compression depth; it then uses this information to control the buzzer and LEDs.

##### Value Criteria:

- Cost
- Size



- Analog to Digital Converter (ADC)
- Digital to Analog Converter (DAC)
- Pulse Width Modulation (PWM)

It is still undetermined the method by which the accelerometer will output data. As such, it is important to have a microcontroller that is versatile enough to convert between digital and analog data. It should also be able to receive and send pulse width modulated (PWM) signals.

Here are the results of the weight assignments:

Table 8. Microcontroller Weight Analysis

Ben	Cost	Size	ADC	DAC	PWM	Total
Cost		0	0	0	0	0
Size	1		0	1	0	2
ADC	1	1		1	0	3
DAC	1	0	0		0	1
PWM	1	1	1	1		4
Eric	Cost	Size	ADC	DAC	PWM	Total
Cost		0	0	1	0	1
Size	1		0	0	0	1
ADC	1	1		1	0	3
DAC	0	1	0		0	1
PWM	1	1	1	1		4
Zoey	Cost	Size	ADC	DAC	PWM	Total
Cost		1	0	0	0	1
Size	0		0	0	0	0
ADC	1	1		1	0	3
DAC	1	1	0		0	2
PWM	1	1	1	1		4
John	Cost	Size	ADC	DAC	PWM	Total
Cost		0	0	1	0	1
Size	1		0	0	0	1
ADC	1	1		1	0	3
DAC	0	1	0		0	1
PWM	1	1	1	1		4
<b>Totals of Totals</b>	<b>3</b>	<b>4</b>	<b>12</b>	<b>5</b>	<b>16</b>	<b>-</b>

As can be seen from the weight assignments, a large emphasis was put on the ability of the microcontroller to send and receive PWM signals and convert analog data to digital data. This is primarily due to the fact that most buzzers and accelerometers operate using PWM signals. However, if the operation of an analog accelerometer far exceeded that of a digital one, it would be preferable to have the option to use it as well. However, analog output devices would not be

likely to vary in function as drastically and therefore the value of a DAC was negligible. It was also important that the microcontroller was small enough so it could be contained easily within the chassis.

Here are the options we evaluated.

1) Arduino Uno



Figure 20. Arduino Uno

The Arduino Uno is a highly versatile general usage microcontroller for testing purposes. It has enough functionality and pins to support most small-scale projects as well as a well-documented and user-friendly IDE. The Arduino Uno is a strong starting point to prototype most devices. In this project, the main concern with using an Arduino Uno is simply its size. At 68.6mm x 53.4mm, it is very large and would easily take up a majority of the available space inside of the device [4].

2) Sparkfun Pro Micro

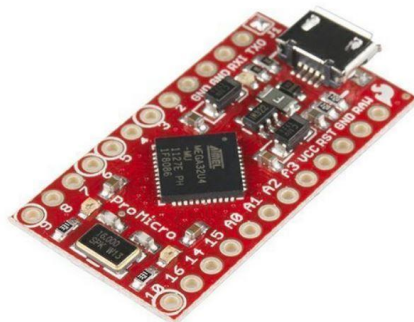


Figure 21. Sparkfun Pro Micro

The Sparkfun Pro Micro is one of the smallest microcontrollers on the market. With ADC and PWM support, it would easily support much of the functionality required for this project. It

measures at about 33mm x 18mm, making it a much smaller and more suitable choice for this project than the Arduino Uno. One concern is its reasonably high price of \$19.95 [40].

### 3) Arduino Mini Nano

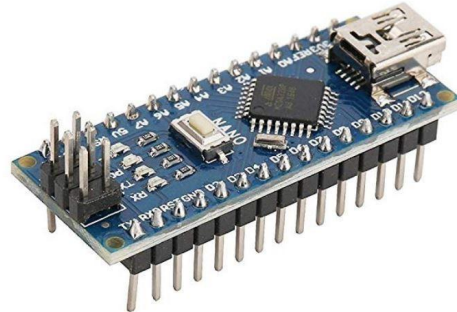


Figure 22. Arduino Mini Nano

The Arduino Mini Nano is the smallest microcontroller created by Arduino. It has similar functionality to the Sparkfun Micro but is slightly smaller at 45mm x 18mm. However, it is also half the price and has integrated support in the IDE. The reduced cost of \$6.99 makes this a superior alternative to the Sparkfun Micro [28].

Here are the results of the value analysis:

Table 9. Microcontroller Value Analysis

Product	Cost	Size	ADC	DAC	PWM	Total
Arduino	2	1	3	1	3	99
Sparkfun Micro	1	3	3	1	3	104
Mini Nano	3	3	3	1	3	110

All of the controllers had standard features including PWM and ADC pins, but none had included a DAC. As such, the final decision came down to the cost and size of the microcontroller. The Sparkfun Micro and Arduino Mini Nano are comparable in size but being that the Arduino Mini Nano is over half the price, it is the microcontroller that we will be using in this project.

#### 3.2.5. Buzzer

The buzzer provides audio feedback regarding the performance of the user.

##### Value Criteria:

- Cost
- Size
- Loudness
- Range of Feedback

If the buzzer is too soft, the user may not be able to hear it as they are performing CPR. If it is too loud, it may become disorienting or distracting for the user.

If the buzzer can output a range of frequencies, the pitch may be altered to send more complex information regarding their performance.

Table 10. Buzzer Weight Analysis

<b>Ben</b>	<b>Cost</b>	<b>Size</b>	<b>Loudness</b>	<b>Range of Feedback</b>	<b>Total</b>
Cost	0	0	0	0	0
Size	1	0	0	1	2
Loudness	1	1	0	1	3
Range of Feedback	1	0	0	0	1
<b>Eric</b>	<b>Cost</b>	<b>Size</b>	<b>Loudness</b>	<b>Range of Feedback</b>	<b>Total</b>
Cost	1	1	1	1	3
Size	0	0	1	1	2
Loudness	0	0	0	1	1
Range of Feedback	0	0	0	0	0
<b>Zoey</b>	<b>Cost</b>	<b>Size</b>	<b>Loudness</b>	<b>Range of Feedback</b>	<b>Total</b>
Cost	1	0	0	0	1
Size	0	0	0	1	1
Loudness	1	1	0	0	2
Range of Feedback	0	0	0	0	0
<b>John</b>	<b>Cost</b>	<b>Size</b>	<b>Loudness</b>	<b>Range of Feedback</b>	<b>Total</b>
Cost	1	0	0	1	2
Size	0	0	0	1	1
Loudness	1	1	0	1	3
Range of Feedback	0	0	0	0	0
	<b>Cost</b>	<b>Size</b>	<b>Loudness</b>	<b>Range of Feedback</b>	<b>-</b>
<b>Totals of Totals</b>	6	6	9	1	

Here are the options we evaluated.

1) Single Tone Buzzer



Figure 23. Single Tone Buzzer

This single tone buzzer produces sound at 3kHz and produces a sound approximately between approximately 78-100 decibels. It operates at a voltage supply between 3V-18V. It has a diameter of 25.1mm and a height of 22.2mm height. It costs \$2.95 [30].

2) Mini Speaker



Figure 24. Mini Speaker

This mini speaker is much smaller than the single tone buzzer with a diameter and height of 12mm. It is also less expensive than the single tone buzzer at a cost of \$1.95. It operates with a supply voltage of 3.3V and produces a sound approximately between 85-95 decibels. This mini speaker also responds best with a square wave, which can be produced by a microcontroller [46].

### 3) 16 Tone Buzzer



Figure 25. 16 Tone Buzzer

This large piezoelectric 16 tone alarm is four times the price of the mini speaker above, however, it can operate with a range of 16 different tones that could be useful in providing the necessary information to the user of the device. This buzzer operates with a supply voltage between 5-16V. Compared to the the mini speaker which is very small at 12 mm diameter, this device is much larger with a diameter of 35.8 mm and a height of 35.7 mm. At a 5V supply, this device can produce a sound registered at approximately 90 decibels. At a 16V supply, this device can produce a sound registered at approximately 103 decibels [29].

Table 11. Buzzer Value Analysis

Product	Cost	Size	Loudness	Range of Feedback	Total
technically speaker	3	3	3	3	66
16 tone buzzer	2	2	3	3	54
regular buzzer	3	2	3	1	58

After performing value analysis on the three different buzzers above, the team decided to use the mini speaker to produce the audio feedback necessary during CPR. At the lowest cost, it will allow the team to produce the device and also market it at a lower cost. This speaker also produces a loud enough sound for a person performing CPR to hear and follow along to while administering compressions. This speaker also requires an input voltage of 5V which will be provided by the use of a 9V battery within the device. This speaker is also toned to respond best with a square wave which can be produced by a microprocessor, which the team will be using within the device.

### 3.2.6. LEDs

The LEDs provide visual feedback regarding the performance of the user.

**Value Criteria:**

- Cost
- Size
- Color
- Brightness

Having a variety of colors available would provide the opportunity to use colored signals to provide feedback to the user.

Depending on the location that the device is used, the LED may not be visible if it is not bright enough. However, if it is too bright, it may also disorient or blind the user.

Here are the weight assignments for the LEDs.

Table 12. LED Weight Analysis

Ben	Cost	Size	Color	Brightness	Total
Cost		0	0	0	0
Size	1		1	0	2
Color	1	0		0	1
Brightness	1	1	1		3
Eric	Cost	Size	Color	Brightness	Total
Cost		0	0	0	0
Size	1		0	0	1
Color	1	1		0	2
Brightness	1	1	1		3
Zoey	Cost	Size	Color	Brightness	Total
Cost		0	0	0	0
Size	1		0	0	1
Color	1	1		0.5	2.5
Brightness	1	1	0.5		2.5
John	Cost	Size	Color	Brightness	Total
Cost		0	0	0	0
Size	1		0	0	1
Color	1	1		0.5	2.5
Brightness	1	1	0.5		2.5
Totals of Totals	Cost	Size	Color	Brightness	-
	0	5	8	11	

As one can see above, the group felt that the cost of the LEDs used for the CPR assistive device did not significant, compared to the brightness of the LEDs. The color of the LEDs was also important because the group wanted LEDs that had the option of different colors, rather than just one. The size of the LEDs was not that important compared to other criteria because most LEDs are relatively small compared to other components used in this device, however, the smaller the better.

Here are the options we evaluated.

1) 10mm Red



Figure 26. 10mm Red LED

The 10mm Red LED is very inexpensive at \$0.55 per LED, but is very large at 10 mm diameter. It is made with an opaque epoxy package that does not produce the brightest of lights, but rather a soft glow. It operates with a supply voltage between 1.8-2.2V [14].

2) RGB Clear

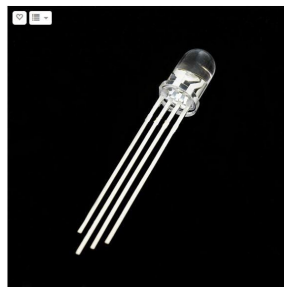


Figure 27. RGB Clear LED

The RBG Clear LED can produce clear and bright light in three colors: red, blue and green. These LED's cost \$2.05 per LED, but are half the size of the 10mm Red LEDs above, at 5 mm. These LEDs also have the feature to pulse width modulation all three colors of red, blue, and green to produce a variety of mixed colors (LED - RGB Clear).



### 3) RGB Addressable



Figure 28. RGB Addressable LED

The RGB Addressable LED is relatively intermediate in size, standing at 8 mm in diameter, when compared to the RGB Clear LEDs and the 10mm Red LEDs. These LEDs require a bit more supply voltage than the previous two however, as they require 4.5-6V. These LEDs outer shell is also an opaque cover, which therefore dims the light that they produce. They do however produce a wide variety of colors just like the RGB Clear LEDs [32].

### 4) 25 pack 5mm red

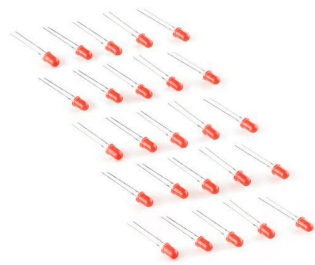


Figure 29. 25 pack 5mm red LED

This 25 pack of 5 mm red LEDs are the least expensive of all 4 options at approximately \$0.14 per LED. They are also the smallest LEDs tied with the RGB Clear LEDs at 5 mm in diameter. These LEDs however produce dim light due to an opaque cover and can only produce red light rather than a variety of colors [31].

Table 13. LED Value Analysis

Product	Cost	Size	Color	Brightness	Total
RGB Clear	1	3	3	3	72
10 mm Red	3	1	1	1	24
RGB Adressable	3	2	3	2	56
25 pack 5mm red	3	3	1	1	34

After performing value analysis on the four different LED options above, the team decided to use the RGB Clear LEDs in order to produce the best possible visual feedback to the user of the device while performing CPR. Although being the most expensive LED, the RGB Clear LED produces the brightest light, the widest variety of light, and is the smallest LED of the 4 (tied with the 5mm red ones).

## Chapter 4. Prototyping

### 4.1. Factor of Safety

When designing a case to support any load, one must first determine standards such as the factor of safety. It is defined as the ratio of the yield strength of a chosen material and the max yield strength of the design. Mathematically it is  $\delta = YS/\sigma_{max}$ , however you must first determine a minimum safety factor value. This minimum value is determined by various factors such as calculation accuracy, use of the device, and chosen materials. Typically, the ranges are from 1.3-1.4 being a design that has very accurate calculations, is made from a uniform material, and has a precise design. The ranges from 1.4 to 1.7 typically indicate your typical working conditions. The ranges from 1.7 - 3.0 will typically indicate unfavorable and nonstandard conditions [36]. When looking at our device, the condition which the device will always see is a load of approximately 100-125 lbs of force [16]. The device itself could be used in any location where someone needs CPR, which could be anywhere. The device will also be printed in PLA, so structurally, the device is not made of a uniform material. These factors shaped our decision to use a safety factor of 3 to properly consider numerous external factors coming into play when using this device. Once we have a minimum safety factor, we can use SolidWorks Simulation on the designs to test for maximum stress values on the device. From there we can use the appropriate governing equations, Eqs (number, as listed in report) to solve for the actual factor of safety, and compare it to our set minimum. If our calculated value is less than the minimum safety factor we determined, then the case does not meet our standards and changes must be made. If our calculated value is greater than the value of our determined safety factor than the case is acceptable from a load bearing standpoint.

## 4.2. Stress Tests

In order to determine if the device design can withstand the forces associated with normal use, we will use SolidWorks Simulation to add loads parts of the design and using the fixation option we are able to make the device seem as if it is sitting on top of a flat platform to recreate the situations as seen in Figures 30-32. In order to properly test the device. The load area is determined by using anthropometric data such as data on the average human hand and the specific location of the load will be where the hand should be placed when using the device as seen in Figure A. The maximum amount of force applied will be the normal average forces applied during CPR, which is 120-125 lbs. of compressive force [16]. In the US, the average palm length of men ranges from 10.35cm to 11.54cm and the average palm length of women is 9.5cm to 9.88cm. For our case, we will use the median of the two average values, 10.95cm and 9.69cm respectively, and test both male and female palm size. Once the loads have been applied in the correct locations and the device has been fixed, we can run the simulation. The simulation then gives back a series of stress values in different locations all over the device. We then pick the highest stress value to plug into our safety factor formula to determine if the device is structurally sound.

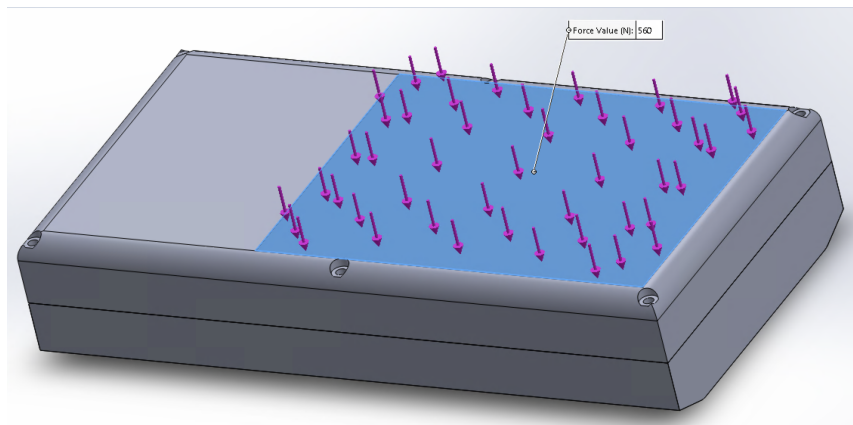


Figure 30. Image of stress test showing force placement

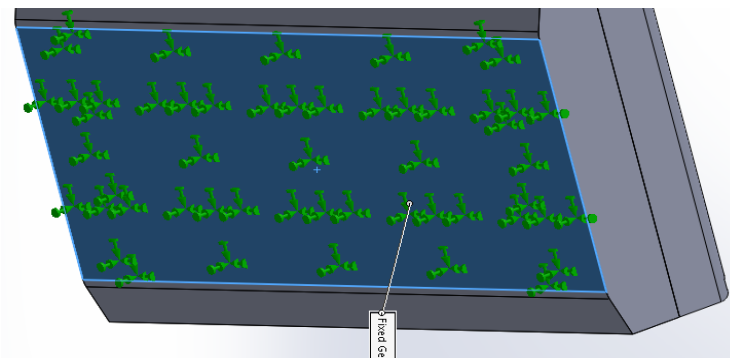


Figure 31. Image of stress test showing location of fixation

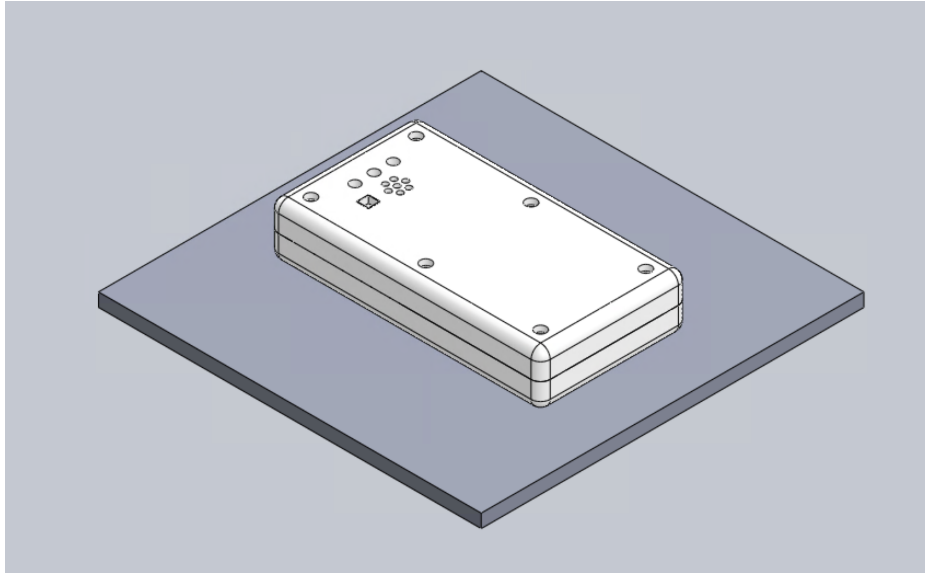


Figure 32. Image of device on a flat surface

### 4.3. Material Selection

The material that the case is made of is also a large factor in designing it to be structurally sound. There are important factors such as how well it molds and its yield strength that are used to help narrow down the search for a material. The first material we considered was Acrylonitrile Butadiene Styrene (ABS). The material is a commonly used in injection molding, however its yield strength was only between 42-46 MPa, making it a fairly weak material. We then looked at a different material called Styrene acrylonitrile (SAN). The material is stronger with a yield strength of 68.4-82.1 MPa making it a much better candidate than the ABS. SAN is similar to ABS, but replacing ABS with SAN allows us to get a larger yield and compressive strength. From there we now look at different versions of ABS and SAN for injection molding. The next material to look at was 30% glass filled ABS and 30% glass filled SAN, both of which are used in injection molding processes. The cost per kg of the two different materials are 3.04 - 3.19 USD 2.97-3.22 USD and respectively. Since their costs are relatively similar, we can eliminate cost as a factor in the decision-making process. Due to the constant loads being applied to this device, we have determined to go with 30% glass filled SAN due to its highest yield strength of the 4 being 110-120 MPa [8].

### 4.3.1. Design 1

The goal of the first device was to determine the relative size of the device and start with component placement as seen in Figure 33. The basic size of the device was adapted from the earlier anthropometric justification. The size of this device is 160mm X 80mm. The starting thickness of the case is 6mm and the screw holes are M2 through holes. We wanted the device to be handheld, so that along with the anthropometric data we designed the case to resemble the dimensions of a cell phone. The bottom of the device is chamfered in order to better fit in the groove of the human chest as seen in Figure 34. In order to determine if the device meets our structural standards, we ran two stress tests on it, one for the average male hand, and one for the average female hand as seen in Figures 35 and 36 respectively. From there we can determine our safety factor for this device. Since our material of choice is 30% glass filled SAN, we have a yield strength of 16679.3 psi, so the safety factor of the device for a woman is  $\delta = 16679.3/737.1 = 22.6$ . Since  $22.6 > 3$  and 3 was our set safety factor, structurally this device is strong enough to support its planned use. For a man the safety factor is  $\delta = 16679.3/711.7 = 23.43$  and  $23.43 > 3$  so the device is strong enough to support its planned use. Since the device passes our safety factor standards for both men and women, its structure is acceptable.

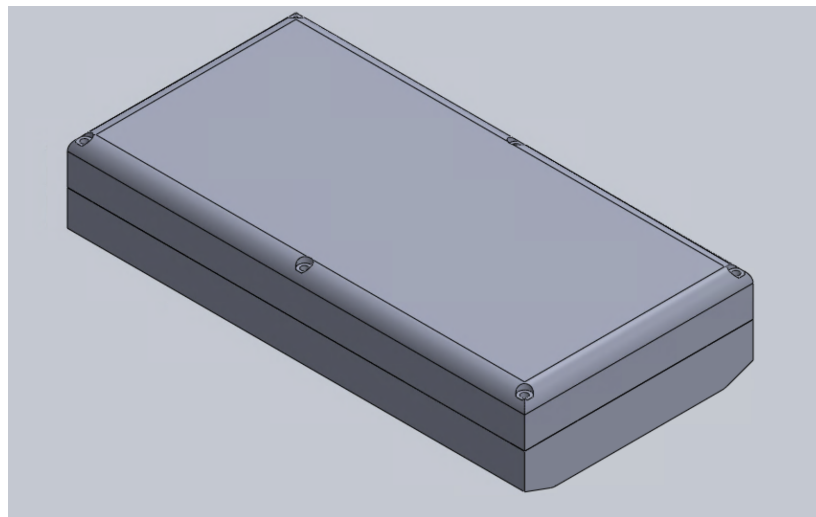


Figure 33. Isometric view of Design 1



Figure 34. Side view of bottom half of device showing chamfer

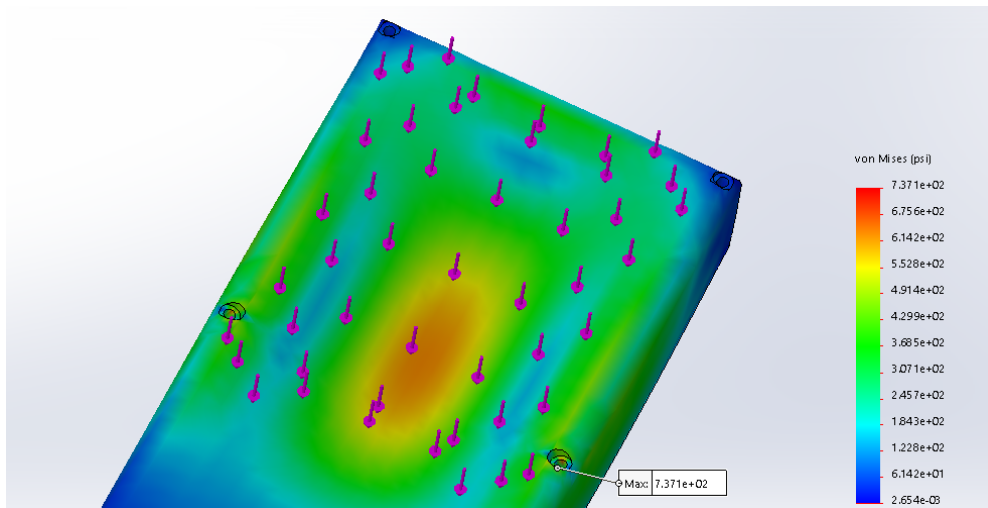


Figure 35. Stress test on Design 1 for average female palm size

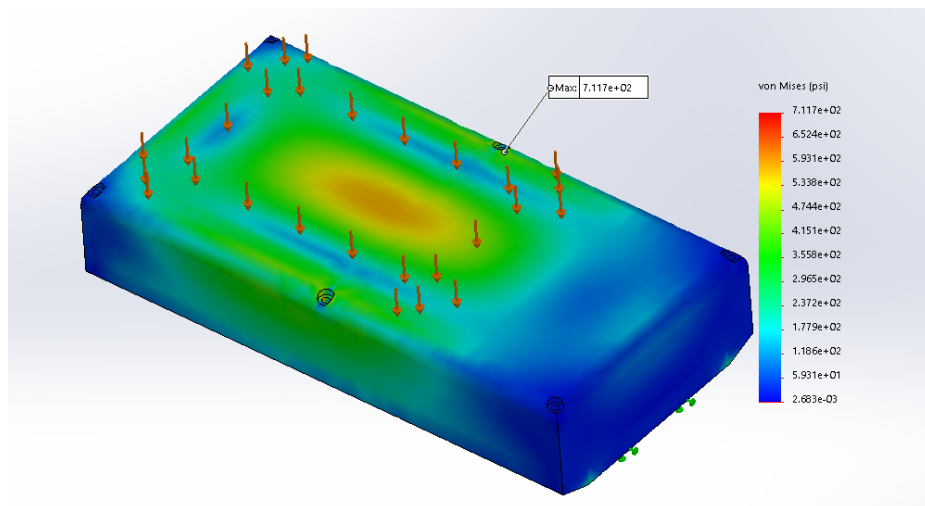


Figure 36. Stress test on design 1 for average male palm size

### 4.3.2. Design 2

A second design was developed concurrently to test the placement of the various components and determine the size requirements of the device. This was done with the intention of maintaining a 2:1 length-to-width ratio, similar to that of a phone, while minimizing the height. The battery, with dimensions of 48mm x 25mm x 15mm, was much larger than the other components, making it the limiting factor in the design. Using these parameters and the measurements of the components from their respective datasheets, a layout to minimize the space required was designed, seen in Figure 37.

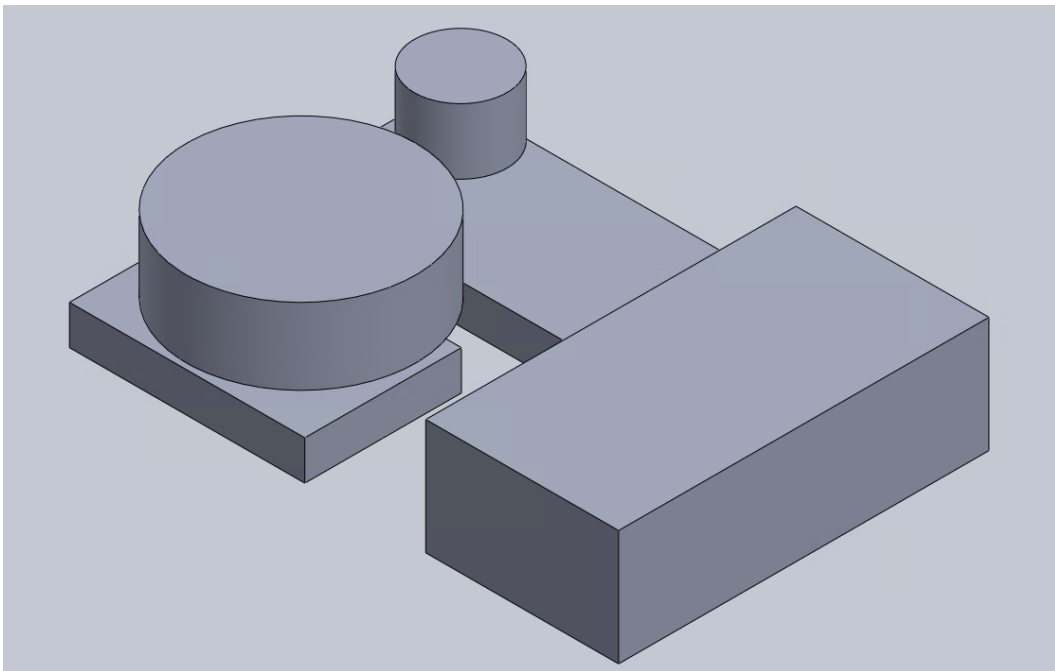


Figure 37. Configuration of Components with 2:1 Length-to-Width Ratio and Minimum Height

The parts fit in a 70.72mm x 48mm x 15mm area. This and a minimum wall thickness of 10mm was used to design a prototype which would adequately house the selected components, seen in Figure 38.

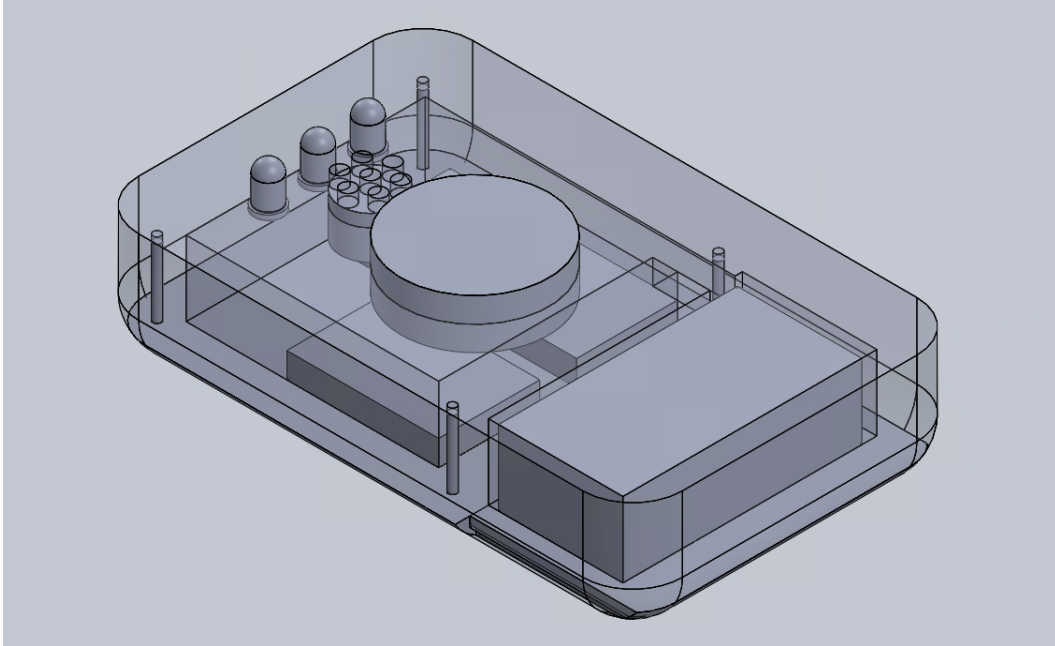


Figure 38. Initial Prototype to House Components with Wall Thickness of 10mm

The size of the button was determined to be too large and a new button was selected. The button was replaced and the wall thickness was adjusted to accommodate for the potential of injection molding. This design was then printed using PLA for additional revision. The design and the printed prototype can be seen in Figure 39 - 41, respectively.

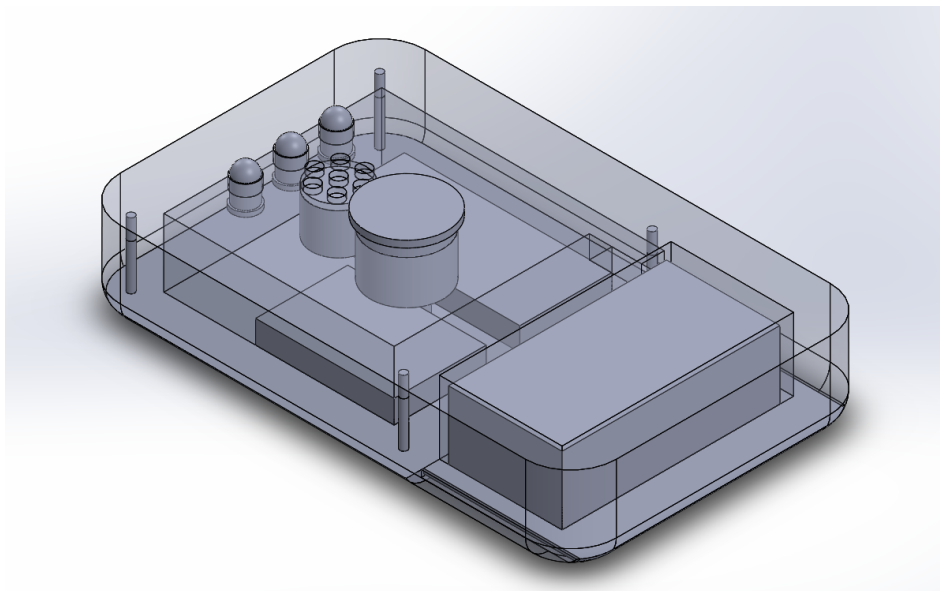


Figure 39. Adjusted Prototype with Smaller Button and Wall Thickness of 4mm





Figure 40. Initial Printed Prototype for Configuration of Electrical Components (Top)



Figure 41. Initial Printed Prototype for Configuration of Electrical Components (Bottom)

### 4.3.3. Design 3

Design 3 is the first draft of the device when we determined that we will be using injection molding to mass produce the case. Using injection molding design guidelines, we were able to further the device. The guidelines recommend using bosses instead of holes in the walls in order to put the two halves together. Doing this helps reduce warpage during the cooling phase of injection molding. Another part of injection molding is cost. In injection molding, one wants to keep everything as thin as possible to keep cooling time and therefore costs down and recommends a thickness between 2-4mm. As seen in Figure 42, the walls of the device were thinned from 3mm to 6mm and the holes for M2 screws inside the walls has been moved to the bosses as seen also in Figure 42. Since we thinned out also in this design, we need to support the weight a different way. The recommended way for helping to support loads is ribbing as seen in Figure 39. Another way to help with support is remove sharp edges, so areas such as when the ribbing meets the walls or when the bosses meet the bottom, we have filleted accordingly as seen in Figure 43. As seen in Figure 44, there are now 3 holes in place at the top of the device for LED. They were placed there instead of on the side of the device because the lights will be more visible facing up and if they were facing sideways there is a potential for more obese individuals to cover the LEDs during compressions. The LEDs are not on the bottom of the device because they would be covered during use and be unable to alert the rescuer. As seen in Figure 44 as well, the speaker has also been placed directly below the LEDs and is facing upwards so that the sound will travel directly from the device, upwards to the rescuer. Also, in Figure 44, the hole for the button has been placed directly below the speaker in order to turn on the device. We have also added a bar across the device in order to divide where you should place your hands while performing CPR from the electronic components and button that can also be seen in Figure 44. In doing this it will help to reduce the chances off accidentally powering off the device while performing CPR. Since there were so many design changes, it is important that we run another stress test for both the male and female palm size to confirm that the case itself is structurally sound. For the female palm,  $\delta = 16679.3/5417 = 3.079$  and for the male palm  $\delta = 16679.3/4941 = 3.38$ . Since both 3.079 and 3.38 are greater than 3, from a structural standpoint this design can be used as planned. See Figures 45 and 46.

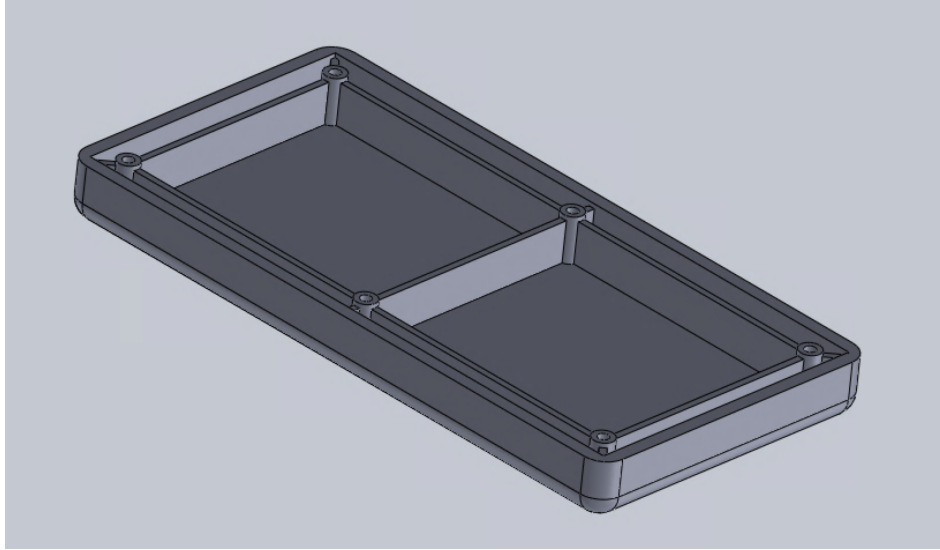


Figure 42. Image of design 3 where bosses, thinning of walls, and ribbing can be seen

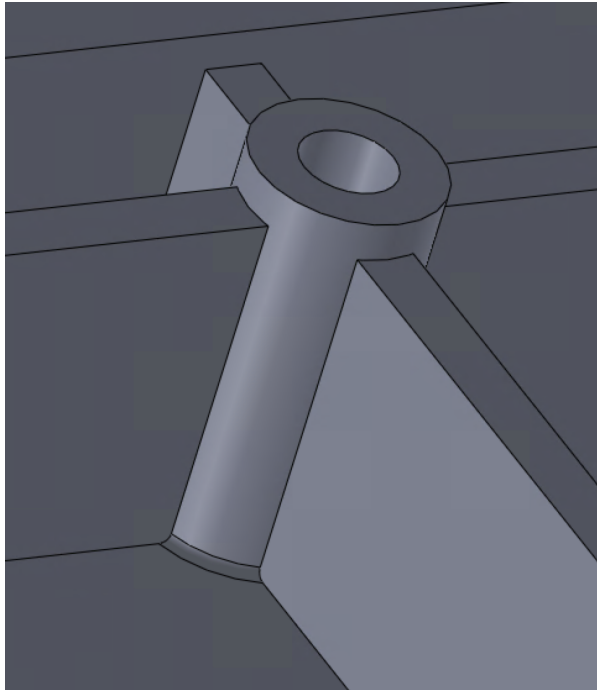


Figure 43. Image of design 3 filleting at bottom of boss

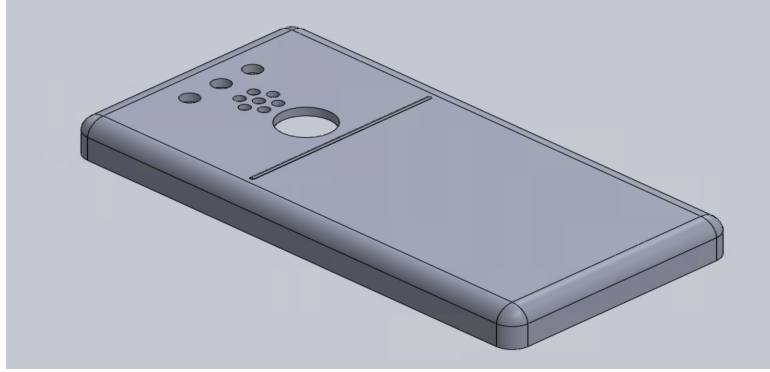


Figure 44. Image of Design 3 displaying LED speaker, button holes, and hand blocking bar

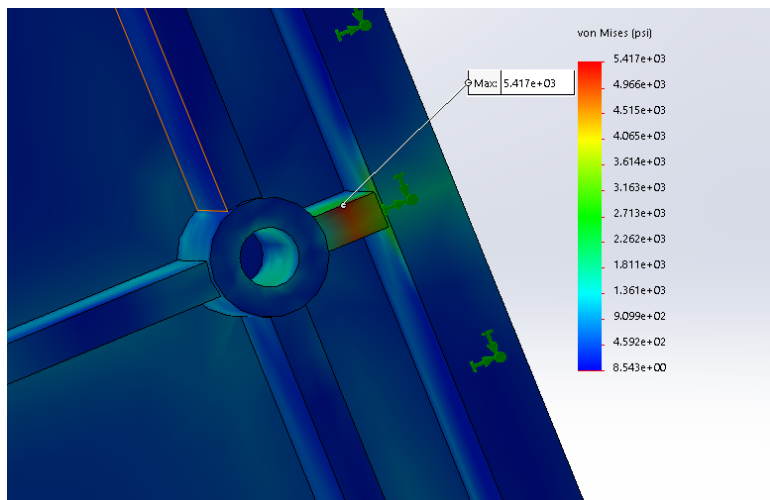


Figure 45. Stress test on design 3 for average female palm size

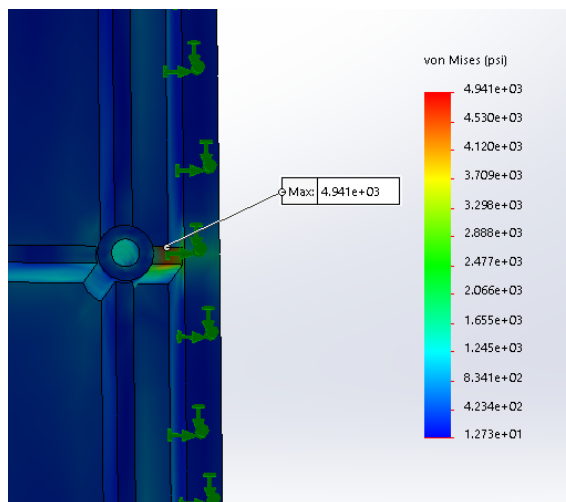


Figure 46. Stress test on design 3 for average male palm size

#### 4.3.4. Design 4

At this point in the design iterations, we are only focusing on the structural support design. While our previous design was able to pass our safety factor, we would like to have a value significantly higher than 3 just to be positive that this device is not going to break. The ribbing was redesigned in two different ways which can be seen in Figures 47 and 48 in an effort to increase the strength of the case. In Figure 48 the ribbing was designed to encase each individual electronic component while also providing support. Next, we need to do stress tests with both male and female palm sizes to confirm that the device can support the necessary weight. For Figure 48 is female palm size safety factor we got  $\delta = 16679.3/2624 = 6.35$  and for the male palm size we got  $\delta = 16679.3/2471 = 6.75$ . Since both 3.079 and 3.38 are greater than 3, from a structural standpoint this design can be used as planned. See Figure 49. In Figure 45, the ribbing was designed purely for support and the electronic component placement would be done in a later revision. After doing the stress analysis for the female palm size we got  $\delta = 16679.3/2562 = 6.51$  and for male palm size we got  $\delta = 16679.3/2618 = 6.37$ . Since both 6.51 and 6.37 are greater than 3, from a structural standpoint this device can be used as planned. See Figures 50 and 51. After discussing the designs with the team however, we determined that the ribbing looked too complex and we wanted a more similar solution to design 2, which led us to Design 5.

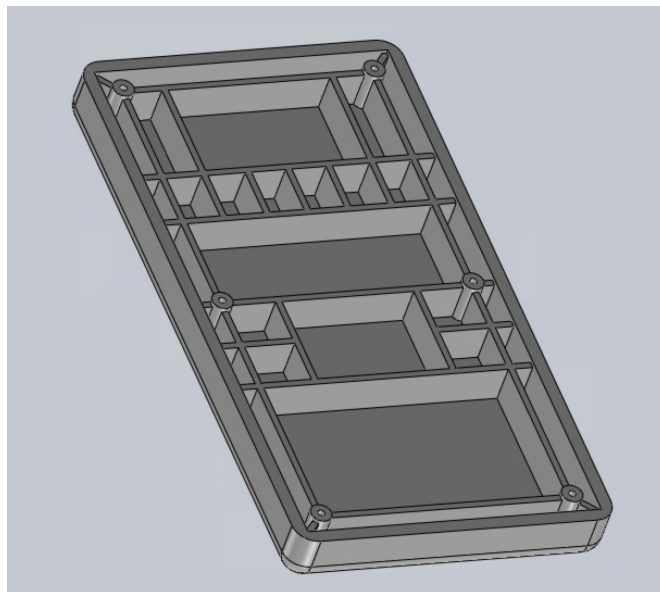


Figure 47. Design 4 ribbing to hold each electrical component in place

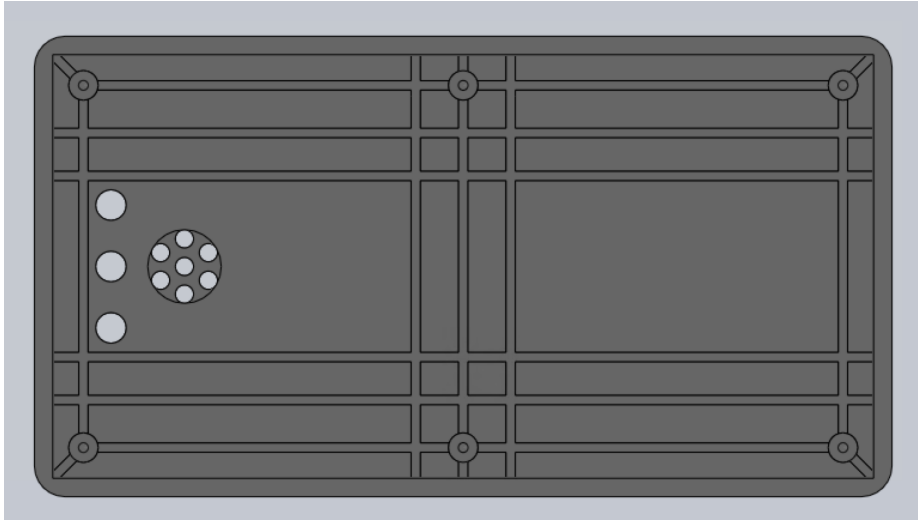


Figure 48. Design 4 ribbing purely for structural support

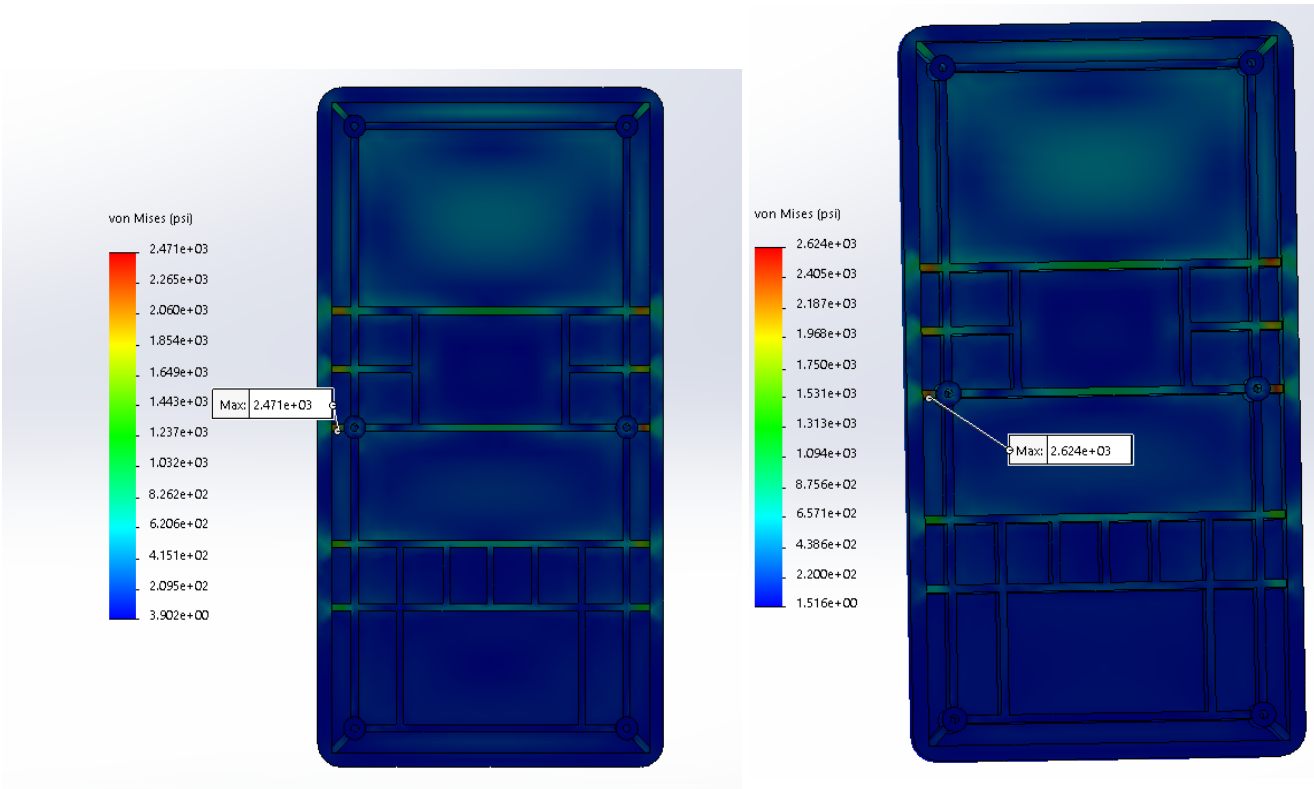


Figure 49. Stress Tests on Design 4 for average female and male palm size respectively

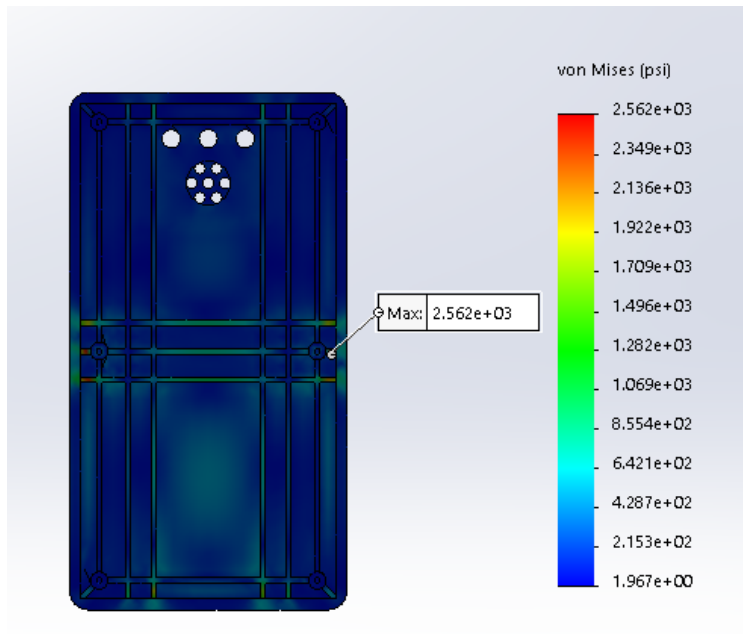


Figure 50. Stress test on design 4 for average female palm size

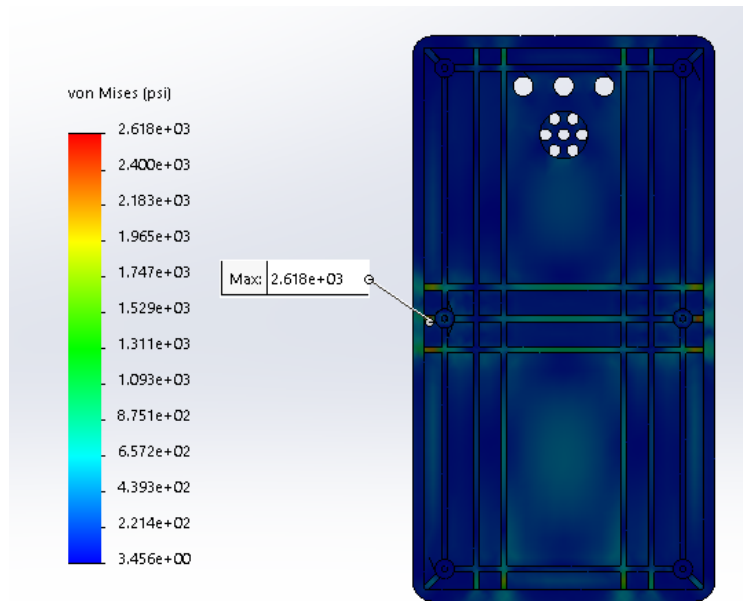


Figure 51. Stress test on design 4 for average male palm size

#### 4.3.5. Design 5

In order to keep the design simple like as in design 2, our first thought as a team was to just increase the thickness of the walls and ribbing, therefore we increased the thickness of the walls and ribbing to be 4mm (See Figure 52). Originally, our ribbing on the top and bottom halves did not touch, so in this design, our differing factor was whether or not there was a gap between the halves of ribbing. Design 5 has a 1mm gap between the ribbing on the two halves and design 4b has no gap between the ribbings as seen in Figures 52 and 53. The next step is to analyze the stresses with the applied force on the designs. For design 5a, we determined that the safety factor for a female palm size is  $\delta = 16679.3/3314 = 5.03$  and for the male palm size  $\delta = 16679.3/3184 = 5.24$ . Since both 5.03 and 5.24 are greater than 3, from a structural standpoint this device is acceptable (see Figures 54 and 55). For design 5b, we determined that the safety factor for a female palm size is  $\delta = 16679.3/381.4 = 43.73$  and for the male palm size  $\delta = 16679.3/365.4 = 45.65$ . Since both 43.73 and 45.65 are greater than 3, from a structural standpoint this device is also acceptable (See Figures 56 and 57). When looking at the two options, technically speaking both will work. Since design 5b has such a larger safety factor than even the first design, we will be using that design for our device.

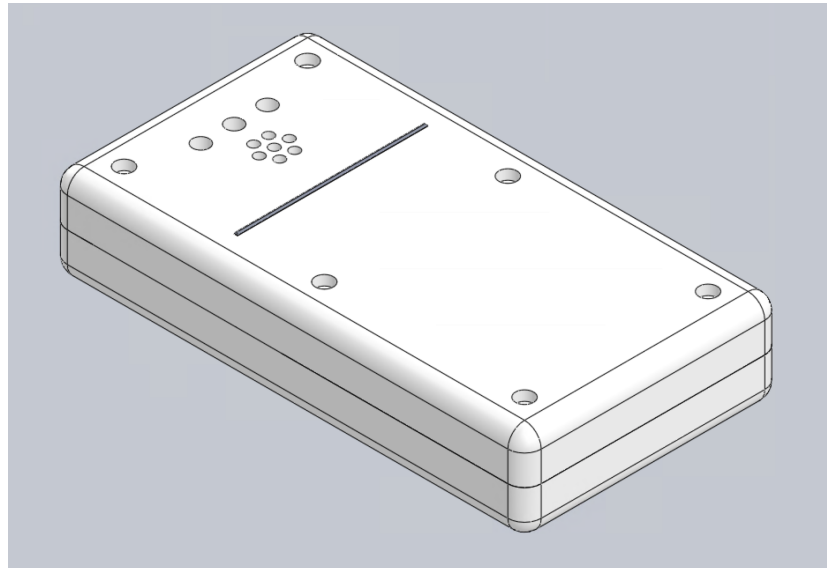


Figure 52. Assembly View of Design 5



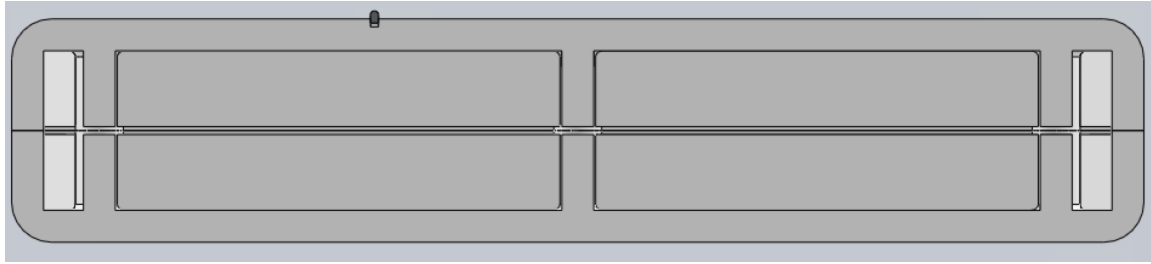


Figure 53. Section View inside design 5a to see 1mm gap in ribbing

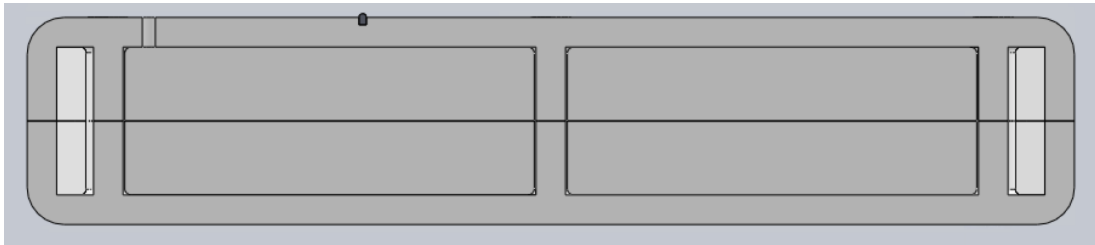


Figure 54. Section view inside design 5b to see no gap in ribbing

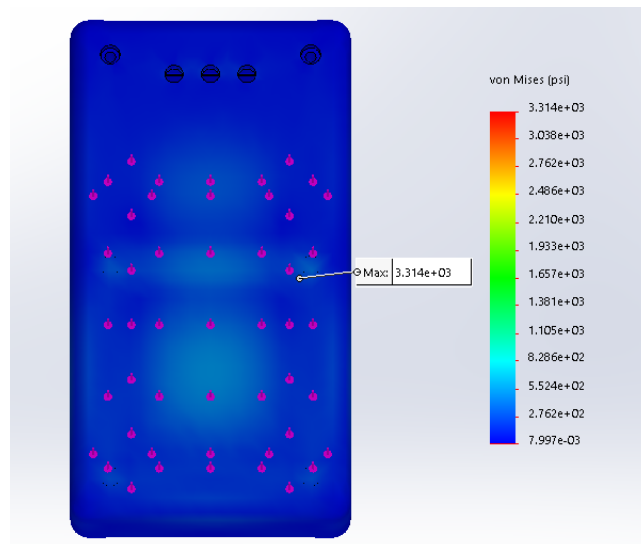


Figure 55. Stress test on design 5a for average female palm size

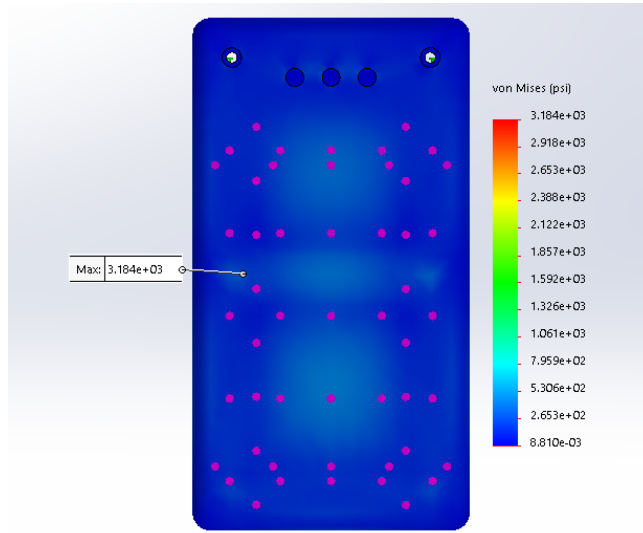


Figure 56. Stress test on design 5a for average male palm size

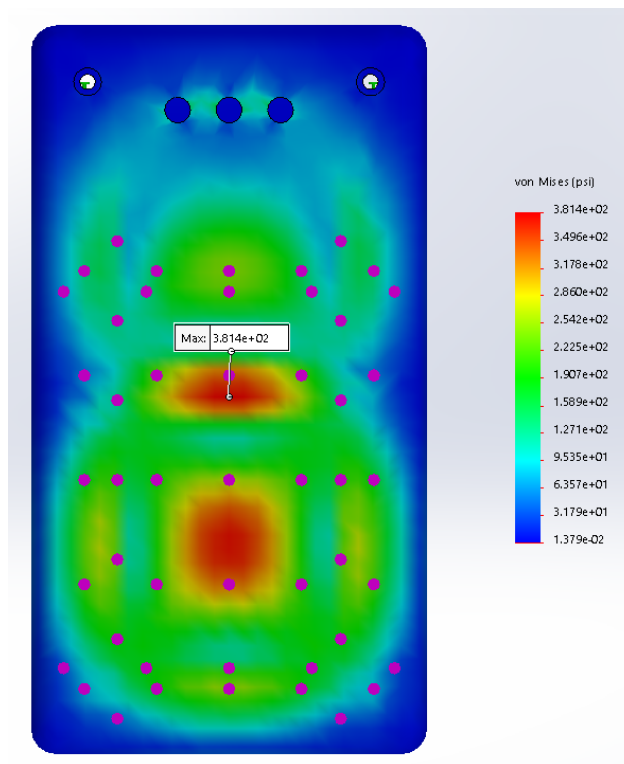


Figure 57. Stress test on design 5b for average female palm size

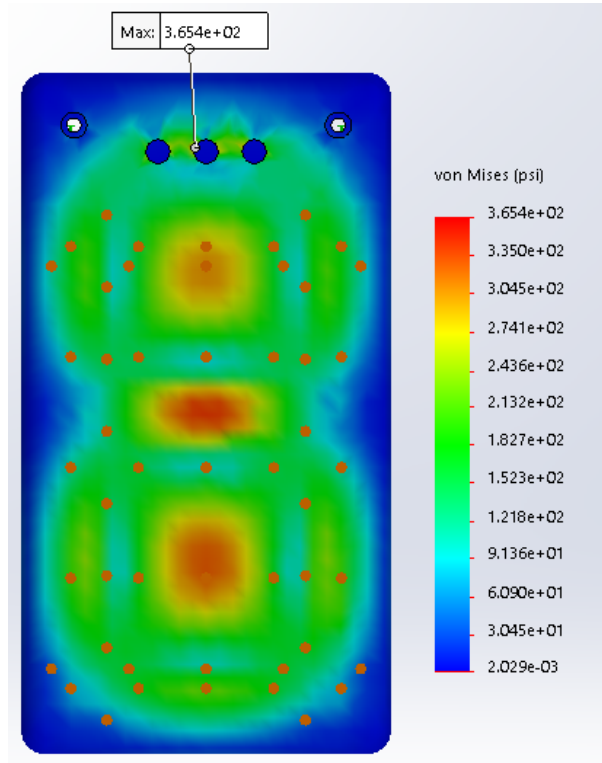


Figure 58. Stress test on design 5b for average male palm size

### Electronic Component Placement Test Plate

Now that we have a determined structure for load support for the case, we can focus on electronic component placement. The first step of which, is making sure that each component properly fits in its location. Since one print of the device takes upwards of 11 hours, our team has designed a test plate as seen in Figure 59. The purpose of this test plate is to place each individual electronic component into its fit and then determine if the fit is good or it needs minor adjustments. In making this plate we can adjust every 2 hours rather than one adjustment a day. Also, on this test plate we brought in a new smaller button and in order to secure the two pieces together, one side of the case will have a hexagonal counterbore to fit an M3 nut. This will allow us to easily secure and take apart our case during the testing phase of the device. Once we were confident that each electric component fits in its spot. We placed the pins and holes in their proper locations on the device to create design 5.

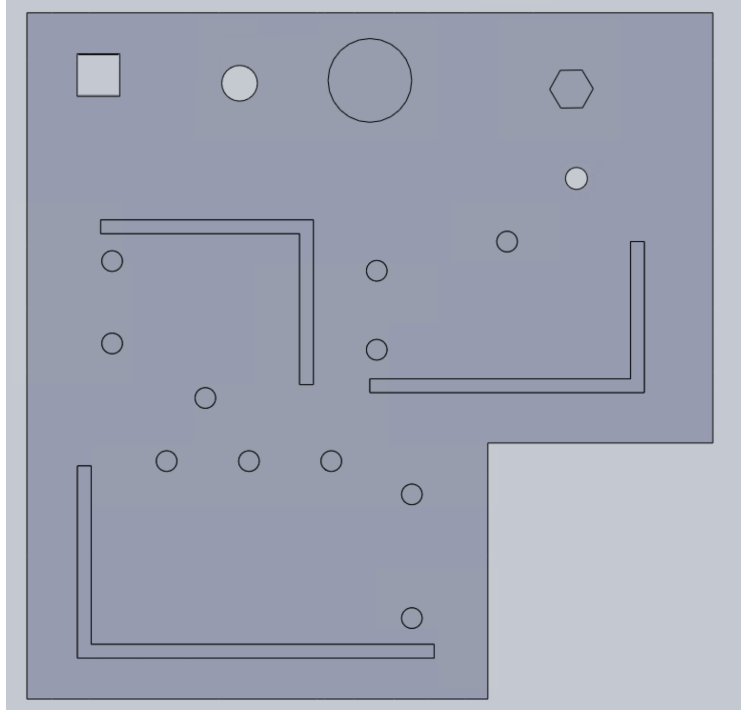


Figure 59. Electronic Component Test Plate

#### 4.3.6. Design 6

The 6th design is a combination of design 5's structure with the component placement from the test plate. In Figure 60 you can see the placement of the accelerometer, microprocessor, and 9V battery. In Figure 60 you can see the placement of the speaker, button, and LEDs. We have also cut a hole in the side of the device by the microprocessor so the device can be plugged into a computer during the testing phase as seen in Figure 61. In order to properly close the device, we decided to put in hexagon counterbores on the bottom of the device where we will press fit M3 nuts. This will add a threading bottom of the device and allow us to take apart the device during testing by simply removing the screws instead of needing a second tool to hold the nut while removing the screw. In order to hide the wires that must go from component to component, we designed 5 upside-down "T" s into the middle rib and they are used to conceal the wires as seen in Figure 62. Since there have been large changes in the structure, we need to run another set of stress test to prove that the device is still within our safety factor. Throughout our testing, the variance between male and female hands is minimal. Therefore, for this stress test will just be an average between the male and female hands, which is 10.32cm. We determined that  $\delta = 16679.3/495.4 = 33.67$ . Since 33.67 is a greater value than 3, from a structural standpoint this device is acceptable. Since the device is structurally sound, and the electronic components fit properly. It is now time for the testing of the device.

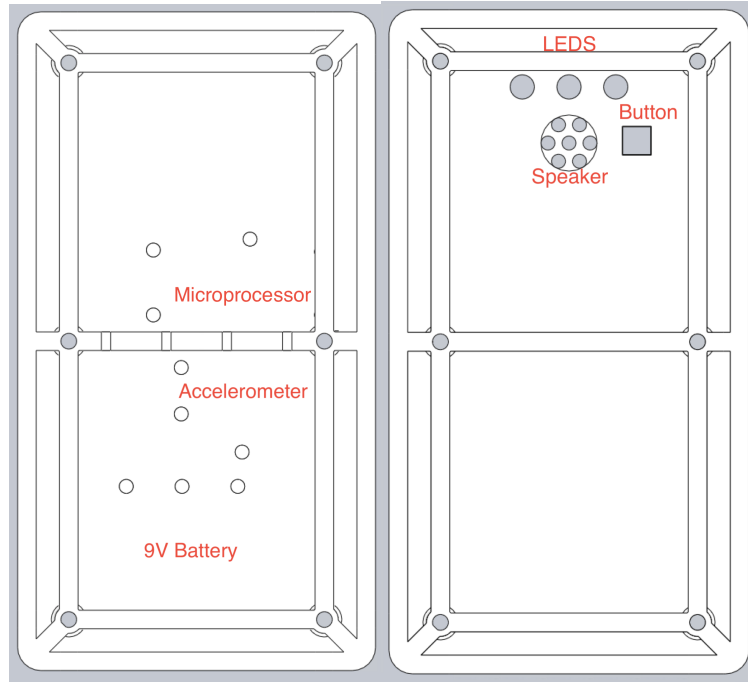


Figure 60. Bottom and top half of design 5 showing electronic component placement

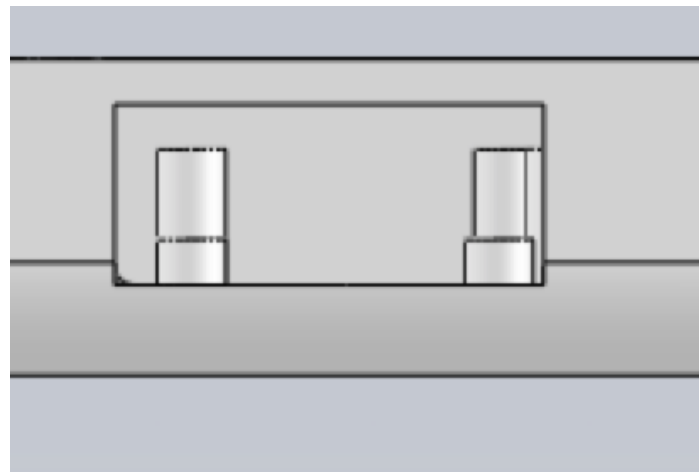


Figure 61. Port cut into side of device for testing

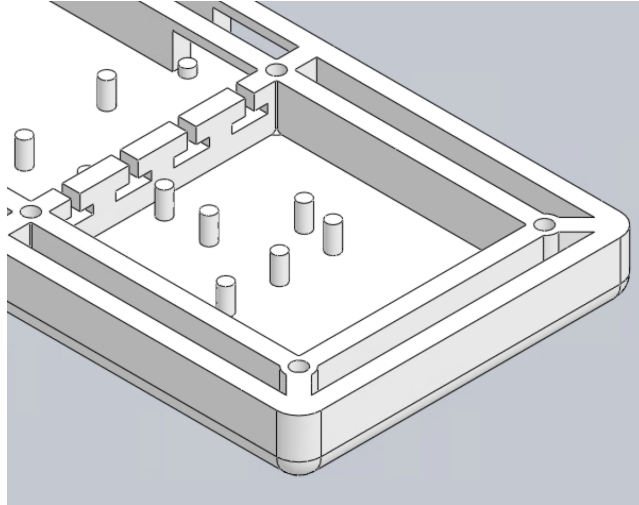


Figure 62. Image of upside down “T” s in middle ribbing of design 6

#### 4.3.7. Design 7

The 7th design is a fine-tuning of the previous design. The “T” wire holding slots were made thinner in Figure 63, the middle ribbing was removed in Figures 64 and 65, and a battery slot was added in Figures 66 and 67.

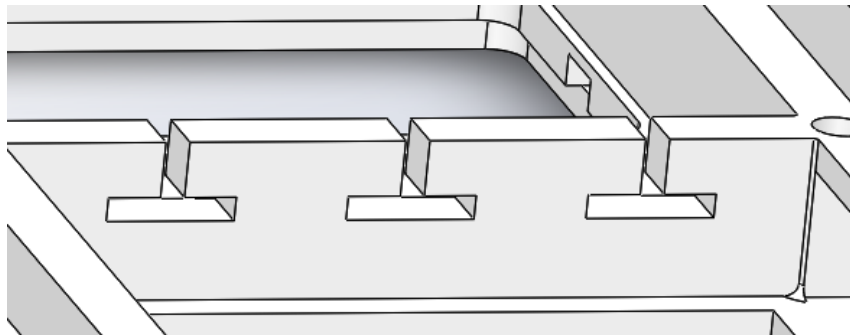


Figure 63. Redesigned "T" for holding wires

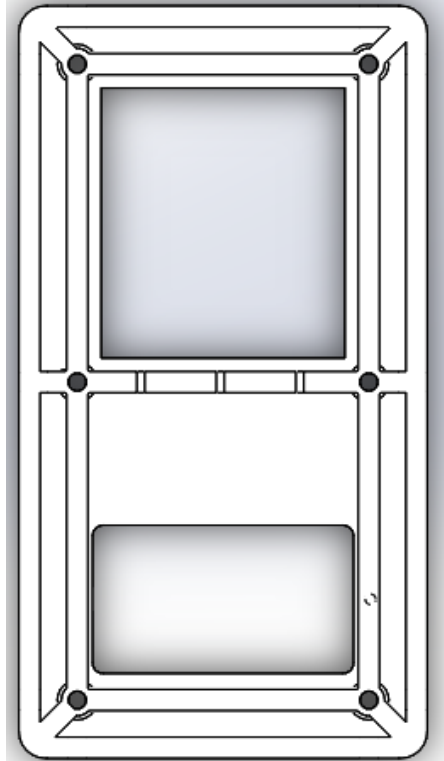


Figure 64. Image of Device Bottom without pins and a porthole for easier wiring

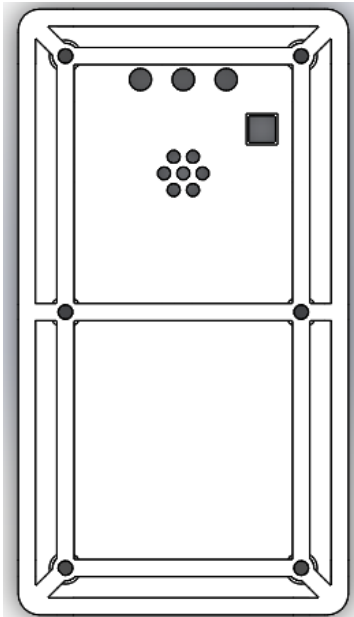


Figure 65. Top half of device with moved speaker and button location to make room for new components

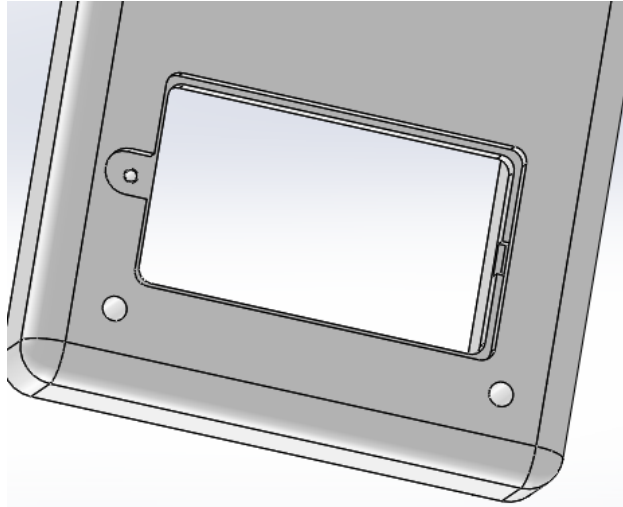


Figure 66. The battery slot on the back of the device

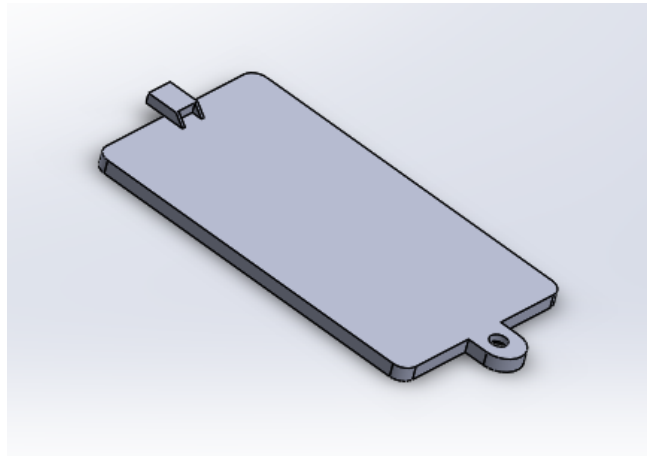


Figure 67. Battery Cover

#### 4.3.8. Final Design

The final case design can be seen in Figure 68. The size of the case is still within our size constraints listed earlier in this section such as anthropometric data of palm width, palm length, and sternum width along with the size of a general first aid kit. As seen in Figure 69 the middle support bar has been removed along with the upside down “T”s in order to make room for the PCB. A new support bar has been placed near the battery port as seen in figure 70 to add support along with preventing the battery from moving horizontally within the device. In order to prevent the battery from moving vertically while in use we placed foam on both sides used to hold it in place. In Figures 69 and 70 you can also see the final electronic component placement inside the case. Also in this final design, we have removed the testing and wiring portholes and close one



side of the screw holes on the bottom of the device. This was done because in our final design, we do not want users to have access to any electronic components besides changing the battery.

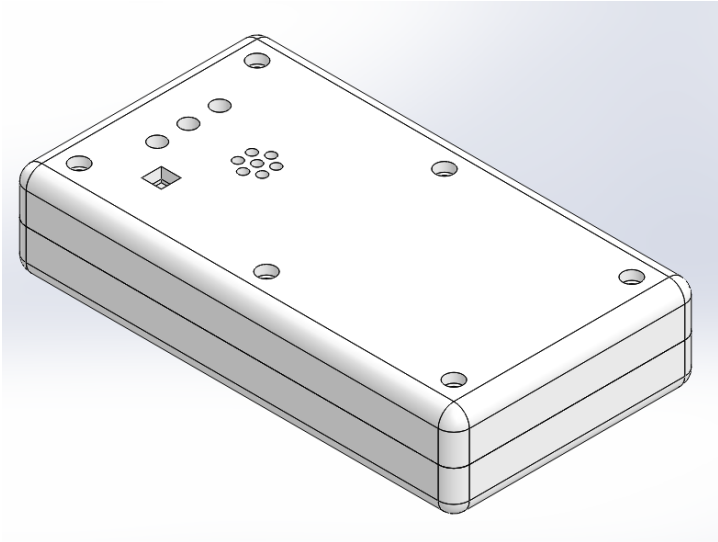


Figure 68. Isometric view of the device

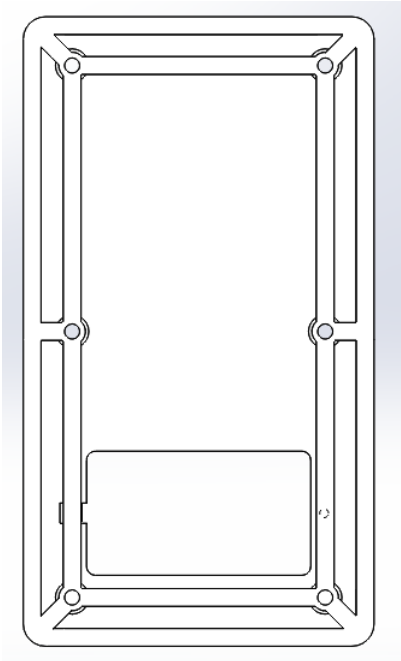


Figure 69. Inside bottom of device where middle bar has been removed and PCB is located

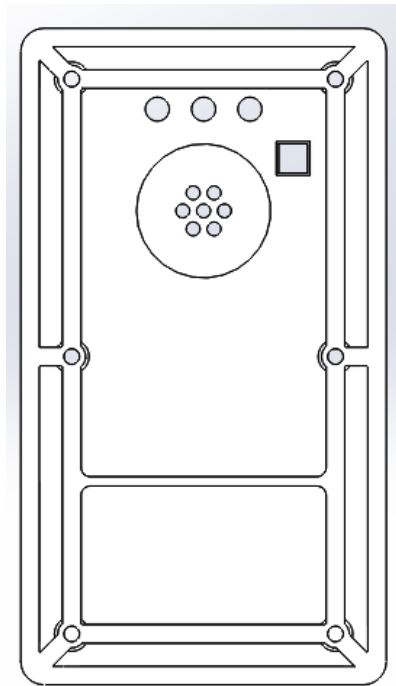


Figure 70. Inside top of device where new bar separating battery from rest of components

Since the middle bar was thinned and moved, more simulated stress tests had to occur in order to make sure that our device was still within our earlier designated safety factor of 3. When testing for the average female palm as seen in Figure 71,  $\delta = 16679.3/1752 = 9.52$ . This value is above the minimum safety factor of 3, therefore it passes our stress test. For the average male palm size,  $\delta = 16679.3/1632 = 10.22$  as seen in Figure 72 This value is greater than the minimum safety factor of 3, therefore it passes our stress test. Since testing for both male and female palm size has concluded that both values are above a safety factor of 3, we can conclude that the case will be able to safely withstand the forces during normal CPR compressions.

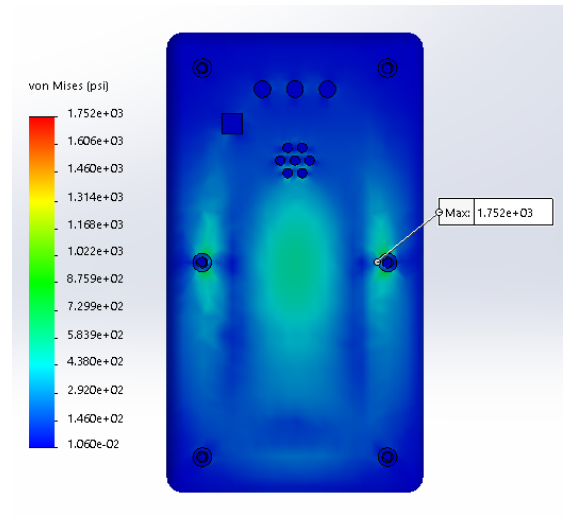


Figure 71. Stress test on final design for average female palm size

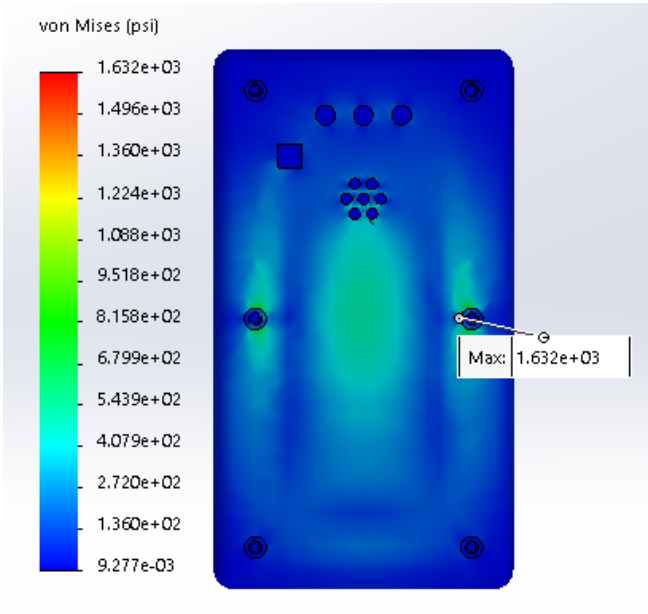


Figure 72. Stress test on final design for average male palm size

## 4.4. Component Modification 1

Through the iterative design of the project, multiple components were replaced to better accommodate the design requirements. This section details the changes, the necessity for the changes, justification for the new selection, and the overall effect on the design of each change.

### 4.4.1. Switch

Originally, it had been decided that a red 22mm pushbutton actuator would be used to turn the device on and off, the Eaton M22-D-R. The pushbutton could be programmed to turn on the device with a single push, but would only turn off when the button was held. Unlike toggle switches, this would prevent the accidental shutdown of the device during operation. The appeal of the red 22mm button was its large size and visibility, making it impossible to overlook for first time users. However, when the first prototype was built, it was clear that the button was too large, relative to the size of the device itself. The user's hand would often be directly over the button, causing discomfort. A smaller pushbutton, the Ulinco U16A1, see Figure 73, was then selected for further prototyping [44]. Although the area of the button was significantly smaller, its height became a limiting factor. After reevaluating, it was determined that size was a primary concern and as such, it was decided that rather than a pushbutton, a tactile switch, the CO-RODE, see Figure 74 [10]. would be used. The usage of the CO-RODE tactile switch greatly reduces the surface area of the I/O components of the device, allowing better positioning of the user's hands on the device.



*Figure 73. Ulinco U16A Pushbutton*



Figure 74. CO-RODE Tactile Switch

#### **4.4.2. Microcontroller**

The microcontroller that had been initially selected for usage in the device was the Arduino Nano. With essentially identical features to the Sparkfun Pro Micro, the deciding factor was the price, with the cost of an Arduino Nano clone at \$6.99 compared to the Sparkfun Pro Micro at \$20.95. When attempting to set up the board, a bug in the board's hardware was discovered which required the bridging of two pins on the processor. The bridge was attempted but failed and it was decided that if a new Arduino Nano were to be purchased, it would not be a clone. However, the price for the Arduino Nano from Arduino themselves was \$22.00, higher than that of the Sparkfun Pro Micro. At this point in the project, nothing was dependent on the specific usage of the Arduino Nano and therefore, the Sparkfun Pro Micro was selected for future usage in the project.

#### **4.4.3. Sensor**

The MPU6050 was first used in the project with the intention of double integrating the discrete acceleration data provided by the accelerometer to determine its relative position since calibration. Once the microcontroller was programmed to perform this integration, it was discovered that the effects of gravity caused significant and unavoidable drift in the relative position over the course of mere seconds of operation. Measures to counter this drift were taken, using calibration to remove the offset created by gravity, acceleration thresholds to decrease noise, and the calculation of relative orientation to determine the contribution of gravity in each axis of acceleration. However, it was ultimately determined that in order to properly adjust for the acceleration of gravity, far greater accuracy of the device's orientation would be necessary than could be provided by the accelerometer. This was caused by the need of highly accurate rotational acceleration readings, highly accurate linear acceleration readings, and constant calibration

conditions, none of which are possible using a 6 degree of freedom (DOF) accelerometer. As such, a 9DOF accelerometer, the BNO055, seen in Figure 75, was used instead [6]. Its fusion mode, which uses an accelerometer, gyroscope, and magnetometer to determine its absolute orientation, can provide linear acceleration vectors which do not include the acceleration of gravity, eliminating the difficulties found in the usage of the MPU6050.



Figure 75. The BNO055 9DOF Accelerometer

#### 4.4.4. Circuitry

In order to use the various components for their intended purpose, they must be connected appropriately to the microcontroller. The pin layout of the Sparkfun Pro Micro, seen in Figure 76, details the functions of each pin on the microcontroller and is used as a guide to determine where each of the components is to be connected. The battery must be connected to the RAW pin which connects to the internal voltage regulator, providing the power in the system. The accelerometer communicates data to the microprocessor using the I2C serial protocol. This component must be supplied a source voltage (VCC,) a ground reference (GND,) the I2C data line (SDA,) and the I2C clock line (SCL.) Other simple components such as the buzzer and LEDs require a ground reference and a GPIO pin to control its operation. This circuit design can be seen in Figure 77.

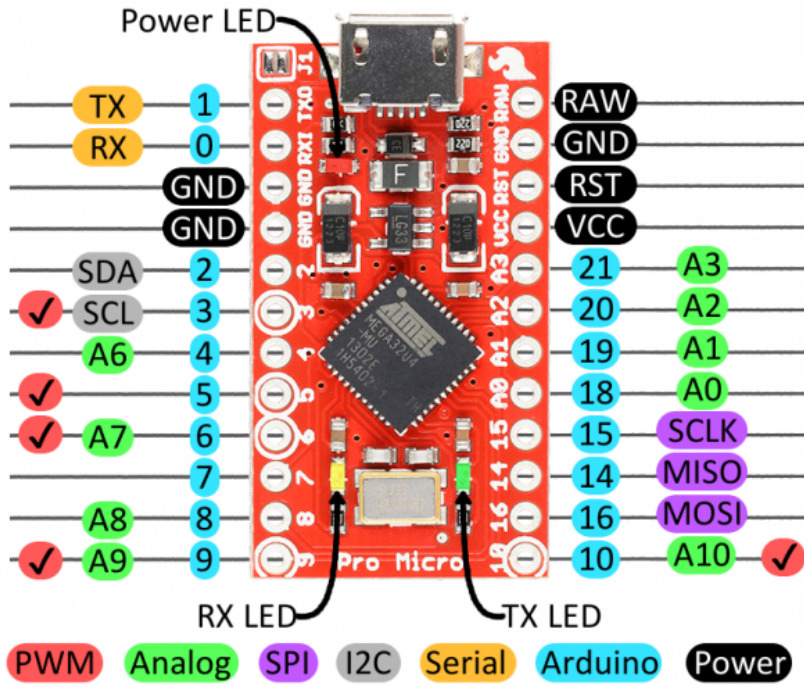
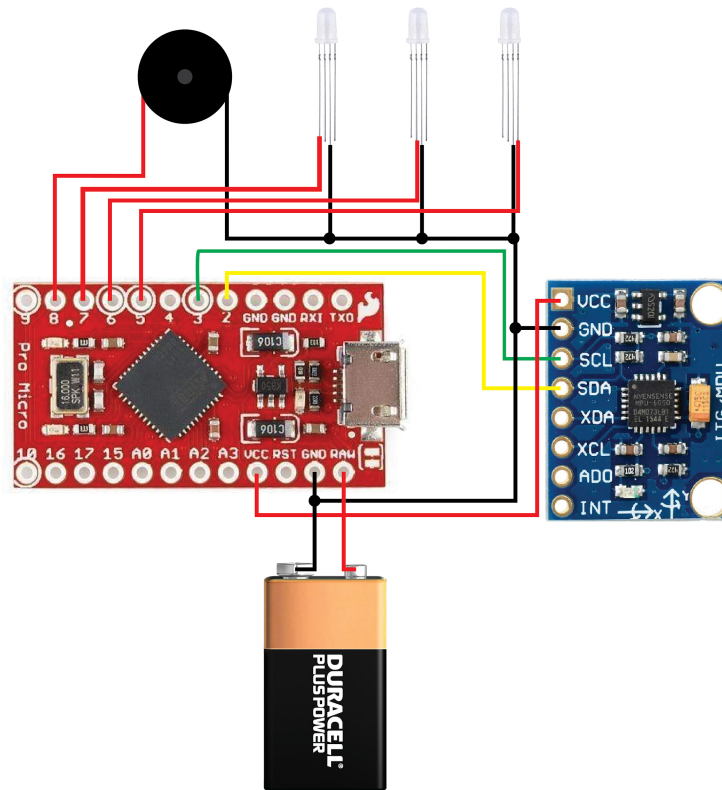


Figure 76. Pin Layout of the Sparkfun Pro Micro



#### **4.4.5. Algorithms**

##### MPU6050

In order to provide depth feedback to the user, acceleration data was read from the MPU6050 in the Z direction, normal to the chest of the person receiving CPR, and integrated twice to calculate the relative displacement since the configuration of the device. In the implementation, all 6 accelerations, X, Y, and Z, both linear and angular, are integrated in expectation that the data may prove to be useful.

When the device is turned on, the microcontroller wakes up the MPU and uses the first 1000 acceleration samples to calculate the offset, which is assumed to be constant, when the device is at rest. The microcontroller then enters the main loop and proceeds to perform the integration. The raw acceleration data is modified and filtered to produce a more accurate final output. The previously calculated offset is first subtracted from the raw acceleration. If this result is lower than a certain threshold, set at 50 for testing purposes, it is considered to be the result of noise and the acceleration is set to zero. The five most recent acceleration readings are stored in a buffer and if they are all equal to zero, the device is determined to be at rest, resetting the velocity to zero as well. Using the filtered acceleration and the difference in time between measurements, the velocity and displacement are calculated. These values are simply printed to the Arduino serial monitor for testing purposes.

In testing the device, significant and consistent drift in the positive Z direction was observed. When the device is turned, the acceleration offset caused by gravity changes. When integrated, even the small amounts of shaking cause large amounts of drift. The exact offset of gravity could be calculated if the orientation of the device was available. However, the MPU6050 does not provide accurate enough readings to derive this information without making a large amount of assumptions regarding its orientation when it is first turned on. As such, further development with this sensor was concluded.

##### BNO055

The code originally written for the MPU6050 was modified to use the acceleration data read from the BNO055. Using simple vector operations, the acceleration in the direction of gravity can be derived. This was then used in the same manner previously described as the acceleration data from the MPU6050. Although the usage of the BNO055 succeeded in reducing the drift caused by gravity, the integration of other noise in the system resulted in drifting displacements



on the scale of a couple centimeters [6]. Without control over the accuracy of the sensor, it was determined that the depth feedback would no longer be determined using the relative displacement, but rather the acceleration itself. The 256 most recent samples acceleration samples are stored within a buffer. This buffer is then searched for the peaks and valleys of the waveform, which are then stored into buffers of their own. For testing purposes, the acceleration samples are printed to the Arduino serial plotter to view both qualitative and quantitative values of the resulting waveform.

## **4.5. Component Modification 2**

After initial testing with the device, a second iteration of component modifications were suggested. Due to time constraints, these modifications were not implemented, but as with the previous section, this one details the changes, the necessity for the changes, justification for the new selection, and the overall effect on the design of each change.

### **4.5.1. Power Regulation**

The device was originally programmed to go into lower power mode if the power button was held for one second. With a 5.0V Sparkfun Pro Micro, the low power mode should consume only 6.2uA. The standard 9.0V battery has a capacity of 500mAh. Therefore, it is expected that in low power mode, the device should last 9.2 years. However, it was found that this is not true in practice. The battery of the device, even in low power mode, lasted only a couple days. In order to reduce the consumption of power over time, a small circuit was designed regulate power through a P-Channel MOSFET. The gate is controlled by a LM555 timer operating in bistable mode as a flip flop. This flip flop is set using a monostable circuit when the power button is pressed. The flip flop is reset after the power button is held for one second by the microcontroller. This circuitry can be seen in Figure 77.

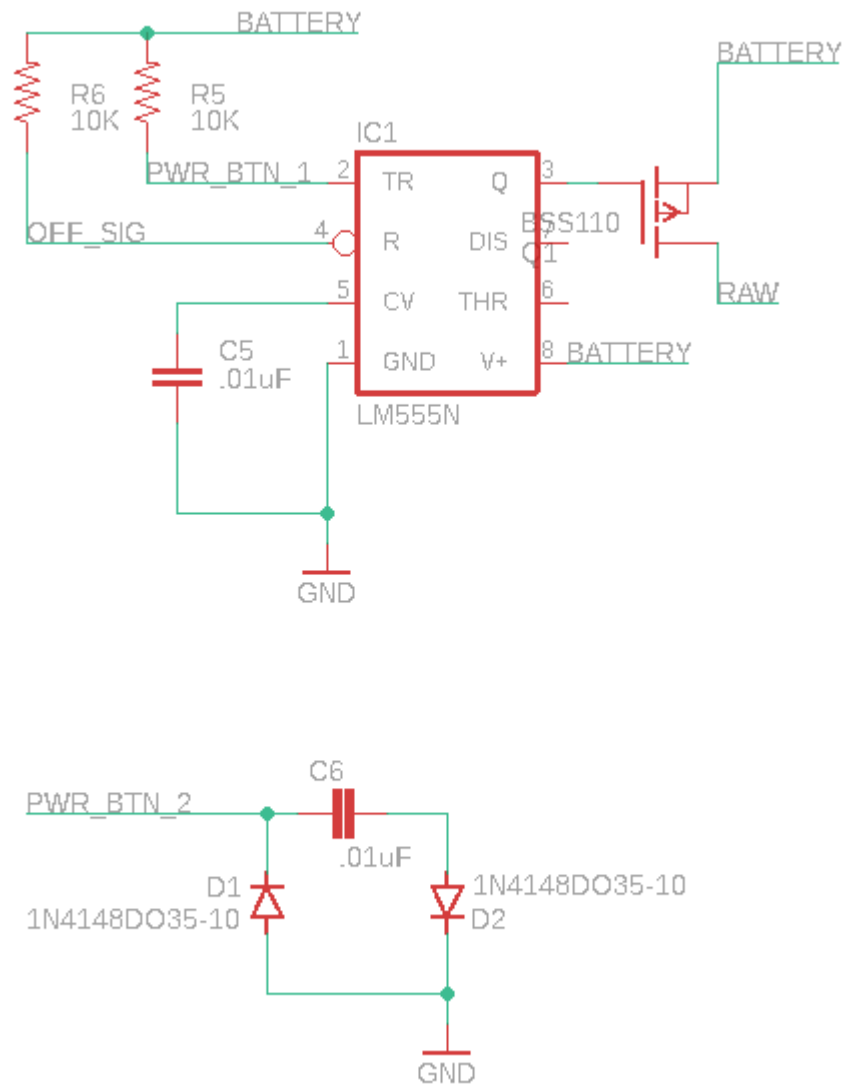


Figure 77. Power Regulation Circuitry

#### 4.4.2. Audio Feedback

Initial testing of the device showed the piezoelectric buzzer to be audible in a controlled environment. However, in practical usage, it is likely that the device could be drowned out by the sounds of crowds, passing cars, and other loud sounds. As such, it was determined that the piezoelectric buzzer should be exchanged for a louder feedback system. An 8-ohm speaker and audio amplifier system was selected as a replacement. Basic tests were performed which showed that the piezoelectric buzzer had an output of 60dB whereas the speaker and audio amplifier had an output of 80dB. The speaker and audio amplifier can be seen in Figure 78 and Figure 79.



Figure 78. 8 Ohm Speaker

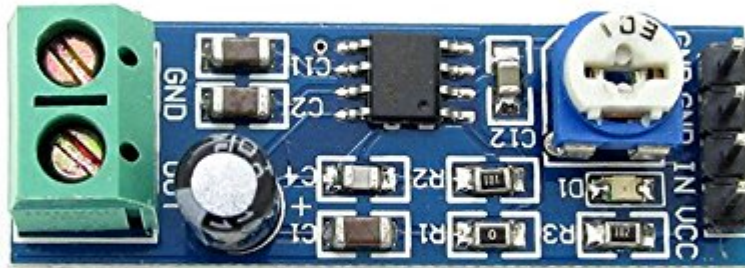


Figure 79. Audio Amplifier

#### 4.4.3. Printed Circuit Board

Having tested and finalized the changes to the circuitry, a printed circuit board (PCB) was designed to further reduce the cost of the device. The PCB combined the microcontroller, accelerometer, power regulator, and audio amplifier into one board. The combined cost of these major components is \$64.29. By switching to a PCB, the price of a single PCB can be reduced to \$34.24, with an expected bulk manufacturing price of \$16.08. The schematic can be seen in Appendix 4. Two iterations of the PCB were created, the second of which improved on the first

by further decreasing the size of the board from a square area of 2.47 square inches to 1.9 square inches. The first PCB can be seen in Figure 80, the second PCB can be seen in Figure 81.

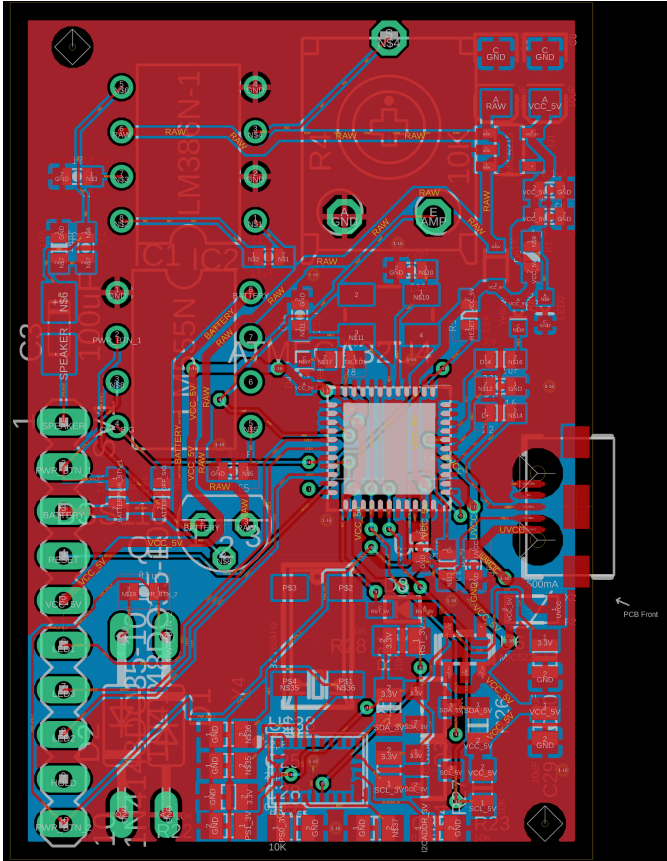


Figure 80. First Design of the Printed Circuit Board

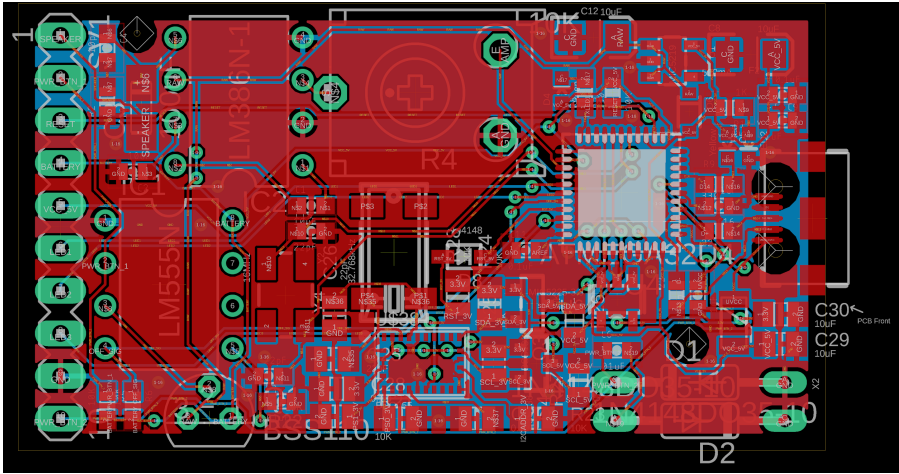


Figure 81. Second Design of the Printed Circuit Board

## **Chapter 5. Testing Apparatus Design**

Performing significant tests that prove a CPR device's effectiveness is very difficult because one cannot test the device on a living person. This is due to the damaging side effects CPR can have on a person such as the possibility of breaking rib bones during chest compressions. For most of the existing devices, the only relevant studies that can be performed are simulation studies. In order to perform the necessary tests that will prove our device's effectiveness, our own test apparatus had to be created. The first step to creating this device was to list the tests we were going to perform. All testing procedures can be found in Appendix 1. Once the tests were defined, criteria needed to be created that most accurately related compressions on the test apparatus to proper CPR performed on a human. This criterion included:

- Applying a force of approximately between 60-120 pounds
- Compressing the device within 2 and 2.4 inches
- Compressing the device under the range of 2-2.4 inches
- Compressing the device over the range of 2-2.4 inches

The combination of a range of depths and approximate force relative to the force needed to compress the average individual's chest will allow the device to be properly tested under conditions that most accurately represent actual CPR performed on a human. Once the necessary criteria for the test apparatus were defined, we had to determine how we were going to design the apparatus. In order for the device to be compressible, springs were determined to be used. In order to replicate the approximate force to compress a chest, 4 springs requiring 120 pounds of force to compress were to be compressed in unison. These springs also had to allow for a range of compression between 0 in and over 2.4 inches in order to simulate the compression of a chest properly. Once the springs were chosen, the first design was created through a CAD software.

### **5.1. First Design Iteration**

A preliminary testing apparatus was designed with the primary focus of reducing the degrees of freedom of the testing surface, the surface on which the device is to be placed, to a single axis. This was done by drilling holes on the four corners of the testing surface, through which dowels are extended. These dowels serve two purposes: to secure the movement of the testing surface, but also hold the springs in place such that the force can be applied to the testing

surface without deviation. Because the testing structure is simply guided by the dowels and not fixed to them, a chassis to house the dowels and testing surface was designed. This chassis was designed to allow the tester to apply a downward force to the testing surface, contain the dowels such that they do not protrude, and fix all dowels in a vertical position even during operation. This design, for ease of prototyping, uses only 3/4" MDF and 1/2" wooden dowels. The Solidworks model can be seen in Figure 82 and prototype in Figure 83.

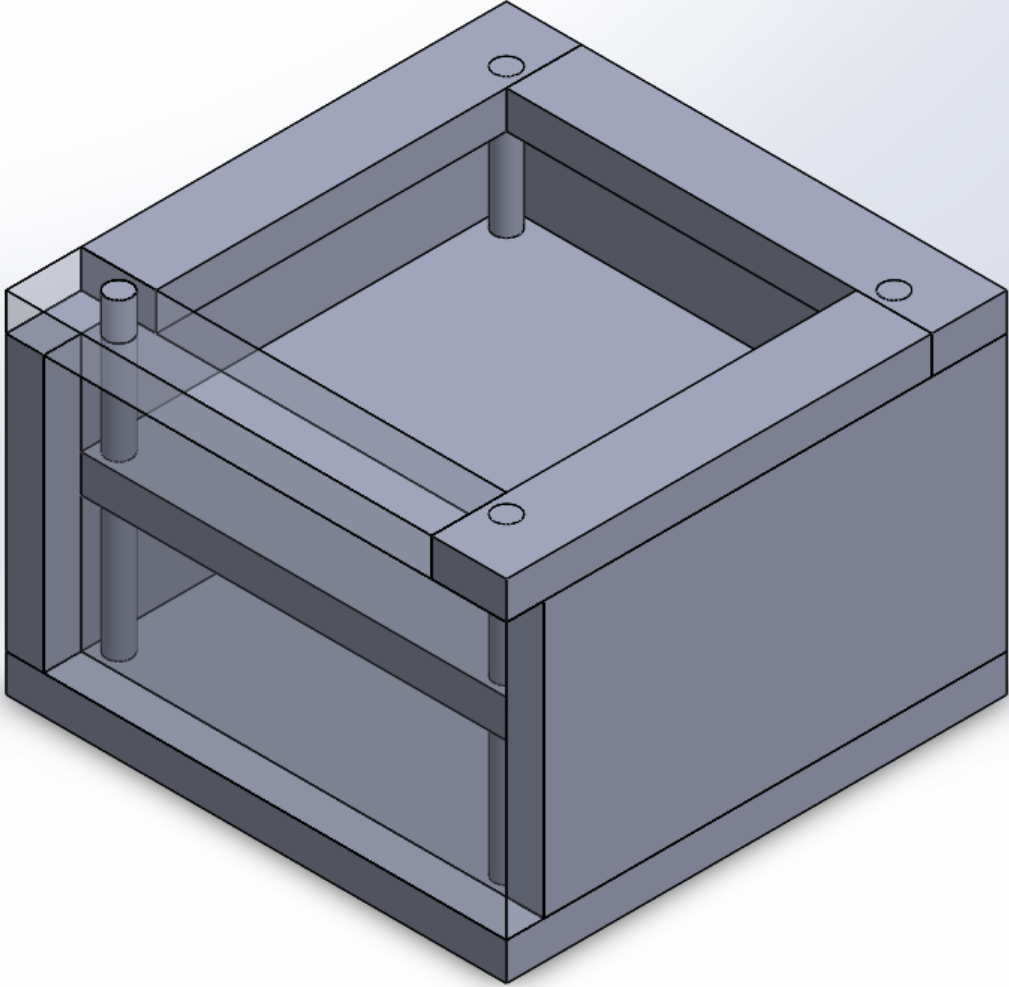


Figure 82. Solidworks Model of the Preliminary Testing Apparatus

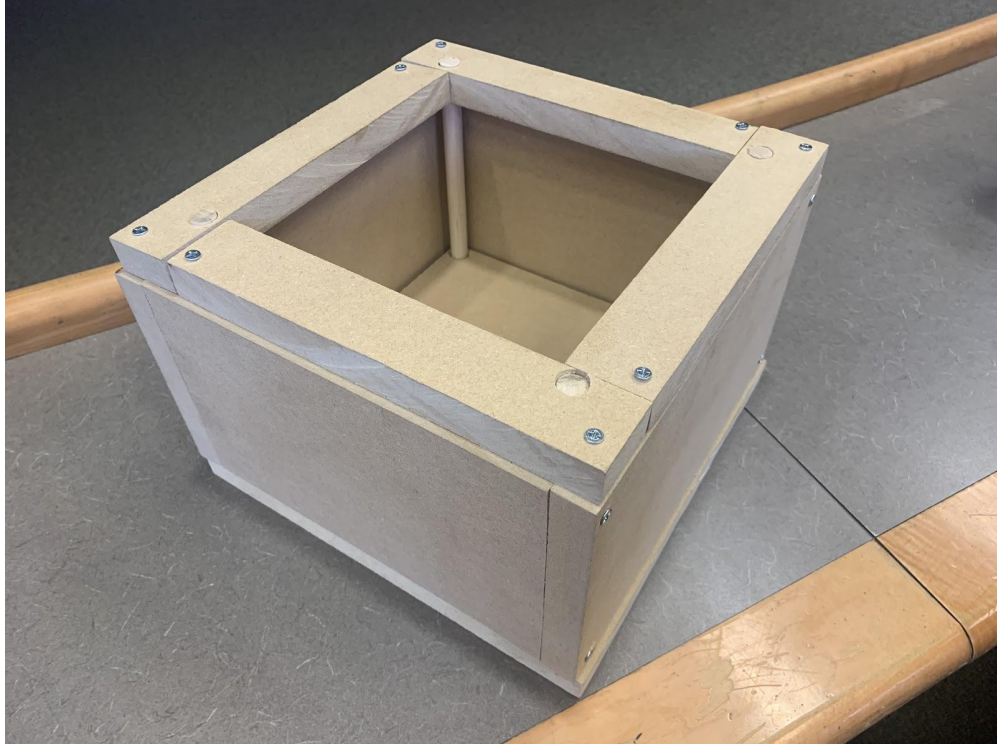


Figure 83. Prototype of First Design

## 5.2. Second Design Iteration

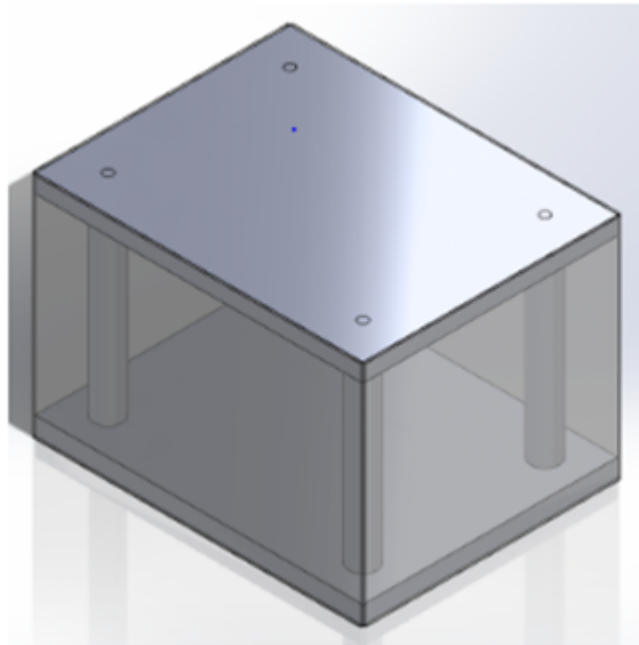


Figure 84. Second Design of Test Apparatus

This design was to be made out of:

- 2 plywood bases sized at 9 inches long, 7 inches wide, and .25 inches thick
- 4 acrylic panels that are all .25 inches thick
  - 2 of which are 9 inches long and 7 inches tall
  - 2 of which are 7 inches long and 7 inches tall
- the 4 springs (designated in the picture by the cylinders) which are 5.5 inches tall and .75 inches in diameter
- 4 metal rods that are 6.5 inches tall and .25 inches in diameter.

The plywood bases would be on the bottom and top of the device and are designated by the solid grey rectangles in the picture. The bottom and top bases would have 4 holes in them that would be used to secure the 4 metal rods in place. The metal rods would press fit into the holes in the bases and sit directly in the middle of the springs, which would hold the springs in place. The 4 acrylic panels would then be secured on each of the 4 sides of the apparatus in order to contain it but to also allow vision of the springs during compression and relaxation. The actual device would sit on top of the top plywood base and the user would push down on the device in order to simulate compression of a chest.

Although this device would have worked for testing, the group found that there were some major theoretical problems with it. As the springs compress and the top base moves down, the metal rods holding the springs in place would stick out of the device and possibly interfere with the compressions. Not only would the metal rods interfere, but based on the size of the test apparatus, the acrylic panels would have also interfered with the user and compressions. In order to address these problems, a second iteration of the test apparatus was designed using the same CAD software.



### 5.3. Third Design Iteration

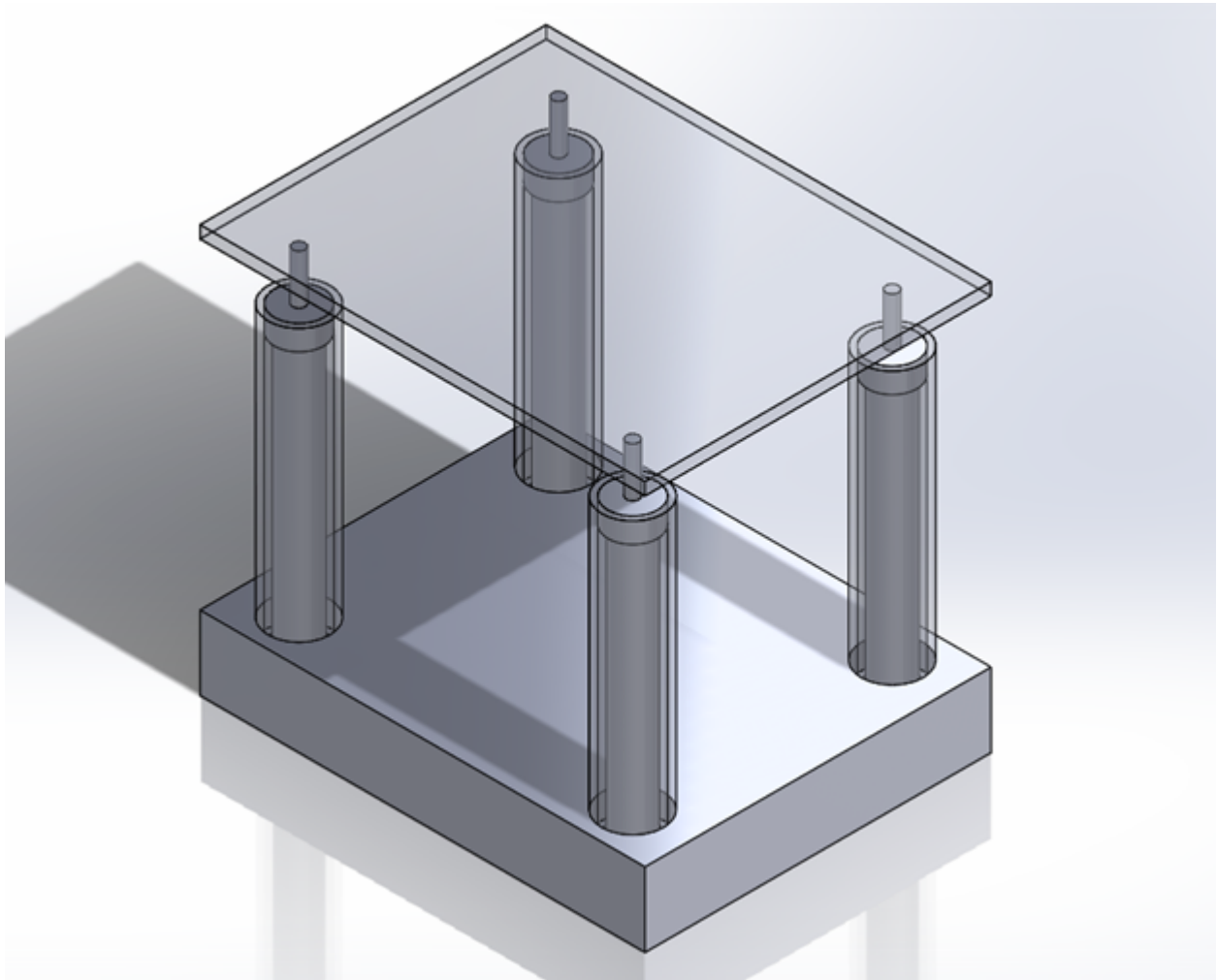


Figure 85. Third Design of Testing Apparatus

In order to address the issues of interference during compressions, this device was designed. This design was to be created using:

- A medium-density fiberboard (mdf) base with a thickness of .75 inches, length of 9 inches, and width of 7 inches
- 4 PVC pipes with an inner diameter of 1 inch, an outer diameter of 1.25 inches, and length of 6 inches
- The 4 springs that were used in the first design iteration
- 4 mdf buffer discs of thickness .5 inches and diameter of 1 inch
- 4 metal rods of diameter .25 inches and length of 3.25 inches
- 1 acrylic top .5 inches thick and the same length and width as the mdf base

The mdf base was to have 4 circular cuts made halfway down into it in order to support the PVC pipes. The PVC pipes would contain the springs. The mdf buffer discs would sit freely on top of the springs, but would also have a single hole in the center of it that would contain the metal rods. The metal rods would then also sit halfway into the acrylic top in order to secure it. The actual device would then be placed on top of the acrylic top and the user would compress the device to simulate chest compressions.

This device would also work for testing the device, however again the group found a theoretical problem with this design. Because the buffer discs are not anchored down and sit freely on top of the springs, the rebound from the compression could possibly cause the acrylic top, metal rod, and buffer disc sub-assembly to fall off the test apparatus. In order to address this problem, a third design was created, found sufficient for testing, and built.

#### **5.4. Fourth Design Iteration**

The fourth iteration of the test apparatus was made of:

- An mdf base with a thickness of .75 inches, length of 9 inches, and width of 7 inches
  - 2 holes were cut .25 inches deep into the base
- 4 PVC pipes with an inner diameter of 1 inch, an outer diameter of 1.25 inches, and length of 6 inches
- 4 threaded PVC fittings with an inner diameter of .75 inches.
- The 4 springs that were used in the previous design iterations
- 4 acrylic buffer discs of thickness .5 inches, diameter of .75 inches, and a hole in the center with diameter and depth of .25 inches
- 4 metal rods of diameter .25 inches and length of 3.25 inches
- 2 metal rods of diameter .25 inches and length of 10.25 inches
- An mdf support backboard of thickness .75 inches, width of 9 inches and height of 10.25 inches.
- 2 mdf constraints

- 1 acrylic top .5 inches thick and the same length and width as the mdf base with 6 holes in it
  - 2 of the holes are straight through the acrylic and .25 inches in diameter
  - 4 of the holes are only .25 inches in depth and .25 inches in diameter

The construction of this device began by first anchoring the 4 threaded PVC fittings to the mdf base using screws and epoxy. Once the fittings were in place, the springs were placed within the 4 PVC fittings in order to secure them, and the 4 PVC pipes were placed over both the springs and fittings thereby covering them. The 4 metal rods of 3.25 inches in length were then secured to both the acrylic buffer discs and the acrylic top using epoxy. The 2 metal rods of 10.25 inches in length were secured to the mdf base using epoxy, but are left to freely move through the acrylic top. The acrylic top, metal rod, and buffer disc sub-assembly was then placed on top of the springs. The mdf backboard was secured to the mdf base using screws and epoxy, and the mdf constraints on top were secured to the backboard using screws. The device is then intended to be placed on top of the acrylic top and the user is to push down on the device in order to simulate chest compressions. This version of the test apparatus properly simulates chest compressions on an average person during CPR, will not interfere with the simulated compressions, and does not allow the acrylic top to fall off the apparatus during the rebound from a compression.

## 5.5 Final Design

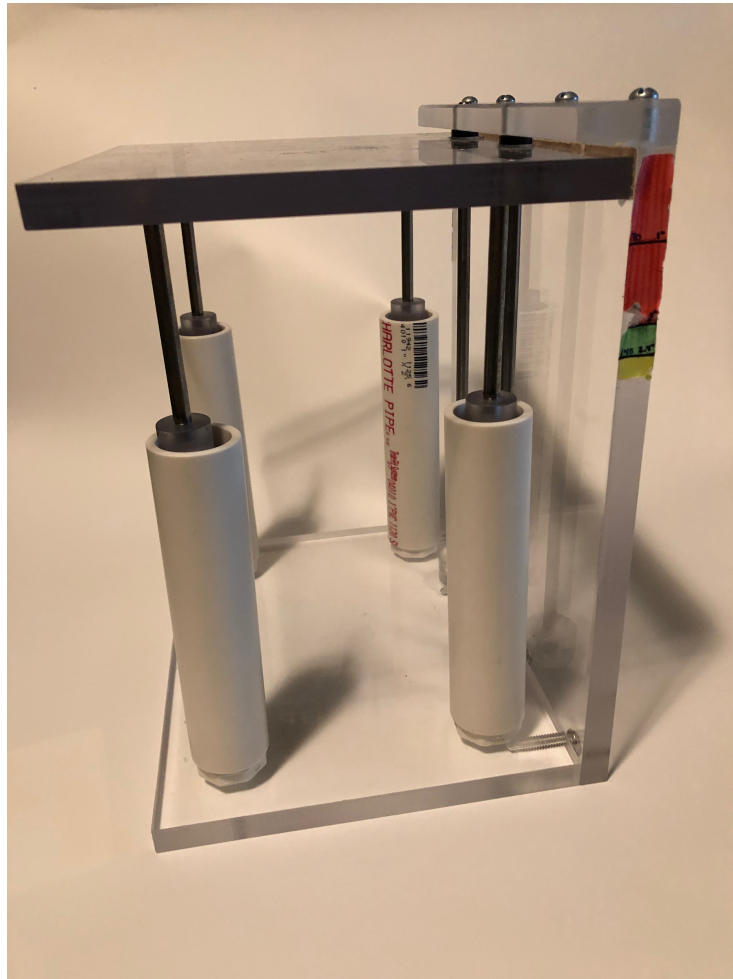


Figure 86. Final Design

The final design of the testing apparatus was created to address the following criteria. The springs that were previously used provided a resistance of 120 pounds of force to compress the device 2 inches. It was found however, that the range of force required to compress a person's chest 2-2.4 inches is between 60-120 pounds of force, varying per person. For this reason, the 4 springs were changed to a set that provided a resistance of 60 pounds of force to compress the device 2 inches. This design iteration also had to adjust for the horizontal shear force being created during compressions due to the metal guide rods. In order to fix this issue, 2 roller bearings were inserted into the compression plate and fitted over the metal guide rods. The following materials were used to create the final design:

- An acrylic base with a thickness of .75 inches, length of 9 inches, and width of 7 inches

- 4 PVC pipes with an inner diameter of 1 inch, an outer diameter of 1.25 inches, and length of 5.75 inches
- 4 threaded PVC fittings with an inner diameter of .75 inches
- 4 springs 5.5 inches tall and .75 inches in diameter that provide a combined force of 60 pounds to compress 2 inches
- 4 acrylic buffer discs of thickness .5 inches, diameter of .75 inches, and a hole in the center with diameter and depth of .25 inches
- 4 metal rods of diameter .25 inches and length of 3.25 inches
- 2 metal rods of diameter .25 inches and length of 10.25 inches
- 2 roller bearings
- An acrylic support backboard of thickness .75 inches, width of 9 inches and height of 10.25 inches
- An acrylic top support strip 9 inches in length
- 1 acrylic top .5 inches thick and the same length and width as the mdf base with 6 holes in it
- Screws
- Epoxy to hold the springs in place

The actual CPR device is then intended to be placed on top of the acrylic top and the user is to push down on the device in order to simulate chest compressions. This final iteration of the test apparatus properly simulates chest compressions on an average person during CPR, will not interfere with the simulated compressions, will not allow for horizontal shear force due to the roller bearings, and will take a force of 60 pounds to compress 2 inches rather than 120 pounds from previous iterations.

## **Chapter 6. Results**

### **6.1. Calibration Results**

After the testing apparatus was constructed, initial tests were performed to determine the accuracy of the device. For the device to be considered accurate, it had to provide the appropriate LED for the appropriate depth compressed, where the red LED appears for shallow compressions or less than 2 inches, the green LED appears for adequate compression depth or between 2 and

2.4 inches, and the blue LED appears for deep compressions or further than 2.4 inches. 100 two-minute samples of compressions simulating bystander CPR were taken. For each sample, the user attempted to perform proper CPR for two minutes on the testing apparatus and each sample was video recorded. The different success and failure rates of the device were then determined by analyzing each sample video and recording whether the appropriate LED appeared for the appropriate depth for every compression that was made.

After analyzing all 100 samples, it was determined that the device operated as expected 73% of the time, displaying the proper LED for the proper compression depth. 8.9% of the time, the device provided false positives where the depth of the compression was either too shallow or too deep but the green LED indicating adequate depth was displayed. 18.1% of the time, the device provided false negatives where the depth of the compression was of adequate depth but either the red or blue LED indicating too shallow or too deep of a compression respectively was displayed. After determining the device's accuracy, subject testing occurred.

## **6.2. Subject Results**

IRB approval, seen in Appendix 4, was obtained to test with subjects on a quantitative and qualitative level to view effectiveness change with the use of the CPR Device. As for subjects, 27 college-aged subjects and 7 CPR-certified subjects were tested, and all First, each subject took a pre-testing quiz in order to get information on their baseline knowledge of CPR and confidence in performing CPR. Only 16 subjects knew that proper CPR depth is between 2-2.4 inches. When asked if the subject felt "not confident", "somewhat confident", or "very confident" in performing CPR, about half of the subjects, regardless of CPR certification, stated that they only feel "somewhat confident" in their CPR skills, 25.9% stated they feel "very confident" and 25.9% stated they feel "not confident".

After the pre-testing quiz was completed, each subject was asked to compress on the testing apparatus at the normal hands-only CPR compression rate and depth for 1 minute on their own. This data was recorded and analyzed and 48.1% of subjects completed compressions at the right rate, but did not have consistent compressions (constantly varied depths between too shallow and good depth), 33.3% of subjects did not have correct depth or rate, with a 130 bpm, shallow compression being the most common mistake. 14.8% of subjects had predominantly adequate compression rate and depth.

After the first testing phase, the subject was given the CPR Device and asked to complete 1 minute of CPR with the device on the testing apparatus. Though there was no change seen in 14.8% of subjects who already performed adequate CPR, in all 81.4% of those who did not provide adequate CPR techniques initially, all performed adequate CPR with the device. In one case, the user actually covered the speaker hole, could not hear the metronome, and actually saw a negative effect in their compression rate. In an exit survey, however, all 81.4% improved effectiveness and 14.8% unchanged effectiveness groups stated that they felt more confident using the device, for a 96% confidence level.

## **Chapter 7. Ethics**

The Biomedical Engineering Society has created a code of ethics that all biomedical engineers must follow that can be found in Appendix 6 of this paper. The most important section of this code for the design and use of a CPR device is research obligations, which include

- Complying fully with legal, ethical, institutional, governmental, and other applicable research guidelines, respecting the rights of and exercising the responsibilities to colleagues, human and animal subjects, and the scientific and general public.
- Publishing and/or presenting properly credited results of research accurately and clearly.

In order for a medical device to be commercially used, it must pass 10 FDA stages of clinical tests. A CPR Device would be classified as a Class III Device, due to the very high risk associated with cardiac arrest, which requires clinical trials, 510 (k) premarket applications, and final FDA approval. The biggest setback is the clinical testing, in order to ensure that the user and patient's safety are mitigated. Clinical testing requires doctor and patient consent and approval, and in the case of CPR Devices, potentially another untrained bystander user's approval as well. To obtain consent from an unconscious patient under cardiac arrest without a Do Not Resuscitate order, and obtain consent from a randomized bystander rescuer and doctor during a critical life-or-death emergency, would be very limiting. A controlled simulation on cadavers or other prosthetic system could be used instead of clinical trials, however these simulations would not take the randomized bystander-user's stressed state during an actual cardiac arrest. If setbacks can be solved, then most of the medical approval ethics will be addressed.

The largest legal issue with a CPR device is that in a tragic case where the user misused the device, the device had a failure to work, or even when everything is done correctly but the patient did not survive, there is a chance that the consumer may find the CPR Device liable for the occurrence.

Although there are no political ramifications associated with the design and use of a CPR device, there are some related legal ramifications. If a patient receives CPR through the use of a CPR device and suffers an injury, they could sue either the person who performed the CPR and/or the company that manufactures the CPR device. If a user of the device injures the patient receiving CPR or fails in resuscitation of the patient, they could also sue the company that manufactures the device. Hence, the ethical issues surrounding bystander CPR assistive devices are still unresolved.

## **Chapter 8. Conclusion**

Our device satisfied all of our design and functional requirements. It is able to withstand 125 lbs of force with a safety factor of 3. The device is economical as its bulk manufacturing cost is below 25 dollars per unit. It is both portable and lightweight as the device is under one pound and is approximately the size of a cell phone. The device is also easy to understand thanks to our label which clearly states the function of the various components of the device. Overall our device provides bystanders with a 96% confidence level when applying appropriate CPR.

## **Chapter 9. Future Plans**

In the device's current state, it is not as accurate as the group would have hoped. For future iterations, the device's depth recognizing algorithm would be adjusted to favor false negatives over false positives. By doing so, this would promote a depth of compressions that are further within the range of adequacy which is favorable to the converse. If the range for adequate depth of compressions was adjusted to favor false negatives, the lower bound of the range would increase and the upper bound of the range would decrease. This would therefore encourage the user to apply an adequate depth of compression more often, rather than returning false positives and encouraging the user to apply improper compressions.



One hundred 2-minute samples simulating bystander CPR with our device were performed. For the future, much more testing should take place in order to better represent both the accuracy and effectiveness of the device. Significantly more samples should be taken and analyzed in order to determine if the aforementioned change in the device's algorithm increased its accuracy or not. This process of collecting samples and editing the algorithm respectively should be repeated until the device attains a high accuracy level.

For the future, more subject testing should also occur. If much more subject testing occurs, the device's effectiveness of providing a 96% confidence level for a bystander performing appropriate CPR techniques can only become more accurate. The more subject testing that also occurs, the more feedback the group receives about the device's physical features such as its portability, its weight, whether it is ergonomic or not, and whether the audio and visual feedback are working effectively. Overall, increased subject testing will provide the group with the necessary information to make changes where necessary based on consumer specifications and will verify its effectiveness.

For future case iterations, the team would like to find a new battery solution in order to make the case less bulky. We would also thin the walls of the case to more closely match the safety factor of 3 as it is currently around 10. With a thinner case, it would reduce injection molding cooling time and material usage, reducing the overall cost of case manufacturing

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# Appendix

## Appendix 1. Testing Procedure

The following procedures are guidelines for the various tests performed on the CPR Device to ensure its viability in giving feedback on compression depth and rate to rescuers during CPR. Table 1 names terms or items used throughout testing and specifically defines the terms.

Table 1. Testing Terms

<b>Term/ Item</b>	<b>Definition</b>
Device	The constructed MQP CPR Device.
Testing Apparatus	The constructed spring-box calibrated at X lb-force to be used to test device compression, shown in Figure 1.
Rescuer	The person performing CPR.
Proper CPR Technique	The rescuer places hands one on top of the other, locks elbows straight, and compresses down 2.0-2.4 inches at 100-120 beats per minute, while allowing for full chest recoil in-between compressions.
Flat Surface	Any table with a level-measured flat surface.
Timer	A timing device with ranges from 01 seconds to 99 minutes.
Stopwatch	A hand-held timing device with ranges from 0.01 seconds to 99 minutes.
Tally Counter	A hand-held counting device with ranges from 0 to 9999.
Recording Device	An excel document of all data observed during testing.

The Testing Apparatus works by compressing the top plate downwards to the desired depth. Four springs inside of the device are calibrated to exert a 60 lb-force resistance, similar to the forces seen during chest compression during CPR. In this testing sequence, the device will be placed in the center of the top plate of the Testing Apparatus, and then the user will compress directly on the device to desired depth.

### Test 1: No Compression Test

#### Materials:

- Flat Surface
- The Device
- Timer

#### Protocol:



1. Place the Device on a flat surface and turn on.
2. Observe and record device feedback for 2 minutes.
3. Turn off device.

Projected Outcome:

The Device should turn on and begin a metronome buzzing. The LEDs on the device should light up throughout the entire test to indicate that the compression depth is not adequate; compressions are too shallow.

Test 2: Held Compression Test

Materials:

- Testing Apparatus
- The Device
- Timer

Protocol:

1. Place the Device in a neutral position on the Testing Apparatus and turn on.
2. Compress device at 2 inches and hold for 2 minutes.
3. Observe and record device feedback.
4. Turn off device.

Projected Outcome:

The Device should turn on and begin a metronome buzzing. The LEDs on the device should light up throughout the entire test to indicate that the compression depth is not adequate; compressions do not allow for recoil.

Test 3: Compression Test

Materials:

- Testing Apparatus
- The Device
- Timer

Protocol:

1. Place the Device in a neutral position on the Testing Apparatus and turn on.
2. Compress device with proper CPR technique for 2 minutes.
3. Observe and record device feedback.
4. Turn off device.

Projected Outcome:

The Device should turn on and begin a metronome buzzing. The LEDs on the device should light up throughout the entire test to indicate that the compression depth is adequate and allows for proper recoil.

Test 4: Shallow Compression Test

Materials:

- Testing Apparatus
- The Device
- Timer

Protocol:

1. Place the Device in a neutral position on the Testing Apparatus and turn on.
2. Compress device with proper CPR technique for 2 minutes.
3. Observe and record device feedback.
4. Turn off device.

Projected Outcome:

The Device should turn on and begin a metronome buzzing. The LEDs on the device should light up throughout the entire test to indicate that the compression depth is not adequate; compressions are too shallow.

Test 5: Deep Compression Test

Materials:

- Testing Apparatus
- The Device
- Timer

Protocol:

1. Place the Device in a neutral position on the Testing Apparatus and turn on.
2. Compress device with proper CPR technique for 2 minutes.
3. Observe and record device feedback.
4. Turn off device.

Projected Outcome:

The Device should turn on and begin a metronome buzzing. The LEDs on the device should light up throughout the entire test to indicate that the compression depth is not adequate; compressions are too deep.

Test 6: Period Test

Materials:

- Flat Surface
- The Device
- Stopwatch

Protocol:

1. Place the device on a flat surface and turn on.
2. With a stopwatch, record the time between one metronome buzz and the next.
3. Record the time and repeat 10 times.

4. Turn off device.

Projected Outcome:

The Device should turn on and begin a metronome buzzing. Since correct compression rate ranges from 100 - 120 beats per minute, the average time between beats should be between 1.67 - 2 seconds. The recorded times should fall within that range.

Test 7: Frequency Test

Materials:

- Flat Surface
- The Device
- Timer
- Tally Counter

Protocol:

1. Set up a timer for 60 seconds.
2. Place the device on a flat surface and turn on.
3. Start the timer, and begin counting the metronome buzzes with a counter.
4. Record the number of buzzes when the timer stops.
5. Turn off device.

Projected Outcome:

The Device should turn on and begin a metronome buzzing. Since correct compression rate ranges from 100 - 120 beats per minute, the number of buzzes in a minute should be between that.

Test 8: 1-Rescuer Longevity Test

Materials:

- Testing Apparatus
- The Device
- Timer
- Tally Counter

Protocol:

1. Set up a timer for 17 minutes.
2. Place the Device in a neutral position on the Testing Apparatus and turn on.
3. Allow Rescuer to compress device with proper CPR technique for 17 minutes.
4. At times 0, 5, 10, and 15 perform Tests 3 and 7.
5. Record all data and turn device off.

Projected Outcome:

The Device should turn on and begin a metronome buzzing. At every time interval, the LEDs on the device should light up to indicate that the compression depth is adequate and allows for proper recoil, and the metronome should buzz between 100- 120 beats per minute.

## Test 9: 2-Rescuer Longevity Test

### Materials:

- Testing Apparatus
- The Device
- Timer
- Tally Counter
- 2 Rescuers

### Protocol:

1. Set up a timer for 16 minutes.
2. Place the Device in a neutral position on the Testing Apparatus and turn on.
3. Allow Rescuer 1 to compress device with proper CPR technique for 2 minutes and then switch positions with Rescuer 2 (repeat switching every 2 minutes).
4. At times 0, 2, 4, 6, 8, 10, 12, and 14 perform Tests 3 and 7.
5. Record all data and turn device off.

### Projected Outcome:

The Device should turn on and begin a metronome buzzing. At every time interval, the LEDs on the device should light up to indicate that the compression depth is adequate and allows for proper recoil, and the metronome should buzz between 100- 120 beats per minute.

## Test 10: Limits Test

### Materials:

- Testing Apparatus
- The Device
- Timer
- Tally Counter

### Protocol:

1. Set up a timer for 42 minutes.
2. Place the Device in a neutral position on the Testing Apparatus and turn on.
3. Allow Rescuer to compress device with proper CPR technique for 42 minutes.
4. At times 0, 5, 10, 15, 20, 25, 30, 35, 40 perform Tests 3 and 7.
5. Record all data and turn device off.

### Projected Outcome:

The Device should turn on and begin a metronome buzzing. At every time interval, the LEDs on the device should light up to indicate that the compression depth is adequate and allows for proper recoil, and the metronome should buzz between 100- 120 beats per minute.

## Test 11: Accelerometer Test

### Materials:

- Testing Apparatus

- The Device
- Timer

Protocol:

1. Set up a timer for 1 minutes
2. Place the Device in a neutral position on the Testing Apparatus and turn on. Leave device for 1 minute.
3. Take the Device off of the neutral position for 30 seconds.
4. Then place the Device back on the original position.

Projected Outcome:

The Device should turn on and begin a metronome buzzing. The Device should recalibrate it's position.

## Appendix 2. Code for MPU6050

```
#include<Wire.h>

const int MPU_addr=0x68; // I2C address of the MPU-6050

uint8_t i2cData[4] = { // Buffer for I2C data
  7, // Set the sample rate to 1000Hz - 8kHz/(7+1) = 1000Hz
  0x00, // Disable FSYNC and set 260 Hz Acc filtering, 256 Hz Gyro filtering, 8 KHz sampling
  0x00, // Set Gyro Full Scale Range to  $\pm 250$ deg/s
  0x10 // Set Accelerometer Full Scale Range (  $\pm 2g = 0x00$   $\pm 4g=0x01$   $\pm 8g=0x10$ ,  $\pm 16g = 0x11$  )
};

const int offsetSamples = 1000, restSamples = 5;
double Acc_threshold[] = {50, 50};

int16_t Tmp;
int16_t input[6] = {0, 0, 0, 0, 0, 0};
int16_t offset[6] = {0, 0, 0, 0, 0, 0};
double Acc[6], Vel[6], Dis[6];

long t, last_t = 0, last_f = 0;

long Acc_FIFO[restSamples];
int Acc_index = 0;
boolean atRest = false;
void setup(){
  Wire.begin();
  Serial.begin(9600);
  while(!Serial);

  //Wait until the accelerometer responds
  while(input[2] == 0){
    initMPU6050();
    readMPU6050();
  }

  //Calculate at rest offset
  for(int i = 0; i < offsetSamples; i++){
    readMPU6050();
    for(int j = 0; j < 6; j++){
      offset[j] += input[j];
    }
  }
}
```

```

}

for(int i = 0; i < 6; i++){
    offset[i] /= offsetSamples;
}
last_t = micros();
}
void loop(){
    if(input[2] == 0) initMPU6050();
    readMPU6050();

    t = micros();
    int16_t d_t = t - last_t;

    //Calculate acceleration and gyro
    for(int i = 0; i < 6; i++){
        Acc[i] = input[i] - offset[i];
    }

    //Determine if the device is moving
    //if(Acc > Acc_threshold[0] && Acc < Acc_threshold[1] ) Acc = 0;
    if(Acc_index >= restSamples) Acc_index = 0;
    Acc_FIFO[Acc_index] = Acc[2];
    atRest = true;
    for(int i = 0; i < restSamples; i++){
        if(Acc_FIFO[i] < Acc_threshold[0] || Acc_FIFO[i] > Acc_threshold[1]){
            atRest = false;
        }
    }
}

//Calculate velocity
for(int i = 0; i < 6; i++){
    Vel[i] += Acc[i] * d_t/1000000;
    if(atRest) Vel[i] = 0;

    //Calculate displacement
    Dis[i] += Vel[i] * d_t/1000000;
}
//Print data
Serial.print("t = "); Serial.print(t); Serial.print(" Acc = "); Serial.print(Acc[2]); Serial.print(" Vel = ");
Serial.print(Vel[2]); Serial.print(" Dis = "); Serial.println(Dis[2]);

```

```

if(t > last_f + 1000){
  last_f = t;
}

last_t = t;

}

void initMPU6050(){
  Wire.beginTransmission(MPU_addr);

// Wire.write(0x19); //Write to register
// Wire.write(i2cData, 4); //Data to be written
  Wire.write(0x6B); // PWR_MGMT_1 register
  Wire.write(0); // set to zero (wakes up the MPU-6050)

  Wire.endTransmission(true);
}

void readMPU6050(){
  Wire.beginTransmission(MPU_addr);
  Wire.write(0x3B); // starting with register 0x3B (ACCEL_XOUT_H)
  Wire.endTransmission(false);
  Wire.requestFrom(MPU_addr,14,true); // request a total of 14 registers
  input[0]=Wire.read()<<8|Wire.read(); // 0x3B (ACCEL_XOUT_H) & 0x3C (ACCEL_XOUT_L)
  input[1]=Wire.read()<<8|Wire.read(); // 0x3D (ACCEL_YOUT_H) & 0x3E (ACCEL_YOUT_L)
  input[2]=Wire.read()<<8|Wire.read(); // 0x3F (ACCEL_ZOUT_H) & 0x40 (ACCEL_ZOUT_L)
  Tmp=Wire.read()<<8|Wire.read(); // 0x41 (TEMP_OUT_H) & 0x42 (TEMP_OUT_L)
  input[3]=Wire.read()<<8|Wire.read(); // 0x43 (GYRO_XOUT_H) & 0x44 (GYRO_XOUT_L)
  input[4]=Wire.read()<<8|Wire.read(); // 0x45 (GYRO_YOUT_H) & 0x46 (GYRO_YOUT_L)
  input[5]=Wire.read()<<8|Wire.read(); // 0x47 (GYRO_ZOUT_H) & 0x48 (GYRO_ZOUT_L)
  Serial.print("GyX = "); Serial.print(input[2]); Serial.print(" GyY = "); Serial.println(input[0]);

// Serial.print(" | Tmp = "); Serial.print(Tmp/340.00+36.53); //equation for temperature in degrees C
// from datasheet
}

//Z offset: 16959

```



## Appendix 3. Code for BNO055

```
#include <Wire.h>
#include <Adafruit_Sensor.h>
#include <Adafruit_BNO055.h>
#include <utility/imuMaths.h>

Adafruit_BNO055 bno = Adafruit_BNO055();

double offset = 0, acc, vel, dis;
long d_t, l_t, t;xx
long l_f;

const int OFFSET_SAMPLES = 50, BUFFER_SIZE = 256, REST_SAMPLES = 5, FRAME_SIZE =
100, EXTREMES_BUFFER_SIZE = 10;
double acc_threshold[] = {-0.1, 0.1};

double acc_buffer[BUFFER_SIZE], peak_buffer[EXTREMES_BUFFER_SIZE],
valley_buffer[EXTREMES_BUFFER_SIZE];
int acc_index = 0;
boolean at_rest = false;

const int BUZZER_FREQUENCY = 100; //Frequency in Hz
const int BUZZER_DURATION = 100000; //Time the buzzer is on in us

boolean buzzer_on = false;

const int BUZZER_PIN = 4, LED_1_PIN = 5, LED_2_PIN = 6, LED_3_PIN = 7;

void setup(void){
  Serial.begin(9600);

  initializeBNO();
  initializePins();

  calculateOffset();

  l_t = micros();
  l_f = l_t;
}

void loop(void){
```

```

//Get change in time since last sample
t = micros();
d_t = t - l_t;

acc = getAcceleration();

//Populate buffer
acc_buffer[acc_index] = acc;
acc_index++;
if(acc_index == BUFFER_SIZE) acc_index = 0;

//Iterate through buffer to determine if the device is at rest
at_rest = atRest();

// //Calculate velocity
// vel += acc * d_t/1000000;
// if(at_rest)vel = 0;
//
// //Calculate displacement
// dis += vel * d_t/1000000;

runBuzzer();

printResults();

//Updates the last time for d_t calculation
l_t = t;
}

void initializeBNO(){

/* Initialise the sensor */
if(!bno.begin()){
/* There was a problem detecting the BNO055 ... check your connections */
Serial.print("Ooops, no BNO055 detected ... Check your wiring or I2C ADDR!");
while(1);
}

bno.setExtCrystalUse(true);

}

```

```

void initializePins(){

    pinMode(BUZZER_PIN, OUTPUT);
    pinMode(LED_1_PIN, OUTPUT);
    pinMode(LED_2_PIN, OUTPUT);
    pinMode(LED_3_PIN, OUTPUT);

}

void calculateOffset(){

    double acceleration;
    for(int i = 0; i < OFFSET_SAMPLES; i++){
        acceleration = getRawAcceleration();
        offset += acceleration;
    }

    offset /= OFFSET_SAMPLES;

}

double getRawAcceleration(){

    //Get data from sensor
    imu::Vector<3> linear_acceleration = bno.getVector(Adafruit_BNO055::VECTOR_LINEARACCEL);
    imu::Vector<3> gravity = bno.getVector(Adafruit_BNO055::VECTOR_GRAVITY);

    //Record magnitude of acceleration, minus gravity
    double magnitude = linear_acceleration.magnitude();

    //Normalize vectors to reduce to unit vectors
    linear_acceleration.normalize();
    gravity.normalize();

    //Return acceleration in the direction of gravity
    return magnitude * linear_acceleration.dot(gravity);

}

double getAcceleration(){

    //Get current acceleration
    double acceleration = getAcceleration();

```

```

//Adjust acceleration for offset
acceleration -= offset;

//Filter low acceleration noise
if(acceleration > acc_threshold[0] && acceleration < acc_threshold[1])acceleration = 0;

return acceleration;
}

boolean atRest(){
boolean rest = true;
for(int i = acc_index; i < acc_index + REST_SAMPLES; i++){
if(i >= BUFFER_SIZE){
if(acc_buffer[i - BUFFER_SIZE] != 0)rest = false;
}
else{
if(acc_buffer[i] != 0)rest = false;
}
}
return rest;
}

void runBuzzer(){

//Determine when to turn on and off the buzzer
if(!buzzer_on && t > l_f + (60*1000000/BUZZER_FREQUENCY) - BUZZER_DURATION){
digitalWrite(BUZZER_PIN, HIGH);
buzzer_on = true;
l_f = t;
}else if(buzzer_on && t > l_f + BUZZER_DURATION){
digitalWrite(BUZZER_PIN, LOW);
buzzer_on = false;
l_f = t;
}
}

void printResults(){

//Text output
// Serial.print("t = "); Serial.print(t); Serial.print(" acc = "); Serial.print(acc); Serial.print(" vel = ");
Serial.print(vel); Serial.print(" dis = "); Serial.println(dis);
Serial.println(acc);

}

```

## Appendix 4. IRB Approved Subject Testing Guidelines

### WORCESTER POLYTECHNIC INSTITUTE

100 INSTITUTE ROAD, WORCESTER MA 01609 USA

#### Institutional Review Board

FWA #00015024 - HHS #00007374

##### Notification of IRB Approval

**Date :** 26-Mar-2019

**PI:** Sullivan, John M  
**Protocol Number:** IRB-19-0397  
**Protocol Title:** User Interface Testing of CPR Device

**Approved Study Personnel:** Sullivan, John M~Foley, Zoey~Abram, Benjamin~Ringuette, John~

**Start Date:** 26-Mar-2019  
**Expiration Date:** 25-Mar-2020

**Review Type:** Initial Review  
**Review Method:** Expedited Review  
**Risk Level:** Minimal Risk

**Sponsor\*:**

The WPI Institutional Review Board (IRB) approves the above-referenced research activity, having conducted a review according to the Code of Federal Regulations (45 CFR 46).

This approval is valid through 25-Mar-2020 unless terminated sooner (in writing) by yourself or the WPI IRB. Research activities involving human subjects may not continue past the expiration date listed above, unless you have applied for and received a renewal from this IRB.

We remind you to only use the stamped, approved consent form, and to give a copy of the signed consent form to each of your subjects. You are also required to store the signed consent forms in a secure location and retain them for a period of at least three years following the conclusion of your study. You are encouraged to use the InfoEd system for the storage of your consent forms.

Amendments or changes to the research must be submitted to the WPI IRB for review and approval before such changes are put into practice.

Investigators must immediately report to the IRB any adverse events or unanticipated problems involving risk to human participants.

Please contact the IRB at [irb@wpi.edu](mailto:irb@wpi.edu) if you have any questions.

\*if blank, the IRB has not reviewed any funding proposal for this protocol

## **Informed Consent Agreement for Participation in a Research Study**

**Investigator: John Sullivan**

**Contact Information: sullivan@wpi.edu**

**Title of Research Study: User Interface Testing of CPR Device**

### **Introduction**

You are being asked to participate in a research study. Before you agree, however, you must be fully informed about the purpose of the study, the procedures to be followed, and any benefits, risks or discomfort that you may experience as a result of your participation. This form presents information about the study so that you may make a fully informed decision regarding your participation.

**Purpose of the study:** To evaluate how easily the MQP CPR Device can be used and understood, as in how long it takes the user to turn on the device and to begin using the device properly. This study will be used to discover any user-interface setbacks so that the device can be improved upon.

**Procedures to be followed:** You will first be asked to perform proper CPR hands-only compressions on a spring-loaded testing device located on the floor for 1 minute. Then, you will be asked to turn on the CPR Device and use it to perform compressions for 1 minute on top of the testing platform, located on the floor.

**Risks to study participants:** This study requires you to be able to sit on your knees on the floor for 2 minutes, which might cause injury, especially to those with pre-existing lower extremity injuries. This study also requires you to be able to compress on a device on a 60-pound spring-device, which might cause injury to the arms, shoulders, or back.

**Benefits to research participants and others:** There are none.

**Record keeping and confidentiality:** You will be videotaped for later quantitative analysis in depth compression and rate. Data collected from this study will be accessed by all members of the MQP team. No identifying data will be used, just quantitative data on the success of the user tests. Records of your participation in this study will be held confidential so far as permitted by law. However, the study investigators, the sponsor or its designee and, under certain circumstances, the Worcester Polytechnic Institute Institutional Review Board (WPI IRB) will be able to inspect and have access to confidential data that identify you by name. Any publication or presentation of the data will not identify you.

**Compensation or treatment in the event of injury:** Should any injury occur, the subject will be assisted to WPI Health Services and/or WPI EMS services will be contacted. Participants will be responsible for the cost of their own treatment, and they

**APPROVED BY  
WPI IRB 1  
3/26/2019 to 3/25/2020**

will not be compensated by the researchers. You do not give up any of your legal rights by signing this statement.

**For more information about this research or about the rights of research participants, or in case of research-related injury, contact:** John Sullivan at [sullivan@wpi.edu](mailto:sullivan@wpi.edu). In addition, you may contact the IRB Chair, Professor Kent Rissmiller, Tel. 508-831-5019, Email: [kjr@wpi.edu](mailto:kjr@wpi.edu), as well as the Human Protection Administrator, Gabe Johnson, Tel. 508-831-4989, Email: [gjohnson@wpi.edu](mailto:gjohnson@wpi.edu).

**Your participation in this research is voluntary.** Your refusal to participate will not result in any penalty to you or any loss of benefits to which you may otherwise be entitled. You may decide to stop participating in the research at any time without penalty or loss of other benefits. The project investigators retain the right to cancel or postpone the experimental procedures at any time they see fit.

**By signing below,** you acknowledge that you have been informed about and consent to be a participant in the study described above. Make sure that your questions are answered to your satisfaction before signing. You are entitled to retain a copy of this consent agreement.

\_\_\_\_\_  
Study Participant Signature

Date: \_\_\_\_\_

\_\_\_\_\_  
Study Participant Name (Please print)

\_\_\_\_\_  
Signature of Person who explained this study

Date: \_\_\_\_\_

**APPROVED BY  
WPI IRB 1  
3/26/2019 to 3/25/2020**

**CPR Testing Pre-Test**

**Have you ever been certified in CPR?                      YES                      NO**

**What is the proper depth to compress on the chest for CPR?**

**At what beats per minute should compressions be performed at?**

**How confident do you feel in your ability to perform correct CPR?**

I Feel Don't Feel Confident in Performing CPR.

I Feel Somewhat Confident in Performing CPR.

I Feel Very Confident in Performing CPR.



## **CPR Device Testing Follow-Up**

These questions are to be asked to every Subject after conducting the CPR User Interface Test. Participation is not required, and there are no consequences to not answer the following questions.

Please answer these questions to the best of your ability. No identifying data will be used in the recording of these answers.

**In your opinion, how did your confidence in performing proper CPR technique change with the use of the MQP CPR Device?**

- I Feel Less Confident in Performing CPR with the CPR Device.
- I Feel Slightly Less Confident in Performing CPR with the CPR Device.
- There was No Change in my Confidence of Performing CPR with the CPR Device.
- I Feel Slightly More Confident in Performing CPR with the CPR Device.
- I Feel More Confident in Performing CPR with the CPR Device.

**During testing, did you find yourself confused by any functions? If so, which functions?**

**Do you have any further comments that you could share on device usage, engineering gaps, ease of use or simplicity?**



## Appendix 6. Biomedical Engineering Society Code of Ethics

Biomedical engineering is a learned profession that combines expertise and responsibilities in engineering, science, technology, and medicine. Since public health and welfare are paramount considerations in each of these areas, biomedical engineers must uphold those principles of ethical conduct embodied in this Code in professional practice, research, patient care, and training. This Code reflects voluntary standards of professional and personal practice recommended for biomedical engineers.

### Biomedical Engineering Professional Obligations

Biomedical engineers in the fulfillment of their professional engineering duties shall:

1. Use their knowledge, skills, and abilities to enhance the safety, health, and welfare of the public.
2. Strive by action, example, and influence to increase the competence, prestige, and honor of the biomedical engineering profession.

### Biomedical Engineering Health Care Obligations

Biomedical engineers involved in health care activities shall:

1. Regard responsibility toward and rights of patients, including those of confidentiality and privacy, as their primary concern.
2. Consider the larger consequences of their work in regard to cost, availability, and delivery of health care.

### Biomedical Engineering Research Obligations

Biomedical engineers involved in research shall:

1. Comply fully with legal, ethical, institutional, governmental, and other applicable research guidelines, respecting the rights of and exercising the responsibilities to colleagues, human and animal subjects, and the scientific and general public.
2. Publish and/or present properly credited results of research accurately and clearly.

### Biomedical Engineering Training Obligations

Biomedical engineers entrusted with the responsibilities of training others shall:

1. Honor the responsibility not only to train biomedical engineering students in proper professional conduct in performing research and publishing results, but also to model such conduct before them.

2. Keep training methods and content free from inappropriate influence from special interests.