BME 400: Biomedical Engineering Design (Fall 2010)
Project: Skin Applicator

Final Report
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Table of Contents

Abstract 3

Problem Statement 3

Background 3

Design Motivation 6

Client Requirements 7

Human Factors & Ergonomics 8

Current Devices 11

Permanent Features of the Design 12

Final Designs 13

Future Work

- Large-scale manufacturing of current device **15**
- Future design alternatives **17**
- Potential testing **18**

References 20

Appendix 21

- A) Calculation – three-point bending **21**
- B) Calculation – cantilevered bending **23**
- C) Product Design Specifications (PDS) **25**
- D) CAD drawings of the 1st design **27**
- E) CAD drawings of the 2nd design **28**
- F) Budget analysis **29**

Abstract

During this semester we were charged with the task of designing an applicator device for a topical drug solution. The motivation for this design comes from our client, Dr. Bill Fahl, who—along with his associates—has developed a drug for the prevention of radiation-induced burns. Our client had a few main requirements for this design, and over the course of this semester we sought to create a prototype which conformed to these standards. In the end, we came up with two separate devices which we believe each have unique and redeeming qualities over current devices on the market. In the future we will attempt to contribute additional design alternatives, test all design prototypes, and pursue large-scale manufacturing of one of these devices via injection molding.

Problem Statement

The goal of this project is to design a disposable applicator for topical, drug-containing solutions. The device should be able to apply 8.0 mL of the topical solution to a cancer patient receiving radiotherapy over the course of 30 days. Some possible issues to address are ergonomics of the device, control of flow-rate, and ease of use. There is a current study under way which could make use of such a device.

Background

The background for this project comes from our client, Dr. Bill Fahl, and his ongoing research as a member of the Department of Oncology at UW-Madison. A major concern in the use of radiation therapy is the development of radiation burns to the skin, and a sufficient treatment

method does not currently exist [1, 2]. Some of the symptoms of these radiation burns include dryness, itching, peeling, or blistering of the skin [4]. While radiation burns are a major side effect, radiation therapy is often a necessity for cancer patients as there is a significantly reduced recurrence rate for patients who choose to undergo radiation [B. Fahl, personal communication, September, 17, 2010, 4]. While there are currently several methods for treating radiation-induced burns to the skin, most of these are aimed at treating the burns after the fact (i.e. attempting to treat the burns once they have already occurred), and few preventative measures exist. Historically, severe radiation burns have been treated by surgical removal of the necrotic tissue [2]. Also, patients have often used many topical treatments post-radiation exposure—including topical lidocaine, aloe vera gel, and topical corticosteroids—to soothe the burns [4]. One particular study examined the use of silver-leaf dressing as a possible treatment for radiation burns [3]. More recently, cellular-based therapies have been combined with standard surgical techniques [1, 2]. These treatments make use of the regenerative capacity of mesenchymal stem cells (MSCs) and have shown favorable results [1, 2]. However, none of these treatments are able to prevent the initial occurrence of radiation burns.

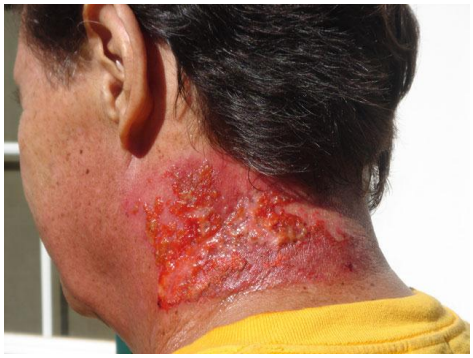

Figure 1
Severe dermatitis
Source: http://www.cancer-throat.com/index.php?s=lubricate

Radiation burns occur by the same mechanism which destroys cancerous tissues and saves lives. During radiation therapy, a high energy electron beam is focused and directed at the affected

tissue [B. Fahl, personal communication, September 17, 2010]. The emitted photons and particles destroy the cancerous tumor by disrupting the DNA of the cancerous cells and causing apoptosis, or cell death [5]. The DNA can be damaged by direct ionization of the DNA strands. Also, the DNA can be indirectly damaged by oxygen free radicals which are formed via the interaction of the electron beam with oxygen-rich tissues, such as blood [5]. While radiation therapy has proven highly effective at reducing the recurrence rates in cancer patients, the side effects are often substantial and painful [B. Fahl, personal communication, September 17, 2010]. One of the main side effects of radiation therapy, as mentioned previously, is radiation burns, or dermatitis. These burns occur when the highly charged particles directed at the cancerous tissue interact with the oxygen-rich compounds in the dermal tissues, causing the formation of oxygen-based free radicals [B. Fahl, personal communication, September 17, 2010]. The free radicals attack or interact with the DNA of the healthy skin cells in much the same way as they do with the DNA of the cancerous cells, and cell death is a direct side effect [5]. These radiation burns can range from moderate to severe, and the regenerative capacity of the skin is often damaged as a result [5]. Thus, while radiation therapy remains a necessary treatment of cancerous tissues, radiation burns are a significant side effect causing long-term discomfort and pain to patients.

Our client and his associates have developed a drug which, in contrast to previous treatments, is aimed at preventing radiation burns. The drug's active compound is the well-known neurotransmitter, norepinephrine (see Figure 2). Most of the oxygen-based free radicals which form in the dermal tissues during radiation therapy and cause radiation burns are derived from the blood stream [B. Fahl, personal communication, September 17, 2010]. Norepinephrine interacts with the smooth muscle lining blood vessels to stimulate the smooth muscle to contract

and the blood vessels to constrict (i.e. it is a vasoconstrictor) [B. Fahl, personal communication, September 17, 2010]. Stimulation by norepinephrine thus restricts blood supply to the regions where it is active. The goal of our client's drug thus is to restrict blood supply to the area of skin immediately prior to radiation treatment and limit oxygen-based free radical formation in the tissue (and thus limit tissue damage). This technique is especially unique in that it has the potential to prevent radiation burns from even occurring and thus has potential widespread applicability and use.

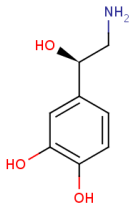

Figure 2
Norepinephrine
Source: http://www.bmrb.wisc.edu/metabolomics/mol_summary/?molName=Norepinephrine

The drug, as currently designed, is supplied to the skin in a 70:30 ethanol: water mixture [B. Fahl, personal communication, September 17, 2010]. Currently our client does not have an efficient and highly reproducible means of applying the drug to patients undergoing treatment. There are devices on the market which are used in the application of topical drug solutions, but due to cost reasons, at the current time do not fulfill our client's demands. Our client therefore has given us the task of designing device which can apply a fixed amount of drug to the skin of patients about to undergo radiation therapy.

Design Motivation

Given the clinical relevance of the aforementioned drug therapy, there is clearly a need for a device which can apply the norepinephrine drug solution in a consistent manner. There are several devices currently on the market which can deliver topical drug solutions, but there are serious costs and functional limitations which warrant the design of a new drug-delivery device

that is more optimally suited for our client's purposes. Based on this pressing need, our client has proposed several requirements to facilitate the design of a new, original device.

Client Requirements

Our client outlined several requirements for the prospective device, as well as some other potential improvements on the current devices being employed. The device should be able to apply 8.0 mL of the 70:30 ethanol:water mixture (containing the drug, norepinephrine) to a skin surface area of approximately 225 cm^2. This drug solution must be stored in a glass container, as this is the standard protocol for storage of such drug solutions in the clinical setting [B. Fahl, personal communication, September 17, 2010]. Furthermore, the device should be single-use (i.e. disposable). Given the clinical application for which the device is intended, it is imperative for such a device to be disposable to minimize the potential risk of transmission of infectious agents between patients. The device should be relatively light-weight and handheld (i.e. similar in size to current devices on the market). Another requirement of the client is for the device to deliver the drug solution in a controlled, consistent manner. This is most readily achieved by the diffusion rate of the drug solution through a foam sponge attached to the end of the applicator. As a potential improvement, some mechanism could be employed to further control the drug-release rate in a user-independent manner. Another possible improvement is to include a method of skin abrasion, to increase skin porosity and improve drug absorption. However, it is likely the friction provided by the foam sponge is sufficient to accomplish this objective.

Human Factors and Ergonomics

In designing our device we had to consider ease of use, as clinicians will be repeatedly using this device in pharmaceutical trials and potentially later on if the drug passes all FDA standards and trials. In fact, it is not unlikely that a nurse will have to perform 30-50 one-to-two minute drug applications during a given day [B. Fahl, personal communication, December 6, 2010]. Our first concern was ease in holding and application of the device after the drug is released. Current devices, with some variation, use a smooth cylindrical handle with no specific modifications to improve gripping capabilities. Thus, in constructing the main design alternatives, we sought to address this issue. For the first, simpler device, we decided to implement four ridges near the base of the device to improve ergonomics (See section entitled 'Final Design'). These ridges should improve gripping capabilities as the present increased surface and contact points for the hand of the clinician applying the drug. In the second device we attempted to take this same concept a step further, and model the handle based around the dimensions of the average human hand [10]. Also, in this more advanced design the handle is oriented parallel to direction of application, which we hypothesize will cause less stress and fatigue to the muscles of the hand and forearm.

In our first design, as we chose to implement a simple bending mechanism for drug release, and thus fracture of the glass ampoule, another concern related to ease of drug release. By this, we mean that we wanted to ensure the handle was easily bendable by the majority of clinicians who would be using the device. As we intend to pursue the large-scale manufacturing of our device out of high-density polyethylene (HDPE), we first needed to obtain the material properties of HDPE. We found, after some minimal online searching, that the published yield stress (the

stress level at which material begins plastic deformation and leaves elastic deformation) of HDPE was between 26-33 MPa [6]. Next, we needed to determine average human strength in the bending objects. While it was difficult to obtain published data on this exact action, we were able to find a human strength data table related to pushing and pulling forces [7]. Based on this data we chose to model the bending action as a pulling force created by an arm or arms acting perpendicular to the body and at a -30° angle from the horizontal (see Figure 3), and found that the average human arm strength in such a position is approximately 129 N (29 lb_f)[7]. Next, for determining the necessary force to be applied in bending the device handle, we made a few assumptions. We assumed that the device handle could be treated as a right, hollow cylinder. We also assumed the device material was homogeneous, isotropic, and behaved linearly elastically, which are all acceptable for the given loading state.

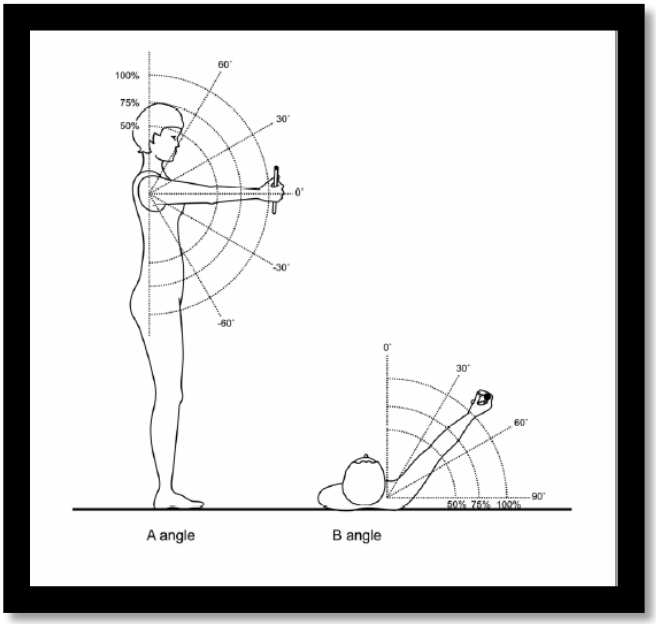

Figure 3 Arm position in bending. We selected Angle A = -30 degrees and Angle B = 0 degrees to model arm position in bending.

Source: http://www.theergonomicscenter.com/graphics/Workstation%20Design/Strength.pdf

We then first chose to model the handle as a hollow, cylindrical beam undergoing three-point bending (see Figure 4) with the hands applying the support forces and the thumbs applying the

central force. Based on this model and using shear/bending diagrams and bending stress equations we were able to determine that each hand would have to apply a force of approximately 138 N (31 lb_f) (see Appendix A). Obviously this result is high, as it exceeds the average human strength noted previously. However, we determined our model was an inaccurate representation of drug activation and adjusted our method accordingly.

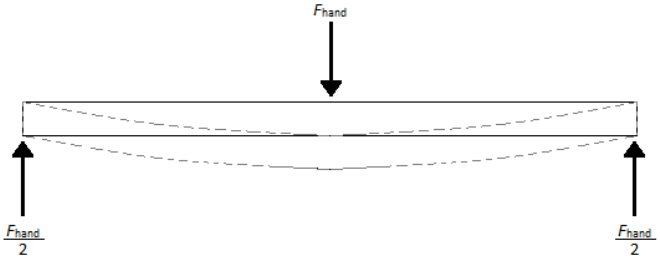

Figure 4

Three-point bending

We then considered the fact that, given the ridge supports added to the handle for ergonomic purpose, bending would likely occur right at the end of these supports and near the base of the handle. This is due to the large reduction in material strength at this point, as the average material thickness is rapidly decreased. Thus, we concluded that it would be more accurate to model device activation as a simple cantilever beam undergoing deflection, with one hand providing the rigid support, and the other hand applying a discrete (simplification used here, actually a distributed force) force at the other end (see Figure 5). Using this model and a similar computational method as described above we determined a required arm strength of approximately 71 N (16 lb_f), which is a significant improvement (see Appendix B).

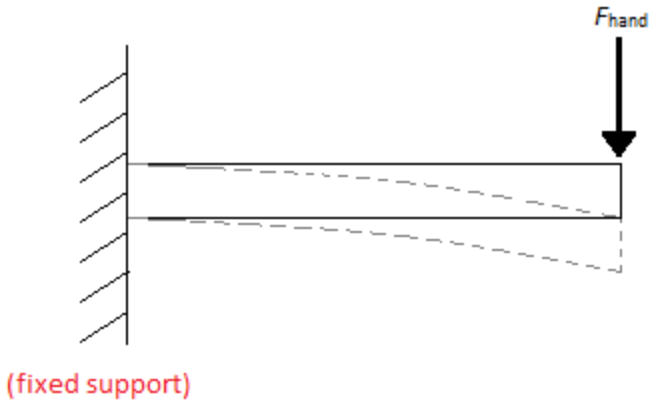

Figure 5

Cantilevered bending

An actual representation of the bending process going on here likely exists somewhere in between our two suggested models. Also, we made several assumptions, such as application of discrete forces, which are unrealistic and could significantly affect our outcome. However, both models found the required force to be near or below the average human arm force in such a scenario. Also, this force was the force required to yield the HDPE material, which will ultimately cause permanent plastic deformation. This force is likely much larger than the force we will require to bend our device, as a small deflection in the handle should be sufficient to fracture the internal glass ampoule, and no permanent, plastic deformation is required. Thus our calculations were based on "upper-bound" values to account for an extreme scenario. Although we will not entirely know until our device has been injection molded in HDPE, we conservatively estimate that the average clinician should have no reasonable trouble in using our first device.

Current Devices

Our client, Dr. Fahl, introduced us to several drug-solution application devices currently on the market (i.e. the ChloraPrep® skin applicators). These skin applicators are designed for different

amounts of drug solution and with various methods of solution release. These applicators apply an antimicrobial and antibacterial drug solution pre-surgery and use crushable glass ampoules as the mechanism for drug containment and delivery. These current designs could be modified to contain and apply our drug of interest, but there are cost-related and functionality-related concerns.

Currently, the ChloraPrep® applicator line comes in five sizes: 0.67mL, 1.5mL, 3mL, 10.5mL, and 26mL [11]. Each of these devices presents relatively creative and unique solutions to our client's requirements. However, we have identified several key issues with these devices: lack of cost-effectiveness, uncontrolled and inconsistent drug application, lack of an abrasive application surface to promote drug absorption, and poor ergonomics (see Figure 6). By designing a novel device, we intend to resolve these issues by proposing a solution which will prove superior over the ChloraPrep® applicators.

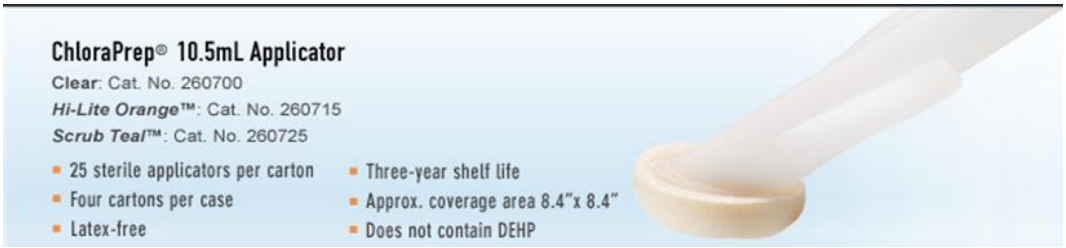

Figure 6

The 10.5 mL ChloraPrep® applicator

Permanent Features of the Design

Based on our discussion with the client, we identified three constant features of the design. First, a glass ampoule should be used for drug containment. This is a standard protocol for containing drug solutions in clinical settings due to the chemical inertness of glass, and our device should therefore incorporate such a glass ampoule accordingly. Second, reticulated polyurethane foam should be used to apply the drug solution because it has soft surface and can provide a sufficient

medium through which the drug solution can diffuse. Lastly, the skin applicator should have a polymer-based, chemically inert handle which will contain the glass ampoule and will be attached to the foam application surface.

Final Designs

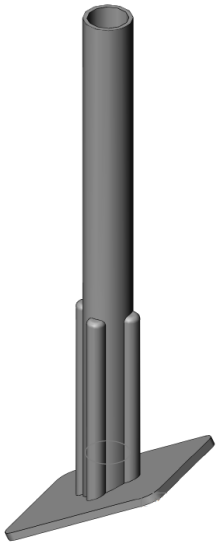

Figure 7

The first design

The first design is based on a very simple, user-friendly approach to deliver the drug solution, per our client's request (see Figure 7). In this design, the glass ampoule is housed in a hollow high-density-polyethylene (HDPE) cylindrical handle. A cap is included with small orifices to promote air flow. A reticulated-polyurethane foam sponge is attached to the base of the handle, and serves as the drug-solution delivery medium. Four symmetrically placed cylindrical grooves are also included near the base of the handle for increased ergonomics when handling the device. These grooves promote griping of the handle and prevent its slipping around in the hand while using the device to apply the drug solution.

In order to activate the device, the device is gripped with one hand near the base (such that the hand covers the cylindrical grips), and the other hand applies a force to the end of the handle.

Based on this cantilevered bending regime, the maximum moment is expected to occur at the end of the reinforcing cylindrical grips (see Appendix B). This force is then transmitted through the HDPE handle to the thin-walled glass ampoule, shattering it in the process. Given the thin wall of the glass ampoule, it should fracture easily without having to significantly yield the HDPE handle (i.e. it will remain intact to deliver the drug solution). Once the ampoule is broken, the drug solution runs through the bottom of the handle (via six hexagonally arranged outlet holes) and onto the reticulated-polyurethane foam sponge, which is attached to the base of the handle. The foam sponge is a critical component of the device, as it stores the drug prior to application on the skin, and serves as the diffusing medium for the drug solution. Once the foam sponge is saturated with the drug solution, the device can then be used to apply the drug solution onto the patient's skin.

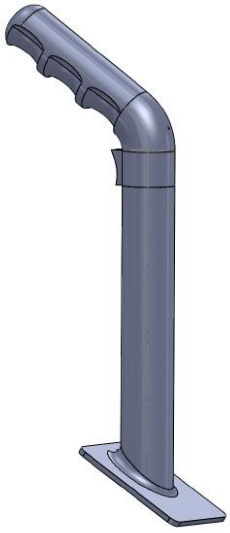

Figure 8

The second design

The second design also aims to present a simple, user-friendly solution to the client's current problem (see Figure 8). However, this design is complex both in means of drug solution release and ergonomic design. The device contains a handle which has been modeled to the average dimensions of the human hand [10]. Specifically, the handle has been modeled to accommodate

an optimal grip diameter. Also, the handle contains grooves modeled to fit the fingers of the average human hand, providing greater user comfort. We hope that this addition will create a more user-friendly design (See Appendix D).

The second main addition to this design in comparison to the first is the means by which the ampoule is fractured and the drug solution is released. Here, the ampoule is stored in a hollow chamber, compressing a polymer-based spring [13]. The device constrains the ampoule against the spring by tab extending from the handle into the hollow chamber. When the 'trigger' of the handle is squeezed, the tab releases the ampoule. Due to the high amount of energy stored in the spring, the ampoule is projected towards a pin at the base of the hollow chamber. This collision causes the tip of the ampoule to fracture, releasing the contents (i.e. the drug solution). The drug is then dispersed to a reticulated polyurethane foam sponge, as in the previous device, and subsequently applied to the skin. We believe that this device provides two main advantages over current devices. Namely, this device is more ergonomically sound than current devices, and also provides a more effective and user-friendly means of drug solution release than current devices (See Appendix E).

Future Work – Large-Scale Manufacturing of Device

As part of our future work for the following semester, we will need to implement large-scale fabrication of one of our design alternatives (on the order of hundreds to a thousand devices) for the disposable use of this device in our client's ongoing clinical trials. After consultation with the Polymers Engineering Department at UW-Madison, it has been determined that we will need to use a process known as injection molding in fabricating our device out of HDPE (see Figure 9)

[M. Jeng, personal communication, December 3, 2010] Essentially, this process consists of heating polymer resins to melting point, and then injecting this material into a mold, which is constantly undergoing cooling to solidify the polymer [8].

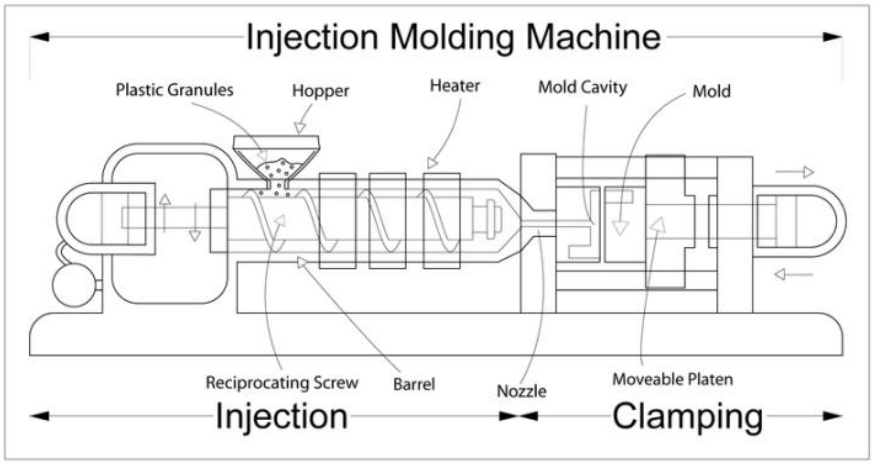

Figure 9 Simple diagram of an injection molding machine.

Source: http://en.wikipedia.org/wiki/Injection_molding

As a requirement of the injection molding process, we will need to have a mold fabricated. We will design this mold using SolidWorks ® software. We will need to get the mold machined out of some type of metal (typically steel or aluminum is used) [9]. We will also need to design the mold to have some type of inner channel through which cold water can flow to provide continual cooling of the mold [M. Jeng, personal communication, December 3, 2010]. This will facilitate solidification of the devices as they are manufactured and prevent warping of the plastic [M. Jeng, personal communication, December 3, 2010]. Ultimately, based on our client's requirements we will attempt to manufacture 500-1000 devices, and potentially more later on. Obviously, such large-scale fabrication makes repetitive 3d-printing, as was used for our initial prototypes, an unrealistic option.

Future Work – Future Design Alternatives

For the future, we would like to come up with at least two more design alternatives to supplement our testing. Also, our client has raised a few substantial and relevant concerns in regards to the current designs. The main issues relate to the second, more complex design. While we believe that this design presents several benefits over the first design, it may present a couple of negative factors which can be easily addressed. First, we are concerned that the location of the handle may be awkward, as it creates such high center of gravity for the device. One possible solution will be to simply shorten the handle, which should be easily doable as we designed the second device to contain at least 12 mL of fluid but only require 8. Also, we may attempt to place the handle lower down the main shaft of the design, which could possibly improve stability of the device and increase user comfort.

The other main issue that we may attempt to address is the location of the device trigger, as its current location may not be ideal for user ease. We will design the alternative devices with the trigger or tab on the front or side of the device, as we believe the thumb may better suited for gripping and squeezing the trigger. See Figure 10 for an example of a possible design alternative which we will pursue.

Figure 10 A potential future alternative design

Source: http://en.wikipedia.org/wiki/Injection_molding

Future Work – Testing

As part of our future work to be completed next semester, we will test the ergonomics of our various designs in comparison to the ChloraPrep® line of applicators. We foresee this testing taking on a three-part approach. First, we would like to measure the pressure gripping forces required in the hand over the application period. Next, we will attempt to measure the level of muscle activity in the forearm over the course of drug application. Lastly, we will have clinicians use the various devices and then fill out a survey to get user feedback.

The first part of our testing will be to discover which device requires the lowest pressure gripping forces over the course of drug application. There are a few main ways to do this, and we will likely only choose one. The first possible technique is to use pressure transducers (i.e. Wheatstone bridges) embedded in some type of gel, but this method is limited in situations where high pressure gradients are present [12]. Another possible method for monitoring pressure with respect to time employs a semi-conductive method [12]. However, there are significant hysteresis loss and creep concerns in these methods [12]. The last method commonly employed to map pressure forces is capacitive technology. A dielectric material is placed between two electrodes, and the deformation of this material changes the capacitance [12]. In fact a recent study by Lemerle et al. created and designed a capacitance-based technology for mapping pressure forces in hand gripping, which we may try and replicate for our testing.

The next method commonly employed in testing ergonomics is the recording of relevant muscle activity via electromyography (EMG). As we anticipate that the muscles most stressed during drug application will be those in the forearm, we will plan on recording EMG data for these muscles over the course of drug application. We will get several test subjects, attach electrodes

to various points on the forearm, and record this data over the course of drug application. We will likely calculate the average force exerted by these muscles over the course of application. We can then compare data for each separate device as recorded by EMG and try to ascertain which device is the 'easiest' to use.

The last way which we intend to test our device is through a subjective questionnaire given to clinicians after having used our devices, as well as the current devices, for the given application period. A sample of the types of questions which we will attempt to address is listed below.

1. Which device did you find provided the easiest means of drug application?
    a. Device A
    b. Device B
    c. Device C
2. Which device did you find provided the easiest means of ampoule fracture?
    a. Device A
    b. Device B
    c. Device C
3. If given a choice, overall which drug application device would you prefer to use on a regular basis?
    a. Device A
    b. Device B
    c. Device C

These are only a few examples of the types of questions which we will seek to have answered. We will plan on having a survey which takes approximately 5-10 minutes to answer, containing 15-20 questions.

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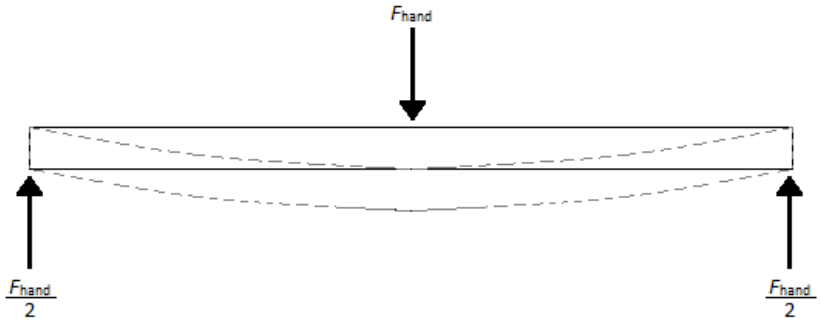
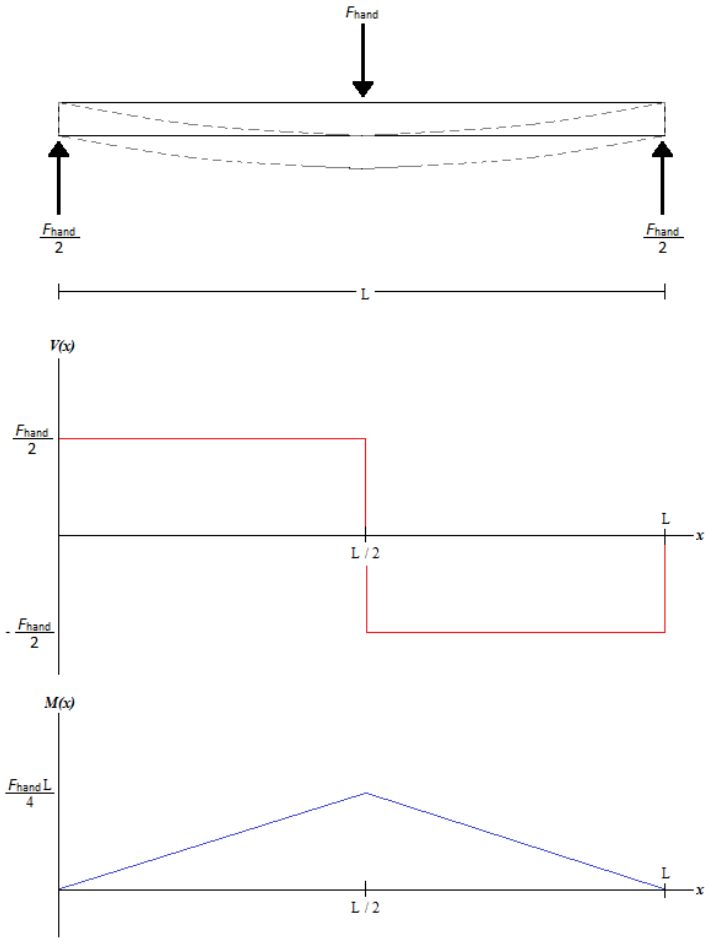
Appendix

A. Calculation – Three-Point Bending

Consider a high-density polyethylene (HDPE) beam undergoing three-point bending:



Under such a loading, the maximum moment occurs at the center of the beam, as can be deduced from the shear and bending-moment diagrams:



$$M_{\max} = M\left(x = L/2\right) = \frac{F_{\text{hand}} L}{4}$$

The bending stress parallel to the axis of the beam can be determined as follows (where $M(x)$ is the bending moment at a position *x* along the beam, y is the distance from the neutral axis of the cross-section of the beam, and I is the moment of inertia of the beam):

$$\sigma_{xx}(x, y) = \frac{M(x) \cdot y}{I}$$

For a hollow-cylinder HDPE beam with outer diameter (d_o) and inner diameter (d_i), the moment of inertia for the cross-section is given as follows:

$$I = \frac{\pi}{64}\left(d_o^4 - d_i^4\right)$$

Thus, the maximum bending stress is (with $y_{\max} = r_o = d_o/2$, the furthest distance from the neutral axis):

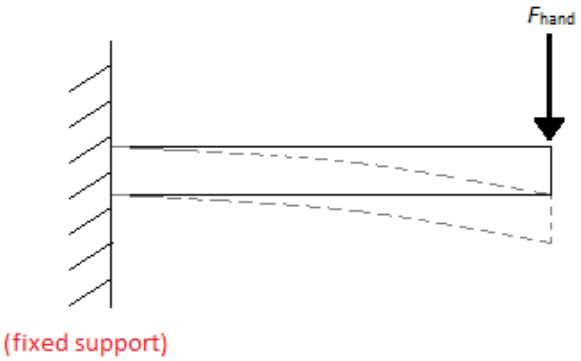
$$\sigma_{xx_{\max}} = \frac{M_{\max} \cdot y_{\max}}{I} = \frac{\left(\frac{F_{\text{hand}} L}{4}\right)\left(\frac{d_o}{2}\right)}{\frac{\pi}{64}\left(d_o^4 - d_i^4\right)} = \frac{8 \cdot F_{\text{hand}} \cdot L \cdot d_o}{\pi \cdot \left(d_o^4 - d_i^4\right)}$$

HDPE has a yield strength of 28 MPa [6]. It is also known that such a particular beam has outer and inner diameters of 18 mm and 15 mm (respectively) and a length of 120 mm. Given this information, the critical force of application to yield the HDPE beam can be determined as:

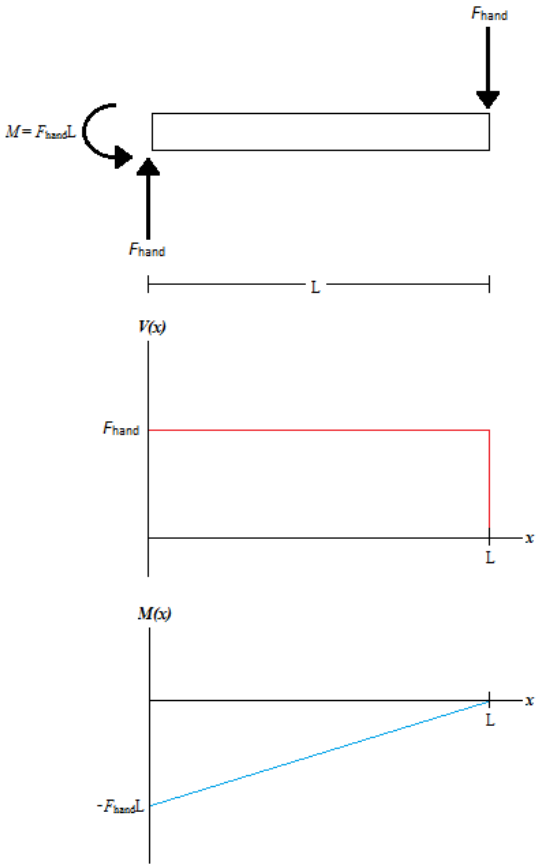
$$F_{\text{yield}} = \frac{\pi \cdot \sigma_{\text{yield}} \cdot \left(d_o^4 - d_i^4\right)}{8 \cdot L \cdot d_o} = \frac{\pi \cdot \left(28 \times 10^6 \ \text{N/m}^2\right) \cdot \left[(0.018\ \text{m})^4 - (0.015\ \text{m})^4\right]}{8 \cdot (0.120\ \text{m}) \cdot (0.018\ \text{m})} \approx 277\ \text{N}\ \left(62.2\ \text{lb}_f\right)$$

B. Calculation – Cantilevered Bending

Consider a high-density polyethylene (HDPE) beam undergoing cantilevered bending:



Under such a loading, the maximum moment occurs at the base of the beam, as can be deduced from the shear and bending-moment diagrams:



$$M_{\max} = M\left(x=0\right) = -F_{\text{hand}}L$$

The bending stress parallel to the axis of the beam can be determined as follows (where $M\left(x\right)$ is the bending moment at a position *x* along the beam, y is the distance from the neutral axis of the cross-section of the beam, and I is the moment of inertia of the beam):

$$\sigma_{xx}\left(x,y\right) = \frac{M\left(x\right)\cdot y}{I}$$

For a hollow-cylinder HDPE beam with outer diameter (d_o) and inner diameter (d_i), the moment of inertia for the cross-section is given as follows:

$$I = \frac{\pi}{64}\left(d_o^4 - d_i^4\right)$$

Thus, the maximum bending stress is (with $y_{\max} = r_o = {}^{d_o}\!/_{2}$, the furthest distance from the neutral axis):

$$\sigma_{xx_{\max}} = \frac{M_{\max}\cdot y_{\max}}{I} = \frac{\left(F_{\text{hand}}L\right)\left(\frac{d_o}{2}\right)}{\frac{\pi}{64}\left(d_o^4 - d_i^4\right)} = \frac{32\cdot F_{\text{hand}}\cdot L\cdot d_o}{\pi\cdot\left(d_o^4 - d_i^4\right)}$$

HDPE has a yield strength of 28 MPa [6]. It is also known that such a particular beam has outer and inner diameters of 18 mm and 15 mm (respectively) and a length of 120 mm. Given this information, the critical force of application to yield the HDPE beam can be determined as:

$$F_{\text{yield}} = \frac{\pi\cdot\sigma_{\text{yield}}\cdot\left(d_o^4 - d_i^4\right)}{32\cdot L\cdot d_o} = \frac{\pi\cdot\left(28\times10^6\ {}^{\text{N}}\!/_{\text{m}^2}\right)\cdot\left[\left(0.018\text{ m}\right)^4 - \left(0.015\text{ m}\right)^4\right]}{32\cdot\left(0.120\text{ m}\right)\cdot\left(0.018\text{ m}\right)} \approx 69.2\text{ N}\ \left(15.5\text{ lb}_f\right)$$

C. Product Design Specifications (PDS)

Project: Skin Applicator

Team Members: Ben Fleming, Beom Kang Huh, Adam Pala

I. Function

The goal of this project is to design a disposable applicator for a topical, drug-containing solution. The device should be able to apply the topical solution to a cancer patient receiving radiotherapy. There is a current study under way which could make use of such a device.

II. Client Requirements

- The device must be able to apply 8.0 mL of the solution per use
- The device should release the drug-containing solution in a controlled, consistent manner
- The device should be able to apply the solution to approximately 225 cm^2 of skin
- The device should be disposable after one use
- The device should function as well as or better than current devices on the market
- The device should be similar in size to the ChloraPrep™ 10.5 mL applicator
- The applicator should be slightly abrasive to increase skin porosity for drug absorption

III. Design Requirements

1. Physical and Operational Characteristics

a) Performance Requirements: The device should be able to successfully apply the solution in a controlled, consistent manner. The device should be disposable after one use.

b) Safety: The device must not harm the patient to which the drug is administered, nor should it harm the individual administering the drug. If a glass ampoule is incorporated into the design, there should be no significant possibility of injury from glass shards. The safety of the device is irrespective of the safety of the drug which it is intended to administer (which is determined by the clinician and other regulatory entities).

c) Accuracy and Reliability: The device should be able to administer 10 ± 0.5 mL of solution (i.e. within 5% of the desired value). The device should be able to administer the drug to a 250 cm^2 area of skin (or larger).

d) Life in Service: The device is intended for a single use.

e) Shelf Life: The shelf life is dependent on the half-life of the drug (norepinephrine).

f) Operating Environment: The device should be able to function correctly in a typical clinical setting (i.e. 25 °C, 1 atm). The device should be able to withstand stresses and strains imposed by the individual using the applicator.

g) Ergonomics: The device should be handheld, portable, and easily used by a single individual with minimal effort.

h) Size: The device will consist of two main components: the handle (approximately 15 cm x 4 cm x 2 cm) and the foam applicator (approximately 8 cm x 4 cm x 2 cm).

i) Weight: The device will weigh approximately 100 to 200 g.

j) **Materials**: The device will consist mainly of glass (i.e. borosilicate or soda lime glass ampoule) and polymer (i.e. plastics, foam).
k) **Aesthetics**: The device should be aesthetically pleasing and should not induce excessive discomfort or fear in the patient. The plastic handle should be reasonably transparent to monitor the solution. The foam should be slightly abrasive in texture to promote application of the solution onto the skin.

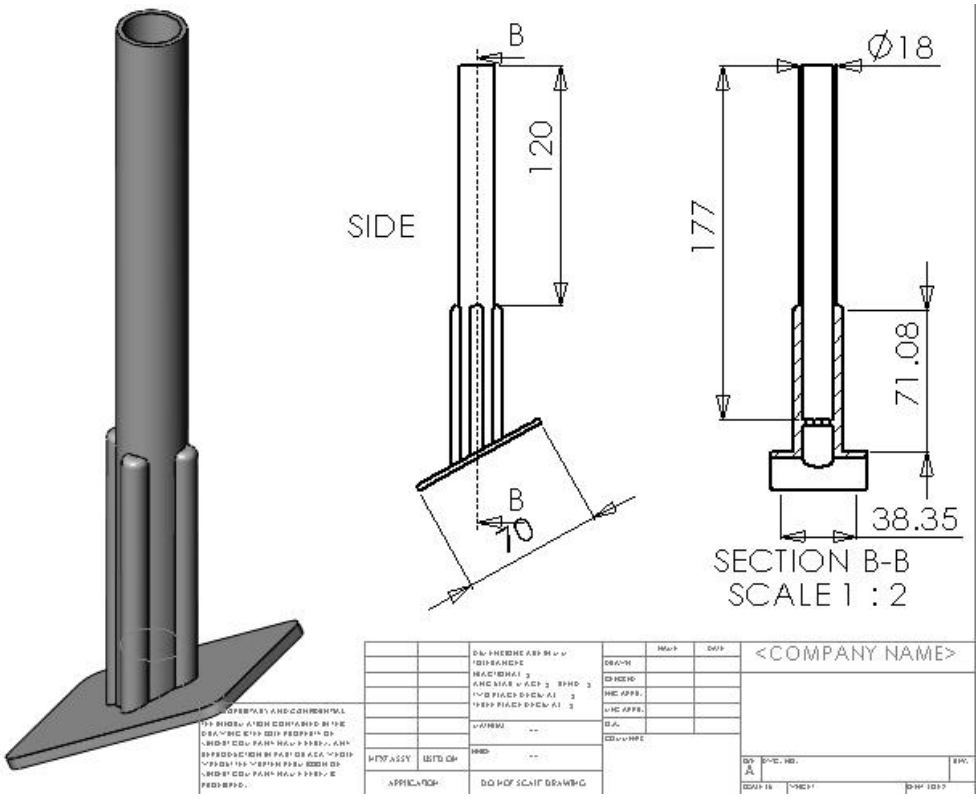
2. Production Characteristics

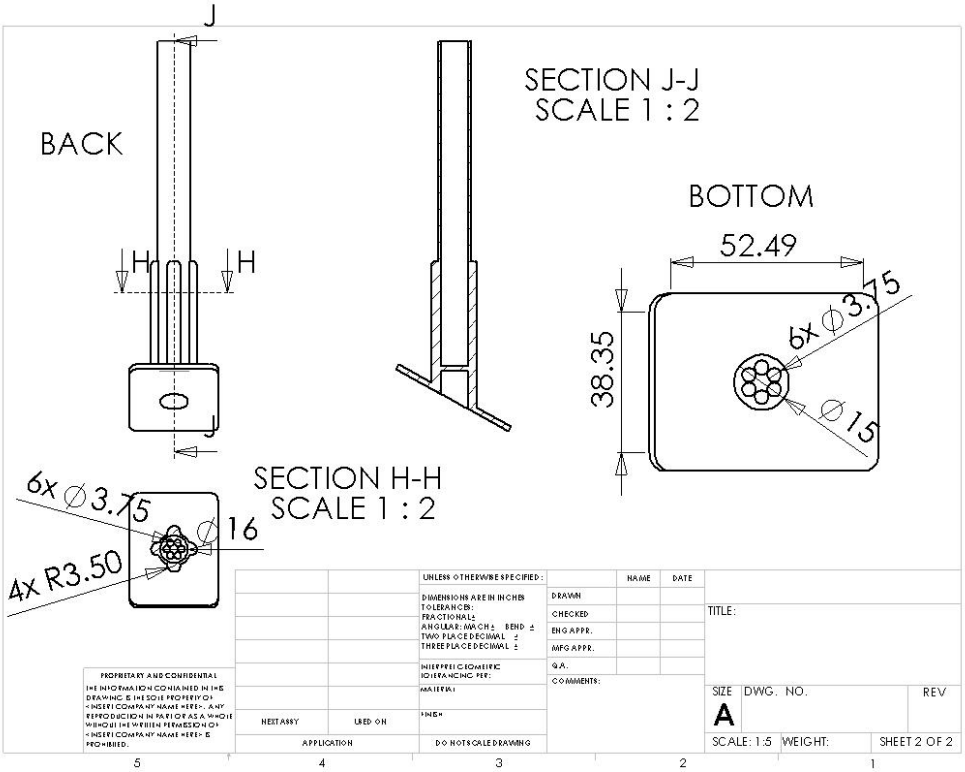
l) **Quantity:** Initially, one functional, disposable device is desired. If the device is successful, then larger scale production may be desired.
m) **Target Product Cost:** The initial prototype should cost under $200-$300 to produce. If the prototype is successful, then a lower per-unit cost would be desired.

3. Miscellaneous

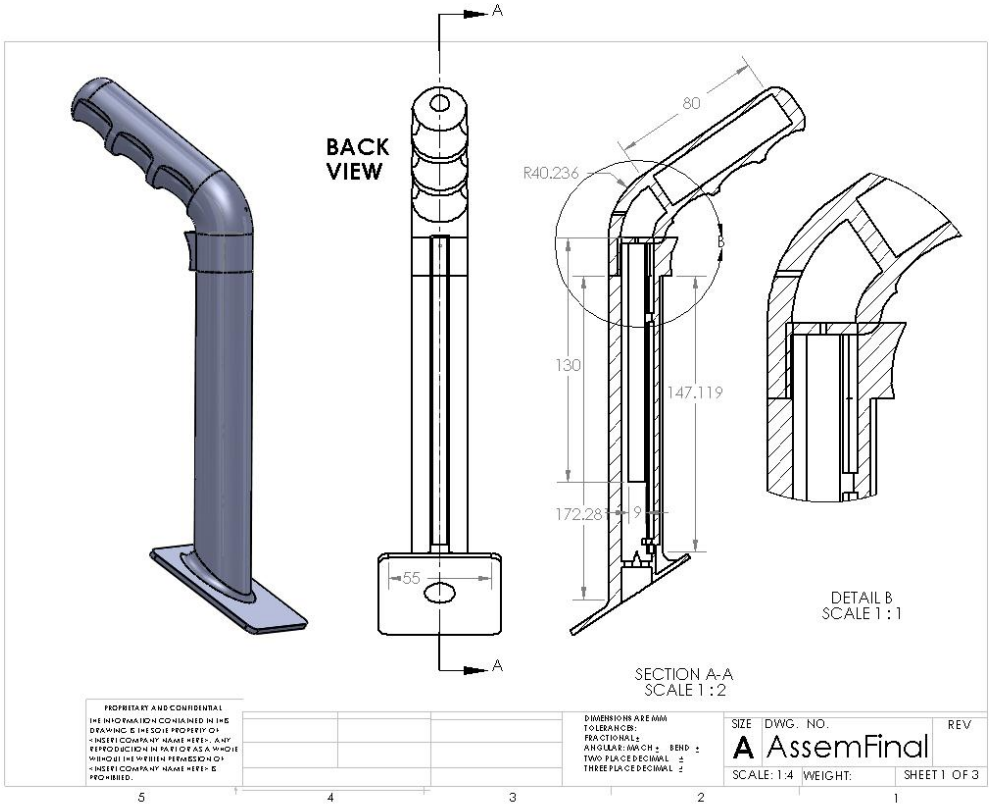
n) **Standards and Specifications:** Currently there are no significant concerns with University, state, national, or international standards, as the device will initially be tested with the drug vehicle (i.e. a 70:30 solution of ethanol/water).
o) **Customers:** The device should be appealing to and easy to use by clinicians applying radiotherapy to cancer patients.
p) **Patient-Related Concerns:** T he cancer patients should not be concerned or harmed by application of the solution by using the device.
q) **Competition**: There are several similar proprietary devices on the market (e.g. CareFusion ChloraPrep™ applicators).

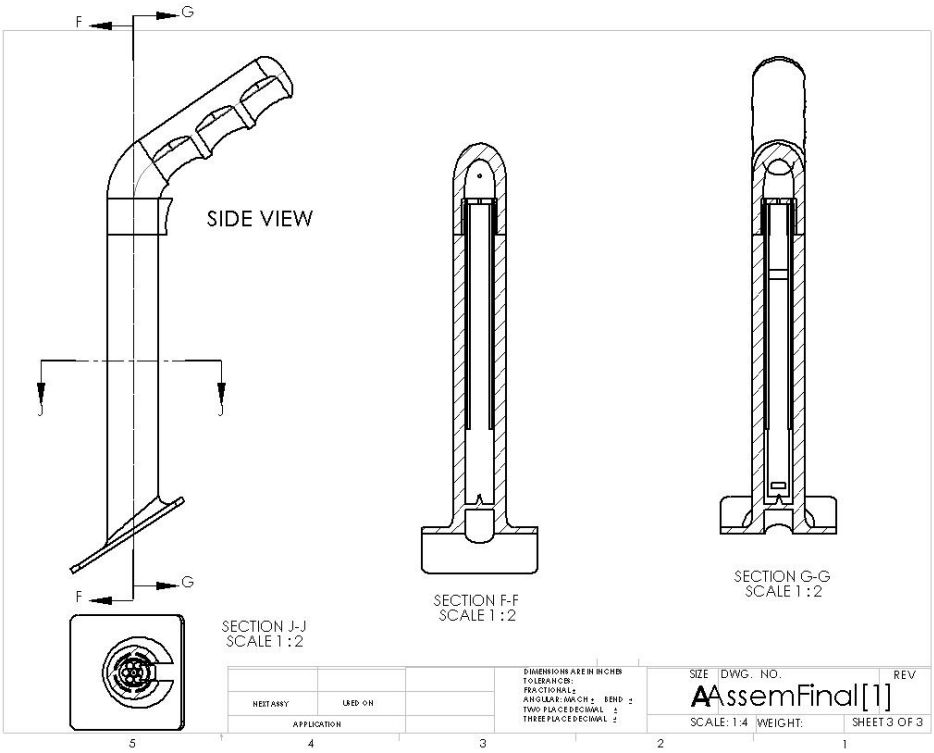
D. CAD drawings of the 1st design





E. CAD drawings of the 2nd design





F. Budget Analysis

Our budge for this semester was fairly minimal. Our only expense was rapid prototyping of the two design alternatives which we came up with, and totaled $75.96.