Final Design Report:
Fetal Movement Simulator

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Abstract
The objective of this project is to develop a wearable fetal movement simulator to help test and verify the fetal movement detection solution from the Sienko Research Group at the University of Michigan. Our simulator should be able to mimic physical attributes of fetal activity in a wearable artificial abdomen to produce similar surface level displacements as actual fetal movements. It is intended to evaluate the accuracy of sensors in their ability to detect fetal movement and distinguish fetal movements from maternal ones with the use of inertial measurement units (IMUs) from the research group.

Executive Summary
In the United States, 24,000 stillbirths occur every year. About one in every 160 pregnancies is a stillbirth. While there were many advancements in obstetrics in the mid-twentieth century, the rate of stillbirth has remained immutable in the past few decades. In addition, stillbirths disproportionately affect minorities compared to white Americans. Current medical antenatal care such as ultrasounds and fetal heart rate monitoring only provide doctors with a snapshot of the fetus' health. Fetal movement is an indicator of fetal health, but it can frequently vary; movement may not occur during a check-up. It is suggested that mothers do not track fetal movement because it can cause unwarranted anxiety. A more complete picture of the fetus' health from fetal movement sensors could prove invaluable for healthcare workers in making important medical decisions. Ideally, this could lead to a decreased rate of stillbirth. Researchers at the Sienko Research Group at the University of Michigan are investigating a wearable solution that would monitor fetal movement and heart rate for longer durations. A multi-nodal system compares angular rotation of the surface of the mother's abdomen caused by fetal movement to the mother's chest in order to differentiate movements. We are tasked with developing a wearable fetal movement simulator to test and verify the wearable fetal health monitor. A fetal simulator that is wearable is crucial to confirm the fetal health monitor's ability to differentiate fetal and maternal movement.

We identified the primary requirements for our device from interviewing our stakeholders. We then crafted engineering specifications to fulfill the requirements. Our project's core requirements include being wearable, mimicking fetal movements, mimicking skin level displacements on the mother's abdomen, and operating continuously for 45 minutes. The research was completed to define these specifications to imitate a fetus. We decided to mimic the mechanical properties of the tissue by averaging the properties of the uterine wall, abdominal muscle, and adipose tissue into one homogeneous material to improve the manufacturability of the device. We plan on generating peak displacements on the surface of our phantom in the range of 0.5mm to 13mm for 0.5-3.5 seconds to mimic the full range of fetal movement.

Applying concept generation using divergent and convergent thinking techniques enables our team to converge down to a single alpha design. Preliminary sketching and CAD outline our plan for the simulator. The final CAD and electrical design has been expanded upon. The process for engineering analysis has also been outlined.

We have developed a functional fetal kick simulator prototype. We tested and verified our device based on the engineering specifications. We also critiqued our device and developed recommendations for improvement that our sponsors can implement in the future.

Problem Description and Background

Social Context

There are on average 24,000 babies stillborn in the U.S. every year, about one stillbirth out of every 160 pregnancies [1]. A stillbirth is a loss of a fetus at 20 weeks of pregnancy or later up to and including delivery [2]. Miscarriage is a loss of a fetus before the 20th week [2]. After the 1940s, improvements in maternity care significantly decreased the occurrence of stillbirth. Although in recent years, the rate of stillbirth has been slow to decrease [3]. See figure 1 below for stillbirth data from 1990-2017 in the United States. In the 27 years shown, despite advancements in medical technology, the rate of early fetal stillbirths has remained approximately constant, and the rate of late fetal stillbirths has decreased but not significantly. There are limitations to current methods for monitoring fetal health. Most of the evaluation of fetal health is sensory from a doctor or the mother, which can be imprecise. Ultrasounds, a commonly used technology to assess fetal health, are limited in the duration they can be used, about 30 minutes [4]. This limitation is because ultrasounds expose the mother and fetus to radiation. Furthermore, ultrasounds must be operated by a trained technician and the equipment is expensive. Active monitoring of the fetus with affordable sensors over longer durations has a promising potential to reduce the rate of stillbirth.

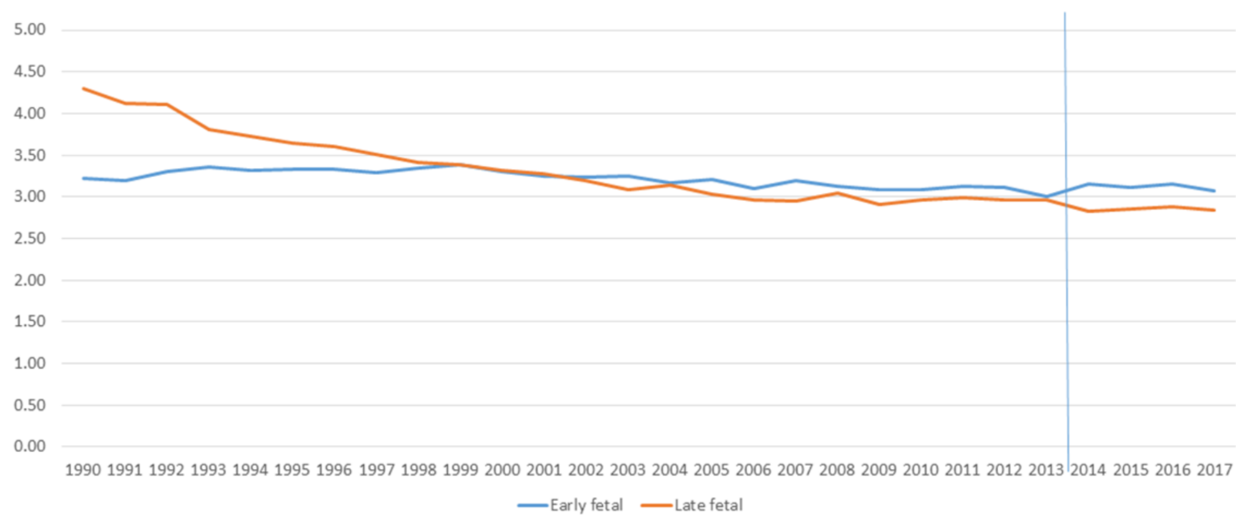


Figure 1. The graphic shown displays the rate of stillbirth per 1,000 births from 1990 to 2017 in the US. The blue line represents "Early Fetal" stillbirths which are between 20 and 27 weeks of pregnancy. The red line represents "Late Fetal" stillbirths which are 28+ weeks of pregnancy. Both of these data sets have largely flatlined in recent years [1].

The rate of stillbirth in the United States affects different races and ethnicities at different rates. As depicted in figure 2 below, the rate of stillbirth for non-Hispanic black women is the highest with 10.32 stillbirths per 1,000 births [1]. This is more than double that of non-Hispanic white women or Hispanic women. American Indian or Alaskan Native mothers also have a significantly higher rate of stillbirth than the total rate of stillbirth [1]. Although stillbirths affect all races in the population, it is important to consider those who are at an increased risk, and when a solution is available we as engineers must ensure the most affected parts of the population can have access to the technology.

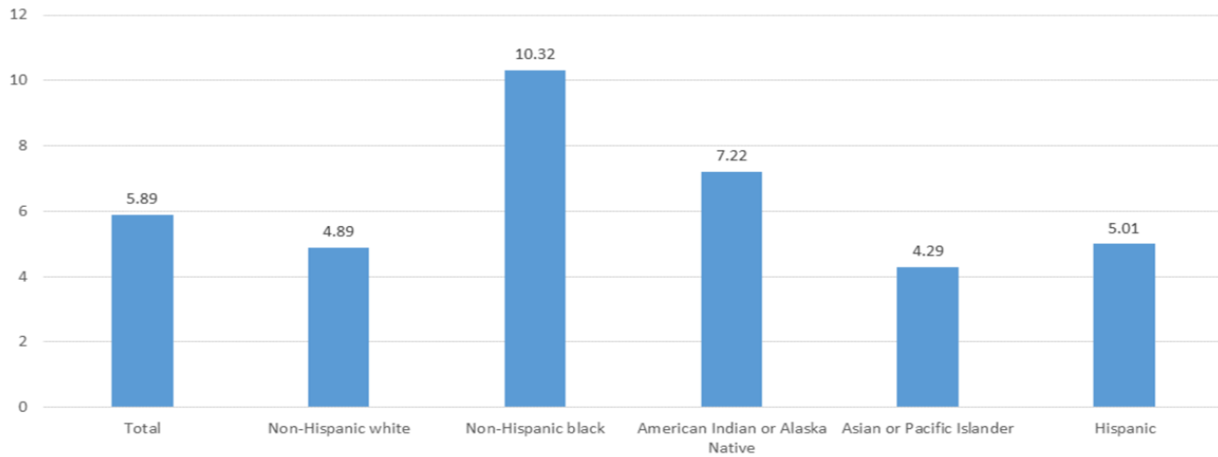


Figure 2. Data from the CDC depicting the rate of stillbirth by ethnic group in the US [1]

Scientific Background

A diagram of a fetus in utero can be seen below in figure 3 [5]. A fetus is an unborn baby from the 8th week after fertilization until birth. For our project, we are focusing on the third trimester of pregnancy, which is from the 28th week of pregnancy until delivery. We are focussing on key elements of the anatomy to replicate in our project, in particular, the amniotic sac, uterus, abdomen muscle wall, and adipose tissue (AT) [5]. The amniotic sac is a thin-walled sac that surrounds the fetus during pregnancy. It is filled with amniotic fluid that is produced by the fetus throughout pregnancy. The volume of amniotic fluid increases throughout gestation. At the 28th week, there is approximately 770 mL of amniotic fluid in the amniotic sac. The uterus is the pear-shaped organ in which the fetus develops after fertilization [5]. The abdomen muscle wall is an additional layer of the mother's anatomy between the fetus and the skin. The skin, and specifically the adipose tissue, known more commonly as body fat, are the final layer.

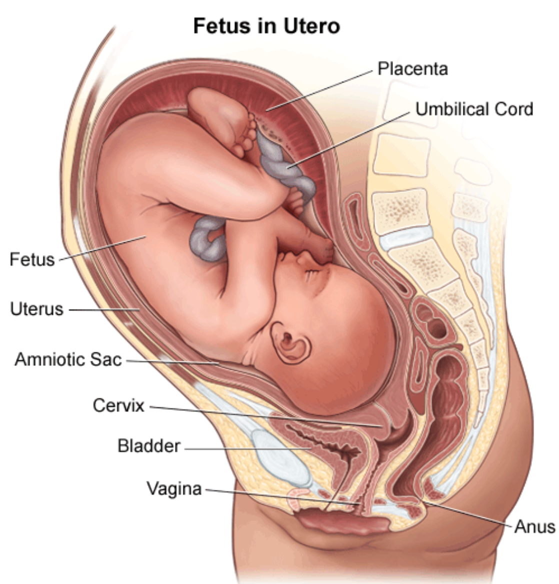


Figure 3. Diagram of the anatomy of the mother with fetus in utero [5]

Current Fetal Health Monitoring

A primary indicator of fetal health and well-being is fetal movement. Fetal movement encompasses a variety of actions and motions that a fetus can perform [3]. Kicking, rolling, head movements, and stretching are some examples [3]. As we learned from our sponsor, Dr. Carrie Bell, at the start of the third-trimester fetal movements average five movements per hour or ten movements per two hours for a healthy fetus. However, the frequency of fetal movements is highly variable, potentially missing movements in the timeframe of clinic visits since the resting cycle of the fetus is about 45 minutes. Mothers are instructed to avoid counting fetal movements because it can lead to unwarranted anxiety since the frequency is so inconstant. However, a lack of fetal movement can be an indication for doctors to take medical action. Doppler Shift Cardiotocography (CTG) and Ultrasonography (US) are two current methods for monitoring fetuses. A CTG is a recording of fetal heart rate via an ultrasound transducer, and an ultrasound creates an image of the fetus using sound waves [6]. While these tests are useful, they are limited in the duration they can be used for, which is only about 30 minutes, as the tests expose the mother and fetus to radiation [4]. The tests also must be operated by a trained technician in a medical facility. The sporadic fetal antenatal checkups provide only a limited snapshot of the fetus' health. The technologies used in modern checkups have existed since the middle of the 20th century, and despite increased use and improvements of these methods, the rate of stillbirth has remained fairly constant. A wearable fetal movement monitor used for longer durations of time could provide a more in-depth view of fetal health.

Current Fetal Monitoring Solutions

There are currently a few solutions for wearable fetal health sensors that are being explored. One of the solutions came from a past ME 450 team that had utilized four iNewtons, an inertial measurement unit (IMU) that can track acceleration and surface rotation. The surface rotation was used to decipher fetal movement from mothers based on comparing values from the sensors at different locations. The iNewtons locations can be seen in figure 4, on the next page, to be placed in a rectangle around a pregnant abdomen. There are other methods that use alternate vibrations sensors such as acoustic sensors and piezoelectric diaphragms [7]. Some of these devices include Novii, Avalon, and Nemo [8, 9, 10]. The goal of all of these wearable technologies is to observe fetal movement over long periods of time. The increased use of fetal monitoring devices would lead to gaining access to the fetus' health. Therefore, important medical decisions could be made with a deeper understanding of the fetus' health.

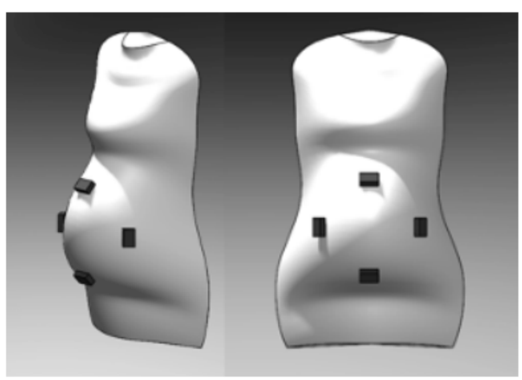


Figure 4. Design to sense and decipher maternal and fetal movements using iNewtons from a previous ME450 project team

The fetal kick simulator that we will be designing will be used to test and verify the current solution coming from the Sienko Research Group at the University of Michigan. OPAL wearable sensors are used to detect fetal and maternal movement. OPAL is an IMU that contains both acceleration and rotation sensors [11]. This sensor can be seen in figure 5 below. One of the sensors is placed on the mother's chest and the other three sensors are placed in a triangle on the mother's abdomen. The purpose of the chest sensor is to receive data from the mother in order to differentiate their movement from the fetus'. The gyroscope will detect rotation from the chest of the mother and the skin displacement of the abdomen.

Figure 5. OPAL wearable sensor used in the Sienko Research Group's solution to detect fetal movement

There is a current need for the simulator to help verify the current wearable sensor solution. A wearable fetal movement simulator is necessary to test the ability of the sensor solution to decipher fetal movement from the mother's. There are a few reasons for this. It is costly to test on human subjects. Fetal movement cannot be reliably controlled. As we learned from Dr. Bell, fetuses can have 45-minute rest cycles. Therefore, fetal movements are highly variable and potentially infrequent. The movements that do occur would also be nearly impossible to exactly replicate in later tests. The wearable simulator will also make it easier to change the current sensor placement or number. This will allow greater concept exploration on fetal movement detection from the Sienko Research Group. As the simulator will be solely used in a laboratory setting, there is limited environmental impact on the resources, manufacturing, and disposal of a single device.

Broader Context

The creation of our device will be instrumental in the beginning stages of testing a wearable fetal health device. Our device alone, will not have a direct affect on the population at large, being a one off testing device. It could have a large impact indirectly, if it benefits the creation of the fetal health monitoring wearable device. The fetal health monitoring device could lead to significant reduction in miscarrage and stillbirth rates. Our product would not greatly benefit from being on the global market. Our device has a niche use in a laboratory setting, testing fetal health monitoring devices. If the fetal health monitoring device that our device will test comes to market, there could be increased access to prenatal care for marginalized communities. The monitoring device has the possibility of reducing the racial divide in miscarrage and stillbirth rates. There is limited economic impact from the manufacturing, use, and displosical of our device and technology because it is a one off device.

Stakeholders

Our primary stakeholders are the faculty within the Sienko Research group. This includes a doctoral student, Lucy Spicher, directly testing with the fetal health monitor, Dr. Carrie Bell an OBGYN within the group, and Prof. Noel Perkins. We rely on these stakeholders for their needs in the project and information about pregnancy. Another primary stakeholder would be the test subjects that have to wear the simulator.

Secondary stakeholders include pregnant individuals who could wear the fetal health monitor. This group of people would greatly benefit from an accurate fetal health monitor. It is vital that the simulator accurately test this device in order for it to be reliable for these women. Ethically, we need to create a device that is able to test the true accuracy of the fetal monitor. The accuracy of detecting movements is very important to both the stakeholders and our team. Tertiary stakeholders could include the CDC and the FDA for approving the wearable fetal monitor. The roles of our stakeholders are depicted in the ecosystem map shown below in Figure 6.

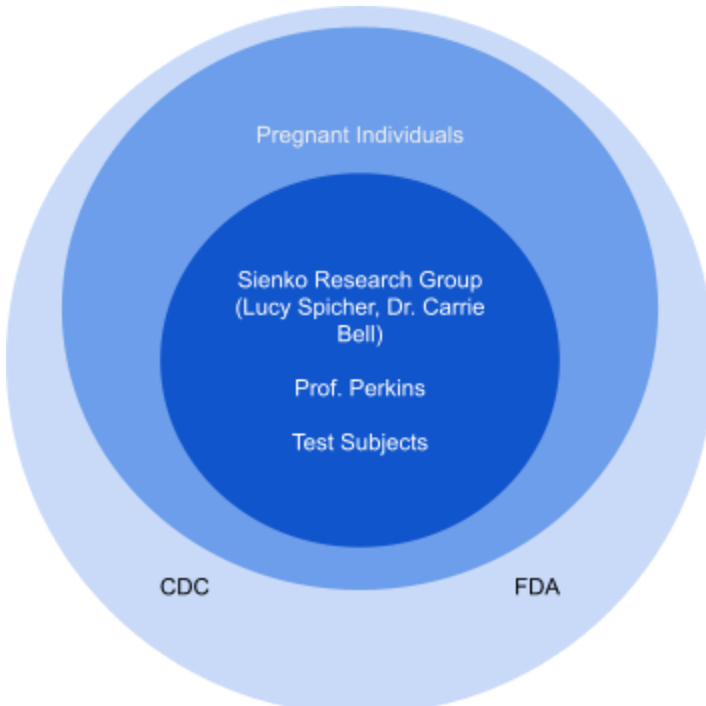
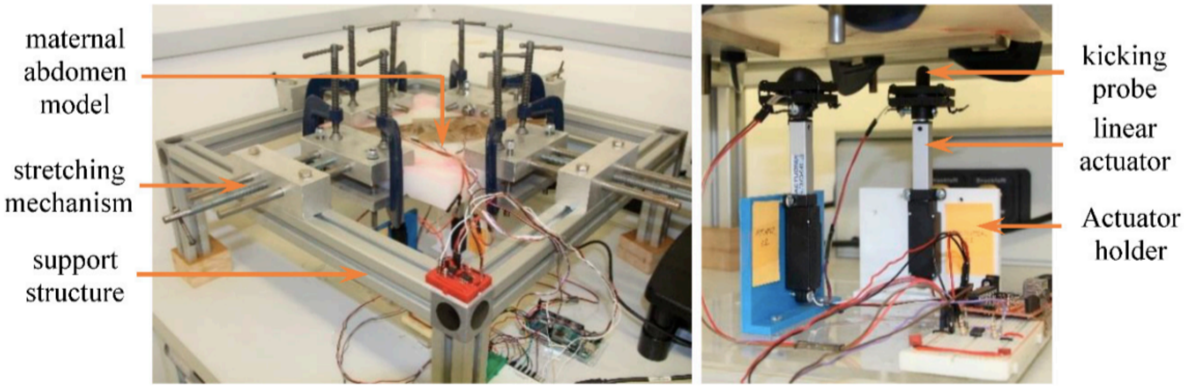


Figure 6. Ecosystem map for our project. The dark blue center are primary stakeholders, the medium blue circle are secondary stakeholders, and the light blue outer circle are tertiary stakeholders.

Existing Fetal Kick Simulator

Our team has been able to find one workbench fetal kick simulator from the research at Imperial College London [7]. The workbench simulator can be seen in figure 6 below. This simulator uses linear actuators to mimic fetal kicks. It uses a single-layer membrane to mimic the mother's abdomen. The single-layer material was chosen by solving for tensile strength and thickness of the different layers in order to make one layer. This research document provided a wealth of information for our project. While very different in its goal, being a workbench simulator, this source proved valuable in finding many of the forces and material properties that can be used in our project.



(a) (b)

Figure 7. Fetal movement simulator from the Imperial College London. **(a)** model of the maternal abdomen with stretched single layer synthetic tissue, **(b)** linear fetal kick similar beneath the synthetic abdomen [7]

Objective

The objective of the wearable fetal movement simulator is to help test and verify the fetal movement detection solution. Our simulator should be able to simulate fetal activity in an abdomen to produce similar surface-level displacements from actual fetal movement impact forces. It is intended to evaluate the accuracy in detecting fetal movement and deciphering movements between mother and fetus with the use of OPAL's angular rate sensors. Our device will be used in a laboratory setting.

Stakeholder Requirements and Engineering Specifications

Following initial background research, our team gathered stakeholder requirements and translated them into engineering specifications. Our team first set up meetings with each of our stakeholders to formulate what requirements they have. We focused on asking engaging questions based on our prior research to not only determine what requirements they had but also the importance of each. Table 1, on the next page, shows our stakeholder requirements and specifications. An asterisk "*" indicates that the requirement/specification came directly from our sponsor. Priority was ranked I - III with "I" indicating the highest priority.

Table 1. The requirements, specifications, and priority of specifications for the fetal movement simulator

Requirement	Specification	Priority
Wearable device	Total weight <= 30 lbs [12]	**II**
	At least 2 straps to mount onto the torso*	**II**
Mimics fetal movement	Mechanism can rotate 100 about its vertical axis*	**II**
	Impact Force can be applied ± 20 degrees from horizontal*	**I**
	Generate a fetal heart rate at 110-160 BPM [13]	**III**
Mimics the displacement profile on the surface of the skin	Generate surface displacements from 0.5mm - 13mm [7]	**I**
	25% accuracy of peak amplitude displacement*	**I**
	Replicate impact durations of 0.5-3s [7][14]	**I**
Operates continuously	Can operate continuously for at least 45 min*	**II**
Fetal movements are controllable	Reference point to "zero" all DOF*	**II**
	Movements are programable*	**I**
	Impact force is centered within 2cm of the desired location*	**II**

Justifying Requirements and Specifications

Wearable device: One key requirement from our stakeholders was that the device should be wearable to see if fetal movement could be isolated from maternal movement. We wanted to ensure that the device wouldn't be too heavy where it impedes the user's movement. We determined that the device should not weigh more than 30 pounds based on the National Institute for Occupational Safety and Health [12]. Our stakeholders also recommended at least two mounting locations to secure the device.

Mimics fetal movement: One of our stakeholders Dr. Bell requested that our device mimic fetal movement specifically, kicking, punching, rolling, and heart rate. To mimic the kicking and punching of the uterine wall our team determined that our device should be able to generate a variable impact force 20 degrees from the horizontal axis. To mimic the rolling of the fetus we decided that the internal mechanism should rotate 100 degrees about its vertical axis. To mimic fetal heart rate we are targeting a pulse generating device that can reproduce frequencies of 110-160 beats per minute [13].

Mimics the displacement profile on the surface of the skin: The movement generated by fetal movements will be detected by inertial measurement units (IMU's) via displacements on the surface of the skin. From discussing with Prof. Perkins and Lucy Spicher it was essential that we closely mimic these surface displacements. Based on our research we concluded that mimicking the exact profile of the displacements would be overly complex and are solely focusing on the peak displacement amplitude. From our research, we should be able to generate surface displacements from 0.5mm - 13.0mm with ± 25% accuracy [15]. Furthermore, these impacts should endure between 0.5 and 3.5 seconds [7].

Operates continuously: The device should be able to operate continuously for at least 45 minutes based on Dr. Bell's request that the device could operate for the typical duration of an ultrasound. We added a safety factor of 1.5 to the typical 30 minute ultrasound time [4].

Fetal movements are controllable: To properly validate the use of IMU's our stakeholders requested that the location of the impact force be controllable alongside the magnitude of the force. To accomplish this we decided that the accuracy of the location of the force should be within 2cm based on advice from Lucy Spicher. To control the magnitude and location of the force we required that the device be programmable and have reference datums to "zero" any degrees of freedom that the device is free to actuate.

Budget: Since our fetal movement simulator will be the only one of its kind, keeping a low budget is more of a goal than a hard requirement. We hope to keep the total budget for the prototype below $400, but if we need to spend more to make the best device our sponsors have voiced that this will be okay.

Recently Removed Specifications: Mechanical properties of the abdominal tissue and amniotic fluid were included to mimic the anatomy of the mother and fetus. These former requirements and specifications can be seen in Appendix A. We have decided to focus on the displacements rather than the anatomical properties.

Problem Analysis

To accurately reproduce fetal movements which could be detected using an IMU sensor, the viscoelastic properties of the tissue and the surface displacements induced from fetal movements must be closely mimicked. During our team's initial research we had to determine which aspects of the fetal movement were paramount to our design and explore potential obstacles that our team may face.

Viscoelastic Properties of Tissue

We explored the option to correctly reproduce the propagation of a pressure gradient through synthetic tissue, the viscoelastic properties of the tissue being closely matched. A viscoelastic model was selected to represent both viscous and elastic characteristics of the tissue when undergoing deformation. The fetal movement will induce a pressure gradient (stress) on the posterior surface of the uterine wall. This gradient will propagate through the tissue and induce a detectable displacement of the anterior abdomen surface. For our model in Appendix A, we focused on the anterior abdomen and the propagation of a pressure gradient through the uterine wall, abdominal muscle, and adipose tissue. The values of the viscoelastic properties for each of these components of the anatomy can be seen in Table 2.

Table 2. Key mechanical properties of the human anatomy

Anatomy	Young's Mod. (kPa)	Viscosity (MPa-s)	Density (kg/m^3)	Thickness (mm)
Adipose Tissue	12	60.6 @25C	916	10 @18BMI 40 @30BMI [16]
Abdominal Wall	21	~0	973	31.3
Uterine Wall	586	0.2	1052	9.1
Amniotic Fluid	-	1.1	1007	**Volume (mL)** 770

While it would be ideal to mimic the mechanical properties of each tissue layer to best model the propagation of a pressure gradient, the overall emphasis was to mimic surface displacements on the epidermis. With this objective in mind, scaling the magnitude or duration of the impact force/displacement on the synthetic posterior uterine wall enables the adjustment of peak displacements on the epidermis surface and relaxes the requirement of mimicking the mechanical properties of the tissue. This enables our synthetic tissue to be composed of one material which greatly decreases the complexity of our design. The calculation of these values can be seen in Appendix A. Table 3 below shows our targeted mechanical properties of a homogeneous synthetic tissue.

Table 3. Key mechanical properties to target in the making of a homogeneous synthetic tissue.

Young's Modulus (kPa)	Viscosity (MPa-s)	Density (kg/m^3)	Thickness (mm)
96	23	962	65.5

Surface Displacements

A previous study investigated the use of an accelerometer to detect the surface displacements induced by fetal movement [7]. The accelerometer unfortunately had a large signal-to-noise ratio (SNR) and was tested on a stationary simulator. The additional material movement would amplify the SNR. The data found in this study can be seen in figure 7 below.

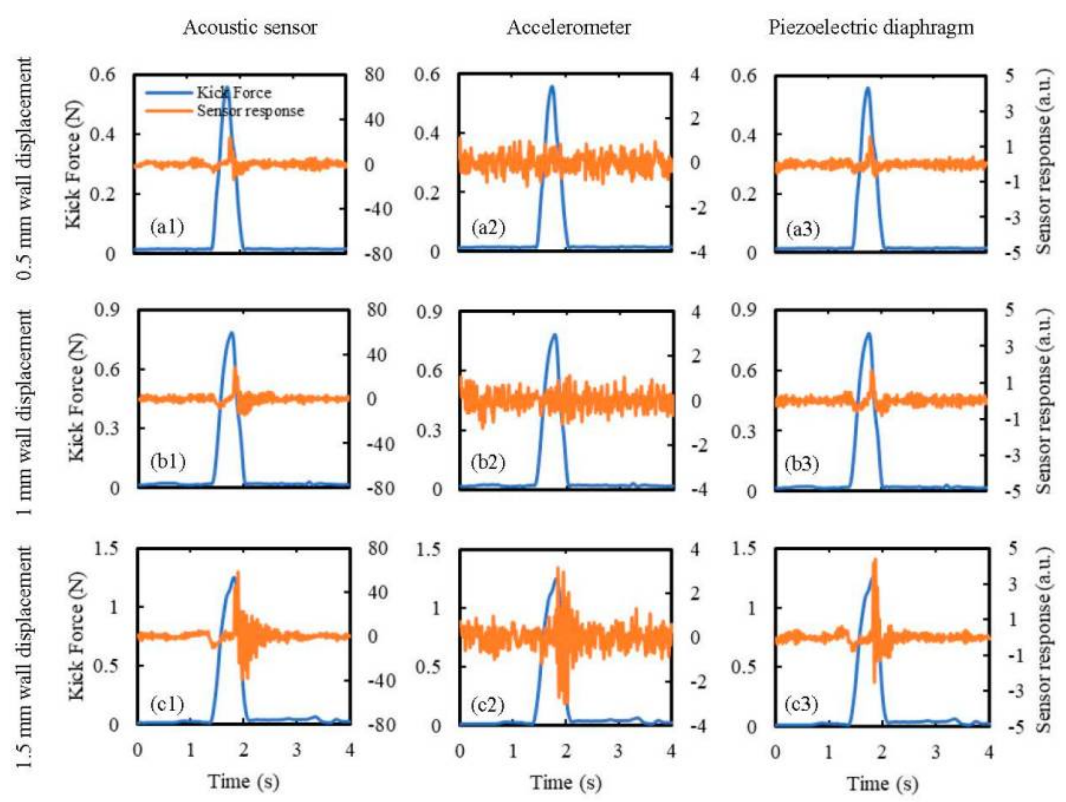


Figure 8. Data from a previous study that compared a kick force to the sensor response. Accelerometer data had a high SNR [7].

A group of university students explored the concept of using differences in angular velocities induced from displacements as a metric for quantifying fetal movement rather than an accelerometer. What these students found was that displacement of the amniotic fluid sac due to fetal movement could be isolated from maternal movement. Figure 8 shows a trial the previous ME 450 team performed where a fetal movement mimicking robot was placed inside a synthetic amniotic fluid sack and commanded to move from 18 to 30 seconds. During this duration peak, angular velocity differences were observed to be much larger on average compared to when the device was off.

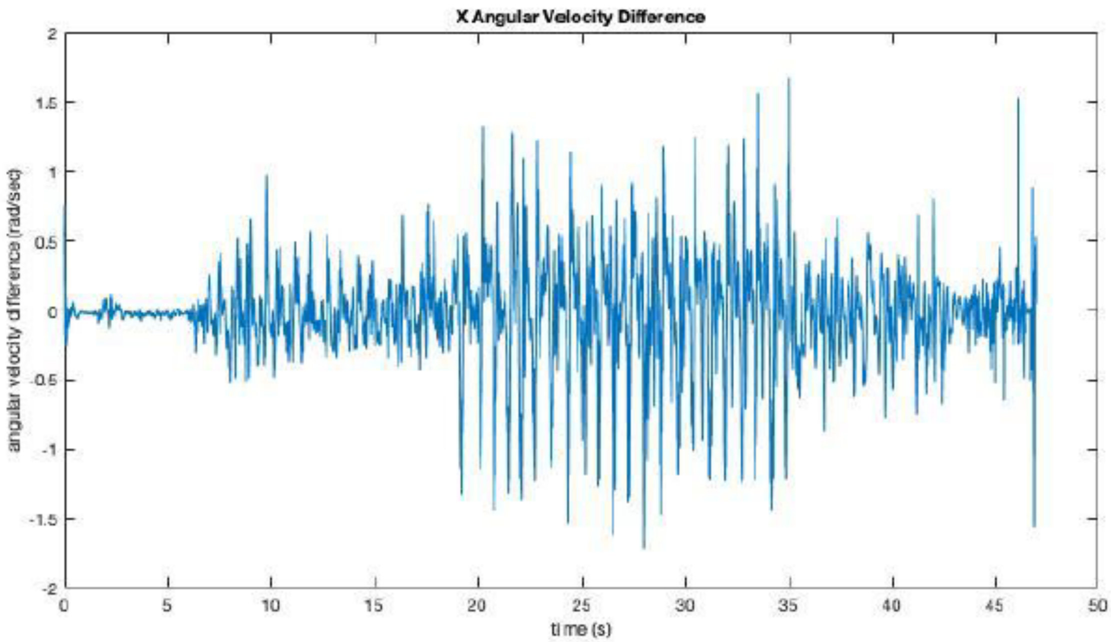


Figure 9. Data from the previous ME 450 team, the X angular velocity difference between two iNewtons, it can be interpreted that in the 20 to 35 second interval, due to the increase in y-axis magnitude, a fetal movement was occurring.

The frequency and pattern of the wavefront of the pressure gradient are going to be extremely difficult to mimic as it depends on the path traveled by the pressure gradient, transmission angle, frequency of the pressure gradient, and location or source of the pressure gradient. Due to this variability, the focus was emphasized on mimicking the magnitude of the surface displacement. By assuming the abdomen is a rigid body all external movements should correspond to similar angular velocity profiles no matter where the IMU is mounted, see figure 9.

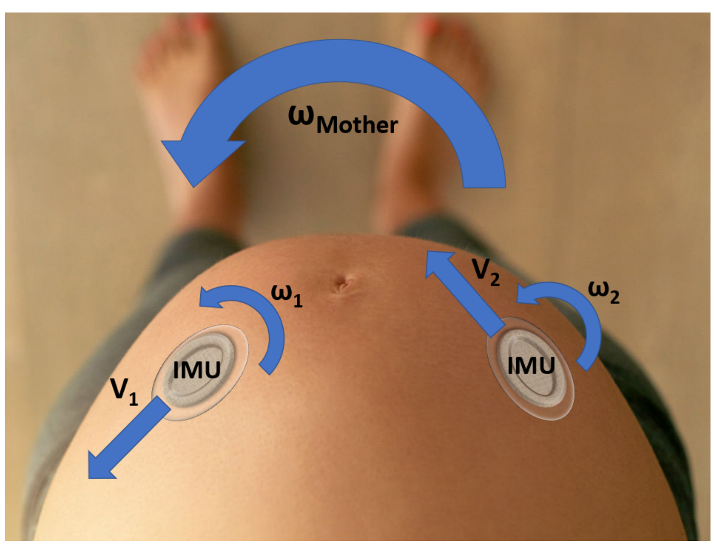


Figure 10. IMU mounting pattern on the mother's abdomen

This assumption would enable us to detect fetal movement by looking at the difference in angular velocity about the vertical axis see figure 10. This difference in angular velocity is induced by epidermis deformation from the propagation of a pressure gradient.

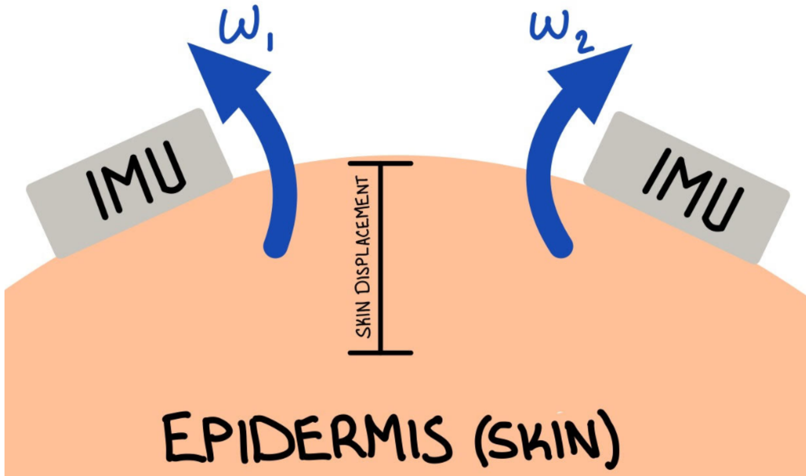


Figure 11. Depiction of how IMU positioning will detect angular rotation on the mother's abdomen from fetal movement

A study conducted to analyze fetal movement using vibration sensors found that a kick for a 30-week old fetus would induce a displacement of 11.52mm ± 1.47mm on the uterine wall [7]. We were unable to locate any data regarding the deformation of the epidermis due to the variability of adipose tissue at different BMIs. To address this problem we will assume that the maximum displacement is 13mm on the surface of the epidermis. These impacts can last between 0.5 and 3.5 seconds [7].

Concept Generation & Selection

We used many concept generation methods in our processes of developing the preliminary design for the fetal movement simulator. First, we created a morphological chart from our requirements and specifications. Every specification was a function that we developed many concepts to fulfill. Being in different regions of the country, we held a group brainstorming session over Zoom to conceptualize as many ideas for each function as we could. We used a google sheet to initially write out our ideas and then sketched the ideas for various concepts in a shared OneNote folder. The platforms we were using allowed us to continue this process asynchronously as well. While brainstorming, we started with our own creativity, but if progress ever stalled or halted we would utilize design heuristics to facilitate the creation of additional ideas. Then we used convergent thinking techniques, specifically a weighted design matrix, to narrow each of the functions to one or two of the best concepts. From there, we combined and iterated on the initial ideas to create our alpha design. We developed sketches and a preliminary CAD model for our design.

Divergent Thinking

In order to generate the best possible preliminary design that fulfilled our requirements and specifications, we used divergent thinking methods. In doing so, we functionally decomposed our design into components. As a team, we had to think as creatively as possible. We used techniques such as developing a morphological chart and design heuristics to foster this. We started with a morphological matrix broken down into categories based on our requirements. The subcategories were broken down further into our design specifications. For each of these specifications, the team brainstormed ideas that would fulfill the function. The basic layout of our matrix can be seen in figure 11 below. We tried to encourage wild ideas to promote the generation of unique concepts. We also tried to build off one another's ideas to create new ones. Once we got stuck we looked to design heuristics to help us think creatively in developing concepts. Once this process was complete we took to sketching to explain our ideas to each other. This also helped us develop more ideas. Finally, we took advice from stakeholders and classmates and added their ideas to our chart. A finalized morphological matrix can be seen in Appendix B.

Requirement	Specification 1	Idea 1.1	Idea 1.2	Idea 1.3	Idea 1.4
	Specification 2	Idea 2.1	Idea 2.2	Idea 2.3	Idea 2.4
	Specification 3	Idea 3.1	Idea 3.2	Idea 3.3	Idea 3.4

Figure 12. The basic layout of morphological chart used in concept generation

Convergent Thinking

After ideating a large number of possible concepts for each engineering specification in our morphological chart, we utilized a weighted decision matrix (WDM). The WDM was a standardized method to evaluate each conceived concept in order to make informed decisions on each function of the design. Each concept was scored from 1 to 10 (1 being the worst and 10 being the best) in five categories: technical feasibility, time to implement, cost, the degree to which it agrees with stakeholder requirements, and how accurately it mimics fetal anatomy. Each category was assigned a weight, from 1 to 5, corresponding to its importance, in regards to design considerations. Our concept most importantly must be practical and fulfill its function, which is the reason technical feasibility was weighted 5. The other most significant consideration for our design is that we can complete the production within our one-semester time frame; therefore, time to implement is also weighted 5. While it is always necessary to consider cost in a design, since our design is intended for laboratory testing, the cost is flexible and not an important consideration. Thus, the cost is weighted 1. We are designing our product for our stakeholders; ergo, stakeholder requirements are an important consideration in our design. However, as engineers we must deviate from stakeholder requests when impractical, hence agreeing with stakeholder requirements has a weight of 4. One of our stakeholders requested that the design accurately replicates fetal anatomy, but the added complexity may be impractical in our time frame. So, the final category, accurately mimics fetal anatomy, was weighted a 3 in our analysis. A portion of the WDM is shown in Figure 12 below to clarify the evaluation process, and the entirety of the WDM is in Appendix C.

Specification	Brainstormed Concept	Technical Feasability	Time to implement	Cost	Agrees with Stakeholder Requirements	Accuratley mimics fetal anatomy	Total Weighted Score
	Weight:	5	5	1	4	3	/180
Generate a fetal heart rate at 110-160 BPM.	Use speaker for FHR	10	10	10	10	4	**162**
	Use animal heart rate toy	10	10	10	4	4	138
	Use research articles wiring diagrams and make one ourselves	6	4	4	9	9	117
	Use our fetal kick mechanism to induce small displacements on the wall that mimic fetal heart rates?	3	3	10	4	4	68

Figure 13. A portion of the weighted design matrix showing the evaluation of the concepts for the specification to generate a fetal heart rate.

The WDM was useful to evaluate each concept and filter out impractical or ineffective concepts, but the highest-scoring concept for each specification was not necessarily the optimal choice. After completing the WDM we, as a team, analyzed the outcomes for each specification and determined if the option with the highest total score was in fact the best choice, or if the optimal design choice was a combination of multiple concepts or an alternate concept alone. Most commonly, the concept or concepts with the highest total score were the optimal choices, but in some instances, this was not the case. For example, for the specification to have a total weight less than or equal to 30 lbs, the highest-scoring concept was to use an 80/20 stock; however, in our alpha design, we plan to use 3D printed parts, machined aluminum, synthetic tissue, and purchased plastic materials since it will be easier and cheaper to configure these materials to our design.

Alpha Design

From our concept generation, our team converged our ideas to an alpha design. One main tool used in developing the alpha design was functional decomposition. Initially, we broke down our high-level stakeholder requirements into three subsections: structures & mounting, electromechanical substem, and synthetic anatomy.

For the structures and mounting, we decided to go with multiple concepts to give us flexibility with the materials used in our design. We plan on using a combination of aluminum, plastic, and 3D printed materials to address our weight constraint, decrease lead time, and decrease manufacturing time. Three straps with two around the shoulders and one around the waste plan to be used to constrain the device to the user and prevent sliding which would add additional noise to material movement data collected, see Figure 13.

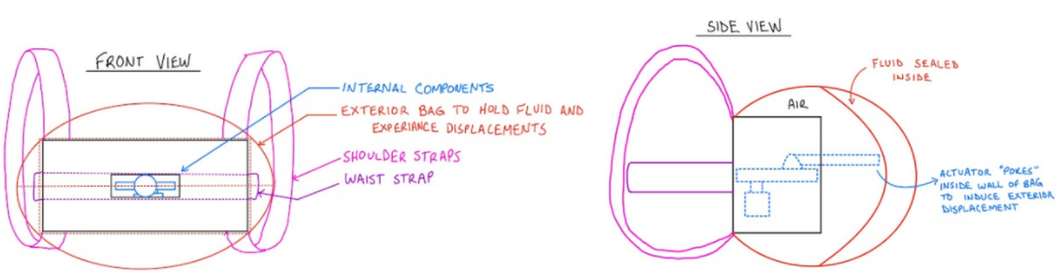


Figure 14. Preliminary sketches of alpha design derived from the weighted design matrix

The electromechanical subsystems primary objective was to induce surface displacements on the outer surface of the synthetic tissue that mimics the amplitude and duration of fetal movement. We ultimately determined that we needed to control three degrees of freedom (two spherical and one radial) shown in Figure 14. A linear actuator is tentatively being used to control the duration and magnitude of the impact force, which is controlled by a microcontroller. The linear actuator should have a full extension of 25mm and have a travel speed of at least 30mm/s. We plan on using empirical testing to determine the displacement profile.

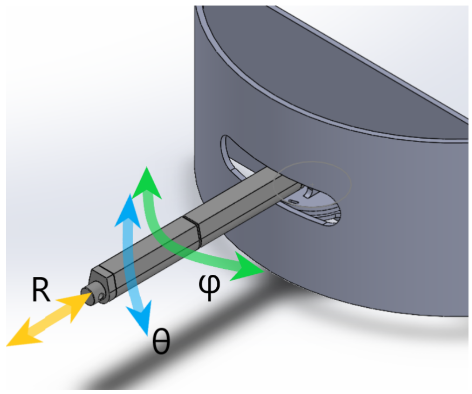


Figure 15. There are three degrees of freedom in our mechanism represented with spherical coordinates.

To control position in the theta and phi directions we plan to use a rotating plate that actuates vertically shown in Figure 15. The top plate will rotate using a gearing mechanism which will control the phi direction. The stage will be then driven by a rotating ⅜" threaded rod which will actuate an acme nut. The end of our actuator will pivot about a hinge but will be constrained in a slot shown in Figure 14. This will enable us to control the theta direction. We are currently exploring the use of a solenoid in place of the linear actuator for quicker actuating times.

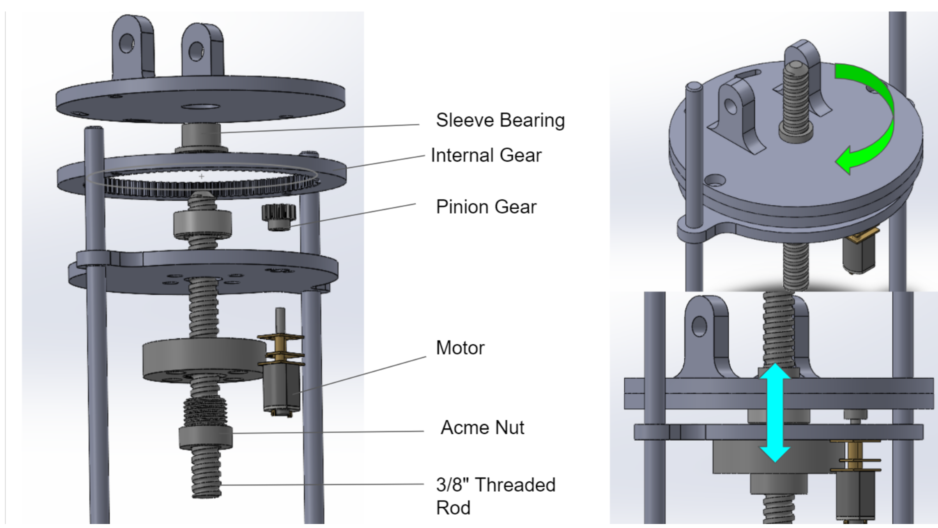


Figure 16. Three views of the internal mechanism: exploded view, top perspective, and side view.

The synthetic anatomy must resemble and maintain the shape of a pregnant abdomen and allow the propagation of impacts to the exterior surface. Since the displacement's magnitude and duration are the key outputs that must mimic the true values, the synthetic anatomy does not need to exactly replicate a fetus in utero and maternal abdomen. However, they should closely imitate the overall mechanical properties of the system. Thus, we simplified the model to decrease the complexity and time to implement. Our synthetic anatomy includes a sealed plastic fluid sac containing mineral water and an exterior layer of suture surgical phantom. The mineral water was chosen because it is a readily available, cheap option whose fluid properties are most similar to amniotic fluid. The suture surgical phantom was selected for its closeness to the approximated mechanical properties of the tissue layers and its availability.

Final Design

Cultural Impact

Each of our team members brought their own unique perspective and idea to the fetal kick simulator design process. Emily was able to provide some great insight with her extensive knowledge on arduino and programing the device. Adam provided lots of additional input on the timeline and ways to increase the cost of the design which kept us on budget. Alex was able to provide input on design for manufacturability, which greatly reduced the number of components and manufacturing time.

One of our sponsors, Dr. Bell, has a medical background. Our other sponsors and the team have an engineering background. Therefore, we came into the project with slightly different expectations and goals. Dr. Bell emphasized realistic anatomical features. While it was recommended to us by Prof. Perkins to focus most on getting the surface displacements correct. We had to balance these expectations and the realities of a short timeline. We did not end up focusing on realistic features, but we did add functionality like a fetal heartbeat to better simulate a fetus in utero.

Mechanical System

In our transition to the final design, we targeted the same three degrees of freedom (DOF) to control, but the mechanisms used to control these DOF underwent some change. The radial position which induces a displacement on the synthetic tissue is now controlled using a more compact linear actuator. The linear actuator has a max stroke of 50mm and is capable of inducing a surface displacement at 40mm/s which is required to accurately mimic fetal movements. The phi angle (rotation in the horizontal plane) is controlled using a Pololu motor which rotates the upper stage using an external 8:1 gear ratio. The theta angle (rotation in the vertical plane) is controlled by translating our rotational stage about the vertical axis using a ⅜”-16 lead screw and acme nut driven by a Pololu motor. The linear actuator rotates about a pin joint and is contained within a slot on our curved plate to achieve rotation about the theta angle. The rationale behind our linear actuator and motor selection is further discussed in the *Engineering Analysis and Iteration* section below.

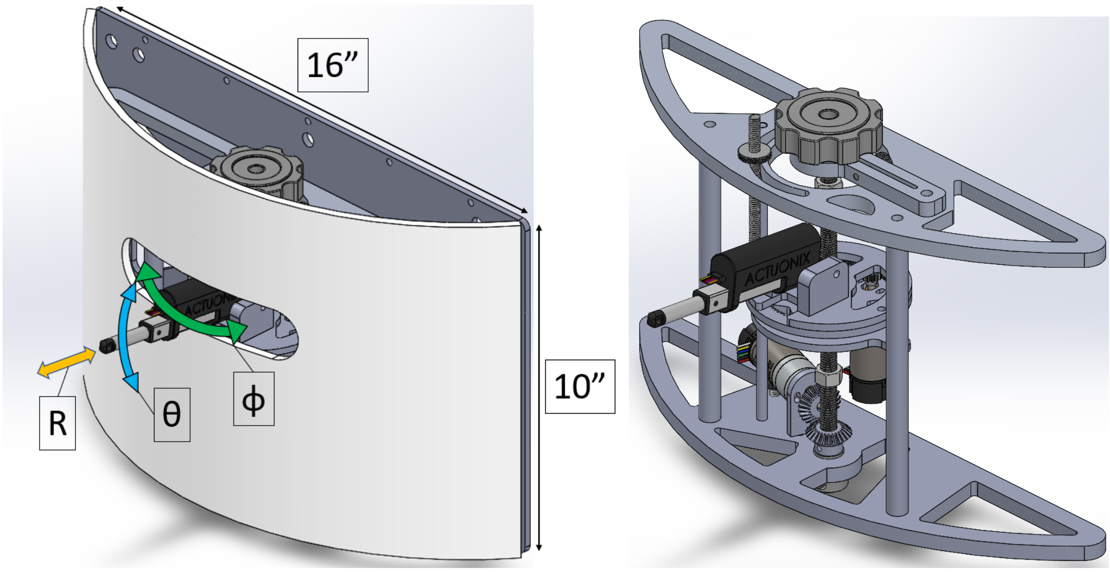


Figure 17. Full mechanical system overview.

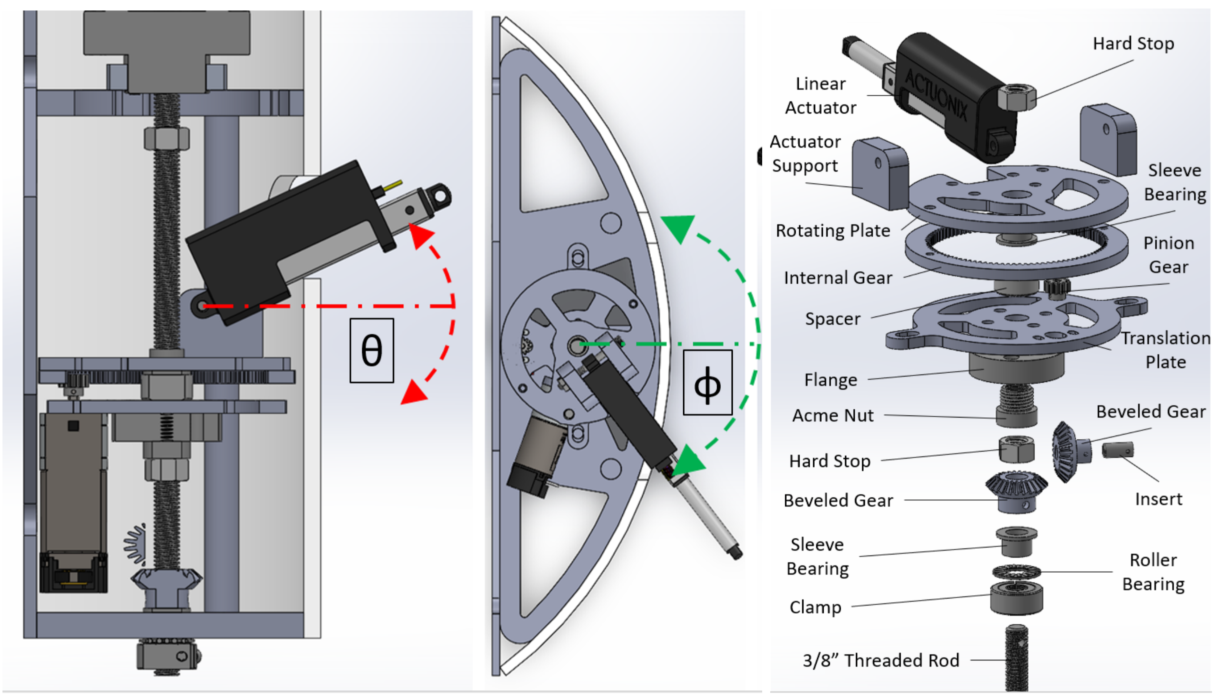


Figure 18. Control mechanism for radial, theta, and phi DOFs

We added functionality to manually control both the phi and theta angles. In discussion with our sponsors, they will be manually positioning the kick at the beginning of the testing. We kept the access for adjusting both the phi and theta angles at the top of the device along with locking mechanisms to prevent any rotation.

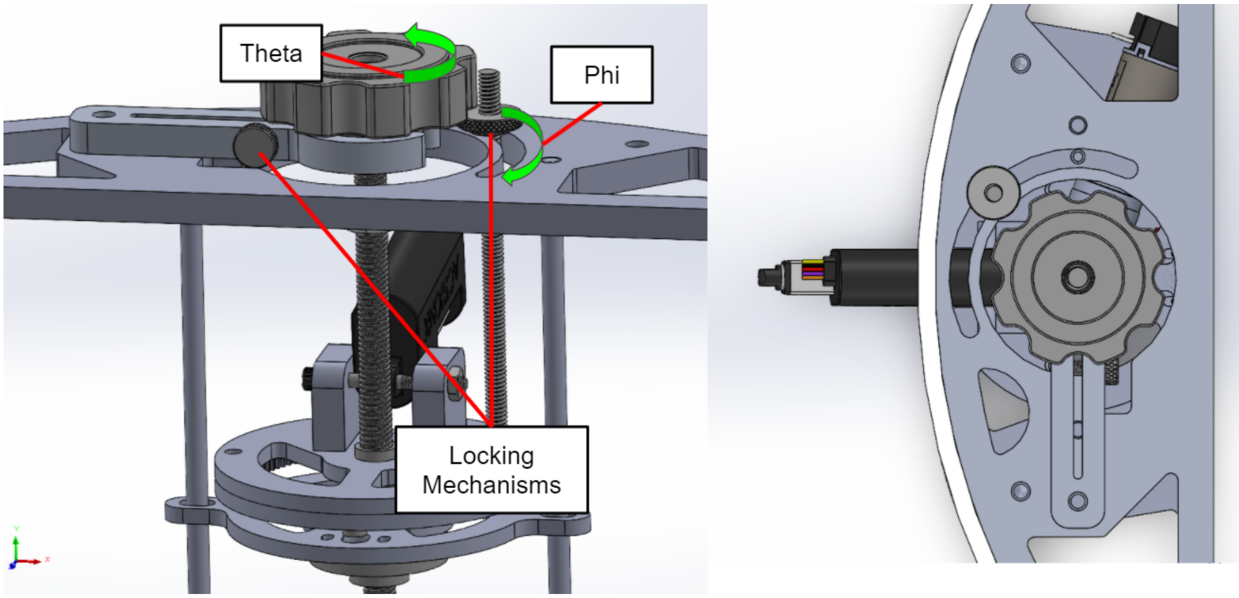


Figure 19. Manual phi and theta DOF control

In addition to manual control, we are able to automatically control the phi and theta angles using a 34:1 internal gear ratio Pololu metal gear motor with encoders, which will enable us to precisely control the position of phi and theta. The phi and theta position can be zero'd using lock nuts on the lead screw and the curved plate slot as hard stops. This will enable a programmable sweep of kick magnitudes, frequencies, locations, and durations.

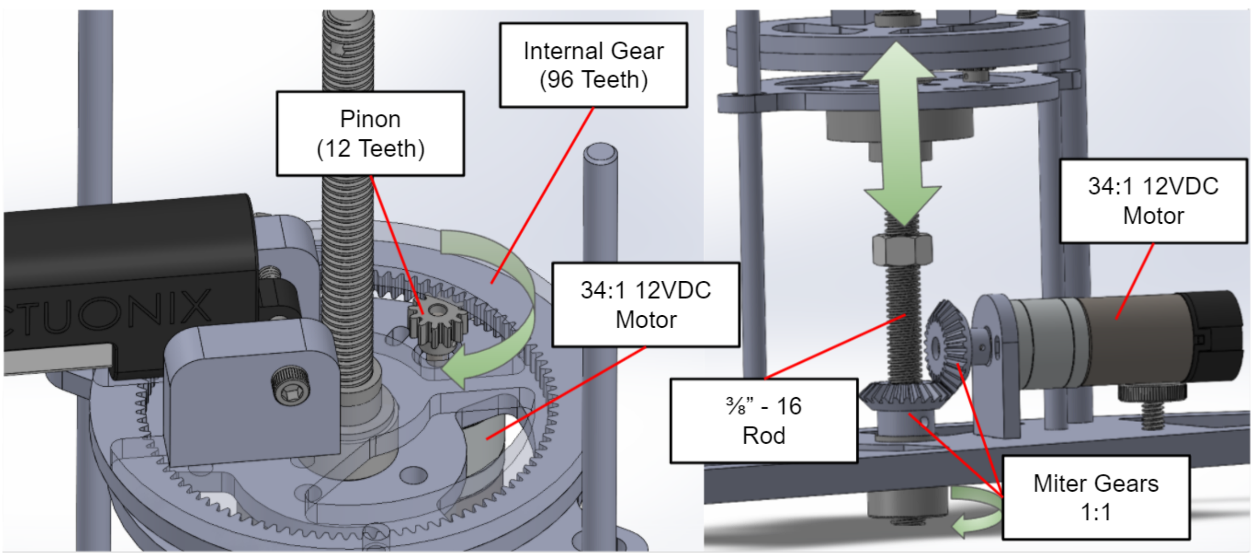


Figure 20. Automatic phi and theta DOF control

Overall this design allowed us to meet all our engineering specifications laid out in table 4. An overview of our systems capabilities can be found in the table below.

Table 4. Mechanical system capabilities

Feature	Description	Specification
Theta (Vertical) Range	26.5° (down) to 24° (up)	±20° from horizontal axis
Theta Angular Velocity	21.5°/sec @no-load	--
Phi (Horizontal) Range	105°	±100° about vertical axis
Phi Angular Velocity	225°/sec @no-load	--
Actuator Stroke	50mm (1.95”)	Minimum: 13mm
Actuator Travel Speed	40mm/s @8.3N	*26 mm/s
System Mass shown in CAD:	3.4kg (7.5lbs)	Weight under 30 lb
Estimated total mass of fetal kick simulator:	11.1572kg (24.6lbs)	Weight under 30 lb

* 26 mm/s is derived from maximum displacement of 13 mm and minimum kick duration of 0.5 sec

Wiring and Electrical

A preliminary wiring diagram can be seen in figure 20 below. As we learn more about the electrical components and needs, the wiring might need to change slightly. This should be a good foundation for the prototype. The electrical components include an Arduino to control and program the device, a remote, an infrared receiver, a speaker, a breadboard, two rotary motors with position encoders, a linear actuator with a potentiometer for displacement feedback, and two h-bridges to control current to the actuators.

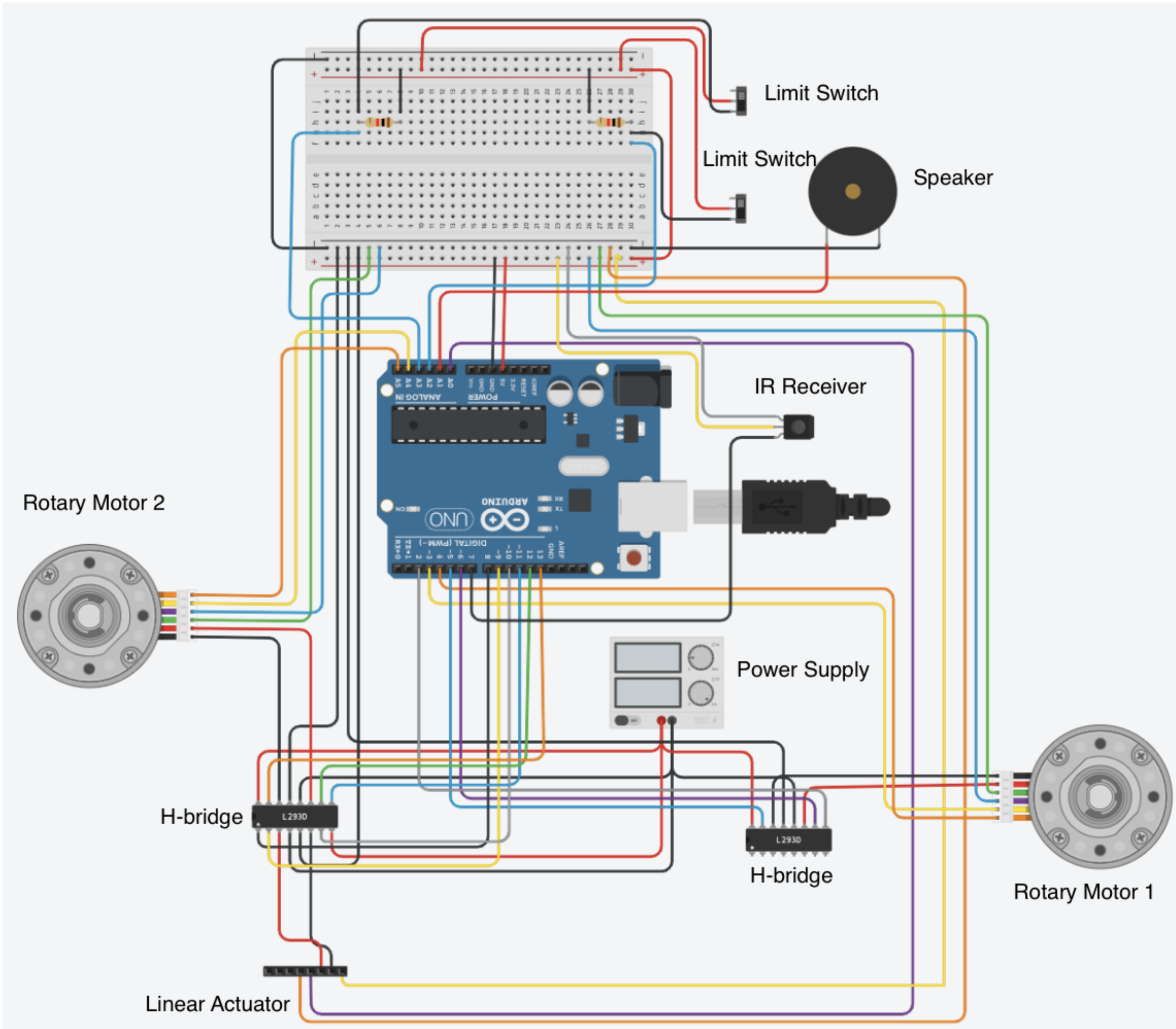


Figure 21. Wiring diagram for fetal kick simulator

The specific h-bridges that will be used can be seen in the figure below. They handle up to 2A at 12V.

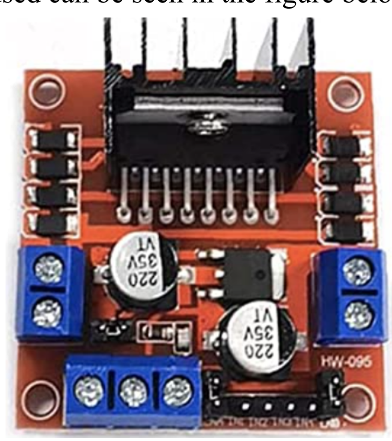


Figure 22. The h-bridge that will be used for the fetal kick simulator

There will be multiple power sources for the simulator as seen in figure 22 below. The 9V battery as seen below will power the Arduino and the breadboard. With powering the breadboard, both encoders, the potentiometer, and toggle switches are powered. The adjustable voltage DC power supply on the right will be used on the h-bridges and therefore the motors and linear actuator.

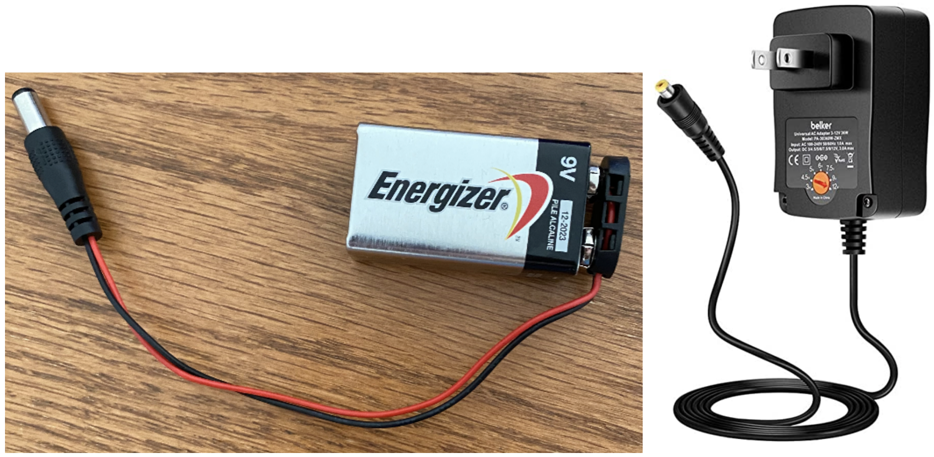


Figure 23. Left: a 9V battery and wire converter to power the Arduino and breadboard. Right: A variable DC voltage wall plug-in power supply used for motors and linear actuator.

Bill of Materials

A complete bill of materials for the mechanical and electrical system can be found at the end of this document. The total project cost was $1,127.71.

Engineering Analysis and Iteration

Different analytical tests will be performed on our prototype based on our design. These tests are performed in order to validate and improve our device with respect to our requirements and specifications. Some of these tests are empirical and iterative, while others will be performed before the features of the device are built. The analyses performed are linear actuator analysis, rotary motor analysis, fluid necessity analysis, and current analysis.

Linear Actuation Analysis

We needed to verify that the linear actuator would be able to displace the skin the largest amount in a half a second stroke. The maximum speed of the linear actuator that we need is 40mm/s. This is given that in the most extreme case, the surface needs to be displaced by about 1 cm in a half-second kick. From the force-speed curve in the linear actuator datasheet, the load at the point on the speed force curve is about 8.33N. The current at that point is 200 mA, and the power consumption is 1.4W given that we run the linear actuator at 12V. At stall force, the maximum current and power will be required from the linear actuator. The max current is 780mA and the power at this point is 9.4W.

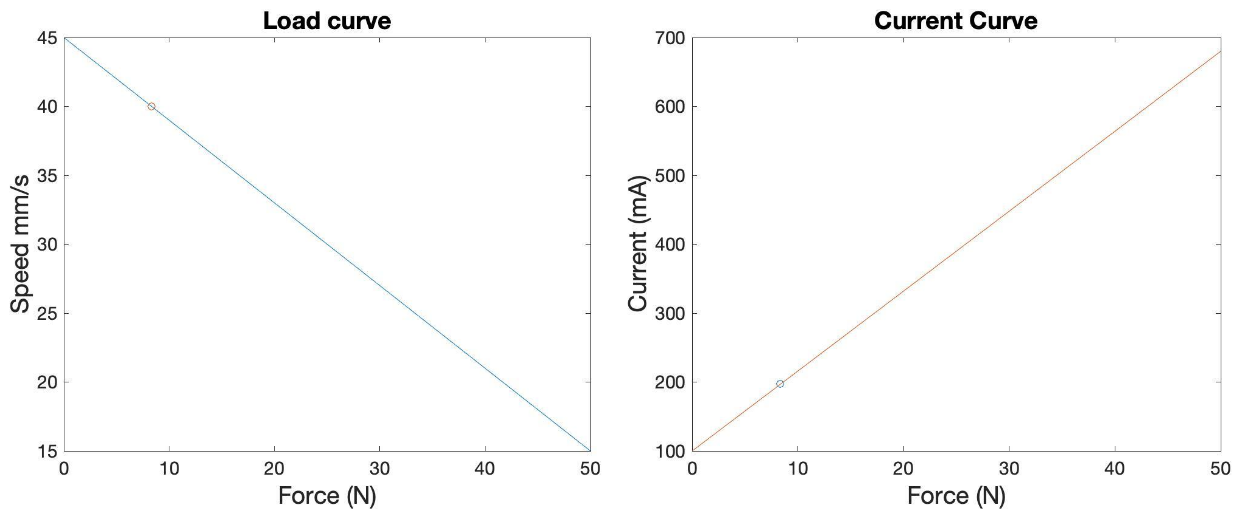


Figure 24. Force vs Speed and Force vs Current curves from the linear actuator datasheet

We used HyperWorks to approximate the displacement of the skin with an 8.5 N force. We decided to round to the nearest 0.5 decimal place because the analysis is an approximation. We applied this force to a rigid body element consisting of multiple elements, creating pressure from the force over these elements. The corners of the rubber were constrained in all degrees of freedom. We used the material properties of silicone rubber and the thickness value of our mat in the model [17]. If we add the mat, the load on the actuator would increase and the displacement would decrease. Therefore this is a worst case test scenario. The maximum displacement we found at the center where force was added was about 9.5mm, which is on par with the displacement we need at the 8.5N loads. The displacement field of the FEA can be seen in Figure 24 below. We believe this maximum displacement to be conservative as the boundary conditions on the corners are stationary, when in reality can also move with the kick because it is attached to a bag. The simulation was also a steady-state analysis. We believe that with acceleration and increased momentum the displacement will be greater. Therefore we should be able to meet the 1 cm displacement in a half-second kick requirement.

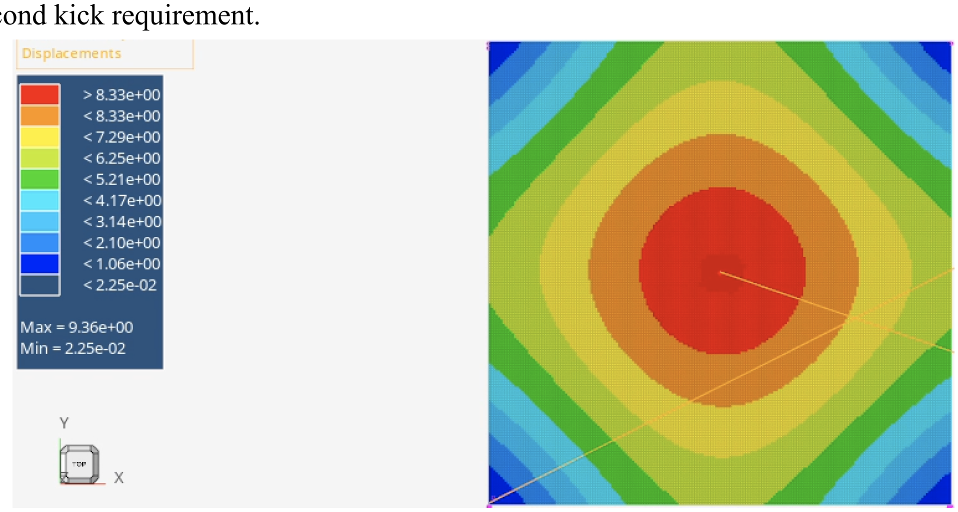


Figure 25. The displacement field of a 8.5N kick of a silicone rubber mat

Rotary Motor Analysis

To actuate the phi and theta angles, two rotary motors with encoders from Pololu can be used to move kick positions. Figure 25 below, is from the motor datasheet with torque-speed, torque-current, torque-power, and torque-efficiency curves. For the theta angle control, the gear ratio is 1:1 to actuate the lead screw. Assuming there is no load, ignoring friction, the maximum speed to drive the lead screw is 7.9 mm/s. This is an overestimate of speed. The actual speed will be slower due to torque caused by the threads of the lead screw, but we assume that the drive speed will still be faster than needed at 12V. Therefore, the power can be set to a lower voltage, decreasing the speed. The phi angle, the horizontal angle, is controlled with an 8:1 gear ratio. At no-load, the max speed is 37.5 rpm. With the larger gear ratio increasing torque handling capabilities we will be able to rotate horizontally with this motor at the cost of speed. Fast speeds for both motors is not a priority. The maximum current of the motor is about 2.9A and the max power is 8.9W at 12V.

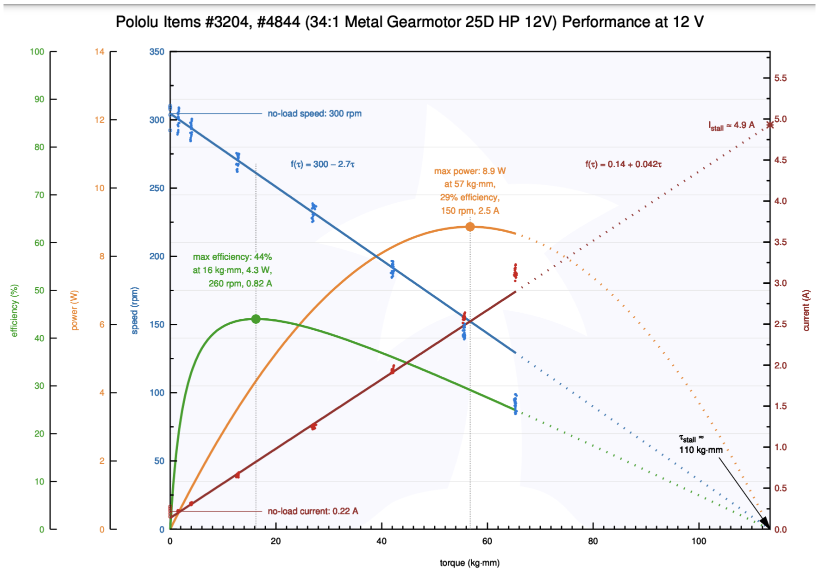


Figure 26. Torque-speed, torque-current, torque-power, and torque-efficiency curves from the rotary motor datasheet

Fluid Necessity & Type Empirical Testing

Originally, one of our primary objectives was to mimic the anatomy of the abdomen during pregnancy, focusing on the viscoelastic properties of tissue and fluid properties of the amniotic fluid. Our team pivoted to focusing on mimicking the displacements of the abdominal skin due to fetal movement. The original plan was to use mineral water in place of the amniotic fluid but our stakeholder, Prof. Perkins, mentioned that incorporating a synthetic amniotic fluid sac into our design may not be pertinent to accurately mimic the displacement profile. Our team explored the necessity of the fluid. We wanted to determine if the fluid could be substituted for other various substrates. Our main concern is the attenuation of a propagating displacement wave will vary substantially depending on the synthetic substrate we use in place of the amniotic fluid, see Figure 26.

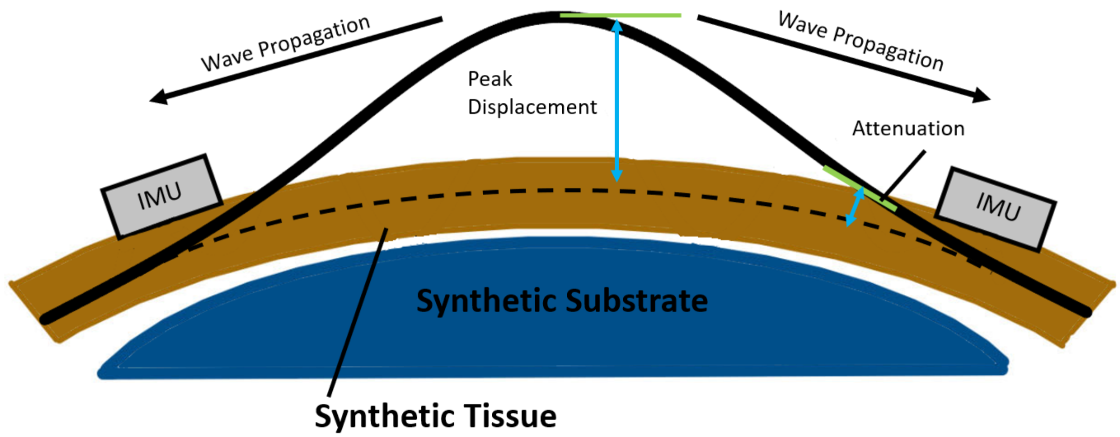


Figure 27. Depicts the wave propagation and attenuation of impact displacements

To further explore this potential concern our team tested two different synthetic substrates: expanded polystyrene foam beads and water. We also tested different types of bags: a reusable silicone bag, an extra large plastic zip-top bag, and a dry bag. This created a total of 6 material combinations. For each combination, a slow motion video was taken as a load was dropped on the bag. The bag was laid flat on the ground. A golf ball was used as the load. We dropped the ball vertically at 10in, 20in, and 30in to vary the load in a controlled way. We visually observed differences in displacements and wave propagation. Example screenshots from the slow motion videos can be seen in Figures 27-29 below.

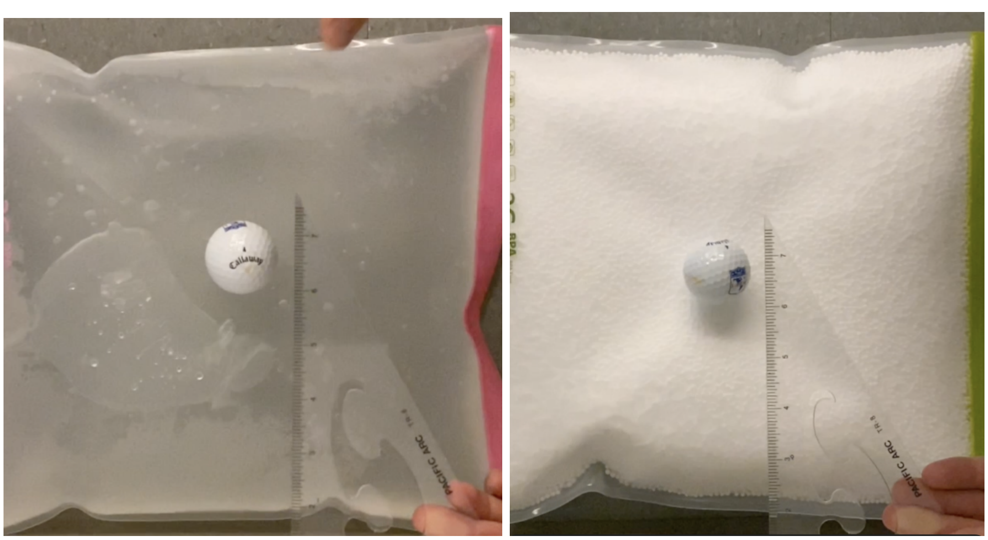


Figure 28. Silicone bag with water and foam fill, load from 10 in height

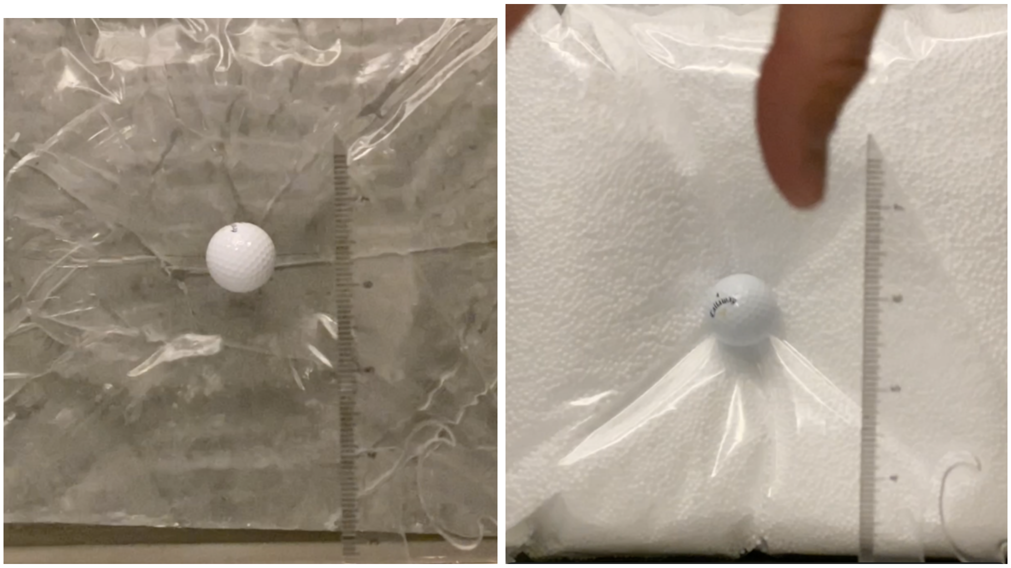


Figure 29. Extra large plastic bag with water and foam fill, load from 20 in height

Figure 30. Dry bag bag with water and foam fill, load from 30 in height

From the tests we found, that foam might actually lead to greater displacements than water. This result could mean that the foam would lead to greater surface displacements with the same force starting on the inside of the bag. In the tests, we also noticed that the drybag and the extra large zip-top bag leaked water. This makes those combinations ineligible to be in the final design. The plastic zip-top also was not durable. There were scratches on the bag at the end of the testing, making this bag a lower level choice for the final design. We think that from these tests, we can conclude that the silicone bag with foam will be the best option for the final design.

Current Analysis

For our simulator, we will need to have electrical circuitry to operate the different actuators and the controller. We want to make sure that in the worst-case scenario if someone were to accidentally ground themselves in the circuit, that there would be no major harm done to them. We want to ensure that the current in the circuit stays below 10 milliamps, therefore not significantly harming the person [18]. Unfortunately, keeping the current below this threshold will be unattainable. We found from the rotary motor at stall torque the maximum current 2.9A is and linear actuator at stall force the maximum current is 780 mA both at 12V operating conditions. To ensure safety, the wiring of these components must be clean. There can be no loose wires or exposed wires in our circuitry.

Verification and Validation

Displacement Analysis

We analyzed the displacement of the linear actuator using the linear potentiometer for two kick durations to gauge the displacements produced by the linear actuator and the time scale at which these displacements occur. We wished to verify that our mechanism can produce large enough displacements at a small enough time scale as dictated by our specifications. Utilizing the potentiometer of the linear actuator we recorded the displacement of the actuator with various voltage inputs for two set kick durations, 0.6 sec and 1 sec. Our results are shown below in Table 5. Kick Distance, as shown in the table, refers to the displacement, but the kick time is the time for a round trip of the actuator, that is both out and back in. Thus the velocity of the actuator is twice the Kick distance over the kick time. From the results in Table 5, it is evident that the linear actuator operates at one speed, approximately 30 mm/s, regardless of voltage. Our specification to produce a 13 mm displacement and smaller is satisfied, and so is our specification to produce a kick in approximately 0.5 sec.

Table 5. Results of the Displacement Analysis Tests we conducted in which Kick Time and Voltage are inputs and Kick Distance is the output.

Kick Time (sec)	Voltage (V)	Kick Distance (mm)
1.00	6	0.00
1.00	7	15.29
1.00	8	15.39
1.00	9	15.44
1.00	10	15.34
1.00	11	15.34
1.00	12	15.39
0.60	6	0.00
0.60	7	9.68
0.60	8	9.73
0.60	9	9.68
0.60	10	9.68
0.60	11	9.68
0.60	12	9.68

Impact Location Analysis

We analyzed the resolution of our manual control to determine the precision of our mechanisms produced kicks. We analyzed the manual control since it has the potential for a smaller resolution. The internal gear driving the horizontal motion has 96 teeth. Since the horizontal motion has a range of 105°, or 29.2% of a circle, the internal gears range spans 28 of its teeth. Therefore the 105° range divided by 28 teeth means the horizontal manual control has a resolution of 3.75°/tooth. Incorporating the linear actuator length, of 3.6 in., and using the equation for arc length, radius multiplied by angle, we calculated the surface level horizontal resolution to be 6.0 mm, which is less than the 2 cm specification we have. We determined the vertical resolution to be negligible since each full rotation of the threaded rod changes the height imperceptibly.

Wearability Analysis

We analyzed the wearability of our device empirically and quantitatively. Once the device was completely manufactured and assembled, we completed a series of simple tasks such as going from sitting to standing, lying to sitting, sitting and reaching, and walking. These are the tasks the simulator will be used for in order to test the wearable IMUs. We needed to ensure that the simulator was stable and allowed for user mobility. In the design and creation of our device we know with 3 straps, we meet the minimum of 2 strap specifications. We also weighed our device to ensure it is below the 30 lbs specification, which it was at 10.1 lbs. In completing the tests, we validated the wearability requirement. We also tested the continuous use requirement while testing the wearability, having the device operate for over 45 minutes, which it was able to accomplish.

Fetal Heart Rate Analysis

When programming the Arduino of the device, we found that the heartbeat sound and rate are variable. Therefore we know that we meet the specification of being within 110-160 bpm. We verified that the heartbeat can sound off at 110 beats per minute. But to validate the device, the sponsors should be able to use a fetal doppler to measure the heartbeats.

Discussion

Problem Definition

If more time was allowed our team would further verify our mimicked fetal kicks accurately replicate nominal fetal kicks. This could be done using IMU sensors placed on the outer surface of the abdomen and comparing the difference in angular velocity profiles as discussed in our problem analysis section. These profiles could be cross referenced with additional empirical testing on patients in the 3rd trimester of pregnancy. In addition, we would simulate other fetal movements like rolling, punching, and hiccuping to name a few. This would cover a broader range of fetal movements but would require modifications to our current design and would increase the overall complexity. Lastly, focusing on an easy way to implement programmable kick sequences would enable our device to more accurately mimic different levels of fetal activity and sleep cycles. This could be done by improving the user interface for our microcontroller to enable quicker programing of kick positioning, duration, surface displacement, and kick frequency.

Design Critique

Through the design and implementation of our fetal kick simulator our team was able to deduce strengths and weaknesses of our design. The ultimate objective of this design was to meet our stakeholders requirements by using our extrapolated engineering specifications. Table 6 below shows our team's reflection on how well we achieved each engineering specification.

Table 6. Stakeholder requirements with an additional column on how closely our team felt that we achieved our engineering specification on a scale of 1-10.

Requirement	Specification	Priority	Achieved
Wearable device	Total weight <= 30 lbs [12]	**II**	**10**
	At least 2 straps to mount onto the torso*	**II**	**10**
Mimics fetal movement	Mechanism can rotate 100° about its vertical axis*	**II**	**6**
	Impact Force can be applied ± 20° from horizontal*	**I**	**8**
	Generate a fetal heart rate at 110-160 BPM [13]	**III**	**10**
Mimics the displacement profile on the surface of the skin	Generate surface displacements from 0.5mm - 13mm [7]	**I**	**7**
	25% accuracy of peak amplitude displacement*	**I**	**10**
	Replicate impact durations of 0.5-3s [7][14]	**I**	**8**
Operates continuously	Can operate continuously for at least 45 min*	**II**	**7**
Fetal movements are controllable	Reference point to "zero" all DOF*	**II**	**3**
	Movements are programable*	**I**	**8**
	Impact force is centered within 2cm of the desired location*	**II**	**8**

Strengths

Our team felt like we hit both wearbility specifications well, where our total device weighed just north of 10lbs and had two shoulder and one torso straps to attach the device to the user. Our team was able to mimic fetal movement quite well and overachieved the requirement of applying a fetal kick at ± 20° from the horizontal, by using a stage which actuated up and down a lead screw achieving ± 24°. We were also able to mimic the full range of fetal heart rates, 110-160 BPM using our speaker system. The displacement profile on the surface of the skin was closely controlled using our linear actuator which was able to generate surface displacements up to 15mm in one second and about 10mm in 0.6 seconds. Using the feedback potentiometer to accurately sense the skin displacement within 5%. The device was powered using an extension cord enabling it to operate continuously for well over 45 minutes. Lastly, we were able to control our device using a remote controller which was able to accurately position the device. We also were able to control the device manually to position the kick well within 2cm of the desired location.

Room for Improvement

The initial design for manual control of the horizontal angle did not work as expected. When the pinion and internal gear were engaged the torque required to move the threaded rod was too much to overcome. The internal gear box of the motor has too much friction to overcome with this rod shown in Figure 31. The motor would have to be switched for one with less internal friction for the original design to work.

We came up with a new solution to manually control the horizontal angle. The backplate of the device needs to be removed. Then the user needs to lower the stage manually with the knob until the threads disengage. The user also needs to try to shimmy the plates apart during this process. Once the threads are completely disengaged, the user needs to rotate the actuator plate to the desired angle. The user then drops the top plate back down making sure the threads are engaged. The user can then manually adjust the vertical position with the knob to get to the desired kick impact location.

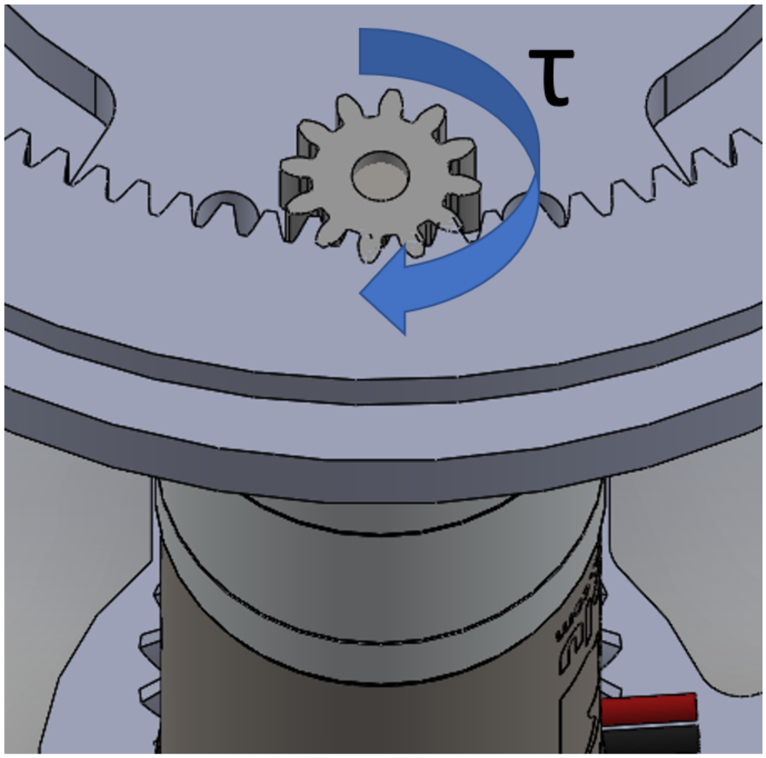


Figure 31. The pinon to internal gear interface to control the horizontal angle

To improve the manual horizontal positioning control, we suggest adding a second slot and bold with thumb screw to drive the mechanism. This might be able to add enough torque to overcome the friction in the system. Changing the motor to one with lower internal friction could also improve this functionality.

In our final implementation of the device we were having lots of issues with the microcontroller board short circuiting due to exposed wire and mounting the arduino right to the aluminium backing plate. For future designs we recommend using an ESD safe mounting pad to set the arduino on as shown in figure 32.

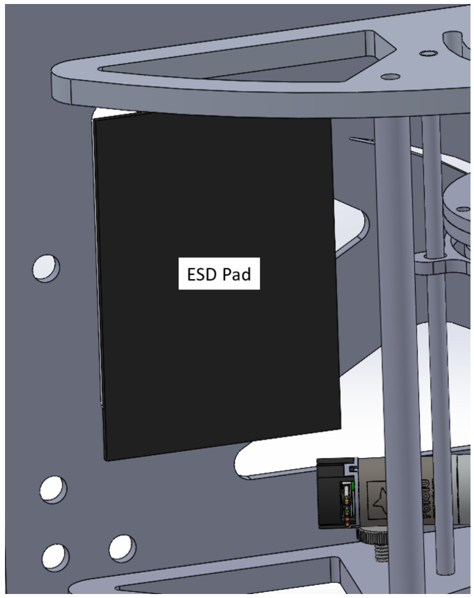


Figure 32. ESD Pad for mounting electrical components

The current electrical circuit is adhered to cardboard as a temporary solution to show the layout and wiring. The cardboard poses a fire hazard if kept as a final solution. The arduino and the IR receiver have shorted and need to be replaced. We suggest buying an arduino starter kit from amazon to replace the arduino and IR receiver.

The limit switches are wired and added into the code for use on the mechanism. The limit switches will set the encoder position to zero when toggled. But due to shorting of the arduino, we were not able to verify their use on the device. This verification will need to be done by our sponsors.

The design also does not fit well on our female wearers. The plate goes up too high on the user's chest. This does not impact the functionality of the design, but it does decrease the overall comfort of the user. Another improvement for comfort would be to add a foam cushion on the back plate during use. We also recommend decreasing the height of the design.

The clamp used to constrain the knob for vertical control is also not functional in the final design. The clamp and the area of concern are highlighted in figure 33 below. Completely loosened, there is too much friction and torque to overcome in order to turn the manual knob. Therefore in the prototype the clamp has been removed. This does not pose an issue with functionality. There is enough friction and torque in the system that the knob will not adjust undesirably from normal wear. To improve this design we recommend increasing the internal diameter of the clamp so that it does not interfere with the hand knob. The interference fit is highlighted in Figure 33.

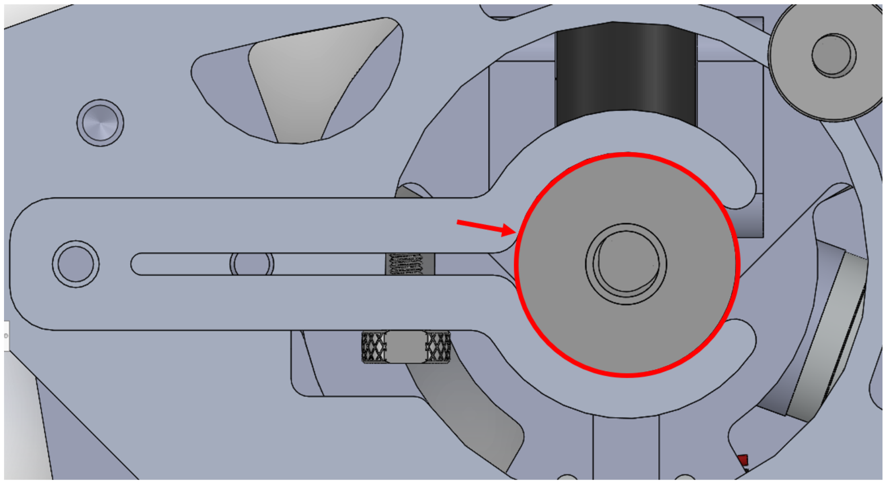


Figure 33. The clamp on the knob for vertical control is has too close of a fit to be functional

Another improvement that the sponsors could make would be to replace the linear actuator with one that is faster or a solenoid. The current one is slightly too slow to fully meet the requirements, but the speed and displacement are well within 25% of desired.

Recommendations

All of the important files such as CAD, manufacturing plans, electrical schematics, and arduino sketches can be found in a shared google folder.

High Level

We recommend adjusting the straps for every user. This will ensure the best fit and comfort. The simulator is to be worn snuggly on the abdomen. We also highly recommend adding a plastic/acrylic/ or ESD pad to mount the electrical component. This will make sure that components do not short and break. As said previously, the arduino board and IR receiver need to be replaced. There can also be additions to the code for added functionality. To slow the kick, a for loop with a time interval for the actuator to move out then another time interval of the actuator being stopped could be implemented. These time intervals would be on the order of 10-100 ms. This will allow the kick to slow down, but still appear to move continuously even though it would be discrete.

Detailed-level

For the electrical circuitry, we recommend connecting either the battery supply or usb to the arduino. We also recommend that no metal from the device touch the circuitry. If there are errors uploading sketches to the arduino such as “error avrdude:stk500() program is not responding” you will need to replace the arduino board. To replace the arduino board, you can remove the digital and analog plastic pin hubs from the old arduino, keeping the wires in the pin slots. You will need to use some sort of lever to separate the hubs from the board. You can do the same with the new board, then place the pin hub with wires from the broken arduino onto the fresh arduino. This will allow you to not have to rewire the board.

For the mechanical system, we recommend using clamping collars as hardstops instead of hex nuts. This recommendation is reflected between the CAD and on the physical prototype. The user may also find it beneficial to replace the internal gear for horizontal positioning. We waterjetted the internal gear, because of this some of the teeth are slightly angled. We recommend buying a new gear which is linked in the Bill of Materials. We believe that this could reduce some of the friction in the system. The bag on the front of the prototype also could be filled with foam beads or foam layers to attenuate the displacement if that is desired by the user. The bag opens and reseals. If the acme nut and acme flange come loose, add Loctite to the threads.

Conclusion

This report contains the background of the need for a fetal health monitor. As well as, our project's role in its development. A fetal movement simulator is needed to test and verify the wearable fetal health monitors in development. Specifically, the simulator will be used to determine if the fetal health monitor has the capability of distinguishing maternal and fetal movement. Requirements and specifications have been developed in order to meet our stakeholders' needs. Some of these requirements include making a wearable simulator, accurately inducing abdomen surface displacements from fetal movements, the ability to control the movements, and making the device have a continuous operating time similar to that of an ultrasound. An in-depth analysis of engineering fundamentals helped quantify the specifications. Concept generation, with processes including divergent thinking to develop many ideas and convergent thinking to narrow ideas down, helped develop our alpha design. A preliminary CAD model and sketches were then formulated. A final design was iterated upon. The prototype was manufactured and assembled. We conducted testing to verify our design. We also created recommendations for future use to be carried out by our sponsors.

Acknowledgements

Our project this semester has been an arduous process that would not have been possible without the contributions of our sponsors Lucy Spicher, Dr. Carrie Bell, and Prof. Noel Perkins. All of them have been helpful as stakeholders identifying the requirements and providing valuable input for this project. Prof. Perkins additionally served as our instructor for this course and provided a plethora of feedback and input from this perspective of our project as well. We would also like to recognize the wonderful machine shop staff, specifically Charlie Bradley and Jonathan Yenkel, for their assistance in the manufacturing of our prototype. Finally, we would like to acknowledge the other members of our section that provided valuable input and asked provoking questions during our design review presentations.

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Appendix A. Specifications to Mimic Skin and Amniotic Fluid Properties

These specifications were previously included in the design of the mechanism. As the team learned more from the stakeholders, we decided that the displacements of the surface are more important to the design than the material being displaced. Table A.1 discusses the former requirements and specifications.

Table A.1. Mechanical properties of abdominal tissue and amniotic fluid

Former Requirements	Former Specifications
Imitates viscoelastic properties of abdominal tissue	Young's modulus, E = 96kPa ± 10% [19]
	Dynamic viscosity, μ = 23Mpa-s ± 10% [20, 21,22]
	Density, ρ = 962kg/m^3 ± 10% [23]
Imitates fluid properties of amniotic fluid	Dynamic viscosity, μ = 1.10 MPa-s ± 5% [15]
	Volume = 770ml ± 5% [15]
	Density, ρ = 1000kg/m^3 ± 2% [24]

Imitates viscoelastic properties of abdominal tissue: To correctly reproduce the propagation of a pressure gradient through synthetic tissue it was necessary to mimic the viscoelastic properties of the abdominal tissue. Our ultimate objective is to mimic surface displacements on the skin, so this stakeholder requirement was relaxed. A compromise was made to ease the manufacturability of the synthetic tissue while still mimicking its mechanical properties by using just one homogeneous material for all of the tissue. A weighted average was used for Young’s modulus, dynamic viscosity, density, and thickness of the material.

Imitates fluid properties of amniotic fluid: The mechanical properties of the fluid were requested to simulate the environment of the fetus at 28 weeks. To accomplish this the drag, volume, and density should be closely matched based on our findings [15].

Appendix B. Viscoelastic Properties of Tissue

We focused on the anterior abdomen and the propagation of a pressure gradient through the uterine wall, abdominal muscle, and adipose tissue. A continuous Kelvin-Vogit model was used to model the tissue of the interior abdomen because of the viscous and elastic properties of the material act in parallel. The stress, σ, induced on the system is:

$$\sigma = \eta \frac{d\varepsilon}{dt} + E\varepsilon \qquad \text{(B.1)}$$

Where η is the viscosity coefficient, ε is the elastic strain, and E is Young's modulus. The equation of motion can then be written using Newton's second law for each component:

$$m_i \frac{d^2 u_i(t)}{dt^2} = \sigma_{i+1} - \sigma_i \qquad \text{(B.2)}$$

Where *m* is the mass of each tissue layer and *i* is the layer of tissue (i = 1 is the most inner layer of tissue).

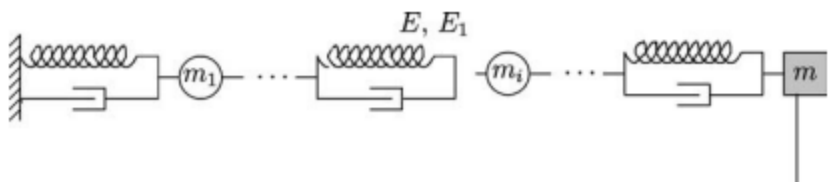


Figure B.1. Model for the pressure propagation through human tissue [24]

From the equations B.1 and B.2 four key properties are required to mimic the propagation of a pressure gradient through the tissue layers: Young's modulus to mimic the compliance of the system, viscosity to mimic attenuation of the system, density to mimic the mass, thickness to mimic the length of the system. These values can be seen in Table B.1 on the next page.

Table B.1. Key mechanical properties of the human anatomy

Anatomy	Young's Modulus (kPa)	Viscosity (MPa-s)	Density (kg/m^3)	Thickness (mm)
Adipose Tissue	12	60.6 @25C	916	10 @18BMI 40 @30BMI [25]
Abdominal Wall	21	~0	973	31.3
Uterine Wall	586	0.2	1052	9.1
Amniotic Fluid	-	1.1	1007	**Volume (mL)** 770

While it would be ideal to mimic the mechanical properties of each tissue layer to best model the propagation of a pressure gradient, the overall emphasis was to mimic surface displacements on the epidermis. With this objective in mind, scaling the magnitude or duration of the impact force/displacement on the synthetic posterior uterine wall enables the adjustment of peak displacements on the epidermis surface and relaxes the requirement of mimicking the mechanical properties of the tissue. This enables our synthetic tissue to be composed of one material which greatly decreases the complexity of our design. A weighted average of these mechanical properties by wall thickness was then calculated for a homogenous material that still mimicked the mechanical properties of the abdominal tissue. Wall thickness was used to weigh each property because it describes the fraction of the total length traveled through each tissue medium. An example calculation (equation B.3) for a weighted Young's modulus, E_{avg}, is shown below where the thickness of the adipose tissue was 2.5cm.

$$E_{avg} = \frac{E_{AT}(25mm)+E_{AW}(31.3mm)+E_{UW}(9.1mm)}{25mm + 31.3mm + 9.1mm} = 96\ kPa \qquad \text{(B.3)}$$

Where *AT, AW, UW* is the adipose tissue, abdominal wall, and uterine wall. Table B.2 below shows our targeted mechanical properties of a homogeneous synthetic tissue.

Table B.2. Key mechanical properties to target in the making of a homogeneous synthetic tissue.

Young's Modulus (kPa)	Viscosity (MPa-s)	Density (kg/m^3)	Thickness (mm)
96	23	962	65.5

Appendix C: Morphological Chart

We developed a morphological matrix that outlines each requirement and specification. This appendix will go into greater depth of each idea generation category and how the ideas were developed. Each of the tables below will show the concepts generated.

Table C.1. There are two specifications that envelope the requirements that the device must be wearable. For the total weight specification, we generated materials that are lightweight. For mounting straps, we used design heuristic to expand upon basic ideas like backpacks and carrying pouches that parents use. Some of the heuristics used repeated, elevate or lower, and reconfigure to vary strap location and amount in order to carry the simulator.

Requirement	Specification	Brainstorm Concepts to Satisfy			
Wearable device	-Total weight <= 30 lbs [12]	Use plastic for fetus mechanism	Use only 3d printed parts for mechanical fetus	Use 80-20 stock	
	-At least 2 straps to mount onto torso *	Two back straps-- reverse backpack	Two cross straps around back	one strap across back of body, one strap around waist	Kangaroo pouch with reverse backpack

Table C.2. There are three subsections for the mimic fetal movement category. We used similar design heuristics to generate ideas on how the kick mechanism can rotate about the horizontal and vertical axis. Some of these include: visually distinguishing functions, rolling, rotating, and redefining joints. For the last category of mimicking fetal movement, we generated ideas to mimic the fetal heart rate. We learned from our sponsors that the heart rate sensor will be acoustic. Therefore we generated ideas that could produce a heartbeat sound.

Requirement	Specification	Brainstorm Concepts to Satisfy				
Mimics fetal movement	-Mechanism can rotate 180 about its vertical axis*	motorized pivot joint for entire mechanism	be able to adjust device manually	Spine thru whole "bag" (top to bottom)	Have a belt system	Have multiple mechanisms to kick
	-Impact Force can be applied ± 30 degrees from horizontal	(Motorzied) pivot joint	Linear Actuator can be dragged to different locations on back wall	Linear vs rotational actuators.		
	-Generate a fetal heart rate at 110-160 BPM [13]	Use speaker for FHR	Use animal heart rate toy	use research articles wiring diagrams and make one ourselves	Could we use our fetus mechanism to induce small displacements on the wall that	

Table C.3. Mimicking skin level displacements had two categories. One to generate and one to evaluate simulated kicks. We started off with the concept of a linear actuator taken from the workbench fetal kick similar found in our research. then used heuristics to generate more ideas. These include allowing users to rearrange, utilize inner space, and redefine joints. For evaluating the accuracy, we looked to our sponsor, Prof. Perkins, for suggestions like a dial indicator.

Requirement	Specification	Brainstorm Concepts to Satisfy							
Mimics the displacement profile on the surface of the skin	-Generate surface displacements from .25mm - 13mm and last 2.5-3.0s [7]	Linear actuator	3 or 4 bar linkage	use off the market battery powered baby doll or toy animal	Slingshot from back wall of uterus	Have a mechanical lever system to cause surface displacements,	Fit wearer's arms inside pouch, they poke/ punch the wall	Pneumatic piston	Solenoid
	-25% accuracy of peak amplitude displacement*	Measure with 3 I-newton and assume displacement has a	Use opals to test displace	Use camera or optical system to measure displacement. Could	Capacitance with magnetic tape	Dial Indicator			

Table C.4. We developed ideas for power sources to fulfill our continuous operation requirement. For example, our source could be battery or solar-powered.

Requirement	Specification	Brainstorm Concepts to Satisfy			
Operates continuously	-Can operate continuously for at least 45 min.*	Battery powered	Plug in to outlet	Solar powered	Rechargeable battery powered

Table C.5. For the imitation of fluid and skin properties, we thought of ideas only based on the requirement rather than the specifications. To imitate viscoelastic and fluid properties we looked to sources that had these phantom materials used for skin and their material properties we well defined and fit within our specifications. We also discussed breaking with our requirements and the implications that could have.

Requirement	Specification	Brainstorm Concepts to Satisfy					
Imitates viscoelastic properties of abdominal tissue	-Young's modulus, E = 96kPa ± 10% -Dynamic viscosity, μ = 23Mpa-s ± 10% -Density, ϱ = 962kg/m3 ± 10% -Thickness = 65.5mm ± 10%	Use agar phantom material	Use gelatin phantom material	Use PVA phantom material	Use silicon as skin, ignore mechanical properties -- just focus on displacement	Use a suture surgical phantom	
Imitates fluid properties of amniotic fluid	-Dynamic viscosity, μ = 1.10 Mpa-s[14] ± 5% -Volume = 770ml[15] ± 5% -Density, ϱ = 1000kg/m[15] ± 2%	Use mineral water	Don't use any fluid	Use tap water	Use a gel	Use foam pellets	Use foam layers

Table C.6. To have our movements be controllable we discussed techniques that would zero our device, such as hard stops and limit switches. For the controller, we looked to Arduinos and raspberry pico controllers. We also looked for ways to test our mechanism later on for verification. We also looked into how we would make our movements controllable. We could conduct experiments to vary voltage or use encoders to move motors to desired counts

Requirement	Specification	Brainstorm Concepts to Satisfy			
Fetal movements are controllable	-Reference point to "zero" all DOF.*	Use water proof limit switches.	have all motors/ actuators wiring separated by wall from	Have hard stops	
	-Movements are programable.*	Use aurdiuno wifi maker 1010 so the movments could be commanded	Use rasberry pico micro controller (Alex Has 10)	Connect right to the PC	Read in commands by programing a good user interface
	-Impact force is centered within 2cm of the desired location.*	Draw grid on inside of the uterine wall	Draw grid on the outside and see if the synthetic tissue is see through	Validate x points based on command from zero datum and if all are in	
	-Impact force is variable in magnitude.*	Control actuator with pneumatics can adjust force with pressure	Conduct experimental evidence of power/displacements	Use encoders to control displacements and forces	

Table C.7. For our requirement to adjust BMIs we looked at design heuristics. These include allowing users to assemble and Reduce material. This allowed us to come up with ideas such as removable layers from a pocket or an inflating balloon layer and no outer layer at all.

Requirement	Specification	Brainstorm Concepts to Satisfy				
Ability adjusts for various BMI's	-Adipose tissue be removable and replaceable to mimic different BMIs (18-30 BMI).*	Velcro attachable layer to front	Use double sided tape to attach layer to front of uterus sac	Pocket for layers to slide into/out of	Have layer be like a "balloon" to fill with a fluid	Attenuate force magnitude

Some of the sketches for the ability to adjust BMI requirements can be seen in figure B.1 below.

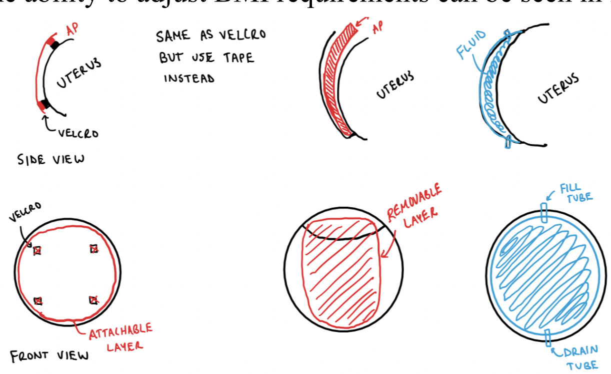


Figure B.1. Sketches of making AT adjustable

Appendix D: Weighted Design Matrix (WDM)

Table D.1. We used a weighted design matrix to evaluate and compare the various concepts we generated for each specification. Each concept was evaluated in 5 weighted categories shown at the top of each of the three pages. For each category, a concept was scored from 1 to 10 with 10 being the best score. The highest score is bolded for each specification.

	Weight:	**5**	**5**	**1**	**4**	**3**	**/180**
Specification	**Brainstormed Concept**	**Technical Feasability**	**Time to implement**	**Cost**	**Agrees with Stakeholder Requirements**	**Accuratley mimics fetal anatomy**	**Total Weighted Score**
Total weight <= 30 lbs.	Use plastic for fetus mechanism	6	5	8	8	4	107
	Use only 3d printed parts for mechanical fetus	3	10	7	5	2	98
	80-20 stock/plate	9	7	8	8	3	**129**
At least 2 straps to mount onto torso.	Two back straps-- reverse backpack	9	8	7	6	6	134
	Two cross straps around back	9	8	7	7	5	135
	One strap across back of body, one strap around waist	9	7	7	7	6	133
	Kangaroo pouch with reverse backpack	7	7	7	10	8	**141**
Mechanism can rotate 180 about its vertical axis.*	Motorized pivot joint for entire mechanism	7	7	7	10	9	**144**
	Able to adjust device manually	8	8	9	7	9	**144**
	Spine thru whole "bag" (top to bottom)	9	8	7	10	3	141
	Have a belt system	6	7	7	10	3	121
	Have multiple mechanisms to kick	6	5	5	8	1	95
Impact Force can be applied ± 30 degrees from horizontal.	(Motorzied) pivot joint	8	7	6	10	7	**142**
	Linear Actuator can be dragged to different locations on back wall	8	8	8	6	5	127
	Linear vs rotational actuators.	8	7	6	10	7	**142**
Generate a fetal heart rate at 110-160 BPM.	Use speaker for FHR	10	10	10	10	4	**162**
	Use animal heart rate toy	10	10	10	4	4	138
	Use research articles wiring diagrams and make one ourselves	6	4	4	9	9	117
	Use our fetal kick mechanism to induce small displacements on the wall that mimic fetal heart rates?	3	3	10	4	4	68
Generate surface displacements from 0.25mm - 13mm	Linear actuator	9	5	8	10	7	**139**
	3 or 4 bar linkage	9	4	6	9	8	131
	Off the market battery powered baby doll or toy animal	7	6	8	8	9	132
	Slingshot from back wall of uterus	5	8	9	5	3	103
	Mechanical lever system to cause surface displacements, be controlled by wearer of simulator. L shaped	7	9	9	6	4	125
	Fit wearer's arms inside pouch, they poke/ punch the wall	10	10	10	1	1	117
	Pneumatic actuator	8	5	5	8	7	123

Specification	Brainstormed Concept	Technical Feasability	Time to implement	Cost	Agrees with Stakeholder Requirements	Accuratley mimics fetal anatomy	Total Weighted Score
	Weight:	5	5	1	4	3	/180
Imitates viscoelastic properties of abdominal tissue	Use agar phantom material	6	7	6	8	9	130
	Use gelatin phantom material	6	7	7	7	7	121
	Use PVA phantom material	6	7	7	7	5	115
	use silicon as skin, ignore mechanical properties -- just focus on displacement	8	8	7	6	5	126
	Use a suture surgical phantom	8	8	6	8	8	**142**
Imitates fluid properties of amniotic fluid	Use mineral water	9	10	8	9	7	**160**
	Dont use any fluid?	7	10	10	5	1	118
	use tap water	8	10	10	6	5	139
Reference point to "zero" all DOF.*	Use water proof limit switches.	9	5	5	9		111
	have all motors/ actuators wiring separated by wall from phantom uterus	6	4	8	9		94
	Have hard stops	10	7	4	9		**125**
Movements are programable.*	Use Ardiuno wifi maker 1010 so the movments could be commanded	10	7	8	10		**133**
	Use rasberry pico micro controller (Alex Has 10)	7	7	8	10		118
	Connect right to the PC	3	3	10	10		80
	Read in commands by programing a good user interface	5	5	10	5		80
25% accuracy of peak amplitude displacement*	Measure with 3 I-newton and assume displacement has a spherical profile to estimate the displacements	10	7	9	9		130
	use opals to test displacememt	10	5	7	10		122
	Dial Indicator	10	9	9	7		**132**
	Capacitance with magnetic tape	8	7	7	7		110
	Use camera or optical system to measure displacement. Could put a bunch of markers and measure displacement in 3d space.	6	8	10	6		104
Can operate continuously for at least 45 min.	Battery powered	8	9	5	10	10	160
	Plug in to outlet	9	8	9	1	10	128
	solar powered	2	2	1	1	10	55
	rechargeable battery powered	9	9	5	10	10	**165**

Specification	Brainstormed Concept	Technical Feasability	Time to implement	Cost	Agrees with Stakeholder Requirements	Accuratley mimics fetal anatomy	Total Weighted Score
	Weight:	**5**	**5**	**1**	**4**	**3**	**/180**
Impact force is centered within 2cm of the desired location.*	Draw grid on inside of the uterine wall	8	10	10	10		140
	Draw grid on the outside and see if the synthetic tissue is see through	10	10	10	10		**150**
	validate x points based on command from zero datum and if all are in cal could assume 2cm accuracy is met?	10	6	10	10		130
Impact force is variable in magnitude.*	Control actuator with pneumatics can adjust force with pressure	10	10	10	10		**150**
	Conduct experimental evidence of power/displacements and calibrate design to this.	10	6	10	10		130
	use encoders to control displacements and forces	10	7	10	10		135
Adipose tissue be removable and replaceable to mimic different BMIs (18-30 BMI).*	Velcro attachable layer to front	10	10	10	10	8	174
	use double sided tape to attach layer to front of uterus sac	10	10	10	10	8	174
	Pocket for layers to slide into/out of	7	5	6	10	10	136
	Attenuate force magnitude	10	10	10	10	9	**177**
	have layer be like a "balloon" to fill with a fluid	6	4	7	10	9	124

Bill of Materials

Part	Supplier Name	Serial Number	Cost	Qty	Total Cost
Rotary motor and encoder	Pololu	4844	$36.95	1	$36.95
Stuffing/ beading	Amazon	B01BUHQN20	$29.74	1	$29.74
Speaker	Amazon	a15080600ux0275	$8.80	1	$8.80
Arduino	Amazon	B07J2QKNHB	$9.99	1	$9.99
Hbridge	Amazon	B099K8DWVV	$11.99	1	$11.99
Power cord adapter	Amazon	4335094611	$7.49	1	$7.49
Toggle switch	Amazon	789659499440	$7.36	1	$7.36
Rubber with Adhesive	McMaster-Carr	1464N262	$209.96	1	$209.96
Bottom Gear	McMaster-Carr	6325K88	$29.71	1	$29.71
Pinion	McMaster-Carr	2664N314	$14.80	1	$14.80
Ring Gear	McMaster-Carr	2696N15	$72.76	1	$72.76
Bag 1	Amazon	B07KFPD8TK	$23.99	1	$23.99
Bag 2	Amazon	B08FR9JGYY	$11.68	1	$11.68
Bag 3	Amazon	B07K1G1M52	$24.99	1	$24.99
Back Strap	Amazon	B075ST5ZDQ	$9.99	1	$9.99
Backpack Straps	Amazon	B078VZXDM5	$11.99	1	$11.99
Strap for bolt	Amazon	B0772V94MC	$9.99	3	$29.97
Linear actuator	Digi-Key	2495-P16-50-22-12-P-ND	$115.00	1	$115.00
Acme Flange	McMaster-Carr	1329K11	$40.32	1	$40.32
Acme Nut	McMaster-Carr	95072A129	$27.48	1	$27.48
Clamping Collar	McMaster-Carr	6698K52	$8.00	1	$8.00
Knob	McMaster-Carr	5532T35	$3.25	1	$3.25
Knurled-Head Thumb Screw	McMaster-Carr	91746A364	$5.64	1	$5.64
Lead Screw	McMaster-Carr	99030A996	$9.28	1	$9.28
Low-friction Sleeve Bearing	McMaster-Carr	1677K325	$1.28	1	$1.28
Nylon Thumb Screw	McMaster-Carr	94955A075	$4.38	1	$4.38
Shaft	McMaster-Carr	6061K101	$5.66	1	$5.66
Steel Hex Nuts	McMaster-Carr	98797A031	$6.46	1	$6.46
Thrust Bearing	McMaster-Carr	5909K25	$3.27	1	$3.27
Thumb Nut	McMaster-Carr	92815A115	$2.82	1	$2.82
Gear - 20 deg Pressure Angle	McMaster-Carr	7880K25	$16.76	1	$16.76

Miter Gear	McMaster-Carr	6529K53	$32.38	2	$64.76
Press-Fit Drill Bushing	McMaster-Carr	8491A421	$10.32	1	$10.32
Press-Fit Drill Bushing	McMaster-Carr	3671N126	$10.88	1	$10.88
Power supply	Amazon	B07N18XN84	$15.90	1	$15.90
Rotary motor and encoder	Pololu	4844	$36.95	1	$36.95
Clamping Collar	McMaster-Carr	6698K52	$8.00	2	$16.00
Knob	McMaster-Carr	5532T35	$3.25	1	$3.25
Low-friction Sleeve Bearing	McMaster-Carr	1677K325	$1.28	1	$1.28
Nylon Thumb Screw	McMaster-Carr	94955A040	$4.08	1	$4.08
Shaft	McMaster-Carr	6061K101	$5.66	1	$5.66
Thumb Nut	McMaster-Carr	92815A115	$2.82	1	$2.82
Support Shaft	*Machined Part*	mp	$0.00	2	$0.00
0.5-in Standoff	*Machined Part*	mp	$0.00	1	$0.00
Actuator Supports	*Machined Part*	mp	$0.00	1	$0.00
Back Plate	*Machined Part*	mp	$0.00	1	$0.00
Base	*Machined Part*	mp	$0.00	1	$0.00
Clamp	*Machined Part*	mp	$0.00	1	$0.00
Curved Plate	*Machined Part*	mp	$0.00	1	$0.00
Motor Support	*Machined Part*	mp	$0.00	1	$0.00
Rotational Plate	*Machined Part*	mp	$0.00	1	$0.00
Stationary Plate	*Machined Part*	mp	$0.00	1	$0.00
Top	*Machined Part*	mp	$0.00	1	$0.00
Internal Gear	*Machined Part*	2696N6	$0.00	1	$0.00
Steel Threaded Rod	*Machined Part*	90322A654	$0.00	1	$0.00

Assembly Plan

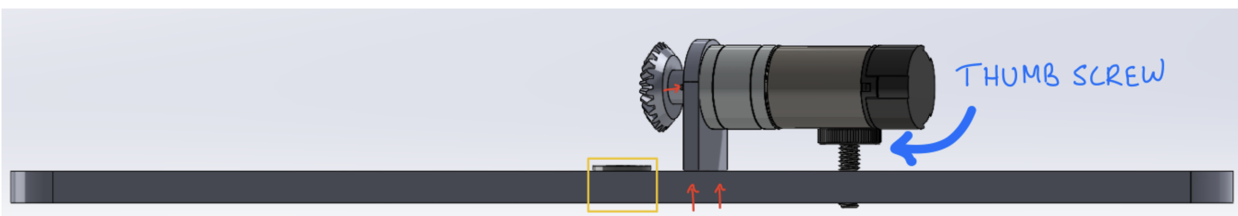
Instructions for assembly are detailed here. In any labeled CAD drawings the colors follow this key: blue = part, red = fastener, yellow = press fit.

Base

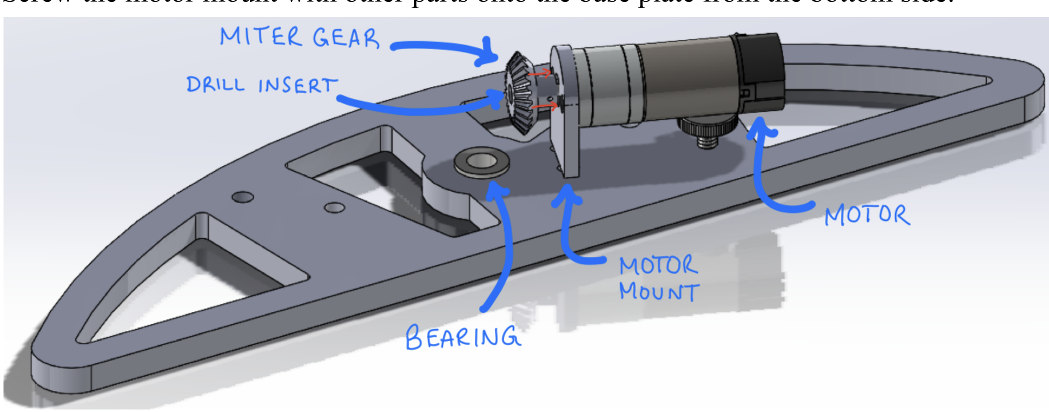
Materials: Base Plate, Motor Mount, Rotary Motor, M3 Screw (x2), Drill Insert, Miter Gear (Motor), Thumb Screw, 1/16” Spring Pin, Sleeve Bearing

Tools: Screwdriver, Allen Key Set (image below), Arbor Press

1. Press the sleeve bearing into the center hole of the base plate.
2. Screw in the thumb screw to the base plate as shown with the head protruding from the top.



3. Press the drill insert into the miter gear.
4. Place the motor into the motor mount and press spring pin through rotary motor, drill insert, and miter gear.
5. Screw the motor mount with other parts onto the base plate from the bottom side.

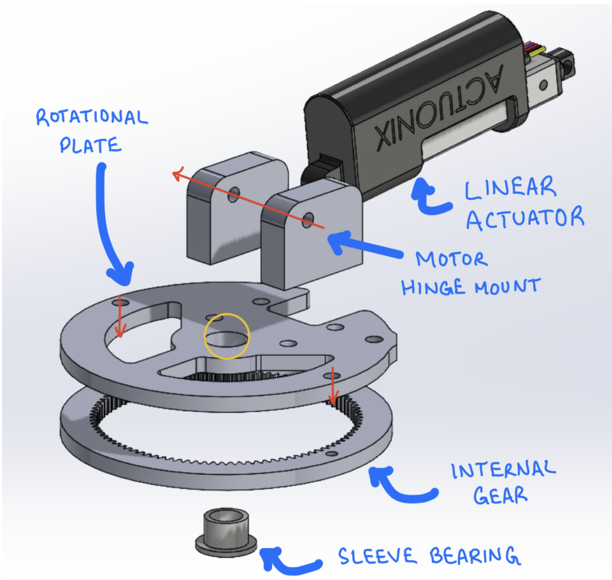


Central Column Mechanism

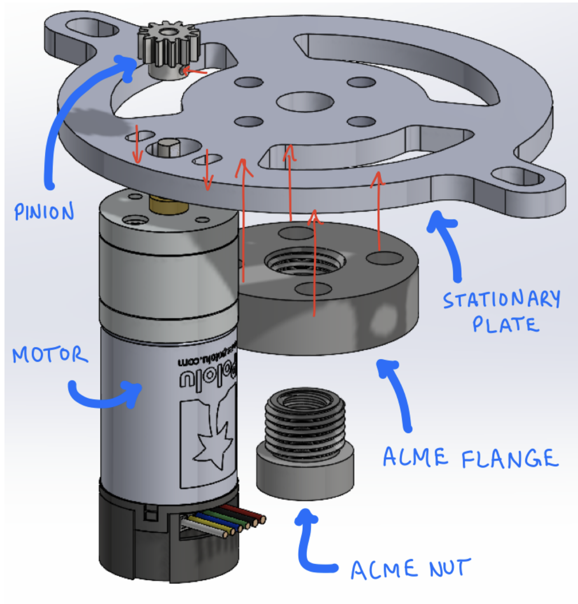
Materials: Threaded Rod, Knob, ¼" Spring Pin, Clamping Collar (x2), Sleeve Bearing, Rotational Plate, Motor Hinge Mount (x2), Linear Actuator, Internal Gear, Bushing, Stationary Plate, Acme Nut, Acme Flange, ¼-20 Bolts, Miter Gear (Lead Screw), Spring Pin, Pinion, Rotary Motor, M3 Screw (x2), Washer (x2), 1/16" Spring Pin

Tools: Screwdriver, Allen Key Set, Arbor Press, Loctite

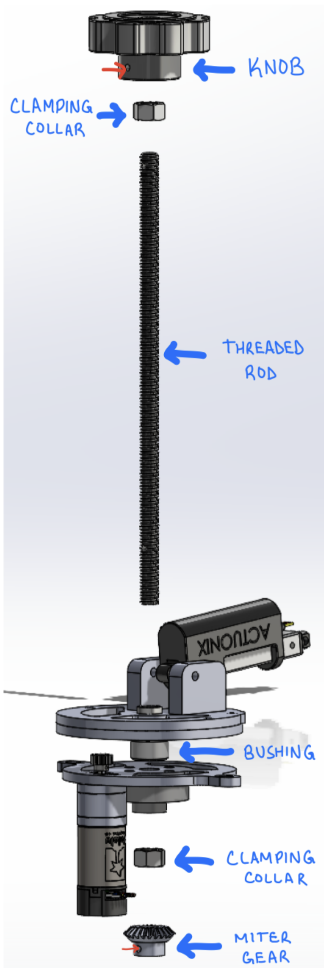
1. Thread a clamping collar on from the top of the threaded rod (side on which the hole is closer to the end).
2. Then place the knob on the top end and press a spring pin to secure it.
3. Press fit a sleeve bearing into the center hole of the rotational plate.
4. Attach the motor hinge mount for the linear actuator onto the rotational plate via screws from the bottom side.
5. Attach the linear actuator with a bolt and nut to the rotational plate.
6. Screw the internal gear onto the rotational plate.

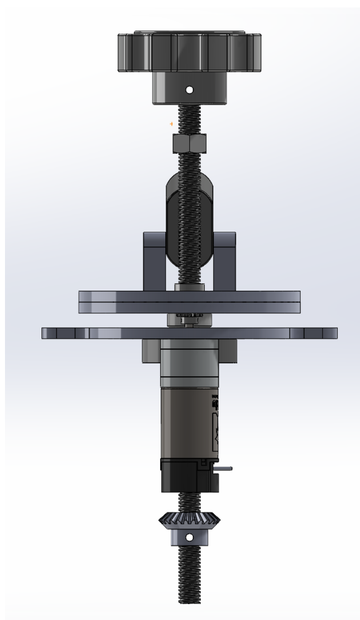
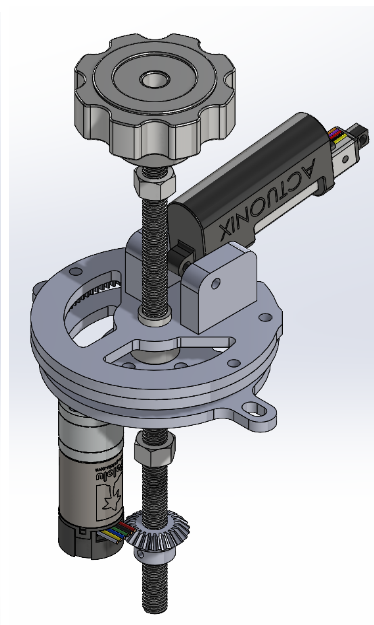


7. Slide the rotational plate with all of its attachments onto the threaded rod from the bottom end.
8. Slide the bushing onto the threaded rod from the bottom end.
9. Use Loctite to fasten the acme nut and acme flange together.
10. Bolt the acme flange to the stationary plate.



11. Slide stationary plate with acme nut attached onto the threaded rod from the bottom side.
12. Thread second clamping collar onto the bottom end of the threaded rod. (The clamping collars are tightened when the front plate is attached before the back plate is to identify the upper and lower limits for the entire stage)
13. Slide miter gear onto the bottom of the threaded rod and press spring pin through the miter gear and bottom hole of the threaded rod.
14. Slide the motor through designated holes on the rear of the stationary plate and fasten onto it with screws.
15. Slide the pinion onto the motor and press spring pin through both of their shafts, and align pinion with internal gear.

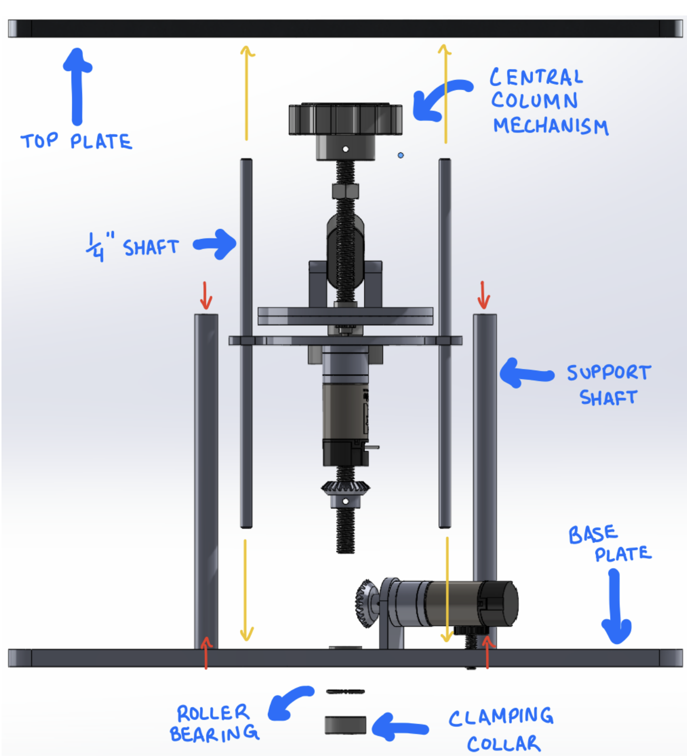


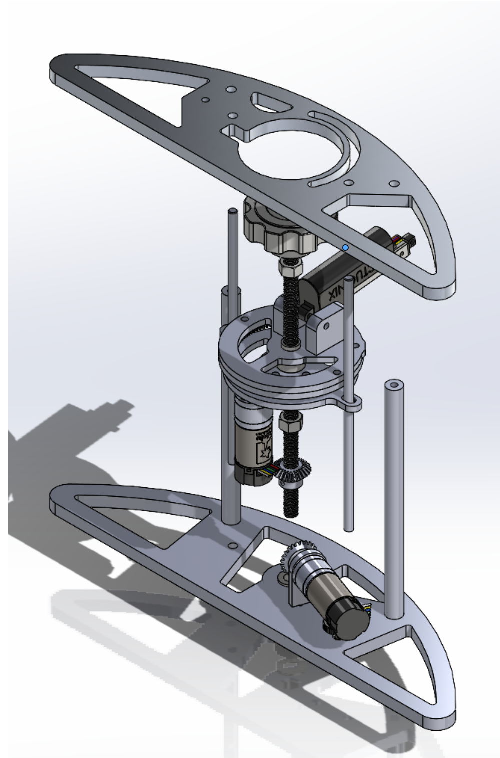



Full Stack
Materials: Central Column Mechanism, Base Plate, Top Plate, ¼" Shaft (x2), Support Shaft (x2), Roller Bearing, Clamping Collar

Tools: Arbor Press, Allen Key Set, Wrench

1. Screw into the two support shafts through the base plate.
2. Press the two ¼" shafts into the top plate and base plate simultaneously until the top plate touches the support shafts.
3. Screw into the two support shafts through the top plate.
4. Slide the top end of the central column mechanism (knob end) through the large center hole in the top plate to align the threaded rod with the center hole in the base plate, and slide threaded rod through the base plate hole with sleeve bearing so the miter gear rests on the bearing.
5. Slide the roller bearing onto the bottom end of the threaded rod below the base plate.
6. Thread the clamping collar onto the bottom end of the threaded rod until the roller bearing is secured, and fasten the clamping collar to the threaded rod.

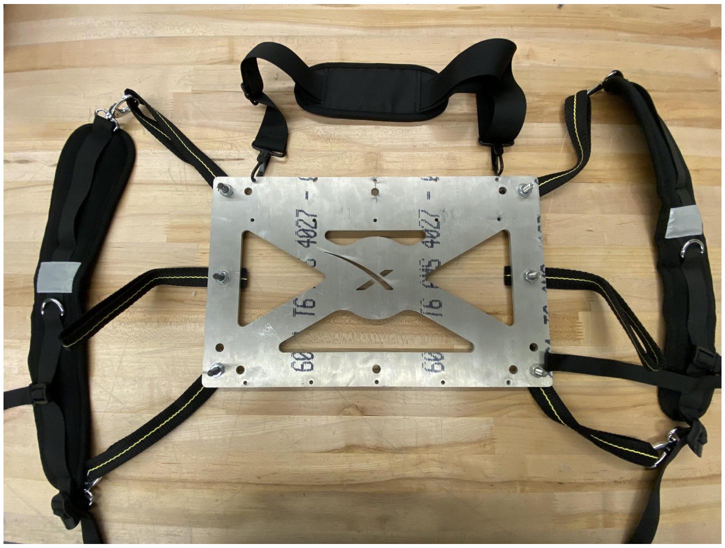




Attaching Straps to Back Plate
Materials: Backpack Straps, Hook Strap, Bolt Straps (x6), 5/16-18 Bolts (x6), 5/16-18 Hex Nuts (x6)

Tools: Wrench

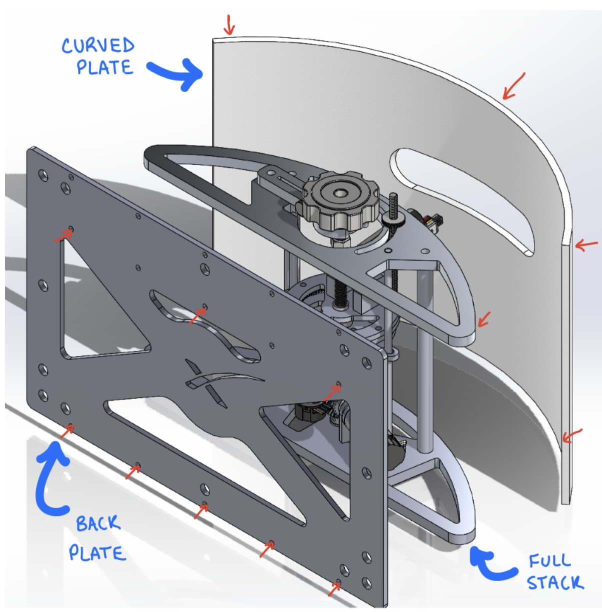
1. Place bolt straps at six locations designated on the perimeter of the back plate.
2. Secure bolt straps to back plate with bolts and nuts. (nuts should be on the interior and bolt straps on the exterior of the back plate.
3. Clip backpack straps onto corner bolt straps.
4. Clip hook strap onto middle bolt straps. (Alternatively: connect hook strap to back plate through small holes at top as shown)



Attaching Curved Plate and Back Plate to Full Stack
Materials: Full Stack, Curved Plate, Back Plate, 8-32 Screws (x14)

Tools: Screwdriver

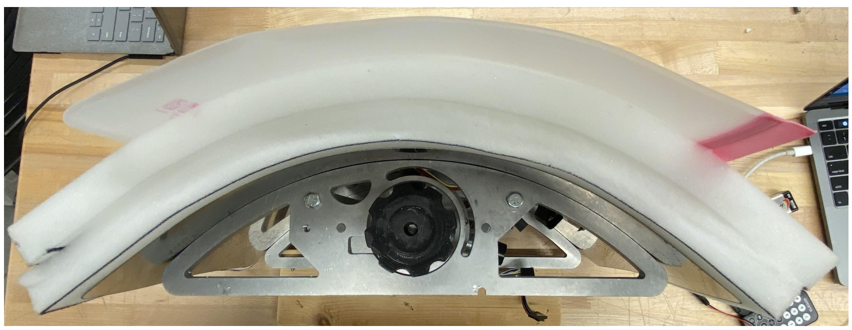
1. Attach the curved plate to the curved edges of the full stack using six screws.
2. Connect electrical hardware to motors and mount it to the curved plate.
3. Using eight screws, fasten the back plate onto the full stack.



Attaching Layers on Curved Plate
Materials: Assembly, Precut 1” Foam Pad (x2), Plastic Bag

Tools: Hot glue gun, Hot glue

1. Plug in and heat up a hot glue gun.
2. Lay assembly on the back side so the curved side is facing up.
3. Use hot glue to attach a pre-cut foam pad to the metal curved plate.
4. Use hot glue to attach a second pre-cut foam pad to the first foam pad.
5. Use hot glue to attach the plastic bag to the second pre-cut foam pad.



Final Assembly:

Authors

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Emily Deming

Emily is a senior at the University of Michigan majoring in mechanical engineering and minoring in computer science. She is from Fenton, Michigan but enjoys living in Ann Arbor. After graduation in December, she will be starting a position at Toyota Motor North America at their Saline, MI facility as a Body Seat Design Engineer. In her free time, she enjoys playing with her cats and watching anything in the Marvel Cinematic Universe.

Alexander Hearn

Alex is a super senior at the University of Michigan majoring in mechanical engineering and finished a minor in computer science. He is from Holland, Michigan but now lives out in the Seattle area where he is working full time for SpaceX on the Starlink Space Lasers team. In his free time, he enjoys hiking, going out with friends, and carpentry.