Design of an Assistive Partial Knee Support Device

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Submitted to:

Professor Selçuk Guçeri

Abstract

This project began as a phase two continuation of a previous MQP endeavor, but with this year's team focusing on reassessing the design choices made by the previous team and developing another design for an active partial knee support device, ultimately taking another path altogether. The prompt for this project was developed due to an identified gap in the market for assistive knee devices that provide active support, but not *full* active support. Most available devices are intended to replace or enhance knee function versus simply providing a helping hand to those with standard knee weakness. Additionally, most devices are not low profile, long-lasting, or affordable to acquire, key elements for the intended users. Research into the biomechanics of the knee joint and human gait as well as conditions that cause knee injury and existing devices guided the design process. A Bowden cable and pulley system, analogous to the knee, was the selected actuation method for the device. The team sought primarily to modify an existing device design, highlighting areas to modify in order to optimize the design for function and usability. Due to the loss of a team member and the restricting circumstances created by COVID-19, the team had limited ability to take desired experimental data, fabricate and test components, and fully flesh out the electronics and computer engineering elements of the design. Due to this the focus of the project shifted to primarily mechanical design, fabrication considerations, and analysis of user needs.

Authorship

The team initially consisted of three mechanical engineering students and one electrical and computer engineering student, and so the elements of the project were split into bio-mechanical design & fabrication and electrical design & controls.

Bio-Mechanical Design & Fabrication

The bio-mechanical part of this project was completed by Emily Bodurtha, Amelia Ring, and Justin Rodriguez. Various background and initial research were accomplished through the joint work of all team members. Amelia and Justin spearheaded investigating the background and forces of the knee joint, including the patellofemoral joint and human gait cycle. Amelia and Emily had extensive experience with Solidworks and CAD, thus both contributed heavily to the design choices and modeling work. Design decisions were made with the input of all members during brainstorming and optimization sessions. Amelia was responsible for the extensive CAD assemblies, while Emily used her manufacturing background to prototype and fabricate the designs.

Electrical Design & Controls

Front-end electrical considerations, circuit design, information on signal noise reduction, and initial component selection was performed by Reed Nowling, who later left WPI. Final power and actuation materials were arranged and completed by Amelia, who attempted to communicate the intended function and methods for these elements, with advice from the advisor, Professor Selçuk Guçeri.

Acknowledgments

We would like to thank, first and foremost, our academic advisor, Professor Selçuk Guçeri, for his support and enthusiasm during our time working on this project. His guidance helped our team overcome the obstacles and limitations we encountered and his encouragement kept us motivated. Also, his personal experience being someone with knee issues allowed us to gain valuable, user-centered input upon which to formulate the goals of our project. We would also like to acknowledge the mechanical engineering department at WPI on the whole, for giving us the opportunity and funding to work on this project. And Barbara Furhman, who helped arrange the parts orders we were able to place during this COVID restricted times. Finally, acknowledgments are due to Reed Nowling, an initial member of the team, who contributed primarily to front-end project work and some content on the power and actuation element, before exiting the project.

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1. Introduction

Whether it be due to injury or age, limited lower limb mobility, specifically surrounding the knee area, is an extremely common occurrence. Those with knee weakness are limited in their ability to stand from a sitting position, ascend and descend stairs, or even walk with ease. The project advisor, Professor Selçuk Guçeri, personally experiences weakness in his right knee which ultimately affects his walking gait and ability to ascend and descend stairs with ease. As such, Professor Guçeri represents a need and market for a device that could provide assistive support for those actions and maintain a low profile and cost. There are many individuals with various mobility issues who could benefit from an everyday, partial support device for their knee joint, especially an affordable and discrete one. Professor Guçeri began a project in 2019 with five WPI undergraduates in the areas of mechanical, robotics, and biomedical engineering; Tina Barsoumian, Jason McGrath, Christina Steele, Kassidy Utheim, and Zachary Zlotnick. This team was assembled to complete a phase 2 continuation of 2019's 'Active Assistive Knee Device,' however, ultimately the project branched off into its own interpretation of the identified problem.

There are two types of assistive knee devices, passive and active. Passive devices are unpowered devices such as crutches, canes, braces, etc., and typically rely on their structure for support. Active devices, on the other hand, provide energy to their system. Most lower limb support devices on the market consist of either structural support only or full to near-full active assistance; contrasting this, this project aims to simply provide a helping hand for motion by supporting only about 10% of the subject's torque at the knee, enough to give a weak knee a helping hand.

The phase 1 design utilized electromyography (EMG) to provide an intuitive control operation coupled with a planetary gear system driven by a pancake motor, whilst representing the motion of the knee as a simple circular hinge joint. The phase 2 team goal was to reassess the design problem and produce another design with the goal of providing partial torque to a user's knee in order to aid those with various mobility limitations in standing, walking, and using stairs. A focus on the size, cost, and lifetime of the device are some of the main goals, as the device is intended to be worn daily and must be affordable to the user.

2. Background

2.1. Phase One Overview & Takeaways

Figure 1: Phase 1 Final Assembly

Above shows the final phase 1 design. The system can be broken into five subsections: the hinge, planetary gears, motor, electromyograph, and sleeve application. Components consisted of two high-powered Maxon motors, 1065 Aluminum blocks, screws, elastic bands, neoprene, and planetary gears.

The hinge subsystem is made up of a male hinge, a female hinge, and the extensions. The male hinge provides housing that facilitates additional torque generation from the planetary gear system. A main element of the male hinge is its circular pocket. This is dimensioned to act as a press-fit with the ring gear and create a barrier for the other gears protruding from the female hinge. The outside of the male also has a mechanical motion stop that restricts a user from over-extending their joint. The female acts as a cover and aids the torque objective by stabilizing the motor system and the planetary gear, and is the largest component in the system. The extensions direct a user's knee joint through the suggested motion of the electromyography and the actuation of the motor. There are a total of four extensions in the device: the upper hamstring and the lower calf extension for both sides of the knee, as shown in figure 1. The hamstring extension is around 50 mm longer than the calf extension due to the need for more power to be generated from the muscle to move the system. The extensions require opposite bending directions during manufacturing in order to mirror each other over the device to account for the natural taper of the user's legs.

For phase 1, the team was provided two high-powered motors to actuate the system through their sponsor, Maxon Group. The motors actuate the system by rotating the planetary gears that then rotate the hinge system. The motor combined with the gear system provides enough torque to achieve the desired 10 Nm of force.

The electromyography was used to monitor the user's muscle movement within the quadriceps to command the actuation of the device and provide appropriate torque to the system. The EMG attaches an electrode to the user's quadriceps to detect when a user is walking, going up stairs, standing up, etc. It sends the signal data to the motor through the wires, and the motor directs the device's movement to aid the user. The EMG was intended to be sewn into a neoprene sleeve to keep it in place and protected.

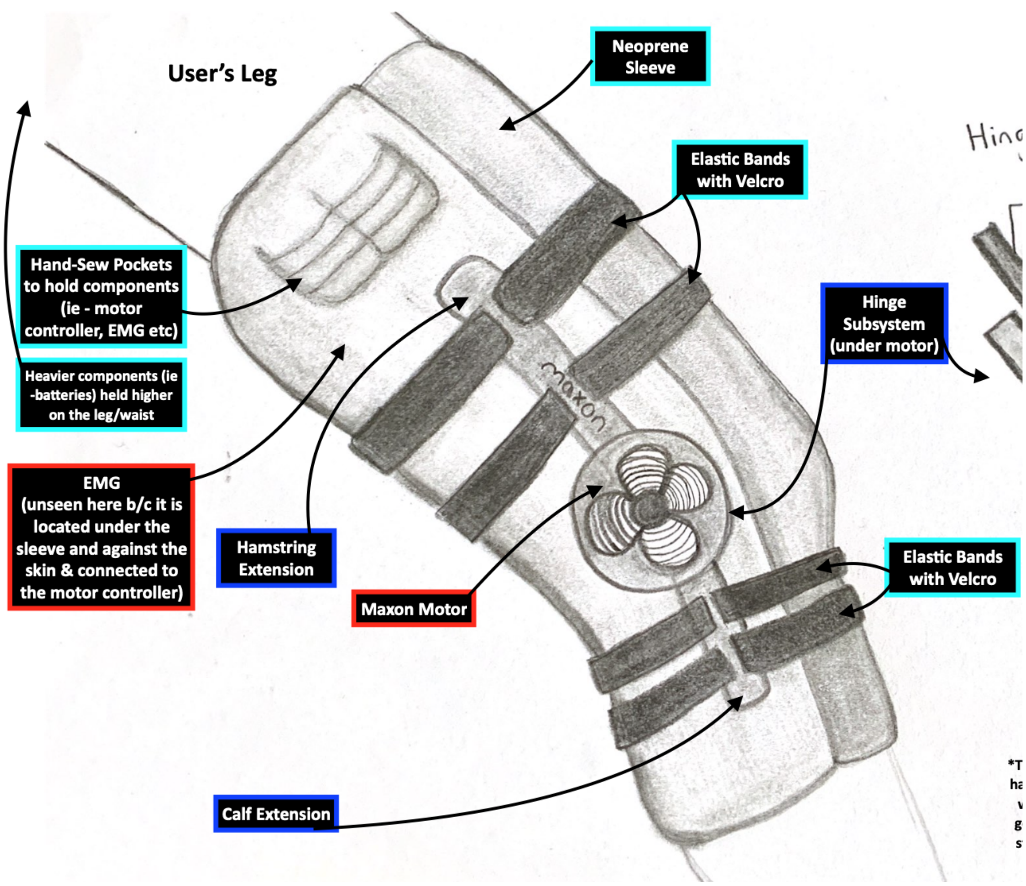

Figure 2: Phase 1 Sleeve Subsystem Assembly Sketch

The above figure shows how the sleeve subsystem brings the whole system together, adhering the full assembly to the user's leg. The sleeve was to be made of a neoprene compression sleeve to increase comfort and will protect the user's skin from the different mechanical and electrical components.

There are several key takeaways and improvements to be made from phase 1. For the gear system, the team was unable to find the exact size and material for the gears due to time and resources. Finding or making more ideal gears would make the system lighter, smaller, and slimmer. The phase 1 team also recommended some changes to the extensions making the system more secure. They also suggest testing at PraceticPoint - a WPI healthcare development and testing facility - to determine if the device truly provides the intended torque.

When testing the device, the phase 1 team also suggests considering the comfort, security, and durability of the device as well as the efficiency of the safety precautions. They also suggested some possible improvements that would result in a smoother and less intrusive experience for the user. A major part of this would be decreasing the weight of the system. Many of the components, especially the gears, can be manufactured out of lighter-weight polymers, and a much smaller, lighter motor should be utilized possibly in a pancake motor fashion. This would create more uniform motion and reduce the amount of lopsided stress on the device. It is also believed that a lower overall weight would reduce the required amount of torque for the motor. Another improvement could be adding further safety factors and ensuring proper and comfortable wearing of the device. Adding a circular spring mechanism between the hinges would also slow the device down as it reaches its mechanical stop which would store energy at the minimum angle of flexion. An interrupt can be used to check if the angle of flexion has exceeded the boundary, which can solve the issue of the device causing an irregular gait. If the boundary is exceeded, the device will allow free rotation until the weight acceptance state is recognized again. Phase 1 determined that their device would need to be sold for around $10,000, so ideally the cost of the device would also be lowered.

2.2. Biomechanics of Lower Limb Motion

As with all engineering endeavors, in order to address the problem, a deeper understanding of the topic must be gained. In this case, one area that requires a thorough background is in regards to the biomechanics of lower limb function. This includes understanding how the knee facilitates motion and in what ways the body moves while walking, standing, and using stairs. Understanding these elements provides a greater perspective and understanding of the nature and impact of the issue as well as what needs to be accounted for in order to solve it. The details of the patellofemoral joint provide inspiration for solutions and knowledge about what kind of forces need to be invoked and protected against in an assistive device. The steps of the human gait and knee flexion will provide useful context for when forces are experienced and at what intervals the device should activate.

2.2.1. The Patellofemoral Joint

The patella, more commonly known as the kneecap, is a unique piece of the body. It is a rounded triangle-shaped bone with the point of the triangle facing down on, making contact with, the trochlear groove of the femur; both contact points are lined with cartilage to disperse joint reaction forces and reduce friction (see Figure 3) (Loudon, 2016). The patella is what's known as a sesamoid bone, small bones that develop inside tendons at joint locations, which help to reduce friction and pressure and redirect muscular forces (Gray & Lewis, 1918); as such sesamoid bones actually act as anatomical pulleys. Through a series of connections to the joint with tendons and ligaments, the patella redirects the mechanical force put out by the quad muscle in order to move the lower leg unit.

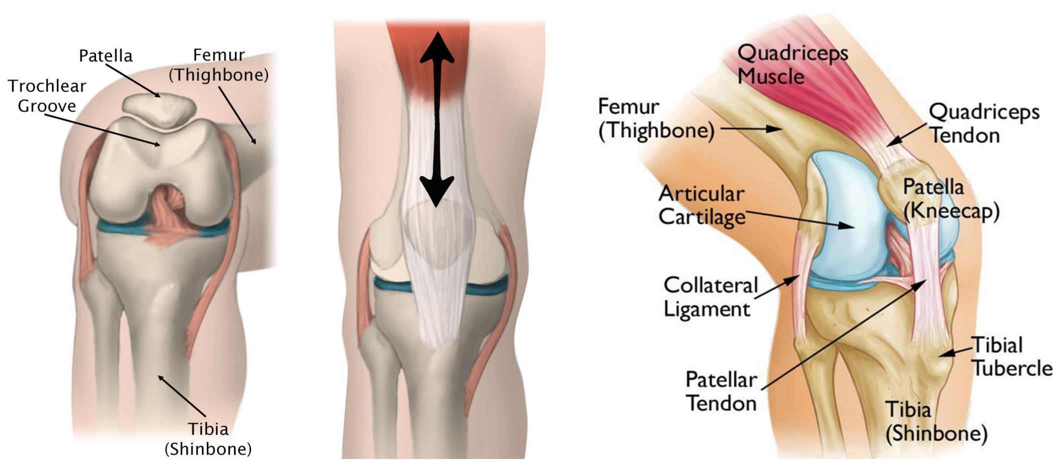

Figure 3: The Patellofemoral Joint (Foran & Grelsamer, 2018)

The patella is bonded to two tendons, the quad tendon, which is also connected to the quad muscle, and the patellar tendon (actually a ligament), which is attached to the tibia as well, together forming the patellofemoral joint (Figure 3). If, for example, the quad muscle contracts, the patella will slide up in the trochlear groove and stretch the patellar tendon. As the patella slides in the trochlea, the area of the patella in contact with the femur changes, ultimately increasing the moment arm and the mechanical advantage of the quadriceps force, and moving the lower leg with ease (Broe et al., 2018). This configuration allows the patella to not only transfer the force from the quad tendon to the patellar tendon to control leg motion, just as a pulley would transfer force but also to magnify this force. As such, the patella is vital to lower limb function. The proper patellar function is most critical in the last 30 degrees of knee extension as here it provides approximately 31% of knee extension torque (Loudon, 2016). It's easy to see how any damage to or weakness in the cartilage, tendons, or patella could severely diminish the knee's ability to create the torque necessary to complete standard activities and handle reaction forces with ease and without pain.

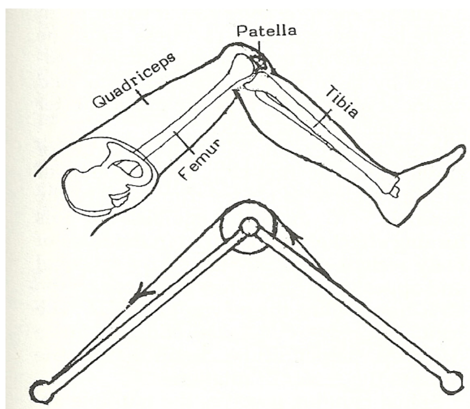

Figure 4: The Knee as an Anatomical Pulley (Schafer, 1983)

Because the patellofemoral joint is transferring and amplifying forces, the knee itself experiences a reaction force. These forces are dependent on both the knee angle and the quadriceps muscle tension (Loudon, 2016). The transfer of the quadriceps force to the patellar tendon at a particular flexion angle creates a compressive reaction force (Figure 5), the magnitude of which will change depending on the magnitude of the force and the angle. As the quadriceps force increases and the knee angle decreases (during knee bending) the corresponding vectors change in size in such a way that the reaction force increases significantly (see Figure 5).

During simple, daily activities, the knee can be put under great joint stress. Given the magnitude and frequency, these forces are experienced, it's no surprise this area sees some of the greatest cartilage deterioration in the body. But this is accounted for, in part with the presence of the cartilage, but also the orientation of the patella during motion. As the patella slides against the trochlear groove during motion, the portion of its contact surface that articulates with the femur changes (Figure 6). The contact area increases throughout degrees of movement; as the contact area between the articulating surfaces increases, the corresponding reaction force is dispersed more completely and joint stress is reduced (Loudon, 2016). A malformity, injury, or misalignment to the patella would alter these contact trends and likely put the knee under undue stress during typical lower limb motion.

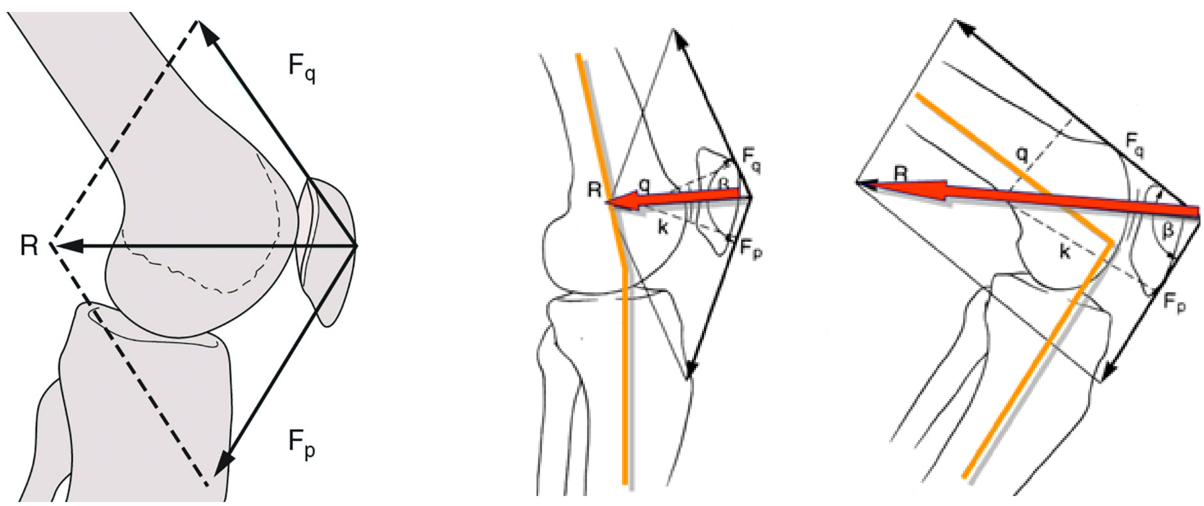

Figure 5: Patellofemoral Joint Reaction Forces (Loudon, 2016 & Robbins, 2015)

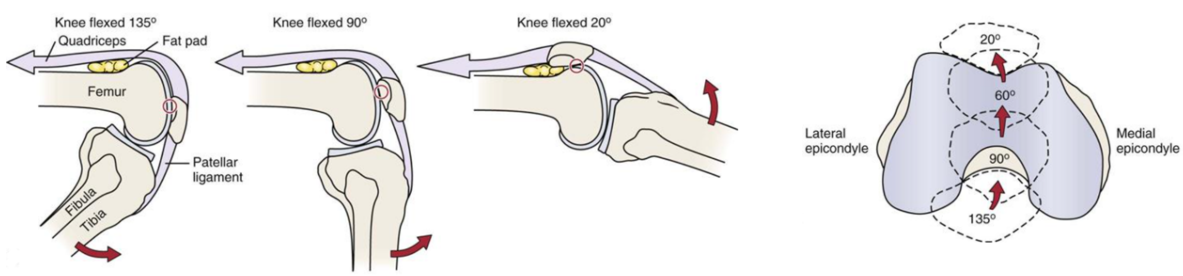

Figure 6: Patella Contact During Flexion (Clinicalgate, 2015)

The patellofemoral joint is a delicate and complex system, one which is absolutely vital to the function of lower limbs. During the subsequent review of human gait, it is key to keep in mind that the patellofemoral joint makes all of the motions possible. An injury, deformity, or simple cartilage deterioration could be enough of an upset to the system to make achieving just the basic 14 - 31 % knee torque contribution of the patella nearly impossible to achieve. Thus despite how simple the gait motion may seem, one without proper patella function would be at a severe loss trying to achieve it in daily life.

2.2.2. The Human Gait

This project will focus on assisting certain common activities that a person uses their lower limbs to accomplish. In order to properly design a device that can assist these movements, it is important to understand the biomechanics behind them. Walking is the process by which a person moves their body toward the desired location. This is accomplished by the foot forming a rigid lever while on the ground in order to propel the body forward with each step. The motion of walking is broken up into two phases that are used to define one of the legs, not both while walking: the stance phase, which makes up 60% of the gait cycle, and the swing phase, which makes up 40% of the gait cycle. The stance phase describes the period of time during the gait cycle when the foot is on the ground. This phase is characterized by 5 distinct stages of motion that the leg goes through during the gait cycle: heel strike, loading response, mid-stance, terminal stance, and pre-swing. Heel strike is the period from which the heel first makes contact with the ground to when the foot is fully touching the ground. Loading response starts the moment the foot is fully on the ground and the leg begins supporting the weight of the body. Midstance defines the period when the leg is fully supporting the entire weight of the body while the other leg is off the ground. Terminal stance occurs when the leg is beginning to support less body weight. Lastly, pre-swing defines when the foot is partially off the ground and about to swing forward. The swing phase consists of initial swing, when the foot fully lifts off the ground, mid-swing, when the leg swings forward, and terminal swing, when the the leg is no longer swinging and the foot is about to make contact with the ground (Katzenschlager & Pirker, 2016). Figure 7 shows the phases of the gait cycle.

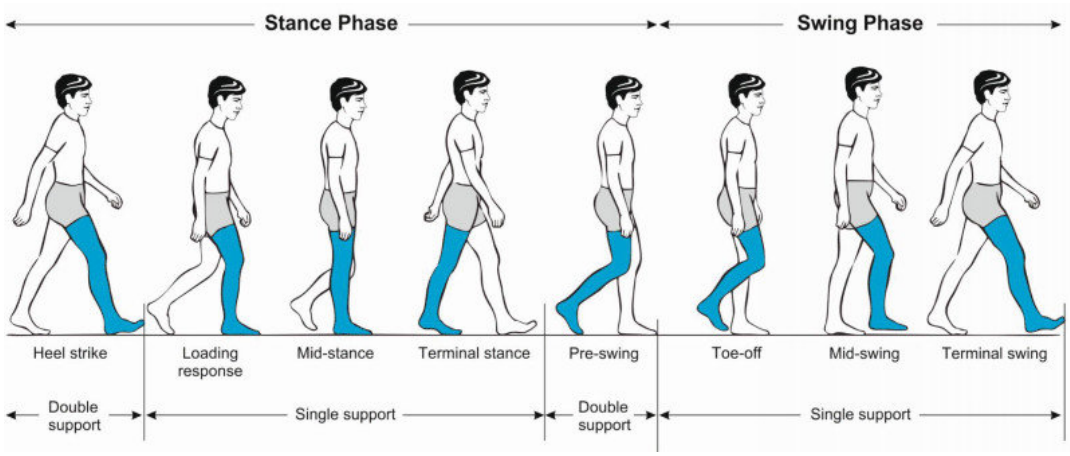

Figure 7: Steps of the Walking Gate (Katzenschlager & Pirker, 2016)

The knee joint is important during the gait cycle because it is the joint that experiences the most torque and it supports the body's weight while walking. When both legs are on the ground at the beginning and end of the stance phase, they both support the weight of the body and are in a "double support" configuration. When only one leg is on the ground while the other is swinging forward, the leg in contact with the ground is in the "single support" configuration and bears the entire weight of the body (Katzenschlager & Pirker, 2016) . This is where issues with the knee typically come into play because if a person's knee is not able to support their entire body weight then they are prone to falling while walking or engaging in other activities.

In addition to bearing weight during the gait cycle, the knee also undergoes a lot of motion, namely extension, and flexion. It serves as the point of rotation between the femur and tibia. During the swing phase, the knee can experience flexion of up to 75 degrees relative to the body's vertical axis. During the stance phase, the knee experiences less flexion, typically 20 degrees or less. At toe-off, which is when the leg transitions from stance phase to swing phase, knee flexion increases to approximately 55 degrees (Morrison, 1970). Figure 8 depicts the full range of motion of the human knee.

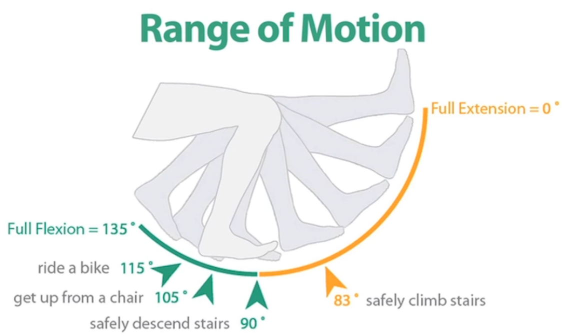

Figure 8: Knee Range of Motion (Flexion and Extension) (Ewing, 2019)

The knee does not undergo much flexion during the stance phase. It is locked during the initial heel strike phase and does not need to flex as it supports the weight of the body. The knee is fully extended during this phase and due to the contraction of the quadriceps muscle as well as other ligaments that tighten and help support and lock the knee. Slight knee flexion occurs after this as the knee moves closer to toe-off. The hamstring contracts causing this slight flexion of 20 degrees at most during this transition from heel-strike to toe-off. One toe off is reached, the knee is no longer fully extended, and begins to enter flexion (Morrison, 1970).

The flexion at toe-off occurs because the leg is about to swing forward. During this initial swing phase, the knee's flexion slightly decreases to 50 degrees but then increases to 60 degrees once the leg has swung in front of the body. The knee then extends beyond 30 degrees as the quadriceps contract and then fully extends and locks once the foot makes contact with the ground (Morrison, 1970).

Similar to the gait cycle, the motion of ascending and descending stairs is cyclical and can be defined by stages that each leg goes through during the cycle. For the stair ascension cycle, these stages are the stance phase and the swing phase. The stance phase consists of weight acceptance, pull up, and forward continuance and makes up and makes up approximately 66% of the stair ascension cycle. The swing phase makes up approximately 34% of the cycle and consists of foot clearance and foot placement (Komistek et al., 2010).

The stair descension cycle has similar phases with slight differences. Like the normal gait cycle, the stance phase makes up roughly 60% of this cycle. It consists of weight acceptance, forward continuance, and controlled lowering. The swing phase makes up roughly 40% of the cycle and consists of leg-pull through and foot placement (Komistek et al., 2010). Figure 9 shows both cycles.

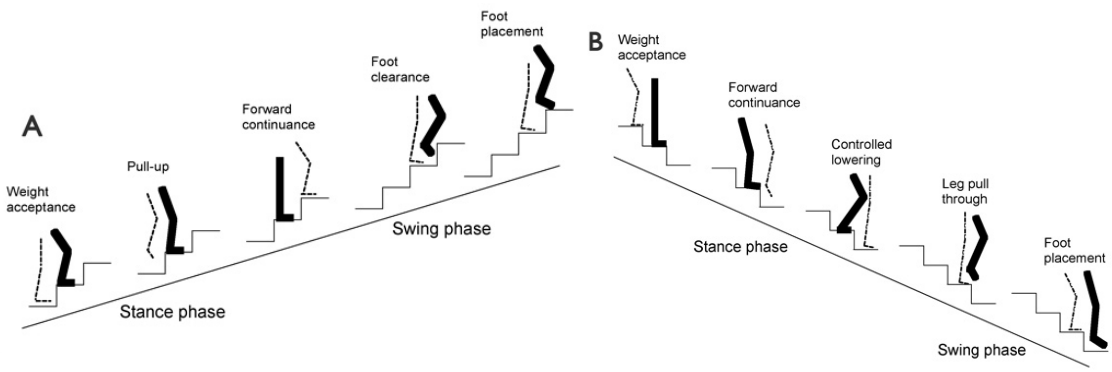

Figure 9: The Stair Ascension Cycle (A) and Descension Cycle (B) (Novak et al., 2010)

While ascending and descending stairs, the knee undergoes a great deal of flexion and experiences significantly higher forces than it does while walking on level ground. During stair ascent, the compressive force on the knee can be as high as 316% of the person's body weight. During the descent, these forces are even higher at 346% of the body's weight (Bender et al., 2010). The knee experiences the most force at the beginning of the stance phase when the muscles around the knee help support the knee and push the body up or down the stairs.

2.3. The User and Their Needs

While the knee is the largest and strongest joint in the body, it is also prone to weakness due damage, which can severely limit the quality of life due to pain and subsequent inability to access the full range of motion of the knee. Knee weakness due to damage typically derives from either injury or overuse. In the case of injury, populations of all ages are afflicted. Common injuries to the knee include sprained or strained ligaments/muscles or torn cartilage (meniscus). Both of these are caused by some kind of blow to the knee or by twisting, bending, or falling incorrectly. Symptoms of these injuries might be pain, swelling, and difficulty walking and they may require significant time and even surgery to fully recover from (Stanford Healthcare, 2021). Injury aside, the other type of damage is typically due to overuse of the body. The most common types of overuse damage are tendonitis and arthritis. Tendonitis is a condition where the tendons become inflamed due to strained overuse of them. A common condition known as 'jumper's knee' is the result of tendonitis of the patellar tendon. Arthritis, and more specifically osteoarthritis (OA) is a condition where the articular cartilage in a joint or joints has worn away over time, leading to pain and stiffness. This is usually caused by significant stress on the joint over time, and is very common at the knee joint, which has been previously shown to experience reaction forces against the cartilage with every motion (Stanford Healthcare, 2021). Osteoarthritis is the developed world's most prevalent joint disease and one of the leading causes of disability due to its devastating impact on hip and knee function. Those who experience knee OA and other knee weakening conditions have to deal with their pain and the effects it has on their lives as their lower limb function, mobility, and productivity are reduced on top of the burden of the cost of medical care. The condition diminishes quality of life in more than one area and affects roughly 25% of the 55+ population, those reported having persistent episodes of knee pain. 10% of adults 55+ had disabling instances of knee osteoarthritis and 25% of those people were considered significantly disabled due to it. Figures for OA have also been increasing over the years, with the increase in obesity being one risk factor highlighted as it greatly increases stress put on the knee (Heidari, 2011).

There is very clearly a large audience of people with varying levels of knee weakness out there. While for some, the right move may involve a knee replacement procedure, many are simply trying to manage their pain and provide additional support to their knee, which is still mostly functional, just weakened. Most people with (or recovering from) knee injury or damage will use some kind of passive brace to support their knee. It is an affordable way to give added support to the weakened area, but not overly effective at helping to perform daily tasks. As will be discussed in the subsequent section, these devices do not actively provide relief to the strain on the knee, and devices that do, do not typically have these users in mind. These fully active devices are typically not low profile, lightweight, or particularly comfortable. They may have large motors, batteries, or electronic connections, and are not overall affordable to the average person. A user like this is seeking a device that is acceptable for 'every-day' use that can be worn comfortably and discreetly; that only supports their missing knee function and won't draw undue attention; that will last all day and be an affordable investment. These are the kinds of features sought to be included in the resulting design for this project. The user needs are a key element and a motivating factor for design decisions.

2.4. Lower Limb Support Devices

With patellofemoral pain or injury being so common, there have been numerous attempts to provide much-needed support to the knee and lower leg, however, there are no commercially available devices that meet the specific goals set for this project. The desired design is low profile, low cost, long-lasting, and with active support which provides the user with support equal to roughly 10% of needed knee torque. With a deeper understanding of the human gait, patellofemoral function, and audience needs, a review of currently available or researched knee support devices and types of suitably actuation devices can be conducted in preparation for the design ideation phase.

2.4.1. Review and Classification of Current Devices

While the concept of assistive knee devices is not new, there is still plenty of room for development and innovation in the area. Disability, injury, and weakness from old age have robbed the affected of their mobility and ability to perform daily tasks with ease. There is most certainly a need and market for devices that provide support to these individuals in order to help them regain their independence. Devices like these are often categorized based on the type of support they offer and the intentions of the design.

The most common form of categorization for assistive devices is whether they are passive, active, or a combination device. This refers to the type of support they offer the user. Passive devices do not use any kind of powered system to alleviate the strain faced by users, they instead use their structure to accomplish this; think canes, crutches, braces, and compression sleeves. Even within this category, the devices offer different types of support. Crutches and canes alleviate the user by taking the weight off of the troubled area while braces and compressions sleeves focus more on encouraging healing and preventing further injury or inflammation by holding bones or muscles in place during activity. Nearly every cheap and easily accessible support device is a passive one; the majority of support devices handed out by doctors or purchased at drugstores are passive devices.

Active support devices, on the other hand, use a powered system to achieve their goals. By inputting energy into the system, the device is actively working to reduce the strain on the user. Some current knee support devices using an active system are the famous HAL (Hybrid Assistive Limb) device, BLEEX (Berkeley Lower Extremity Exoskeleton), or Honda's Walking and Body Weight Assist Devices which all use active supports to aid or improve a person's walking capabilities (Berkeley Robotics…, n.d.; Cyberdyne, n.d.; & Honda, n.d.). The HAL device is billed as the world's first robotic medical device to improve a patient's ability to walk. It uses non-invasive EMG sensors to read muscle signals in order to direct their single motor to perform the desired action; then the motor provides a force at the knee, aiding the wearer in their motion. The development of active devices is good news for the future of increasing mobility, but it also means the devices will likely require a lot more hardware, bulking it out and increasing the weight and the limitations that a power supply brings in terms of use time. For an active device, HAL is remarkably low profile, but not quite enough to be concealed during daily use (see Figure 8), it also suffers from battery life limitations. It has an unfortunate battery life of only 2 hrs.

Figure 10: People Testing HAL in the Streets of Japan (Harding, 2009)

Some device development has been made with the concepts of a semi-active semi-passive device such as the 'variable non-active interval' knee support device coming out of New Zealand in 2018 (Babin et al., 2018) or the 'quasi-passive lower limb exoskeleton' from Singapore 2 years prior (Collo, 2016). The non-active interval device chose the unique strategy of using active support with a triggered non-active period where the device only provides passive support, in order to reduce power consumption and avoid disrupting the user's gate. Devices utilizing this semi-active semi-passive design have the added benefit of energy efficiency, meaning they can reduce their power supply sizing and/or be used for longer periods of time since they do not rely entirely (or primarily) on powered support. The 'quasi-passive lower limb exoskeleton' takes advantage of this lesser power requirements and designed a more discrete, lightweight, and affordable exoskeleton (Figure 9), more in line with the goals of the project, although even this piece is not complete; there is still plenty of work off of devices like these to be made.

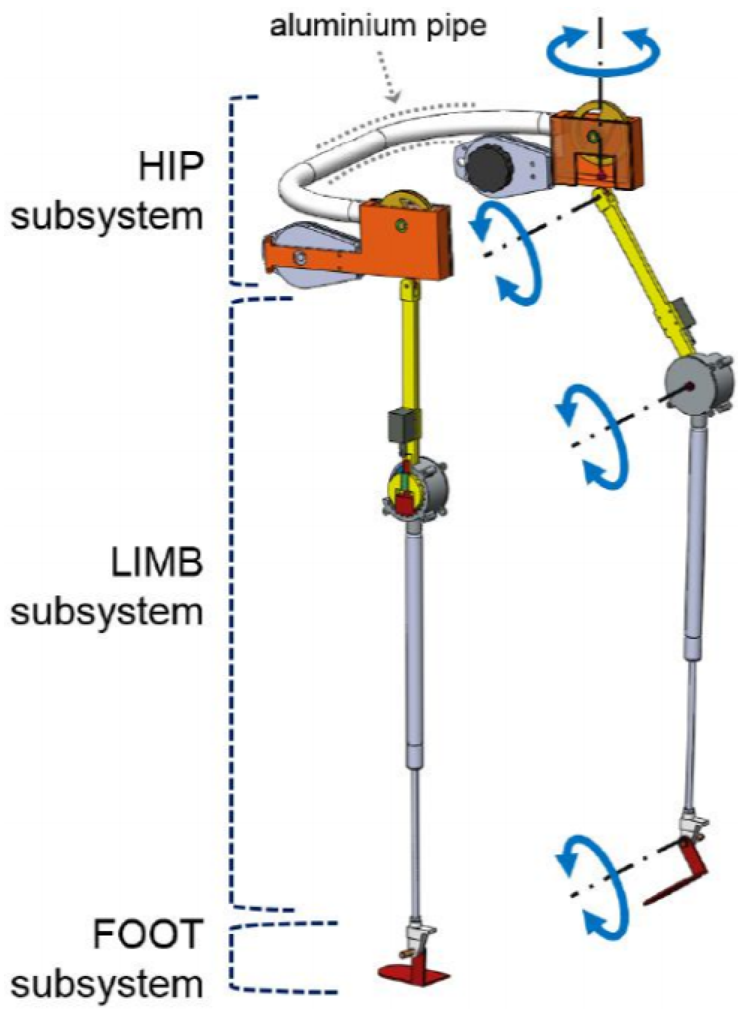

Figure 11: Quasi-Passive Lower Limb Support Exoskeleton

The motivation or goal behind the design can also serve as categorization criteria. Different motivations for the development of the design can impact the criteria and design choices made by the team behind it. Support devices like these are typically designed for the purpose of either rehabilitation assistance (the shorter term as the subject's strength will come back) or for those with a chronic disability or experiencing the effects of old age (long term use with strength likely not returning). Some work in the area is also focused on devices meant to completely replace the body's contribution, bordering on the prosthesis, as well as those used to enhance the abilities of an already healthy individual, but this will not be the intended use of the proposed project design. An example of such motivations is seen in the HAL and BLEEX devices. HAL is designed for rehabilitation and strength replacement for those with certain spinal cord injuries. This means HAL intends to support and replace the majority of hip and knee contributions, mainly aiming to re-imbue the user with their strength, controlled by their own EMG signals. The BLEEX device (Berkeley Lower Extremity Exoskeleton) was designed with the purpose of providing soldiers and emergency personnel with the extra strength needed to carry heavy loads and preserve their strength and endurance. As such their methods are far too intense and the results too bulky for the purposes of this project (see Figure 10). The presented solution is intended to alleviate some of the burden of standing, walking, climbing stairs, and other daily activities, but not to fully replace the work done by the body. In other words, the device is meant for frequent use in everyday tasks by those with the need for additional support, due to weakened physical abilities. This means the design goals will focus more on keeping size, weight, and costs down, maximizing the life of any power source, and the durability of the design for long-term use.

Figure 12: The BLEEX Device (Berkeley Robotics & Human Engineering Laboratory, n.d.)

There are numerous current support devices, more still than examined, and plenty of research being done into further developing these for public availability. The classification of knee support devices is important as it helps define precisely the function of the project design and prototype, but also to demonstrate the current need for such a device as proposed by this project. Totally passive devices provide physical stability at a low cost and have the low profile qualities desired, but are incapable of providing the type of active torque relief necessary. On the other hand, most active devices in development are far more intense than necessary for this project. Many of them are designed to fully or near fully replace the human torque contribution, while the goal here is to replace only a fraction of torque to give a 'helping hand.' Additionally, these active devices are bulky, expensive, and typically have atrocious battery life. Some semi-passive/semi-active devices are attempting to work around these issues, and while a combination device is not the specific aim, their work towards reducing device size, cost, and energy consumption is something that is valuable to the direction of this project.

2.4.2. H. Park et al

One type of active support device not yet discussed is the soft wearable device designed by Hyunjun Park, Jian Lan, Jiwen Zhang, Ken Chen, and Chenglong Fu (H. Park et al, 2019), which was of particular interest to the team and so is worth review. Many of the previously active and quasi-active devices reduce the knee motion to a single hinge using a motor mounted at the knee to support the user; the researchers here chose a more complex, pulley-based system to support the joints. As was previously discussed, the patellofemoral joint acts as an anatomical pulley. The patella transfers the force from the quad tendon to the patellar tendon while increasing the moment arm and maximizes output torque through stages of extension/flexion. These anatomical elements can be replicated mechanically for the purpose of developing knee support which is something the team recognized, whether intentional or not, in the H Park et al project.

The researchers of this project took advantage of this mechanical-anatomical relationship in their design. Their knee support device effectively replaces all key elements of the patellofemoral joint with mechanical counterparts: the quad muscle is replaced by a Bowden cable, which can transfer mechanical force; the patella is a system of pulleys, and the quad and patellar tendons become a single cable called the 'tendon' cable. H. Park et al settled on a fixed and moveable pulley system to achieve the motion as well as increase the mechanical advantage of the Bowden Cable force, just as the patella does for the quadriceps force. The concept is appealing given its parallels with the natural system our body uses as well as its potential for a low profile design. As can be seen in Figure 13, the design focusing on small pulleys and flexible cables means that an exclusively rigid linked exoskeleton design is not necessary for function, it also means that the control hardware for the devices is not required to be situated explicitly at the knee. With this flexibility, the design has great potential to not only support the knee during extension and flexion but to do so with a much lower profile and comfortable design than many other devices have not been able to achieve thus far.

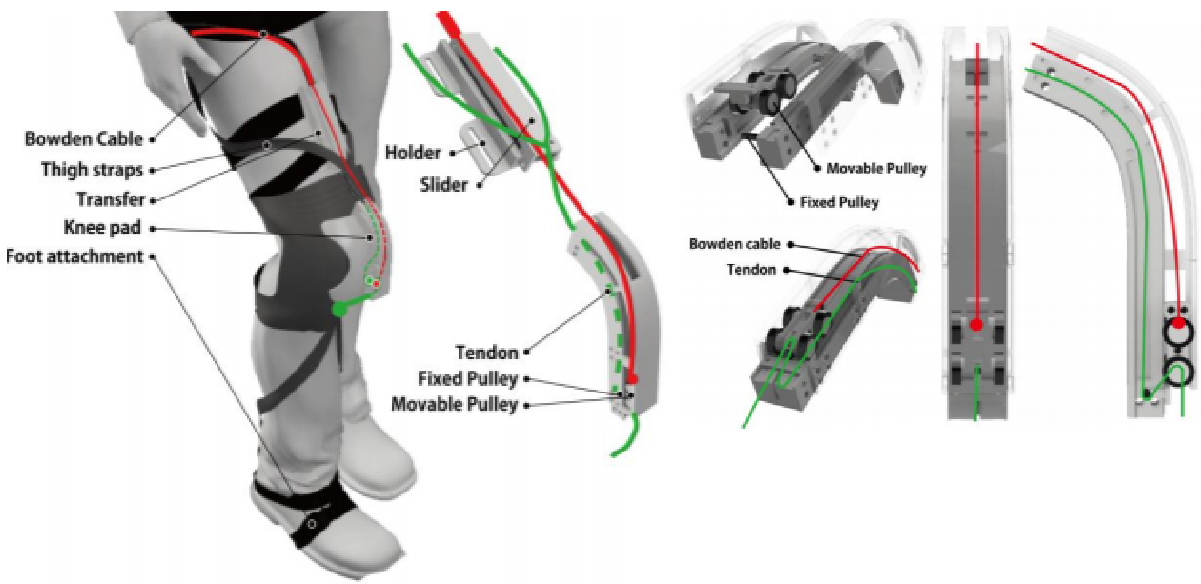

Figure 13: 3D Model of the H. Park. et al Device

While an extremely promising start, very similar to the goals of this project, the H. Park et al design is not without room for improvement. The research concluded that the H. Park et al design has the ability to support knee and hip extension and flexion, so there are not currently any logical fallacies with regards to the function of the knee support. The identified issues concern more the optimization of the device for users. One such issue is related to the bulkiness around the knee pad and pulley track. The knee pad with a hard outer shell supports the track. The inner part of the knee pad consists of fabric that surrounds the knee and a cushion over the knee that supports the knee and protects it from the reaction forces caused by the Bowden and tendon cables (recall Figure 5). The track, which is also a rigid material, is on top of this shell.

The outer shell and track causes the device to stick out a significant amount from the knee. While slimmer than many other designs, it still has unnecessary bulk that makes this device more inconvenient and obtrusive to wear for everyday use.

Another concern noted with the design is the amount of fabric and straps used to secure the device to the body and distribute force throughout the wearer's leg. It features a padded velcro waist belt, thigh straps, a knee pad, and a foot attachment composed of additional straps that must be secured around the foot of the wearer. The device is made for people with existing mobility issues, many of them being older. People with severe mobility issues especially may have trouble putting the device on and making sure all of the straps and other components are secured, fitted, and in the correct place. Additionally, the complexity of these components could make putting on the device confusing for any wearer. A streamlined design that is easier to put on without assistance would make the device more accessible and user-friendly.

Figure 14: Photograph of the H. Park. et al. Prototype

2.5. Detecting, Processing, and Classifying EMG Signals

Something from phase 1 the team wanted to utilize were electromyography (EMG) sensors to gather data about leg motion and inform the timing for the actuation of the Bowden cable. Due to the departure of our ECE specializing team member, many power and actuation elements never fully came to fruition. One such element was the use of non-invasive EMG sensors to gather data from the user. The EMG sensors are key to the actuation of the mechanism; they read the electrical signals from the body in order to time the motor for the desired motions. They also help provide important data about leg motion and gait timing which helps inform design decisions. As such, understanding their function and how to obtain an accurate signal is key background regardless of the realization of this component.

Electromyography (EMG) refers to the muscular electrical signals controlled by the nervous system which are produced during muscle contractions (Chowdhury, 2013). The contraction of a muscle is triggered by an electrical impulse sent from either the brain, brain stem, or the spinal cord (Fleischer, 2008). This impulse then travels through the nervous system to the specific muscles that are to contract. These electrical impulses can be read through the use of needle or surface electrodes, with needle electrodes outperforming surface electrodes due to the added interference through the layers of the skin surface electrodes experience. In addition, surface electrodes suffer more from interference effects due to the movement of the electrode on the surface of the skin, but have the significant benefit of being a non-invasive measurement system. While filtering out this noise is a key problem with surface electrodes, in recent years, the use of surface electrodes to read EMG signals has been increasing as we find new ways to increase the signal-to-noise ratio of the electrodes (Chowdhury, 2013). Needle electrodes are invasive and require the insertion of one or more wire into the muscle that is being read, therefore the non-invasive surface electrodes are ideal for ease of use and patient comfort.

EMG signals are received by the muscle about 20-80ms before the muscle begins to contract, giving ample time for signals to be read and reacted to for real-time addition of torque to specific joints (Fleischer, 2008). As surface electrodes have become more studied to reduce interferences for use in moving applications, there has been an increasing number of exoskeletons being produced and studied utilizing surface electrodes, such as the aforementioned HAL and the phase 1 design. One of the primary sources of interference comes from the movement of the muscles, the interface between the electrode and the surface of the skin, and the skin itself moving between layers as the muscle contracts (Chowdhury, 2013). This causes an electrical interference around “1–10 Hz and has a voltage comparable to the amplitude of the EMG” (Chowdhury, 2013). This can be reduced by utilizing recessed electrodes, an electrode with a layer of gel between it and the skin, reducing the skin impedance by roughing the skin or removing top-layer skin with tape, or filter algorithms to remove the noise dynamically (Chowdhury, 2013). Conforto et. al. tested four different filter methods and found that the use of an adaptive filter based on Meyer wavelets gave the best time accuracy while keeping all necessary information to remove movement artifacts in the EMG (Conforto et. al., 1999).

Electromagnetic noise throughout the body also plays a significant role in interfering with EMG signals. This includes electromagnetic noise introduced to the body from the environment, internal noise of the human body, cross-talk of EMG signals, and the inherent instability of EMG signals (Chowdhury, 2013). The human body is constantly absorbing EMF radiation from a multitude of sources, compounded with interference introduced by the amount of tissue between the electrode and the target muscle (Chowdhury, 2013). The interfering signal caused by power lines (power line interference, PLI) generates a 60Hz (50Hz in some parts of the world) EMF signal coming off of the body. This can be removed with a simple high-pass filter as EMG signals are below this cutoff. In addition, cross talk occurs when the electrode reads an EMG signal from a muscle group that is not being targeted (Chowdhury, 2013). The effect of cross-talk can be minimized with proper electrode sizing and spacing to ensure the proper muscle groups are targeted (Chowdhury, 2013). However, EMG signals are inherently difficult to interpret and unstable due to the quasi-random behavior of signal distribution throughout the muscle group (Chowdhury, 2013). Electrical noise is introduced in a quasi-random-like behavior when motor units are fired. The number of motor units, density, the location from the electrode, and much else affect the amplitude and frequency of this noise, leaving frequencies 0 to 20Hz largely unusable while motor units are firing (Chowdhury, 2013).

The single largest source of interference for reading an EMG signal with a surface electrode is the electrocardiograph (ECG) signal, the electrical impulse responsible for keeping the heart beating (Chowdhury, 2013). The effects of the ECG can be reduced by common-mode rejection by measuring the ECG across the axis of the heart (Chowdhury, 2013). This removes the ECG signal from the EMG signal based on the measured ECG signal. However, it has been found that a high-pass filter with a corner frequency around 100Hz essentially removes noise from the ECG (Chowdhury, 2013). Lastly, ECG signals have been found to not be visually detectable at muscle contraction greater than 25%, leaving room to be able to read the full signal without much interference from the ECG signal (Chowdhury, 2013).

3. Goals & Methodology

3.1. Goals

The major goal of this project is easy to define: to fully design an active knee support exoskeleton device, capable of supplementing approximately 10% of maximum knee torque to the average user. While this can be put rather succinctly, numerous sub-goals were set in pursuit of the overarching goal. Some of these sub-goals pertain primarily to the mechanical and biomechanical elements of the project while others are focused on electrical and controls, and so were often performed simultaneously by different group members. Subgoals include:

1. Determine maximum and average torque values through the knee during sit-to-stand, walking, and climbing stairs situations of various users.
2. Fabricate designed components, for initial testing and proof of concepts.
3. Utilize surface electrodes to monitor the activation of the quadriceps femoris muscle group with an error rate of less than 5% throughout the operation.
4. Evaluate novelty, marketability, and intellectual property concerns.

Even still numerous tasks were involved in the completion of each of these sub-goals. Unfortunately, due to the lost team member and circumstances presented by COVID, the subgoals and tasks had to be modified to accommodate changes. The initial list of tasks can be found in Appendix A. Items in red are those that had to be put on pause or otherwise adapted due to the unique situation.

As can be seen, the tasks that were halted or restructured were primarily related to the power and actuation design, testing, and some fabrication elements. This is a direct result of losing the ECE team member and COVID restrictions, which made achieving these tasks to a desirable degree difficult. For the many other tasks that were completed, The subsequent sections on the biomechanical design process, fabrication considerations, and electronics and control systems will shed more light on the work completed in each of these areas in pursuit of the goals discussed. Sub-goal 4, evaluation of usability, marketability, and intellectual property concerns will be later addressed in the conclusions section.

3.2. Bio-Mechanical Design Methods

The design of the device should meet several criteria set by the team and based on the previous MQP team's criterion determined with the input of Professor Guçeri. The criteria outline the features the design should have when completed, to be considered a success. The primary criteria are that the design should be able to provide 10% active torque relief to the knee. It should be as lightweight and low profile as possible so it can be worn under clothing with ease. It should also, ideally, have a long operation life, be affordable, and not interfere with the user's natural movements in a negative way. One of the major constraints, or limitations, of the team, was working with a budget of $1,000 provided by WPI. Any design would need to be able to be fabricated within this budget, and ideally below that, as the funds also go towards materials for prototyping and testing. Other limitations include time, the project must be designed and completed within the 2020-21 school year. But, additionally, the COVID situation created further limitations on what in-person meeting, investigation, and fabrications could be completed in this time frame, and so a greater focus on complete design over functional prototype was had. Finally, the loss of our ECE member midway through the project was even further restriction, as it was no longer feasible to achieve certain tasks anymore. Criteria and constraints are vital boundaries to keep in mind at every step of the design process. The team's design process was split into approximately three phases, the investigation, ideation, and optimization phases; the methods for which will be detailed individually below along with the intended fabrication and electrical and control system methods.

3.2.1. Investigation

The first step in the biomechanical design process was to conduct a thorough investigation of the issue, to gain vital knowledge that informs design decisions. Other than conducting the research necessary for the in depth background investigation, the team also conducted a simple force calculation on the leg at key frames during motion. The initial sketches can be seen in Appendix B, however updated versions are shown in the figures below. The purpose of drawing these diagrams was to, for one, garner a better understanding of what was actually happening during each of the three motions, and primarily to investigate what exactly '10% of maximum torque to the average user' really looked like (recall sub-goal 1). Table 1 below shows the variables that are used in the drawings and subsequent calculations along with the average of the sample values used in the investigation.

Variable	Description	Unit	Avg. Value Used
W	Body weight	Newtons (N)	735.00
W_{Upper}	Upper body weight (87.8% of W)	Newtons (N)	645.33
W_{Lower}	Lower body weight (6.1% of W)	Newtons (N)	44.85
L	Length of leg segment	Meters (m)	0.56
r	Approx. distance from knee to W_L center of gravity	Meters (m)	0.19
β	Angle between upper and lower leg segment	Degrees	35 - 180
$\ltimes$	= 180 degrees - β	Degrees	0 - 145
θ	Angle between upper leg segment and body	Degrees	35 - 180
φ	= 180 degrees - θ	Degrees	0 - 145

Table 1: Variable Guide for Force Diagrams

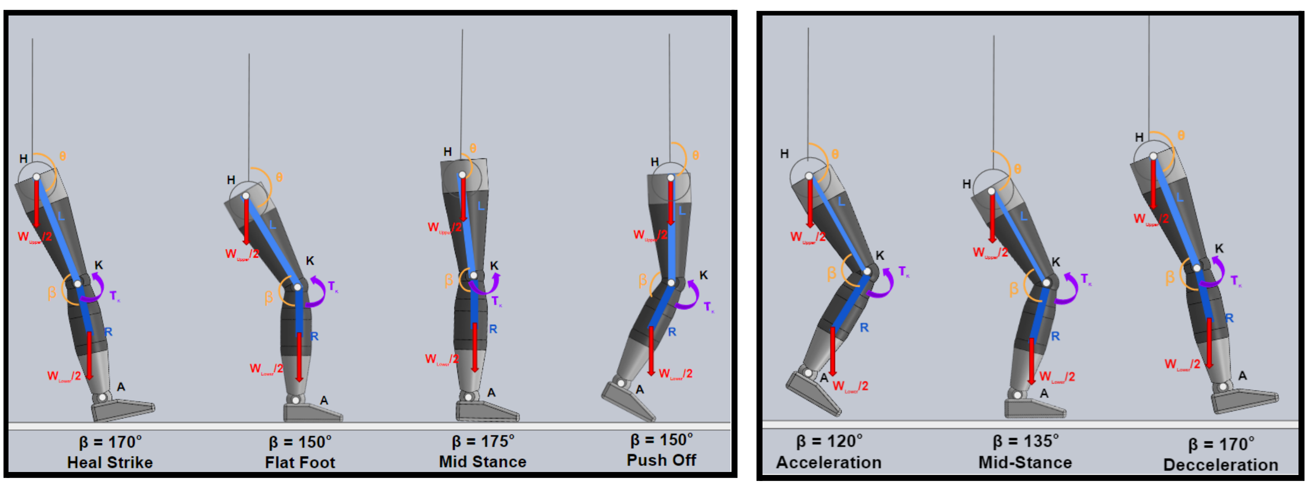

Figure 15: Updated Force Diagrams for Walking Gait (Stance and Swing Phases)

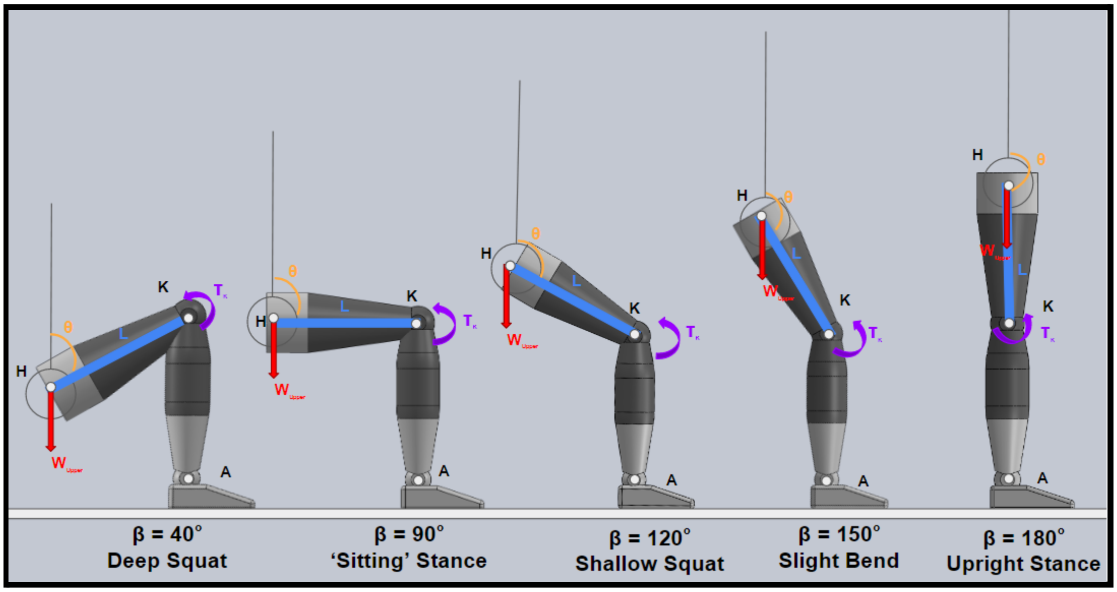

Figure 16: Updated Force Diagrams for Rising from a Squat

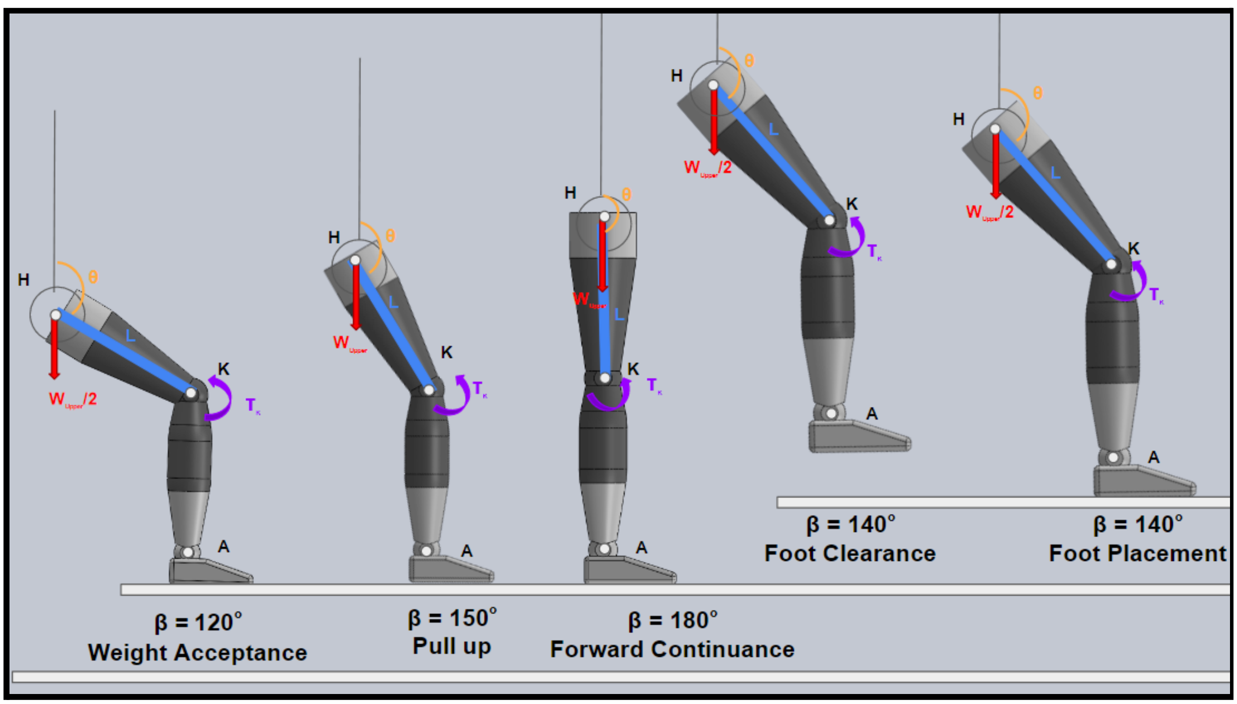

Figure 17: Updated Force Diagrams for Ascending Stairs

While the diagrams are helpful for visualizing the forces, some hand calculations were also done to determine effective knee torque at each interval. Table 2 below shows several variations of the standard torque equation that can be applied at different instances as appropriate. For example, in instances where the leg element is in line with the force, the force contributes no moment, so no torque is produced, as such the equation involving the lower leg is not applicable.

$\tau_k = (L * \frac{W_{Upper}}{2}) * sin(\phi)$
$\tau_k = (r * \frac{W_{Lower}}{2}) * sin(\propto)$
$\tau_k = (L * \frac{W_{Upper}}{2}) * sin(\phi) + (r * \frac{W_{Lower}}{2}) * sin(\propto)$

Table 2: Equations for Single Knee Torque at Various Instances of Knee Motion

Applying these equations to the instances in each of the three motion types, the team isolated which activities were the most strenuous on the knee in question as well as what an average maximum torque value might look like. The team determined that the 'sitting stance' and 'pull up' instances, in the rising from a squat and stair gait diagrams respectively, showed the most strenuous moments. In the sitting stance instant (ie. the moment you rise from the chair or hold a squat) the two knees support 100% of the upper body weight, as the 90 degree angle makes the sine component of the equation equal to one. Each individual knee produces half that total torque. In the pull up instance while climbing the stairs, this is the moment that the leg being examined pulls the entirety of the upper body weight up the step into the straight leg position. This whole upper body weight is at a slight angle, so the moment is reduced, but it is not divided between legs like before. Ultimately the two values come out to require roughly the same torque at that instance, although rising from a squat is the most strenuous over all as a large amount of torque is required at nearly every step. Overall, the team took from this that an average amount of maximum torque necessary throughout these activities is roughly 182 Nm, meaning that 10% supplemented would require approximately a maximum 18.2 Nm of torque from the device. With this and other background information, the team sought to begin the informed ideation process.

3.2.2. Ideation

The key design choice to make, which shaped the rest of the device, is the actuation method. How exactly does the device relieve the 10% of knee torque? There are numerous appropriate actuation devices that could be used, each with its own pros and cons which could add or take away from the design. As such, a brainstorming session to decide on an actuation method was held. Actuation methods such as a motor at the knee, linkages, springs (gas, torsion), pulley systems, worm gear systems, linear actuators, and pneumatic/fluidic muscles were all discussed (Figure 18). Each had unique benefits, but the method that caught the team's eye most was the pulley system. As previously discussed, the patellofemoral joint acts as an anatomical pulley, and so the use of a pulley system mimics the human body and is sure to be capable of achieving the end goal.

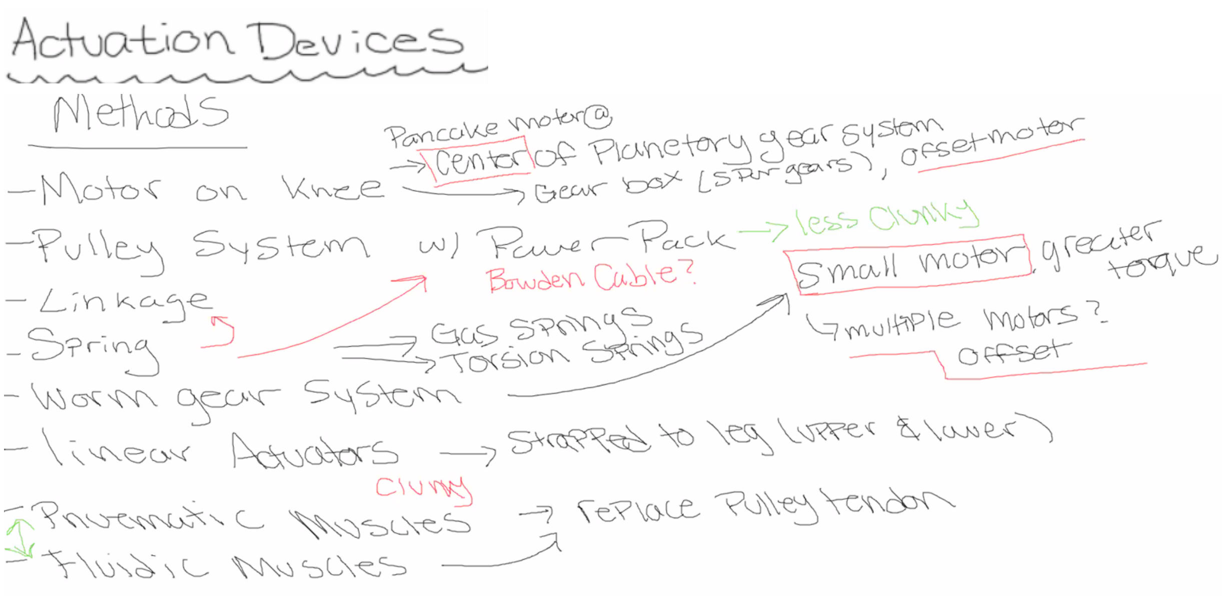

Figure 18: Notes from the Discussion of Actuation Device Possibilities

This actuation method also shows great potential to be lightweight, low profile, and have limited energy use, which would help to achieve the other defined criteria for success. The use of cables and very small pulleys allows for a significantly reduced firm exoskeleton, which keeps the bulkiness and hindrance of the device down. The use of cables also allows for the power/control system to be mounted away from the knee itself, further helping to reduce the bulk and hindrance of the device as it can instead be strapped to the waist, like many daily medical devices (insulin pump, EKG, etc.), moving the weight of the motor and battery off of the leg itself.

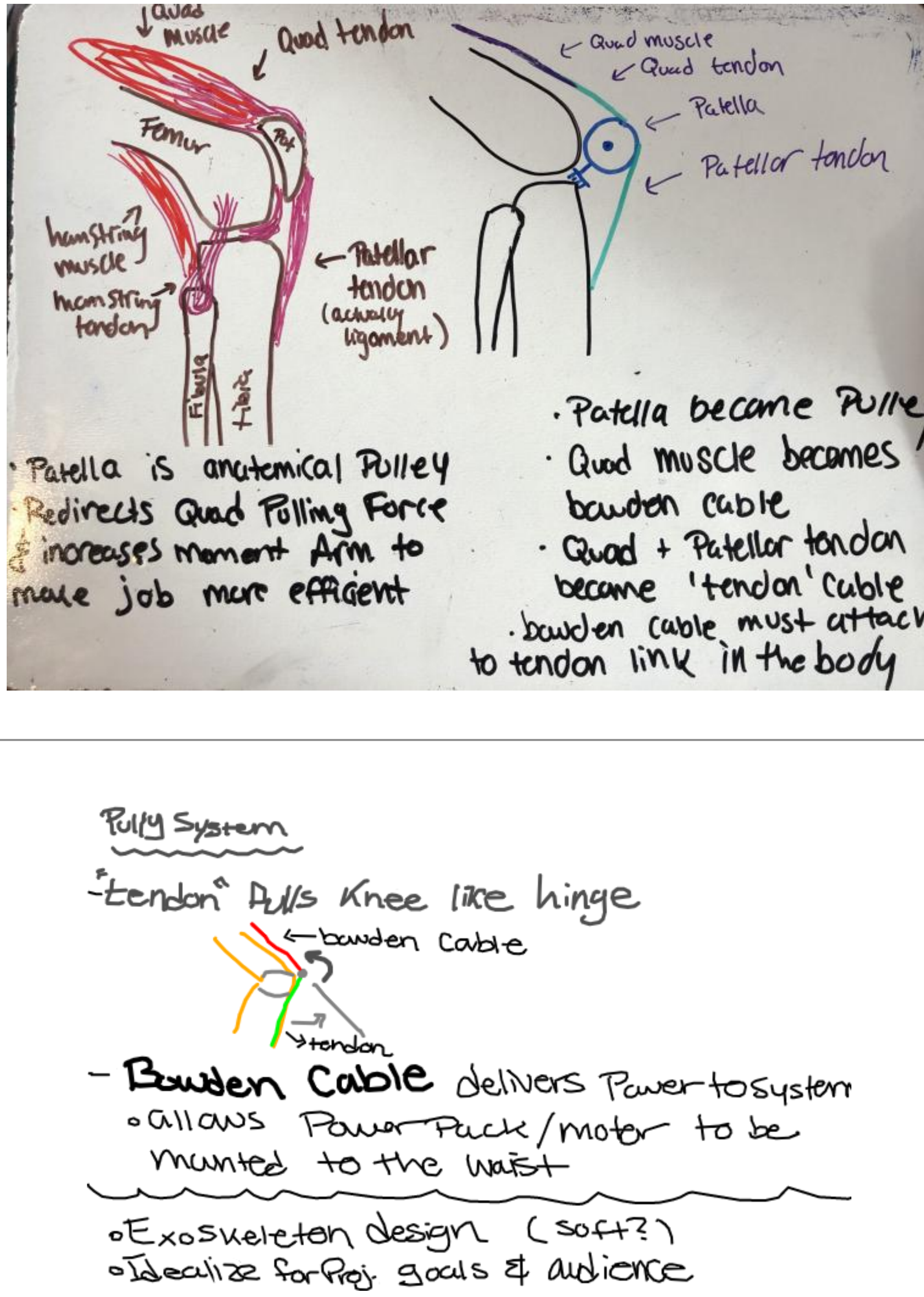

Figure 19: Notes from Brainstorming about Pulley Actuation

The H. Park et al paper, previously discussed, proved to be a great jumping-off point for the team following the decision to use this actuation device. They too chose a pulley system to run their device with. In fact, they took the opportunity to reduce the hindrance of the device and went for a soft exoskeleton for their user. Their soft exoskeleton holds a knee pad and cable track. The knee pad protects the knee against the reaction forces brought on by the pulley mechanism, just as the patella and femur's cartilage protects from patellofemoral reaction forces. The track provides a guide and mounts for the movable and fixed pulleys, respectively, and the attached cables, which mimic the tendons and the quadriceps muscle. The H. Park et al design demonstrated great potential for further exploration and development, and so during ideation, the team decided to lock on to this idea versus the phase 1 use of a motor and planetary gearbox mounted at the knee. The subsequent development of design ideas and improvements were based on the information presented in the H. Park et al paper, in the hopes of improving and idealizing that concept to meet the criteria and constraints and desired user set for the project.

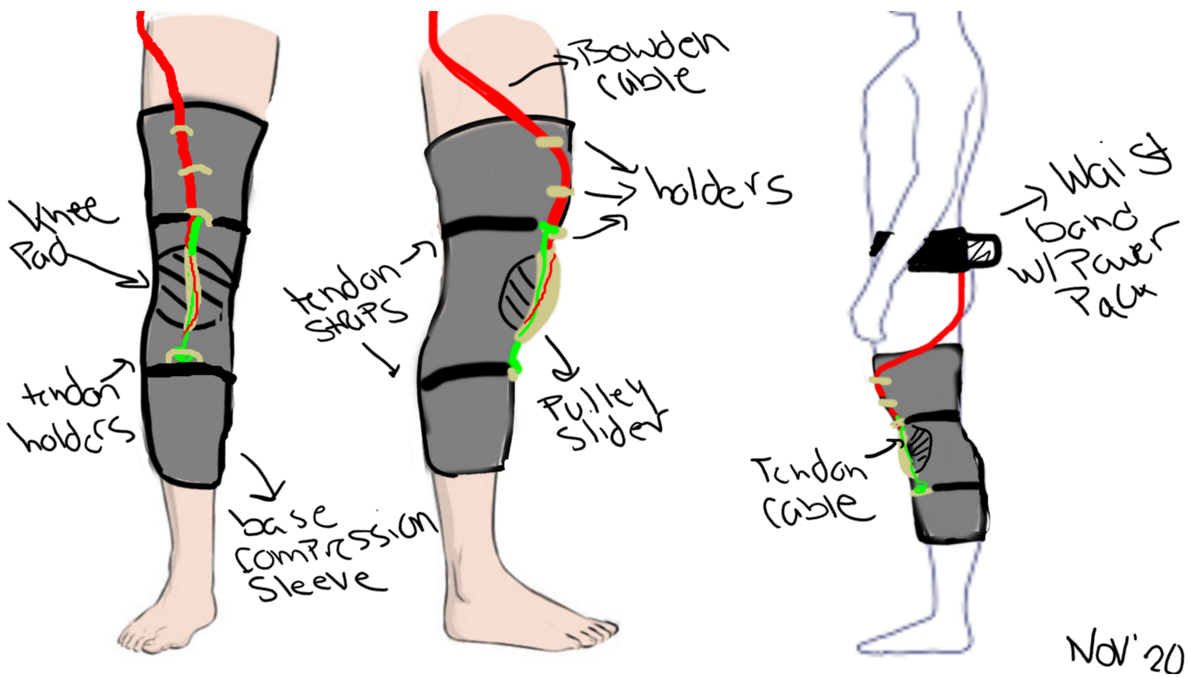

Figure 20: Sketch of the Initial Design Concept

3.2.3. Optimization

After deciding on a direction for the project, the category of optimization became key, because the team had chosen to focus on improving a previously attempted design concept. Optimizing both the H. Park et al design in theory and the team's concepts in practice became a large part of the methodology. The methods by which the team optimized theoretical components was to examine the function they served and the manner in which it occurred and try to identify elements that conflicted with the goals and criteria/constraints of the project. Then substitute these elements with more suitable designs. In practice, such as with the pulley track, for which the optimization is thoroughly detailed later on, optimization was an iterative process, where the team presented a design, attempted to realize it, then came together to examine the benefits and faults of the design. This process can be repeated over and over until a desirable result has been reached.

3.3. Fabrication Methods

Fabrication and manufacturing of the knee support device will only be partially fulfilled for this project. Due to COVID restrictions, full access to WPI's labs and shops was not available and nor entirely feasible for the team. As such, the team used a lot of tools and resources available through friends, personal means, and employers. The 3D printed material the team has access to mostly consists of PLA and ABS. Components that need to be fabricated or manufactured would be the Bowden cable guides and cable drum; tendon cable mounts; pulley track components, slider block, and mounts; test rig components; and the axles.

For the components that were to be fabricated and manufactured for the time duration of this project, these consist of the guides, mounts, and pulley track. The guides and mounts were prototyped in PLA. Ideally, the final guides and mounts would be made out of aluminum or a hard sturdy plastic, but 3D printed concepts are sufficient for proof of concept. In order for the pulley track to flex naturally, it would need to be made out of a flexible material. The team intended to achieve this via fabrication methods by using low-pressure injection molding. Many other components, such as the pulley mount, slider block, wheels, etc. were not able to be prototyped during the time period of the project, as the precision machining that needed to be done was not feasible given the team dynamics and tools consistently available to the manufacturing team member at that time.

3.4. Electrical & Control System Methods

The methods for electrical and control systems are primarily theoretical because the team member responsible for their execution left the project. The team had intended to create a test rig using 3D printed components and a potentiometer. This test rig could be affixed to a user's leg along with a number of EMG electrodes. The electrodes would read the user's EMG signals as they execute leg motions during walking, standing up, and ascending or descending stairs. These could be used, along with the angle position from the potentiometer to inform the timing for the actuation system. This test rig could then be adapted to to recognize when to actuate the device and in what capacity.

The electrodes never got to be set up, but the former team member did draft a circuit design for collecting and filtering the EMG signals. This circuit, shown in Figure 21 will connect to a microcontroller which interprets the signals and delivers instructions to the actuating motor. Data collected from the test rig setup was intended to be used to determine the specific linear velocity desired during motion and help inform the selection of the motor and power pack. As this was not completed, the motor and power pack selections detailed in the final design section are simply suggested possibilities.

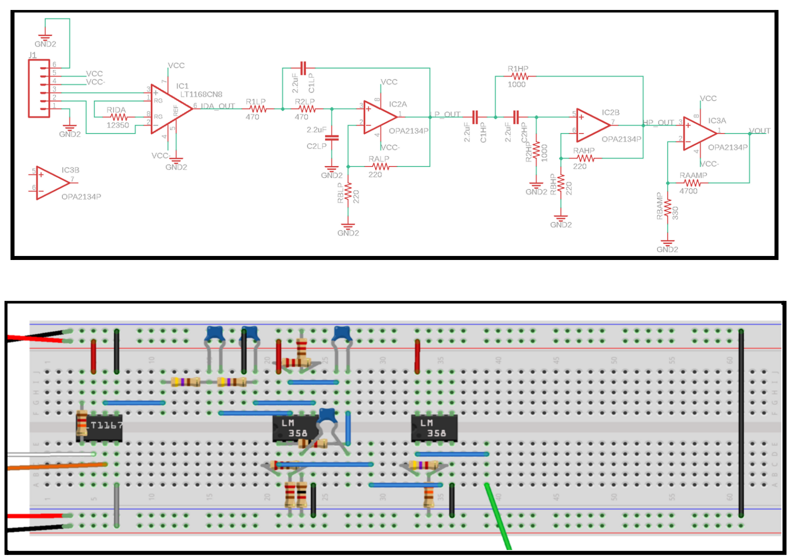

Figure 21: EMG Circuit Design

4. Fabrication & Prototyping

Due to restrictions previously discussed, the fabrication focus shifted towards the pulley track element, as this was one of the main and unique components of the design. The team also had the resources and materials available to prototype the design and test proof of concept. The following relay the team's design and fabrication decisions, as well as recommendations for future manufacturing of the pulley track.

4.1. Pulley Track Design Optimization

The team based the design of the pulley track on the H. Park design shown in Figure 13. The first iteration is shown in the figure below. The design is meant to sit on the knee, with the resting position being bent at 90 degrees. The component is made of an upper and lower level to the track. The upper level has the interior track for the moveable pulley to move along with a cutout between the two levels. This cutout allows for the cables to take the curve without running into the bottom of the track. Along the lower level, the tendon cable will be strung through the fixed pulley at the bottom of the curve. The team wanted to make use of a flexible material for the track, and so design changes were made according to the feasibility of the fabrication and operation of it.

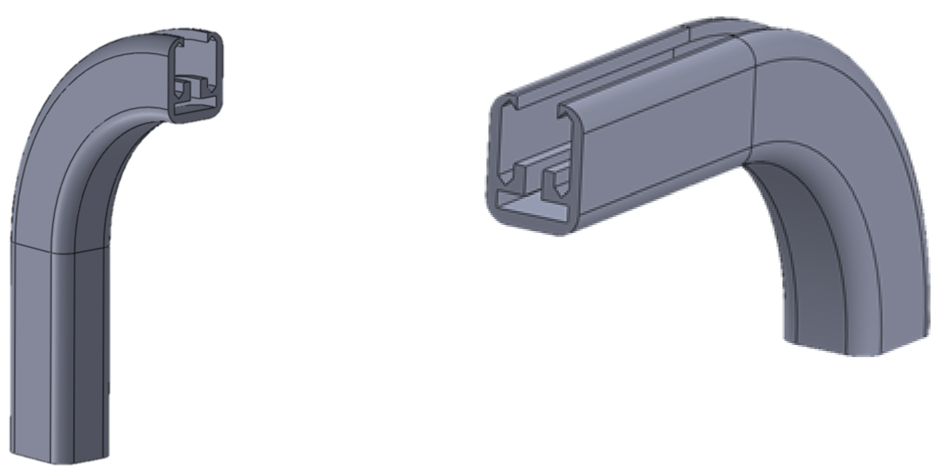

Figure 22: Initial Track Design

After considering the first design, changes were made for another iteration. Major changes included closing the roof of the track. This was for fear that the flexing of the material and this gap would cause the track to collapse inward. Others were the addition of a platform on which to mount the fixed pulley as well as a cap at the top end of the track to prevent the pulley block from being pulled clean out of the track, and 'mounting wings' to allow the track to be sewn down to straps and the base of the design. This second track received a rapid prototyping test, being 3D printed for a hands-on example of how the track geometry would work out (Figure 24).

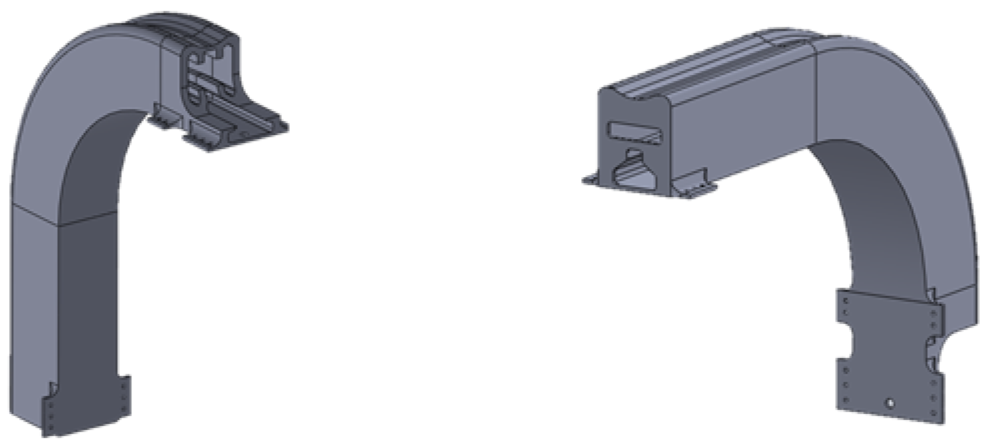

Figure 23: Second Track Iteration

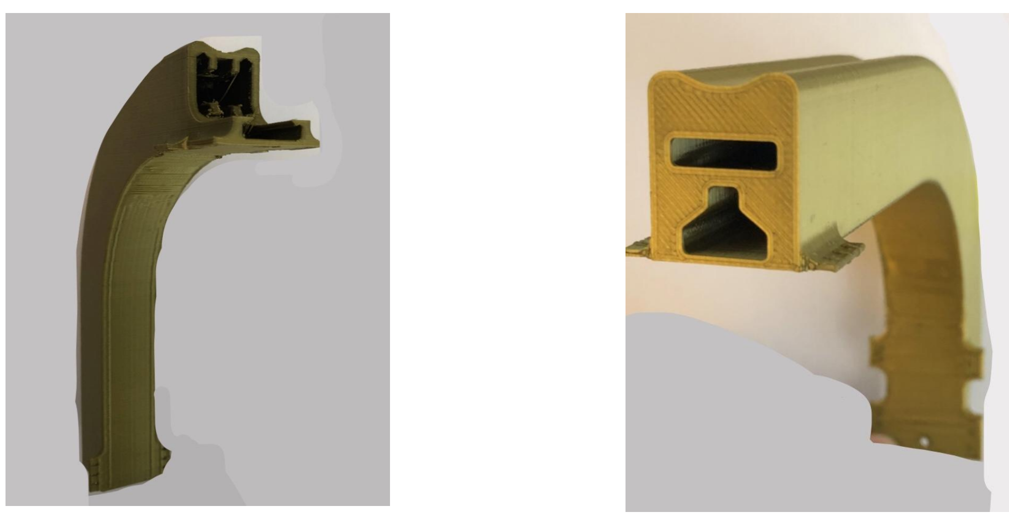

Figure 24: 3D Print Test of the Second Track Iteration

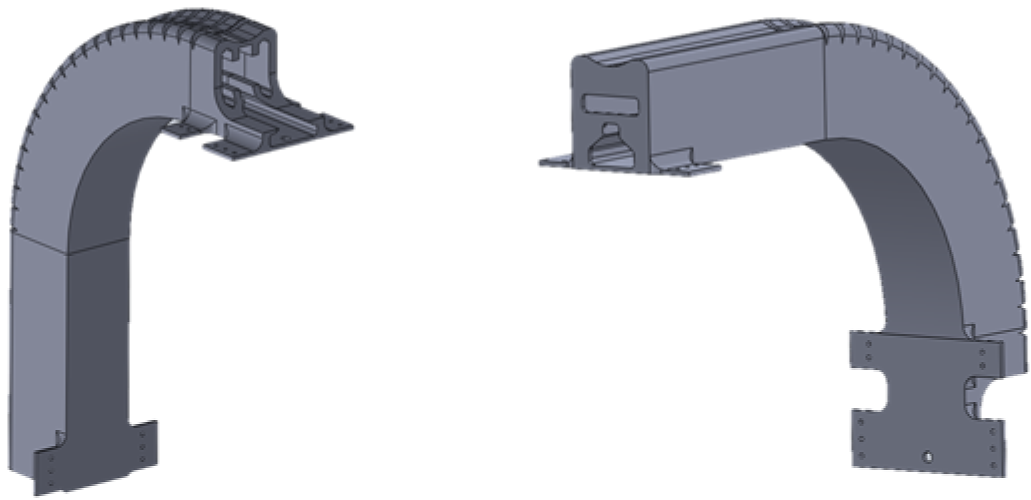

Figure 25: Final Track Iteration

From the second iteration to the final pulley track design, the only major change in design was the addition of the strain relief cutouts along the top of the track. This was added to provide extra room for the track to compress on itself when flexing with the user's knee motion in an effort to fully reduce deformation of the part that would hinder the pulleys. Some smaller changes to the mounting wings were made in anticipation of size issues.

While the geometry of the model was idealized for the moment, the printing tests revealed that the design is not ideal for 3D printing. While the final product would not be made out of a hard plastic like this, the design would be difficult to manufacture even if not 3D printed. Thus, when testing out the final design, the track was split in half, as shown below to make for more manufacturable parts.

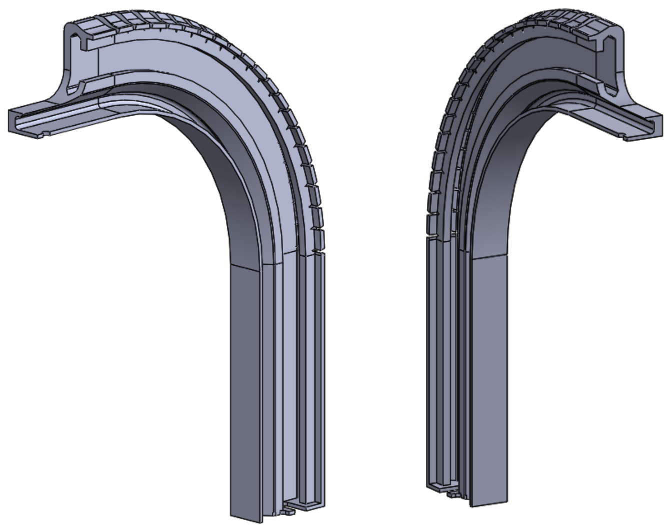

Figure 26: Split Model of the Final Track Design

The mounting tracks shown on the final design were also not included due to manufacturability. For the final system, those mounting tracks should also be a harder plastic and can be molded or attached to the pulley track for a final product. This is often referred to as insert molding, and the mounting tracks could easily be incorporated into the mold design. Material bonding has to be considered when utilizing insert molding, because depending on material choice, the tracks could be pulled out of the pulley track with minimal force if not designed properly or correctly accounted for. Hard plastics, such as high pressure injection molded parts, or even 3D printed material, would have a decent chance of bonding to the molding material, making for a less complex design that is more manufacturable. The 3D print test is shown below. The supports were still quite hard to get out the part and as can be seen some of the track broke off during the support removal process.

Figure 27: 3D Print Test of the Final Track Model

4.2. Pulley Track Fabrication

As noted previously, the team wanted the pulley track to be able to flex with the knee as much as possible in order to reduce the size and rigidity of the component and provide maximum comfort. Low-pressure injection molding material has the consistency and flex desired for a product like this, and the team had some access to some outside molding resources so the track prototype was fabricated using a Gluco handheld low-pressure molding system with Technomelt material. The main system is shown below in Figure 28 and a hose is attached to an injection gun, not shown in the picture. The team also had access to an industry-size ABS 3D printer. ABS has better thermal and mechanical properties over PLA when it comes to prototyping molds, but it isn't exactly ideal for molding, but a good material when it comes to getting an idea of what the product will eventually look like.

Figure 28: Gluco Handheld Low-Pressure Molding System

In order to fabricate this track through injection molding, it needed to be split in half - similarly to how the final design had to be split in half to 3D print due to the complex geometry and hollow cavities. When something is injection molded, there are usually two mold halves with milled-out cavities that when pressed together and filled with material - form the desired product. For 3D printing, the large amount of hollow space in the design was troublesome due to support removal, but for injection molding, the same thing is troublesome for other reasons. The cavity within the track is quite large and oddly shaped. In injection molding, when a part needs a negative space, there is often some type of insert that can be molded around and removed post-molding. Due to the design, removing any type of insert post-molding would be inefficient, if even feasible. Thus, the need to manufacture the track into two separate pieces that can be attached together further along the fabrication process.

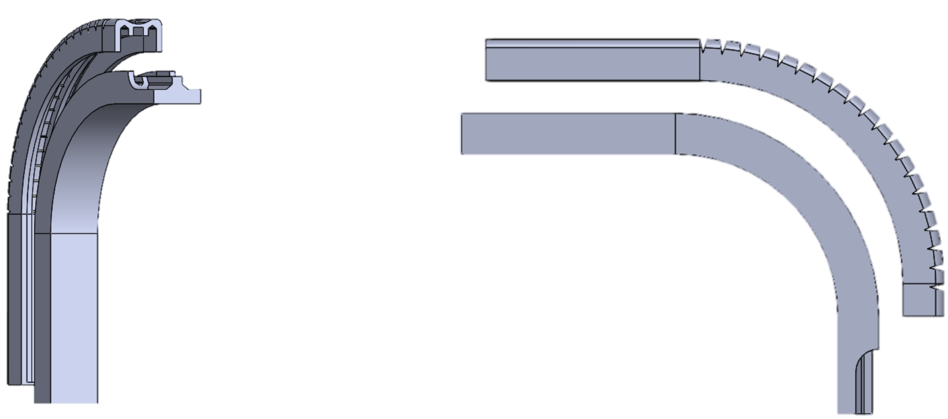

Figure 29: Split 3D Model of Final Pulley Track Prototype Design

When 3D printing the final design in two halves the way shown in Figure 26, the geometry was so complex, that a mold would never easily be able to be machined to match. The geometry also featured several undercuts - which is a term in molding that refers to a part feature that would prevent the part from being easily removed from the mold. Figure 29 above shows a different way of splitting the final design in half. There were still undercuts in the bottom half, but for the sake of seeing if the material and overall design would achieve the flex desired, the cavity where the fixed pulley and the tendon cable would be located, was filled.

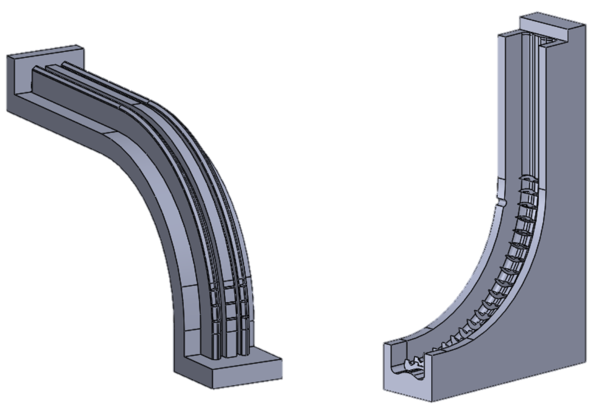

Figure 30: 3D Model of Top Pulley Track Piece Mold Halves

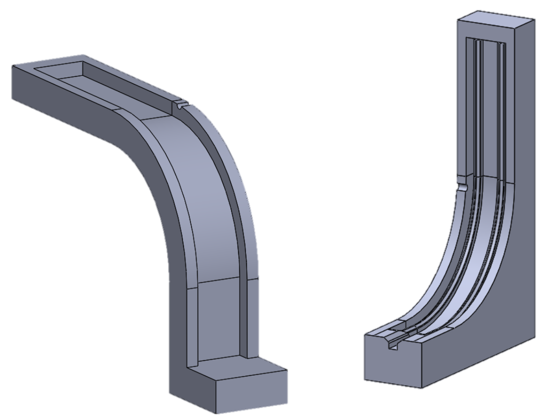

Figure 31: 3D Model of Bottom Pulley Track Piece Mold Halves

Figures 30 & 31 above show the four mold halves for the final prototype. As mentioned earlier, these will be printed in ABS. Each set has an injection site hole to allow for the material to flow in. The parting line of the molds was designed strategically so that enough of the final molded pieces would protrude from each half that the part could be gripped well enough to remove the part from the mold with minimal trouble.

Figure 32: Wax Mold Release

Figure 33: Post Wax

Figure 34: Spray Mold Release

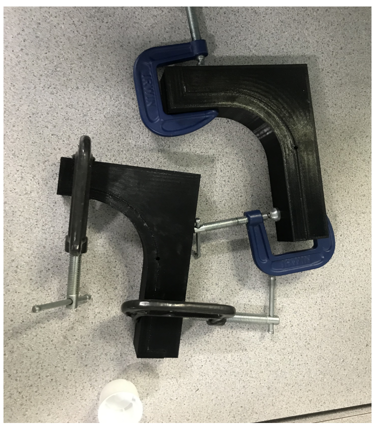

Figure 35: Mold Halves Clamped Together

Figure 36: Mold Being Injected With the Gluco System

To further ensure that the molded parts could be removed from the molds, and so that the molding material would not fuse to the ABS mold halves, two types of mold release were used. The first one is shown in Figure 32, and is a wax consistency that also helps to fill some of the layer gaps that come with 3D printing. This was applied with a Q-tip and simply rubbed in along the grain of the print. After all of the cavity faces have received a good coverage of the wax mold release, an air nozzle was used to remove some of the excess material. Then all four mold halves were sprayed with dry film mold release and clamped together shown in Figures 34 & 35 above, respectively. When prototyping molds this way, the clamps should be placed on opposite ends of the mold to ensure equal pressure is distributed. Once clamped together, the parts are ready to be molded. The molds should be set flat and the injection gun set perpendicular to the mold. The injection time should be relative to the size of the mold, in this case the injection time ended up being a little over 10 seconds. Once injected, the molds were placed in a freezer for 15 minutes to solidify before being removed from the mold halves.

Figure 37: Top View of the Bottom Half of the Molded Pulley Track

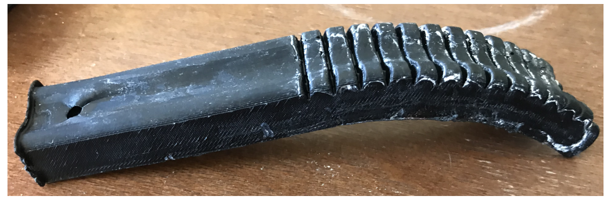

Figure 38: Top View of the Top Half of the Molded Pulley Track

Figure 39: Top View of the Bottom Half of the Molded Pulley Track

Figure 40: The Molded Pulley Tracks Together

Due to the complex geometry, large size of the part, and some complications with the mold, when molding the top half of the pulley track, material started leaking out the sides of the mold before the full cavity could be filled. This is referred to as an undershot in molding - when the part is not fully formed. The final molded parts can be seen in Figures 37 to 40 above.

4.3. Fabrication Considerations

As this was a prototype made out of a printed mold, the molded pulley track pieces were not expected to be super polished, but the information desired from these parts was provided. The general shape was able to (mostly) be formed (and would be fine with a few changes to wall thicknesses). The flex of the material was exactly what the team was looking for. The top half of the pulley track flexed with very minimal strain. The bottom half was able to bend and flex as well, but was a lot stiffer. Some strain relief cuts in the bottom of the bottom half of the track would help to decrease the stiffness.

In terms of actually producing this product, an aluminum or steel mold would be milled, and an automated molding system would be used to mold the parts. This would create a much smoother surface finish of the molded part, as well as smoother geometry on the interior cavities. Molded pieces are also much easier to remove from metal molds than plastic molds. Some basic re-design is also necessary to truly make this a manufacturable part. The bottom cavity that was filled for prototyping purposes, would need to be incorporated back into the design and made so there are no undercuts.

The team believes that fabricating the pulley track in two halves is the best way to create a complex part like this. The halves would have to be attached to one another in a sturdy fashion to ensure a stable overall system. A great way to seamlessly connect two plastic parts of the same material together is through an ultrasonic welder. An ultrasonic welder is an industrial machine that uses high-frequency acoustic vibrations that are locally applied to work pieces that are held together under pressure to create a solid-state weld. Small triangular sections are extruded from one half of the part to form an energy director to focus the ultrasonic energy. A fixture would be milled so that the two halves are lined up, the ultrasonic welding horn would be lowered, the initial contact of the energy directors would melt to the other half, and the two halves would fuse together. The current design would most likely need to be widened to provide a thicker surface where energy locators can be located so this process could be achieved.

5. Results

5.1. Final Design

Due to the limitations the team was faced with, the fabrication of parts and implementation of EMG was limited, but the primary goal of the team, to fully design the desired device, was achieved. In the figures seen below, annotated CAD models of the final design (excluding the cables which could not be modeled fully) are shown. The team utilized 3DS Solidworks 2020 to model the parts and achieve a final frame by frame animation of the device's response to knee flexion, seen later in this section.

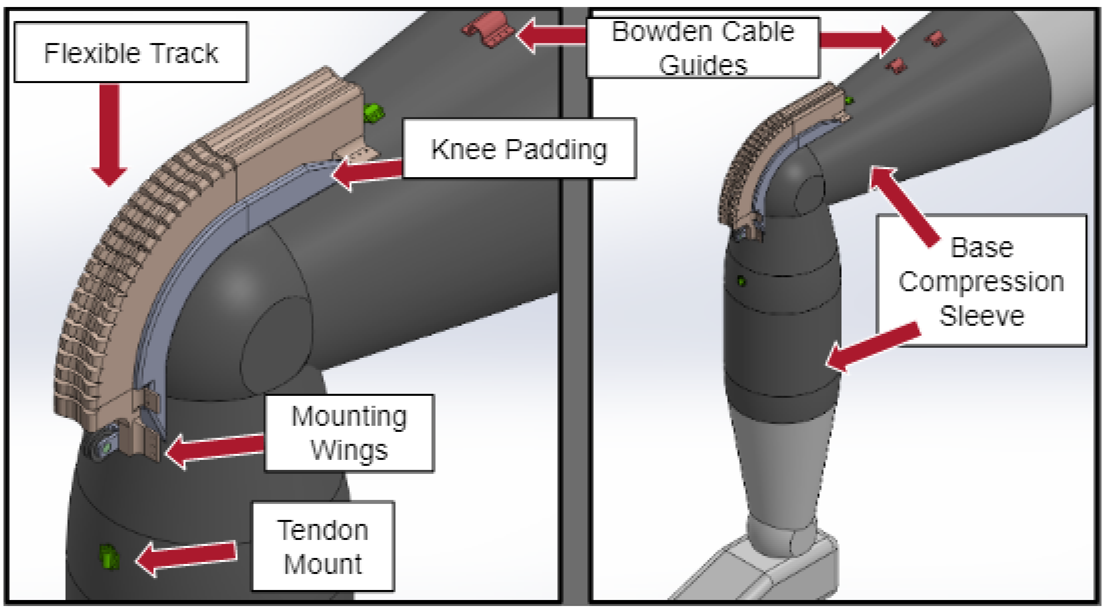

Figure 41: Final Design

The above figure shows the mechanism model, affixed to a stand in model of the human leg. Annotated in the image are the compression sleeve base (indicated by a color difference on the leg model), the flexible pulley track, knee padding, Bowden cable guides, tendon mounts, and the mounting wings for the track. Other elements are detailed in subsequent images, but some, such as the velcro straps and full bowden and tendon cables have not been modeled.

The track is affixed to the compression sleeve as well as two straps that can velcroed under the knee to increase support and hold the track steady. The Bowden cable guides can be affixed directly to the track, as there are no forces pulling on these, they simply hold the Bowden cable in place. The tendon mounts require their own set of straps, like the track, that grip the leg to avoid pulling on the elastic compression sleeve. While the design does still have straps, the reduced number and complexity of them, as well as the simple compression sleeve base, is an improvement on the H. Park setup.

Note that the cutouts on the curvature of the flexible track, which allows for maximum flexibility with minimal deformity. Figure 42 shows another set of annotated images of the final design, this time up close and with the leg hidden. Here, the knee padding, fixed pulley, top of track cover are shown more closely. The knee pad tapers down to a point at the bottom. This is to allow for the track to bend at the end. This could not be demonstrated in the Solidworks model, but the fixed pulley part of the track will lay flush with the taper when it has been mounted on the whole assembly. This also helps to provide extra clearance for the tendon cable, which is threaded through the fixed and moveable pulley, without increasing the design profile vertically.

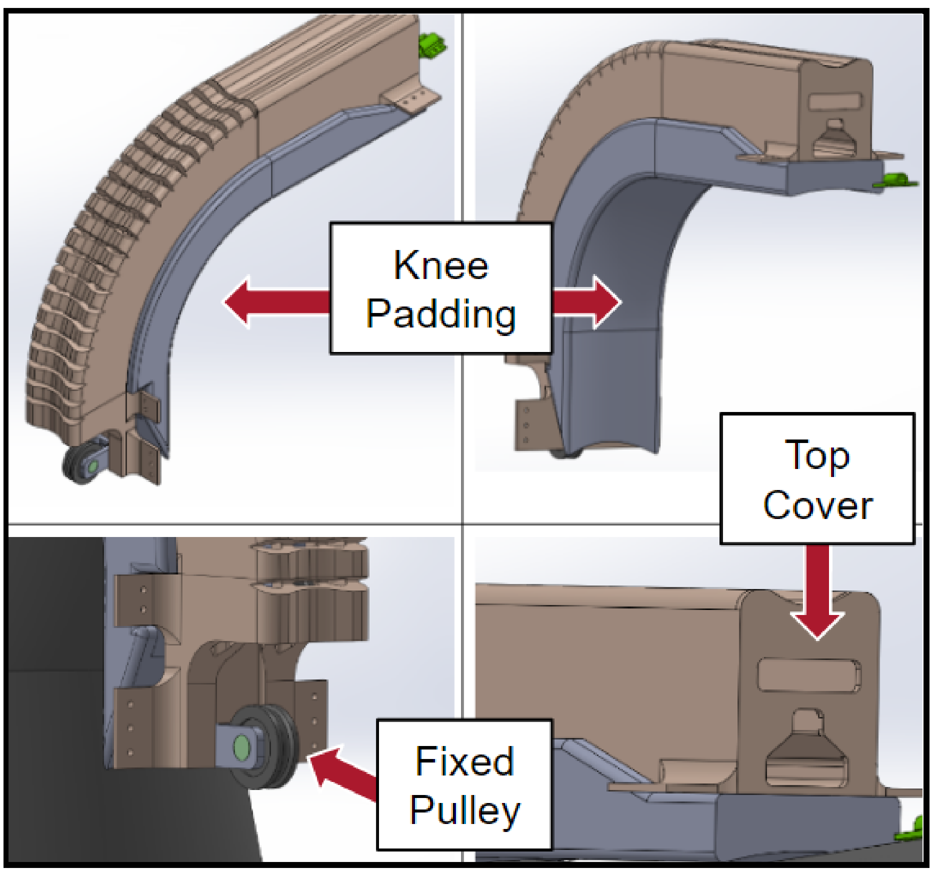

Figure 42: Final Design Up Close

The top portion of the track has a cover component built in. This element is meant to prevent the sliding pulley from being pulled clean out of the track system when the pulley is fully engaged (see Figure 43). The slider must be inserted into the track from the bottom, before affixing the fixed pulley. Then The unsheathed portion of the bowden cable is strung through the block at the top, in the fully engaged position. Adding in the fixed pulley locks the moveable pulley inside the track. Figure 43 shows the track up close in the fully disengaged (bottom) and engaged (top) positions. Annotations point out the mounting wings and fixed pulley once again, as well as the position of the slider pulley and the tracks it engages with.

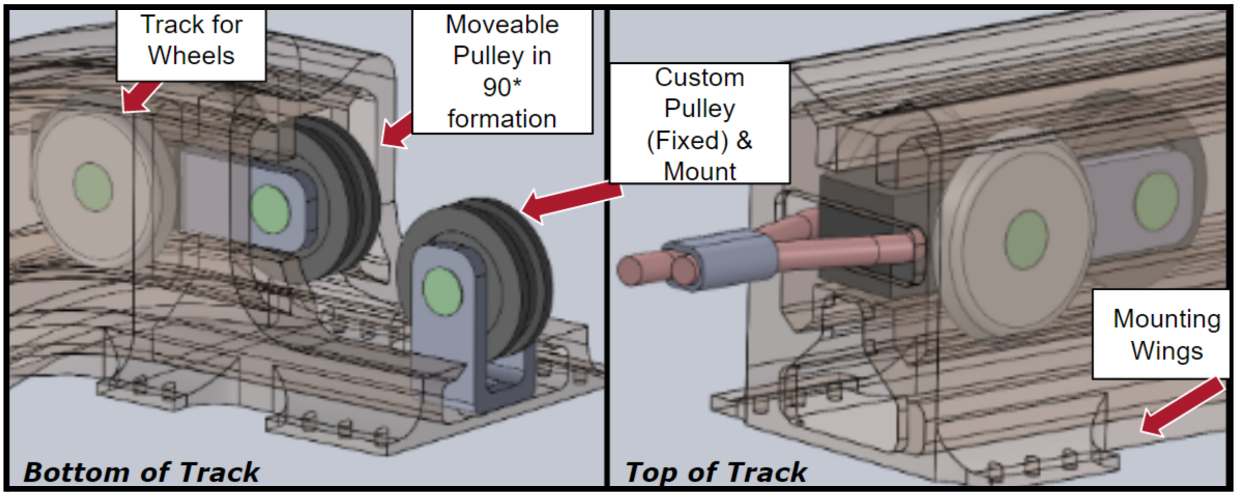

Figure 43: Pulley in Fully Disengage and Fully Engaged Positions

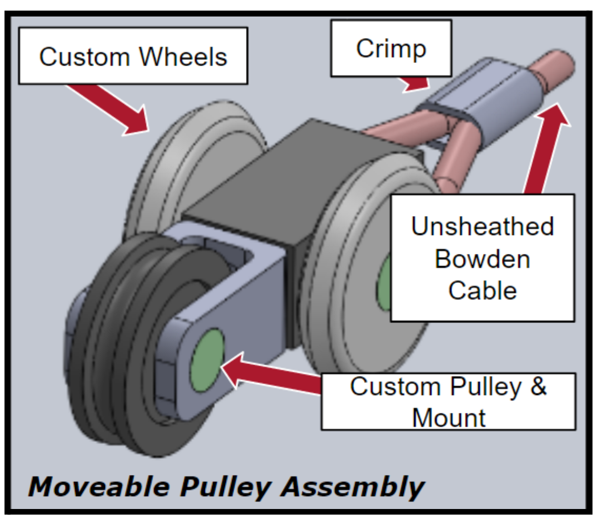

Figure 44: 3D Model of the Slider (Moving Pulley) Assembly

The final figure, Figure 44, shows the movable pulley up close. All of the parts in this design are as small as possible for reference, the Bowden and tendon cables are 2mm in width and the axles for these pulleys are 4mm in radius. Every effort to make these components as small as possible was taken, which requires several components to be customly designed, such as the pulley mounts, block, and wheels. Components for the slider that could be ordered are axle stock, Bowden cable, and crimps that hold the cable in place. A complete list of parts that were ordered (not customly produced or planned to be customely produced) can be found in the Appendix C.

5.1.1. Final Animation

In order to demonstrate the full range of motion of the knee under the system, specifically the knee pad and pulley track, an animation was created. Due to the flexible nature of the pulley track, which is extremely difficult to demonstrate in a Solidworks analysis, the animation was constructed via more traditional methods. Adjustments were made to the angle of the leg model and track, and screen caps at key time periods were taken. These were strung together into an animation that can be found at the following link: https://drive.google.com/file/d/1DH9HBJD0Wh8RcNf9L3dOE9X4tt2Fbm4z/view?usp=sharing. The animation shows the leg model exercising the full range of motion. The track is in resting position at 90 degrees flexion. At maximum flex of the track (roughly 45 degrees) the part is shown as stretched to meet the required flex of the knee. At maximum extension of the leg (180 degrees) the track is shown to have compressed without deformation. The figure below shows instances from this range of motion arranged to show a user rising from a deep squat to stand. This extension and compression of the track can be seen in this figure as well.

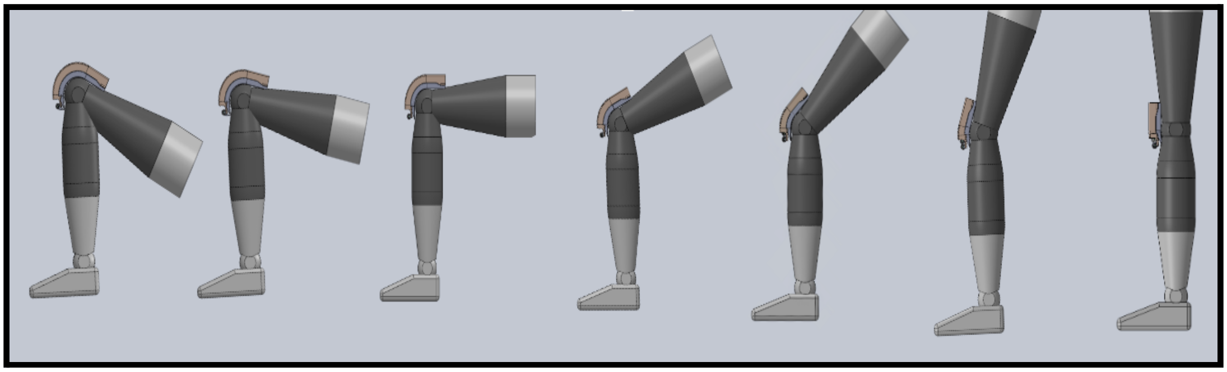

Figure 45: Motion of User Rising from Deep Squat to Stand

The animation and figure shows a shrinking and growing effect of the knee pad and pulley track due to complications with the software design, but in reality the effect would be more stretching and compression of the track as it flexes and straightens out with the knee. Since a great amount of focus was on the pulley track throughout this project, the animation did not feature the Bowden and tendon cables or other components in order to clearly show the pulley track model.

5.1.2. Power and Actuation

For reasons discussed, the power and actuation elements of the final design were not able to be finalized in their entirety. For this reason, it is assumed that all components function as desired, and the reality of that must be examined with a future team. These power, electronics, and controls components are key to realizing the function of the device. Non-invasive EMG sensors would be sewn into the fabric of the compression sleeve, taking muscle signals from the quadriceps region. These signals are collected, cleaned, and categorized, before being sent to the Arduino, a microcontroller. The Arduino code will interpret these signals and deliver the information to the motor, much like how the brain delivers the EMG signals to the muscles. Recall the motor, and powering battery is mounted on a velcro waistband. This motor activates, and with the help of a pulley, actuates the Bowden cable, which then sets the mechanism in motion as needed. The flow diagram shown in Figure 46 demonstrates the path of the signal from muscles all the way back to the leg and Figure 47 shows the motor and pulley actuating system as it would have operated.

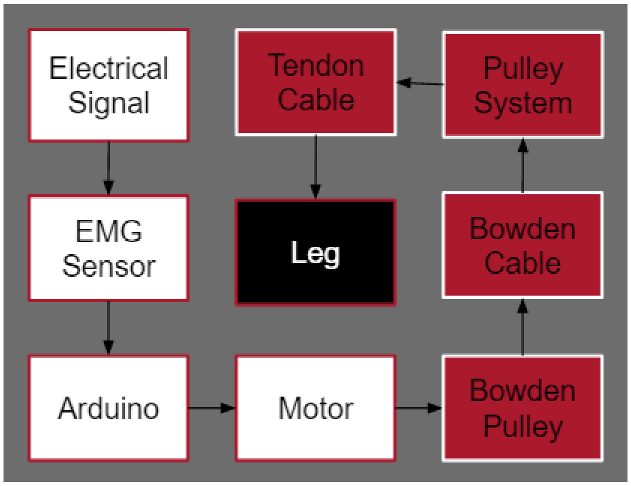

Figure 46: Signal Flow Diagram

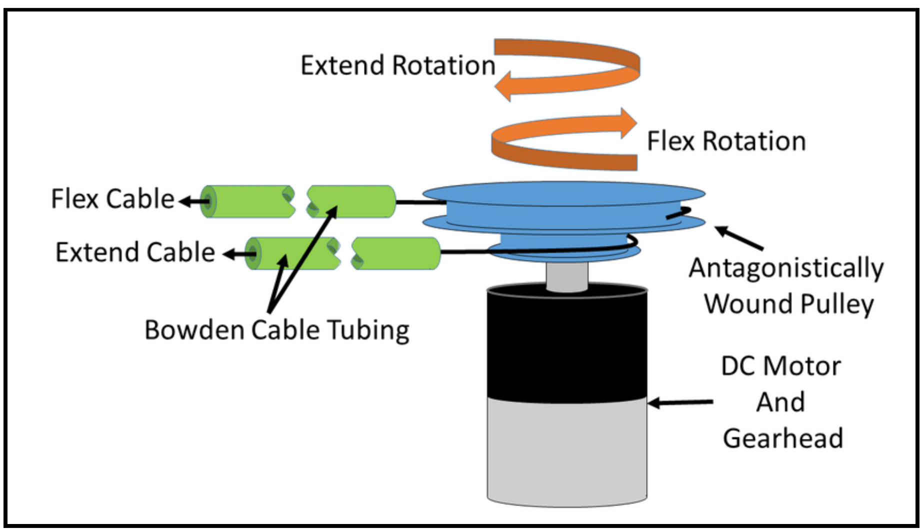

Figure 47: Motor and Pulley Actuation System

The waistband essentially houses all of the electrical elements. The battery and motor and pulley, wound with the Bowden cable are fundamental to drive the system, and, as with all other components, must be kept as small and light as possible. A possible choice for battery and motor are shown in Figure 48 and again in Figure 49 in hand for size comparison. The data sheet for the 12 V battery is available in Appendix D, where the exceptionally small and light dimensions can be seen. The battery also has a very admirable capacity, important if the device is to be operated for most of the day. The 12v DC motor and gearhead shown is an example of a possible motor. This 12v motor has a suitable torque capability, which could be increased even more if a slower

RPM motor is selected. While these portions of the design are far from finalized, they do make adequate stand-ins, showing some reasonable options for power and actuation could be selected by a team member familiar with their operation.

Figure 48: Possible Battery and Motor Choice

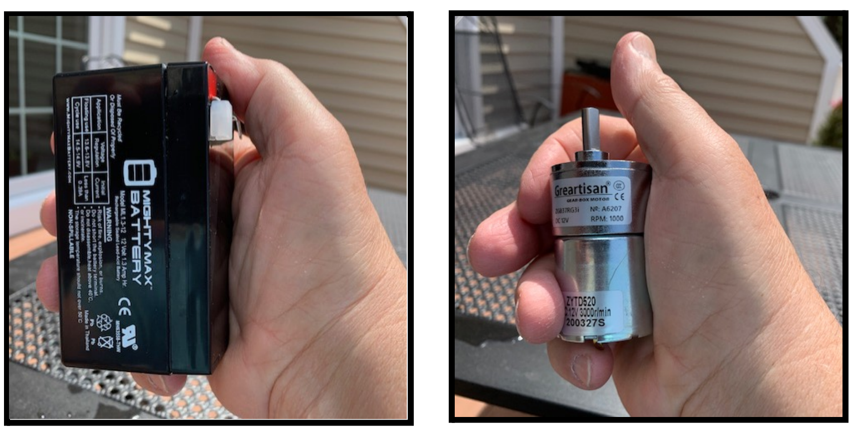

Figure 49: Possible Battery and Motor Choice (Hand Reference)

6. Conclusions

Throughout the duration of the project, the team was able to improve an existing assistive device concept and fabricate prototypes and order parts for some major components. Extensive research of biomechanics and user-centered design guided the team in the goal of creating a device that mimics the biomechanics of knee motion while also being accessible and user-friendly. Due to the COVID-19 pandemic and extenuating circumstances regarding a former group member who has since left WPI, the team was unable to fully manufacture a prototype and complete the electrical elements of the project. However, in the pursuit of designing a device that met these goals, much was learned about the biomechanical topics and design for manufacturability and with the user in mind. A future team, under better circumstances, could take this project to the next level and produce a fully functioning model of the concept.

6.1. Improvements & Next Steps

The original plan for this project included creating a test rig using a potentiometer and a circuit board to record the position and EMG measurements of a user's knee to help the team determine the necessary muscle movements and linear velocity which would in turn help to determine the torque needed for a user to walk, stand, and ascend/descend stairs experimentally. The team hopes that this project can be continued in a phase 3 continuation, hopefully in a full on-campus environment with access to more resources. A team member with ECE or robotics experience would be needed for this continuation to bring the power, actuation, electrical, and coding elements together. Full assembly and fabrication of the whole system should be completed. Testing of the system can be performed at WPI's Practicepoint healthcare facility. Phase 1 had the chance to do some initial testing at Practice Point to measure the torque of a user's various knee motions, so further utilization of this facility would be useful.

A major goal of this project was to reduce the bulk and weight of the system and add less strain to the users leg compared to the phase 1 design. The phase 2 design achieves this, but improvements could further be made. Ideally the leg components would be able to fit underneath loose pants comfortably. The current pulley track design sits one inch tall and combined with the compressive sleeve and knee pad the whole system will sit around 2 inches off the knee. Some redesign the pulley layout to decrease height would also be needed. Possible changes to make the fabrication of the pulley track in one piece would also majorly increase the manufacturability of the product.

6.2. Novelty & Marketability

This pulley-based knee support design is novel because of its unique properties which incorporate the positive aspects of previous assistive knee devices. One aspect of this design is its unique way of providing additional torque to the knee through a pulley and tendon system which mimics the patellofemoral joint. This system differs from those used in most other active support devices. Most devices simply apply additional torque at the knee without considering the intricate biomechanics of the knee. On the other hand, the pulley system replicates the muscles and tendons used during knee motion with mechanical components in order to provide additional torque in a way that is mechanically similar to the natural motion.

This design also incorporates the simplistic, slim, low-profile style of other designs. Devices that are simple and slim are easy for users to wear with minimal impact on their daily lives. They are much more accessible than devices that are complex and difficult to put on and wear in day-to-day situations. Our design bridges the gap between designs that are accessible and designs that replicate the knee's motion in a biologically accurate way.

There is some risk for intellectual property concerns associated with our design that would need to be addressed if the design were to be patented and distributed. Since this design is based off of another group's design, the H. Park et al group, it would be important to consult with a patent attorney and the members of the project to examine the viability of a patent. However, since this team is not considering any further action with this design beyond the scope of this project, this will not be an issue. If future teams were to finish the electrical components of the device, manufacture it, and were at the stage where they were considering a patent, this would need to be addressed.

This knee support device has the potential to do well in current markets. Biomedical research suggests that up to 25% of adults experience some sort of frequent knee pain (Nguyen et al, 2011). This means that there are millions of people who are in the market for products that provide knee support or comfort. Furthermore, the frequency of knee pain in the population increases with age, which means that older people are more likely to experience knee pain due to causes such as osteoarthritis, a leading cause of knee pain in the older population, or injury caused by common accidents. The global market for knee braces is expected to reach $1.9 billion by 2025 and is growing at a rate of 4.25% annually (PR Newswire, 2020). Market data for active knee devices is currently not available as it is an emerging market that has not yet been deeply analyzed, but these statistics are promising and show that the market is expanding and that there is potential for new knee devices, such as this one, to be successful.

An important question that would need to be addressed if this device, and others like it, were to be sold on the consumer level would be if the cost of an active knee support device would be worth it for a consumer as opposed to the reduced cost of a passive knee support device. Active devices are typically much more complex and have components that could be expensive to manufacture, whereas passive devices are typically simple to manufacture and do not have any advanced mechanisms, motors, or electrical components. An active device such as this one would need to be low-cost in order to become a viable option for consumers and compete against passive devices. For example, the phase one device had an estimated price point of $10,000. A device that is that expensive would likely not be something a typical consumer in this market would consider buying as the price is very high and there are cheaper alternatives. Such devices would likely need to be priced at $1,000 or less to be accessible and reasonable for most users to purchase them.

6.3. Learning Components

While working on this project, the team learned valuable information and skills that can be applied to future project experiences. Through research on knee motion, the team learned about the intricate details of the biology and biomechanics of the knee as well as the human gait cycle. This included the patellofemoral joint and its unique functionality as an anatomical pulley which was the foundation of the eventual design. This also gave the team the necessary information to work towards producing a design that mimicked biomechanics as opposed to a design that did not account for the knee's intricate biology and function.

The teamed also gained insight into designing for fabrication. As the focus shifted to fabricating the device's pulley track, some unexpected obstacles became apparent with the component's manufacturability. Initially, the team intended for the track to be manufactured as one piece. However, when considering fabrication methods, it became apparent that this needed to change in order to preserve the important properties of the component, specifically the intricacies of the inner track and the part's flexibility. Injection molding was chosen as the ideal way to manufacture the track, but the track design had to be modified in order to account for the limitations of injection molding. The team was able to redesign the track so that it could be made with this method, and in the process, learned how to design with manufacturability in mind.

The team also learned to work with limitations. One major obstacle the group faced was the COVID-19 pandemic. Remote learning and social distancing measures meant that team members were unable to meet in person consistently or have in-person meetings with the advisor. It also meant that manufacturing a full prototype would be difficult due to reduced machine shop availability and limited access to other resources. The team also had to formulate a new plan of action due to a former team member leaving WPI halfway through the year. This team member's background in electrical engineering was critical to the electrical component of the anticipated design, so the team had to restructure the goals of the project due to the loss of their expected contributions. However, the team was able to pivot and learned a valuable lesson in working with unexpected limitations during an engineering endeavour.

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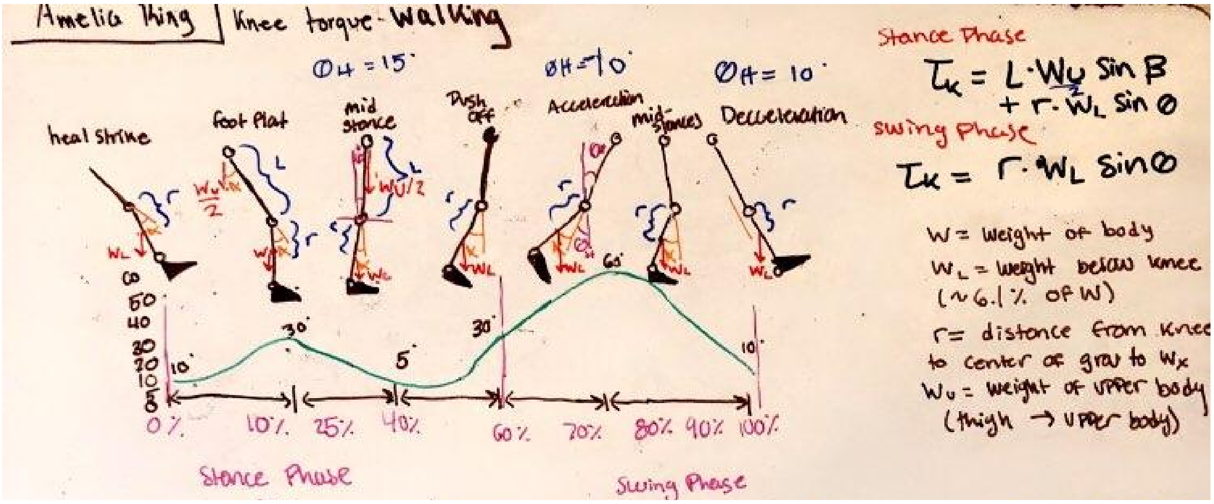
Appendices

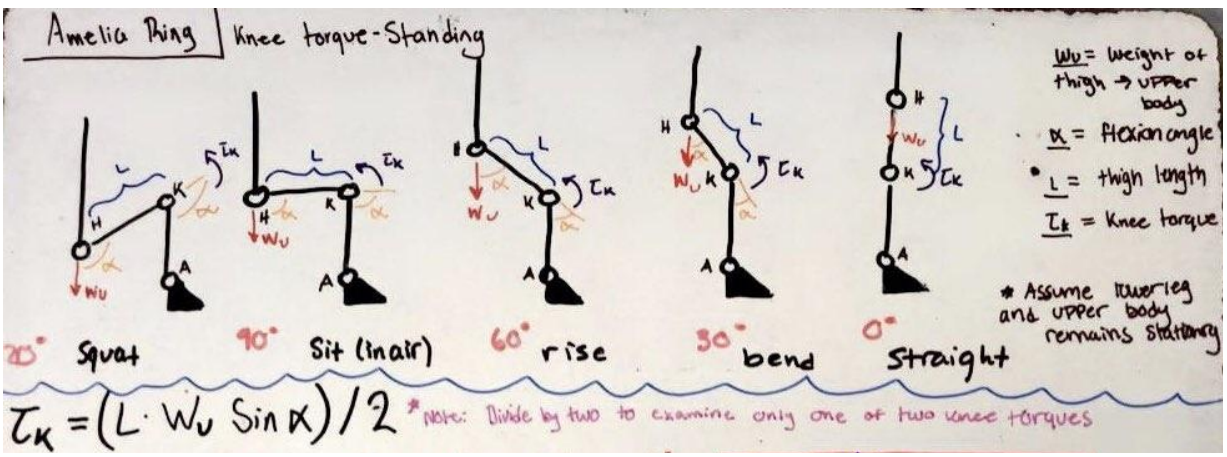
Appendix A. Initial List of Tasks

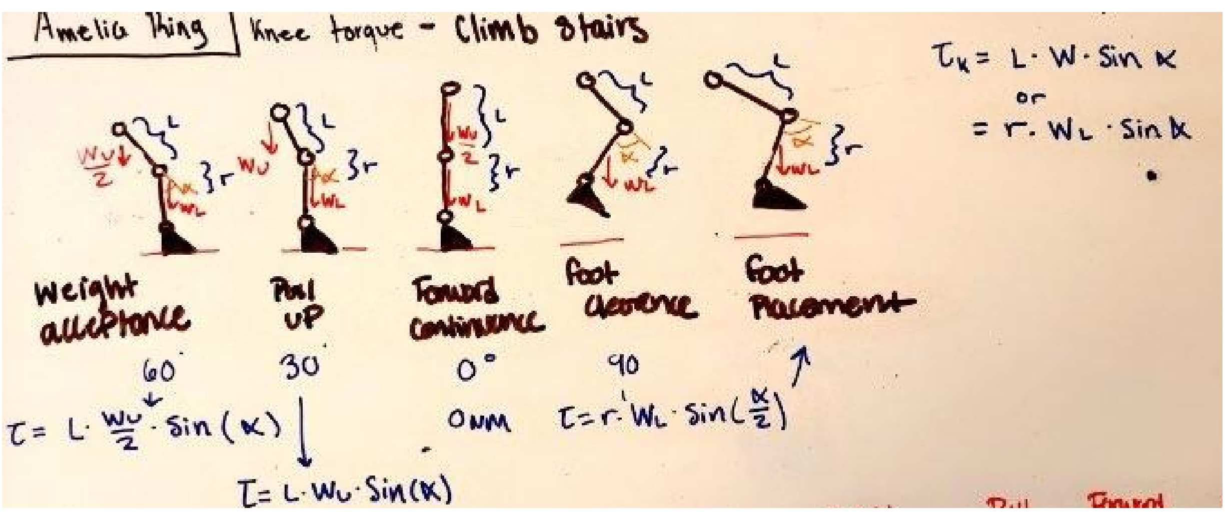
1. Research the topic thoroughly
    a. Phase 1 report
    b. Journal articles on similar knee support exoskeletons
    c. Research on the human gait
    d. An in-depth look at H. Park et al.
    e. Research into the patella-femoral joint function
    f. Conditions causing knee weakness
    g. EMG Surface Electrodes
    h. Bowden cable control
2. Determine goals and direction
    a. Problem statement
    b. The Goals/Sub-Goals
3. Calculate the expected max torque
    a. Do free body diagrams for the human gait for standing up, walking, and climbing
    b. Use diagram and rough figures to calculate knee torque throughout the phases
    c. Determine how much torque is expected to be supplemented
4. Design the prototype (initial)
    a. Determine actuation method (pulley and control cable)
    b. Determine direction (H. Park et al)
5. Order parts & supplies
    a. Order electronic parts
    b. Order parts for initial fabrications/assembly
6. Set up the test rig and collect data
    a. Complete the test rig
    b. Set up electrodes
    c. Create movement graphs
    d. Calculate the angular velocity
        i. Add current numbers to MATLAB script
7. Fabricate parts
    a. Bowden Guides
    b. Tendon Cable Mounts
    c. Track 3D Print
    d. Injection Mold Track
    e. Slider Block
    f. Cut axles
    g. Pulley Mounts
    h. Test Rig
    i. Wheels

   j. Knee Pad
   k. Bowden Cable Drum
8. Design power pack and select motor
   a. Select the ideal motor
   b. Design/select the Bowden Cable drum
   c. Select the power supply
   d. Design waistband configuration
   e. Order remaining parts
9. Design Knee Pad
   a. Design a slim, pressure-relieving knee pad molded to the track
10. Order remaining parts
    a. Parts for the waistband powerpack & cable system
    b. Parts for the knee pad/a knee pad
    c. The remaining parts to finish the prototype
11. Assemble the prototype
    a. Assemble the straps (velcro, mounts/guides)
    b. Assemble the pulley system and thread the cables
    c. Affix the track to the knee pad
    d. Affix straps and the knee pad to the sleeve
    e. Set up the waistband
12. Test prototype
    a. Set up the actuation system
    b. Set a user up with the system
    c. Do the three activities and measure torque (if possible) to determine the effectiveness
13. Write the paper
14. Redesign the prototype
    a. If time...
    b. Make changes to the parts based on design conferences and tests
15. Prepare for presentation

Appendix B. Force Diagram Sketches

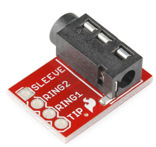






Appendix C. Parts Purchased

Part	Picture	Cost	Quantity (purchased)
Bowden Cable		$2.80./ft	6 ft
5/64th in (2 mm) Cable		$0.65/ft	10 ft
U Groove Bearings 4x13x4mm		$1.02	10
4mm Diameter Axle Stock		$0.0176/mm	150 mm
Wire Cable Crimps (5/64th/2mm)		$0.10	50
Compression Sleeve (Medium)		$6.00	2

1-inch Nylon Strap Material		$0.55 / ft	15 ft
Velcro for Straps		$0.53	15 ft
Disposable Surface Electrodes		$7.95	1 Pack
Electrode Sensor Cable		$4.95	1 Set
Audio Jack Breakout Board		$3.95	1
Arduino MKR Zero		$25.90	1
22 AWG Solid Wire		$12.98	1 Set
Assorted Header Kit		$10.99	1 Kit

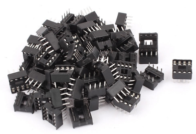
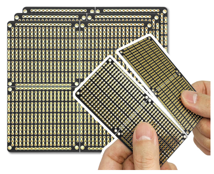
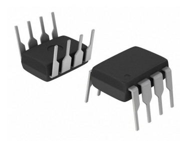
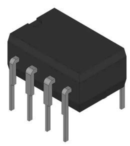
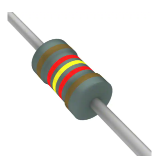
8 Pin DIP IC Socket		$25.90	1Set
Snapple Strip Boards		$11.99	1 Pack
Instrumentation Amplifier		$9.23	1
RC4580P Operational Amplifier		$0.43	8
2.2uF Ceramic Capacitor		$0.34	4
12.4k Through-Hole Resistor		$0.10	1
20k Linear Potentiometer		$1.69	5

Table 3: Parts Ordered

Appendix D. Battery Data Sheet

MIGHTYMAX BATTERY

ML1.3-12*(12V1.3Ah/20hr)*

The rechargeable batteries are lead-lead dioxide systems. The dilute sulfuric acid electrolyte is absorbed by separators and thus immobilized. Should the battery be accidentally overcharged producing hydrogen and oxygen, Special one-way valves allow the gases to escape thus avoiding excessive pressure build-up. Otherwise, the battery is completely sealed and is, therefore, maintenance-free, leak proof and usable in any position.

Battery Construction

Component	Positive plate	Negative plate	Container	Cover	Safety valve	Terminal	Separator	Electrolyte
Raw material	Lead dioxide	Lead	ABS	ABS	Rubber	F1	Fiberglass	Sulfuric acid

General Feature

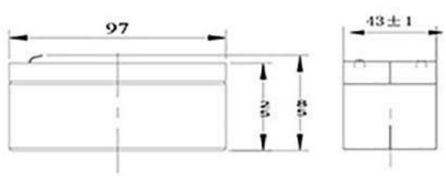
- Absorbent Glass Mat(AGM) technology for efficient gas recombination of up to 99% and freedom from electrolyte maintenance or water adding.
- Computer designed lead, calcium tin alloy grid for high power density.
  Long service life, float or cyclic applications.
  Maintenance-free operation.
- Low self discharge.

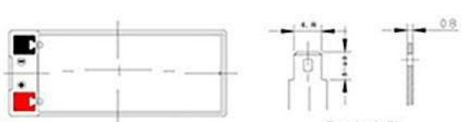
Performance Characteristics

Capacity 77°F(25℃)	20 hour rate (0.06A、10.5V)	1.3Ah
	10 hour rate (0.115A、10.5V)	1.15Ah.
	5 hour rate (0.22A、10.5V)	1.1Ah
	1 hour rate (0.81A、9.6V)	0.81Ah
Internal Resistance	Full charged Battery77°F(25℃):120mΩ	
Capacity affected by Temperature (20 hour rate)	104° F(40℃)	102%
	77° F(25℃)	100%
	32° F(10℃)	85%
	5° F(-15℃)	65%
Self-Discharge 68°F(20℃)	Capacity after 3 month storage	90%
	Capacity after 6 month storage	80%
	Capacity after 12month storage	60%
Max. discharge current77°F(25℃): 18A(5S)		
Charge (Constant Voltage)	Float: 13.6~13.8 V/77° F/(25℃)	
	Cycle:14.5~14.9 V/77°F/(25℃) Max. Current: 0.39A	

SPECIFICATION

Nominal voltage ·················· 12V
Number of cell 6
Length(mm/inch) ·················· 97/3.82
Width(mm/inch ·················· 43/1.69
Height(mm/inch) ·················· 52/2.05
Total Height(mm/inch) ············ 58/2.28
Approx. Weight(kg/lbs) ············ 0.56/1.23





Discharge Constant Current (Amperes at 77° F25 ℃)

End Point Volts/Cell	5min	10min	15min	30min	1h	3h	5h	10h	20h
1.60V	5.20	3.50	2.43	1.35	0.81	0.35	0.24	0.126	0.07
1.65V	4.93	3.33	2.32	1.30	0.78	0.34	0.23	0.123	0.06
1.70V	4.65	3.16	2.21	1.24	0.75	0.33	0.23	0.118	0.06
1.75V	4.36	2.98	2.10	1.18	0.72	0.31	0.22	0.115	0.06
1.80V	4.07	2.80	1.98	1.12	0.69	0.30	0.21	0.113	0.06

Discharge Constant Power (watts at 77° F 25℃)

End Point Volts/Cell	5min	10min	15min	30min	45min	1h	2h	3h	5h
1.60V	9.00	5.67	4.67	2.67	2.07	1.63	0.88	0.66	0.48
1.65V	8.44	5.34	4.41	2.53	1.97	1.56	0.85	0.65	0.47
1.70V	7.88	5.01	4.16	2.40	1.87	1.49	0.81	0.63	0.46
1.75V	7.33	4.68	3.90	2.26	1.77	1.42	0.76	0.61	0.45
1.80V	6.79	4.35	3.64	2.12	1.67	1.34	0.71	0.60	0.44

(Note)The above characteristics data are average values obtained Within three charge/discharge cycles not the minimum values.

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