Soft, Passive, Energy Recovering Exosuit for Augmenting Lower Body Motion

Anand Parwal, Abhishek Kashyap, Michael Chan, Noah Bennett, Omar Hoblos

***Abstract*— A soft, passive, multi-joint exosuit was developed for the purpose of lower body augmentation to aid the user with the performance of walking, running, and jumping. Through OpenSim simulations, key muscles at specific joints were identified and analyzed for major contributions in total fiber force and moment created amongst other key factors. Torque results obtained from OpenSim were verified using MATLAB through the implementation of the Newton-Euler method to ensure reliable results. Following simulation, a design concept for the initial prototype was developed. The suit is comprised of all soft, conforming materials which include webbing anchor points and elastic actuators, both of which allow for user adjustability. The exosuit uses elastics of varying stiffness at optimal anchor points, similar to the natural anchor points of biological muscles, to provide the user with an assistive torque for hip extension and knee extension. The suit takes advantage of a persons kinetic energy and momentum to stretch the elastics and gain potential energy. The elastics are then released as an assistive torque at a specific moment on the gait cycle to augment the users motion without hindering their range of motion. The suit was tested using the Tracker software and analyzed for velocity and acceleration with and without the suit to test the performance of user motion. The results show a potential performance enhancement with the suit on for the acceleration of the hip, however more trials need to be analyzed for further validation of results.**

Overall, the team was able to successfully establish a stable proof of concept platform, ready for further testing and analysis in future works.

I. INTRODUCTION

Exoskeletons on the market today showcase a variety of ways to aid both healthy and disabled humans perform or excel in daily tasks. Current exoskeletons vary in application ranging from rehabilitation to assistive and augmentation, with commercial exoskeletons targeting nearly every joint in the body. Many of these products tend to be rigid and bulky in design and come at a hefty price [1]. Lower extremity exoskeletons are becoming increasingly popular with specific designs looking to assist users in walking, running, and/or jumping. These types of devices focus around the hip, knee and ankle joints either exclusively or in combination. For healthy users, the challenge of improving an already efficient human body makes for a daunting task [2]. Improving one′s natural ability can come in the form of enhancing physical ability directly or reducing the fatigue factor during certain activities, allowing a user to perform a task for longer periods of time.

To truly observe reductions in metabolic cost and improve on the user′s natural ability, the weight of these exoskeletons must be minimized [3,4]. The location of the weight itself should also be strongly considered. The closer the weight is to the user′s center of mass, the less increase in metabolic effort required [3]. Harvard′s Biodesign lab which is part of the Wyss Institute has been developing a soft exosuit which utilizes soft textiles to better integrate with the human body. This exosuit connects to the user with Bowden cables actuated by motors to provide assistive torques at the hip and ankle joints to aid the user′s gait cycle. This design is lightweight at only 6.6 kg for the entire system, located at the wearer′s center of mass, and can reduce metabolic cost by about 7% on average [3,5,6]. A similar reduction in metabolic cost of walking was observed by Collins et al. with the development of an unpowered exoskeleton which uses only a passive clutch and spring system weighing between 0.4 and 0.5 kg for the entire device, per leg. This system relies on the body′s tendency to waste metabolic energy during walking and other activities and stores that energy in a spring to be used as positive energy [2]. While this device is lighter due to its passive design which does not require motors and controllers, the frame which attaches to the calf and ankle region is very rigid and, therefore, does not integrate well to the body. We have designed a system that can assist the lower body during walking, running, and jumping by utilizing the soft textiles and multi-joint assistance of the Harvard exosuit while incorporating the lightweight, passive actuation that Collins et. al. demonstrated.

The concept of unpowered assistance of human locomotion is made possible by the fact that the majority of energy is wasted in the process [2]. This allows a system to harness this energy and put it to use as an assistive force. Several key factors must be addressed to ensure the force is actually assisting the user. The first major factor is the alignment of the device along the biological muscles [3]. The human lower body has about fifty pairs of muscles, each with a specific function and range of motion. The muscles that contribute the most to the specific movement are the agonist or prime mover muscles. For example, the prime movers used with jumping are the gluteus maximus, quadriceps, hamstrings, and calves. Prime movers are assisted by a group of muscles known as assisting muscles. Alone, assisting muscles are unable to perform the motion at a functional level but in combination with prime movers, the motion is seamlessly accomplished. Some muscles perform the opposite motion of prime movers so as to prevent, slow down, or control a motion. This group of muscles are known as antagonist muscles. The quadriceps, for example, will contract eccentrically to slow down knee flexion during slow squats which makes the quadriceps the antagonist muscle. Finally, muscles can be grouped as stabilizers which means the muscles contract to hold the body in place while another body part is moving.

During calf raise exercises where the ankle is held in plantar flexion, the quadriceps would act as the stabilizer muscle to stabilize the knee during this extended position [7]. Aligning the force parallel to the prime mover muscles would be the most beneficial location since these muscles contribute the most to specific movements. Just as precisely placing the device is essential to observing the most significant impact, it is equally important to prevent misalignment of the device. Proper alignment of the device along the correct biological joints can be the difference between positive, assistive torques and kinematic restrictions on the users natural motion [3]. Alignment is especially challenging since the biological joints, unlike mechanical joints, do not have a fixed center of rotation [3]. Once the device is properly aligned with the joints, static and dynamic misalignment must be prevented to ensure the results are consistent and repeatable, and to ensure no hindrance on the users motion throughout the trial.

Another major factor that must be considered is the timing of actuation. A force can only be assistive if the timing is just right during the users motion. Malcolm et al. from Ghent University determined that the ideal range to assist ankle plantar flexion was between 40-50% of the gait cycle [8]. This range agrees with the range given by the Harvard Biodesign lab which states the ideal time range to aid the calf muscles is between 40-60%. Harvards exosuit is a multi-joint system so ideal timing for the hip was also determined which was at a range of 50-75%. To aid both the hip and the ankle, they determined that actuation should primarily occur between 40-60% of the gait cycle [3,5]. This timing range can be useful to providing the user with the most ideal assistive torque. Unlike the hip and ankle joints, the ideal actuation timing of the knee was not referenced in the literature as a direct joint receiving assistance. However, with the back-to-back negative work phases that the knee experiences, it would be possible to harness this energy to put to use at another joint. A negative work phase followed by a positive cannot be used for storing energy into a system since this energy is used by the biological muscles for storage and return of elastic tendon energy [8,9]. The timing of actuation is essential, but equally important is the direction of actuation. The assistive torque must be in the positive direction so as not to prevent the users motion which would result in an increase in metabolic effort to overcome the resistance. Collins et al. ensure their exoskeleton only provides positive torques through the use of a custom passive clutch which only engages when the users foot is on the ground and disengages when the foot is in the air during the swing phase to allow for natural motion [2].

Since passive devices do not rely on motors or power supplies to provide an assistive torque, the components used must be able to act as artificial muscles to store the energy from one motion and use it in another. Common substitutes of artificial muscle are elastics and springs, both of which have the ability to store kinetic energy into potential. A key factor to consider with these components is the stiffness. The stiffness can only be within a very finite range to be able to provide the user with assistance as opposed to resistance. Collins et al. determined that an intermediate spring stiffness was optimal for reducing metabolic cost [2]. The relationship between stiffness and percent change in net metabolic rate is parabolic; with increasing stiffness, the net change in metabolic rate decreased at first until reaching the optimal spring stiffness before the net change started to increase again until the metabolic rate showed an increase in metabolic effort rather than a reduction [2]. Finding the optimal range of stiffness is essential to ensuring the device is aiding the user and not providing additional resistance to range of motion. There can be a wide range of stiffness values that may aid in reduction of metabolic cost, however, testing should ensure the stiffness used is the most ideal for maximizing the results of metabolic cost reduction.

This paper focuses on the development of a multi-joint, passive exosuit that assists the user's lower extremities to perform specific motions with a reduction in metabolic effort. The exosuit is created of all soft materials that can conform to the body much more seamlessly in comparison to rigid components, and will be much more comfortable for the wearer with the possibility of adjusting the fit of the device. The use of soft materials increases the chances of misalignment which must be accounted for to ensure the device is providing assistive torques at the proper location. Elastics are placed at specific locations, and the hip and knee joints have been targeted to provide the most assistance.

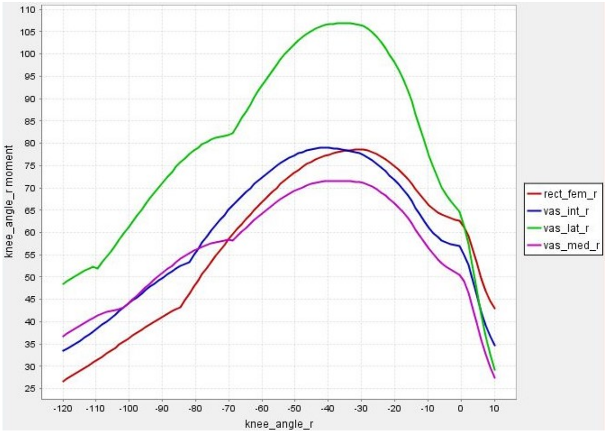
Section II provides the description and results of simulation. Section III talks about the concepts involved in designing the exosuit while Section IV describes the prototype that was fabricated. Results of the experiments to test its efficiency are provided in Section V and conclusions drawn from the project are mentioned in Section VI. Finally section VII provides the recommendations for future work in passive exosuits.

II. Simulation

A. OpenSim

After going through the literature, OpenSim models were analyzed to get a better understanding of lower limb dynamics. Walking and running models [13] are 23 DOF systems with 92 muscles. To understand which muscles contribute the most to a particular motion, muscles at each joint were analyzed for different variables. OpenSim′s built-in Inverse Dynamics solver was used to plot the dynamic variables such as moments and fiber forces versus gait cycle. The plots for the knee joint for both flexion and extension during walking are shown in the Fig 1.

Similar plots for other joints provided the information about where along each gait cycle each joint provided the most force for the motion as well as understanding the correlation between joint angles and general range of motion compared to location on the gait cycle. These plots were analyzed to target which muscles should be targeted for design. For the knee joint, the suit will specifically target knee extension so the vastus lateralis muscle was targeted since it provides the most force and generates the highest



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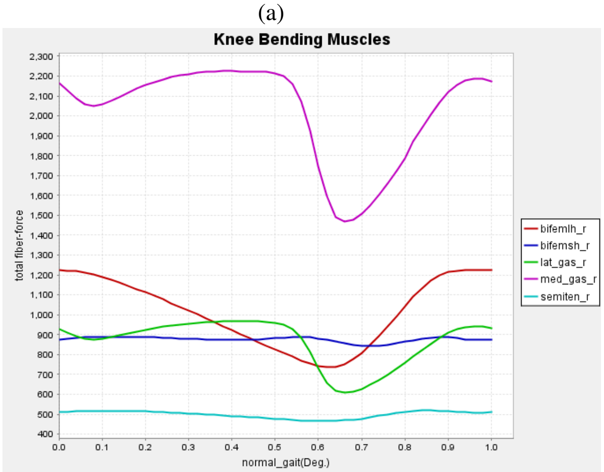
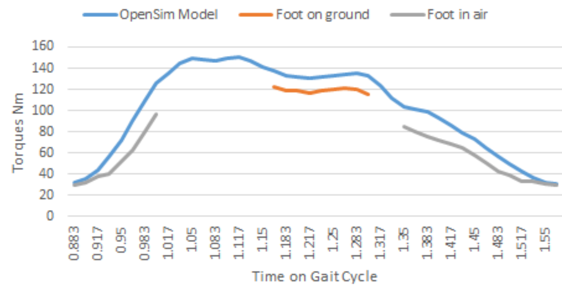


(b)

Fig. 1: Knee Extension and Knee Bending Moments vs Gait Cycle from OpenSim



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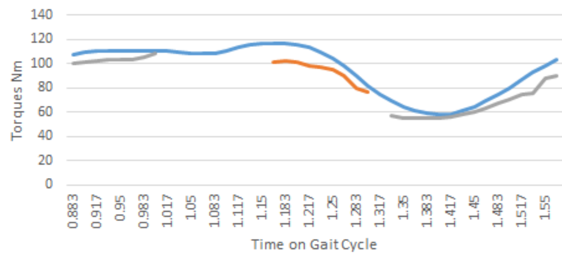

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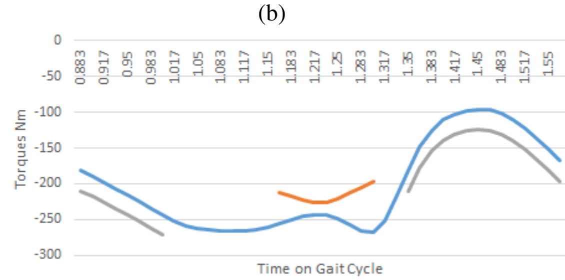


(c)

Fig. 2: Joint Torques versus one gait cycle time compared from both of our MATLAB models with OpenSim results (a) Knee joint torques (b) Hip joint torques (c) Ankle joint torques

moment during this motion. When it comes to designing the suit, the anchor points are located similarly to the anchor points of these two natural muscles.

B. MATLAB Model

The team created a simplified kinematic model of human walking in order to validate the results that were obtained from OpenSim. Similar to the homeworks that we completed in class, the team simulated the human leg as a 3 link arm. By assuming the leg only acting in the X-Y plane, and disregarding motion in the z-axis, the act of simulation became much easier. By following along with what we accomplished in class, the forward kinematics of a single leg were derived. From this, the velocity Jacobian was derived, which allowed for the team to solve for torques and forces on individual links. By taking the outputs of torques and positions from OpenSim, we then used the equation

$$Tau = J^T \times F \tag{1}$$

This did not give satisfactory results as the massless link assumption was an oversimplification.

A simplified model was created to approximate dynamics of human running. The modeling of one leg was divided in two phases. First phase is when the foot is in the air and the second when the toes are on the ground.

The first system has the hip joint as the base and the foot acts as an unloaded end effector. The other phase has the foot as base and the hip joint as the end effector with the rest of the body approximated as external load.The Newton-Euler formulation was used to calculate the dynamics of these two different systems and verify the OpenSim results using the joint angle information for running models of OpenSim.

Figure 2 shows the results obtained via the two MATLAB models compared with the OpenSim results. Here the results near the heel strike time are not shown as they were being estimated incorrectly by both the phases. Otherwise, the results follow the trend from OpenSim. As seen in figure 2c foot on ground phase failed to give satisfactory ankle torques results. We believe this was because of most of the simplifications made in our model affect the ankle joint.

III. Design Concept

A. Contributing Muscles

To increase the efficiency of the suit by ensuring it worked in parallel with our own muscles, we had to determine the muscles that had the greatest contribution to specific human motions. We studied the literature on running, walking, and jumping, and came up with a list of muscles for each of the joint motions. Some of the muscles have more than one attachment point; they are indexed by OpenSim as if they were a number of discrete muscles all belonging to one group. Also, a few of the muscle groups are not exclusive to only one type of motion. Fig 11 in the Appendix lists all the relevant muscles.

B. Optimal Anchor Points

Next, we determined the optimal locations to anchor our actuators. Through millions of years of evolution, the human body has the optimal design in regards to location, alignment, and anchor points of muscles that makes it possible to achieve a number of different kinds of motions. So we decided to mimic the biological anchor points of the muscles for our own actuators anchor points. As shown in Fig 3, the Biodigital Human software was used to pinpoint exact anchor points of the muscles we were focused on. For the actuation to be the most effective, the anchor points have to remain in the same location so that the actuation is consistent and always in parallel to the biological muscles. Therefore, static and dynamic misalignment have to be reduced to successfully implement assistive torques.

Fig. 3: Biodigital Human depicting two key muscles associated with knee extension. The natural anchor points of these muscles were mimicked in the design of the prototype

C. Passive Actuation

Springs are one of the most readily available candidates for passive actuation. Springs are very easy to control and can effectively store energy in a linear application. However, unlike springs, elastics are more compliant and therefore better able to bend at joints to maintain comfort and adjustability. Biological muscles themselves store energy as elastic energy, so to keep with the biomimicry of human muscles, elastics are the best option. For our design, we went with readily available exercise resistance bands. Resistance bands have different force ratings, and selecting the right ones play an important role in providing an ideal amount of assistive torque for the users specific application.

D. Timing of Actuation

Unlike an active system that has algorithms implemented on microcontrollers to control actuation timing, passive systems rely on their own natural dynamics. Therefore, other clever means have to be used to make sure the elastics start engaging only when they can provide assistance to the wearer. In our suit, the length of the elastics was adjustable based on the users height. The elastics will have to be relatively slack for the majority of the users gait cycle and at a specific moment in the users motion, the stretch of the elastic will be maximized to store the potential energy which will be converted into kinetic energy to aid the users motion.

E. Adjustability and comfort

We envision our suits to be able to worn by any person, which is why adjustability and comfort are the two key factors to be considered. Adjustability is ensured by designing the suit in a way that the wearer can alter the effective length of the elastics and and also their tension. Additionally, the suit has buckles to firmly secure the webbing straps to the body, which allows the wearer to adjust how tightly the straps wrap around the thighs and calves.

To make the suit as comfortable to put on as possible, weve used mostly soft materials in its construction; in fact, the only rigid elements are the buckles, and they are not even directly on the skin. Comfort is another motivation why elastics were preferred to springs. Elastics could be placed along the joints as they conform to the bending and flexing of the joints. By using soft materials, the system uses the human skeletal system as the rigid frame to anchor the actuators. With this in mind, the user must feel as comfortable as possible despite elastics tugging and pulling at specific locations.

IV. Prototype

The main goal of the initial prototype is to create a soft, wearable exosuit that can demonstrate the concept of assistance using optimally placed elastics of varying stiffness. All the components of the suit are soft and relatively conformable to the human body. The exosuit can be seen in figure 4. The main components of the suit are the rock climbing harness, webbing, elastics, elastic fabric, compression knee braces, and plastic buckles. Having all soft components helps with comfort, but to ensure the suit is comfortable for users of different shapes and sizes, adjustability was essential to the design. The major factors that influence the effectiveness of the suit are the location of the anchor points on the body, the timing of actuation and the actual assistive torque provided by the elastics.

(a)

(b)

Fig. 4: The prototype (a) Ventral view (b) Dorsal view

A. Optimal Anchor Points

The harness acts as the main anchor and sits on the waist, aligned by the iliac crest for maximum support. The closer the weight is to the center of mass, the less metabolic effort required to compensate for the added weight [5], therefore anchoring at the waist will minimize the effort required to carry the additional weight. Since rock climbing harnesses are meant to hold a significant amount of weight suspended, it should act as a decent main anchor of the suit.

Webbing was used to create the other anchor points. Buckles were sewn to the webbing to allow for adjustability. The webbing has two major issues as an anchor point. First, the webbing is not very comfortable and digs into the skin. To solve this problem, an elastic fabric was sewn into the webbing at a specific location at the back of the leg as shown in figure 4. Ideally, the elastic will be unstretched when the user stands up straight, but will stretch when the user moves to help with comfort. Another issue with the webbing is its tendency to slip on the skin. To help with this issue, sports tape was sewn to the inside of the webbing seen in figure 4 at anchor points where misalignment was prevalent to reduce slippage. Also, to keep the anchors from any rotational misalignment, straps of webbing were sewn at angles, both medially and laterally.

The elastics were placed through the two anchor point buckles which allows for each elastic to be doubled in the amount of assistive force that can be provided. The light green elastics over the knee were anchored at the top of the thigh, near the hip joint and below the knee, near the calf. These anchor points are similar to the vastus lateralis muscle which is the main contributor of knee extension motion. To keep the elastic consistently aligned over the knee joint, another webbing strap was placed right above the knee joint which has a loop to ensure the elastics do not slip laterally. Further, for keeping the elastics from rubbing against the skin directly and further keeping elastics in place, a fabric compression knee brace was used.

The hip anchor points are parallel to the biological muscles: gluteus maximus and biceps femoris long head. The grey elastic band is placed through two buckles that were sewn onto the main anchor point at the harness and then both elastics converge to the buckle located on the lateral side of the knee, as shown in 4. The placement of these elastics aid in the motion of hip extension.

B. Actuation Timing

Unlike active systems which can control, precisely, the moment of actuation, passive systems must rely on clever design to control actuation. This system relies on the lengths of the elastics, the users typical stride length and range of motion, and gait cycle events of common physical movements such as walking, running, and jumping. Precise control of the timing

The initial prototype assists the user at two joints for two specific motions: knee extension and hip extension. For walking, the hip is fully extended at the end of the gait cycle during the swing phase. To best utilize the users kinetic energy, the elastic will begin to stretch near the end of the gait cycle at about 95%. By using the users momentum during swing phases, the energy from the leg swing will increase the potential energy of the elastic. Once the users leg is at the maximum range of motion (the end of the gait cycle), the elastic potential energy will be converted into kinetic energy to assist the user with the motion of hip extension. The plot in figure 5 shows the total fiber force of the hip extension muscles, in this case the agonistic muscles, in purple, and the hip flexion muscles, the antagonistic muscles, in green. The plot shows roughly two gait cycles of running. To time the stretching and releasing of the elastic to an exact time on the gait cycle, the location where the agonistic muscles start to decline in force produced and the antagonistic muscles start to increase was thought to be the ideal location to start the stretching of the elastic. For the case of assisting hip extension, at about 0.5 on the gait cycle in figure 5, the hip extension muscles start to decline in total force while the hip flexion muscles start to increase. The elastic will be stretched until the 0.6 mark and will be subsequently released as an assistive torque up until the 0.7 mark. This timing will ensure that the users momentum is utilized as much as possible without hindrance to the motion.

A similar concept can be applied to the other joints. The main difference between assisting different motions will be the timing of actuation compared to the gait cycle. The second motion assisted by the suit is knee extension. For the most effective assistance of knee extension, the stretching will have to begin at the exact moment that the knee flexion muscles (antagonist) start to generate more force than the knee extension muscles (agonistic). The plot in figure 5 shows the knee extension muscles in red and the knee flexion muscles in blue. To best assist knee extension, the elastic will begin to stretch at 0.3 on the gait cycle labeled in the figure and release an assistive torque at about 0.4 when the extension muscles start to increase in force again. Once again, this will ensure the users momentum is optimally utilized and the elastic energy is released at a point that will only assist, not hinder.

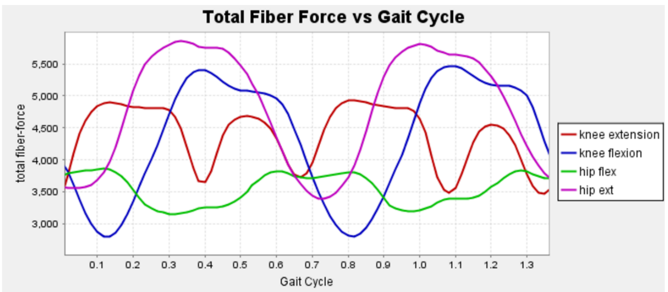


Fig. 5: Total Fiber Force in newtons versus the running gait cycle for the muscles contributing to knee extension, knee flexion, hip extension, and hip flexion

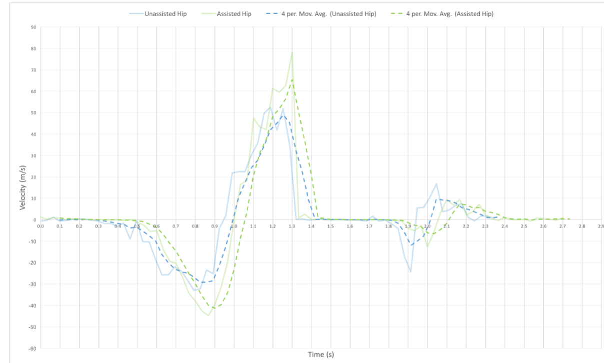
C. Elastic Assistance

For the design to be fully effective, the elastics must provide the user with an assistive torque while at the same time, not hindering the motion of the user in any way. The hip joint contributes a significantly higher amount of force for walking and running [5] and therefore the elastics used for the hip had a higher force rating of 30 pounds for each leg with the knee elastics rated at 15 pounds. An exact elastic stiffness and ideal actuation force was not directly tested due to time constraints. Future prototypes of the suit will test a variety of elastic stiffnesses and force ratings for average user weight and heights to find an optimal combination. However, for the current design, the buckles allow for enough adjustability for the user to tension the elastics as much as they feel fit, although an ideal elastic length and tension can be recommended for the user based on limb lengths and stride lengths.

D. Challenges

Several challenges were experienced during the actual fabrication of the system. Both static and dynamic misalignment of the anchor points was an issue. The user donning the suit featured many inconsistencies as far as anchor point locations with each trial. And also during trials with the suit, anchor points would slip against the user's skin.

Another challenge experienced was user comfortability. Upon tightening the exosuit, teammates reported discomfort due to the plastic buckles used digging into their skin. Since the exosuit was slipping so much, the team resorted to tightening the straps on the user as much as possible, which could cause potential circulation problems over extended periods of time.



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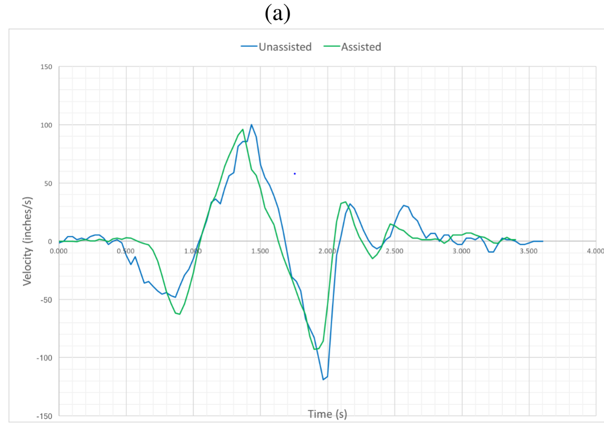

(b)

Fig. 6: Tracker Data (a) Hip Velocity(b) Wall jumping comparison

During the actual design phase of the prototype, it was realized that it would be nearly impossible to incorporate the ankle joint with the current system. With the way that attachments were made, for the current prototype to provide actuation to the ankle, the elastics would end up providing assistance to the wrong part of the foot, which could cause discomfort for the user or actually hinder them.

V. Results

The ultimate goal of this project was to demonstrate that the prototype we developed is able to augment the motion of the user in a significant way. Considering the time and resources available to us for this project the determination was made to analyze the motion of the user with and without the exosuit prototype using motion capture software.

Initially our intention was to take advantage of the motion capture system in professor Jane Lis lab however our timeline for completion of this project, assembly of the suit and testing did not allow this. Instead we elected to use a piece of software called Tracker to perform the motion analysis of the videos we captured. Tracker is a freely available software designed for analysis of physics and astronomy created by Douglas Brown as part of the Open Source Physics collaborative.

For use with Tracker our team took video of our test subject, Omar Hoblos, squatting, jumping, walking, running, and sprinting. Due to the design of the suit the motions that have the most significant augmentation factor are those that use the greatest range of motion i.e. running and jumping. Initially our plan was to analyze the motion of running first however in order for Tracker to accurately measure the motion of the user the camera needs to be a stable and perpendicular to the user as possible to avoid skewing the image and ultimately the data. The treadmill we had access to either prohibited us from taking video perpendicular to the test subject or obstructed the view of the subject limiting the data we were able to capture. For this reason the team elected to analyze the jumping and squatting motion over the running motion despite the reduced effect the suit would have on this motion.

To do this we took two videos, one of the test subject with the suit on performing a standing squat, a vertical jump, and a wall jump and one without totaling in six different video analyses. These videos benefited from no obstructions, and a nearly unskewed video angle that would lead the the most accurate motion analysis possible.

After loading the video files into Tracker we used the frame rate to determine the time interval between frames and used a calibration feature to assign a set number of pixels to a measured distance. With these two elements of the video identified we were then able to begin the analysis of the user's motion.

Our ultimate goal of the analysis was to determine the force and or acceleration of the user with and without the suit to determine the augmentation level provided by our prototype. Tracker has the ability to determine the angular velocity and acceleration as well as the linear velocity and acceleration. This is determined by placing markers on key locations in the video, in our case the toe, ankle, knee, hip, and shoulder were used.

Upon first glance the velocity and acceleration graphs produced by Tracker did not appear to match the motion of the user and its results were called into question. For this reason we used only the position information (x,y), produced by tracker and calculated from this the length of each limb of the user using the distance formula and the angle between the joints using the law of cosines. After calculating the internal angles of the joints we were able to calculate the angular velocity and acceleration of the discrete data points by taking the average slope between position and velocity data points. Knowing the angular velocities and accelerations we then used the vectorial form of the limb lengths and cross multiplied them with the angular velocities and accelerations to determine the i, j, k components of linear velocity. Knowing the linear acceleration of each of the joints we were then able to compare the change in acceleration seen during each of the three motions. Due to the fact that the knee and ankle accelerations are combined to achieve the hip acceleration the focus was placed on the change in acceleration of the hip in the j direction for each joint.

In the future more analysis will be performed on the individual joint motions and their contributions to the acceleration of the user.

The first analysis performed was on the squatting motion of the user. Our test subject wore the suit and performed the different motions (squatting, jumping, and running) and then then upon completion of these actions removed the suit and performed the same motions. This was done due to the time constraints we were under otherwise the test subject would have performed a single action unassisted, stopped, put the suit on and repeated the same action immediately thereafter. For this reason there is a potential for the test subject to have experienced minor fatigue while wearing the suit which could have influenced the results of the test. Furthermore the jump test and squat tests are somewhat subjective as the rely not only on the endurance of the subject but also on the consistency that the test subject executed the motion. If for some reason the test subject was to exert slightly more or less energy during one motion than another this could influence the test.

In the future multiple trials will be taken and averaged to ensure there is a consistent difference between the test with the suit and without.

In the graph below we compare the results of the assisted and unassisted squatting motions, specifically with respect to the acceleration of the hip in the j direction.

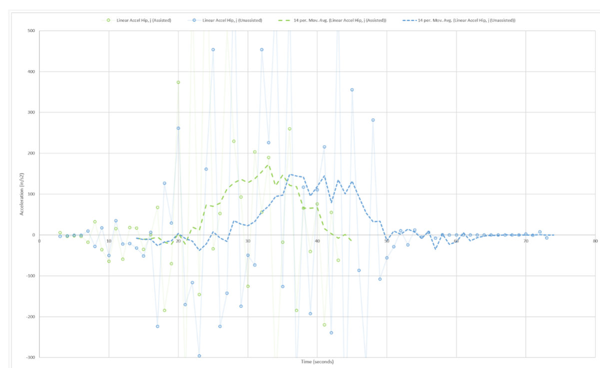

Fig. 7
Hip Acceleration - Squatting

In the figure above the trend lines created from taking a moving average of the acceleration data show pretty similar results in the overall amplitude. Due to the large fluctuations in the acceleration data it was difficult to determine which had a greater improvement. One notable difference in the data was the length of time it took the user to complete the squatting motion. The shortened length of the motion may be an artifact of the data processing but may also indicate that the user was able to complete the motion in a shorter duration indicating a positive change. See velocity comparison data below.

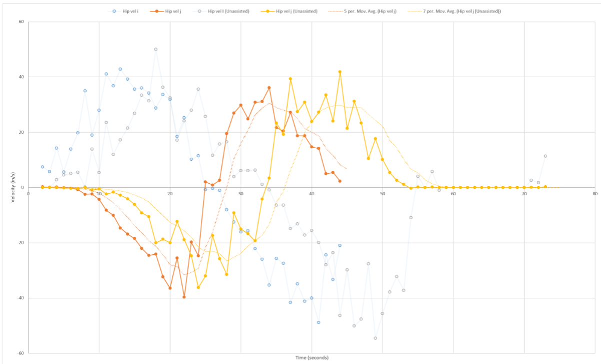

Fig. 8
Hip Velocity - Squatting

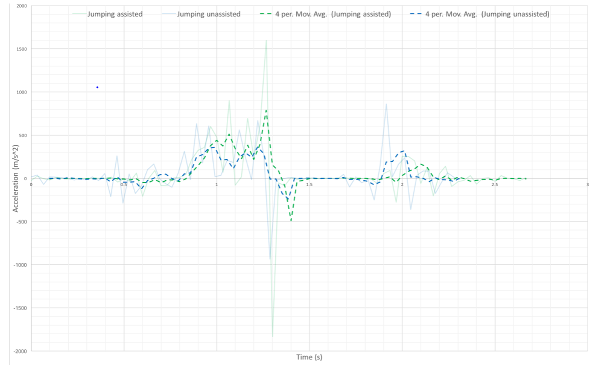

Fig. 10
Hip Acceleration - Vertical Jumping

Next we looked at the motion of the wall jump and the acceleration changes of the hip in the j direction.

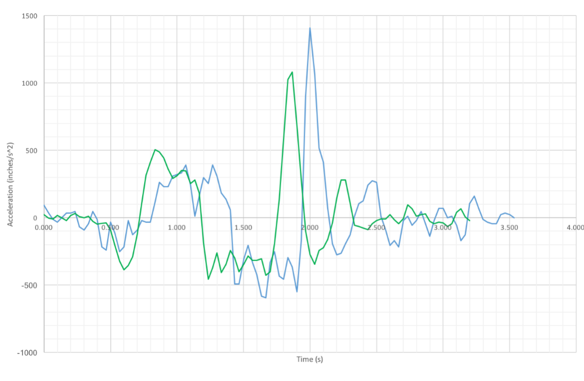

Fig. 9
Hip Acceleration - Wall Jump

This too did not show conclusive evidence that the suit made an appreciable difference in the jump height. One of the items that was of note was the velocity comparison of the hip during the wall jump. The assisted motion despite being the same overall level as the unassisted motion at the peak appears to show a significant reduction in the velocity upon impact which would indicate the suit is able to absorb motion and potentially protect the user's joints.

Finally we compared the straight vertical motion of jumping assisted and unassisted which seemed to show conclusively that an improvement was made as a result of the suit.

In this figure it is very clear that the suit augmented the users motion which was our first validation of the results. Despite the clear difference in acceleration of the hip this was only a single data point and more data will need to be taken to fully validate the results. As the authors see it this lays a solid groundwork for the expansion and further validation of our design in the future.

VI. Conclusions

Overall, the team has accomplished a lot since the initial proposal. The project began the project with a vague idea of what we wanted to accomplish, refined it into a solidified idea, and then moved on to actually build it. Through our extensive literature review, we as a team gained a strong background knowledge on the research completed so far in the fields of soft passive exoskeletons. And by using what we learned, were able to successfully design and fabricate our physical system.

With a strong base knowledge established, the team successfully identified and determined which muscles play the largest roles in the act of jumping, running and walking. We then quantified the contribution each muscle has within each motion through the use of OpenSim. The team then found which muscles played the largest role in locomotion, and which ones played mostly support roles. Once this data was gathered, the team then reduced the number of muscles needed in the model to the minimum number needed, i.e, only the most important muscles were modeled. The team also utilized our familiarity with OpenSim to create a simulation capable of performing a running and walking motion. While the team originally aimed to model both loaded and unloaded motion, the team ran into some problems in the development of custom actuators with the software.

While OpenSim is a very well known and documented software, the team decided to also write a MATLAB script that would simulate a simplified version of our OpenSim model. This was done in order to help us determine which constraints were redundant in our OpenSim model as well as verify that the dynamic results from our OpenSim model were correct. The team was successful in this endeavor, accurately writing a forward simulation script for a human leg, and utilizing the Newton-Euler method to verify the results gathered from OpenSim.

The team then began work on fabricating the original prototype of the device. It was decided that the most optimal placements of the elastics were found by mimicking the muscle distribution of the human leg. Elastics and webbing were bought and a first design was created, ready for preliminary testing.

With the prototype complete and ready, the team successfully ran several tests, showcasing a team member running, walking, jumping, and squatting, with and without the exosuit. Then, through extensive analysis and processing of our data, an estimate of the assistance the exosuit provided was able to be calculated.

All in all, the team was able to successfully complete almost all of the tasks originally set out for this project, thus resulting in a solid proof of concept prototype that is ready for future tests.

VII. Future Work

For future work, there are several aspects that can be further improved upon. These improvements would aid in helping to quantify and perhaps improve the amount of assistance that the exosuit provides to the user.

A. *Creating more realistic simulations*

The team recommends making more realistic simulations in order to better know what to expect from concepts before physical design. By being able to accurately simulate things such as actuators acting on joints, optimal placement points of the elastics would be able to be found. In addition to this, an accurate simulation of the exosuit on a human would enable the ability to make an estimate on the amount of assistance provided by the exosuit before the actual fabrication of it.

B. *Create an entire garment*

To ensure consistency of anchor point location and minimize the chances of webbing and elastics slipping, the team recommends the fabrication of an entire garment. By having one singular garment opposed to the disjoint pieces of fabric that we have now, the amount of slipping experienced by the user would be greatly reduced. The team recommends the following points to be taken into consideration when designing the full length exosuit.

1) Fabricate and sew on leggings of webbing
2) Custom sizes
3) tunnels/sheath design for elastics to travel through to ensure proper location of actuation
4) Custom 3d printed buckles for ergonomic fit

C. *Development of Ankle Joint Actuation*

The team recommends the extension of elastic actuation to the ankle joint as well. Currently, only the hip and knees are actuated on by the elastics. While this is indeed useful for wider and more general motions, for directed and specific movements, elastic assistance should be given to the ankle joint as well.

D. *Testing more elastics*

The rigorous testing of more elastics is also recommended. The choice of elastics made for the prototype could be improved on. Due to time and budget constraints, the team is unsure if the optimal elastics were purchased for the prototype. Future teams should put more emphasis and analysis on trying to find the best possible type of elastics for this purpose.

E. *More Quantitative Testing*

The team recommends more thorough and quantitative testing methods to be implemented. While a lot of work was done on the validation of the exosuit, more can be done to achieve more detailed numbers. This more detailed testing would include things such as validation through motion capture technology, metabolic testing, and more.

F. *Use of Sensors*

To gather more data on the exosuit, sensors can be embedded into the garment to help with validation. These sensors can be a variety of motion sensors, such as load cells, accelerometers, and more. The data provided through this would be extremely helpful in conjunction with future data collected through motion capture technology.

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VIII. Appendix

Hip Flexion	Hip Extension	Knee bending	Kne
Gluteus Medius 1 Rectus Femoris Psoas Major Iliacus	Gluteus Maximus 1,2,3 Gluteus Medius 3 Adductor Magnus 1,2,3 Semitendinosus Biceps Femoris Long Head (LH)	Semitendinosus Biceps Femoris Long Head (LH) Biceps Femoris Short Head (SH) Gastrocnemius Medalis (GM) Gastrocnemius Lateralis (GL)	Vastus Vastus Vastus Rectus

Fig. 11: Muscles contributing to each joint′s actuation

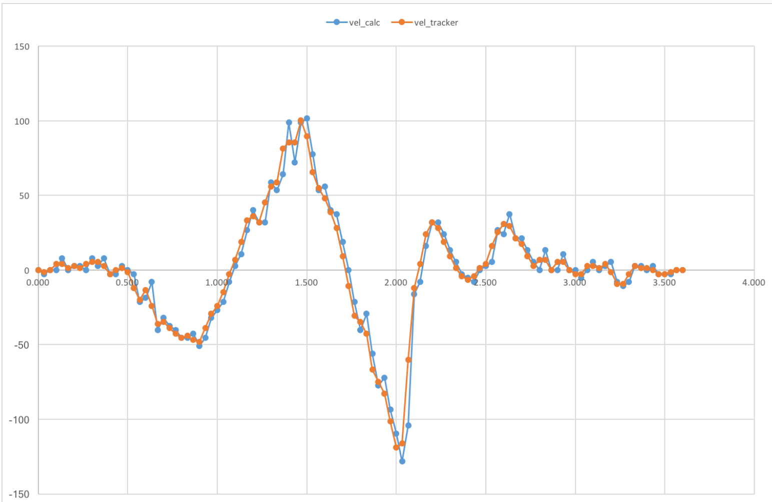

Fig. 12: Velocity check between excel data and Tracker. Confirms Tracker results are correct