COUNTER-BALANCING MECHANISM FOR IMPROVING INDEPENDENCE WHEN USING AN EXOSKELETON

by

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ABSTRACT

An exoskeleton is a robotic device used in assisting paraplegics with standing and walking. Existing designs use a series of DC motors and brakes to move the different parts of the device. Some exoskeletons mimic the musculoskeletal system by sending signals to a computer that tells the motors to rotate the knee, ankle, and hip joints appropriately for correct forward movement of the device. Users of the devices have a walker or crutches with controls to aid in balance. The goal of this project is to provide complete independence by removing the need for these walking aids. A new leg orthotic has been designed that may be implemented on any exoskeleton device to maintain balance in the fore-aft direction. A series of fast-acting electric actuators respond to the individual's movements. If at any point the device begins to tip, the actuators engage in such a way that the user's leg is brought back to an up-right position allowing balance to be recovered. As this movement takes place, the normal actions of the device's DC motors and brakes are also engaged to avoid falling (the reactions from the motors and brakes are already a feature of current exoskeleton designs.) This is a counter-balancing mechanism and could provide more independence to paraplegics in the future.

DEDICATION

This thesis is dedicated to my family, friends, and the staff of the University of Alabama for always pushing me and believing that I can accomplish tough goals through incredible odds. I cannot express how proud I am to be given this opportunity and I am forever grateful to those who made this possible.

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I would like to thank Dr. Beth Todd for her mentoring and believing that a biologist could actually get a master's in Mechanical Engineering. She showed me that staying focused and believing in me would prevail and in fact, it has. In addition, I would like to thank my other committee members Dr. Juan Lopez-Bautista and Dr. Keith Williams for their support through this long process.

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I would also like to thank Mike Poulton with Aker Solutions for taking time away from his job in helping me to learn and create my model in SolidWorks®.

Lastly, I thank my parents for always standing behind me on all of my decisions and that putting my full trust and faith in God would create a wonderful path for me, it has and I couldn't be more thankful and blessed.

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CHAPTER 1

INTRODUCTION

"The dream of regaining the ability to stand up and walk has come closer to reality for people paralyzed below the waist who thought they would never take another step" [4]. Imagine being paralyzed for 20 years and then, because of some incredible ingenuity, being able to walk once again and live a semi-normal life. That is what the robotic exoskeleton is providing some paraplegics around the world. The "Ekso™" is an example of a mechanical bodysuit invented to provide relief for individuals with spinal cord injuries. Ekso Bionics® is one of the leading companies to develop this piece of technology. These devices can allow a patient to sit, stand-up, walk, and climb stairs with a marginal range of independence. Furthermore, it is light weight and its compact size allows easy mobility. The military is already using it and soon it may be available for personal use, as soon as insurance companies create the correct policies [11].

Figure 1: Ekso™ Body Suit (www.eksobionics.com)

Some exoskeleton designs mimic the musculoskeletal system found in the human body. Different parts of the exoskeleton can carry out the many functions associated with the brain, muscles, and the sensory nerves. The exoskeleton often comes accompanied with a walker or crutches to provide balance and stability. This may be a flaw when it comes to complete independence for the user. Like the Ekso™ body suit shown in figure 1, exoskeletons may contain embedded sensors that detect differences in the user's hand movements, which are signals of the wearer's intention of what he or she would like to do. The control system detects and translates the sensory elements into a signal that moves the orthotic legs. Finally, a motor translates these signals into actual movements. For example, if pressure is exerted on both crutches, this may indicate to the sensors that the user wants to stand up or turn. Moving one crutch forward causes the opposite knee to flex which, in turn, causes the opposite leg to move forward or backward. The design of the sensors allows the user to walk over uneven ground and semi-difficult terrain. An experienced user can get up from one's wheelchair or other sitting position and strapped into the exoskeleton in a relatively fast amount of time.

It seems the greatest challenge with these exoskeletons is balancing without the use of a walker or crutches. The research that will be presented was performed to address this very issue. In the following chapters, one will learn how previous orthotics has been designed and the methods that were used to do so. In addition, two examples of exoskeletons will be discussed and how their functionality is utilized to help an individual walk. Finally, a newly designed orthotic with a counter-balancing mechanism will be described in how it may be implemented on any existing exoskeleton.

CHAPTER 2a

METHODOLOGY- BACKGROUND

A human knee primarily moves in one direction in the sagittal plane. The sagittal plane is a vertical plane dividing the body in right and left halves. Its range of motion depends on the range of motion of one's connective tissues, muscles, tendons, and ligaments that attach the upper leg to the lower leg. The joint in one's knee is made of the patella (kneecap), articulations at the ends of the tibia and femur, ligaments that hold the bones together and tendons that attach the bone to the leg muscles. The knee joint is coated in synovial fluid and has cartilages that provide smooth gliding when rotating. The knee moves through flexion and extension. Normal flexion of the knee (decreasing the angle of the knee) is between 120 and 150 degrees and normal extension is between 5 and 10 degrees. Engineers who design exoskeletons and other lower limb prosthetics must take into account and understand how an individual's knee rotates in order to create an intelligent device that mimics the musculoskeletal system of the user. In a 2001 survey of 435 lower-limb amputees, nearly half reported falling in the previous year. 40.4% of those individuals suffered injury [2]. A demand for better quality of lower-limb prosthetics arose and intelligent mechatronics seems to be the answer.

A combination of intelligence, that results from advanced control algorithms implemented on microcontrollers with high computational capabilities, and power in the form of high-capacity batteries and DC motors, implemented on a prosthetic provides an opportunity for better stability for lower-limb amputees. Dr. Goldfarb and his team have developed a way to detect a stumble in the swing phase of gait and stumble classification of intelligent transfermoral prosthesis. If disturbances such as stumbles or slips can be correctly identified, a powered prosthesis can be designed to initiate an active recovery response. Before this active response occurs, a method of identification must be developed. The human response to stumbling has been well-characterized in healthy subjects. Most studies identify two to three recovery strategies. These include lowering, elevating, and on occasion a delayed-lowering strategy when the elevating strategy is unsuccessful. The elevating strategy is commonly seen as a response to perturbations in early swing. In this strategy, a human flexes the hip, knee, and ankle to raise the swing leg producing two different effects. In the event that the leg is still hindered by the obstacle after the response is activated in the leg, the elevation helps the foot clear the obstacle to allow free swing. The second effect is that the flexion reduces the amount of inertia of the leg about the hip joint and allows a faster and longer step to be taken in order to support the trunk, which has traveled forward and rotated even more due to the momentum during the extended time of the perturbed stride. After this flexion period, a complementary extension phase is initiated to prepare the leg for stance at the next heel strike. The lowering strategy usually occurs in mid to late swing. This response is a premature exit from swing at the point where swing was hindered. The stance leg then executes an exaggerated step to clear the obstacle and to properly position the foot for the next stance phase. A delayed-lowering strategy occurs when an elevation strategy is performed but the foot does not clear the obstacle before the forward rotation of the trunk becomes too severe. When this happens, a human aborts the active flexion while the foot is still behind the obstacle and performs a lowering strategy instead. All strategies are on the order of 100ms. The research team conducted experiments on 10 healthy individuals in order to create an algorithm to classify stumble detection. They performed this by creating a walkway with obstacles mounted. Each subject had three low-pass filtered 50 Hz accelerometers strapped to them on their foot, shank and thigh. In addition, the authors used a 90Hz video camera to track the movements of each individual. Each accelerometer signal was high and low-pass filtered using second order filters with 3 and 40 Hz cut-off frequencies, respectfully. When a swing occurred, the filtered-data produced frames of 64 samples of data with the starting point's successive frames separated by 10 samples. The biomechanics of the swing phase contain primarily low frequency information. Furthermore, since stumble events contain high power at higher frequencies, a Fast Fourier Transform (FFT) was used by the authors to measure the amount of power at frequencies between 10 and 40 Hz. They used stumble flags that would be raised when the power at these higher frequencies exceeded a certain threshold, determined relative to normal swing. If 4 or more signals raised stumble flags within a 100ms interval, the algorithm reports a stumble. Their research provided an excellent method and algorithm to detect and classify stumbling when using lower-limb prosthesis [3].

In order to provide an individual with an orthosis to improve mobility, one must understand how a person's gait must be thoroughly analyzed. Gait is the pattern of movement of the limbs as an individual moves over a solid surface. A person's gait is defined by that individual's footfall pattern. However, the engineering field prefers gait to be defined based on mechanics. The term gait does not usually refer to limb-based propulsion through fluid mediums, but propulsion across a solid surface by generating reaction forces due to motion. Human gait is described as bipedal, biphasic forward movement of the center of gravity of the human body in which there are curved movements of different sections of the body with the least amount of expelled energy. Different gaits are related by differences in limb motions, overall velocity, kinetic and potential energy, forces, and fluctuations on the contact surface [5, 6]. Dr. Farris and his research team have developed a hybrid orthosis to aid in restoring gait back to individuals with spinal cord injuries. The device uses FES or functional electrical stimulation which stimulates the quadriceps muscle group of the legs. Along with FES, a controllable orthosis unidirectional couples hip to knee flexion, aids in hip and knee flexion with a spring assist, and incorporates sensors and modulated friction brakes. Combined, these elements help control joint and limb trajectories. Because of the unidirectional coupling, a knee extension does not generate a hip extension. The purpose of the spring and joint coupling is so that the knee joint is biased toward an equilibrium position where both the knee and hip joint are flexed. This combination enables knee flexion, hip flexion, and knee extension which all are from surface stimulation of the quadriceps muscle group of both legs. The joint coupled orthosis (JCO) also contains controllable friction brakes at both knees and hips. These brakes can independently lock the joints as well as modulate the resistive torque at each joint in order to control limb motion. The JCO also contains sensors at both hips and knees to detect angles of motion, providing feedback control of limb motion. The last element of the JCO is its ability to constrain motion along uncontrolled degrees of freedom (hip adduction and ankle flexion) which will enhance to stability and control of an individual’s gait. Probably the most important part of this device is the wafer disc brakes. The brakes provide added safety via the normally locked design of the knee brake, preventing the individual from falling if the device loses power, provided the device does not tip over. Also, the brakes increase muscle efficiency by locking joints during different phases of gait when they are normally static. This action takes the pressure of support off the leg muscles which reduces muscle fatigue and extends walking time. Finally, the brakes provide a smooth and controlled leg trajectory for a more natural and repetitive gait by using the brakes as dampers controlled in relation to joint angle feedback [7]. Researchers in the past have used magnetic particle brakes when building an orthosis. Since these magnetic brakes require electrical power in order to operate, if the power was to fail the brakes remain unlocked and the individual using the device could risk injury [8]. In order to remedy this potential problem, the researchers developed a new braking system using wafer disc brakes (WDB). The authors claim that the WDB has 45 times the torque-to-weight ratio of magnetic particle brakes. In addition, the WDB has the option of being locked or unlocked. For their orthosis, they have designed it in a way that the knee brakes are locked and the hip brakes are unlocked. The fact that hip brakes are used for trajectory control is the reason they remain unlocked. Designing the brakes in this manner allows for less power consumption and the added benefit that if electrical power fails, there is no risk of collapse. The WDB in the hip joint consists of a stack of thin high-strength plastic wafers which are splined to fit the brake rotor and stator. The stack of wafers receives a compressive force through a ball screw from a brushless DC motor located inside the brake shaft. This compressive force is proportional to the motor current. The hip brake contains 61 discs and because of this, the effective hip torque increases which provides greater torque-to-hip ratio than the magnetic particle brakes. The design of the brake is similar to a normally unlocked type with the exception that the discs are preloaded with a compressive spring. When current is applied to the motor, the preload proportionally unloads and maximum brake torque occurs at zero motor current and minimum brake torque occurs at full motor current. The ball screw is back-drivable which means the brake torque remains inversely proportional to the motor current. The knee brake designed has already been constructed and tested. The brake's mass is 0.73 kg and was experimentally measured to provide a maximum torque of 50.7 N-m which gives a resistive torque-to-weight-ratio of 69.4 N-m/kg. The torque varies linearly with input current for both brakes (inversely for the knee brake). Using the JCO requires the user to have a walker that provides assistance with balance. The authors state that two controllers are active at the same time. These controllers are the JCO controller (brakes and electrical stimulation) and the user controller, which governs the interaction with the walker through the arms. Once the authors had their prototype complete they tested using an individual 1.7m in height with a weight size of 65 kg. The cadence of that individual was 34 steps per minute with an average velocity of 0.2 m/s. They state that the upper body never leans backward and the maximum forward inclination is 25 degrees. This test also indicates that almost all of the weight is carried by the legs, resulting in minimal weight being applied to the arms. The authors have developed a JCO that works relatively well in restoring gait to spinal cord injured individuals. The remaining question is how can one develop a way to minimize the use of walkers and crutches all together? The answer to this underlying problem will become more interesting as one begins to study further into the other designs of orthoses [7].

In the previous sections, one has learned how a lower limb orthosis can aid in the mobility of some paraplegics. In addition, a stability control system has been described that aids in balancing while using an orthosis. Now, one should learn how a full-body exoskeleton can be built and used effectively by individuals that may be paralyzed from the chest down. These exoskeletons work using the same type of elements (computers, sensors, actuators, etc.) to provide outstanding mobility for paralyzed individuals. There are approximately 200,000 spinal cord- injured (SCI) individuals in the United States. Each year there are approximately 11,000 new SCI's, most reporting a reduced quality of life [9]. Being confined to a wheelchair for most SCI's is an improved means of mobility but psychologically, an individual would much rather be on his or her feet. Furthermore, because of architectural and environmental restrictions, there are many limitations associated with using a wheelchair. The ReWalk® powered exoskeleton is a device created to address this very issue. Their goal was to develop a device that not only provided a means of mobility, but mobility over a longer distance than most other lower-limb orthoses and FES gait enhancing systems. The ReWalk® is a lower limb powered exoskeleton that allows thoracic or lower level motor-complete individuals with spinal cord injuries to improve their independence of walking. The device has independently controlled bilateral hip and knee joint motors, rechargeable batteries, and a computerized control system that is carried by means of a backpack. Users of the ReWalk® control their walking and movements through trunk motion and changes in center of gravity. The angle of the torso is determined by a tilt sensor as well as generates a preset hip and knee displacement (angle and time) that will result in a step. The ankles use a double-action orthotic joint with limited motion and spring-assisted dorsiflexion that is adjustable through screw tension. The ReWalk® is adjustable in height and in width. In addition, the device has padded interfaces for calves and thighs and the limbs are linked via a rigid pelvic frame. An individual using the exoskeleton is secured to the device by a waist belt, shoes, and Velcro closures with pads. As with other devices, standing stability for the ReWalk® is achieved by the use of crutches. The user can remotely interact with the system with a user-operated wrist pad controller that can command sit-to-stand, stand-to-sit, and walk-activation. This unique design allows the user to actively be involved with controlling the device. The software algorithm was specifically designed to interpret signals from the torso tilt sensor. Furthermore, this algorithm then generates alternating limb-coordinated motion to produce bipedal walking. One function of the ReWalk® is that it does not allow two sequential steps to be taken by the same leg. During training, an external computer may be used to adjust hip and knee joint angle displacements to optimize the walking characteristics or implement a training mode. Also equipped, is a manual mode of operation used by the individual that is used to trigger steps, bypassing the tilt sensor. This same mode may be used to trigger sit-stand-sit transfers. Walking using the ReWalk® exoskeleton is similar to upright bipedal walking, offering a way in which SCI's can overcome some of the physical psychosocial problems that are common in these individuals.

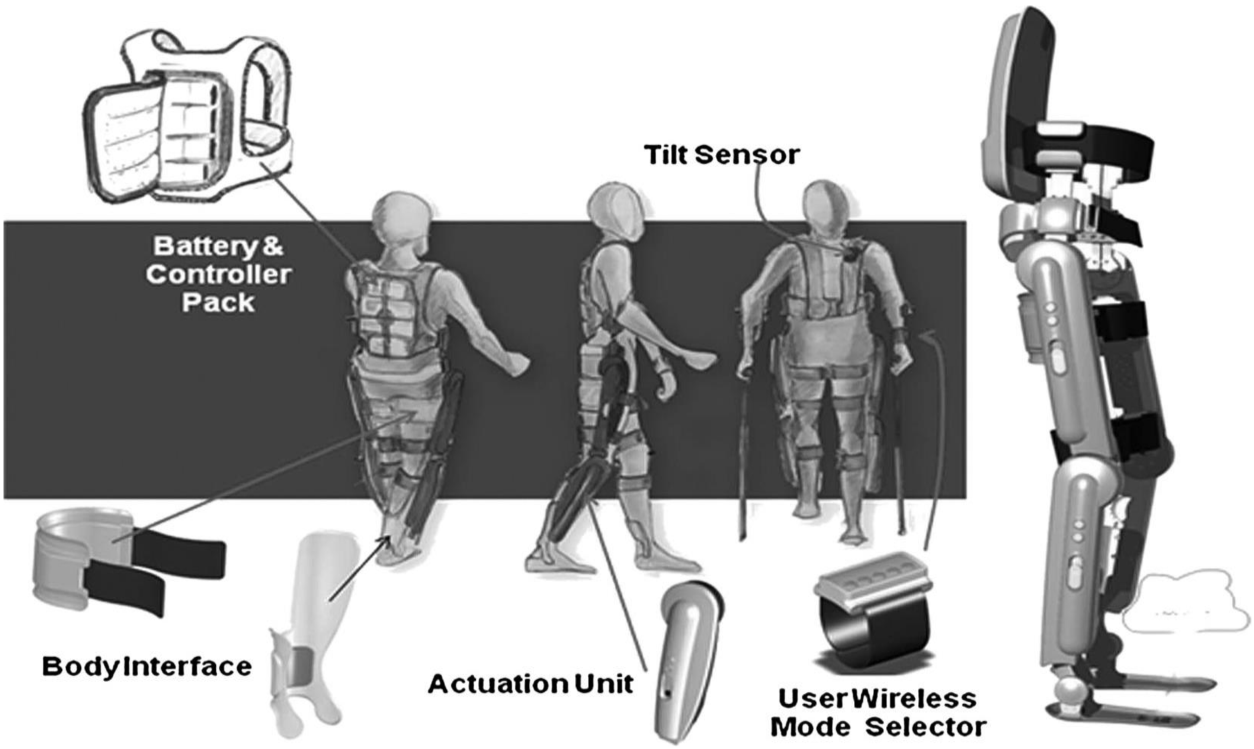

Figure 1a: ReWalk® Exoskeleton (Esquenazi et al. 2013)

ReWalk® researchers conducted tests on 13 individuals using the device. The average range of speed was between 0.03 and 0.45 m/s. The test patients were evaluated over a 10 m walk although many subjects were able to walk much greater distances (greater than 150m). One of the negative aspects of the tests was several patients not being able to consistently and comfortably use the remote wrist controller. These subjects were still in control of their walking by self-triggering steps; however, they were not able to consistently change from walking and sitting routines by themselves as they were more comfortable using the crutches for stability [10].

Another powered exoskeleton that has been developed is the Ekso™, engineered by Ekso Bionics®. The Ekso™ is a mechanical bodysuit invented to provide relief for individuals with spinal cord injuries. Ekso Bionics® is one of the original companies to develop this piece of technology. The Ekso™ can allow a patient to sit, stand-up, walk, and climb stairs with a marginal range of independence. Furthermore, it is light weight and its compact size allows easy mobility. The military is already using it and soon it will be available for personal use, as soon as insurance companies create the correct policies [11].

Figure 1b: Ekso Bionics® Exoskeleton (www.eksobionics.com)

The Ekso™ mimics the musculoskeletal system found in the human body. Different parts of the exoskeleton can carry out the many functions associated with the brain, muscles, and the sensory nerves. The exoskeleton often comes accompanied with a walker or crutches to provide balance and stability. Some may say that this is its flaw when it comes to complete independence for a patient. Also, the Ekso™ may contain embedded sensors that detect differences in the patient's hand movements, which are signals of the wearer's intention of what he or she would like to do. The control system detects and translates the sensory elements into a signal that moves the bionic legs. Finally, a motor translates these signals into actual movements. For example, if pressure is exerted on both crutches, this may tell the sensors that the patient wants to stand up or turn. Moving one crutch forward causes the opposite knee to flex which, in turn, causes the opposite leg to move. The design of the sensors allows a patient to walk over uneven ground and semi-difficult terrain.

An experienced user can get up from a wheelchair or other sitting position and strapped into the Ekso™ in five minutes. With the Ekso™, a patient gets the choice of three walk modes:

- FirstStep™- A physical therapist actuates steps with a button push. The user progresses from sit to stand and from using a walker to walking with crutches, often in their first session of training.
- ActiveStep™- Users take control of actuating their steps through buttons on the walker or crutches.
- ProStep™- The Ekso™ can sense the patient's gestures and body shift when the patient is in the right position. In other words, the device knows the patient wants to take a step and takes a step for them.

The Ekso™ also has a training mode where the device can be set to provide audio feedback when the patient has achieved the ideal positioning required for a step. Training mode is used to determine the position targets for a patient that wants to move to ProStep™. The exoskeleton is also equipped with EksoPulse™- a data module that automatically gathers and transmits statistics and device information during walking sessions. There are limitations to using the Ekso™. The device may be used by people with lower limb weakness or paralysis due to spinal injury or disease, multiple sclerosis, and Guillain Barre Syndrome. The height range of the device is 5'2" to 6'2" with a maximum weight of 220 pounds. Also the maximum hip width is 16.5". Some other requirements include proper upper body strength to balance with crutches or a walker, ability to transfer oneself from a wheelchair or other chair, and a complete evaluation and screening by a medical provider [12]. As with the ReWalk®, this device requires the user to use crutches for balance as well as giving commands to the Ekso™. So, what can one do to provide an individual with even more independence while using an exoskeleton? This purpose of this thesis is to show how a simple series of actuators can be implemented on many existing exoskeleton designs to provide further independence for an individual with a SCI.

CHAPTER 2b

METHODOLOGY- APPROACH

Designing a new exoskeleton was not a goal of this research. Instead, the focus was creating a new design to the lower limbs that could be implemented on any existing device. A design goal was to keep the device relatively simple and inexpensive. This led to a mechanical system that could run off of the same power and signals as the rest of the exoskeleton, performing the desired tasks that are interpreted as signals being sent to the computer that is built in the many current designs of the exoskeleton. If an individual is on an unstable or uneven surface, without the assistance of crutches or a walker, how would that person be able to make the proper motions necessary to keep him or her from falling and getting injured? This was the inevitable question that needed to be answered. One idea was to make a device that is heavier with a lower center-of-gravity. This idea may be simple but definitely made a valid point as one tends to have much better balance when the greatest amount of weight and force is closer to the ground. The problem with this idea is the fact that most that use the device is paralyzed from the chest down and their sense of gravity is obscured when considering their lower limb movements. Ultimately, it is quite a convoluted process trying to figure out how to lower the center of gravity on a device that has already proven to work adequately the way it is already designed. It was determined that a series of actuators could be used in an attempt to counter-balance an individual who is using the exoskeleton. The preliminary designs are as shown in Figure 2a.

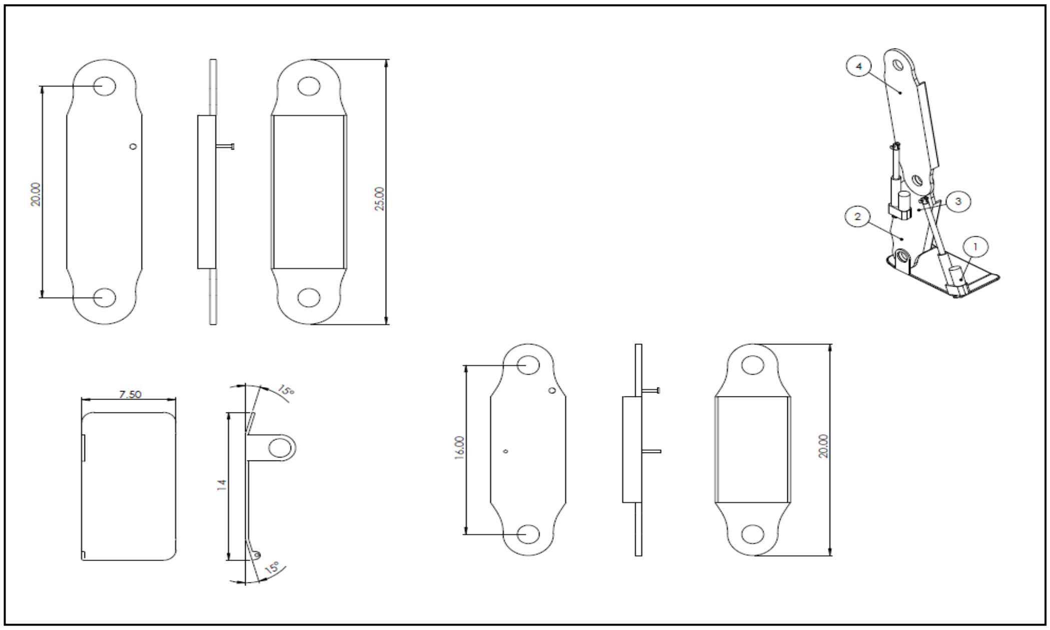

Figure 2a. Additional Components for the Orthotic: 1. Connection of the actuator to the footplate. 2. Lower orthotic link. 3. Knee joint. 4. Upper orthotic length

CHAPTER 3

RESULTS

The final design created using SolidWorks® is shown in Figure 3a. From the figure, the design has the actuators attached as well as circular mounts for the ankle, hip, and knee motors.

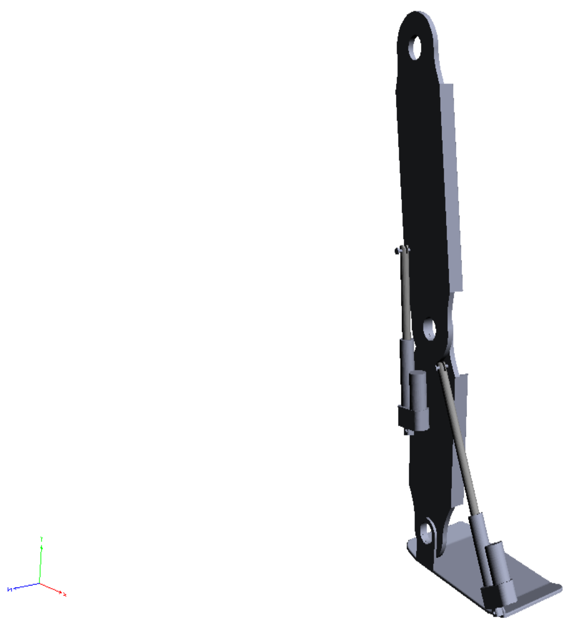

Figure 3a. Final Assembled Design

The new orthotic presented operates in a similar fashion as the previously mentioned orthotic and exoskeleton designs. There are mounts at the ankle, knee, and hip joints to allow motors and a braking system to be installed for functionality that may be preferred by a potential buyer for the design. In addition, a tilt sensor (not shown in figure) is located at the bottom of the ankle bracket. This tilt sensor operates in a way that if and when the foot plate is rotated past 15 degrees up or down in a vertical plane, a signal is sent to the computer as a stumble flag. This flag is interpreted as a signal and engages all actuators, bringing the user back to an up-right position. As the actuators engage, the user will regain the sense of stability being exerted on his or her center of gravity, allowing the user to regain control in an amount of time before the exoskeleton makes the next step. In addition, if an exoskeleton loses all power, the brakes automatically engage and the actuators lock with hopes of preventing the user from falling and risking injury.

It was determined that the center of gravities (CG's) for each part of the leg needed to be found in order to show that the device was balanced properly. The CG's are listed below in Table 1.

Table 1. CG Location

Center of Gravity			
Coordinate	**Top Leg Link (in)**	**Bottom Leg Link (in)**	**Foot Plate (in)**
X-coordinate (from left)	2.93	3.01	5.5
Y-coordinate (from top)	16.45	11.14	6.02

The free body diagrams for finding relative forces and center of gravities are shown in Figure 3b.

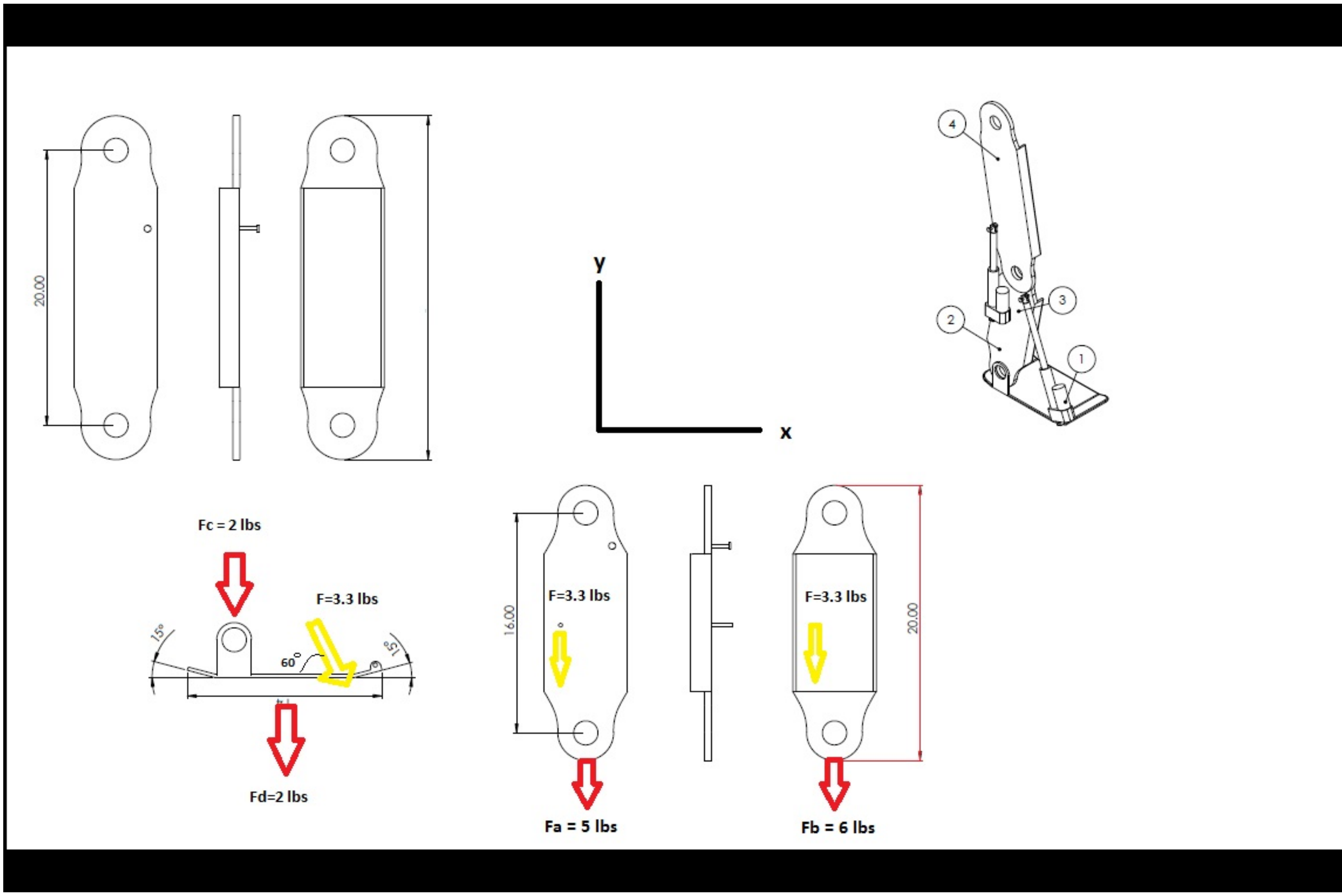

Figure 3b. Free Body Diagrams: Fa, Fb, Fc, Fd= Internal Forces. F= Actuator Forces

The red arrows represent internal forces of each part of the device. In addition, the yellow arrows represent the forces created on each part by the actuators. The specifications of the entire design are shown in Table 2.

Table 2: Component Lengths

Top Leg Link Length	20 inch
Bottom Leg Link Length	16 inch
Ankle Bracket Length	4 inch
Foot Plate Length	14 inch
Foot Plate Width	7.5 inch
Front/Rear Foot Plate Angle	15 degrees

The actuators used on the orthotic were 18 inch stroke, 12 Volt DC, 200 lbs max force actuators produced by Firgelli Automations®. The actuators are capable of movements as fast as 1 inch per second. This is a good reaction time because of the short rate of time that signals are sent from the tilt sensors to the computer, and back again to the actuators.

Further analysis was done to determine the loads that the new exoskeleton legs would see. Static equilibrium formulas were used to find the amount of force exerted on each joint of the new orthotic. It was estimated that the average approximate mass that would be evenly exerted on one leg is described below in Table 3.

Table 3: Relative Masses and Loads

Average Exoskeleton Mass	45 lbs
Max Mass Exo. Can Hold	220 lbs
Distrib. Force on Each Leg	132.5 lbf
Actuator Mass	3.3 lbs
Total Orthotic Mass	15 lbs

Given the data for Table 3, it was determined that the actuators chosen for the new orthotic would in fact work and provide the proper amount of force to help keep a user from potentially falling and risking injury.

The goal here was to show how many extra amp hours are needed in order to be implemented and properly function on an existing exoskeleton design. Another task that needed to be addressed was power consumption. Most of the orthotics and exoskeleton designs mentioned in the previous section have custom-made battery packs capable of producing enough power and battery life for up to 8 continuous hours of usage. The extra power that was needed to run 4 actuators at maximum load was determined. The following equation represents the amount of amp hours (Ah) that is needed to run 4 actuators for 8 hours:

I_{drawn}= (5A*4)*8hrs=**160 Ah** at V=12 volts; where I is the current in amps that is drawn by 4 actuators and thus is required as a minimum from the battery source that will be powering a full-size exoskeleton. The total Watts of the design is 240W.

CHAPTER 4

CONCLUSION

An orthotic has been modeled and the results showed that the modifications provide adequate force to prevent tipping in the sagittal plane. Though the new mechanism that has been proposed has not been tested thoroughly in a facility, it does offer hope and a new idea for gaining the fore-mentioned independence a user may want. There are fail-safes built into the design, should the device lose power. As mentioned previously if an exoskeleton loses all power, the brakes engage as well as the actuators leaving the user stranded in what could be an awkward position for an unknown amount of time, however; this action may also keep the user from getting injured. Also, if there is malfunctioning in the tilt sensor at any point, the actuators may not react to signals from the computer effectively resulting in un-proper gait of the exoskeleton. These problems may need to be addressed once a full-scale test is conducted.

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